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► To cite this version:

Jamie Frame, Pierre-Yves Rohan, Laurent Corte, Rachele Allena. Simulating the Remodelling of Bone around Implants. 23rd Congress of European Society of Biomechanics, Jul 2017, Séville, Spain. hal-02269991

HAL Id: hal-02269991

<https://hal.science/hal-02269991>

Submitted on 23 Aug 2019

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SIMULATING THE REMODELLING OF BONE AROUND IMPLANTS

Jamie Frame (1), Pierre-Yves Rohan (1), Laurent Corté (2,3) and Rachele Allena (1)

1. LBM – Institut de Biomécanique Humaine Georges Charpak, ENSAM Paris, France

2. Centre des Matériaux – CNRS : UMR7633, MINES ParisTech, PSL Research University, France

3. Matière Molle et Chimie – CNRS : UMR 7167, ESPCI Paris, PSL Research University, France

Introduction

Improper osseointegration of implants leading to poor mechanical anchoring or embrittlement of neighboring bone is a major concern in orthopedic surgery [1?]. This integration is known to depend on the complex interplay between the mechanical environment and the cell activity in the tissues surrounding the implant. In order to accurately predict the success of an implant a robust description of the remodeling behavior of bone is required. Building upon previous research modeling osteogenesis around implants [2] a mechano-biological Finite Element (FE) model is proposed to describe the remodeling processes involved when bone, cartilage and fibrous tissue are submitted to mechanical loads.

Method

In this work, we describe the mechanostat (the interrelationship between loading conditions and remodeling) of bone [1] by modelling the net effect of cellular activities at the tissue level. For that, we distinguish the immature tissues resulting from the early proliferation steps (growth and diffusion) from the mature tissues obtained after a consolidation of the extra-cellular matrix (mineralization for bone). In each elementary volume element, the creation of new (immature) tissue is dependent upon the level of applied strain and is described by a reaction-diffusion equation as follows

$$\frac{\partial \varphi_i^{im}}{\partial t} = (1 - \varphi_{TOT})D\Delta\varphi_i^{im} + \alpha(1 - \varphi_{TOT})\varphi_{TOT}h(\varepsilon_i^G)h(tact_i) - \beta\varphi_i^{im}h(\varepsilon_i^R) - \gamma h(tact_i^M)\varphi_i^{im} \quad (1)$$

where φ_i^{im} is the immature tissue volume fraction (subscript i refers to bone, cartilage or fibrous tissue), φ_{TOT} is the total volume fraction occupied by tissues, D is the diffusion tensor, α , β and γ are the rates of immature tissue growth, resorption and maturation, respectively, ε_i^G and ε_i^R are the normalized strain thresholds required for immature tissue to grow or resorb, respectively, and h represents a special Heaviside function. The function $tact_i$ represents the activation time for each tissue type and is defined as

$$\frac{\partial tact_i}{\partial t} = h(\varepsilon_i^G)h(tact_i^U) - h(\varepsilon_i^R)h(tact_i^L) \quad (2)$$

where $tact_i^U$ and $tact_i^L$ are the upper and lower bounds for the activation time, respectively.

An activation time $tact_i^M$ is required to begin differentiation into mature tissue. The rate of change of mature tissue volume fraction is expressed as

$$\frac{\partial \varphi_i^M}{\partial t} = \gamma h(tact_i^M)\varphi_i^{im} - \beta \varphi_i^M h(\varepsilon_i^R) \quad (3)$$

[A sentence is missing here to explain how volume fractions are related to the mechanics no?]

Results

Using these equations a simple cantilever cyclic bending simulation was created and loaded to recreate a range of physiological strains experienced during bone remodeling.

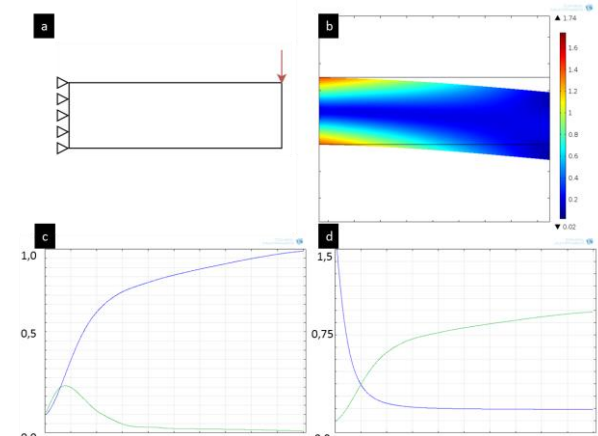


Figure 1: (a) The cantilever boundary conditions and (b) the maximum normalized shear strain distribution after 5 hrs producing (c) the evolution of immature (green) and mature (blue) bone tissues and (d) the evolution of normalized Young's modulus (green) and maximum normalized shear strain (blue) over time.

Preliminary results for bone tissue only are presented in Figure 1. This shows the cantilever boundary conditions and maximum normalized shear strain distribution which produce the evolution of immature and mature bone tissues over time. As the Young's modulus increases proportionately with the increase in mature tissue density the strain under constant loading conditions is observed to reduce, therefore altering the generation of new tissue types.

The model proposed here may offer numerous perspectives as a predictive tool for implant design or for the new therapies against bone resorption.

References

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2. Schmidt et al 2015. Diffusion model to describe osteogenesis within a porous titanium scaffold, *Computer methods in biomechanics and biomedical engineering*.

Acknowledgements

Funding by the Fédération Francilienne de Mécanique (Coup de pouce 2015) is acknowledged.