COMPARING THE EFFECTS BETWEEN ORTHOTROPIC AND ISOTROPIC FINITE ELEMENT BONE MODELS ON HIP JOINT STRESSES DUE TO CAM FEMOROACETABULAR IMPINGEMENT

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INTRODUCTION
Cam type femoroacetabular impingement (FAI), characterized by an enlarged aspherical femoral head-neck junction, has been recognized as a pathomechanical process associated with osteoarthritis (OA) [1, 2]. The selection of material properties for modelling the hip joint in finite element analysis (FEA) has been long disputed and often depends on the physiological applications and mechanical assembly. Linear elastic isotropic models have been used in more recent studies for quasi-static conditions under low frequencies, where the femur has been considered as two separate, linear elastic, isotropic entities representative of the cortical and trabecular structures [3, 4]. It was further suggested that trabecular bone can be neglected in FEA, due to its low functional stiffness in the joint assembly and also in attempts to reduce processing time [5].

Since bone has been long considered to be an anisotropic composite, it was postulated that the femur would be better modelled in FEA as orthotropic rather than isotropic at physiological range of motions [6, 7]. Orthotropic properties could account for both cortical and trabecular structures in the femur, thus eliminating the need to segment the trabecular component and alleviate the overuse of computer memory. Very little work has been established to compare and discuss the effects with using an orthotropic femur model. The aim of this study was to examine the stress distributions in the hip joint with cam FAI and compare three different hip models: (1) solid orthotropic femur, (2) isotropic model comprised of both cortical and trabecular bones, (3) isotropic cortical shell.

METHODS
Patient-specific hip joint reaction forces were acquired from a previous study that examined the effects of cam FAI on pelvic hip motion [8], where two patients with severe cam FAI (with alpha angles greater than 80° [2]) from the initial study were selected for this FEA. The intersegmental forces were calculated, using inverse dynamics, from 3-D joint kinematics and kinetics for a dynamic squat motion. The matching patient-specific pelvic CT data were acquired and compiled using 3D-DOCTOR (Able Software Corp, MA, USA) to manually segment the symptomatic femur and pelvis, in addition to the contours of the trabecular structures. A cartilage layer of varying thickness was created using an offset method extruded from the acetabulum. Subsequently, the objects were all resurfaced using SolidWorks (Dassault Systèmes, MA, USA) to minimize geometric artefacts and reduce the number of triangular surfaces. The models were assembled in ANSYS (ANSYS, PA, USA) for FEA and meshed with tetrahedral SOLID187 elements. In the first case, bone was modelled as a linear elastic orthotropic solid, with material properties defined by \( E_1 = 11.6 \text{ GPa}, E_2 = 12.2 \text{ GPa}, E_3 = 19.9 \text{ GPa}; G_{12} = 4.0 \text{ GPa}, G_{13} = 5.0 \text{ GPa}, G_{23} = 5.4 \text{ GPa};\) \( v_{12} = 0.42, v_{13} = v_{23} = 0.23 \) (where \( 1, 2, 3 \) denotes the radial, tangential, and longitudinal directions, respectively) [6, 7]. In the second case, cortical bone was modelled as a linear elastic isotropic model, defined by \( E = 17 \text{ GPa} \) and \( v = 0.3 \) [5, 9]; and trabecular bone was defined by \( E = 1 \text{ GPa} \) and \( v = 0.3 [9] \). The cortical shell in the third case maintained the same geometry and material properties as the second case. Cartilage was assumed as a linear elastic isotropic material for all cases [3].

The intersegmental forces were applied through the hip joint centre of each respective model, where two quasi-static scenarios were considered: (1) stance and (2) maximum force endured during the impinged squat. Using the patient-specific kinematics data, the femur was oriented with respect to the pelvis according to the squat interval. Maximum-shear stress (MSS) was determined at the surface of the acetabular cartilage and on the underlying acetabulum to determine the adverse loading conditions within the symptomatic hip joint.

RESULTS AND DISCUSSION
In terms of computer processing times from the software, the orthotropic models required substantially less time to reach convergence (9 times faster than the two-component isotropic models and 3.5 times faster than the cortical shell models). During the standing position, MSS distributions on the cartilage were similar for all three models, with the peak MSS localized postero-superiorly (Figure 2). The orthotropic model’s peak MSS magnitudes on the cartilage and underlying acetabulum were slightly higher than the peaks on the other two models (Figure 1 and 2). The rigidity of the orthotropic model in the longitudinal direction created a stiffer geometry during standing conditions, thus produced a higher peak MSS and more evenly distributed MSS on the acetabulum, in comparison with the other two models (Figure 2).
Increases in MSS on the cartilage, when proceeding from the standing to the squatting position, were marginal. The orthotropic model’s peak MSS on the cartilage were similar to the peaks on the other models, localized at the antero-lateral regions of the cartilage. However, substantially higher peak MSS magnitudes were found behind the acetabular cartilage layer (Figure 1), localized at the underlying antero-lateral acetabulum, during the squatting position.

**Figure 1:** Comparison of peak maximum-shear stress (MPa) on the cartilage and on the acetabulum during the standing and squatting positions for each assembly.

The orthotropic model’s peak MSS on the acetabulum was within range of the other two models. During the squatting position, the hip oriented the forces antero-superiorly, causing the cam deformity to come into contact with the hip, which may affect the less rigid isotropic cortical shell model to deform considerably, thus exerting a higher contact load and overestimating the peak MSS.

**CONCLUSIONS**

The solutions for the orthotropic model required considerably less time to reach convergence than the cortical shell model, proposed by Anderson and associates as a means to reduce solution time [5]. Currently, with little knowledge on the material properties of the cam lesion, the orthotropic models were able to function under physiological loading conditions, suggesting its suitability towards hip modelling and predicting the locations of peak mechanical stimuli in the joint [6]. Although the resultant magnitudes were not representative of in vivo contact stresses, the results indicated the relative joint reactions due to cam FAI. The inclusion of soft tissues and muscle forces may yield a higher MSS and a more accurate result of contact stresses. In the three hip joint assemblies, the resultant peak MSS were found to be much higher during the squatting position, agreeing with a recent study that implemented an idealized parametric FAI model [3]. Moreover, the peak MSS for all three assemblies were found on the surface of the acetabulum, re-iterating the postulated theory that mechanical factors of joint degeneration may be due to the local stiffening of the underlying subchondral plate [10] and perhaps not limited to direct shear stresses to the articulating cartilage. This further suggests that cartilage delamination and the onset of OA would be secondary to the altered rate of bone remodelling and consequent changes in the subchondral bone’s apparent density, and thus its stiffness.

**REFERENCES**