

EXPERIMENTAL AND NUMERICAL MODELS IN ORTHOPAEDIC BIOMECHANICS

CARMINE PAPPALETTERE¹

Abstract: *The article outlines the main results of different research activities, in the field of Orthopaedic Biomechanics, recently conducted at the Experimental Mechanics Laboratory, Politecnico di Bari.*

Key words: *Bone Mechanics, Mechanobiology, Mechano-Regulation Algorithm, Finite Element Method.*

1. Introduction

Faced with the task of understanding a complex system, it is useful to extract its most essential features and use them to create a simplified representation of the system or, a 'model' of the system. A model allows one to observe more closely the behaviour of the system and to make predictions regarding its performance under altered input conditions and different system parameters. Modelling is widely used in Biomechanics; both, physical models that use real constructions and mathematical models that use conceptual/virtual representations. Modelling plays a role of crucial importance especially in the field of the Orthopaedic Biomechanics where: (i) the morphology of the anatomic structures is highly irregular, (ii) the mechanical behaviour of materials is highly nonlinear, (iii) the physical, chemical and biological factors often involved in the regeneration/remodelling process of anatomic regions act simultaneously and are difficult to control [20].

The physical/experimental models adopt real constructions to assess the structural response of anatomic regions subjected to specific boundary and loading conditions. They reproduce in vitro mechanical/chemical/biological environment surrounding the anatomic region and try to assess how the response of the region itself changes for different input conditions. Experimental models in Orthopaedic

Biomechanics present often the limitation that the boundary and loading conditions simulated in vitro are significantly different with respect to those acting in vivo on the region under investigation. On the other hand, the mathematical models that allow to simulate the structural response of very irregular anatomic regions subjected to very complex boundary and loading conditions however, present the limitation that the solution they provide is the result of a mathematical formulation that never will be capable of correctly capturing the real physics of the problem.

Among the mathematical models, certainly the finite element models are those mostly utilized in Orthopaedics. They are numerical mathematical models – numerical because they rely on computers to find approximate solutions to large sets of equations. In 1972, about fifteen years after the finite element method (FEM) initiated a revolution in stress analyses of structures in engineering mechanics, this 'new method to analyse the mechanical behaviour of skeletal parts' [15], was first introduced in the orthopaedic literature. Different techniques of Experimental Mechanics have been employed in Orthopaedic Biomechanics: (i) ultrasounds for the assessment of the anisotropy of the cancellous and cortical bone [18]; (ii) strain gauges for in vivo [1] and in vitro [17] measurements of the strain in loaded bones; digital image correlation for the evaluation of

¹Dipartimento di Ingegneria Meccanica e Gestionale, Politecnico di Bari, viale Japigia 182, 70126 Bari - Italy

the structural response of trabecular bone samples under specific boundary and loading conditions [23] or for the contouring of anatomic regions [21]; structured light for the reconstruction of the morphology of the human head [24]; speckle interferometry for the reconstruction of dental elements [16]; projection moiré [25] and holography [22] for the high accuracy reconstruction of the shape of a human face, etc.

The article outlines the main results of different research activities, in the field of Orthopaedic Biomechanics, recently conducted at the Experimental Mechanics Laboratory, Politecnico di Bari. It will be structured into two principal sections: the first section will be devoted to the description of different physical models that have been developed to assess the stress state within different anatomical regions or orthopaedic devices; computational biomechanical models combined with finite element models will be the subject of the second section.

2. Experimental Models

In this section three different experimental models will be described that were developed to: (i) evaluate the error done in the reconstruction of finite element models from CT scan data; (ii) assess how the contact pressure distribution at the bone-implant interface can be measured with finite element models generated from CT data; (iii) estimate the in-plane displacement field in a transverse diametral section of the tibial plate of a knee prosthesis.

