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Effect of position and alteration in synergist muscle force contribution on hip forces when performing hip strengthening exercises

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Abstract

Background—Understanding the magnitude and direction of joint forces generated by hip strengthening exercises is essential for appropriate prescription and modification of these exercises. The purpose of this study was to evaluate hip joint forces created across a range of hip flexion and extension angles during two hip strengthening exercises: prone hip extension and supine hip flexion.

Methods—A musculoskeletal model was used to estimate hip joint forces during simulated prone hip extension and supine hip flexion under a control condition and two altered synergist muscle force conditions. Decreased strength or activation of specific muscle groups was simulated by decreasing the modeled maximum force values by 50%. For prone hip extension, the gluteal muscle strength was decreased in one condition and the hamstring muscle strength in the second condition. For supine hip flexion, the strength of the iliacus and psoas muscles was decreased in one condition, and the rectus femoris, tensor fascia lata, and sartorius muscles in the second condition.

Findings—The hip joint forces were affected by hip joint position and partially by alterations in muscle force contribution. For prone hip extension, the highest net resultant force occurred with the hip in extension and the gluteal muscles weakened. For supine hip flexion, the highest resultant forces occurred with the hip in extension and the iliacus and psoas muscles weakened.

Interpretation—Clinicians can use this information to select exercises to provide appropriate prescription and pathology-specific modification of exercise.

Keywords

hip joint force; hip pain; prone hip extension; straight leg raising

1. INTRODUCTION

Hip rehabilitation exercises are commonly prescribed to patients with joint pain and muscle imbalance or weakness. Knowledge of the magnitude and direction of joint forces generated during these exercises is essential for appropriate exercise prescription. For example, high joint forces are associated with the development of hip osteoarthritis (Mavcic et al., 2004). Therefore, clinicians should modify exercises to reduce the magnitude of the joint force in patients with or at risk for hip osteoarthritis. Modification may include changing the position in which the exercise is performed. Furthermore, it has been suggested that the musculoskeletal system is finely optimized to minimize stresses in bones and muscles and that any alteration in this system, such as muscle imbalance or weakness, may significantly increase the joint forces (Bergmann et al., 2004). Therefore, it is important to investigate how joint forces are affected by muscle weaknesses, especially during performance of common strengthening exercises. An improved understanding of these joint forces, including the direction of the force, is essential for appropriate prescription and pathology-specific modification of exercise, and may improve rehabilitation outcomes (Heller et al., 2001).

The purpose of this study was to use a musculoskeletal model to evaluate the hip joint forces created across a range of hip flexion and extension angles during two standard hip strengthening exercises: active hip extension in prone and active hip flexion in supine. As these exercises are normally strengthening exercises and alterations in muscle balance may affect joint forces, we also artificially induced weakness in our model by reducing the strength of synergist muscles to investigate the effect of changes in muscle force contribution on the hip joint forces.

2. METHODS

2.1. Musculoskeletal model

We used a 3-dimensional musculoskeletal model to estimate the hip joint force. A musculoskeletal model is a mathematical representation of bone and muscle, and illustrates how external forces (i.e. ground reaction force, gravity) and internal forces (i.e. muscle contraction, joint reaction forces) affect joint movement. Using a model allows us to artificially manipulate components of the model to test hypotheses. In this study, we manipulated the hip joint position in the sagittal plane and the maximum isometric force of specific muscles in order to test the effect of hip position and muscle force contribution on hip joint forces.

The musculoskeletal model we used was based on a bilateral model developed by Carhart to study the feasibility of utilizing functional neuromuscular stimulation to effect single-step compensatory movements in paraplegics (Carhart, 2003). This model does not take into account properties of the muscle-tendon unit nor forces due to passive response of the muscular tissue. As in another study (Lewis et al., 2007), we simplified Carhart's bilateral model to include only 4 segments: the pelvis, thigh, shank and foot of the right leg. The model contains 6 degrees of freedom (DOF) to represent the primary motions at the hip, knee and ankle as follows: (i) 3 DOF at the hip to model adduction-abduction, internal-external rotation and flexion-extension, (ii) 1 DOF at the knee to model flexion-extension, and (iii) 2 DOF at the ankle to model inversion-eversion and dorsiflexion-plantar flexion (Carhart, 2003). The definition of the kinematics of each joint was based on work by Delp (Delp, 1990).

