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Hip Joint Muscle Forces during Gait in Patients with Femoroacetabular Impingement Syndrome are Associated with Patient Reported Outcomes and Cartilage Composition

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Abstract

Femoroacetabular impingement syndrome (FAIS) consists of abnormal hip joint morphology and pain during activities of daily living. Abnormal gait mechanics and potentially abnormal muscle forces within FAI patients leads to articular cartilage damage. Therefore, there is a necessity to understand the effects of FAI on hip joint muscle forces during gait and the link between muscle forces, patient reported outcomes (PRO) and articular cartilage health. The purposes of this study were to assess: 1) hip muscle forces between FAI patients and healthy controls and 2) the associations between hip muscle forces with PRO and cartilage composition (T_{1P}/T_2 mapping) within FAI patients. Musculoskeletal simulations were used to estimate peak muscle forces during the stance phase of gait in 24 FAI patients and 24 healthy controls. Compared to controls, FAI patients ambulated with lower vasti (30% body-weight, p=0.01) and higher sartorius (4.0% body-weight, p<0.01) forces. Within FAI patients, lower peak gluteus medius, gluteus minimus, sartorius and iliopsoas forces were associated with worse hip joint pain and function (R = 0.43 – 0.70, p=0 – 0.04), while lower muscle forces were associated with increased T_{1P} and T_2 values (i.e. altered cartilage composition) within the hip joint cartilage (R = −0.44 – −0.58, p=0.006 – 0.05). Although FAI patients demonstrate abnormal muscle forces, it is unknown whether or not these altered muscle force patterns are associated with pain avoidance or weak musculature.

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Conflict of Interest Statement

Alan L. Zhang is a paid consultant for Stryker Orthopaedics yet this relationship was not related to the outcomes of this study.

Further investigation is required in order to better understand the effects of FAI on hip joint muscle forces and the associations with hip joint cartilage degeneration.

Keywords

Femoroacetabular Impingement; Gait; OpenSim; Hip Joint; Musculoskeletal Simulation; Muscle Force; $T_{1\rho}/T_2$ mapping

1. Introduction

Femoroacetabular impingement syndrome (FAIS) is a morphological abnormality of the hip joint with corresponding clinical symptoms of hip joint impingement (Griffin et al., 2016). FAI is associated with severe hip joint disability and pain during activities of daily living (Ganz et al., 2003). The abnormal hip joint contact area present in FAI patients is associated with articular cartilage abnormalities (Meermans et al., 2010; Samaan et al., 2017b) and labral injuries (Lavigne et al., 2004). If not properly managed, FAI may lead to hip joint osteoarthritis (OA) (Ganz et al., 2003). In particular, quantitative magnetic resonance imaging (QMRI) based techniques such as $T_{1\rho}$ and T_2 mapping have been shown to be sensitive enough in detecting early stage alterations of the hip joint articular cartilage composition (higher $T_{1\rho}/T_2$ values) within FAI patients compared to asymptomatic controls (Karupppasamy et al., 2013; Samaan et al., 2017a).

Abnormal hip joint cartilage health in FAI patients may be due to altered lower extremity joint loading patterns exhibited during walking yet the results of these previous studies are not consistent (Diamond et al., 2016a; Hunt et al., 2013; Kennedy et al., 2009; Kumar et al., 2014; Samaan et al., 2017b). More specifically, in one previous study (Hunt et al., 2013), FAI patients exhibited reduced external hip flexor moments during walking but this result was not supported by similar studies (Diamond et al., 2016a; Kennedy et al., 2009; Kumar et al., 2014; Samaan et al., 2017b). When compared to asymptomatic controls, an increase in the external hip flexion moment impulse during the first half of the stance phase of gait was strongly associated with increased hip joint pain, dysfunction and severity of acetabular cartilage abnormalities within FAI patients (Samaan et al., 2017b). In addition, FAI patients exhibited lower isometric hip flexor, extensor, adductor and abductor strength compared to healthy controls (Casartelli et al., 2011; Diamond et al., 2016b; Kierkegaard et al., 2017).

Musculoskeletal simulations allow for the estimation of subject specific muscle force patterns during walking and can provide clinicians with the information needed to develop better intervention protocols aimed at restoring normal muscle function and reducing clinical symptoms. Previous studies using musculoskeletal simulations demonstrated lower hip muscle forces during gait in patients with hip dysplasia (Harris et al., 2017; Skalshoi et al., 2015) and patellofemoral joint OA (Crossley et al., 2012). In the current study, a combined approach consisting of musculoskeletal simulations and quantitative MRI was used to determine the effects of FAI on muscle forces and the associations of these muscle forces with patient reported outcomes and cartilage composition within FAI patients. It is hypothesized that: 1) FAI patients will ambulate with less hip joint muscle forces compared to healthy controls and 2) FAI patients that ambulate with decreased lower extremity muscle

forces during gait will exhibit more hip joint pain, dysfunction and worse hip joint cartilage composition.

2. Materials and Methods

2.1. Participants

Twenty-four FAI patients from our University's Hip Arthroscopy Clinic as well as 24 age-, sex- and BMI-matched healthy controls were recruited for this study (Table 1). All FAI patients demonstrated both morphological and clinical signs of impingement. Patients with an alpha angle of > 55° (Domayer et al., 2011), measured on oblique axial MR-images were considered to have the cam impingement, while patients with a lateral center edge (LCE) angle of > 35° (Philippon et al., 2012) on anterior-posterior (AP) radiographs were considered to have pincer impingement. Patients that met both of these morphological-based criteria were classified as mixed-type FAI. Each FAI patient demonstrated positive clinical signs of impingement (i.e. pain reproduction with flexion adduction and internal rotation [FADIR] test) (Klaue et al., 1991) during physical examination by an orthopaedic surgeon (A.L.Z.). All control participants were recruited from the local community and were part of a longitudinal study on hip OA. None of the control participants used in this study exhibited clinical signs of impingement (i.e. negative FADIR test). Control participants underwent an AP weight-bearing pelvic radiograph in order to assess radiographic signs of hip OA bilaterally. Study participants were excluded from this study if they had: 1) total joint replacement of any lower extremity joint; 2) previous hip surgery on the affected side; 3) pain at any other lower extremity joint except the study hip; 4) neurological, spine or lower extremity conditions that would affect movement; 5) contraindications to MRI; 6) radiographic signs of hip OA on either side (Kellgren-Lawrence score of >1) (Kellgren and Lawrence, 1957) and 7) body mass index (BMI) > 30 $kg{\cdot}m^{-2}$. All participants provided written informed consent prior to testing. This study was approved by the University Committee on Human Research.

