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The risk of loosening of extramedullary fracture fixation devices

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¹ The risk of loosening of extramedullary fracture fixation

² devices

3

4 Abstract

5

6	Extramedullary devices that use screws, pins or wires are used extensively to treat
7	fractures in normal and diseased bone. A common failure mode is implant loosening
8	at the bone-screw/pin/wire interface before fracture healing occurs. This review first
9	considers the fundamental mechanics of the bone-fixator construct with focus on
10	interfacial strains that result in loosening. It then evaluates the time-independent and
11	time-dependent material models of bone that have been used to simulate and predict
12	loosening. It is shown that the recently developed time-dependent models are
13	capable of predicting loosening due to cyclic loads in bone of varying quality.
14 15	Key words: locking plates; unilateral fixators; ring fixators; time-dependent
16	behaviour; cyclic loading; inter-fragmentary motion; plasticity and viscoplasticity
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18 **Conflict of interest:** The authors declare that there is no conflict of interest.

19 **1** Introduction

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Extramedullary devices that use screws, pins or wires are used extensively to treat 21 fractures in normal and diseased bone. These devices carry most of the load, 22 particularly in cases where there is a fracture gap, before callus formation occurs. 23 The load is transmitted from the bone-screw/pin/wire interface to the plate or an 24 external frame. It has been well documented that these devices need to fulfil three 25 clinical requirements [1,2]: (a) they must support fracture healing; (b) they must not 26 fail during the healing period; and (c) they should not loosen or cause patient 27 discomfort. Requirement (a) depends on the stiffness of the bone fixator construct 28 and the load applied by the patient, which determine the relative movement between 29 fractured fragments or interfragmentary motion (IFM). Requirement (b) relates to 30 stresses within the implant and potential failure before healing occurs. Strains at the 31 bone-screw/pin/wire interface should not be too high to ensure that requirement (c) is 32 met. 33

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There have been a number of studies that have considered requirements (a) and (b) 35 [3–10] and shown that fulfilment of these depends on factors such as fracture 36 location, device used and its configuration (e.g. where the screws are placed in a 37 38 locking plate or how much tension is applied to the wires in ring fixators). Interestingly it has been found that device stiffness (or resulting IFM) and stresses 39 within the device are not strongly effected by bone quality [3–5,11]. In other words, if 40 the aim of a biomechanical study is to determine IFM alone then bone quality does 41 not have a significant role to play. Whereas, loosening at the bone-implant interface 42 strongly depends on bone quality in addition to the factors that influence IFM [3,4]. 43

Loosening is reported frequently as a complication in implant usage and some previous studies have noted that mechanical forces initiate it before any contribution from biological processes [12]. Since biomechanical prediction of loosening requires modelling the complex bone material, it is much more complicated; consequently, influence of bone properties to examine mechanical environment at interface has received relatively little attention [3,4,11].

50

The first aim of this review is to present the fundamental mechanics of the bone-51 fixator construct with focus on interfacial strains that result in loosening. The second 52 aim is to consider the constitutive material models of bone used to predict loosening, 53 in particular recently developed novel time-dependent models that are capable of 54 predicting loosening due to cyclic loads [13–15]. While most discussion presented is 55 in the context of extramedullary devices such as locking plates, unilateral fixators 56 and Ilizarov rings, many of the concepts presented are equally applicable to other 57 fixation devices. 58

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60 2 The mechanics of extramedullary devices

61 **2**

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2.1 Interfragmentary motion and stresses in the implant

We first consider the mechanics of extramedullary devices. Figure 1a shows a bonelocking plate construct, the mechanics of which is not too dissimilar to unilateral fixators. A number of biomechanical responses arise due to the application of load P (due to partial or full load bearing by the patient). Firstly load bearing causes interfragmentary motion (IFM) between the fractured fragments (Figure 1b) which is known to aid callus formation [16,17] – too much or too little inhibits fracture healing [2]. IFM can vary across the thickness of the bone; for example from Figure 1a and

