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Human Proximal Femur Bone Adaptation to Variations in Hip Geometry

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ABSTRACT

The study of bone mass distribution at proximal femur may contribute to understand the role of hip geometry on hip fracture risk. We examined how bone mineral density (BMD) of proximal femur adapts to inter individual variations in the femoral neck length (FNL), femoral neck width (FNW) and neck shaft angle (NSA). A parameterized and dimensionally scalable 3-D finite element model of a reference proximal femur geometry was incrementally adjusted to adopt physiological ranges at FNL (3.90-6.90 cm), FNW (2.90-3.46 cm), and NSA (109-141º), yielding a set of femora with different geometries. The bone mass distribution for each femur was obtained with a suitable bone remodelling model. The BMDs at the integral femoral neck (FN) and at the intertrochanteric (ITR) region, as well as the BMD ratio of inferomedial to superolateral (IM:SL) regions of FN and BMD ratio of FN: ITR were used to represent bone mass distribution. Results revealed that longer FNLs present greater BMD (g/cm^3) at the FN, mainly at the SL region, and at the ITR region. Wider FNs were associated with reduced BMD at the FN, particularly at the SL region, and at the ITR region. Larger NSAs up to 129° were associated with BMD diminutions at the FN and ITR regions and with increases of the IM:SL BMD ratio while NSAs larger than 129° resulted in decrease of the IM:SL BMD ratio. These findings suggest hip geometry as moderator of the mechanical loading influence on bone mass distribution at proximal femur with higher FNL favoring the BMD of FN and ITR regions and greater FNW and NSA having the opposite effect. Augmented values of FNL and FNW seem also to favor more the BMD at the superolateral than at the inferomedial FN region.

Introduction

Hip fracture is an important public health and personal burden which is anticipated to continue to rise due to increased life span [1,2]. The ability of bone to resist fracture depends on its material composition (i.e., amount and properties of the materials that form the bone) and the spatial distribution of the bone mass [3]. Moreover, it is well established that the risk of fracture is multi-factorial and many independent risk factors have been identified to enhance the specificity and sensitivity of predictive fracture risk models [1,2,4]. Despite the suggestion of hip geometry as risk factor for hip fracture [5–9], no consensus has been achieved so far about which geometric parameters improve the prediction of fracture risk. The inconsistency among studies of femoral geometry and hip fracture can be attributed to several factors, including study design (retrospective vs prospective), variability in measurement techniques, sample size limitations and the fact that most studies have combined all the types of hip fracture (femoral neck - cervical, trochanteric or intertrochanteric fractures) [10]. Currently, even the most complex fracture risk prediction tools do not integrate hip geometric characteristics as risk factors in clinical practice [4].

A reasonable approach to link hip geometry and hip fracture may be to analyze first if isolated variations in certain proximal femur geometric parameters give rise to particular spatial bone mass distribution patterns, especially as it has been shown that bone mass distribution at the proximal femur is associated with hip fracture risk [3,11]. Thus, with the objective of testing if similar mechanical load generates different effects on bone mass distribution at the proximal femur according to the hip geometry, this study proposes to map, by means of a three-dimensional finite element method and a pre-validated bone remodeling model, the distribution of bone mass at the proximal femur and examine its association with the geometric parameters that are commonly assessed, namely the femoral neck length (FNL), the femoral neck width (FNW) and the neck-shaft angle (NSA).

Material and Methods

Development of a parameterized 3-D finite element model (FEM)

The left proximal femur geometry of the adult 'Standardized Femur' 3-D model [12], derived from a CT-scan dataset of a composite human femur replica, was discretized using tetrahedral elements, giving rise to a refined and uniform size mesh (207502 elements, 39285 nodes,

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Abaqus FEA® element type C3D4, 4-node linear element, average edge-length of 2 mm, Joe-Liu mesh quality metric (0-1) [13] of 0.85) (Fig. 1 A). The type and number of elements were decided in order to minimize the error on stress and strain evaluation (convergence study) and, at the same time, to have a good resolution on the bone apparent density distribution which is computed assuming bone density is constant within each element. The Cartesian coordinates of the nodes forming the finite element mesh were imported to Matlab R2013a (8.1.0.604) as a point cloud, from which the following measures were calculated according to Mahaisavariya et al. [14]: femoral head centre; neck isthmus centre; femoral neck axis; femoral shaft axis; intersection of the femoral neck axis and the proximal shaft axis; FNL; FNW and NSA (Fig. 1 B). Three geometric parameters frequently assessed at the proximal femur were investigated: FNL, FNW (measured at the narrowest part of femoral neck) and NSA [10]. The parameterized model geometry was scaled by applying an affine transformation to the nodes within the finite element mesh belonging to the proximal femur region R (Fig. 1 B). The inter-distance of the mesh nodes not belonging to that region were not changed [6]. Sixteen individual finite element models (FEMs) were created based on the reference FEM to represent the three geometric features and their physiological variances in the human population: six FEMs for FNLs and five FEMs per each of the remaining geometric features (Fig. 1 C).

