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Simulated hip abductor strengthening reduces peak joint contact forces in patients with total hip arthroplasty

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ABSTRACT

Lower extremity muscle strength training is a focus of rehabilitation following total hip arthroplasty (THA). Strength of the hip abductor muscle group is a predictor of overall function following THA. The purpose of this study was to investigate the effects of hip abductor strengthening following rehabilitation on joint contact forces (JCFs) in the lower extremity and low back during a high demand step down task. Five THA patients performed lower extremity maximum isometric strength tests and a stair descent task. Patient-specific musculoskeletal models were created in OpenSim and maximum isometric strength parameters were scaled to reproduce measured pre-operative joint torques. A pre-operative forward dynamic simulation of each patient performing the stair descent was constructed using their corresponding patient-specific model to predict JCFs at the ankle, knee, hip, and low back. The hip abductor muscles were strengthened with clinically supported increases (0–30%) above pre-operative values in a probabilistic framework to predict the effects on peak JCFs (99% confidence bounds). Simulated hip abductor strengthening resulted in lower peak JCFs relative to pre-operative for all five patients at the hip (18.9–23.8 ± 16.5%) and knee (20.5–23.8 ± 11.2%). Four of the five patients had reductions at the ankle (7.1–8.5 ± 11.3%) and low back (3.5–7.0 ± 5.3%) with one patient demonstrating no change. The reduction in JCF at the hip joint and at joints other than the hip with hip abductor strengthening demonstrates the dynamic and mechanical interdependencies of the knee, hip and spine that can be targeted in early THA rehabilitation to improve overall patient function.

1. Introduction

Total hip arthroplasty (THA) is the most common surgery performed for patients with hip osteoarthritis (Daigle et al., 2012; Kurtz et al., 2005), which generally leads to improvement in overall physical function and high patient satisfaction (Jones and Pohar, 2012; Lau et al., 2012). However, after surgery, patients often do not attain full functional capacity (Fortin et al., 2002), with functional deficits remaining for years after surgery (Rasch et al., 2010). Rehabilitation following THA is designed to reduce these deficits and to optimize overall functional recovery. Lower extremity strength training is a common focus of rehabilitation because post-operative strength loss has been strongly associated with decreased overall function that inhibits the ability to comfortably perform tasks of daily living (Skoffer et al., 2015). Involved limb lower extremity strength gains from rehabilitation can range from 0 to 30% (Suetta et al., 2008), with more common gains of 15–20% (Judd et al., 2014). While strength deficits relative to the uninvolved limb may persist, early stage strength gains may be beneficial to long-term function and in reducing the loading experienced by the implant.

Targeting strength deficits in hip abductor muscles may improve the recovery of movement ability following surgery by influencing the loading at the hip joint and potentially at joints other than the hip. There is a growing body of literature demonstrating that interventions applied to one anatomical region of the body can influence the outcome and function of other regions of the body that may be seemingly unrelated to the applied intervention. This is a concept known as regional interdependence that

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has emerged primarily in the clinical literature (Sueki et al., 2013; Wainner et al., 2007). The strength of the hip abductor muscles is an important predictor of overall function following THA (Judd et al., 2014; Kamimura et al., 2014; Vaz et al., 1993). The hip abductors are made up of the gluteus maximus, gluteus medius, gluteus minimus, tensor fasciae latae, piriformis, and gemellus. Musculoskeletal simulations of gait have identified the hip abductor muscles as influencers of the joint contact force (JCF) at the hip and knee, where weakness in the hip abductors may result in greater hip JCFs (Valente et al., 2013). Weakness in the abductors results in increased demand on the flexor and extensor muscles to provide compensatory muscle force in positions and activities when they would not normally be active, which can result in greater contact forces compared to when hip abductor strength is healthy (Valente et al., 2013). Increased joint loading following THA can lead to loosening of the implanted components (Long et al., 1993) and progression of osteoarthritis in joints other than the hip resulting in overall functional deficits during tasks with high muscular demand (Griffin and Guilak, 2005).

The clinical relevance of the hip abductor muscle strength extends to both the knee and low back. Adequate strength in the hip abductor group has been associated with slower progression of knee osteoarthritis (Chang et al., 2005), reduced pain in patients with patellofemoral pain syndrome (Lee et al., 2012; Powers, 2010; Salsich and Long-Rossi, 2011), and lower incidence of low back pain (Nelson-Wong et al., 2008; Reiman et al., 2009). Additionally, three weeks of hip abductor strengthening in patients with patellofemoral pain syndrome resulted in strength gains of approximately 30% that altered frontal plane knee kinematics and decreased pain (Ferber et al., 2011). However, the relationship between hip abductor strength and JCFs in the lower extremity (ankle, knee, and hip) and low back has not been fully investigated, particularly during tasks with high muscle demand. Further, identifying which abductor muscles have the greatest impact on JCF can help direct rehabilitation strategies and inform surgical approach. Recent probabilistic musculoskeletal simulations have quantified the effect of variability in model parameters and identified the simulation inputs with the greatest impact on muscle and joint loading (Lamberto et al., 2017, Myers et al., 2014, Navacchia et al., 2016). Using probabilistic tools, we can also explore the effects of clinically important phenomena such as strengthening on quantitative musculoskeletal biomechanics, which are not connected well in biomechanical literature.

The purpose of this study was to investigate the effects of simulated increases in hip abductor strength following rehabilitation on JCFs in the lower extremity and low back during a step down task within a probabilistic framework. We hypothesized that simulated increases in abductor muscle strength would reduce peak hip JCFs and influence peak JCFs at joints other than the hip.