Musculoskeletal parameters, including muscle path and maximum isometric force, were adapted from work by Delp (Delp, 1990) for the 43 muscle units included in the model. Delp subdivided large or complex muscles such as the gluteal muscles into multiple muscle units to more accurately represent their muscle paths and functions than would single muscle units. We modified the path of the iliacus and psoas muscles via an iterative process to be more consistent with the muscle moment arms as determined in a recent magnetic resonance imaging (MRI) study of their architecture (Arnold et al., 2000). We compared the muscle moment arms calculated by our model and found them to be in agreement with those calculated by SIMM (MusculoGraphics, Inc, Santa Rosa, CA, USA) for the published models (Arnold et al., 2000;Delp et al., 1990) from which the muscle data was obtained. The published models were validated previously by comparing the calculated muscle moment arms from their model with moment arms measured on magnetic resonance images (Arnold et al., 2000) and from cadavers and cross-sectional anatomy texts (Delp et al., 1990). We used Kane's Method (Kane and Levinson, 1985) and AUTOLEV 3.1 (OnLine Dynamics, Inc., Sunnyvale, CA, USA) to generate the equations of motion. In this study, because we were interested in the hip joint force only when the limb was held in a hip flexed or hip extended position, we simplified the equations of motion to include only the torques due to muscle force and gravity. Thus, the set of equations was simplified to:

$$\vec{\mathbf{T}}(\vec{\mathbf{Q}}) = -\vec{\mathbf{G}}(\vec{\mathbf{Q}})$$

(1)

In this equation, the position of the limb is defined by $\vec{\mathbf{Q}}$, which is a column vector of the six angles, one for each degree of freedom at each modeled joint. Similarly, $\vec{\mathbf{T}}$ is comprised of the net joint torques generated by the muscles at each degree of freedom. $\vec{\mathbf{G}}$ is comprised of the torques due to gravity at each degree of freedom, and is also affected by the position of the limb. Equation 1 indicates that the net torques due to muscle across all joints have to be equal and opposite the torques due to gravity. The torques due to gravity were estimated based on limb position, anthropometric parameters, and gravity (9.81 m/s^2). In a method similar to Yamaguchi and colleagues (Yamaguchi et al., 1995), we used an optimization routine (*fmincon* in MATLAB 6.5.1 (The MathWorks, Inc, Natick, MA, USA)) to solve for the percentage of maximal force contribution (P_{Force}) from each muscle to generate net muscle torques which were equal and opposite to the torques due to gravity. P_{Force} represents the level of force that the muscle is contributing as a percentage of the muscle's maximal force, and was constrained between 0% (no force) and 100% (maximal force). These constraints ensured that a muscle could not push (have a negative P_{Force}) nor exceed its maximum isometric force (P_{Force} greater than 100%). The optimization routine minimized the sum of the squared P_{Force} of the system. This routine is a scaled equivalent of minimizing muscle stress, which has the goal of maximum muscle endurance (Crowninshield and Brand, 1981).

In this study, we manipulated the maximum muscle force values for selected muscles in order to test the effect of decreased muscle strength. Manipulating the maximal muscle force also allows us to indirectly test the effect of decreased muscle activation.

Once the optimized P_{Force} for each muscle was solved simultaneously across all joints, the model estimated the total resulting force in the hip joint due to the muscles at their percentages of force. This net resultant force was also resolved into its three force components in the pelvic reference frame. The pelvic reference frame was defined by a vertical (superior/inferior) axis in line with the trunk when in a standing posture, a sagittal (anterior/posterior) axis perpendicular to the vertical axis and in line with movement in the anterior direction, and a transverse (lateral/medial) axis defined as the cross product of the other two axes. Forces were always calculated with regard to the pelvic reference frame, and from the perspective of the force of the femur on the acetabulum. For example, an "anterior force" indicates a force which

is imparted from the femur onto the acetabulum, and is in the anterior direction without regard for the position of the femur.

2.2. Exercises

The hip joint forces generated during the simulation of two hip exercises were evaluated. We selected these exercises as they both are often used as a rehabilitation exercise for patients with a variety of conditions (Hall and Brody, 2005;Moffat, 2006;Prentice and Voight, 2001) The first exercise simulated was prone hip extension. For the prone hip extension simulated exercise, gravity was specified as acting from posterior to anterior in line with the pelvic reference frame (Figure 1). The knee joint angle and ankle joint angles were set at zero so that both joints were in the neutral position and had to be maintained in neutral through a balance of muscle forces. The hip joint adduction/abduction and internal/external rotation angles were also set and maintained at zero. The hip joint angle was increased in one degree increments from 10° of hip flexion to 20° of hip extension. The hip joint angle range started at 10° of hip flexion because we recommend starting patients in 10° of hip flexion when performing prone hip extension in order to avoid hip hyperextension (Sahrmann, 2002).

The second exercise simulated was hip flexion in the supine position, or straight leg raising. For the supine hip flexion simulated exercise, gravity was specified as acting from anterior to posterior to simulate the supine position (Figure 1). Again, the knee and ankle joint angles as well as the hip adduction/abduction and internal/external rotation angles were set to zero. The hip joint angle was increased in one degree increments from 10° of hip extension to 30° of hip flexion. The range of simulated hip joint angles started at 10° of hip extension because this position is the presumed position of the hip when the lumbar spine is against the mat (Kendall et al., 1993) and is the typical starting position for a straight leg raise.

2.3. Conditions

We simulated 3 different conditions for each exercise to estimate the hip joint force when the maximum muscle force value for selected muscles was reduced. The first condition (Normal Condition) served as a control condition to which the other conditions were compared. For this condition, the hip joint force due to muscle was estimated using the maximum muscle force values specified in the original model (Delp, 1990). For the two altered conditions, the maximum muscle force value for selected muscles were reduced by 50%. This level of reduction has been used by other researchers (Goldberg and Neptune, 2007) and has been noted clinically in runners with overuse injuries (Niemuth et al., 2005). We induced weakness in the model to simulate the conditions under which these strengthening exercises are performed as it has been suggested that alterations in muscle force balance may increases joint forces (Bergmann et al., 2004).

