DEVELOPMENT OF AN ORTHOTROPIC BIOFIDELIC MODEL OF THE WHOLE FEMUR

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INTRODUCTION
Bone adapts its architecture in response to the loading it is subjected to, resulting in an optimized structure [1]. This mechano-adaptation is mediated by osteocytes [2], which are thought to be activated by strain-driven flow caused by changes in their surrounding strain gradient as a result of an ever varying loading environment [3]. This optimization is driven towards a remodelling plateau around a target strain where no bone is either gained or lost [4]. Recently, finite element studies have incorporated remodelling algorithms in an attempt to simulate bone’s response to altered loading conditions, such as bone-implant interfaces [5], or its morphogenesis [6]. In order to simplify these analyses, bone is usually considered to be isotropic. However, this assumption does not explain the directionality of its internal structures that have been observed [7] or the orthotropic properties that have been measured in other studies [8]. The remodelling signal is dependent on bone’s loading configuration and studies have shown that the inclusion of muscle forces is a key factor in examining the femur [9]. The use of simplified loading, although time efficient, is considered to be unrealistic [10] and the need for more physiological loading has been stressed [11]. A free boundary approach to modelling the pelvic and femoral constructs produced more physiological stress and strain distributions and was considered to be a closer approach to modelling the in vivo environment [12, 13]. Previous work by the authors [14] introduced a 3D bone remodelling algorithm with bone modelled as a strain-adaptive continuum with local orthotropic material properties. This approach showed promising results, with stiffness distribution and bone structure directionality being physiologically predicted for the proximal femur, under simplified loading. This study describes the algorithm’s application to a more physiological, free boundary model of the femoral construct. Predictions for the proximal femur and condyles compare well with previous clinical observations of material property and directionality distributions. This indicates that the loading conditions modelled are closer to the ones experienced in vivo in comparison to fixed boundary condition models. Development of this approach to modelling and bone structure prediction can lead to a better understanding of bone’s mechanical behaviour. The development and public release of orthotropic heterogeneous models for different constructs can be of great interest to researchers in orthopaedic biomechanics.

METHODS
A 3D orthotropic bone remodelling algorithm [14] was applied to a free boundary condition model of the femur [13] modified at the knee joint and undergoing single-leg stance with a body weight (BW) load applied at the L5S1 joint. A complete heterogeneous orthotropic model of the femur was produced with artificial hip and knee joint structures, muscles and ligaments explicitly included according to the linear muscle model proposed [13]. The geometry was extracted from the Muscle Standardised Femur [15] and meshed with 326026 tetrahedral C3D4 elements with the same initial local orthotropic material properties (E1, E2, E3 = 3000 MPa, ν12, ν13, ν23 = 0.3, G12, G13, G23 = 1500 MPa) and orientations. An isotropic elastic layer representing the cartilage was included at the femoral head and the condyles (E = 10 MPa, ν = 0.49). Small sliding penalty contact conditions were applied to both the hip joint and knee joint surfaces. At each iteration, every element’s strains and stresses were extracted and processed. The local element material orientations were matched with the local principal stress orientations and their associated strain stimuli found. The local directional Young’s moduli of the elements outside the remodelling plateau of 1000-1500 µstrain were then updated proportionally to the absolute value of their associated strain stimuli, in order to achieve a target normal strain value of 1250 µstrain, E1, E2 and E3 were limited between 10 MPa and 20 GPa, and associated with the minimum, medium and maximum principal stresses respectively. Poisson’s ratios were restricted [16] and shear moduli taken as a constant fraction of the average of the appropriate Young’s moduli. The model was run until convergence was achieved [14].

RESULTS AND DISCUSSION
Convergence was achieved after the 49th iteration with all femoral strains below 2600 µstrain, in agreement with [13]. The predicted hip contact force was 211% BW and the knee contact force 263% BW, in the range of the contact forces measured by Bergmann [17] and D’Lima [18], respectively. Furthermore, a 51.3%-48.7% medial-lateral split in joint reaction forces in the tibial tray was found, matching the data collected [18]. Fig. 1 shows the material orientations associated with E1 and E3 with compression in blue and tension in red, and line thickness, width and colour proportional to the local Young’s moduli values for each element. The areas where medium or high stiffness values can be found for the proximal femur (top) are the great trochanter, along the lateral epiphysis, towards the articular surface of the femoral head, the inferior side of the femoral neck, the calcar femorale [19] and along the surface of the diaphysis (cortical region of the femoral shaft).
These are areas of high compression or tension stresses due to bending of the proximal femur because of the application of the hip contact force and the action of the attached muscle groups on the femur [1, 20]. The lower value regions observed included the Ward’s triangle, Babcock’s triangle and the intermedullary canal, regions known for either being composed of thin and loosely arranged trabeculae or where virtually no trabecular bone can be found [21]. The main documented trabecular groups [21] are also clearly represented. In blue (i.e. compression), the primary and secondary compressive groups; in red (i.e. tension) the primary and secondary tensile groups as well as the great trochanter group. These arise as a structural response to the necessity to transfer load along the femur from an oblique to vertical direction [19]. Finally, the perpendicular arrangement of trabeculae along the articular surface of the condyles (bottom) is clearly present, particularly in the medial side, as response to the compression caused by the weight bearing function of the femur [19]. The high stiffness distribution around the epiphyseal line is a result of the thicker and coarser trabeculae that can be found in this region; in contrast, finer and denser trabeculae were also observed towards the metaphysis. Lastly, the trabeculae radiating from the intercondylar notch towards both condyles were also predicted. The low stiffness distribution in the lateral condyle matches the lower density expected for the region [23].

Because of the orthotropic assumption, the intersection of trabeculae in the region below the epiphysis can be seen to be orthogonal. Although this agrees with Wolff’s trajectorial theory, other studies have measured the angle of decussation to be acute [19, 20]. This is the limitation of assuming bone to be a continuum with local orthotropic symmetry. Nevertheless, orthotropy has been shown to be a closer approximation to bone’s anisotropy [22]. The use of a single load case is limiting when describing the complex mechanical environment the whole femur is subjected to physiologically. Inclusion of more load cases for a variety of activities will allow for a more accurate prediction of the distribution of the mechanical properties and associated orientations, particularly in the condyles, the distal part of the femur adapted to both mechanical compression and rotary movements [23].

CONCLUSION
This study combined two developed approaches: considering bone to be a strain-adaptive continuum material with local orthotropic symmetry [14] and inclusion of free boundary conditions in the model [13]. The stiffness distribution and structure directionality for the whole femur can be correctly assessed, with results approaching those observed and measured in vivo. Further developments to this method will be the inclusion of more load cases, thus creating a loading envelope more representative of what bone is subjected to in vivo. Nonetheless, the models obtained from this approach can be applied in structure and directionality dependent research areas such as fracture mechanics (with applications in impact protection devices and blast biomechanics) and bone-implant design improvement.

REFERENCES