2. Methods

2.1. Patients

A cohort of five patients undergoing THA (2 M, 3F; age: 63 ± 7.5 yrs; BMI: 27.5 ± 2.0), performed through a posterior approach, were selected from a larger prospective study that included 26 patients (Judd et al., 2014). Patients were eligible if they were between the ages of 45 and 80 years old, had no history of uncontrolled hypertension or diabetes, body mass index <40 kg/m^2, and no additional orthopaedic pathology or neurologic disorders that impaired daily function. Each patient participated in a preoperative (Pre-op) experimental testing session and provided written, informed consent. The study was approved by the Colorado Multiple Institutional Review Board.

2.2. Experimental testing sessions

Isometric torque was measured to quantify strength of the hip flexors, extensors, and abductors, as well as the knee flexors and extensors using an electromechanical dynamometer (HUMAC NORM, CSMI Solutions, Stoughton, MA) connected to a Biopac Data Acquisition System (Biodex Medical Systems, Inc., Shirley, NY). Strength was measured in the affected limb. For hip flexor and extensor strength assessment, patients were positioned supine with the hip flexed to 40°. Hip abductor strength was measured while patients were positioned side-lying with 0° of hip flexion/extension and 0° of hip abduction/adduction. Knee extensor and flexor strength was measured in a seated position with a shoulder harness and waist strap for stabilization. Patients were placed in 85° of hip flexion and 60° of knee flexion for testing.

Patients were fitted with 32 reflective markers used to define anatomical landmarks for 3D motion capture. Following a standing static trial, patients were instructed to perform a single step down task leading with their involved limb from a step height of 20 cm onto a Bertec (Columbus, OH) force platform. The force platform was embedded beneath the step and collected force data at 2000 Hz. An 8 camera Vicon motion capture system (Centennial, CO) collected motion data at 100 Hz.

2.3. Musculoskeletal modeling

Muscle forces and JCFs were calculated using musculoskeletal modeling for each patient using OpenSim (Delp et al., 2007) in a two-stage approach. In the first stage, musculoskeletal models were calibrated to generate models with patient-specific muscle strength at the time of Pre-op testing. In the second stage, these models were used to calculate lower extremity and low back peak JCFs during the step down considering Pre-op measured strengths, and using a probabilistic framework (Myers et al., 2014) to assess the effect of simulated hip abductor strengthening on JCFs during the step down (Fig. 1). Hip JCF results from the Pre-op simulations were also compared to data collected from patients implanted with telemetric hip implants during step down (Bergmann et al., 2010).

2.3.1. Stage 1: Patient-specific strength scaling

Patient-specific lower extremity muscle strength calibration was performed using a musculoskeletal model that included detailed knee and hip musculature (Myers et al., 2018, Navacchia et al. 2016, Shelburne et al., 2010). Muscles and wrapping were added to a generic musculoskeletal model with 10 rigid bodies, 23 degrees of freedom, and 92 actuators (Arnold et al., 2000; Arnold and Delp, 2005; Delp et al., 2007, 1990). Analysis focused on muscles surrounding the hip that included: gluteus medius, gluteus maximus, gluteus minimus, rectus femoris, semimembranosus, semitendinosus, and tensor fasciae latae. The dimensions of the body segments, mass properties (mass and inertia tensor) of the segments, and the elements attached to the body segments, such as muscle actuators and wrapping objects were all scaled. In addition, for each patient-specific model, moment arms and maximum isometric torques were calculated for flexion/extension, internal/external rotation, and adduction/abduction of the hip.

Forward dynamic simulations of each patient performing maximum isometric hip abduction, extension, and flexion, as well as knee extension and flexion were generated in OpenSim using the joint position of the laboratory isometric tests and setting the abductor muscle activation level to 1.0 for the muscles involved in each task (Table 1) while setting all other muscle activations to 0. Patient-specific maximum isometric strength parameters of each hip muscle were increased or decreased to minimize differences between model-predicted and measured joint torques for each isometric task. Muscles in each group were scaled by the