All study participants were asked to provide self-reported measures of hip joint pain and function using the Hip disability and Osteoarthritis Outcome Score (HOOS) (Nilsdotter et al., 2003). HOOS scores range from 0 to 100, where a score of 0 and 100 indicate severe pain or dysfunction and no pain or dysfunction, respectively.

2.2. Experimental Data Collection and Processing

A marker set consisting of 45 retroreflective markers were used to collect 3-dimensional position data. Calibration markers were placed bilaterally at the greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli and first metatarsal head. Pelvic tracking was performed using individual markers placed at the anterior superior iliac spines, iliac crests and the L5/S1 joint. Torso tracking was performed using markers placed at the acromion processes, C7 and sternal notch. Thigh and shank segment tracking was performed using rigid clusters consisting of four markers each, while foot segment tracking was performed using markers placed at the fifth metatarsal head and clusters consisting of three markers placed on the heel shoe counters. A 10-camera motion capture system (Vicon, Oxford, UK) and two in-ground force plates (AMTI, Watertown, MA) were used to collect

three-dimensional marker position and ground reaction force (GRF) data at 250 Hz and 1000 Hz, respectively. A one-second static calibration trial was obtained and all calibration markers were then removed.

All study participants performed gait trials at a fixed walking speed of $1.35 m \cdot s^{-1}$, which is the mean of the average walking speeds of male and female adults on a level surface (Perry and Burnfield, 2010). Five successful trials were obtained and analyzed for each participant where a successful gait trial consisted of the participant's entire foot making a clean strike on one of the two force plates and their speed being within $1.35 m \cdot s^{-1} \pm 0.07 m \cdot s^{-1}$.

All raw marker position and force plate data were filtered using a fourth order, Butterworth filter at 6 Hz and 50 Hz, respectively. An eight segment kinematic model composed of the torso, pelvis, bilateral femurs, shanks and feet were created using Visual3D (C-Motion Inc., Rockville, MD) from the standing calibration trial. The hip joint centers were defined as one-quarter of the distance from the ipsilateral to the contralateral greater trochanters. The knee joint center was defined as the midpoint between the lateral and medial femoral epicondyles. The ankle joint center was defined as the midpoint between the lateral and medial malleoli. Segment coordinate systems were defined using an unweighted least-squares approach (Spoor and Veldpaus, 1980). An inverse kinematics algorithm designed to reduce joint motion artifact was used to determine joint kinematics (Lu and O'Connor, 1999).

Electromyography (EMG) data were collected at 2000 Hz using a wireless EMG system (Delsys Trigno, Delsys Inc., Boston, MA). Prior to electrode placement, the skin was shaved and cleaned with isopropyl alcohol. Skin preparation and EMG electrode placement for the gluteus medius, vastus lateralis, vastus medialis, medial and lateral hamstrings, medial and lateral gastrocnemii muscles was performed according to the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) guidelines (Hermens et al., 1999). After electrode placement each participant performed a 5 second maximal voluntary isometric contraction (MVIC) for the gluteus medius, quadriceps, hamstrings and gastrocnemii. The gluteus medius MVIC was performed with the participant lying on the contralateral side with their torso and lower extremities in a fully extended position. Manual resistance was then applied at the lateral femoral epicondyle and lateral malleolus while the participant performed a maximal hip abduction contraction. The quadriceps MVIC was performed with the participant seated on a plinth and the knees flexed to 70° using an adjustable strap. A strap was used to stabilize the pelvis during the quadriceps MVIC. The hamstrings MVIC was performed with the participant lying prone on a plinth with the knees flexed to 70° (secured using an adjustable strap) and the pelvis stabilized using a strap. The gastrocnemius MVIC was performed by asking the participant to stand up and to maximally contract the gastrocnemius through maximal ankle plantarflexion, while using a table to help maintain balance. Finally, a one-second resting trial was obtained with the participant lying prone on a plinth.

The average resting voltage for each muscle was determined from the resting EMG trial and subtracted from its respective dynamic EMG data during walking. Next, all dynamic EMG data were bandpass filtered using a 4th order Butterworth filter (20 – 500 Hz), full-wave

rectified and low-pass filtered using a 4th order Butterworth filter with a cut-off frequency of 6 Hz. These filtered EMG profiles were then normalized by the peak MVIC values.

2.3. Musculoskeletal Modeling

A generic eight segment (Gait 2392), 19 degree of freedom (DOF) OpenSim musculoskeletal model (Delp et al., 2007) consisting of 92 musculotendon actuators was used to create scaled models for each participant using the anthropometric data determined from the standing calibration trial. The torso segment was modeled as a 3-DOF ball and socket joint. The pelvis was modeled using 6-DOF consisting of 3 translations and 3 rotations. The hip joint was modeled as a 3-DOF ball and socket joint while the knee and ankle joints were each modeled using 1-DOF. Tibiofemoral translations were described as a function of knee flexion angle (Yamaguchi and Zajac, 1989).

An external numerical optimization algorithm (Samaan et al., 2016; Weinhandl et al., 2013) was used to determine the optimal task weights for each DOF that the model used in the residual reduction algorithm (RRA) within OpenSim. The optimal task weights and the adjusted mass properties of each segment were used within RRA to minimize the residual forces and moments applied to the pelvis segment and to closely replicate the experimental kinematics during the gait simulations. Residual forces and moments were normalized by bodyweight (%BW) and bodyweight multiplied by height (%BW*Ht), respectively. Muscle forces were estimated using computed muscle control (CMC), which computes the muscle excitations required to produce the forces that are necessary to accelerate each of the model's DOFs while accounting for muscle activation dynamics (Thelen and Anderson, 2006). Previously published guidelines in regards to RRA performance (Hicks et al., 2015) as well as a qualitative comparison of the CMC estimated muscle activation and EMG data were performed to assess the accuracy of the musculoskeletal simulations. Peak muscle forces (normalized to BW) of the gluteus maximus (GMAX), gluteus medius (GMED), gluteus minimus (GMIN), adductors (ADD: summation of adductor magnus, brevis, and longus), sartorius (SART), iliopsoas (summation of iliacus and psoas), piriformis, rectus femoris (RF), vasti (summation of vastus Lateralis, medialis and intermedius) and hamstrings (summation of biceps femoris short and long heads, semitendinosus and semimembranosus), during the stance phase of gait were assessed. The stance phase was defined as initial contact (vertical GRF exceeds 20 N) to toe-off (vertical GRF below 20N).

