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Proximal femoral growth plate mechanical behavior: Comparison between different developmental stages

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ABSTRACT

In long bones the growth plate is a cartilaginous structure located between the epiphysis and the diaphysis. This structure regulates longitudinal growth and helps determine the structure of mature bone through the process of endochondral ossification. During human growth the femur's proximal growth plate experiences changes in its morphology that may be related to its mechanical environment. Thus, in order to test this hypothesis from a computational perspective, a finite element analysis on a proximal femur was performed on which we modeled different physeal geometries corresponding to the shapes acquired for this structure in a child between the ages of five to eleven. Results show augmented Von Mises stress values with increasing irregularities in physeal geometry, whereas displacement decreased with increased irregularities in the growth plate's morphology. Such observations suggest that growth plate's shape changes follows a possible mechanical adaptation on imposed loads to sustain a person's increasing body mass during growth.

1. Introduction

Endochondral ossification is responsible for long bone longitudinal growth occurring at the growth plate. The cartilaginous mold is continuously replaced by bone as a result of sequential events including cell proliferation, extracellular matrix (ECM) synthesis, chondrocyte hypertrophy, ECM mineralization, vascular invasion, and apoptosis [1]. Therefore, based on these functions the growth plate is histologically organized in three zones: resting, proliferating, and hypertrophic. Each zone distinguishes specific morphological and biochemical stages during chondrocyte differentiation. In the resting zone, chondrocytes are in a quiescent state and present a rich ECM. In the proliferative zone, chondrocytes divide by mitosis and assume a flattened phenotype, and organize in columns. Last, in the hypertrophic zone chondrocytes stop dividing and terminal cell differentiation takes place [2]. Growth plate morphology in the proximal femur has been described by Kandzierski et al. [3] in an attempt to present a possible influence of the proximal femoral growth plate as a risk factor for the incidence of slipped capital femoral epiphysis (SCFE). In their work they carried out morphological studies of the proximal femur based on radiological images, computed tomography scans (CT), and magnetic resonance images (MRI) obtained from 100 children between the ages of three to thirteen. With their results they demonstrated that growth plate shape experiences different changes throughout development, and suggest these changes could be implicated in SCFE [3]. However, data reported by these authors does not provide information regarding mechanical stress present for each distinct physeal shape. Additionally, at present no computational model associated with the study of these morphological changes exist. Based on the above, we propose the hypothesis that with increasing body mass, both proximal femoral growth plate size and shape changes follow mechanical load adaptation, in the way that they reduce experienced mechanical stress. Under this view, the main objective of this work was to describe through a finite element (FE) analysis the state of stress associated with growth plate morphology, expressed in terms of size and shape. A mechanical FE model of a child's femoral head was established to simulate loading conditions during the toe off phase of the gait cycle. To this end, we employed a tridimensional (3D) reconstruction of the femur, where we modeled the growth plate during different stages of development. Results obtained

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provide qualitative and quantitative data of stress produced within the physis for distinct shapes. Thus, different geometries result in different stress patterns to support increasing body mass as a child grows.

2. Materials and methods

Due to anatomical variability inherent to the human being, resulting in uncertainties related to bone mechanical properties behavior, load positioning, and bone geometry among patients, here we performed a finite element analysis based on an idealized geometric model of a femur. This model was obtained from a 3D reconstruction of a collection of images obtained from CT scans accessed online through the 3D Content Central repository (http://www.3dcontentcentral.com/default.aspx). This model was scaled (Fig. 1), assuming the same external proximal femoral morphology, according to anthropometric data for children at the ages of 5, 7, 10, and 11 [4–6]. Global Coordinate system was aligned according to the one used by Gómez-Benito et al. [14]. Thus, the longitudinal axis of the femur was assigned to the *Y*-axis; the inplane axis orthogonal to it was assigned to the *X*-axis; and the perpendicular one to both *X* and *Y* axes was assigned to the *Z*-axis. This global coordinate system was also maintained in the definition and application of loads, as in Gómez-Benito et al. [14]. Model scaling was performed simultaneously in all three directions (*X*, *Y* and *Z*) in order to maintain the aspect ratio.

On this model, and following a similar methodology from a previous work [7], we generated different growth plate geometries (Fig. 2) in a 3D CAD program (SolidWorks, Dessault Systèmes, Waltham, MA USA) based on observations made by Kandzierski et al. [3].

All physes had a constant thickness of 1.2 mm and a physis-diaphysis angle (PDA) of 41.9° [7]. All growth plate geometries meshing and FE analysis were performed on ANSYS 14.5 (ANSYS, Inc. Canonsburg, PA USA). Femoral head, physis, and femoral shaft were meshed with 10 node structural elements (SOLID92). The ring of LaCroix was meshed with 4 node shell type element (SHELL41). We performed a convergence analysis in order to determine the adequate number of elements for model solution. Table 1 shows the number of elements employed for growth plate models.

