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Impact of the Femoral Head Position on Moment Arms in Total Hip Arthroplasty: A Parametric Finite Element Study

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A B S T R A C T

Background: Although the importance of accurate femoral reconstruction to achieve a good functional outcome is well documented, quantitative data on the effects of a displacement of the femoral center of rotation on moment arms are scarce. The purpose of this study was to calculate moment arms after nonanatomical femoral reconstruction.
Methods: Finite element models of 15 patients including the pelvis, the femur, and the gluteal muscles were developed. Moment arms were calculated within the native anatomy and compared to distinct displacement of the femoral center of rotation (leg lengthening of 10 mm, loss of femoral offset of 20%, anteversion $\pm$10°, and fixed anteversion at 15°). Calculations were performed within the range of motion observed during a normal gait cycle.
Results: Although with all evaluated displacements of the femoral center of rotation, the abductor moment arm remained positive, some fibers initially contributing to extension became antagonists (flexors) and vice versa. A loss of 20% of femoral offset led to an average decrease of 15% of abductor moment. Femoral lengthening and changes in femoral anteversion ($\pm$10°, fixed at 15°) led to minimal changes in abductor moment arms (maximum change of 5%). Native femoral anteversion correlated with the changes in moment arms induced by the 5 variations of reconstruction.
Conclusion: Accurate reconstruction of offset is important to maintaining abductor moment arms, while changes of femoral rotation had minimal effects. Patients with larger native femoral anteversion appear to be more susceptible to femoral head displacements.

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The importance of the reconstruction of moment arms in total hip arthroplasty (THA) to achieve a good joint function is well documented [1-4]. Displacement of the femoral head center might occur by surgeon's preference (eg, correction of an excessive native femoral anteversion), by an inaccurate reproduction of a preoperative template, or due to the limited choices offered by off-the-shelf implants. Quantitative data on the effects of nonanatomical reconstructions on muscle moment arms are scarce, complicating preoperative and intraoperative decision making.

Some clinical data are available regarding a displacement of the femoral offset in the mediolateral direction (ie, changes of femoral offset). For example, a loss of offset of 15% to 20% [2-4] appears to be associated with a significantly worse outcome in terms of abductor strength and gait. However, these data do not take into account eventual concomitant displacements of the femoral head center in anteroposterior or craniocaudal directions, which are very common in clinical practice [5]. In a cohort of 45 patients undergoing THA for developmental dysplasia, Liu et al [6] described a significant increase in abductor moment arm by restoring leg length. However, analysis was performed solely on a standard anteroposterior pelvic x-ray. Two-dimensional (2D) evaluations of moment arms are, however, flawed, as abductor muscle fibers are not parallel to the frontal plane, are not straight, and deform during hip motion [7]. Sakai et al [8] evaluated preoperative and postoperative computed tomographic (CT) scans of patients undergoing THA using a modular neck stem, including displacement of the femoral center of rotation in all dimensions. However, moment arms were then only calculated in 2D (ie, on scout views). We are not aware of studies evaluating the effect of a displacement of the femoral head center in anteroposterior direction on abductor moment arms in 3 dimensions (3D).

The purpose of this study is to assess the effect of a displacement of the femoral center of rotation in craniocaudal, mediolateral, and anteroposterior directions on moment arms of abductor muscles.

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Various displacements of the femoral center of rotation have been evaluated, including a decrease of femoral offset (−20%), femoral lengthening (+10 mm), changes of native anteversion (±10°), and fixed anteversion (15°). Using a finite element model based on preoperative CT scans of 15 individual patients, we have previously reported that the effect of a displacement of the center of rotation depends on individual anatomy and is hence variable among patients [7]. Adapting those models, abductor moment arms were calculated within a range of motion typically observed during a normal gait cycle. These values were compared to those calculated after a displacement of the femoral head center in 3D separately (mediolateral, craniocaudal, and torsional) around the femoral axis.

Materials and Methods

Preoperative CT scans of 15 patients undergoing THA were used. Abductor moment arms were calculated in each patient for 6 different positions of the femoral center of rotation:

(P0) Anatomical reconstruction of the hip as reference,
(P1) Loss of femoral offset (−20%),
(P2) Femoral lengthening (+10 mm),
(P3) Anteversion (ie, native anteversion +10°),
(P4) Retroversion (ie, native anteversion −10°),
(P5) Femoral anteversion at 15°.

The CT scan included the entire pelvis and the proximal and distal femur. The pelvis; the femur; and the gluteus medius, minimus, and maximus were segmented using the imaging software Amira (FEI Visualization Sciences Group, Bordeaux, France). A cloud of points at the border of all segmented objects was extracted, and a surface model of each structure was built using Geomagic software (Geomagic, Research Triangle Park, NC). Femoral offset, the trochanteric height, and the femoral anteversion of each patient were measured based on these reconstructions (Fig. 1 and Table 1). These anatomical measurements were performed with the CAD software Catia (Dassault Systèmes Simulia Corp, Providence, RI) and user-written scripts in Matlab (MathWorks, Inc, Natick, MA).