2.4. MRI Acquisition and Analysis

All FAI patients underwent an MR-exam of the symptomatic hip joint using a 3-Tesla MR-scanner (MR750, GE Healthcare, Waukesha, WI) and an 8 channel cardiac coil (GE Healthcare, Waukesha, WI). Each FAI patient was positioned supine in the MR-Scanner and secured with straps. In addition, each FAI patient's feet were secured to minimize any hip rotation during scanning. The MR-protocol included a combined $T_{1\rho}/T_2$ sequence used to assess cartilage composition (Li et al., 2014; Wyatt et al., 2015). For this study, acetabular and femoral cartilage $T_{1\rho}$ and T_2 relaxation times were estimated and used to provide an indirect measurement of the proteoglycan content and collagen structure, respectively, where an increase in $T_{1\rho}$ or T_2 relaxation times indicates an alteration in the proteoglycan content or collagen network within the articular cartilage. An atlas-based algorithm was used to

perform automatic segmentation of the acetabular and femoral cartilage segmentation and corresponding $T_{1\rho}$ and T_2 relaxation time estimation (Gallo et al., 2016). Acetabular and femoral segmentations were then divided into eight sub-regions (Karupppasamy et al., 2013), where sub-regions with less than 50 pixels over all segmented slices were not analyzed (Figure 1).

2.5. Statistical Analysis

Group differences in demographics, alpha and LCE angles as well as HOOS scores were assessed using independent t-tests. Peak muscle forces during stance were compared using a multivariate analysis of variance. Mann-Whitney U-tests were used to compare variables that were non-uniformly distributed. Partial correlation coefficients (R), adjusting for age, gender and BMI, were used to assess the associations between peak muscle forces, HOOS scores, $T_{1\rho}$ and T_2 relaxation times within the FAI group. All statistical analyses were performed using SPSS (v21, IBM Corp., Armonk, NY) and alpha was set a priori at the 0.05 level. In addition, a previously described voxel-based relaxometry technique (Pedoia et al., 2017), which implemented statistical parametric mapping, was used to visualize the correlation coefficients between peak muscle forces, $T_{1\rho}$ and T_2 relaxation times on a voxel-by-voxel basis within the FAI group.

3. Results

There were no differences in group demographics ($p>0.05$). The FAI patients exhibited higher alpha angles ($p<0.001$) and reported more severe hip joint pain ($p<0.001$) and dysfunction ($p<0.001$) compared to controls (Table 1). All musculoskeletal simulations closely tracked the experimental kinematic data with root mean square (RMS) positional differences of less than 1.1cm for pelvic translations, less than 0.60° for pelvic rotations and less than 1.45° for lower extremity joint angles. The RMS magnitudes of the residual forces and moments applied to all simulations fell below 0.66%BW and 0.52%BW*Ht, respectively. In addition, a good qualitative match was found between the EMG and CMC estimated muscle activations (Figure 2).

An overall effect of FAI was observed on muscles forces during gait (Wilk's $\lambda=0.59$, $F(10,37)=2.56$, $p=0.02$, partial $\eta^2=0.41$). FAI patients exhibited lower Vasti ($p=0.01$) and higher SART ($p=0.004$) forces during gait compared to controls (Table 2; Figure 3). Lower peak muscle forces within FAI patients were associated with worse HOOS pain and function sub-scores (Table 3). More specifically, lower GMIN, SART and Iliopsoas forces were associated with more severe hip joint pain ($R=0.44–0.63$, $p=0.002–0.05$), while lower GMED, GMIN, SART and Iliopsoas forces were associated with more severe hip joint dysfunction ($R=0.48–0.70$, $p=0–0.03$). Also, a trend was observed where lower GMED force was associated with worse hip joint pain ($R=0.40$, $p=0.08$).

An overall negative association between muscle force with $T_{1\rho}$ (Figure 4) and T_2 (Figure 5) values was observed within the FAI group. Within FAI patients, lower RF force was associated with higher $T_{1\rho}$ values within the anterior femoral cartilage (region 6; $R = -0.58$, $p=0.006$), while lower Iliopsoas force was associated with higher $T_{1\rho}$ values within the superomedial (regions 3 and 4; $R = -0.46 – -0.50$, $p=0.02–0.04$) and anterior (regions 6 and

7; R = −0.48 – −0.51, p=0.02–0.03) femoral cartilage. Lower GMED force within FAI patients was associated with higher T_2 values within the anterior femoral cartilage (region 6; R = −0.47, p=0.03), while lower Vasti force was associated with higher T_2 values within the posterior (region 2; R = −0.48, p=0.03) and anterior (region 6; R = −0.44, p=0.05) femoral cartilage. In addition, lower Vasti force was associated with higher T_2 values within the posterior (region 2; R = −0.45, p=0.04) and anterior-superior (region 5; R = −0.47, p=0.03) acetabulum, while lower Iliopsoas force was associated with higher T_2 values within the anterior-superior acetabulum (region 4; R = −0.45, p=0.04). Scatterplots of the statistically significant muscle force and average sub-regional $T_{1\rho}/T_2$ correlations are displayed in the supplementary material.

