Distinguishing Fact from Fiction in Finite Element Analysis: A guide for clinicians

Distinguishing Fact from Fiction in Finite Element Analysis – A Guide for Clinicians?

Chloe E.H. Scott[1], A. Hamish R.W. Simpson[1], Pankaj Pankaj[2]

[1] Department of Orthopaedics, The University of Edinburgh, UK

[2] School of Engineering, Institute for Bioengineering, The University of Edinburgh, UK

Finite element (FE) analysis is a well-established computational technique, which is routinely used by engineers to build complex structures such as bridges, skyscrapers and planes and to model the effects of earthquakes and avalanches. It has been successfully applied to examine the mechanical environment of the musculoskeletal system over the past four decades, and appears increasingly in orthopaedic journals but there are key questions orthopaedic surgeons should ask about FE modelling, namely:

1. Can the clinical question be answered by a FE model?
2. Is the particular FE model that has been used appropriate for the clinical question being addressed?
3. Do the results pass the basic sanity check: e.g. does bone deform on load application as expected?
4. Has the model been experimentally validated? And if so was the experiment a good representation of *in vivo* reality?
5. Are the output variables being examined appropriate to the research question: e.g. are principal strains a good indicator of bone or implants failure?
6. Are the findings clinically important?
7. Is the model description and reporting adequate in terms of input parameters employed, the mechanics/physics used and the output considered [1,2].

FE models simulate the mechanical response (stress, strain, deformation) of bone and bone-implant constructs. An FE analysis requires four input ingredients : (a) the geometry of each part of the structure being evaluated, (b) the material properties of each part, (c) the interactions between these parts and how they are restrained (boundary conditions) and (d) how they are loaded. Each of the four inputs (geometry, material properties, boundary conditions and loading) need to be specified. These are not always easy to define and therefore need to be represented by *appropriate* approximations.

FE models can be subject-specific (e.g. developed from cadaveric specimens or a single patient's scans) or generic (e.g. using average representative values). It is important for the

clinician to appreciate that for a model to be *truly* subject-specific, all four inputs must be subject-specific. As it is rarely possible to measure all parameters experimentally, assumptions pertaining to some aspect of the model are made, even in "subject-specific" models. It is important to note that although the results from a subject-specific model may represent that individual's situation more accurately, the results will be less generalizable to other patients.

Geometry: The geometry of the part to be examined in a FE simulation is converted into a set of elements with simple shapes (e.g. tetrahedra), much like pixelating the part in 3 dimensions. This generates a conceptual "mesh" of the part. The fineness of this mesh (i.e. the dimensions of each element) can have a major effect on the result the model predicts. The accuracy of the FE analysis increases with increasing number of elements used in the model which also increases the computational time/expense. The appropriate element size needs to be determined by conducting a mesh convergence study. This provides the element size beyond which further mesh refinement has little benefit in terms of accuracy. By importing data directly from CT and MRI scans, it is possible to create patient-specific geometries and meshes [2, 3].

Material properties: Bone is an inhomogeneous and anisotropic (properties vary with direction) composite material. Its response to load is both time dependent (it does not deform instantaneously on load application) and non-linear. However, in most orthopaedic FE models, the behaviour of bone is assumed to be isotropic, time-independent and linearly elastic. So, is this acceptable? Many clinical questions pertaining to bone implant constructs are adequately addressed by modelling bone in this simple way [4-7].

Inclusion of nonlinear geometry is warranted when large displacements are likely to arise [8] and nonlinear material properties if large strain are expected [8, 9]. Time-dependent properties should be incorporated if effect of cyclic loading [10, 11] or when sudden impacts are being considered. When considering implant or bone failure, wear, fatigue and fracture are all difficult outcomes to predict using the FE method due to these requirements for time dependent and non-linear material properties and nonlinear geometry. This is often not practical in terms of computational effort and thus proxies are often used. For bone and implant materials this has included regions where strain concentrates excessively [17, 18]; or the volume of elements within a defined region experiencing strain above the yield point for that material [4, 6].