Finite element modeling: boundary conditions

The procedure to define the boundary conditions consisted of an adaptation of the musculoskeletal model and derived load profiles developed and validated by Bergmann [15] and Heller et al. [16,17], from which hip contact and muscle forces and the position of their acting points were taken. The muscle system included the abductors (*gluteus minimus*, *gluteus maximus*, *gluteus medius* and *tensor fascia latae*) and the. *vastus lateralis* (Fig. 1 A). Due to anatomical differences between the available musculoskeletal model and the used finite element model, a manual procedure was used to match the position of the acting points defined in the musculoskeletal model with the element surfaces of the reference finite element geometry.

A load case corresponding to the instant of peak hip contact force observed in the stance phase of a “typical” gait cycle was used (Table 1).

In order to generate a physiological loading condition, the forces applied to each individual surface element of the FEM's were not concentrated at the attachment locations but distributed over neighboring surfaces [18]. Thus, muscles forces were uniformly distributed over elements surface based on previously measured physiological muscle attachment areas [19]. Regarding hip contact force, a physiological hip contact surface area was also considered [20], in which the load decreases linearly with the distance from the acting point to zero on the edge of the highlighted area (Fig. 1 A). The general surface traction – Dsload *Abaqus 6.9* feature was used to define loads in a user-defined non-normal surface direction. The components of the traction vector load were calculated given the force vectors presented in Table 1 and the area of the elements surface.

The femoral bone was constrained in the mid-diaphysis to suppress rigid body motion. Since we intended to study the femur bone adaptation in different femur geometries, it was important to determine whether changes in the studied geometric parameters influence hip loading, which consequently would influence bone strains and thus bone remodeling [21,22]. Indeed, estimates of hip loading have been shown to be sensitive to femoral geometric features [24,25]. Of particular interest in this work are the observations of Lenaerts et al. [26] who found that increased femoral neck length resulted in increased peak hip

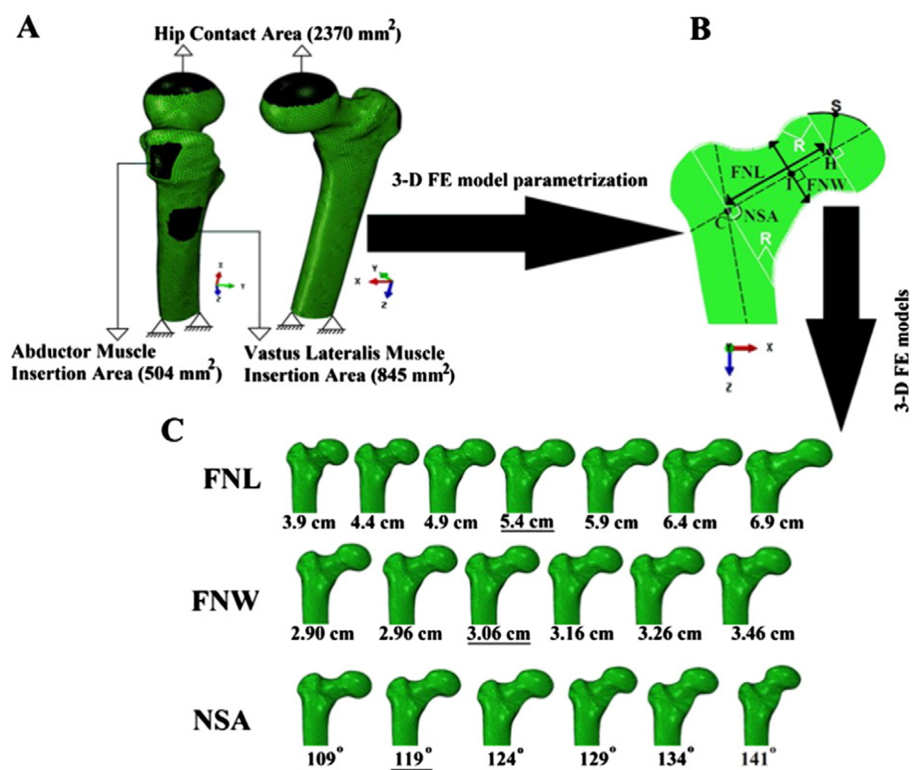

Fig. 1. Finite element model approach: A) Three dimensional (3-D) finite element model of the reference proximal femur. The areas highlighted in black represent the surfaces of application of the muscle and contact forces. The axis + z and + x is directed downwards and medially, respectively, while the axis + y is directed posteriorly. B) Schematic representation of geometric parameters: H – femoral head centre; I – neck isthmus centre; S - center of the femoral head surface; H-S – hip contact force direction; H-I – femoral neck axis; C - intersection of the femoral neck axis with the proximal shaft axis; FNL – femoral neck length; FNW – femoral neck width; NSA – neck shaft angle; R – proximal femur region wherein inter-distance of mesh nodes were changed. C) Whole set of finite element models separated by geometric characteristics. The underlined models represent the reference model for the corresponding feature.