1b it can be seen that the largest IFM is at the far cortex and given by x-x'. Secondly 70 the plate and screws experience bending causing stresses within the implant. The 71 amount of bending and IFM depend on factors such as dimensions and materials of 72 the locking plate, bone-plate offset, load applied and the manner in which bone 73 experiences load and screw configuration particularly the working length (also known 74 as the bridging span and defined as the distance between the two innermost screws 75 76 on either side of the fracture). In cases with a fracture gap, higher working length results in larger plate stresses (primarily in the plate portion bridging the fracture) 77 78 and larger IFM [2,4]. Some studies have incorrectly reported larger stresses with shorter working lengths [18], but the reasons for this erroneous interpretation have 79 been discussed in Macleod and Pankaj [2]. Plate bending also results in pull-out and 80 81 push-in forces as shown in Figure 1b; these have been previously discussed in the 82 context of unilateral fixators [11]. As the applied load increases the lever arm d (Figure 1a) increases to $\Delta > d$ (Figure 1b) which increases the bending forces even 83 further. In engineering mechanics this is often referred to as P - Δ effect and causes 84 the relationship between load and IFM to become nonlinear [19]. Nonlinear load-85 displacement behaviour also arises in Ilizarov fixators (Figure 2a) due to sagging 86 wires [3]. Studies on locking plates [4], unilateral fixators [5,11] and Ilizarov fixators 87 [3] have shown that bone quality has a relatively small influence on IFM and implant 88 89 stresses.

90

91 **2.2 The mechanics of loosening**

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Let us now consider strains at the bone-screw interface due to forces along the axis
of the bone (as shown in Figures 1a and 2a); these strains are responsible for
loosening, which is the primary focus of this review. It is important to note that we

deliberately employ the response parameter strain rather than stress for three 96 reasons. Firstly it is now well recognised that bone fails due to strain rather than 97 stress [20]. Secondly, failure strain does not vary significantly with bone quality or its 98 anisotropy (this is further discussed later in this review). Lastly, while stresses have 99 peak values beyond which they cannot rise due to yielding/failure, strains can 100 continue to increase. Typical large strain regions for locking plates are shown in 101 102 Figure 1c and for Ilizarov fixators in Figure 2b. It has been shown that the maximum bone strains at the interface of the screw/pin/wire closest to the fracture (e.g. screws 103 104 2 and 3 rather than screws 1 and 4 in Figure 1a) [3–5,11,21]. For locking plates and unilateral fixators the strains are the largest at the periosteum of the near cortex and 105 progress towards the endosteum with increasing load [11]. The volume of bone that 106 goes beyond the yield level increases considerably with poor bone quality [3,11]. The 107 pattern of bone yielding is different between unilateral and Ilizarov fixators. For 108 unilateral fixators and locking plates bone yielding can progress through the full 109 cortex as shown in Figure 1c for screw 2, where bone superior to the screw 110 experiences large strains. If the depth of yielded bone is greater than thread height, 111 then loosening can be initiated due to loss of screw thread purchase. For Ilizarov 112 fixators, on the other hand, bone yield remains concentrated separately at the 113 periosteum and endosteum, superior and inferior to the wire, respectively [3] as 114 shown in Figure 2b. This is a possible reason for Ilizarov wires being associated with 115 lower rates of loosening than half pins [22,23]. 116

117

It has also been shown that reduced stiffness (or increased flexibility) of the bone
fixator construct, which increases IFM, also results in larger interfacial strains
[3,4,11]. Flexibility can be increased by using materials with lower elastic modulus