In subject-specific models, inhomogeneity can be included by assigning elastic moduli on the basis of CT attenuations [2, 3] i.e. Young's modulus is estimated from Hounsfield units.

Generic models assign properties to defined regions (e.g. implant, trabecular bone, cortical bone) but do not include point-to-point variations. Thus, the results should be interpreted with

caution if there is likely to be variability within each of these materials e.g. subchondral trabecular bone versus mid metaphyseal trabecular bone. Results of mechanical tests conducted on cadaveric long bones compared with the results of corresponding FE models have shown that while subject-specific property assignment provides a closer match generic property assignment also gives fairly good results [3]. While inhomogeneity can be included in subject-specific models [3], anisotropy is not readily incorporated [8, 9] and studies that examine its influence are limited [12].

Boundary conditions and loading: In vivo bones are restrained by their adjacent joints, muscles and ligaments. In a laboratory set-up or in a numerical model the bone-implant system must be restrained before mechanical forces can be applied to it. Such boundary conditions rarely reflect *in vivo* reality, but approximate it to different degrees. The importance of boundary conditions depends on the problem and the outcome variable being investigated [13]. In most experimental mechanical testing studies and FE models the contribution of muscles and ligaments is ignored [4, 6]. Where soft tissues are incorporated there are two possible approaches: 1) force vectors are applied to the bone at areas of muscle attachments; 2) muscles/ligaments are included as part of the system as simple springs running from origin to insertion [14-16]. Loading protocols similarly vary in complexity from simple distributed loads applied to a given surface in a monotonically increasing manner to complex patterns of loading reflecting physiological activities as obtained from instrumented prosthesis data (www.orthoload.com, [5, 7,16]).

Outputs and their validation: FE simulations provide a multitude of output parameters including displacements, stresses and strains from any location within the model. Normal (or principal), stress/strain refers to tension (positive values) or compression (negative values)). Simulations can show how the bone or bone-implant system is likely to become unstable or fail, though, as already discussed, predictingfracture propagation, fatigue or wear is complex. Consequently, they can be used to design new implants or optimise treatments. In musculoskeletal science, it is expected that FE models be validated, usually against *in vitro* biomechanical testing. However, it is important to recognise that such experimental set-ups are themselves an approximation of reality. Often a properly developed computer model can simulate the *in vivo* scenario much better than a lab set-up. Hence, sanity checks can be a more important and reassuring form of validation than laboratory experimental validation.

What are FE models good at? The FE method is well established for investigating the mechanical behaviour of implants in single bones (generic or subject-specific) [4-7] and of bone itself. Patient specific modelling undoubtedly has a role in the management of complex

patients including modelling bone metastases [17, 18] and complex revision scenarios as part of custom implant design. Modelling whole joints needs to include associated soft tissue structures and constraints in addition to implant and bone [14]. This can be important when addressing certain clinical questions. While commonly used elastic models can provide the location at which yielding/fracture would be expected to initiate [11], more complex nonlinear models are required to predict implant loosening, fracture propagation or wear [8-11, 17-20] .

In conclusion, FE can readily incorporate complex anatomical shapes, loading scenarios, boundary conditions and bone properties; these parameters can be readily varied for examining their effect on the predicted outcome. Such variations are very difficult to achieve in *in vitro* experiments. FE simulation can be undertaken at a fraction of the cost of a lab set-up, and provides a wealth of output data (e.g. stresses and strains at any location) impossible to obtain in the lab. The key is the interpretation of this data in the context of the research question and ensuring that the assumptions that have been made fit the clinical condition.