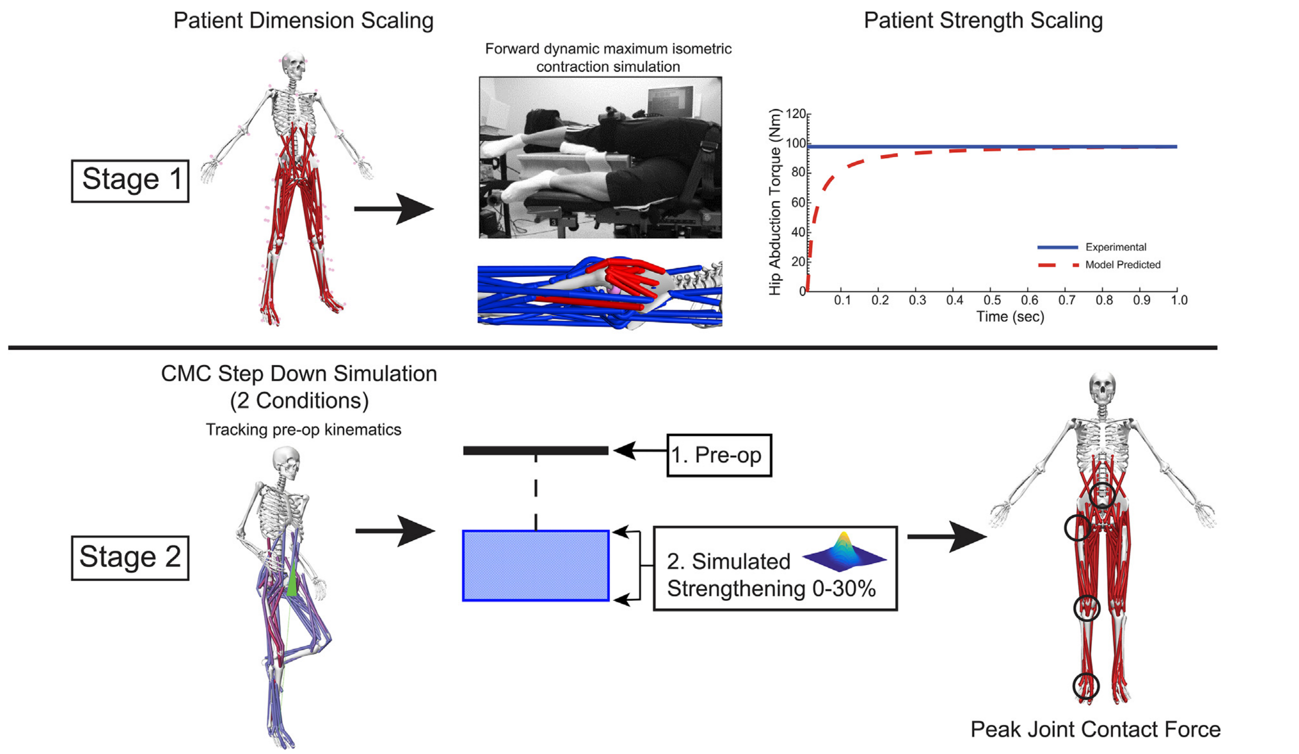

Fig. 1. Musculoskeletal simulation analysis was performed in two stages. In the first stage, patient-specific strength calibration was accomplished by first scaling models to patient mass and segment dimensions. The patient-specific muscle maximum isometric strength values for each model were calibrated to minimize differences between model-predicted and measured pre-operative maximum isometric joint torques in hip flexion, extension, and abduction, as well as knee flexion and extension. In the second stage, a pre-operative forward dynamic simulation of each patient performing the stair descent was constructed using their corresponding patient-specific model to predict JCFs at the ankle, knee, hip, and low back (graphically illustrated with a solid black line). The hip abductor muscle strength was increased relative to Pre-op in a probabilistic framework using the advanced mean value (AMV) method. A range of possible strength increases was simulated with a mean of 15% and a standard deviation of 5% to result in a ± 3 standard deviation range of 0–30% of possible increase in abductor muscle strength (upper and lower bounds graphically illustrated with a blue box). (For interpretation of the references to colour in figure legends, the reader is referred to the web version of this article.)

Table 1
The muscles that make up the abductor, extensor, and flexor groups of the hip and extensor and flexor group of the knee with abbreviations for each muscle. The abbreviations are consistent with those used in OpenSim.

Hip Abductors	Hip Extensors	Hip Flexors	Knee Extensors	Knee Flexors
Gluteus Maximus: 1 fascicle	Adductor Magnus:	Adductor Longus	Vastus Medialis (vas_med)	Biceps Femoris long Head (bifemlh)
Anterior (glut_max1)	3 fascicles	(add_long)	Vastus Lateralis (vas_lat)	Biceps Femoris Short Head (bifemsh)
Gluteus Medius:	Superior (add_mag1)	Iliacus	Vastus Intermedius	Semimembranosus (semimem)
3 fascicles	Middle (add_mag2)	Pectineus (pect)	(vas_int)	Semitendinosus (semiten)
Anterior (glut_med1)	Inferior (add_mag3)	Psoas		
Middle (glut_med2)	Gluteus Maximus: 2 fascicles	Rectus Femoris (rect_fem)		
Posterior (glut_med3)	Middle (glut_max2)	Sartorius (sar)		
Gluteus Minimus: 3 fascicles	Posterior (glut_max3)			
Anterior (glut_min1)	Gracilis			
Middle (glut_min2)	Quadratus femoris			
Posterior (glut_min3)	(quad_fem)			
Piriformis (piri)				
Tensor Fasciae Latae (tfl)				
Gemellus (gem)				

same factor to maintain strength ratios between muscles of the same group (see Table 2).

2.3.2. Stage 2: Step down task and simulated hip abductor strengthening

Using the model of each patient and their measured kinematics and ground reaction forces as input, a forward dynamic step down (Thelen and Anderson, 2006) was simulated with computed muscle control to predict Pre-op lower extremity muscle forces and JCFs at the ankle, knee, hip, and low back. Inverse kinematic total RMS was below 4 cm for all simulations with mean total RMS 2.76 ± 0.78 cm. Residual forces and moments were low with mean RMS values across five patients: Forces A/P = 19.7 ± 5.5 N; S/I = 19.0 ± 4.1 N; M/L = 21.7 ± N: Moments A/P = 15.2 ± 6.7Nm; S/I = 16.5 ± 7.1 Nm; M/L = 14.2 ± 3.2 Nm. In the simulations, the summed square of muscle stresses were minimized. JCFs were calculated using the joint reaction algorithm in OpenSim (Demers et al., 2015). Hip joint contact force results were compared to data collected from patients implanted with telemetric hip implants performing the step down (Bergmann et al., 2010).

Strengthening was simulated with a probabilistic approach by increasing the maximum isometric force parameter for each mus-

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Table 2
Anatomical data on the collected patients with THA.

Patient	Gender	Involved limb	Height (cm)	Mass (kg)	BMI (kg/m^2)	Age (yrs)
1	F	L	167.6	81.8	29.1	53
2	F	R	157.4	66.75	26.9	71
3	F	R	162.5	67.9	25.6	60
4	M	R	182.8	90.2	26.9	54
5	M	L	182.8	101.57	30.3	67
Avg			170.7	81.6	27.8	61.0
SD			11.7	14.8	1.9	7.9

cle of the hip abductor muscle group between 0 and 30% relative to the values determined in the Pre-op model. The abductor strength increase was based on the range measured in THA patients (Judd et al., 2014; Suetta et al., 2008) and characterized by a mean increase of 15% and a standard deviation of 5% (so that a ± 3 standard deviation range captured 0 to 30%). Probabilistic analyses to predict JCFs were performed using the advanced mean value method (AMV) (Wu et al., 1990). AMV is an approximation probabilistic method that offer a means to perform probabilistic studies with greater efficiency. Models that are highly complex can prohibit the use of sampling techniques, such as Monte Carlo and Latin hypercube, because of long model run times. While AMV is an approximation, it has been shown to be accurate in comparison to Monte Carlo (Laz and Browne, 2010).