2.1. Evaluation of reconstruction errors in FEM models from CT-scan images

In general, CAD models can be generated from CT scans by following two distinct approaches: “geometry based” (GB) and “voxel based” (VB). The former method defines a geometric model comprised of curves and surfaces that is finally discretized into finite elements. The strength point of the geometry based approach is the possibility of creating smooth surfaces and hence to simulate any kind of interface. The voxel based approach is more

diffused than the geometry based approach and relies on the principle that each group of voxels (the base unit of 3-D imaging) is directly converted into hexahedral elements. The presence of “ramp effects” in the reconstructed contour surfaces makes it very difficult to simulate the actual behaviour of bone-implant interfaces.



Fig. 1. RP Cure 600 ND benchmark specimen built with stereolithography and region of interest considered in the study.

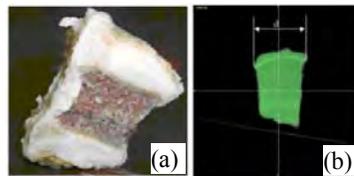


Fig. 2. (a) Lumbar rachis segment subjected to CT scanning and (b) resulting CT scan image

The aim of the study was to improve the overall efficiency of the geometry based approach by pursuing these three objectives: (i) to evaluate accuracy of FE meshes generated from CT scans; (ii) to individuate the most important sources of geometric errors and hence minimize their effects; (iii) to estimate the errors made in computing the stress state with FE models reconstructed from CT data and affected by known amounts of geometric errors.

Tasks (i) and (ii) have been accomplished by considering a benchmark specimen made of epoxy resin and manufactured using Stereolithography (SL) (Fig. 1). The SL model was CT-scanned and the corresponding FE model was hence reconstructed. Geometric errors entailed by the CT-FEM transformation process were evaluated and a methodology for minimizing these errors was developed. The successful application of this procedure to the generation of a finite element model of a human

lumbar vertebra reconstructed from CT-scans (Fig. 2) was documented. Finally, errors on computed stress values were evaluated by comparing results obtained from FE models reconstructed from CT data and their counterparts obtained from FE models entirely generated in virtual CAD environment [9].

2.2. Accuracy of FEM predictions on bone/implant interface contact pressures for models reconstructed from CT data

In vivo evaluation of stress state at bone/implant interface is a very complicated task. Finite element analyses can give indications on stress distributions in the contact region. However, FE predictions must be corroborated by experimental data or, at least, by comparing numerical results with analytical/theoretical solutions. The procedure presented in the study to evaluate accuracy of bone/implant interface FE models relies on a rather simple analytical formulation derived from Persson's general theory [19] of cylindrical contact adequately validated through experimental analyses. The process for evaluating FE models accuracy included the four steps listed below.

1. Formulation of the cylindrical contact problem between conformal surfaces (Fig. 3). Two different cases have been analyzed: a titanium (Ti) alloy pin ($E=110000 \text{ N/mm}^2$; $\nu=0.3$) and a chromium-cobalt (Cr-Co) alloy pin ($E=230000 \text{ N/mm}^2$; $\nu=0.3$).

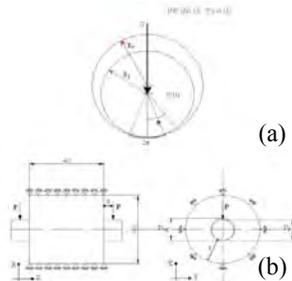


Fig. 3. (a) Principle scheme of Persson's theory for cylindrical contact; (b) Schematic of the pin/muff coupling analysed in the study.

2. Generation of an "ideal" FE model for the

cylindrical contact problem formulated in Step 1. This model - denoted as MCAD - reproduced the nominal geometry and was entirely constructed using virtual CAD tools. The model MCAD was experimentally validated.

3. The CAD model generated in Step 2 was given in input to a stereo-lithographic machine which built an epoxy resin muff. The muff was coupled with a metallic pin and the resulting assembly was submitted to CT scanning (Fig. 4). Scan data were used to reconstruct a new FE model, denoted as MCT.

4. Numerical results obtained from models MCAD and MCT were finally compared in order to estimate accuracy of FE predictions on contact pressure. Sensitivity of contact pressure computation errors to geometric errors included in the reconstructed model were also evaluated.