Several studies have reported the anisotropic behavior of cortical and trabecular bone [8,9], characterizing mechanical properties such as strain and compression [10,11]. However, since growth's plate anisotropic nature has not been completely elucidated [12,13], mechanical properties herein were considered linear, elastic, and isotropic [14,15] as follows: growth plate Young's modulus $E_{\text{physis}}=5$ MPa, Poisson's coefficient $\nu_{\text{physis}}=0.45$ [14,15]; cortical bone Young's modulus $E_{\text{cortical}}=17.2$ GPa, Poisson's coefficient $\nu_{\text{cortical}}=0.2$ [14,15]; trabecular bone Young's modulus $E_{\text{trabecular}}=7$ GPa, Poisson's coefficient $\nu_{\text{trabecular}}=0.2$ [14]; and ring of LaCroix Young's modulus $E_{\text{ring}}=775$ MPa, Poisson's coefficient $\nu_{\text{ring}}=0.3$ [14].

Mechanical load on hip joint depends on body mass and physical activity. This load results from abductor and adductor muscle action, in addition to proximal femoral epiphysis contact force by the acetabulum. Mechanical load can be represented with sufficient accuracy as concentrated forces [16,17]. Thus, in this study forces considered for simulations were applied as concentrated on the femoral head ($\vec{P}_{acetabulum}$) and major trochanter ($\vec{P}_{abductor}$) (Fig. 3, Table 2). Such forces corresponded to rectangular components of reaction forces generated by the abductor muscles ($\vec{P}_{abductor}$) and the contact force produced by the acetabulum ($\vec{P}_{acetabulum}$), in the following stances of the support phase of the gait cycle as modeled by Gómez-Benito et al. [14]: *Heel Strike*, *Midstance* and *Toe Off*. Force application points were maintained constant for each case and stance to allow comparison among obtained results. Thus, (X, Y, Z) rectangular components of the aforementioned forces are expressed in Table 2

Where BW represents body weight in Newtons (N). Forces were escalated according to corresponding physiological body weights for each age (Table 3) [18].

Boundary conditions were applied to the distal nodes of each computational model as follows: total movement restriction on the leftmost node and partial restriction on remaining nodes (Fig. 3). Based on force values obtained from Table 2, tissue mechanical properties, and problem boundary conditions, we calculated stress values on the growth plate.

We used the equivalent Von Mises stress to evaluate physeal stress [19] despite the fact that it is usually employed as a yield criterion for ductile, elastic, isotropic materials that exhibit a linear behavior. The reasons behind this decision are that we modeled tissue behavior as linear, elastic and isotropic (as stated above), and because of the fact that this stress, which is a scalar magnitude proportional to the cartilage energy of distortion, indicates potential areas of failure in a domain, in this case the growth plate. Mathematically, Von Mises stress gathers into a single value all the information regarding the state of stress at a given point of the domain. Therefore, from the stress tensor present in each point of the domain, we calculated the invariant J_2 of this tensor [19], from which the equivalent stress is obtained from the following equation:

$$\sigma_{VM}=\sqrt{3J_2}\ ; \tag{1}$$

where σ_{VM} is the Von Mises equivalent stress and J_2, the deviatoric component of the stress tensor. In addition, Von Mises stress can be related to octahedral shear stress in the following way:

$$\sigma_{VM}=\frac{3}{\sqrt{2}}\tau_{oct}\ ; \tag{2}$$

where σ_{VM} is the Von Mises equivalent stress and τ_{oct} is the octahedral shear stress. Furthermore, due to the relationship

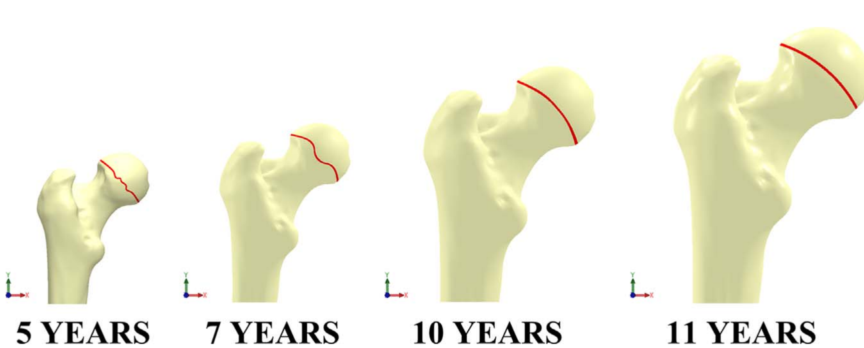

Fig. 1. Frontal view of the geometrical models for the proximal femur, scaled according to age.