For the prone hip extension exercise simulation, one altered condition was the Gluteal Reduced Condition in which the maximum force values of the gluteal muscles (gluteus maximus, medius, and minimus) were decreased by 50%. The second altered condition was the Hamstring Reduced Condition in which the maximum force values of the hamstring muscles (semimembranosis, semitendinosis, and biceps femoris) were decreased. We altered the maximum muscle force of the gluteal and hamstring muscle groups because both groups are major contributors to hip extensor torque and are strengthened by performing prone hip extension.

For the supine hip flexion exercise simulation, one altered condition was the Iliopsoas Reduced Condition, in which the maximum muscle force values for the iliacus and psoas muscles were decreased by 50%. These muscles were selected as they are primary hip flexor muscles and are strengthened by performing supine hip flexion. In the second condition, the Rectus Reduced

Condition, the maximum force values for the rectus femoris, tensor fascia lata (TFL) and sartorius muscles were decreased. These three muscles were selected because they contribute to hip flexion torque and may compensate for weakness of the iliacus and psoas muscles. For each altered condition, the optimization routine recalculated the optimal set of muscle forces with the altered constraint of the reduced maximum muscle force values.

2.4. Sensitivity Analysis

We also conducted an analysis to determine how sensitive the results of the study were to the minimization parameter used. We tested two additional minimization routines, one which minimized the sum of P_{Force} to the third power, and one which minimized the sum of P_{Force} to the fourth power.

2.5. Experimental Model Validation

We collected surface electromyographic (EMG) data from five subjects who provided written informed consent as approved by the University of Michigan Medical School Institutional Review Board. Surface electrodes (1.1 cm diameter) with an inter-electrode distance of 2.5 cm were placed over the muscle bellies of the medial hamstrings, lateral hamstring, gluteus maximus, gluteus medius, tensor fascia lata, rectus femoris, and iliopsoas muscles per guidelines (Konrad, 2005). The iliopsoas electrode was placed lateral to the femoral pulse, medial to the rectus femoris, and inferior to the inguinal ligament (Gottschall and Kram, 2005) and verified with movement tests (Cram and Kasman, 1998). Data was collected at 1200 Hz while the subject held the leg at the midpoint of the range of motion for each exercise as determined by a physical therapist. The midpoint for prone hip extension was approximately 5 degrees of hip extension while 10 degrees of flexion was used for supine hip flexion. The midpoint was used to avoid the influence of joint limitation or muscle length limitations at the end range of motion. Data were collected while each position was held for at least two seconds and repeated three times. Data were also collected during maximal voluntary contractions in standard manual muscle testing positions (Kendall et al., 1993). All emg data were high pass filtered with a 10 Hz zero-phase lag Butterworth filter, rectified and low pass filtered at 500 Hz. The root mean square (RMS) of the emg data was calculated using a window length of 12.5 msec. For each repetition of each exercise, we calculated the average RMS emg while the leg was held stationary. We then normalized the data using the peak RMS value during the maximum voluntary contraction for each muscle. Finally, we averaged the data across subjects for each muscle in each position to compare with the value estimated by the model.

2.6. Data analysis

The two focuses of this study were the effect of hip position and the effect of muscle force contribution on hip joint forces. To evaluate the effect of hip position, we calculated the net resultant force due to muscle at their level of P_{Force} throughout the range of hip angles for each simulated exercise. We also divided the resultant force into each force component to determine the effect of hip angle on the force components in each direction.

In order to evaluate the effect of altering muscle force contribution on hip joint forces, we calculated the joint forces for all three conditions. We also evaluated the change in P_{Force} for each muscle between the normal condition and the altered conditions to determine how the P_{Force} of each muscle in the model would be modified to compensate for the altered muscles. As a measure of the overall efficiency of the system of muscles, we calculated the sum of the P_{Force} values across all muscles which cross the hip or knee joints. The change in the P_{Force} and the sum of the P_{Force} values were all calculated at the neutral position (0°) of hi p flexion/ extension to allow comparisons between simulated exercises.

3. RESULTS

3.1. Prone hip extension

Effect of position—The net resultant hip force increases with increasing hip extension angle during simulated prone hip extension (Figure 2). The resultant force increased 181 N (22.7% body weight) with a 30° change in hip angle (10° of hip flexion to 20° of hip extension). The vertical force is the lar gest component of the hip force, followed by the force in the sagittal plane, and then in the transverse plane.

Effect of muscle force alteration—Decreased force contribution from the gluteal muscles when performing prone hip extension resulted in an increase in the net resultant and vertical hip joint forces compared to the normal condition. The magnitude of the sagittal force increased at the end ranges of hip motion. Compared to the normal condition, the hip force in the anterior direction was higher when in hip extension while the posterior hip force was higher when in hip flexion. Decreased force contribution from the gluteal muscles also decreased the transverse force component.