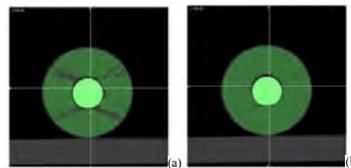


Fig. 4. CT scan images of the pin/muff coupling: (a) Cr-Co alloy pin; (b) titanium alloy pin.

The study demonstrated that it is possible to assess accuracy of finite element predictions on coupling models reconstructed from CT data by evaluating errors on contact pressure distribution. Remarkably, in the titanium pin case, a rather clear correlation was found between the geometric error made on reconstructed muff diameter and the error made in the computation of the maximum contact pressure. No clear correlation between geometric and stress errors was found in the chromium-cobalt pin case since the higher X-ray attenuation generated image artefacts thus preventing a correct reconstruction of coupling geometry [4].

2.3. Analysis of the displacement field in the tibial plate of a knee prosthesis.

An experimental model was developed to assess, through the moiré technique, the in-plane displacement field in a transverse diametral section of the tibial plate of a knee prosthesis (Fig.

5). A grating of points was inserted within a polyurethane disc – representing the simplified model of the tibial plate – and a stainless steel sphere was pushed against the disc. The in-plane displacement field of the transverse section placed in front of the CCD camera, was determined for different loads P (Fig. 5).

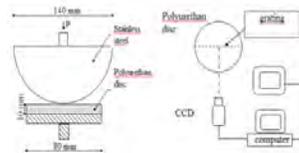


Fig. 5. Schematic of the experimental set-up used to assess the in-plane displacement field of a diametral transverse section of the tibial plate.

3. Numerical Models

This second section of the article describes how mechanobiological models can be utilized to predict the spatial and temporal patterns of the tissues differentiating within a fracture site during the healing process. Specifically, two examples will be given showing how a mechano-regulation model, can be utilized to assess bone regeneration in an osteotomized mandible submitted to distraction osteogenesis and in a fractured lumbar vertebra. The article will finally outline how a mechano-regulation model can be utilized to design a bone tissue scaffold as well as to optimize its morphology and performance.

3.1. Fracture healing

Fracture healing is a physiological process that initiates immediately after the fracture event and occurs by following a primary or a secondary fracture healing. Primary healing involves a direct attempt by the cortex to re-establish itself once it has become interrupted. Secondary healing involves responses within the periosteum and external soft tissues and subsequent formation of an external callus. Secondary fracture healing occurs in the following stages. Blood emanates from the ruptured vessels. Macrophages remove the dead tissue and generate initial granulation tissue for the migration of undifferentiated mesenchymal stem cells. These cells proliferate and migrate from the surrounding soft tissue.

Then, mesenchymal cells may differentiate into chondrocytes, osteoblasts or fibroblasts, depending on the biological and mechanical conditions.

In this study a mechano-regulation model was developed to predict the patterns of tissue differentiation formed in the fracture gap of an osteotomized mandible or of a vertebral body. The model assumes the bone callus as a biphasic poroelastic material. The biophysical stimulus S regulating the tissue differentiation process is hypothesized to be a function of octahedral shear strain γ and interstitial fluid flow velocity v .

3.2. Mechanobiology of the mandibular symphyseal distraction osteogenesis

Mandibular distraction osteogenesis is a common clinical procedure aimed to modify the geometrical shape of the mandible for correcting problems of dental overcrowding and arch shrinkage [2, 5, 13]. In spite of consolidated clinical use, the process of tissue regeneration and bone regrowth in an osteotomized region remains poorly understood. The mandible is first osteotomized and then instrumented by applying a distraction orthodontic device. After a latency period of seven to ten days, the distractor is progressively opened by giving the same daily expansion for between 5 and 10 days. The expansion process is then terminated and the level of aperture is kept constant in order to facilitate the formation of new bone tissue.

