COMBINED MUSCULOSKELETAL AND FINITE ELEMENT MODELLING OF THE FEMUR

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SUMMARY

In the 1870's, Wolff formulated a 'trajectory theory' about trabecular bone architecture which can be succinctly written as follows: bone adapts its structure to loading conditions in a way that follows principal stress trajectories. In this study, it was assumed that the human femur is optimally adapted to the loading conditions experienced during daily activities such as walking or climbing stairs. Hence, an initially randomized structural mesoscale model of a femur was iteratively adapted to the loading conditions experienced during a range of daily activities. The resulting structure shows a good visual comparison with clinical observation and the model proved computationally efficient.

INTRODUCTION

As bone is responsible for load bearing throughout the body, a comprehensive knowledge of its structure allows for estimations of stress and strain distributions within the skeletal system. This in turn informs prediction, prevention, or repair of skeletal disorders and design of protection equipment as well as prosthetics. Although bones are highly anisotropic porous structures, it has been common to use a continuum level approach when attempting to resolve the distribution of dense cortical and porous trabecular bone within long bones such as the femur. Macroscale continuum models can run in a matter of minutes but present a limited resolution and overlook the anisotropic properties of bone [1]. On the other hand, microscale models allow for good resolution, but they are extremely computationally demanding [2]. Structural approaches idealise bone as a combination of structural elements with the aim of combining resolution with efficiency. The structural optimisation approach consists in predicting the structure of bones based on the assumptions that bones are optimised structures, and that they adapt in response to the load environment towards a target stress or strain state. This is the basis of the 'mechanostat' introduced by Frost [3]. The aim of this study was to build a predictive mesoscale structural model of the human femur [4]. A preliminary study was conducted and the results are presented.

METHODS

Loading conditions:

Gait cycles describing walking, stair climbing and sit to stand, were recorded on a volunteer. A 10 infrared camera 3D motion capture system (*Vicon*) was used to track reflective markers positioned on bony landmarks of the subjects according to the ISB recommendations [5]. A force plate recorded the ground reaction force. An *OpenSim* [6] musculoskeletal model of the lower limb [7] based on an anatomical dataset [8] was used to simulate the gait cycles recorded. Using static optimisation of muscle recruitment, muscle forces were computed and loading conditions experienced by the bones were derived.

Structural femur model:

Based on a CT scan of a *Sawbones* femur (#3403), a semi-structured node point cloud was obtained from a volumetric mesh made of 113103 four noded tetrahedral elements using *Mimics*. The surface nodes were used to generate 10410 shell elements representing cortical bone in a finite element (FE) model. 218717 truss elements were generated between pairs of nodes within the volume to model trabecular bone, by connecting each node to at least its 16 closest neighbours. The loading conditions previously derived were applied to the FE model. 'Load applicators' made of beam and layered solid wedge elements were designed to apply the joint loads (defined at the centre of the joint by *OpenSim*) as well as the inertial forces (necessary to maintain the equilibrium in the model). The strain distribution was analysed using *Abaqus*. Having defined a target strain from the literature, the arbitrary initial thicknesses of the shell elements (0.1mm) as well as the radii (0.1mm) of the truss elements were adapted to direct the strain towards the target. This operation was iterated until convergence, according to the methodology proposed by Phillips [4].

Analysis of the features of the model:

A *MATLAB* algorithm was written to compute bone density following a cuboid partial volume method; the entire bone was partitioned into identical

1.65×1.65×1.65mm cubic regions. The ratio of bone volume to cube volume was estimated for each of them. To display the results, the tetrahedral elements of the *Mimics* mesh were once again introduced and were assigned a porosity value depending on the position of their centroid within this partition. Results were consistent with literature although further work is required to validate the method.

RESULTS AND DISCUSSION

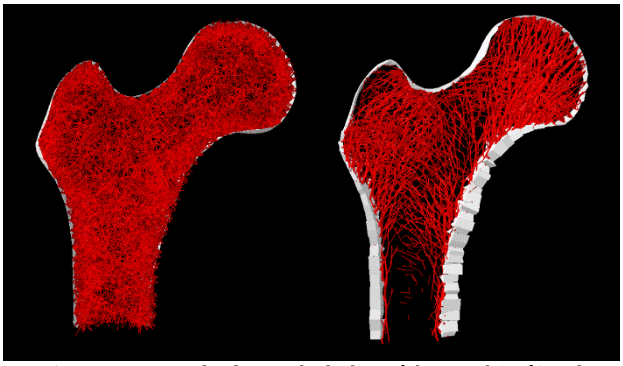

Figure 1: Longitudinal 5mm thick slice of the initial (Left) and converged (Right) proximal femur. Trabecular and cortical bone are displayed in red and white respectively.