The overall muscle efficiency of the system was decreased when the maximum muscle force values for the gluteal muscles were decreased by 50%. At neutral hip flexion, the sum of the P_{Force} was 123, an increase of 47.2% over the normal condition which had a P_{Force} of 83.8.

Decreased force contribution from the hamstring muscles when performing prone hip extension resulted in a decrease in the resultant and vertical hip forces along with a decrease in the magnitude of the sagittal force compared to the normal condition. The transverse force was also slightly decreased.

When the maximum muscle force values for the hamstring muscles were decreased, the muscle efficiency of the system was also decreased compared to the normal condition. The sum of the P_{Force} was 122, an increase of 45.5% over the normal condition when in neutral hip extension.

In the prone hip extension simulation, when the gluteal muscle strength was reduced, the model increased the P_{Force} of the semimembranosis, gluteus medius (posterior fibers), TFL, and vasti muscles (Supplemental Figure 1). When the hamstring muscle strength was decreased, the gluteus maximus (all fibers), and the anterior portions of the gluteus medius and minimus demonstrated increased P_{Force}.

3.2. Supine hip flexion

Effect of position—The resultant hip force increased with increasing hip extension angle during simulated supine hip flexion (Figure 2). The resultant force increased by 195 N (24.5% body weight) with a 40° change in hip a ngle (30° of hip flexion to 10° of hip extension). Again, the vertical component of the force was the largest. The sagittal force became more anterior with increasing hip extension angle. The transverse force was less affected by position and only increased after 20° of hip flexion.

Effect of muscle force alteration—The effect of decreased force contribution from the iliacus and psoas muscles was dependent on hip position. The resultant and vertical forces were increased when in less than 11° of hip flexio n compared to the normal condition. The transverse force was also increased when in less than 7° of hip flexion compared to the normal condition. The sagittal force was consistently more anterior when the iliacus and psoas muscle forces were decreased.

The overall muscle efficiency of the system was significantly decreased when the maximum muscle force values for the iliacus and psoas were reduced by 50% compared to the normal

condition. At neutral hip extension, the sum of the P_{Force} was 382, an increase of 58.1% over normal which had a P_{Force} of 242.

Decreased force contribution from the rectus femoris, TFL, and sartorius muscles when performing supine hip flexion consistently resulted in an increase in the resultant, vertical and transverse forces when compared to the normal condition. The sagittal force was slightly more posterior in the Rectus Reduced Condition. The overall muscle efficiency was only slightly affected by reduced force contribution from the rectus femoris, TFL and sartorius muscles, increasing the sum of the P_{Force} only 6.8% above the normal condition.

In the supine hip flexion simulation, when the maximum force values for the iliacus and psoas muscles were reduced, the model increased the P_{Force} of a number of muscles: gluteus minimus (all fibers), adductor longus, pectineus, sartorius, TFL, and rectus femoris (Supplemental Figure 2). When the maximum force value of the rectus femoris, TFL, and sartorius were reduced, the model increased the P_{Force} of the gluteus minimus (all fibers), gluteus medius (posterior fibers), piriformis, iliacus and psoas muscles.

3.3. Sensitivity Analysis

In addition to the sum of the squared P_{Force} of the system, we tested two other minimization parameters (P_{Force} to the third and to the fourth power) to determine the sensitivity of the model to the optimization used. Although the different minimization parameters did result in slight changes in individual muscle stress and joint force values, the conclusions of the study were not affected by the minimization parameter used.

3.4 Experimental Model Validation

To validate the model, we compared the calculated P_{Force} from the model to the measured EMG data from our subjects. We found that the measured muscle activation during prone hip extension and supine hip flexion follows the pattern of the muscle force estimated by the model. For prone hip extension, the hamstring muscles are activated at the highest percentage of their maximum, with the gluteus maximus and medius activated at approximately half the activation of the hamstrings. The tensor fascia lata has the smallest activation of the muscles measured. The musculoskeletal model predicts this same pattern of activation (hamstring greater than gluteal and gluteal greater than tensor fascia lata). Similarly, both the experimental data and the model estimations during supine hip flexion display a pattern of greater activation of the iliopsoas than the rectus femoris or tensor fascia lata.

4. DISCUSSION

The results of this study indicate that hip joint forces are affected both by position and by alterations in synergist muscle force contribution, with position having a much greater effect. The highest forces are when the hip is in extension, and are greater during hip flexion in supine than hip extension in prone. The sensitivity analysis demonstrated that the conclusions of this study are not affected by the minimization parameters tested. Experimental data follow the relative contributions of muscles during prone hip extension and supine hip flexion.

4.1 Effect of position

The effect of hip joint position was varied in this study to determine if modifying the position of the exercise would significantly change the forces at the hip. We found that the net resultant hip force increased with increasing hip extension angle whether performing supine hip flexion or prone hip extension. Similar to the net resultant force, the vertical force was higher when in hip extension. These findings have implications for the prescription and modification of strengthening exercises for certain patient populations. For example, high joint forces may

contribute to the development of cartilage degeneration and osteoarthritis (Mavcic et al., 2004); therefore, when prescribing prone hip extension exercises for a patient with hip osteoarthritis, the hip should start in flexion and be limited to neutral extension to decrease the compressive force on the joint.