4. Discussion

When compared to the healthy asymptomatic controls, FAI patients ambulated with lower vasti and higher peak SART forces, suggesting potential multi-joint effects of FAI on muscle force production during walking. Within FAI patients, lower peak GMED, GMIN, SART, and Iliopsoas forces were associated with more severe hip joint pain and dysfunction. Also, lower peak Iliopsoas, GMED, RF and Vasti forces were associated with higher $T_{1\rho}$ and T_2 values of the hip joint cartilage, indicating a relationship between muscle force production and cartilage composition within FAI patients. Although it is not feasible to determine whether or not these altered muscle forces are compensatory mechanisms of the abnormal hip joint morphology present in FAI patients, the results of the current study provide novel information into the effects of FAI on hip muscle forces during gait and the potential link between muscle forces, patient reported outcomes and cartilage health.

Previous work has demonstrated that FAI patients possess lower isometric hip flexor strength (Casartelli et al., 2011; Kierkegaard et al., 2017) yet the FAI patients in the current study exhibited higher peak SART forces during the second half of stance compared to the CONT group. Previous studies have suggested that in order to reduce hip joint pain caused by excessive anterior hip joint loading, patients tend to ambulate with reduced hip extension (Lewis et al., 2010; Skalshoi et al., 2015). The FAI patients in the current study may be exhibiting this pain-avoidance mechanism by avoiding excessive hip extension and reducing anterior hip joint forces through higher SART force during the second half of stance. Despite the SART forces being larger (4%BW) in the FAI group compared to the CONT group, it is possible that a 4% difference may not be clinically significant. Future studies focusing on the function of the SART in FAI-related gait mechanics would provide insight into the clinical relevance of this muscle during late stance. In addition, FAI patients exhibited increased sagittal plane hip joint loading during the first half of stance (Samaan et al., 2017b), which may be due to potential alterations in distal joint mechanics. In the current study, FAI patients ambulated with lower peak Vasti force during the first 30% of the stance phase, which may suggest an inability to extend and stabilize at the knee joint, potentially placing a larger demand on the hip joint during loading response.

Direct associations between muscle forces and patient reported outcomes were observed within FAI patients. More specifically, FAI patients with lower peak GMED, GMIN, SART and Iliopsoas forces during gait reported worse hip joint pain and function. It is difficult to

thoroughly interpret the relationship between muscle forces and clinical symptoms due to the cross-sectional nature of the current study. It is possible that the FAI patients with hip joint pain have adopted a method to avoid high muscle force production in order to limit hip joint loading. On the other hand, it is also possible that FAI patients that exhibit lower muscle forces may be placing the hip joint at a higher risk of impingement during gait. More specifically, lower GMED and GMIN forces may lead to less hip abduction during gait, potentially causing hip joint impingement which may lead to increased hip joint pain.

The current study demonstrated a direct relationship between hip muscle forces during gait and articular cartilage composition within FAI patients. FAI patients that produced lower peak RF forces during gait exhibited higher $T_{1\rho}$ values within the anterior femur, while those FAI patients that ambulated with lower peak Iliopsoas forces demonstrated higher $T_{1\rho}$ values within the anterior and medial femur. FAI patients that produced lower GMED and Vasti forces exhibited higher T_2 values within the posterior and anterior femur. In addition, FAI patients that produced lower Vasti and Iliopsoas forces demonstrated higher T_2 values within the posterior and anterior-superior acetabulum. It can be suggested that lower RF, Vasti, GMED and Iliopsoas forces during gait may be detrimental to the cartilage health particularly within the anterior femur and acetabulum. Lower RF and Iliopsoas force leads to a more extended hip joint and in combination with lower Vasti forces (reduced ability to stabilize at the knee joint), these FAI patients may be excessively loading the anterior portion of the hip joint, which may lead to higher $T_{1\rho}$ and T_2 values. The Iliopsoas muscle was found to be a large contributor to the anterior hip joint contact force in normal walking patterns (Correa et al., 2010) and abnormal function of the Iliopsoas may lead to altered anterior hip joint contact forces, which may be detrimental to anterior hip joint cartilage health (regions 6– 7). In addition, previous work has demonstrated that FAI patients with more severe hip pain and dysfunction exhibited increased $T_{1\rho}$ and T_2 values within the anterior superior femoral cartilage layer (Grace et al., 2018). The FAI patients in the current study that produced lower peak Iliopsoas force reported worse hip joint pain, function and exhibited increased $T_{1\rho}$ and T_2 values within the anterior-superior femoral cartilage. Combining the results of the current study and those of Grace *et al.* (2018) and Correa *et al.* (2010), may suggest that the Iliopsoas is an important muscle to consider when assessing hip joint symptoms and cartilage health in FAI patients.

Previous work has shown increased $T_{1\rho}$ and T_2 cartilage heterogeneity within the anterior-superior acetabulum in FAI patients compared to asymptomatic controls (Samaan et al., 2017a). The relationship between RF, Vasti and Iliopsoas force with $T_{1\rho}$ and T_2 values within the anterior-superior acetabulum observed in the current study suggests a potential relationship between these three muscles and cartilage health within the anterior-superior acetabulum. More specifically, lower RF, Vasti and Iliopsoas force may lead to an overloading of the anterior superior acetabular cartilage potentially due to a reduced ability to flex the hip joint and may increase anterior hip joint loading, leading to increased $T_{1\rho}$ and T_2 values within the anterior-superior acetabulum. In addition, the Vasti are a substantial contributor to the superior hip joint contact force (Correa et al., 2010) and abnormal function of the Vasti may lead to abnormal superior hip joint contact forces in FAI patients, which may be detrimental to cartilage within the weight-bearing region (regions 3–5) of the hip joint. The overall pattern identified in this study is that FAI patients that exhibit higher RF,

Vasti, GMED and Iliopsoas forces during gait may have adopted a compensatory mechanism to avoid excessive hip joint loading, thereby reducing the forces experienced by the hip joint cartilage.

This exploratory study is not without its limitations and should be considered when interpreting the results of the current study. Future studies should incorporate a larger cohort size, more dynamic activity (i.e. squat) and be performed after hip-arthroscopy in order to assess the effects of surgical intervention on hip muscle forces. Although the associations between muscle forces, HOOS sub-scores and cartilage composition may suggest that overall lower extremity strength is important in FAI patients, we did not assess muscle strength to determine whether or not FAI patients in the current study presented with weaker musculature compared to healthy controls. In addition, a similar study incorporating self-selected walking speeds should be performed to assess the potential effects of speed on muscle forces in the FAI population.