References

1. Erdemir A, Guess TM, Halloran J, Tadepalli SC, Morrison TM. Considerations for reporting finite element analysis studies in biomechanics. J Biomech 2012. 45:625-633
2. Pankaj, P., *Patient-specific modelling of bone and bone-implant systems: the challenges.* Int J Numer Method Biomed Eng, 2013. **29**(2): p. 233-49.
3. Taddei F, Cristofolini L, Martelli S, Gill HS, Viceconti M (2006) *Subject-specific finite element models of long bones: An in vitro evaluation of the overall accuracy.* J Biomech 2006. 39:2457-2467
4. Scott, C.E., et al., *Metal-backed versus all-polyethylene unicompartmental knee arthroplasty: Proximal tibial strain in an experimentally validated finite element model.* Bone Joint Res, 2017. **6**(1): p. 22-30.
5. Conlisk, N., C.R. Howie, and P. Pankaj, *The role of complex clinical scenarios in the failure of modular components following revision total knee arthroplasty: A finite element study.* J Orthop Res, 2015.
6. Danese, I., P. Pankaj, and C.E.H. Scott, *The effect of malalignment on proximal tibial strain in fixed-bearing unicompartmental knee arthroplasty: A comparison between metal-backed and all-polyethylene components using a validated finite element model.* Bone Joint Res, 2019. **8**(2): p. 55-64.
7. Xie, S., et al., *Metaphyseal cones in revision total knee arthroplasty.* Bone Joint Res, 2020. **9**(4): p. 162-172.
8. Donaldson, F.E., P. Pankaj, and A.H.R.W. Simpson, Investigation of factors affecting loosening of ilizarov ring-wire external fixator systems at the bone-wire interface. J. Orthopaedic Research, 2012. 30(5): p. 726-732.

9. Donaldson, F.E., P. Pankaj, and A.H.R.W. Simpson, Bone properties affect loosening of half-pin external fixators at the pin–bone interface. Injury, 2012. 43(10): p. 1764-1770
10. Xie, S., K. Manda, and P. Pankaj, *Effect of loading frequency on deformations at the bone-implant interface.* Proc Inst Mech Eng H, 2019. **233**(12): p. 1219-1225.
11. Pankaj, P. and S. Xie, The risk of loosening of extramedullary fracture fixation devices. Injury, 2019. 50 Suppl 1: p. S66-72
12. Hellmich C, Kober C, Erdmann B. Micromechanics-based conversion of CT data into anisotropic elasticity tensors, applied to fe simulations of a mandible. Annals of Biomedical Engineering 2008; 36(1):108–122
13. MacLeod, A.R., A.H.R.W. Simpson, and P. Pankaj, Experimental and numerical investigation into the influence of loading conditions in biomechanical testing of locking plate fracture fixation devices. Bone Joint Res, 2018. 7(1): p. 111-120.
14. Kang, K.T., et al., *A computational simulation study to determine the biomechanical influence of posterior condylar offset and tibial slope in cruciate retaining total knee arthroplasty.* Bone Joint Res, 2018. **7**(1): p. 69-78.
15. Phillips, A.T., et al., *Finite element modelling of the pelvis: inclusion of muscular and ligamentous boundary conditions.* Med Eng Phys, 2007. **29**(7): p. 739-48.
16. MacLeod, A.R., et al., *The effect of plate design, bridging span, and fracture healing on the performance of high tibial osteotomy plates: An experimental and finite element study.* Bone Joint Res, 2018. **7**(12): p. 639-649.
17. Eggermont, F., et al., *Can patient-specific finite element models better predict fractures in metastatic bone disease than experienced clinicians?: Towards computational modelling in daily clinical practice.* Bone Joint Res, 2018. **7**(6): p. 430-439.
18. Sternheim, A., et al., *When and where do patients with bone metastases actually break their femurs?* Bone Joint J, 2020. **102-B**(5): p. 638-645.
19. Goffin, J.M., P. Pankaj, and A.H. Simpson, *Are plasticity models required to predict relative risk of lag screw cut-out in finite element models of trochanteric fracture fixation?* J Biomech, 2014. **47**(1): p. 323-8.
20. Pankaj, P. and F.E. Donaldson, *Algorithms for a strain-based plasticity criterion for bone.* Int J Numer Method Biomed Eng, 2013. **29**(1): p. 40-61.