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Relationship between Physical Impairments and Movement Patterns During Gait in Patients With End-stage Hip Osteoarthritis

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Abstract

Patients with hip osteoarthritis demonstrate limited range of motion, muscle weakness and altered biomechanics; however, few studies have evaluated the relationships between physical impairments and movement asymmetries. The purpose of this study was to identify the physical impairments related to movement abnormalities in patients awaiting total hip arthroplasty. We hypothesized that muscle weakness and pain would be related to greater movement asymmetries. Fifty-six subjects who were awaiting total hip arthroplasty were enrolled. Pain was assessed using a 0 to 10 scale, range of motion was assessed with the Harris Hip Score and isometric hip abductor strength was tested using a hand-held dynamometer. Trunk, pelvis and hip angles and moments in the frontal and sagittal planes were measured during walking using three dimensional motion analysis. During gait, subjects had 3.49 degrees less peak hip flexion and 8.82 degrees less extension angles ($p < 0.001$) and had 0.03 Nm/k*m less hip abduction moment on the affected side ($p = 0.043$). Weaker hip muscles were related to greater pelvis ($r = -0.291$) and trunk ($r = -0.332$) rotations in the frontal plane. These findings suggest that hip weakness drives abnormal movement patterns at the pelvis and trunk in patients with hip osteoarthritis to a greater degree than hip pain.

Keywords

biomechanics; arthroplasty; trunk lean

INTRODUCTION

Hip osteoarthritis (OA) is a chronic disease that affects one in four people who live to the age of eighty-five ¹. This disease is characterized by joint pain, reduced range of motion, and muscular weakness ². Patients with hip OA also demonstrate abnormal movement

patterns. Biomechanical changes in the sagittal plane are characterized by reduced joint excursions and moments ³, while proximal changes in the frontal plane include increased trunk lean toward the affected limb and contralateral pelvic drop, in which the contralateral iliac crest moves inferiorly during stance on the affected limb. It has been suggested that these changes are compensatory strategies to reduce hip joint compression forces and pain by reducing the demand on the hip abductor muscles ⁴; however, it is also possible that these changes at the trunk and pelvis arise as a result of significant hip abductor muscle weakness ⁵.

Although compensatory movement patterns may reduce pain and hip muscular demand in the short term, they may put excessive stress on the contralateral joints and trunk. Identifying the underlying physical impairments that contribute to these altered movement strategies is essential to developing rehabilitation approaches that normalize movement patterns in patients with hip OA. To date, few studies have evaluated interlimb differences in frontal plane biomechanics in patients with hip OA and no studies have comprehensively evaluated factors that may contribute to these movement asymmetries in the sagittal and frontal plane. Therefore, the purpose of this study was to quantify biomechanical asymmetries in the sagittal and frontal planes at the hip, pelvis and trunk in patients with end-stage hip OA and to identify the underlying physical impairments that contribute to these biomechanical abnormalities. We hypothesized that patients with hip OA would ambulate with greater trunk lean and greater pelvic drop during the stance phase of gait on the more affected side. We also hypothesized that greater pain and weakness would be related to greater trunk lean, pelvic drop and hip adduction during gait.

METHODS

Subjects analyzed in this study were recruited from a larger longitudinal study aiming to quantify changes in function and biomechanical movement patterns before and after total hip arthroplasty. Subjects between 40 and 85 who were scheduled for total hip arthroplasty between March 2012 and May 2014 received a letter in the mail from their referring surgeons informing them about the study. Potential subjects were then screened for eligibility via a telephone interview with research staff. Subjects were excluded from the longitudinal parent study if they had: 1) neurological, vascular or other lower extremity musculoskeletal conditions that affected gait or functional performance, 2) self-reported lack of sensation in the foot or lower extremity, 3) uncontrolled hypertension, 4) history of cancer in the lower extremity, 5) were unable to walk short distances (10 m) without an assistive device, or 6) were moving within the next year. Additionally, only subjects with primarily unilateral hip OA were included in the current analysis. Therefore, subjects were also excluded from this analysis if they had 1) previous contralateral total hip arthroplasty; 2) plan for a future contralateral total hip arthroplasty; and 3) pain in the contralateral hip greater than 5 out of 10. All participants signed an informed consent form that was approved by the Human Subjects Review Board prior to participation in any portion of the study.