In conclusion, the current study demonstrated that FAI patients ambulate with altered lower extremity muscle forces compared to healthy controls and that these muscle forces are directly associated with patient reported outcomes and cartilage health within FAI patients. The results of this study indicate that the RF, Vasti, GMED and Iliopsoas muscles are important to study in the FAI population as these muscles are associated with hip joint symptoms and cartilage composition. Lower extremity muscle strengthening may be important in the FAI population and should be highly considered during pre-surgical intervention protocols. In addition, a larger focus should be placed on late stance mechanics in the FAI population as the loading patterns experienced during this period of the gait cycle may be associated with more severe hip joint pain and dysfunction.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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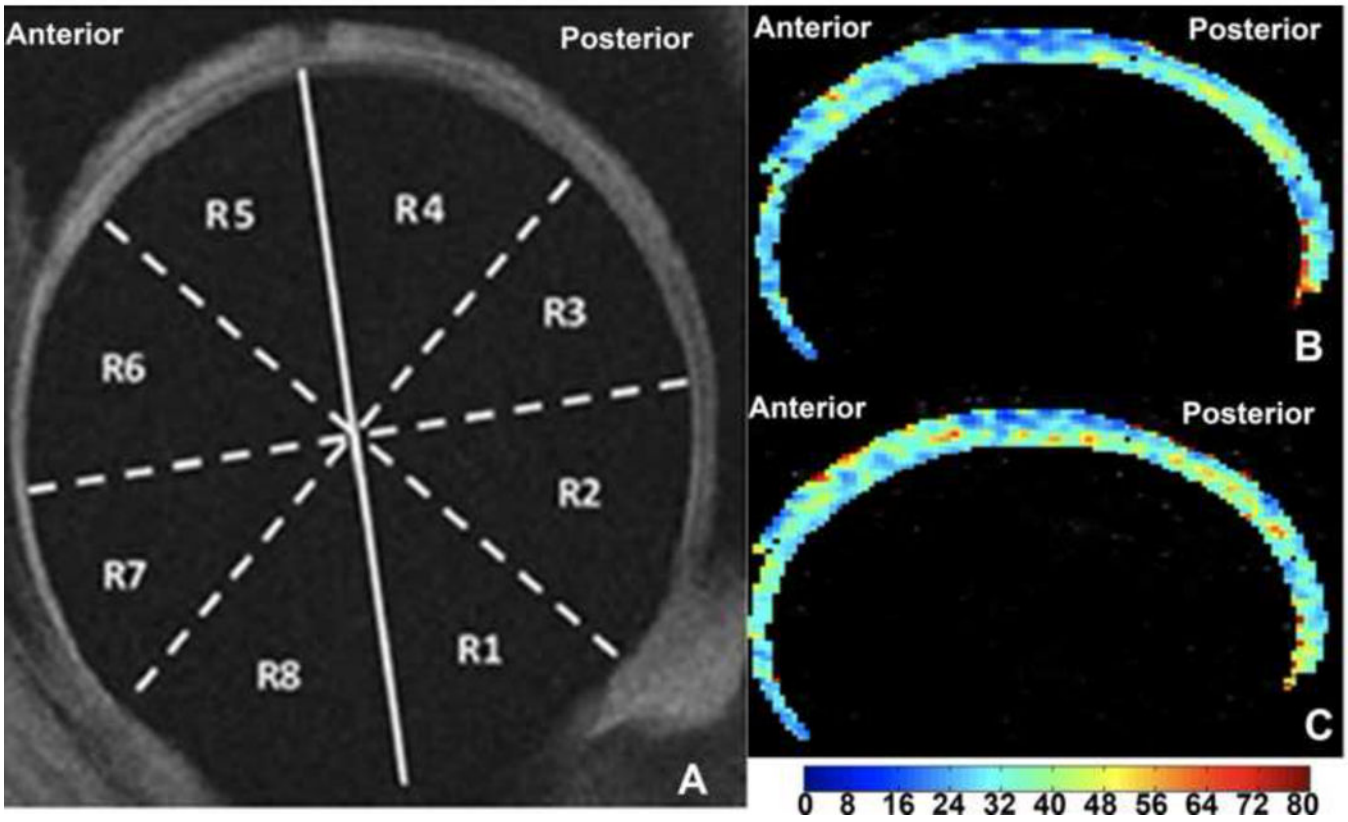

Figure 1:

The acetabular and femoral cartilage of the femoroacetabular impingement (FAI) patients were divided into 8 regions (R; Figure 1A). $T_{1\rho}$ (Figure 1B) and T_2 mapping (Figure 1C), measured in milliseconds, was performed within R2 – R5 of acetabular and R2 – R7 of the femoral cartilage.