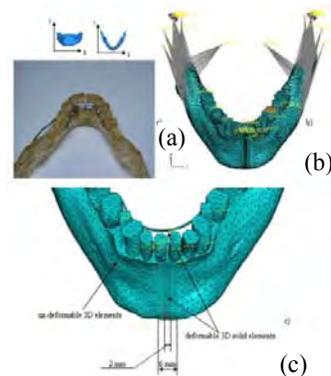


Fig. 6. Finite element model of the osteotomized mandible submitted to distraction osteogenesis

A 3D model of the mandible was reconstructed from CT scan data (Fig. 6) and meshed into finite elements. A distractor was also modeled. The model was used to determine (i) the optimal rate of expansion [3] (Fig. 7) (ii) the optimal duration of the latency period [6] and (iii) how the latency period changes for differently aged patients [7].

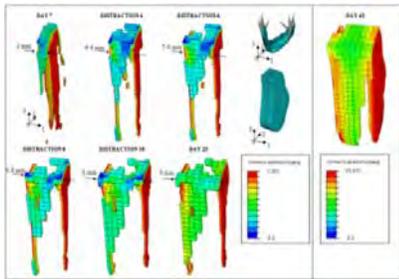


Fig. 7. Patterns of bone tissue forming during the distraction period.

3.3. Mechanobiology of fracture repair in vertebral bodies

A novel multi-scale mechano-regulation model is developed in order to investigate the mechanobiology of trabecular fracture healing in vertebral bodies. A model that, known the morphology of the micro-architecture of the spongy bone, allows to assess the risk of failure of vertebral bodies under specific boundary and loading conditions was developed in a previous study [8].

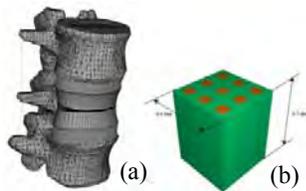


Fig. 8. Finite element model of the spinal segment L3-L4-L5 (macro-scale) (a) and of a portion of fractured trabecular bone (micro-scale) (b).

A macro-scale finite element model (Fig. 8a) of the spinal segment L3-L4-L5, including a mild wedge fracture in the body of the L4 vertebra, was used to determine the boundary conditions

acting on a micro-scale finite element model (Fig. 8b) of a portion of fractured trabecular bone. The micro-scale model, in turn, was utilized to predict the local patterns of tissue differentiation within the fracture site and then how the equivalent mechanical properties of the macro-scale model change with time [10]. The spatial and temporal patterns of tissue differentiation predicted by this model (Fig. 9) were in general agreement with those observed experimentally. The model was further developed by including the effects of a minimally invasive percutaneous fixation device for the stabilization of the fracture [14].

3. 4. Design of new scaffolds for bone tissue engineering

Defects of critical dimensions in bone, whether induced by primary tumor resection, trauma, or selective surgery have in many cases presented insurmountable challenges to the current gold standard treatment for bone tissue repair. The primary purpose of bone tissue scaffolds is to offer a structural framework that favours and accelerate the bone defect healing process by allowing cell attachment, proliferation and differentiation into a controlled phenotype. The scaffold has to promote biological processes such as the production of extra-cellular matrix and vascularisation, furthermore must withstand the mechanical loads acting on it and transfer them to the biological tissues within which it is located. Bone tissue engineering is an emerging area in bioengineering at the frontiers between biomaterials, biology and biomechanics. Scaffold design for bone tissue engineering applications involves many parameters that directly influence the rate of tissue regeneration onto its microstructural surface. To improve scaffold functionality, increasing interest is being focused on in vitro and in vivo research in order to obtain the optimal scaffold design for a specific application.

Among the disciplines involved in the designing of new scaffolds for bone tissue engineering Mechanobiology certainly plays a role of crucial importance. Mechanobiology allows to determine the biophysical stimulus

acting on the mesenchymal tissue and how this stimulus will trigger or inhibit the formation of osteoblasts, chondrocytes and fibroblasts.

