Spatial Influence Function Response to Multi-load Case Bone Remodelling Simulation with Contact Interaction

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Summary

This is the first study to implement a strain energy density (SED)-driven bone remodelling algorithm combined with a spatial influence function for a full 3D bone. Subject-specific gait data are used to define a multi-load case, and finite-element (FE) modelling simulations run to predict proximal femoral density distribution. Density prediction trends agree with radiography.

Introduction

SED-driven algorithms combined with the spatial influence function in adaptive FE methods enable simulation of the bone remodelling process [1]. However, no studies have implemented this approach in full 3D bone models and with realistic contact boundary conditions at joints and loading conditions stemming from muscle forces. This programme of work aims to investigate the bone mineral density (BMD) of the femur under loads experienced by above-knee amputees with different types of prosthetic sockets and implants. Here, a sensitivity analysis was conducted to evaluate the effect of the spatial influence function on BMD in a model of a femur.

Methods

Subject-specific MRI and gait data of an able-bodied participant (28 years, male, 192 cm height, 85 kg mass) were used as inputs. The FE model consisted of the femur, surrounding soft tissue, and acetabulum (Figure 1a), with the remaining leg defined as a rigid body. The boundary conditions were set for five different load cases (heel-strike/HS, foot-flat/FF, midstance/MS, heel-off/HO, and toe-off/TO) during gait. Force and moment measurements were calculated using inverse dynamics musculoskeletal modelling (Table 1).

Table 1: Knee joint reaction (KJR) force (N) and moment (Nm)

	HS	FF	MS	HO	TO
F med(+)/lat(-)	-10.2	70.9	77.7	84.5	-3.27
F post(+)/ant(-)	-64.0	-96.6	25.9	128	18.1
F sup(+)/inf(-)	88.9	641	632	785	136
M sagittal	-40.9	10.3	-12.1	-3.33	35.6
M frontal	15.2	-5.59	-7.92	-13.1	7.48
M transverse	2.86	-1.91	1.82	2.07	0.58

Fixed displacement was set for proximal soft tissue and a specific pose per load case defined by the hip joint angle. Soft tissue was assigned hyperelastic material properties (Mooney Rivlin; C_{10}=85kPa; C_{01}=21.4kPa). The governing rule for bone density adaptation, ρ was defined by Equation 1, with parameter values derived from established literature [2].

$$\frac{d\rho(x)}{dt} = \tau \sum_{j=1}^{N} f_j(x)\left(\frac{SED_j}{\rho_j} - (1 \pm s)k\right) \qquad (1)$$

The function $f_j(x) = e^{-dist(x,j)/D}$ represents Mullender's spatial influence function [1], which characterizes the cellular communication network among osteocytes. τ=1 is a fixed time constant, k=0.0145 J/g is a constant reference value, and s=0.1 is a lazy zone parameter. Two distance parameters (D_1=0.2mm and D_2=0.5mm) were applied for sensitivity analysis. Changes in BMD were translated into changes in mechanical properties, E=3790ρ^3 MPa for the next iteration. The load cases were applied in consecutive analysis steps of bone remodelling, following HS, FF, MS, HO, and TO before repeating successively.

Results and Discussion

Figure 1b shows BMD after iteration 125, which covers the primary compressive and primary tensile arcade, ensuring alignment between the computational model and bone tissue behaviour shown on radiographs (Figure 1c [3]). A larger distance parameter increased connectivity between neighbouring elements. If surrounding elements resorb, adjacent elements tend to follow, as observed in the case of D_2. To ensure trabecular distribution at the secondary arcade, muscle contractions, for example the dominant muscle at the greater trochanter, should be added.

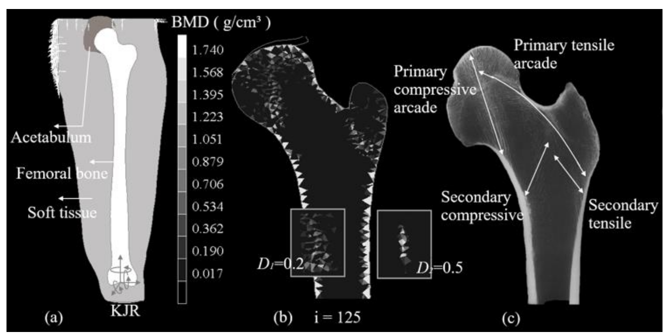

Figure 1: (a) 3D model, (b) BMD, and (c) radiograph image

Conclusions

The integration of spatial influence functions in bone remodelling simulation provides a promising approach for accurately predicting BMD distribution. This macroscopic model has the potential to simulate contact interactions, such as BMD changes from stump-prosthetic interaction.

References

[1] Mullender et al (1994) *J Biomech*. **27**(11): 1389-1394
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[3] Jacobs et al (1997) *J. Biomech*. **30**: 603-613