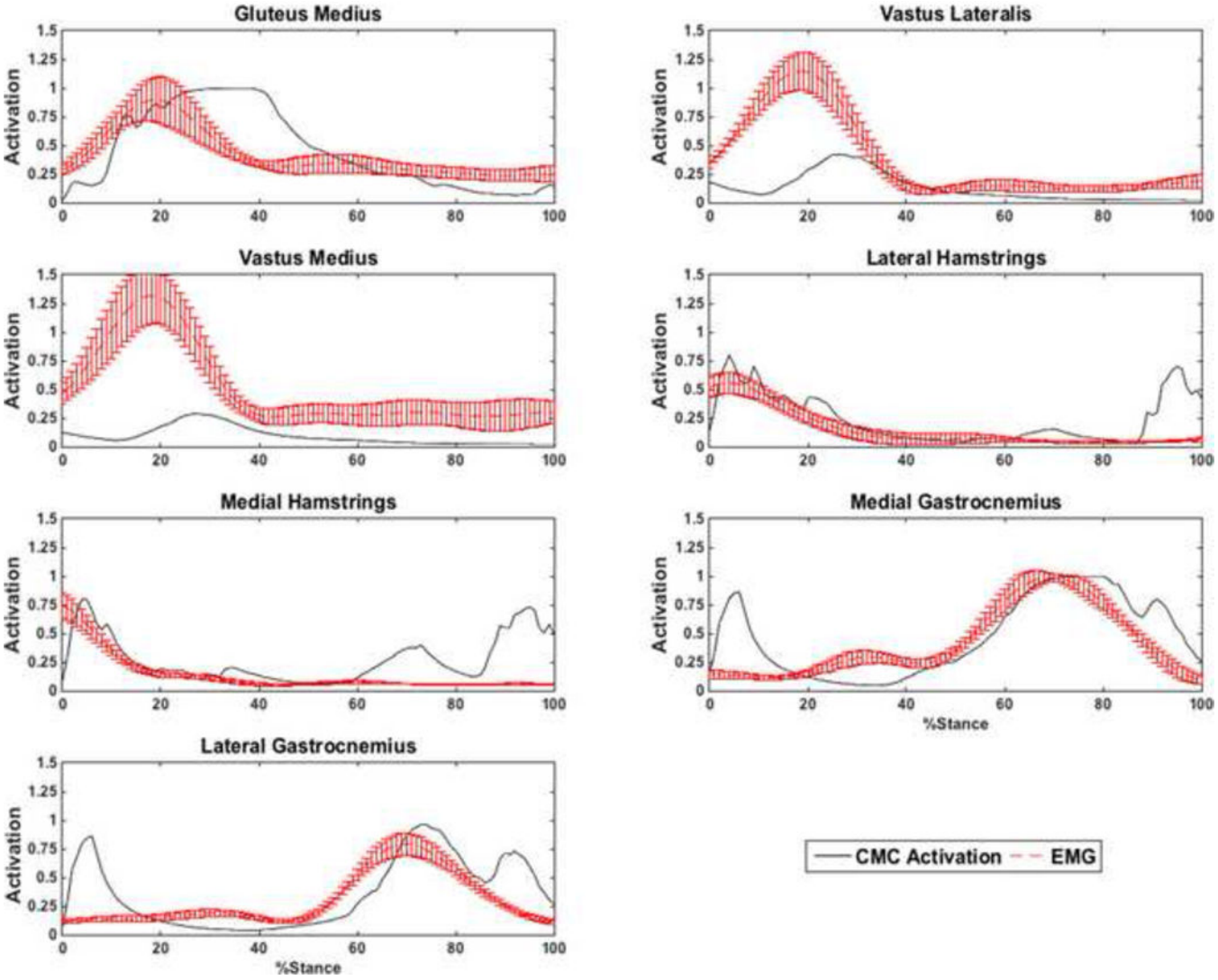

Figure 2.

Average electromyography (EMG) and computed muscle control (CMC) estimated muscle activations during the stance phase of gait for one representative study participant. EMG profiles represent ±1 standard deviation of the average EMG profiles for the one study participant.

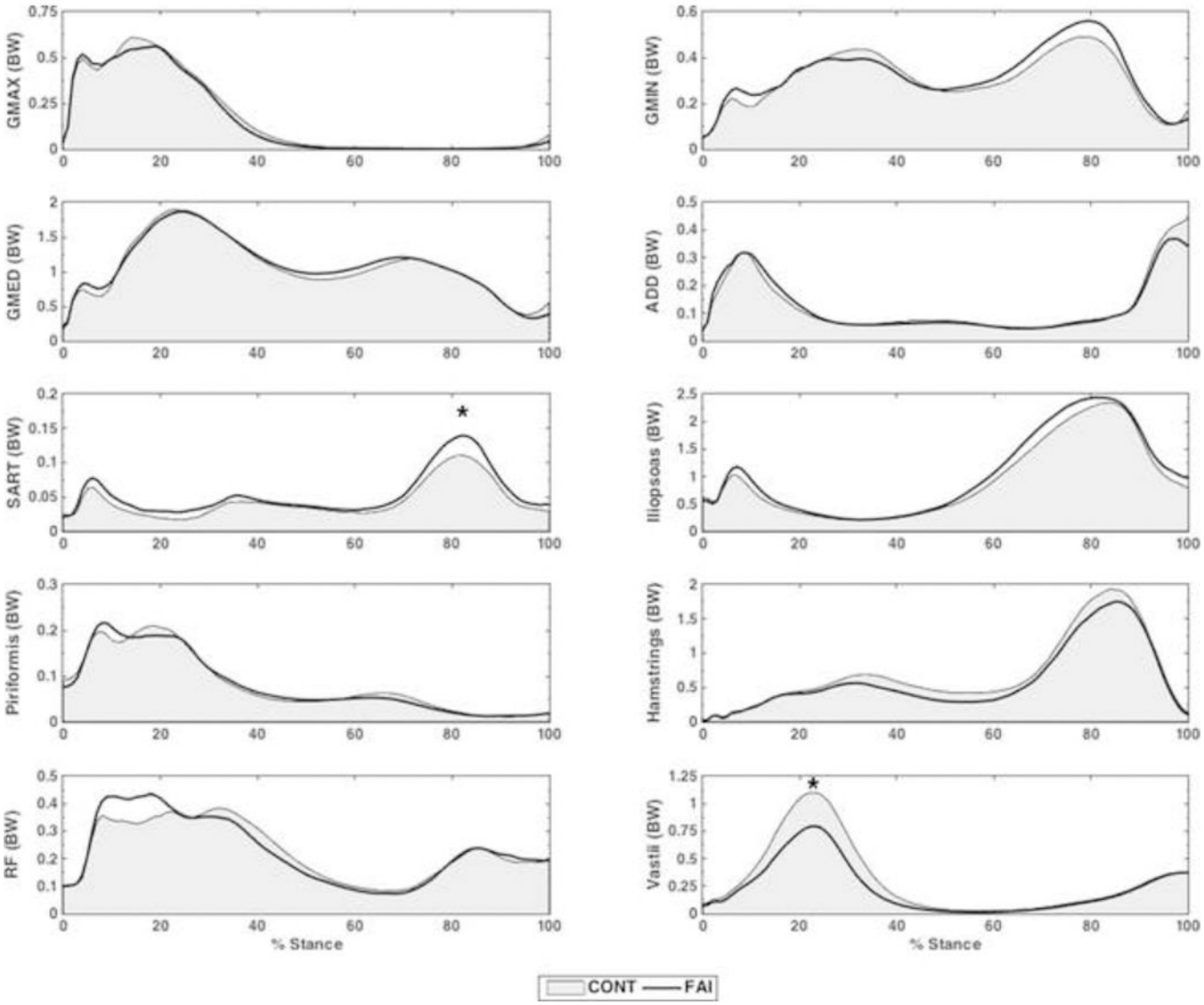

Figure 3.