AMV predicts performance for a specified probability level (Langenderfer et al., 2008; Pal et al., 2007) and can significantly reduce computational time compared to the Monte Carlo method when certain time points in the movement cycle are isolated. In this study, AMV was used to predict peak joint contact loading. To verify convergence, data from one patient were analyzed using both the AMV and Monte Carlo methods and joint contact force at the 0.5% and 99.5% probability levels were compared. After confirming agreement between probabilistic methods, the AMV method was used in the remainder of the simulations.

2.4. Data analysis

The range of peak JCF at each joint was generated by calculating the 99% confidence bounds (0.5% to 99.5% probability). These bounds represent the possible range without including the extreme tails of the output distribution. The predicted peak JCFs using each patient's Pre-op were compared to the predicted range from the probabilistic analyses.

Sensitivity factors from AMV were calculated for each muscle to assess the relative impact of increased strength of each muscle on the peak JCF. Sensitivity factors were calculated in the standard normal variate space as the unit vector from the origin to the point that represents the combination of input parameter values that predict performance at the two specified probability levels. The sum of squares of all sensitivities for each joint will equal one.

3. Results

After calibration to the isometric strength data, patient-specific models with corresponding applied kinematics established the Pre-op JCF and muscle forces for each patient. The predicted JCFs for the five patients in the current study compared well in magnitude and profile to direct measurements on a cohort of patients with telemetric implants (Bergmann et al., 2010) for the step down task (Fig. 2). Notable differences were present at the point of weight acceptance and toe off ranging from 17.4 ± 3.4% between groups, likely due to slight anthropometric differences in the groups. Maximum isometric strength data for each muscle group are reported in Table 3.

Joint contact forces from the AMV analysis closely matched those from a Monte Carlo simulation of 3000 trials. The 0.5% and 99.5% bounds calculated from AMV were on average 97.6% accurate for joint contact force estimations when compared to Monte Carlo. Computational time was approximately two orders of magnitude less for AMV (36 simulations) compared to Monte Carlo (3000 simulations).

Intersubject variation in the predicted JCFs were associated with measured Pre-op strength (Fig. 3), as predictions are influenced by anatomy, strength (Table 3) and kinematics of individual patients. Simulated hip abductor strengthening resulted in peak JCFs at lower and upper bounds that were smaller than Pre-op peak JCFs for all five patients at the hip (18.9–23.8 ± 16.5%) and knee (20.5–23.8 ± 11.2%) (Fig. 3). In general, simulations in the weakest patients resulted in the greatest reductions in JCF in response to simulated strengthening. For example, simulated strengthening in the three patients with the lowest Pre-op hip abductor strength (Patients 1, 3 and 5) resulted in peak hip JCFs that were on average 35.3% smaller than Pre-op peak JCFs. In addition, the impact of strengthening throughout the stance phase of the step-down task is illustrated in Fig. 4 for patient 2. These results indicate that when simulating the effects of strengthening, individual differences in Pre-op strength may have a pronounced influence on changes in joint loading.

Simulated strengthening resulted in reductions in JCF at the ankle (7.1–8.5 ± 11.3%) and low back (3.5–7.0 ± 5.3%) for four of the five patients, with one patient demonstrating no change. Reductions at the ankle and low back were smaller than at the hip and knee, but demonstrated the ability of the hip abductor group to influence loading at these joints in some patients (Fig. 5).

Reductions in JCFs during step down is accomplished by the resulting cascade of changes in muscle forces caused from simulated strengthening of the abductors. For example, a 15% strengthening of the gluteus medius resulted in a 16.7% decrease in peak gluteus maximus muscle force relative to Pre-op at the hip, while also causing an average 8.9% increase in peak quadriceps force and 25.8% decrease in peak hamstrings force at the knee (Fig. 6). Additionally, simulated strengthening lead to a redirection of JCFs at the lower extremity joints that was caused by these changes in muscle forces. This was demonstrated by changes to the force components at each joint as a result of simulated strengthening (Table 4) The largest differences occurred in the vertical component at each joint and accounted for 82.5 ± 13.1% of the JCF reductions.

The two posterior sections of the gluteus medius (glut_med2 and glut_med3) had a 20.3% greater effect on low back JCF than any other joint, while the anterior section (glut_med1) had a 46.3% greater effect on knee JCF than any other joint. The smaller muscles (tensor facia latae, gemellus) had the greatest influence overall for the relative increase in hip strength. Knee JCFs demonstrated sensitivity factors of 0.24 ± 0.8 and 0.26 ± 0.8 for the tensor facia latae and gemellus, respectively, and were the highest of any individual muscle-joint relationship. However, sensitivity factors varied between patients, likely due to differences in anthropome-

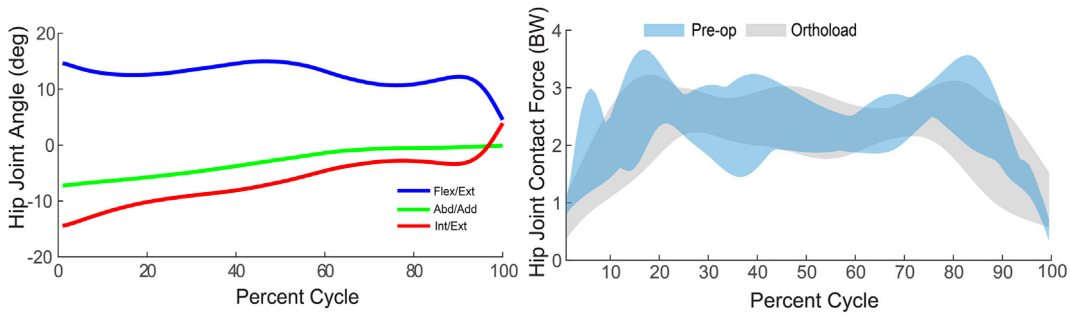

Fig. 2. Left: Representative Pre-op kinematics from one patient for the step down. Right: Comparisons between the magnitude of hip joint reaction force between the group of Orthoload patients (grey) and the Pre-op hip joint contact force for the group of five patients that participated in this study (blue). Each shaded region shown captures all of the patient data.