Functional ability of the subjects was assessed using previously validated measures, including the Harris Hip Score (HHS) and the Hip Outcome Survey Activity of Daily Living Scale (HOS) ⁶. Subjects also rated their overall functional ability on the Global Rating Scale

(GRS), which is a scale from 0 to 100, where 0 represents complete disability and 100 represents normal functional ability before they had hip pain. Pain was measured on a numerical scale from 0–10 where the subject responded to the question “Rate your average pain over the past week from 0 to 10, where 0 is no pain and 10 is the worst imaginable pain.” This question was asked for the affected hip and the un- or less-affected hip.

Hip abductor muscle strength was measured during an isometric hip abduction contraction using a hand-held dynamometer (Lafayette Manual Muscle Testing System; Model 01165; Instrument Company, Lafayette, IN) ⁷. Subjects were positioned in sidelying and a non-elastic strap was positioned around the subjects’ distal thigh to provide resistance. The hand-held dynamometer was positioned proximal to the lateral femoral condyles and its position was held constant between trials to avoid changes in the resistance moment arm. Subjects were asked to push into the strap (abduct their hip) as hard as possible. The maximal trial from 3 attempts was used as the maximal isometric contraction. This method has also been shown to be valid and reliable in older adults ⁷. Active-assisted range of motion (ROM) was measured for hip flexion, abduction, adduction, internal rotation and external rotation as part of the HHS using a goniometer. Subjects were asked to move their limb into end range and the investigator provided support and a slight overpressure. Hip flexion was measured supine with the knee flexed. Hip abduction and adduction were measured supine with the knee extended. Hip internal and external rotation were measured in a seated position ⁸. The total hip ROM was quantified as the sum of all individual range of motions measured in the HHS.

Eight infrared cameras (Vicon Motion Systems Ltd, Oxford, UK) were used to detect the position of retro-reflective markers at 120Hz through a collection volume that was approximately 1.2m wide, 1.5m long, and 2.3m high. Sixteen-millimeter spherical retro-reflective markers were placed bilaterally on the acromion process, iliac crest (aligned vertically with the greater trochanter), greater trochanter, lateral and medial femoral condyle, lateral and medial malleolus, head of the 1st and 5th metatarsal bone, and 2 markers on the heel. Rigid thermoplastic shells with 4 markers were secured bilaterally on the shank, thighs, and upper-back and were used to track the motion of these segments during the dynamic walking trials. Pelvic motion was tracked using a rigid thermoplastic shell with 3 markers placed below the line between the 2 posterior superior iliac spines. A standing calibration trial was taken to identify knee and ankle joint centers and create the segment coordinate systems. Functional hip joint centers were determined using a built-in algorithm that calculates the most likely intersection of all axes (effective joint center) and most likely orientation of the axes (effective joint axis) between the pelvis and femur based on a separate dynamic calibration trial in which subjects performed hip flexion, extension, abduction, and circumduction during single leg stance ⁹. Joint angles for the ankle, knee, hip, and trunk joints were calculated using Euler X-Y-Z sequence corresponding to sagittal, frontal, and transverse rotations sequence. Two force platforms (Bertec Corporation, Columbus, OH, USA) were embedded into the floor and recorded synchronous ground reaction forces at 1080Hz.

Subjects walked along a 10m walkway at their normal self-selected speed. Subjects were shod in their own footwear, but subjects were instructed not to wear sandals for the testing.

Self-selected speed was measured during 3 practice trials prior to data acquisition. Subjects completed 5 successful trials for each leg. A successful trial was defined as a walking trial that was within 5% of the initial self-selected speed in which at least one foot landed completely within the force plate area and there was no apparent targeting towards the force plate by the subject.