Muscle force profiles, normalized by body weight (BW), during the stance phase of gait for the control (CONT) and femoroacetabular impingement (FAI) groups. Statistically significant differences are indicated with an *.

Abbreviations: Gluteus Maximus (GMAX), Gluteus Minimus (GMIN), Gluteus Medius (GMED), Adductors (ADD), Sartorius (SART), Rectus Femoris (RF)

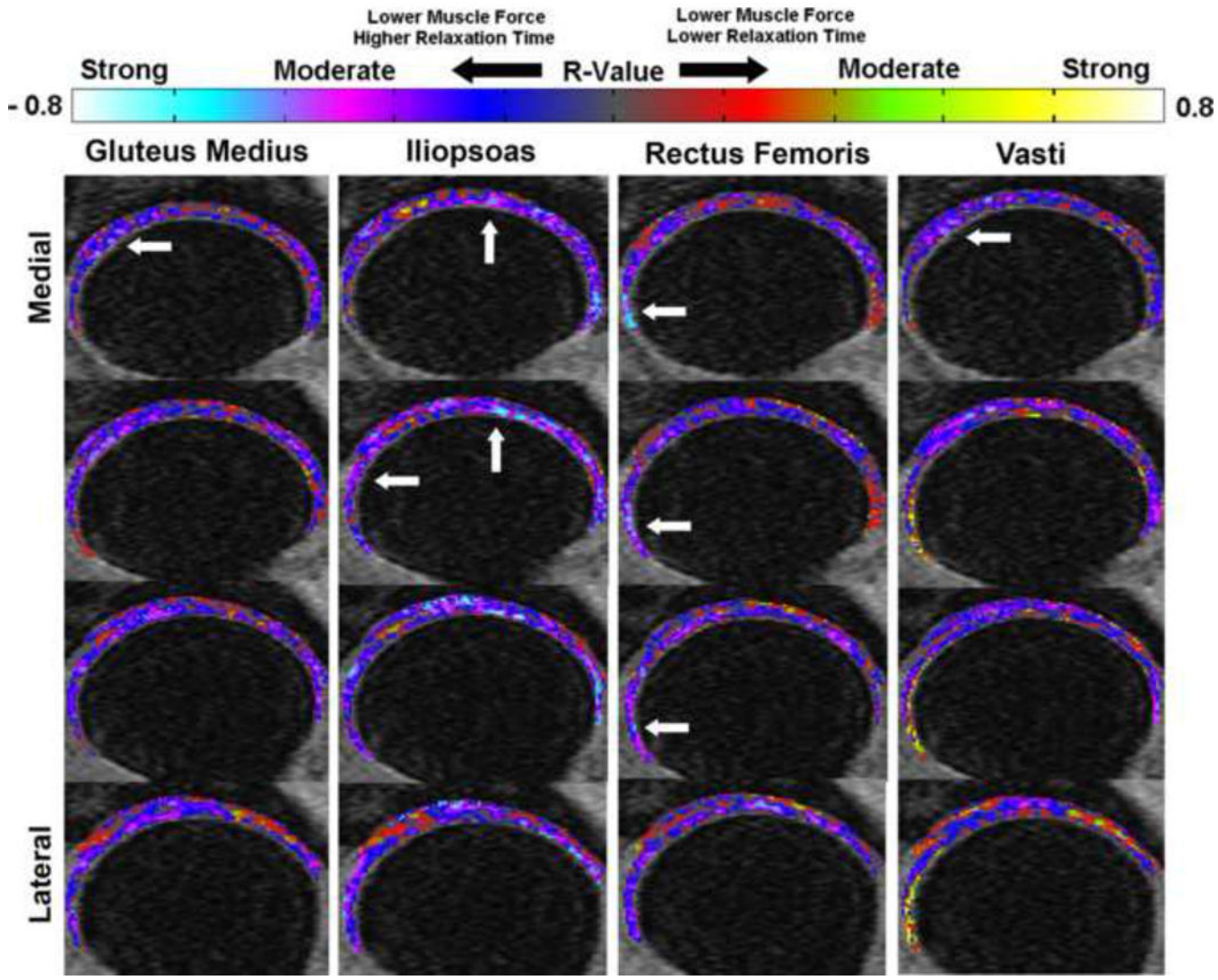

Figure 4:

Partial correlation coefficient maps between muscle forces, acetabular and femoral cartilage $T_{1\rho}$ relaxation times within femoroacetabular impingement patients for the Gluteus Medius, Iliopsoas, Rectus Femoris and Vasti muscles are displayed. White arrows indicate clusters of significantly correlated voxels within the hip joint cartilage.