Table 3
Measured maximum isometric torque (N/kg) in each muscle group for all patients.

Patient	Quadriceps	Hamstrings	Flexors	Extensors	Abductors
1	1.40	0.43	0.91	0.33	0.81
2	1.70	0.76	0.73	0.78	0.96
3	1.08	0.42	0.86	0.73	0.85
4	2.69	1.09	1.70	0.77	1.56
5	1.42	0.50	0.78	1.04	0.61
Avg	1.66	0.64	1.00	0.73	0.96
SD	0.62	0.29	0.40	0.25	0.36

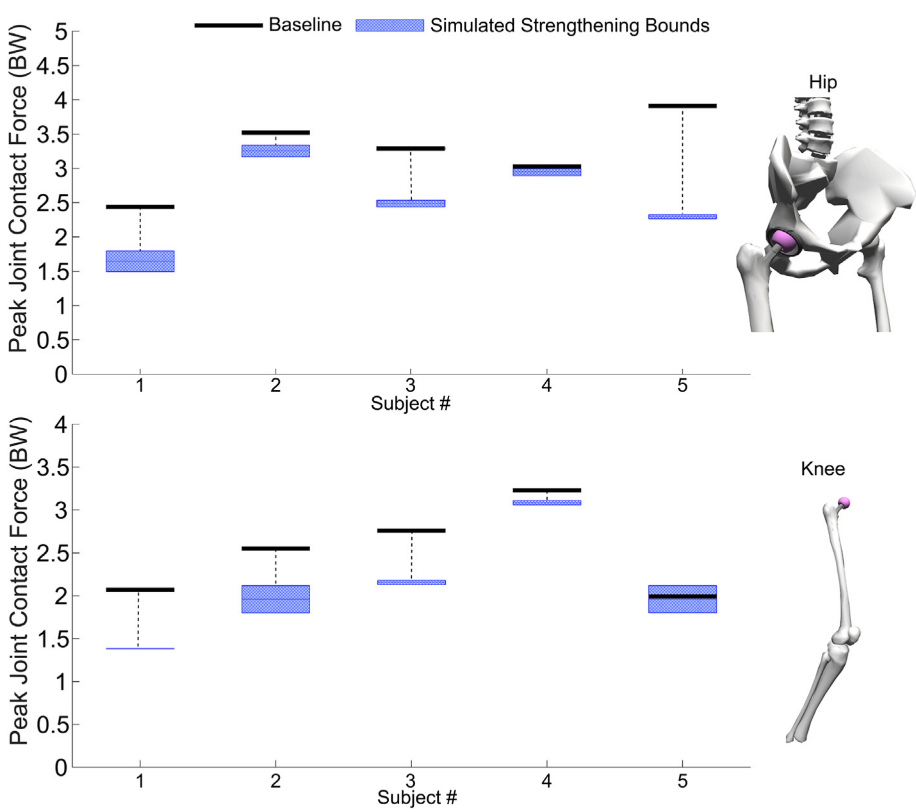

Fig. 3. Hip and knee joint contact forces (JCFs) during step down with pre-operative strength (black). Blue shaded regions indicate the upper and lower bounds from simulated hip abductor strengthening. Reductions in JCF resulting from strengthening were greatest for the weaker patients (patients 1, 3, 5).

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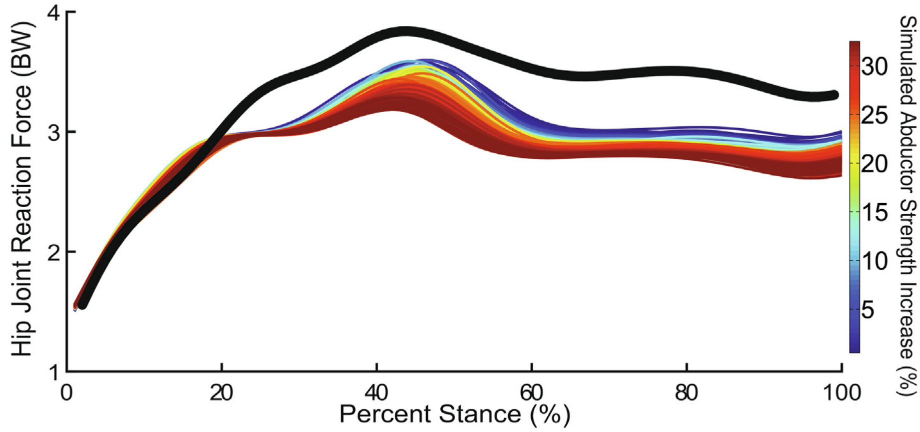

Fig. 4. Representative hip joint contact force from Patient 2 throughout the stance phase of the step down with variation caused by simulated strengthening of the hip abductor muscle group. The black line represents the pre-operative condition. Forces taken from a single patient following the Monte Carlo simulation used to establish AMV convergence.

