Finite Element Modelling of Tibial Defects Managed with a Fine Wire Fixator

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Introduction

Clinical Challenge

- Tibial defects as consequence of trauma are commonplace in limb reconstruction
- Circular frames utilised frequently for primary trauma and salvage
- Variations in frame constructs for the same indications exist due to
    - Modularity of constructs
    - Surgeon experience/preference
    - Interpretations/assumptions of frame construct
    - Multiple manufacturers and consignments
    - Patient physique and functional demands
- As a result, there of these limited objective information on the mechanobiological environment within a defect as a result of variations in frame construction

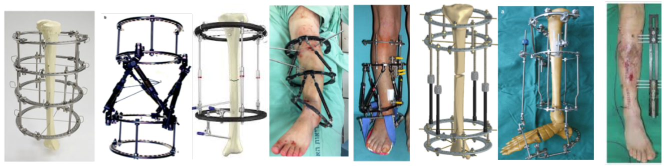


Figure 1 – Example range of fixators employed to address tibial defects

Finite Element Analysis (FEA)

- Generates numerical representation of each element that matches mechanical testing
- Computationally simulates behaviour of the structure under various configurations
- Identify the behaviour of each component e.g. of the Ilizarov system

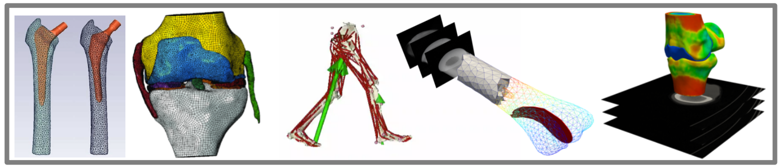


Figure 2 – FEA applications in orthopaedics [1,2]

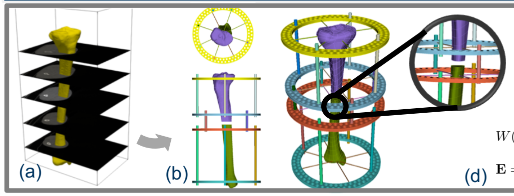
Aims

Develop an *in silico* framework to model circular frame fixators

Objectives

1. Parameterise an experimental model of tibia & Ilizarov frame
2. Construct an *in silico* FEA model to simulate physiological loading
3. Collect predictive (stochastic) data based on archetypal clinical models of a 4cm tibial defect.

Methods



(c)

$$\nabla_{\mathbf{X}} \cdot \mathbf{P} + \mathbf{b} = 0$$

$$\mathbf{u} = \bar{\mathbf{u}} \quad \text{at } \partial_{\mathbf{u}}\mathscr{B}$$

$$\mathbf{P} \cdot \mathbf{N} = \bar{\mathbf{t}} \quad \text{at } \partial_{\mathbf{t}}\mathscr{B}$$

$$W(\boldsymbol{E}) = \frac{\lambda}{2}[\mathrm{tr}(\boldsymbol{E})]^2 + \mu\,\mathrm{tr}\left(\boldsymbol{E}^2\right)$$

(d)

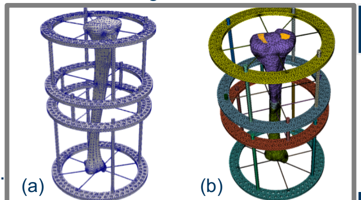
$$\mathbf{E} = \frac{1}{2}\left[(\nabla_{\mathbf{X}}\mathbf{u})^T + \nabla_{\mathbf{X}}\mathbf{u} + (\nabla_{\mathbf{X}}\mathbf{u})^T \cdot \nabla_{\mathbf{X}}\mathbf{u}\right]$$

Figure 3 – (a) tibial segmentation from CT (b) parameterised surface model of frame with tibial defect. Key equations for (c) nonliner elasticity model and (d) Saint Venant-Kirchoff material.

1. CT segmented and surface model created. 4cm tibial defect created *in silico*. Fine wire fixator generated using reference material parameters (e.g. 316l stainless steel), with 1.8mm wires introduced in recognized corridors[3] and digitally tensioned to 1200N (Figure 3).

2. Finite element mesh constructed and axially loaded to 800N using a nonlinear elasticity model of a Saint Venant-Kirchhoff material[4] (Figure 3). This was solved using an open source parallel finite element library (MOFEM) (Figure 4).

3. Data for deformation under load was normalised to a standard bone density. Longitudinal deformation at various defect sizes and bone densities.



$$\rho_{\mathrm{QCT}}\left[\mathrm{gK_2HPO_4/cm^3}\right] = 0.000705 \cdot \mathrm{HU} - 0.000581$$

$$E[\mathrm{MPa}] = 10{,}200 \cdot \rho_{\mathrm{qCT}}^{2.01}\left[\ \mathrm{g/cm^3}\right] \quad \text{(c)}$$

Figure 4– (a) FEA mesh (b) loading set up. (c) power-law for density to modulus

References

1. MA Kadir, N Kamsah. Interface Micromotion of Cementless Hip Stems in Simulated Hip Arthroplasty VL - 6 DO - 10.3844/ajassp.2009.1682.1689 JO - American Journal of Applied Sciences ER
2. K Lewandowski, L Kaczmarczyk, I Athanasiadis, C Pearce. Numerical investigation into fracture resistance of bone following adaptation (unpublished results)
3. M. A. Catagni, A. Bianchi Maiocchi Atlas For The Insertion Of Trans-osseous Wires And Half-Pins December 2001 In book: Atlas For The Insertion Of Trans-osseous Wires And Half-Pins.
4. B. Helgason (2006) Mathematical relationships between bone density and mechanical properties: A literature review.

Results

- Following generation of 100,208 tetrahedral elements, approximately 3000 lines of code were required to generate the model (Figure 5)
- Loading across various defected demonstrated similar deformation and therefore varying strains (Table 1).

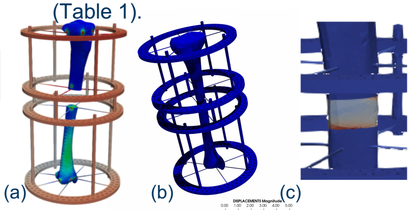


Figure 5 – (a) stress (b) displacement (c) strain field

Size of defect	Max displacement:
5 mm	5.00 mm
10 mm	5.75 mm
20 mm	5.76 mm
40 mm	5.78 mm

Table 1 – *Displacement & defect size*

- The magnitude of this deformation was such that bone contact was observed with defects less than 5.75mm.
- When adjusting for bone density, displacement remained statistically similar (Table 2).

Youngs modulus:	Max displacement:
10 GPa	5.76 mm
5 GPa	5.81 mm
Image-Based	5.78 mm

Table 2 – *Displacement & bone Modulus*

Limitations

- Due to current restrictions, this model has only yet been validated on historic parameters
- As with any FEA model, there are a range of assumptions including modular interface, bone interface, repetitive loading and fluid dynamics within the defect

Conclusions

1. Accounting for standard assumptions, a mathematically rigorous model has been parameterised to represent a clinical challenge
2. Outputs are compatible with in physical observations
3. Using a standard frame set up, a 5mm defet will 'bottom out'.
4. Further work to validate and refine this model using mechanical testing is warranted