Visual3D software (C-Motion Inc., Germantown, MD, USA) was used for kinematic and inverse dynamic analyses. Kinematic and kinetic data were filtered at 6Hz and 40Hz, respectively using a second-order phase corrected Butterworth filter. Joint moments for the ankle, knee, and hip joints were calculated using inverse dynamics and were normalized by body mass (kg). Gait speed was measured using the “temporal distance” pipeline command of Visual3D. Peak vertical ground reaction force and frontal and sagittal plane hip angles and moments were calculated for each limb during the stance phase of gait. Peak hip flexion angle, peak hip extension angle, peak internal hip flexion and extension moments, and peak internal hip abduction moment were assessed on each limb. Frontal plane trunk movement was measured in two ways. First, peak trunk angle was defined as the maximum trunk angle towards the stance side during the stance phase of gait. This angle was the resultant angle between the trunk segment and the pelvis segment. Second, frontal plane trunk position in the lab coordinate system was measured to remove the effect of altered pelvis position on the trunk angle calculation. This was calculated as the trunk angle in the plane perpendicular to the walking direction and represents what is clinically referred to as trunk lean. Due to the nature of observational gait analysis, clinical observation of trunk lean may not delineate pelvic contribution. Trunk angle in the frontal plane, independent of pelvis position, was defined as lateral trunk lean for the purpose of this paper. Similar to peak trunk angle, peak lateral trunk lean in the lab coordinate system was calculated as the maximum trunk lean angle towards the stance side. Positive values indicate lateral trunk lean toward the stance side for both trunk angle and lateral trunk lean.

Pelvic drop was also measured two ways. First, the vertical motion of a marker on the contralateral iliac crest was measured throughout the stance cycle. Therefore, for the affected limb, the iliac crest height of the unaffected side was measured throughout the stance phase on the affected side. Because the contralateral iliac crest rises throughout the first 50% of the stance phase as the knee and hip extend in mid-stance, both the peak contralateral iliac crest height and excursion were used in this analysis. Excursion was calculated as the (maximum height – height at initial contact) of the iliac crest marker and was reported in meters. Second, pelvic drop was also evaluated by measuring the rotation of the pelvis in the lab coordinate system. This angle was calculated as the angle of the pelvis segment about an axis parallel to the direction of walking. This conveys information of the frontal plane pelvis rotation, irrespective of the position of the femur segment. This was defined in this paper as pelvis rotation. Negative values indicate rotation in which the contralateral iliac crest is depressed relative to the hip (akin to hip adduction). The minimum value of the iliac crest height during the first 50% of the stance phase and the pelvic rotation excursion were used for the analysis. Excursion was calculated as the absolute value of the (minimum rotation angle (trough) in the first 50% of stance - position at heel strike) and reported in the degrees.

Means, standard deviations and ranges were calculated for subject demographic variables and for discrete biomechanical variables. Differences between limbs for pain, strength and the biomechanical variables were assessed using paired t-tests. The relationships between biomechanical variables and pain, ROM and strength on the affected side were measured using bivariate correlation. Hierarchical linear regressions were created for biomechanical variables that showed significant correlations with measures of clinical impairment. Separate regression models were created for each biomechanical variable. The predictors of pain, strength and ROM were entered in that order in subsequent steps of the model. Significance of the change in R^2 value between each step of the regression model was calculated.

RESULTS

Fifty-six subjects (28 male/28 female; aged 65 ± 7.8 years) completed functional and motion analysis testing and were included in this analysis (Figure 1). On average, this subject pool was moderately impaired with self-reported scores on the HHS, HOS and GRS of 57.5, 55.5 and 56.6, respectively. The hip abductors on the affected limb were 28% weaker than the unaffected side ($p < 0.001$) and subjects reported significantly more pain in the affected limb ($p < 0.001$) (Table 1). On the affected side, subjects had 6% less vertical peak ground reaction force ($p < 0.001$) and significantly less peak hip flexion and peak hip extension angles and moments ($p < 0.001$) (Table 1; Figure 2). There was no difference between limbs for peak hip adduction angle ($p = 0.365$), but the affected limb had 7% less peak hip abduction moment ($p = 0.043$).