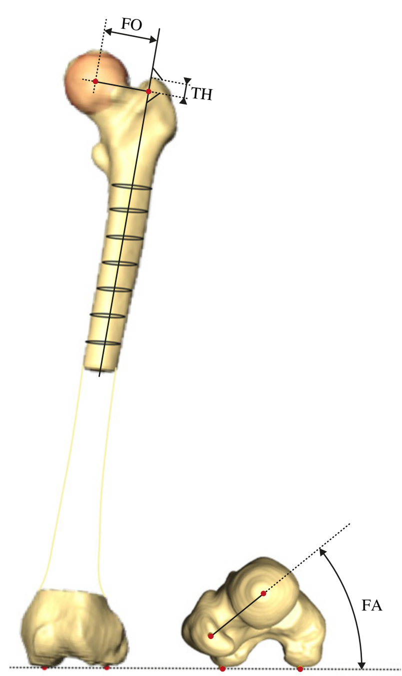

Fig. 1. The femoral offset (FO), trochanteric height (TH), and femoral anteversion (FA) were measured on 3D reconstructions of the proximal and distal femur. The femoral axis was fitted through center points of diaphyseal cross sections. FO was the distance from the femoral head center to the femoral axis. TH was the length of the tip of the greater trochanter relative to the femoral head center. FA was the angle between the axis of the condyles and the axis of the femoral head center and neck.

Table 1
List of the 15 Patients, With the 3 Femoral Preoperative Parameters: Femoral Offset, Trochanteric Height, and Femoral Anteversion.

Patient No.	Sex	Age (y)	FO (mm)	TH (mm)	FA (°)
1	F	65	37	13	30
2	M	61	39	13	30
3	M	68	41	14	36
4	M	77	36	16	36
5	F	74	41	13	38
6	M	73	52	15	24
7	F	60	43	19	25
8	F	72	45	8	18
9	F	71	31	7	30
10	M	39	48	11	11
11	F	47	28	7	34
12	M	36	38	9	15
13	M	52	30	6	11
14	M	57	31	9	42
15	F	39	28	2	20
Mean	–	59	38	11	27
Median	–	61	38	11	30
Minimum	–	36	28	2	11
Maximum	–	77	52	19	42

Abbreviations: FO, femoral offset; TH, trochanteric height; FA, femoral anteversion; F, female; M, male.

We then performed THA within this virtual model using Catia. To avoid limitations of reconstruction options imposed by a chosen implant brand, reconstruction was simplified by positioning the femoral center of rotation either anatomically (P0) or including a defined displacement (P1-5). The acetabular and femoral centers of rotation were superimposed in each case, avoiding subluxation.

Finite element models were created for each of the 15 patients. A local coordinate system was defined for the pelvis and femur [9]. A passive hip motion was simulated within the range of a normal gait cycle [10]. Abduction was performed in the coronal plane, from 6° of adduction to 6° of abduction. Extension was performed in the parasagittal plane, from 38° of flexion to 5° of extension. The finite element method predicted the deformation of the muscles during passive motion. Muscles were assumed incompressible and transversely isotropic. A hyperelastic anisotropic constitutive law based on the following strain energy potential W was used to model muscle tissue [11]:

$$W = C_{10}(I_1-3) + C_{01}(I_2-3) + C_{20}(I_1-3)^2 + C_{11}(I_1-3)(I_2-3) + C_{02}(I_2-3)^2 + C_{\nu}(\boldsymbol{\lambda}_2-1)^2$$

where $\boldsymbol{\lambda}$ is the stretch in the fiber direction, and I_1 and I_2 are the first and second modified invariants of the Cauchy-Green deformation tensor, respectively [11]. The material constants were determined experimentally ($C_{10} = 64.3$ kPa, $C_{01} = -38$ kPa, $C_{20} = 9.4$ kPa, $C_{11} = -0.043$ kPa, $C_{02} = 0.005$ kPa, $C_{\nu} = 200$ kPa) [12]. The common surfaces of the gluteus minimus, gluteus medius, and gluteus maximus were tethered together. A sphere was used to model the contact wrapping of the gluteus maximus around the pelvis. The anisotropy was aligned along the directions of the muscle fiber. It was assumed perpendicular to the origin and insertion surfaces and following the outer surface of the muscle from origin to insertion. Bony structures as well as implants were assumed rigid. Moment arms were calculated for 10 fibers per muscle, which were evenly distributed within the gluteus medius and minimus (Fig. 2). The finite element analysis was performed in Abaqus (version 6.13; Dassault Systèmes). The constitutive law of the muscles was implemented in Fortran with user-defined subroutines (UANISOHYPER_INV and ORIENT) available in Abaqus. Muscles were meshed with nearly 120,000 hybrid linear tetrahedral elements (C3D4H). The 10 fibers used to evaluate the moment arms were modeled as soft neo-Hookean