The direction of the sagittal force was also dependent on the position of the hip. The force was directed anteriorly when in hip extension, while it was directed posteriorly when in hip flexion. Again, this finding can be used to appropriately modify exercises. For a patient with anterior hip pain that is aggravated by anterior hip forces, supine hip flexion should not be initiated with the hip in extension.

4.2. Effect of alteration of muscle force contribution

Although the joint position affects the magnitude of the hip force more than alterations in muscle force contribution, the effect of muscle weakness or decreased activation is important to understand, especially for strengthening exercises. If the gluteal muscles are weak, it may be even more important to strengthen in neutral hip extension rather than full hip extension as weakness of the gluteal muscles increases the resultant force by 46.3 N at 20° degrees of hip extension compared to the normal condition. Limiting the total force may be important for a patient with osteoarthritis (Mavcic et al., 2004). Similarly, when prescribing supine hip flexion for a person with weakness of the iliacus and psoas muscles, the exercise should be initiated in greater hip flexion than if the muscles were not weak in order to reduce the anterior hip force.

Evaluating the effect of altered muscle force contribution also provides insight into the specificity of muscle function and the compensations for certain muscle weaknesses. Reduction of the maximum muscle strength of the gluteal muscles results in increased activation of the semimembranosis over the other hamstring (semitendinosis and biceps femoris) muscles. The semimembranosis muscle has the smallest rotational and adduction moment arms relative to the extension moment arm, making it a more efficient hip extensor than the other hamstring muscles (Supplemental Figure 3). The hip rotation and adduction moment arms of the semimembranosis muscle are 2.4% and 28.2% respectively of its hip extension moment arm. For the semitendinosis and biceps femoris muscles, the hip rotation moment arms are 0.3% and 9.9% and hip adduction moment arms are 43.8% and 38.8% of their hip extension moment arms.

In both prone hip extension and supine hip flexion, the increased P_{Force} of one muscle to compensate for weakness of another muscle results in extraneous torques or torques not directly related to the desired task. For example, during supine hip flexion, the TFL compensates for the hip flexion torque lost by the weakened iliacus and psoas muscles; however, it also creates extraneous abduction torque (Supplemental Figure 3). The extraneous abduction torque is 149% of its hip flexion torque. In other words, for each 100 units of hip flexion torque the TFL produces, it produces 149 units of abduction torque, making it a more efficient hip abductor than hip flexor. Other muscles, therefore, must contribute more force. To compensate for the extraneous abduction torque from the TFL, the P_{Force} of the adductor longer muscle is increased. The additional muscle forces result in an overall increase in the sum of the P_{Force} of the system of muscles.

Weakness of certain muscles affects the system differently. When the maximum muscle strength of the gluteal muscle group was reduced, the sum of the P_{Force} increased by 39.5 units over the normal condition. When the maximum muscle strength of the hamstring muscles were reduced, the sum increased by 38.2 units. It seems reasonable that the change in the overall P_{Force} would be similar for these two alterations as the total hip extension torque for the gluteal muscle group is similar to the hip extension torque for the hamstring muscle group (111.7 Nm

and 121.0 Nm respectively). However, during supine hip flexion, reduction of the maximal force the iliacus and psoas muscles resulted in an increase of 140.4 units over the normal condition while reduction of the rectus femoris, TFL and sartorius muscles resulted in only a 16.4 unit increase despite similar total hip flexion torques (28.0 Nm for iliacus and psoas combined and 38.8 Nm for rectus femoris, TFL and sartorius combined.) This extreme effect of weakness of the iliacus and psoas muscles emphases the unique role these muscles play in straight plane hip flexion.

Although other studies have investigated hip joint forces during gait and stairs (Bergmann et al., 2001;Heller et al., 2001;Stansfield and Nicol, 2002), and contact pressures during exercises (Tackson et al., 1997), our study is the first to evaluate the force components in each direction during performance of hip strengthening exercises. In this study, the net hip joint forces were divided into the component forces in the vertical, sagittal, and transverse directions relative to the pelvis. The individual forces in each direction are important when determining appropriate interventions for patients with particular pathologies. For example, abnormal or excessive forces in the anterior direction have been recognized as a potential cause of anterior hip pain and subtle hip instability (Shindle et al., 2006) and may lead to a tear of the anterior acetabular labrum even in the absence of a traumatic event (Mason, 2001;McCarthy et al., 2001;Shindle et al., 2006). Therefore, in patients with anterior hip pain and an anterior labral tear, it may be more important to limit anterior force than to limit the net resultant force.

