Clinical Biomechanics 24 (2009) 35–42

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

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

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ARTICLE INFO

Article history:
Received 28 January 2008
Accepted 9 September 2008

Keywords:
Hip joint force
Hip pain
Prone hip extension
Straight leg raising

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.

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 three-dimensional musculoskeletal model to estimate the hip joint force. A musculoskeletal model is a mathematical representation of bone and muscle, and illustrates how

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doi:10.1016/j.clinbiomech.2008.09.006

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 four segments: the pelvis, thigh, shank and foot of the right leg. The model contains six degrees of freedom (DOF) to represent the primary motions at the hip, knee and ankle as follows: (i) three DOFs at the hip to model adduction–abduction, internal–external rotation and flexion–extension, (ii) one DOF at the knee to model flexion–extension, and (iii) two DOFs 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 (1990).

Musculoskeletal parameters, including muscle path and maximum isometric force, were adapted from work by 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

$$\vec{T}(\vec{Q}) = -\vec{G}(\vec{Q}) \quad (1)$$

In this equation, the position of the limb is defined by $\vec{Q}$, which is a column vector of the six angles, one for each degree of freedom at each modeled joint. Similarly, $\vec{T}$ is comprised of the net joint torques generated by the muscles at each degree of freedom. $\vec{G}$ is comprised of the torques due to gravity at each degree of freedom, and is also affected by the position of the limb. Eq. (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 (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 rehabilitation exercises 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 (Fig. 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 (Fig. 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 three 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