Peak frontal plane trunk angles (defined as the trunk segment movement relative to the pelvis segment) were 3.1 degrees greater during stance on the affected limb than during stance on the unaffected limb (Table 1; $p < 0.001$). There was no significant difference in peak lateral trunk lean (trunk motion in reference to the lab coordinate system; $p = 0.079$) (Figure 3). Pelvis rotation relative to the lab coordinate system had 1.8 degrees greater rotation ($p = 0.003$) and 0.6 degrees greater rotation excursion ($p = 0.019$) on the affected side. The pelvis rotation in the lab coordinate system was indicative of greater “pelvic drop” on the affected side (Figure 3). There was no significant difference between sides for the peak height or excursion of the contralateral iliac crest (Figure 3).

There were no significant correlations between pain and biomechanical variables during gait on the affected side (Table 2). However, hip abductor strength was significantly correlated with peak internal hip flexion moment and peak vertical ground reaction force. As strength increased, so did the flexion moment and ground reaction force. Strength was also significantly and negatively correlated with the peak and the excursion of pelvic rotation in the laboratory coordinate system and with the peak trunk rotation in the laboratory coordinate system. As strength decreased there was an increase in the pelvis rotation (towards hip adduction) and an increase in the lateral trunk lean toward the affected limb.

The regression models highlighted that pain did not significantly contribute to the variance in any biomechanical measure. Total hip ROM significantly predicted peak hip extension angle after accounting for pain and strength ($R^2 = 0.170$) and explained an additional 13.9% of the variance in the model for peak hip extension. Abductor strength predicted peak

internal hip flexion moment ($R^2 = 0.217$), ground reaction force ($R^2 = 0.173$), pelvis rotation ($R^2 = 0.110$), trunk rotation ($R^2 = 0.158$) and pelvis excursion ($R^2 = 0.114$), after accounting for pain. The addition of strength into the model explained an additional 7.1% to 20.8% of the variance for these gait biomechanical variables, although the addition of hip total ROM did not improve the models (Table 3).

DISCUSSION

Although trunk lean, pelvic drop and other frontal plane movement asymmetries are clinically described in patients with hip pathology, few studies have examined the magnitude and causes of these biomechanical abnormalities. We hypothesized that subjects with end-stage hip OA would ambulate with greater trunk lean and greater pelvic drop during the stance phase of gait on the affected side. The results of this study support this hypothesis. We also hypothesized that greater pain and weakness would be related to greater trunk lean, pelvic drop and hip adduction during gait. The results of this study partially support our hypothesis. Although pain was not related to any of the biomechanical variables, weaker subjects had greater movement asymmetries in the frontal plane.

Subjects in our study walked with reduced sagittal plane excursions and moments, most notably during terminal stance where there was a substantial difference between sides for peak hip extension angle (Figure 1). This is consistent with previous work that found sagittal plane asymmetries are typical of patients with hip OA^{3,10} and even persist after joint replacement¹¹. Interestingly, we did not find significant differences between sides with respect to hip adduction angle, although subjects did have significantly lower internal hip abduction moments on the affected side (Figure 1). The difference in the frontal plane kinetics may be a consequence of reduced vertical ground reaction force (Figure 1) or increased trunk lean and subsequent lateralization of the ground reaction force on the affected side (Figure 2).