(e.g. titanium rather than steel), smaller plate or screw dimensions, larger working 121 length or in case of Ilizarov fixators smaller wire tensions. So flexibility is detrimental 122 from the point of view of large strains at the interface but it may result in an IFM that 123 causes faster healing before any ill effects of high interfacial strains come to the fore. 124 Thus need for maintaining adequate IFM needs to be balanced with the risk of 125 loosening. It is also important to note that compressive and tensile strains often 126 occur simultaneously as shown in Figure 1d for the near cortex of screw 2. In this 127 case compressive strains due to screw pushing up in the radial direction are 128 accompanied by tensile strains in the circumferential direction due to screw hole 129 being enlarged. 130 Figure 1 131 It is also important to note that drilling (prior to screw insertion) causes interfacial 132 damage which has been estimated to extend up to 300 µm around the circumference 133 [24]. Moreover, large interfacial strains also result from an interference fit when the 134 drilled pilot hole has a smaller diameter than the screw being inserted [19]. 135 Figure 2 136 Push-in and pull-out forces discussed in the context of unilateral fixators and locking 137 plates can cause loosening which is resisted by screw threads. It has been shown 138 that the bone at the interface of the first thread from the screw entrance carries the 139 largest load [6] and this load carrying demand decreases for screws deeper inside 140 the bone. As bone is not homogeneous, local microarchitecture can play an 141 important role in determining whether the device may become loose [25]. 142 143 144 145

147 **3** Material models of bone to predict loosening

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As discussed above bone quality (varying from healthy to osteoporotic) plays a major 149 role in the distribution of strains at the bone-screw/pin/wire interface. In order to 150 predict loosening using principles of biomechanics it is important to use appropriate 151 material models of bone. The most commonly used mechanical models of bone are 152 time-independent i.e. they assume that any deformation due to loading occurs 153 instantaneously. Almost all research on bone-implant systems assumes bone 154 behaviour to be time-independent [26] though it is well recognised that bone 155 deformation on load application increases with time or is time-dependent [13-156 15,27,28] In the following sections we first discuss time-independent models that 157 have been employed to examine loosening; these include use of elasticity and 158 elastoplasticity. We then go on to consider time-dependent models that have been 159 recently developed by the authors and employed to evaluate fixator loosening. 160

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162 **3.1 Modelling bone as an elastic material**

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In computational biomechanics the most common assumption for modelling bone is 164 that it is linear, isotropic and elastic. The term elastic implies that any deformation 165 experienced by the material on application of forces is fully recovered when the 166 forces are removed. Addition of the term linear means that the mechanical response 167 (e.g. deformation) is proportional to the load applied and isotropic material is one 168 which has the same mechanical properties in all directions and requires two elastic 169 constants to relate stresses to strains (e.g. Young's modulus and Poisson's ratio). In 170 most computational studies with generic bone geometries it is a common practice to 171

further assume that the material is homogeneous (i.e. properties do not vary from 172 point to point), though distinctly different regions (e.g. cortical and trabecular) may be 173 assigned different properties [29]. In subject-specific studies for which CT data is 174 available inhomogeneous material properties are often assigned [30,31] by 175 empirically converting CT attenuations to Young's modulus. It is arguable as to 176 whether answers obtained from subject- or patient-specific models have a limited 177 applicability and whether generic or "average" models are more suitable for 178 answering general questions. 179

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While the assumption of isotropy serves well for many biomechanical studies, it is 181 well recognised that both cortical and cancellous bone are better represented by 182 183 orthotropic or transtropic elasticity [32] requiring many more properties for relating stresses to strains. Materials that are not isotropic do not have the same properties 184 in all directions. For example, orthotropic materials have three orthogonal planes of 185 186 elastic symmetry and stress-strain relations are defined by using 9 elastic constants. Orthotropic properties of bone have been determined using experimental [33] and 187 numerical approaches [34,35]. 188

189

In computational modelling to evaluate loosening of fracture fixation systems two
questions arise. The first is whether an isotropic bone model is adequate for
obtaining reasonable answers and the second is whether elasticity can be used to
predict loosening. Let us consider each of these questions in turn.

