WALKING IN GREATER HIP EXTENSION INCREASES PREDICTED ANTERIOR HIP JOINT REACTION FORCES

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INTRODUCTION

Acetabular labral tears are a recently recognized source of anterior hip pain. Excessive forces have been implicated as one cause of these tears. A clinical sign of a tear is anterior hip pain during late stance phase of gait. We propose that this pain is due, in part, to excessive force on the anterior structures of the hip, especially when the hip is extended. Our hypothesis is supported by the fact that, in patients with this pain, the pain is reduced when the patient is instructed to walk in less hip extension. This gait modification may help to decrease the anteriorly directed force, thus decreasing the pain.

The purpose of this study is to use a 3D dynamic musculoskeletal model to estimate the hip joint reaction forces while walking. We hypothesize that, within subjects, the gait trials with the greatest maximum hip extension will have a greater maximum anterior hip joint reaction force when compared to trials with the least maximum hip extension.

METHODS

A six degree of freedom, 3D musculoskeletal model of a lower leg was developed to calculate joint reaction forces in the hip, knee, and ankle. Musculoskeletal parameters were adapted from Delp [1]. Kane's Method [2] and AUTOLEV 3.1 (OnLine Dynamics, Inc., Sunnyvale, CA) were used to generate the dynamic equations of motion. The general form of the dynamic equations of motion is:

$$\mathbf{M}\left(\vec{Q}\right)\ddot{\vec{Q}} = \vec{T} + \vec{P}\left(\vec{Q},\dot{\vec{Q}}\right) + \vec{V}\left(\vec{Q},\dot{\vec{Q}}\right) + \vec{G}\left(\vec{Q}\right) + \vec{E}\left(\vec{Q},\dot{\vec{Q}}\right)$$

where M is the mass matrix, T is the net joint torques contributed by the force in the muscles spanning all joints, P is the torques developed passively in the joints due to viscoelastic damping and passive joint structures, and V, G, and E are the instantaneous segmental torques caused by the inertial, gravitational, and external forces, respectively. Q is the column vector of joint angles. Using kinematic and kinetic data from gait trials, the required segmental torques due to muscles were obtained for each trial. At each time point, a pseudoinverse optimization routine was used to solve for the optimal set of muscle forces [3] to create the needed joint torques. Once the optimized muscle stresses were solved simultaneously across all joints, the model calculated the resulting 3D reaction forces in the hip due to muscular contraction.

The gait data used in this study was collected from 5 healthy college-aged male subjects who participated in a previous study [4]. Each subject walked at a self-selected speed while 3D kinematic and kinetic data were collected. As we were most interested in anteriorly directed joint reaction forces, we only analyzed the terminal stance phase of gait. Within each subject, the reaction forces at the hip for the 2 trials with the most hip extension (MHE) range of motion and the 2 trials with the least hip extension (LHE) range of motion were each averaged. Paired t-tests were used to detect differences ($p < 0.05$).

RESULTS AND DISCUSSION

The average anterior reaction forces are presented in Table 1. Despite only a 2 degree difference between the MHE and LHE trials on average, ambulating with greater hip extension results in anteriorly directed forces that are significantly higher than ambulating with less hip extension.

Table 1: The average maximum hip angle and average anterior joint reaction force for the two groups.

Variable:	MHE	LHE	p
Max Hip Extension Angle (°)	13.5	11.5	0.01
Anterior Joint Reaction (N)	1791	1428	<0.01

CONCLUSIONS

A 3D musculoskeletal model was modified to estimate joint reaction forces in the hip during ambulation. During gait trials with greater maximum hip extension, anterior joint reaction forces are higher than with less hip extension. Instructing patients with anterior hip pain to ambulate in less hip extension may help to decrease forces on the anterior hip joint and thereby decrease pain.

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