We anticipated that individuals with weaker hip abductors would ambulate with reduced internal hip abduction moments. Contrary to this hypothesis, we did not find a relationship between hip strength and frontal plane hip moments. However, hip abductor weakness was a predictor of compensatory motions at the trunk and pelvis. Weaker individuals had greater pelvis rotation measured in the lab coordinate system, they also had significantly greater frontal plane trunk rotation, and a trend towards greater trunk lean towards the affected side. These proximal changes in the frontal plane support the theory that patients with hip weakness ambulate with a “Trendelenburg gait pattern” that is characterized by trunk lean towards the affected side and pelvic rotation, or pelvic drop, on the contralateral side. It should be noted that although we did see significantly greater pelvic rotation in the laboratory coordinate system, we did not see an evident drop measured using the height of the contralateral iliac crest. This measure of pelvic drop may be confounded by the rise of the pelvis segment and whole center of mass of the body during the first 50% of the stance cycle. This normal rise in the height of the pelvis may mask the small changes in pelvic drop. Measures of pelvic rotation in the frontal plane may provide a better understanding of how the pelvic segment compensates or responds to muscle weakness during unilateral stance.

Hip weakness is a substantial concern in patients with hip OA ¹². Patients with weaker hip abductor muscles have worse functional performance ¹³. Weakness may also put patients at risk for additional joint pathologies. Amaro et al. ¹⁴ found that subjects with hip OA who also had gluteus medius muscle atrophy had a greater risk of having radiographic signs of OA on the contralateral hip. It is possible that the asymmetric movement patterns and compensatory proximal movement strategies associated with hip weakness overload the contralateral limb and predispose these joints to OA. This would support biomechanical studies that found increased joint loading in contralateral joints in patients with hip OA ¹⁵ and the observational findings that OA progresses in a non-random fashion in which the contralateral cognate joint is most likely to be replaced next ¹⁶. Improving the strength of the hip abductors may offer a protective effect against asymmetrical movements and excessive contralateral forces, although longitudinal intervention studies are needed.

Although strength played an important role in frontal plane trunk and pelvis kinematics, pain was not related to any biomechanical variable. One previous study found that pain ¹⁷ was related to abnormal sagittal plane movement patterns. However, this study included relatively few patients (n=19) and pain was quantified using the scale from the HHS. The pain measure used in this study asked patients to quantify average pain over the last week. Pain in our cohort may be related to a variety of factors, other than walking. It is possible that if we asked patients to specifically rate their pain during walking, we may have found a relationship between pain and movement strategies. Similarly, although there were significant relationships between strength and frontal plane movements, strength explained a minority of the variance in these variables (R^2 values for strength ranged from 0.113 – 0.208; Table 2). We only assessed hip isometric abductor strength in a sidelying position with the lower leg positioned in 5 degrees of abduction. During gait, the abductors function over a range of joint positions and in a manner that is not likely solely isometric. Future work should quantify how different measures of muscle strength (power, eccentric, different hip positions) influence movement strategies.

Individuals with end-stage unilateral hip OA utilize compensatory proximal strategies and demonstrate abnormal movement patterns in the pelvis and hip in the frontal and sagittal planes. Hip abductor strength is related to movement asymmetries, including trunk lean and pelvis rotation that is consistent with clinical observations of pelvic drop. Although the effect of these biomechanical changes is not well understood, they may increase the demand on other lower extremity joints and place the individuals at greater risk for additional OA. Future studies evaluating interventions that address the underlying muscle impairments are warranted for older adults with hip OA.

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Figure 1.
Flowchart of subject recruitment and enrollment

Figure 2.

Subjects had less hip extension (A), less internal hip extension and flexion moments (C), and less internal hip abduction moment (D) on the affected side. There was no difference between limbs for hip adduction angle (B).

Figure 3.

Subjects had greater trunk angle towards the affected limb (A). When evaluated in the lab coordinate system, subjects had greater trunk lean toward the affected limb although this was not significant (B). Subjects had significantly different greater pelvis rotation that favored a more adducted hip position (C). There was no difference between limbs for the height or excursion of the contralateral iliac crest (D).