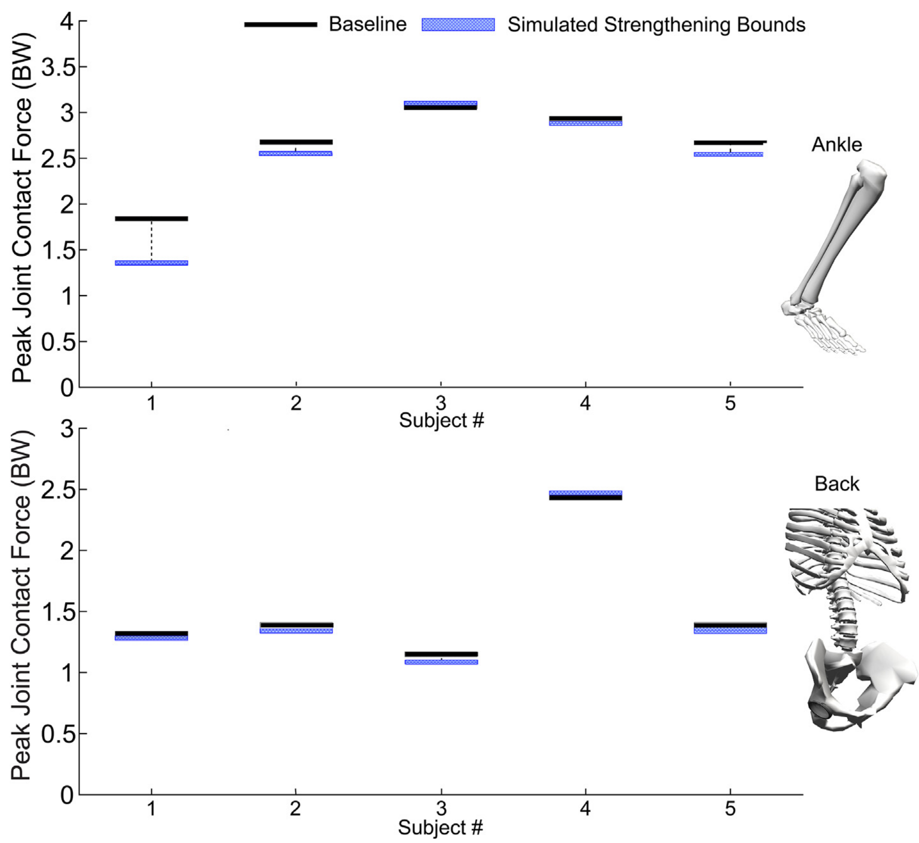

Fig. 5. Ankle and low back joint contact forces (JCFs) during step down with pre-operative strength (black). Blue shaded regions indicate the upper and lower bounds from simulated hip abductor strengthening. Reductions in JCF at the ankle and low back were smaller than at the hip and knee but were still apparent for four of the five patients.

try and stair descent kinematics that can influence moment arm and muscle mechanics (Fig. 7).

4. Discussion

Simulated strengthening of the hip abductor muscle group produced reductions in JCFs for all joints (ankle, hip, knee, and low back) during a high demand step down, which implies targeting the hip abductors in early THA rehabilitation may reduce loading on the implant and improve overall patient function. The reductions in JCF at the hip and knee were larger and more consistent than reductions at the ankle and low back, reinforcing regional interdependence. In addition, JCF was most sensitive to simulated strengthening in hip muscles that are often considered minor muscles, but may require more attention in both surgical approach and rehabilitation planning.

Strengthening of the hip abductor group is capable of reducing JCF during a step down at joints other than the hip, which confirms our initial hypothesis. While simulated strengthening of the hip abductors had the greatest influence on the hip JCF (18.9–23.8%), reductions in JCF ranging from 3.5% to 20.5% were also demon-

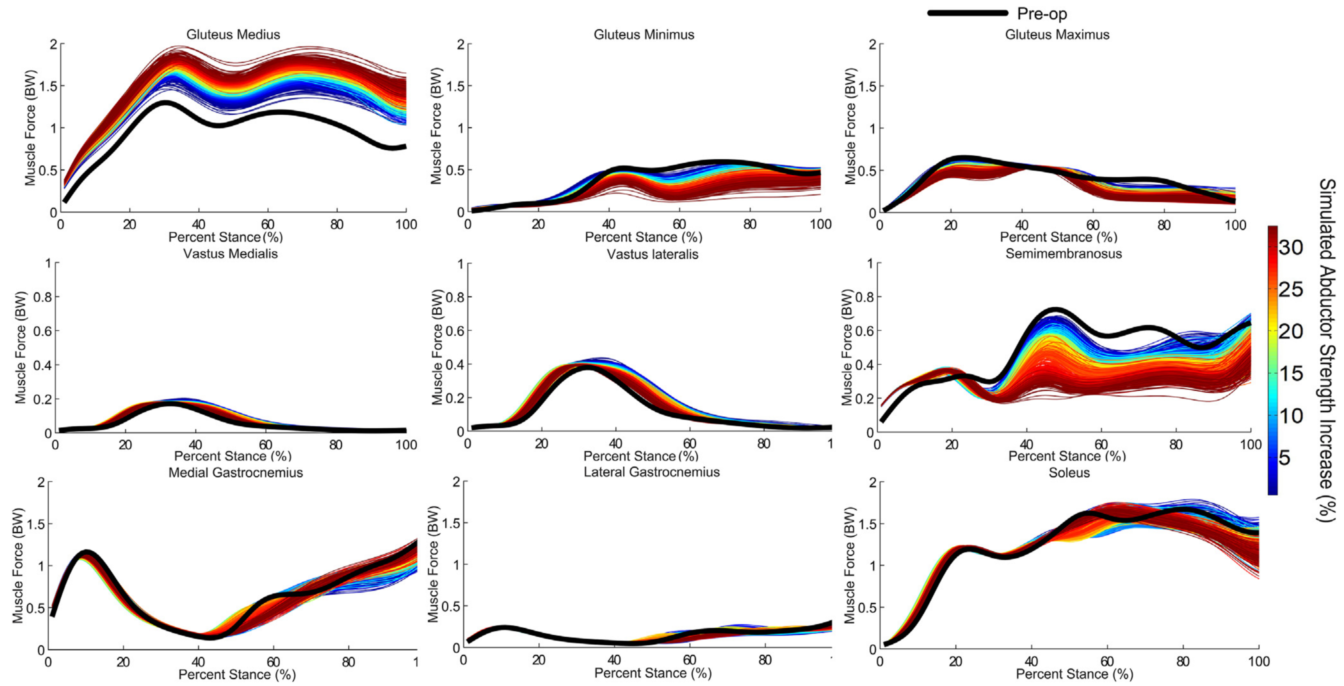

Fig. 6. Representative muscle forces from Patient 2 throughout the stance phase of the step down for selected muscles at the hip, knee, and ankle joint levels with variation caused by simulated strengthening of the hip abductor muscle group. The black line represents the pre-operative condition. Muscle forces taken from a single patient following the Monte Carlo simulation used to establish AMV convergence.