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To our knowledge there have been no studies that have compared isotropic and anisotropic models in fracture fixation studies. It can be argued that the use of

orthotropic material properties increases the complexity of the model, and if these 197 are not accurately assigned, they may introduce more prediction errors than a simple 198 assumption of isotropy. However, Young's moduli for both cortical and cancellous 199 bones in one principal orthotropic direction can be around three times the other 200 direction [35]. Therefore, same force acting in one direction will cause much larger 201 strains than in the other. Donaldson et al. [35] showed that in the femoral mid-shaft 202 203 the elastic modulus of cortical bone in the proximal-distal direction was not only higher than that for endosteum-periosteum direction but also decreased less rapidly 204 205 with age i.e. bone became more anisotropic with age. Considering this finding in conjunction with the mechanics of unilateral and locking plate fixation in which axial 206 loading of bone is accompanied by pull-out and push-in forces it can be concluded 207 208 that half-pin or screws apply forces in the direction least adapted to loading, and therefore most at risk of failure in patients with osteoporosis [11]. 209

210

Let us now consider use of elasticity in the estimation of loosening. It has been 211 suggested that loosening is caused by large irreversible strains at the bone implant 212 interface that enlarge the screw/pin/wire hole [3,11]. Since elasticity implies that 213 deformations are recovered on load removal it is argued that it cannot be used to 214 model loosening. However, researchers often use elasticity wherein they assume a 215 216 threshold output variable (e.g. yield strain in compression) and evaluate the volume of material that exceeds this threshold value, which is then taken as an estimate of 217 the volume susceptible to yielding [4,36,37]. In reality, when a small region bone 218 goes beyond its yield limit and cannot carry additional loads, considerable 219 redistribution of stresses occurs resulting in the yield region becoming localised; 220 these phenomena cannot be captured by elasticity. In spite of this shortcoming, it 221

has been shown that in the case of hip screws prediction of regions likely to yield 222 using elasticity are similar to those obtained from more complex models [38]. 223 MacLeod et al. [4] used orthotropic elasticity with equivalent strain threshold to 224 examine screw placement to reduce loosening risk in locked plating. They found that 225 the use of titanium in comparison to steel increased the volume of bone exceeding 226 the threshold; results similar to those obtained with plasticity models [11]. MacLeod 227 228 et al. [4] also showed that larger working lengths increase the predicted volumes of bone above the threshold (Figure 3). Therefore, simple elastic models can be 229 successfully used to, at least, ascertain trends, though they are unable to predict 230 propagation of yielding or damage. 231

232

Figure 3

- **3.2 Modelling bone as an elastoplastic material**
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It has been shown that load bearing causes strains at the bone-screw/pin/wire 235 interface that are larger than the elastic limit for bone [3,11] resulting in irreversible 236 deformations and these are responsible for loosening. Simulation of this irreversible 237 deformation response requires inclusion of post-elastic material behaviour for bone 238 which has been most commonly modelled using elastoplasticity. Elastoplasticity 239 implies that the material remains elastic when loaded up to a certain limit (yield value 240 defined in terms of stresses or strains) and has irreversible deformations when 241 loaded beyond this limit. A wide range of yield criterion are available in commercial 242 finite element codes and several of these have been used for bone [26], often with 243 little thought to their suitability. Most models available in commercial codes are 244 based on stress i.e. a material is considered to have yielded when a combination of 245 stress components reaches a yield value (i.e. elastic limit). In addition to anisotropic 246

elasticity, bone is also anisotropic in terms of yield strength, which varies with bone 247 quality. So, specifying yield parameters for stress-based criteria cannot be readily 248 achieved. Interestingly relatively recent experimental [39] and computational [40] 249 research has shown that bone yields at relatively isotropic strains and yield strain is 250 not dependent on apparent elastic stiffness or density. In other words, it is much 251 simpler to model bone of varying quality and microstructure using strain-based 252 criteria in comparison to stress-based approaches. Strain-based plasticity was first 253 discussed about four decades ago by Naghdi and Trapp [41] but has received little 254 attention in comparison with stress-based theories. Algorithms to achieve these are 255 now available [42]. 256

257

Donaldson et al. [3] used orthotropic elasticity in conjunction with strain-based plasticity to determine loosening in Ilizarov fixators. They used asymmetric yield strain limits, 0.5% in tension and 0.7% in compression, and showed that the pattern of yielding in ring fixators was as shown in Figure 2. They found that: increasing wire tension reduces volume of yielded bone and the volume increases as the bone quality decreases; and that there is significant reduction bone yield volume when the number of wires on either side of the fractures are increased.