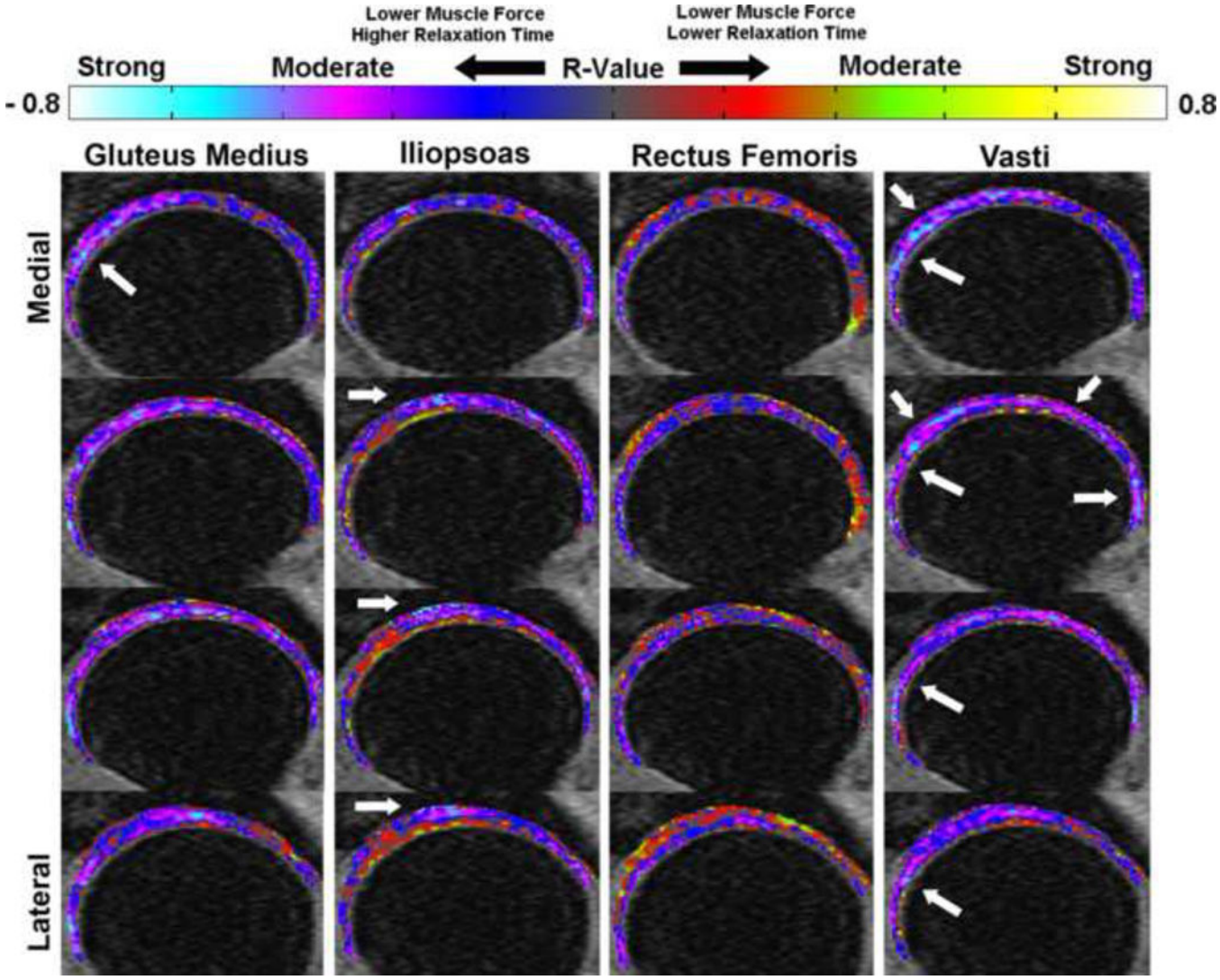

Figure 5:

Partial correlation coefficient maps between muscle forces, acetabular and femoral cartilage T_2 relaxation times within femoroacetabular impingement patients for the Gluteus Medius, Iliopsoas, Rectus Femoris and Vasti muscles are displayed. White arrows indicate clusters of significantly correlated voxels within the hip joint cartilage.

Table 1.

Group demographics and Hip disability and Osteoarthritis Outcome Scores (HOOS) for the control (CONT) participants, and femoroacetabular impingement (FAI) patients are presented as Mean±Standard Deviation. An * indicates a statistically significant difference between CONT and FAI ($p < 0.05$).

	CONT (N=24)	FAI (N=24)	p-value
Age (years)	42.0±18.1	35.6±8.54	0.12
Males:Females	14:10	14:10	1.0
Body Mass Index ($kg \cdot m^{-2}$)	24.1±3.28	24.9±3.70	0.43
Alpha Angle (°)	**47.7±11.4**	**61.5±5.1**	**<0.001***
Lateral Center Edge Angle (°)	31.1±8.6	32.7±6.1	0.46
Cam Type:Mixed Type	X	16:8	X
HOOS Pain	**98.7±3.36**	**63.3±17.3**	**<0.001***
HOOS Function	**99.4±2.76**	**63.3±19.7**	**<0.001***

Table 2.

Peak muscle forces during the stance phase of gait, normalized by body weight, for the control (CONT) and femoroacetabular impingement (FAI) groups are reported as Mean±Standard Deviation. An * indicates a statistically significant difference between CONT and FAI ($p < 0.05$).

Muscles	CONT	FAI	p-value
Gluteus Maximus	0.73±0.17	0.71±0.23	0.73
Gluteus Medius	1.98±0.27	1.97±0.37	0.87
Gluteus Minimus	0.59±0.17	0.65±0.16	0.16
Adductors	0.54±0.16	0.49±0.14	0.22
Sartorius	**0.13±0.05**	**0.17±0.04**	**0.006***
Iliopsoas	2.44±0.56	2.54±0.40	0.47
Piriformis	0.28±0.08	0.27±0.08	0.71
Hamstrings	1.89±0.32	1.98±0.32	0.32
Rectus Femoris	0.61±0.18	0.65±0.17	0.46
Vasti	**1.17±0.37**	**0.87±0.39**	**0.01***

Table 3.

Partial correlation coefficients (R) of muscle forces with Hip disability and Osteoarthritis Outcome Score (HOOS) for pain and function within femoroacetabular impingement patients. Statistically significant associations are denoted with an *.

Muscles	HOOS Pain	HOOS Function
Gluteus Maximus	R = −0.13/p = 0.59	R = −0.06/p = 0.81
Gluteus Medius	R = 0.40/p = 0.08	**R = 0.52/p = 0.02***
Gluteus Minimus	**R = 0.44/p = 0.05***	**R = 0.48/p = 0.03***
Adductors	R = −0.09/p = 0.71	R = 0.03/p = 0.91
Sartorius	**R = 0.63/p = 0.002***	**R = 0.70/p < 0.001***
Iliopsoas	**R = 0.55/p = 0.01***	**R = 0.53/p = 0.01***
Piriformis	R = 0.21/p = 0.37	R = 0.28/p = 0.23
Hamstrings	R = 0.27/p = 0.24	R = 0.32/p = 0.15
Rectus Femoris	R = 0.13/p = 0.57	R = 0.22/p = 0.35
Vasti	R = 0.13/p = 0.57	R = 0.19/p = 0.40