Table 4
Predicted mean (SD) joint contact forces in body weight for ankle (A), knee (K), hip (H), and low back (B) in anterior-posterior (x), vertical (y), and medial-lateral (z) components across 5 patients. Included is the difference between the lower and upper (L/U) probability levels.

	Ax	Ay	Az	||A||	Kx	Ky	Kz	||K||	Hx	Hy	Hz	||H||	Bx	By	Bz	||B||
Pre-op	−0.31	−3.36	−0.34	3.64	0.21	−3.42	−0.31	3.51	0.03	−3.30	0.81	3.43	0.08	1.56	0.04	1.56
	(0.31)	(0.68)	(0.38)	(1.05)	(0.77)	(1.35)	(0.26)	(1.36)	(0.50)	(0.34)	(0.31)	(0.38)	(0.08)	(0.60)	(0.06)	(0.60)
Lower	−0.33	−3.25	−0.30	3.53	0.22	−2.65	−0.22	2.84	0.19	−2.61	0.65	2.80	0.07	1.51	0.06	1.51
	(0.31)	(0.67)	(0.33)	(0.93)	(0.71)	(0.92)	(0.19)	(0.86)	(0.35)	(0.37)	(0.16)	(0.45)	(0.07)	(0.65)	(0.06)	(0.65)
Upper	−0.35	−3.28	−0.31	3.50	0.19	−2.76	−0.24	2.73	0.16	−2.71	0.59	2.70	0.06	1.46	0.05	1.46
	(0.31)	(0.67)	(0.33)	(0.93)	(0.69)	(0.82)	(0.19)	(0.96)	(0.35)	(0.44)	(0.15)	(0.40)	(0.07)	(0.66)	(0.06)	(0.66)
L/U Diff	0.02	0.03	0.01	0.04	0.03	0.11	0.02	0.11	0.04	0.10	0.05	0.10	0.01	0.05	0.01	0.05

strated in the low back, knee, and ankle. Increasing the strength alone of an important muscle group, while maintaining kinematics and anthropometrics, resulted in a redirection of contact forces and redistribution across muscles that lead to potentially beneficial force reductions. Biarticular muscles, which are capable of influencing multiple body segments, can explain a portion of transferring force from one segment to another; however, muscles important to the transfer of force among the body segments do not have to be biarticular (Zajac et al., 2002). It may also be due to dynamic coupling between the body segments where each muscle force contributes to the angular accelerations of all the joints at each instant of the task (Pandy, 2001; Zajac and Gordon, 1989). Because contact forces are influenced by angular accelerations of joints proximal and distal to the joint of interest, it follows that each muscle force also contributes to the contact force transmitted by each joint. For example, during gait, the vasti, soleus, and gastrocnemius contribute greater than 0.5 BW to hip contact force (Correa et al., 2010).

Patient-specific strength scaling in combination with sensitivity factor analysis provides clinical insight on beneficial muscles to target when designing strength-based rehabilitation strategy. Sensitivity factors demonstrated that all three sections of the gluteus medius were capable of influencing loading at the knee, hip, and low back. The most anterior section of the gluteus medius had the largest influence on the knee JCF, while the two posterior sections had a greater influence on the low back, which is a result of the architecture and moment arm of each section. Interestingly, knee JCF demonstrated the greatest sensitivity to the gemellus and tensor fascia latae, which may be considered minor muscles of the hip due to their size in comparison to the prime mover gluteal muscles. THA patients may rely on minor hip muscles to serve a compensatory role and account for a greater percentage of loading compared to healthy patients due to the muscle weakness that results following the surgery (Horstmann et al., 2013; Sutter et al., 2013). Sensitivity data such as this can be used when assessing the muscles which are most affected during various surgical approaches. Typically, the anterolateral approach affects muscle function of the gluteus minimus, gluteus medius, TFL, and vastus lateralis muscles, while the posterolateral approach affects the gluteus maximus, piriformis and gemellus (Madsen et al., 2004). With either approach, decisions to preserve and or repair a particular muscle are made on a case-by-case basis.

While this work simulated the effects of muscle strengthening on joint loading, similar observations have been reported in in-vitro studies. Using servo-hydraulic dynamic testing simulators, in-vitro studies have recreated the reaction forces and muscle