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3.3 Bone modelled as a time-dependent material

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As discussed loosening at the bone-screw/pin/wire interface has been considered by examining strains on load application using time-independent elastic or elastoplastic constitutive models for bone. A number of studies [43,44,45] have shown that loosening of connecting screw/pin is a function of loading cycles. Time-independent models are unable to capture this phenomenon as cyclic loading (with the same

magnitude and direction) merely reproduces the mechanical response from the first
cycle. Here we consider a recently promulgated theory which explains loosening due
to cyclic loading via time-dependent behaviour of bone [46].

276

Bone is recognised as time-dependent material and its time-dependent properties 277have been measured experimentally using: creep tests [13–15] in which time-varying 278 strain due to applied constant load is measured over time; relaxation tests [47,48] in 279 which time-varying force due to applied constant deformation is measured over time; 280 281 and dynamic tests [49,50] in which the lag between sinusoidal stress and strain is measured over a frequency range. Although time-dependent behaviour of bone has 282 been studied extensively, most experimental studies were not developed into 283 284 computational models or employed in modelling of bone-implant systems. Recently studies employed multiple-load-creep-unload-recovery experiments [13] to 285 characterise time-dependent behaviour of trabecular bone, and developed BV/TV-286 287 based linear viscoelastic [14], nonlinear viscoelastic [15] and nonlinear viscoelasticviscoplastic [51] constitutive models – models with increasing complexity and 288 289 consequent accuracy.

290

Xie et al. [46] considered the influence of cyclic loading in an idealised unicortical
bone-screw system (Figure 4a and 4b). In this the screw was subjected to 500
cycles of lateral loads (Figure 4c) with loading frequency f = 1 Hz followed by 1000
sec recovery. The trabecular bone modelled as time-dependent material. The study
examined the accumulation of strain at the bone-screw interface with increasing
number of cycles and after recovery.

Figure 4

Figure 5 shows the minimum (compressive denoted negative) and maximum (tensile 298 denoted positive) principal strain contours from the symmetry surface (Figure 4a) 299 and Section A-A (Figure 4b). Figures 5a and 5b show the compressive strain 300 contours at time points when the load is at its peak and when it has been reduced to 301 zero respectively at different loading cycles. Similarly, Figures 5c and 5d show the 302 303 tensile strain contours at time points when the load is at its peak and when it has been reduced to zero respectively at different loading cycles. Figures 5e and 5f show 304 305 the compressive and tensile strain contours respectively after 1000 sec of recovery following 500 cycles of loading. It is clear that the strain experienced by bone 306 increases with increasing number of cycles, similar to that reported in previous 307 studies [43,44,45]. It is important to note that with time-independent models the 308 variation with number of cycles cannot be captured. Moreover, time-independent 309 elastic models will show zero strains upon unloading. For the nonlinear viscoelastic-310 viscoplastic simulation [46], not all of the strain is recovered upon unloading and the 311 strain experienced by bone increases with applied loading cycles. A residual strain 312 exists even after 1000s of recovery. This increase in strain with increasing number of 313 loading cycles and residual strain indicates that the mechanical environment at the 314 bone-screw interface will change as physiological activities are undertaken by the 315 patient and will accentuate screw loosening. 316

317

Figure 5

By assigning time-dependent material properties for different bone densities based on recent experiential studies [14], permits simulation of bone-screw interface strain/micromotion similar to that reported experimentally [43]. This has only become possible recently.