Table 1

Paired comparisons between affected and unaffected limb

	Affected Limb		Unaffected Limb		<i>p-value</i>
	Mean	Standard Dev	Mean	Standard Dev	
Peak hip flexion angle, °	24.5	6.55	27.99	8.73	<0.001
Peak hip extension angle, °	-0.9	9.2	-9.72	7.74	<0.001
Peak hip adduction angle, °	4.72	3.7	4.05	3.58	0.365
Peak internal hip flexion moment, Nm/kg*m	-0.36	0.13	-0.44	0.15	<0.001
Peak internal hip abduction moment, Nm/kg*m	-0.43	0.11	-0.46	0.09	0.043
Peak vertical ground reaction force, N/Kg	1.04	0.11	1.11	0.13	<0.001
Peak frontal plane trunk angle, °	5.69	3.75	2.58	3.9	0.002
Peak contralateral iliac crest height, m	1.053	0.06	1.055	0.06	0.410
Excursion contralateral iliac crest, m	0.028	0.011	0.033	0.029	0.150
Peak pelvis rotation in the lab coordinate system, °	-1.87	3.01	-0.06	2.87	0.019
Excursion pelvis rotation in the lab coordinate system, °	2.78	1.66	2.2	1.27	0.003
Peak lateral trunk lean in the lab coordinate system, °	4.41	4.24	2.69	3.57	0.079
Pain, 0-10	5.8	2.28	0.45	1.06	<0.001
Hip abductor strength, N/kg	0.146	0.081	0.203	0.08	<0.001

Table 2

Relationship between functional and biomechanical measures

	Affected Limb		
	<u>Pain</u>	<u>Strength</u>	<u>ROM</u>
Peak hip flexion angle	0.048	0.091	-0.069
Peak hip extension angle	0.015	-0.175	-0.396*
Peak hip adduction angle	0.049	-0.068	0.068
Peak internal hip flexion moment	0.091	-0.465**	-0.201
Peak internal hip abduction moment	-0.084	0.014	-0.061
Vertical Ground Reaction Force	-0.184	0.396*	0.178
Peak frontal plane trunk angle	-0.116	0.023	-0.091
Peak contralateral iliac crest height	-0.177	-0.078	0.092
Excursion contralateral iliac crest	0.089	0.151	0.140
Peak pelvis rotation in the lab coordinate system	0.195	-0.291 [†]	0.018
Excursion pelvis rotation in the lab coordinate system	0.072	-0.379 ^{††}	-0.077
Peak lateral trunk rotation in the lab coordinate system	-0.008	-0.332 [‡]	-0.021

*
p=0.003,**
p<0.001,[†]
p=0.031,^{††}
p=0.006,[‡]
p=0.015

Table 3

Regression model to predict biomechanical variables with the functional variables.

	R	R²	R² change	p-value change
Peak hip extension angle				
Pain	0.015	0.000	0.000	0.912
Pain + Strength	0.175	0.031	0.030	0.208
Pain + Strength + ROM	0.412	0.170	0.139	0.005*
Peak internal hip flexion moment				
Pain	0.091	0.008	0.008	0.507
Pain + Strength	0.465	0.217	0.208	<0.001*
Pain + Strength + ROM	0.477	0.228	0.011	0.394
Peak ground reaction force				
Pain	0.184	0.034	0.034	0.179
Pain + Strength	0.416	0.173	0.140	0.005*
Pain + Strength + ROM	0.424	0.180	0.007	0.519
Peak pelvis rotation (lab coordinate system)				
Pain	0.198	0.039	0.039	0.148
Pain + Strength	0.331	0.110	0.071	0.047*
Pain + Strength + ROM	0.346	0.068	0.010	0.457
Excursion pelvis rotation (lab coordinate system)				
Pain	0.173	0.030	0.030	0.207
Pain + Strength	0.398	0.158	0.129	0.007*
Pain + Strength + ROM	0.398	0.159	0.000	0.893
Peak lateral trunk lean (lab coordinate system)				
Pain	0.008	0.000	0.000	0.957
Pain + Strength	0.337	0.114	0.113	0.015*
Pain + Strength + ROM	0.339	0.115	0.002	0.763