As with all musculoskeletal models, there are limitations inherent in attempting to model complex human movement with simplified lines of action for muscles and computerized optimization routines for motor control. One major limitation specific to this study was our inability to accurately model the stabilizing effect, if any, of the iliacus and psoas muscles as they pass over the front of the femoral head. The iliacus and psoas muscles may be similar to the rotator cuff muscles of the shoulder, applying forces to the femur not only through their muscle insertion, but also through their muscle bellies and tendons as they pass over the underlying structures (Figure 3). It has been suggested that the psoas muscle tendon adds strength to the anterior hip joint capsule when in hip extension (Philippon, 2001), especially in the presence of capsular laxity (Shindle et al., 2006). While wrapping of the iliopsoas may be more accurately modeled in software such as OpenSim (https://simtk.org/home/opensim), our model uses via points to modify the path of the muscle around the femoral head.

The optimization criterion used to determine P_{Force} is another limitation. We used an optimization routine which minimized the sum of the squared P_{Force} over the system of muscles. Theoretically, this optimization function captures the physiological properties of muscle (muscle moment arm and maximum muscle strength), as well as the goal of maximum muscle endurance and produces solutions with significant temporal agreement between predicted muscle forces and recorded muscle electromyography (Crowninshield and Brand, 1981). A similar optimization routine has been used with this model in the prediction of muscle forces during gait and have resulted in temporally consistent and "intuitively reasonable" solutions (Carhart, 2003). We also tested two other minimization parameters (P_{Force} to the third and to the fourth power) and found that the conclusions of the study were not affected. This is in agreement with the study by Crowninshield and Brand (Crowninshield and Brand, 1981), which demonstrated that calculated muscle stress using minimization of the sum of muscle stresses to the nth power was not sensitive to small changes in n. Furthermore, the pattern of measured muscle activation was similar to the muscle forces predicted by the model. We also modeled only static behavior. Because both of the exercises used in this study are typically performed at a slow velocity (30° per second) and within the normal range of motion, we did not include torques from inertia, passive joint structures or viscoelastic damping.

Another limitation is in our choice of exercises. Although prone hip extension and supine hip flexion are not functional tasks, they are commonly prescribed exercises. The joint forces created when performing these exercises should be understood so that appropriate modifications can be made. Furthermore, analysis of these simple exercises may provide insight into the forces that muscles create when active in other tasks such as walking.

5. CONCLUSION

In this study, the musculoskeletal model predicted that hip joint forces are greatly affected by the angle of hip flexion or extension and partially by alterations in muscle force contribution. During prone hip extension, the highest net resultant forces occurred when the hip was in extension and when the gluteal muscles were weakened. During supine hip flexion, the highest net resultant forces occurred when the hip was in extension and when the iliacus and psoas muscles were weakened. Clinicians can use this information to modify exercises to provide the most appropriate intervention for a patient based on the specific location of hip joint pathology.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgements