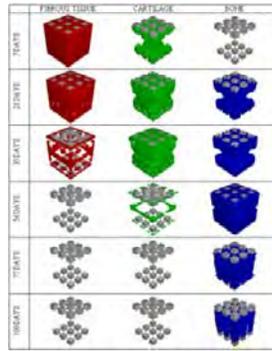


Fig. 9. Patterns of the issues differentiating during the fracture healing process.

It will be shown how in silico models, through a combined approach, i.e. finite element method/mechano-regulation algorithms/numerical optimization, can be used to determine the optimal parameters governing the scaffold performance [12].

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References

- Atzori B., Pappalettere C. in *Biomeccanica Ortopedica e Traumatologica*. Pipino and Quagliariella (Eds.), UTET, Torino, 1995.
- Boccaccio A., Lamberti L., Pappalettere C., Carano A., Cozzani M. *Journal of Biomechanics*, 39: 2907-2918, 2006.
- Boccaccio A., Pappalettere C., Kelly D.J. *Annals of Biomedical Engineering*, 35: 1940-1960, 2007.
- Boccaccio A., Lamberti L., Pappalettere C., Quagliariella L. *Journal of Mechanics in Medicine and Biology*, 8(2), 161-182, 2008.
- Boccaccio A., Lamberti L., Pappalettere C., Cozzani M., Siciliani G. *American Journal of Orthodontics and Dentofacial Orthopaedics*, 134(2), 260-269, 2008.
- Boccaccio A., Prendergast P.J., Pappalettere C., Kelly D.J. *Medical and Biological Engineering & Computing*, 46: 283-298, 2008.
- Boccaccio A., Lamberti L., Pappalettere C. *Journal of Mechanics in Medicine and Biology*, 8(2), 203-225, 2008.
- Boccaccio A., Vena P., Gastaldi D., Franzoso G., Pietrabissa R., Pappalettere C. *Proceedings of the Institution of the Mechanical Engineers, Journal of Engineering in Medicine, Part H*, 222(7), 1023-1036, 2008.
- Boccaccio A., Lamberti L., Pappalettere C., Quagliariella L. *Journal of Mechanics in Medicine and Biology*, 9, 301-327, 2009.
- Boccaccio A., Kelly D.J., Pappalettere C. *Journal of Orthopaedic Research*, 29, 433-443, 2011.
- Boccaccio A., Ballini A., Pappalettere C., Tullo D., Cantore S., Desiate A. *International Journal of Biological Sciences*, 7, 112-132, 2011.
- Boccaccio A., Pappalettere C. *Theoretical Biomechanics*, ISBN 978-953-307-851-9. Intech, 2011, Book edited by: Vaclav Klika. Invited Book Chapter, page 21-48, 2011.
- Boccaccio A., Cozzani M., Pappalettere C. *European Journal of Orthodontics*, 33, 113-120, 2011.
- Boccaccio A., D.J. Kelly, C. Pappalettere. Submitted.
- Breklemans W.A.M., et al. *Acta Orthopaedica Scandinavica*, 43, 301-317, 1972.
- Genovese K., Lamberti L., Pappalettere C. *Journal of Biomechanics*, 38, 2375-2389, 2005.
- Huiskes R., et al. *Mechanical properties of bone* (Edited by Cowin S.) AMD 45, 211-234, American Society of Mechanical Engineers, New York, 1981.
- Pascal L., Guillaume H. Eds. Springer. First Edition 468 pages, 2011.
- Persson A. *PhD Dissertation*, Chalmers Tekniska, Goteborg (Sweden), 1964.
- Prendergast P.J. *Clinical Biomechanics*, 12(6): 343-66, 1997.
- Spera D., et al. *Strain*, 47, e572-e587, 2011.
- Thelen A., Frey S., Hirsch S., Ladrrière N., Hering P. *Medical Laser Application*, 21, 9-14, 2006.
- Verhulp E., et al. *Journal of Biomechanics*, 37, 1313-1320, 2004.
- Wu X.F., et al. *Optics and Laser Technology*, 36, 387-391, 2004.
- Yuen K., et al. *Otolaryngology- head and neck surgery*, 117, 567-572, 1997.