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Original article

No effect of femoral offset on bone implant micromotion in an experimental model

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ABSTRACT

Background: In total hip replacement (THR), the femoral offset (FO) is assessed preoperatively, and the surgeon must determine whether to restore, increase, or decrease the FO based on experience and the patient's clinical history. The FO is known to influence the abductor muscle strength, range of motion (ROM), gait, and hip pain after THR; however, the true effect of FO on bone implant micromotion is unclear. Therefore, we investigated to assess: (1) the muscle loading response during gait, (2) whether FO affects bone implant micromotion during gait.
Hypothesis: A variation of $\pm$ 10 mm from the anatomical FO affects the muscle loading forces.
Materials and methods: We modified a personalized musculoskeletal model of the lower extremity to determine the 3-dimensional contact forces at the hip joint in the presence of a stem with varying offsets during a gait cycle. A detailed finite element (FE) model was then constructed for increased, restored, and decreased FOs. The maximum load obtained during normal walking gait from the musculoskeletal model was applied to the respective FE models, and the resultant stem-bone micromotion and stress distribution were computed.
Results: Increasing the FO to +10 mm decreased the peak force generated by the abductor muscles during the cycle by 15.0% and decreasing the FO to $-$10 mm increased the von Mises stress distribution at the distal bone by 77.5% ($P < 0.05$). A variation of the offset within 10 mm of the anatomical offset showed no significant differences in micromotion ($P > 0.05$) and peak stresses ($P > 0.05$).
Discussion: Coupling the musculoskeletal model of the gait cycle with FE analysis provides a realistic model to understand how FO affects bone implant micromotion. We found that there was no effect of FO on bone implant micromotion; thus, a surgeon does not need to evaluate the implications of FO on micromotion and can determine a FO that best decreases the work load of abductor muscles, increases ROM, and reduces hip pain.
Level of evidence: IV, biomechanical study.

1. Introduction

Total hip replacement (THR) has become an effective way to alleviate pain and improve mobility in patients with severe arthritic conditions of the hip [1]. The increased life expectancy and demand among younger patients has required improvements in the durability of hip arthroplasty [2]. The cementless approach has rapidly become accepted as the ideal surgery in younger patients, and its success relies on the initial stability of the bone-prosthesis interface [3]. The femoral offset (FO) is defined as the perpendicular distance between the femoral head's center of rotation and the long axis of the femur, which is estimated using an anteroposterior (AP) radiograph of the pelvis as shown in Fig. 1 [4].

Since the modularity of implants has increased over the last two decades, surgeons can precisely control the FO to optimize abductor muscle strength, range of motion (ROM), and stability [4,5]. Clinical studies and computer modeling have indicated a correlation between an increase in ROM and abductor muscle strength with increased FO [6–8]. However, some authors have suggested that if the anatomical offset is not restored, the patient may develop fatigue, impingement, limp, or prosthesis dislocation [4,5,9]. A decrease of the FO in clinical studies has been implicated in reduced ROM and abductor muscle strength, gait abnormalities, and increased polyethylene wear [6,10,11].

Stability of the stem can be evaluated by measuring the micromotion between the bone-stem interface [12]. Increased micromotion prevents biological fixation of the prosthesis,

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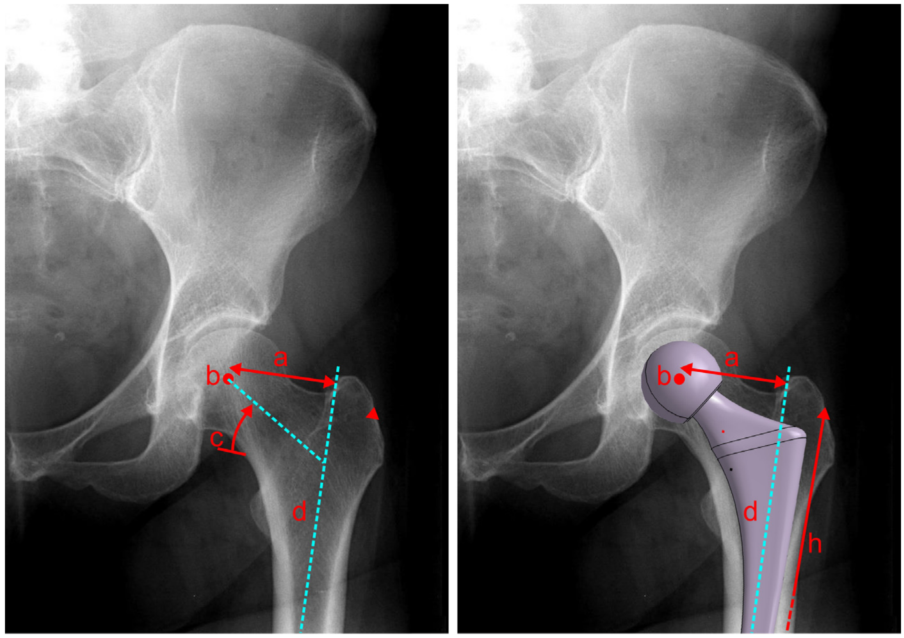