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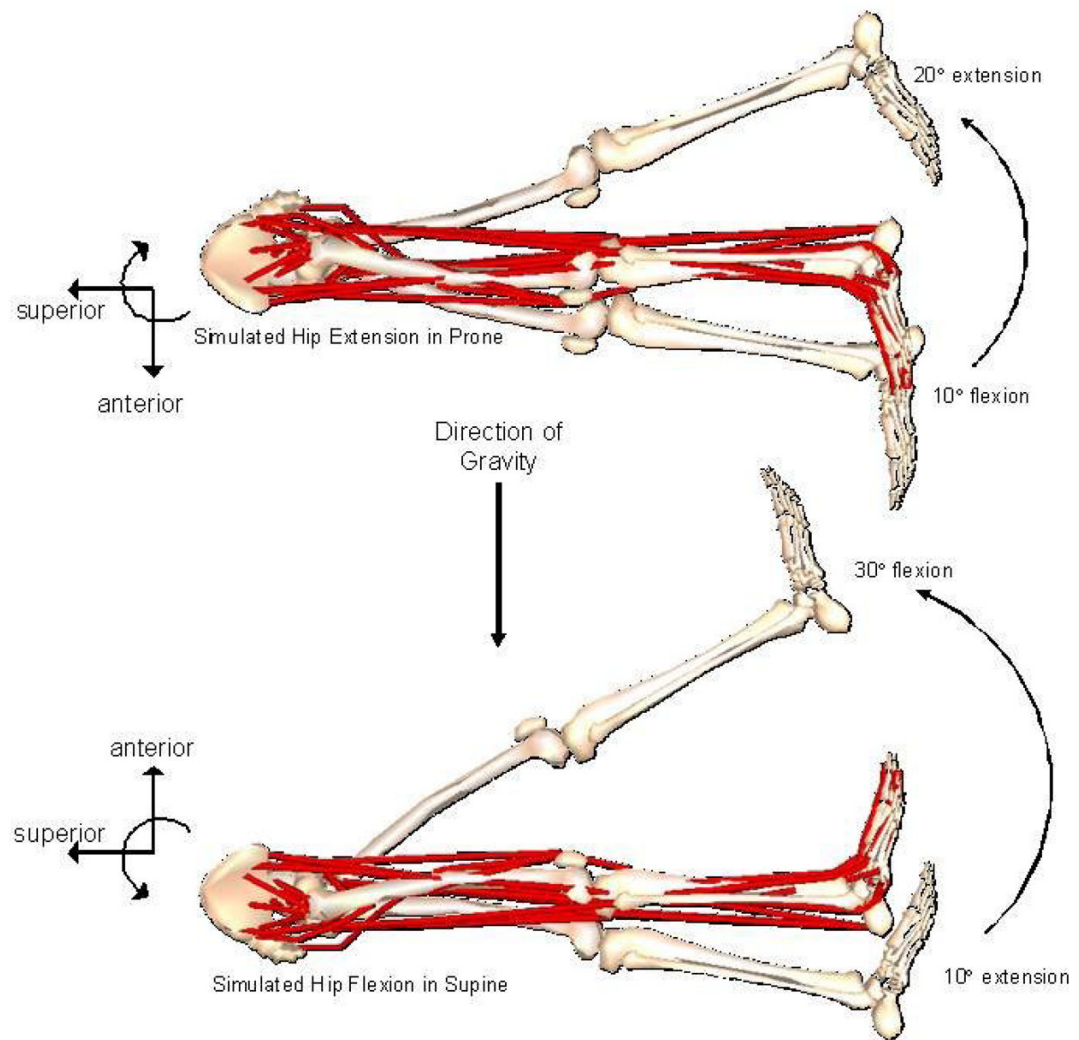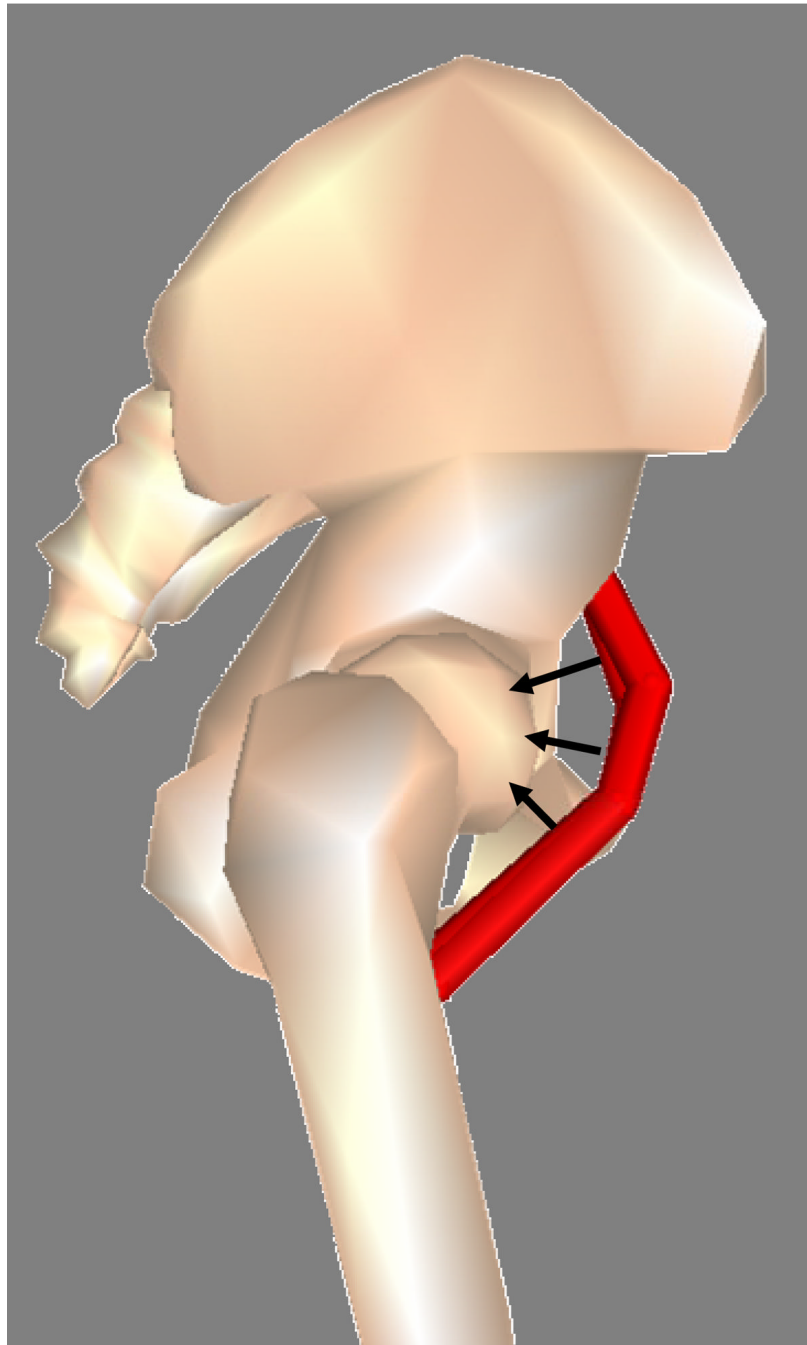
FIGURE 1.
Exercises simulated using the musculoskeletal model. Red lines indicate all the muscles included in the model. For hip extension in prone, gravity was specified as acting from posterior to anterior in line with the pelvic reference frame. The hip started in 10° of hip flexion and moved through 30° to a final position of 20° of hip extension. For hip flexion in supine, gravity was specified as acting from anterior to posterior in line with the pelvic reference frame. The hip started in 10° of hip extension and was moved through 40° to a final position of 30° of hip flexion. (Image created in SIMM (MusculoGraphics, Inc, Santa Rosa, CA, USA))
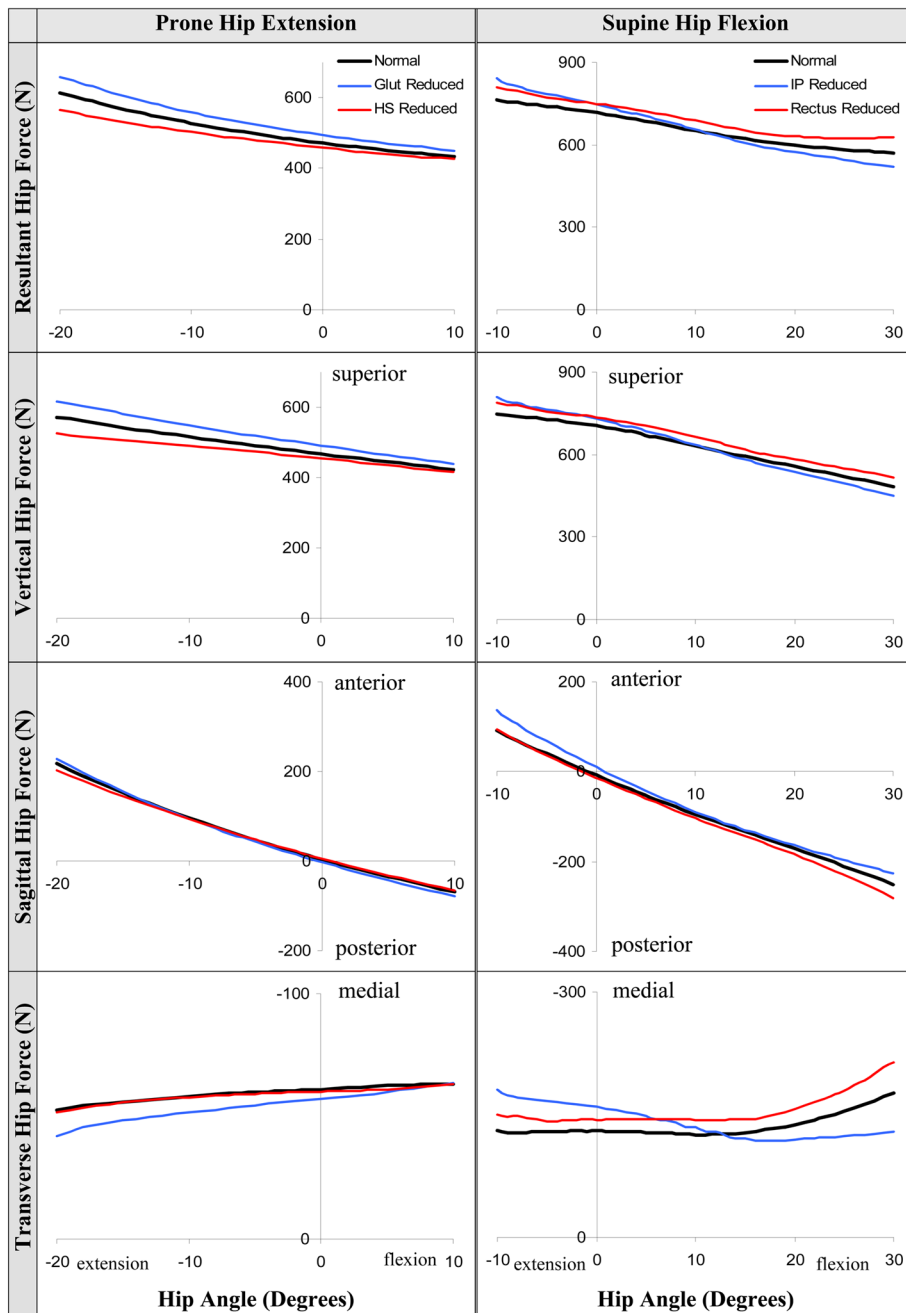
FIGURE 2.
Joint forces due to muscle activity during simulated prone hip extension and supine hip flexion. The vertical force is always in the superior direction and is the force component with the highest magnitude. The direction of the force in the sagittal plane is affected by hip position and muscle force contribution. The transverse force is always in the medial direction. Hip joint forces are greatly affected by hip joint position and partially affected by alterations in muscle force contribution.

FIGURE 3.
Stabilizing effect of iliacus and psoas muscles crossing the hip joint. The iliacus and psoas muscles (thick red lines) may apply forces (black arrows) to the femoral head through their muscle bellies as they pass over the underlying anterior hip joint structure. These forces may help maintain the femoral head positioning within the acetabulum similar to the rotator cuff muscles of the shoulder. This effect has been suggested to add strength to the anterior hip joint capsule when in hip extension (Philippon, 2001).