Convergence was reached after 20 iterations. The converged model effectively contained 176688 trabecular elements. It presented a total volume of cortical and trabecular bone of 112,059mm^3 and 47,391mm^3 respectively. Shell element thickness ranged from 0.1mm to 8mm (mean 1.8mm, STD 1.7 mm). Trabecula radius ranged from 0.1mm to 1.4 mm (mean 0.13mm, STD 0.08mm).

Figure 1 displays a longitudinal slice of the initial and converged proximal femurs. When comparing the converged proximal femur with clinical images, such as the proximal femur CT scan in Figure 2 (Left), a good comparison is observed in the distribution of the cortical layer thickness. Similarities are particularly noticeable in the proximal extremity of the shaft.

The main trabeculae groups identified in the proximal femur by numerous studies [9, 10] can also be observed in the converged result of this model, such as the greater trochanter group, the primary compressive group from the hip to the medial cortex, and the primary tensile group arching between the lateral cortex and the femoral head.

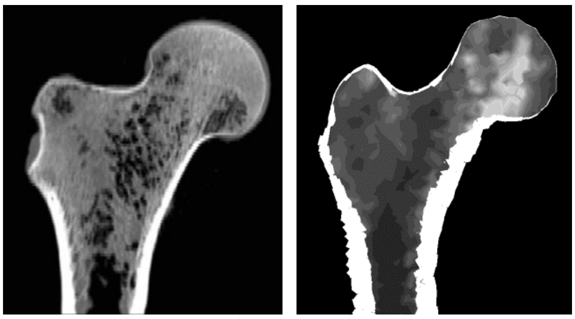

Figure 2: (Left) CT scan of a proximal femur; (Right) Normalised density plot (Black=0; White=1) for the converged model

Figure 2 (Right) displays the computed normalised density of the trabecular bone in the proximal femur. The stiffening observed in the femoral head is consistent with clinical observations, and so are the comparatively porous lateral part of the proximal femur, and the relatively empty regions in its centre and in the shaft. Features of the shaft and the distal end of the femur showed a good comparison with clinical observations as well. Some limitations to this study have to be mentioned. Among them is the use of truss elements to model trabecular bone, which restricted the choice of strain stimulus to a single axial measure. In vivo, trabeculae would also be subjected to shear and bending actions. Further work on the model will include the replacement of truss elements by beam elements. Further limitations concern the representation in the FE model of the loads derived by muscles whose paths include wrapping surfaces and via-points, e.g. the *gluteus maximus*, as the force directions derived from their bone attachments will be representative of their anatomical but not of their effective lines of action. This issue is overlooked in many musculoskeletal models. In addition, it would be beneficial to define an exhaustive, or at least comprehensive, list of the daily activities influencing the femur structure. Indeed, the present study already shows significant differences in the resulting structure compared to the preliminary study done by Phillips [4] using a simple case of single legged stance: the number of effective trabecular elements was then about 64,000, and the volume of trabecular and cortical bone, calculated using the method outlined here, were about 17,000 mm^3 and 82,000 mm^3 respectively.

Applications for such a model include simulation of fracture for the prediction of failure load and fracture patterns. Rapid prototyping based on the predicted bone structure could provide limb surrogates for mechanical testing. On a longer term, implications of this study could inform the design of prosthetics and scaffolds for tissue engineering.

CONCLUSION

Structural optimisation has been used to predict the structure of the human femur based on the loading conditions experienced during daily activities. The resulting model shows good visual comparison with clinical observations. It is also computationally efficient. There are thus strong indications that this type of modelling has the potential to become a powerful tool in biomechanics research as well as in a range of experimental and clinical applications.

REFERENCES

1. Phillips, A.T.M, *Medical engineering & physics,* 2009. **31**(6): p. 673-680.
2. Tsubota, K., et al., *Journal of biomechanics*, 2009. **42**(8): p. 1088-1094.
3. Frost, H., *The anatomical record*, 1987. **219**(1): p. 1-9.
4. Phillips, A.T.M, *Engineering and Computational Mechanics*, 2012. **165**: p. 147-154.
5. Wu, G., et al., *Journal of biomechanics*, 2002. **35**(4): p. 543-548.
6. Delp, S.L., et al., *Biomedical Engineering, IEEE Transactions on*, 2007. **54**(11): p. 1940-1950.
7. Modenese, L., *Journal of Biomechanics*, 2011. **44**(12): p. 2185-2193.
8. Klein Horsman, M.D., et al., *Clinical Biomechanics*, 2007. **22**(2): p. 239-247.
9. Singh, M. A.R.et al., *The Journal of Bone & Joint Surgery*, 1970. **52**(3): p. 457-467.
10. von Meyer, H., *Arch Anat Physiol Wiss Med*, 1867. **34**: p. 615-628.