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The impact of hip implant alignment on muscle and joint loading during dynamic activities

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ABSTRACT

Background: Component alignment is an important consideration in total hip arthroplasty. The impact of changes in alignment on muscle forces and joint contact forces during dynamic tasks are not well understood, and have the potential to influence surgical decision making. The objectives of this study were to assess the impact of femoral head/stem and cup component placement on hip muscle and joint contact forces during tasks of daily living and to identify which alignment parameters have the greatest impact on joint loading.
Methods: Using a series of strength-calibrated, subject-specific musculoskeletal models of patients performing gait, sit-to-stand and step down tasks, component alignments were perturbed and joint contact and muscle forces evaluated.
Findings: Based on the range of alignments reported clinically, variation in head/stem anteversion-retroversion had the largest impact of any degree of freedom throughout all three tasks; average contact forces 413.5 (319.1) N during gait, 262.7 (256.4) N during sit to stand, and 572.7 (228.1) N during the step down task. The sensitivity of contact force to anteversion-retroversion of the head/stem was 31.5 N/° for gait, which was similar in magnitude to anterior-posterior position of the cup (34.6 N/m for gait). Additionally, superior-inferior cup alignment resulted in 16.4 (4.9)° of variation in the direction of the hip joint contact force across the three tasks, with the most inferior cup placements moving the force vector towards the cup equator at the point of peak joint contact force.
Interpretation: A quantitative understanding of the impact and potential tradeoffs when altering component alignment is valuable in supporting surgical decision making.

1. Introduction

In total hip arthroplasty (THA), alignment of the femoral components and acetabular cup influences the mechanics of the joint, including the functional range of motion of hip articulation and the joint positions in which impingement can and cannot occur (Patel et al., 2010). Further, component alignment has been associated with poor clinical outcomes such as impingement (Renkawitz et al., 2012), dislocation (Higa et al., 2011), increased liner wear and fracture, osteolysis (Kennedy et al., 1998), edge loading (Kwon et al., 2012) and increased metal ion in the blood (Harris, 2012). Alignment includes implantation parameters of the three dimensional position of the stem and cup relative to the femur and pelvis, respectively, as well as orientation in version and abduction (Widmer and Zurfluh, 2004). With surgical aims of improving joint mobility and restoring the ability to safely perform activities of daily living (Vissers et al., 2011), the mechanics of the joint are a critical consideration, with many studies assessing the effects of component alignment on range of motion and the likelihood of impingement (Petrella et al., 2009). Component alignment also directly affects hip joint loading during typical activities of daily living, with changes in femoral anteversion capable of resulting in 30% increases in hip joint contact force (Heller et al., 2001).

Native hip anatomy following THA is typically not fully restored compared to the contralateral hip (Tsai et al., 2014). A wide range of variation has been reported in the placement of the femoral component in particular. Wines and McNicol (2006) used CT measurements to show femoral anteversion averaged 16.8 (11.1)° with a range from −15.0° retroversion to 45.0° anteversion. Higher amounts of femoral

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version following THA have been associated with pain and decreased quality of life (Liebs et al., 2014). In addition, vertical elevation of the femoral component, which results in leg length changes, has been shown to vary between −2.5 to 12.6 mm relative to the contralateral side (Tsai et al., 2014). Placement of the acetabular component has shown smaller amounts of variability in comparison to the femoral component, but cup placement has a direct influence on the location of the center of rotation of the hip and can also result in leg length changes. Cup position variability has been shown to be similar in each degree of freedom and can vary by up to approximately 10 mm (Tsai et al., 2014). Recently, computer-aided surgery systems that improve the placement of the cup relative to native hip geometry have become available for use during THA (Renkawitz et al., 2009). However, these systems have not been widely adopted because they can add considerably to both cost and time of the surgery and, currently, there is no assistive surgical technology for the placement of the stem.

The influence of component position on joint loading can be assessed non-invasively with the use of the musculoskeletal modeling software platforms, such as OpenSim (Delp et al., 2007) or Anybody (AnyBody Technology, Aalborg, Denmark). Musculoskeletal modeling is used to calculate joint kinematics and moments, as well as intersegmental joint loads and muscle forces. Musculoskeletal simulation offers valuable data to clinicians and researchers assessing pathological conditions and understanding human movement. Simulation of human movement has significantly impacted approaches to clinical treatment of osteoarthritis (Fregly et al., 2007) and total joint replacement (Gaffney et al., 2015; Navacchia et al., 2016), as well as basic science related to the understanding of movement progression and control during dynamic tasks (Anderson et al., 2004; Neptune et al., 2009; Zajac et al., 2002). There have been a number of impactful innovations in simulation methods from sophisticated subject-specific models with highly accurate anatomic detail (Arnold et al., 2010), to creation of efficient forward dynamics simulations using computed muscle control (Thelen and Anderson, 2006) that make it possible to address variation in component positioning across multiple patients.