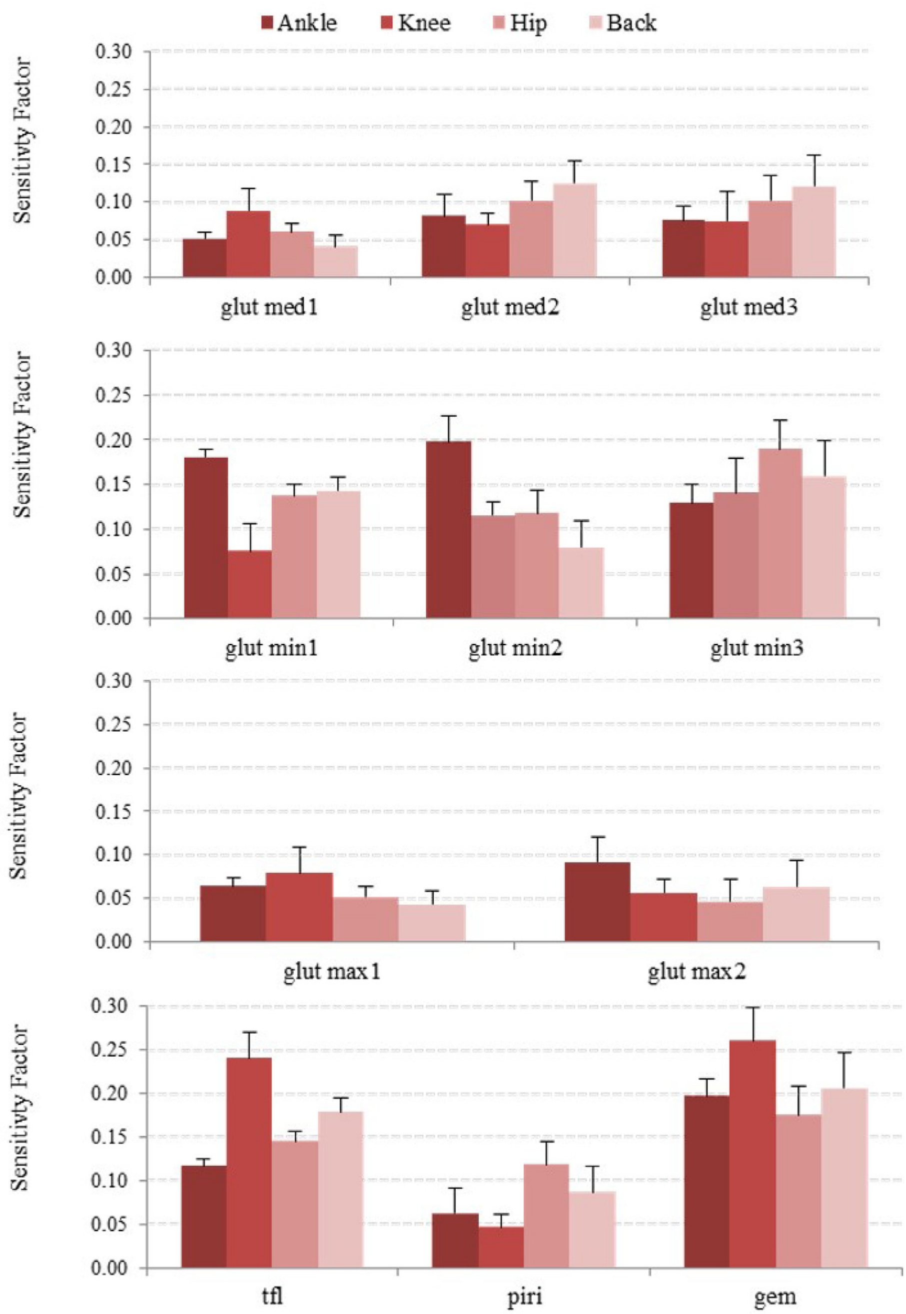

Fig. 7. Sensitivity factors for hip abductor muscles with respect to ankle, knee, hip, and low back joint contact forces. Averages are reported with error bars based on intersubject variability.

loads experienced at the knee during high-demand tasks, such as kneeling and landing (Abo-Alhol et al., 2014; Hashemi et al., 2010; Shalhoub and Maletsky, 2014). By simulating kinematics and applied loads during landing with cadaveric knees, while increasing quadriceps force over a physiological possible range, Hashemi et al. (2010) demonstrated a redirection of ground reaction forces and reductions in ACL strain. While not at the hip joint, this study provided quantitative evidence of the ability of increases in muscle forces to redirect contact forces during high-demand tasks. Cristofolini et al. (1995) simulated the forces of ten thigh muscles during early stance in gait on cadaveric femurs and found that the gluteus medius and minimus had over two times greater influence on vertical femur strain than the gluteus maximus, quadriceps muscles, and adductor magnus.

The current study implemented the AMV probabilistic method to consider variability in musculoskeletal simulation in an efficient and accurate way. The method has been utilized previously in structural, aerospace and recently biomechanics applications (Langenderfer et al., 2009, 2008; Laz and Browne, 2010). AMV has benefits in applications with high computational costs, like the forward simulations used in this study, because it requires fewer evaluations than Monte Carlo to generate similar results. However, the AMV method may not be appropriate in every musculoskeletal modeling application. The number of trials needed for AMV is dependent on the number of random variables and specified probability levels. As study complexity increases with increasing number of random variables and outputs of interest, computational savings is reduced and may be comparable to the robust Monte Carlo method. Additionally, when multiple combinations of input parameters result in the same output, the method may have difficulty with accuracy. Benchmarking AMV to Monte Carlo simulation is recommended in new applications.

There are limitations to this study that should be considered. First, this investigation assessed only the influence of increased muscle strength on JCF in isolation while leaving all parameters the same as the Pre-op condition. Following a strengthening rehabilitation protocol we would expect changes in kinematics, ground reaction forces, and other anthropometric variables that we cannot currently predict in this population with certainty and will be an interesting area for future investigation. Second, the simulated strengthening assumed that the maximum isometric strength of each muscle was independent. While it is not known how different muscles of the hip abductor group respond to typical strengthening rehabilitation, this approach enabled sensitivity factors for each muscle in the abductor group to be assessed. Finally, the maximum isometric tasks were performed in one position for flexion, extension, and abduction, and therefore, predicted maximum isometric torques at positions other than those tested may not be patient-specific. However, it has been shown that the shape of torque–angle relationship in the hip is consistent in flexion, extension, and abduction across each patient (Anderson et al., 2007).

In summary, simulated hip abductor strengthening produced reductions in JCF when muscle demand was high at the hip joint, as well as at the knee and low back. This is evidence of the dynamic and mechanical interdependencies of the knee, hip, and spine that can be targeted in early THA rehabilitation, potentially leading to higher overall patient function with reduced JCF on the implant. In addition, JCF was most sensitive to simulated strengthening in what may be considered minor muscles of the hip, which may play an important role in surgical approach and rehabilitation planning.

Declaration of Competing Interest

None of the authors had financial or personal conflicts of interest that could inappropriately influence this study.

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