Fig. 1. Femoral offset is measured in plain antero-posterior radiographs as the perpendicular distance from the femoral head's center of rotation and the long axis of the femur: (a) radiological offset as a projection of the actual offset, (b) center of rotation, (c) neck angle, (d) anatomical axis, (h) constant stem length.

resulting in failure. Computer modeling and in vitro testing are used to evaluate the prosthesis' mechanical stability. A major limitation of previously performed in vitro studies [13] of femoral stem stability is that the dynamic nature of the contact forces at the hip joint during gait was not incorporated.

Several studies have reported on whether FO affects the muscle strength and ROM, but limited data exists on whether the FO affects stem stability during gait [14]. Therefore, our objective was:

- to determine the change in muscle activity at the hip joint during gait for different FOs in the context of THR;
- to evaluate the bone implant micromotion of the femoral stem during gait.

Since gait has a significant impact on the contact forces at the hip joint, we used a multibody dynamic computer simulation of gait coupled with finite element (FE) analysis to evaluate bone implant micromotion. We aim to shed light on how FO should be managed to optimize stem stability and understand the resultant changes in the gait cycle with varying FOs. We hypothesized that a variation of ± 10 mm from the anatomical FO affects the muscle loading forces.

2. Materials and methods

2.1. OpenSim model for gait

The model was developed in OpenSim (v2.2.1, http://simtk.org) and consisted of the pelvis, thigh, shank, and foot segments, which were inter-connected by hip, knee, and ankle joints [15]. The eight-segment, 23-degree-of-freedom (dof) musculoskeletal model with 22 muscles was modified from a previously validated Gait2392 musculoskeletal model [15,16].

The hip center was defined as the center of the sphere that best fitted the femoral head surface, and the center of the intercondyloid eminence was defined as the femur's rotation center. The restored FO was defined as the configuration in which the head center of the prosthesis is coincident with that of the femoral head prior to bone dissection.

Preparation of the femur was performed by a Boolean subtraction of 0.165 kg of bone from the femoral neck in a commercial solid modeling software (Rhinoceros, Robert McNeel & Associates, Seattle, WA, USA). A trabecular femoral stem prosthesis (Trabecular Metal^{TM}, Zimmer, USA) was used in our model (length 160 mm, mass 0.295 kg). Fixation of the prosthesis resulted in a new mass center. Five case studies were analyzed for resultant forces during gait: decreased (−10 and −5 mm), increased (+5 and +10 mm), and restored (0 mm) FOs. A positive and negative 1.18-degree angular variation of the mechanical axis produced a +5 and −5 mm FO, respectively. Muscle attachment points of the 22 considered muscles were recalculated in computer-aided design (CAD) and imported into OpenSim (Fig. 2). Using the computed muscle control (CMC) algorithm [17], musculoskeletal dynamic simulations were run.

2.2. FE model for bone implant micromotion

Three-dimensional mesh model of the femur's contours was built from CT data available of the Visible Man (Visible Human Project®, National Library of Medicine, Bethesda, MD) in 3-Matic software package (Materialise, Leuven, Belgium). Methodology was adapted from a previously validated FE model [18]. Linear [19] and power [20] relationships were used to compute the Young's modulus for every element of mesh from the Hounsfield Unit value. A Poisson's ratio of 0.3 was assumed since the femur was modeled to be linear elastic [21]. We imposed a range of 1.2 to 28.2 GPa for the Young's Modulus, which was in range of previously established values [18]. The CAD parametric model of the stem was based on the Trabecular stem's material properties, which consisted of a rigid core with softer coating material (Young's Modulus 3.0 GPa) to mimic coating. The geometry of the reconstructed bone was scaled to that of the femur in the OpenSim model. Contact between the bone-stem interface was modeled with penalty formulation [22]. Friction coefficients of 0.5 for the coated part [23] and 0.01 for the uncoated part were used to define the tangential friction properties. The FOs were varied with unaltered stem length (h) as shown in Fig. 1; therefore, maintaining the footprint on the long axis of the femur. Boundary conditions and corresponding vertical peak load