Musculoskeletal modeling has been previously used to assess how variation in the location of the hip joint center can result in changes to the lines of action and moment generating capabilities of the hip muscles (Delp and Maloney, 1993). However, the extent of the impact of positioning THA components on hip joint loading during daily activities is not known. Quantifying relationships between component alignment, muscle forces and joint loading can support surgical practice in assessing the tradeoffs that exist in joint mechanics and loading when altering component placement for considerations of bone quality or fixation. Accordingly, the objectives of this study were to assess the impact of femoral head/stem and cup component alignment on hip joint contact forces and muscle forces during tasks of daily living and to identify which alignment parameters have the greatest impact on hip joint loading.

2. Methods

2.1. Patient data collection

A cohort of five patients who had undergone THA to treat end stage osteoarthritis (2 M: 60.5 (9.2) years, 94.5 (9.9) kg; 3 F: 61.3 (9.1) years, 72.2 (8.4) kg) performed through a posterolateral approach, were selected from a larger prospective study that included 26 patients. Patients were eligible if they were between the ages of 45 and 80 years, had no history of uncontrolled hypertension or diabetes, body mass index < 40 kg/m^2, no additional orthopaedic pathology, or neurologic disorders that impaired daily function. Each patient provided written, informed consent and participated in a laboratory testing session that was approved by the Colorado Multiple Institutional Review Board. The laboratory testing session was performed a minimum of 10 weeks post-operatively and averaged 11.7 ± 1.4 weeks for these five patients.

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 three activities of daily living. Activities consisted of gait at a self-selected pace, a sit-to-stand task in which patients stood from a chair 43 cm in height and achieved a fully upright posture, and a step down task from a height of 20 cm. Each task was performed onto a Bertec (Columbus, OH, USA) force platform embedded in the floor with their surgical limb while force data was collected at 2000 Hz and an 8 camera Vicon motion capture system (Centennial, CO, USA) collected at 100 Hz (Judd et al., 2016).

Isometric strength of the hip flexors, extensors, and abductors, as well as the knee flexors and extensors, was assessed using an electromechanical dynamometer (HUMAC NORM, CSMI Solutions, Stoughton, MA, USA) connected to a Biopac Data Acquisition System (Biodex Medical Systems, Inc., Shirley, NY, USA) running AcqKnowledge software (Biodex Medical Systems, Inc., Shirley, NY, USA). Strength was measured in the affected limb. For hip flexor and extensor strength assessment, participants were positioned supine with the hip flexed to 40°. Hip abductor strength was measured while participants 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 the measurement (Judd et al., 2016). Patients underwent 3 trials of maximal effort contractions and the highest of the three was used in further analysis.

2.2. Musculoskeletal modeling

Patient-specific lower extremity muscle strength calibration was performed using a musculoskeletal model that included detailed hip musculature (Shelburne et al., 2010) to be used in simulations of each task of daily living. 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., 1990, 2007). 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 each segment in the model were scaled so that the distances between the virtual markers on the model matched the distances between the experimental markers. 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. Calibration of muscle maximum isometric parameters was performed by increasing or decreasing the maximum force for each muscle to minimize differences between model-predicted and measured preoperative maximum isometric joint torques in hip flexion, extension, and abduction, as well as knee flexion and extension. Muscles in each group were all scaled by the same factor to maintain the strength ratios between muscles of the same group.

Baseline simulations for each patient performing the three activities were constructed using the patient-specific scaled model and corresponding measured kinematics and ground reaction forces to predict hip joint contact forces (JCFs) and muscle forces using static optimization, in which the sum of muscle activation squared was minimized (Anderson and Pandy, 2001). Baseline refers to the prescribed neutral hip implant alignment that was created by exactly replicating the joint center of rotation in the patient-specific scaled model with the cup placed in 15° of anteversion and 40° of inclination and the stem in 10° of anteversion. Results from the baseline simulations were compared to data collected from patients implanted with telemetric hip implants performing the same three tasks (Bergmann et al., 2010).

Two types of evaluation were performed to assess the impact of