A recent study has also shown that the strain/displacement experienced at the 323 interface is also loading frequency dependent [51]. In the first few cycles the larger 324 strain is observed if bone-screw system is loaded at a lower frequency; while the 325 interface experiences larger strain at higher loading frequencies after a large number 326 of loading cycles have been applied. In the first few cycles, a lower loading 327 328 frequency has a relatively longer loading time and relatively smaller loading rate. Therefore, larger displacement occurs when bone-screw system is loaded at a lower 329 330 frequency during the loading and unloading phases as the bone is provided more time to deform or recover. When the bone-screw system is loaded at higher 331 frequencies, the loading/unloading time is shorter (in comparison to lower frequency 332 loading) and the bone is loaded again by the next cycle before it can recover from its 333 last loading cycle. 334

335

336 4 Conclusions

337

Implant loosening is initiated by strains at the bone-screw/pin/wire interface. These 338 strains are generally larger in low density bone. The interfacial strains increase with 339 decrease in the stiffness of the bone fixator construct which can be caused by 340 features such as increased working length, use of implant materials with lower 341 stiffness (e.g. titanium rather than steel) or reduced wire tension in ring fixators. The 342 reduction of the construct stiffness also causes increased interfragmentary motions 343 between fractured segments which may be beneficial for healing. Therefore, risk of 344 loosening needs to be balanced by the need of maintaining adequate 345 interfragmentary motion. Computational simulation/prediction of loosening requires 346

347	appropriate models of bone behaviour. For this most previous studies have
348	employed time-independent models. These are unable to capture loosening that is
349	accentuated due to cyclic loading. Recently developed time-dependent models are
350	extremely promising in this respect.
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354 **References**

- 355 [1] MacLeod A, Pankaj P. Computer simulation of fracture fixation using
- 356 extramedullary devices: an appraisal. In: Doyle B, Miller K, Wittek A, Nielsen
- 357 PMF, editors. Comput. Biomech. Med., Springer, New York, NY; 2014, p. 87–
- 358 99.
- Macleod AR, Pankaj P. Pre-operative planning for fracture fixation using
 locking plates: device configuration and other considerations. Injury
 2018;49:S12–S18.
- 362 [3] Donaldson FE, Pankaj P, Simpson AHRW. Investigation of factors affecting
- 363Ioosening of ilizarov ring-wire external fixator systems at the bone-wire
- 364 interface. J Orthop Res 2012;30:726–32.
- MacLeod AR, Simpson AHRW, Pankaj P. Age-related optimization of screw
 placement for reduced loosening risk in locked plating. J Orthop Res
 2016;34:1856–64.
- 368 [5] Donaldson FE. On incorporating bone microstructure in macro-finite-element
 369 models. PhD Thesis, The University of Edinburgh, 2011.
- 370 [6] MacLeod AR, Simpson AHRW, Pankaj P. Reasons why dynamic compression

371 plates are inferior to locking plates in osteoporotic bone: a finite element

- explanation. Comput Methods Biomech Biomed Engin 2015;18:1818–25.
- 373 [7] MacLeod AR, Simpson AHRW, Pankaj P. Experimental and Numerical
- 374 Investigation into the Influence of Loading Conditions in Biomechanical Testing
- of Locking Plate Fracture Fixation Devices. Bone Jt Res 2018;7:111–120.
- [8] Lenz M, Windolf M, Mückley T, Hofmann GO, Wagner M, Richards RG, et al.
- 377 The locking attachment plate for proximal fixation of periprosthetic femur
- 378 fractures A biomechanical comparison of two techniques. Int Orthop

2012;36:1915–21.

380	[9]	Matres-Lorenzo L, Diop A, Maurel N, Boucton MC, Bernard F, Bernardé A.
381		Biomechanical Comparison of Locking Compression Plate and Limited Contact
382		Dynamic Compression Plate Combined with an Intramedullary Rod in a
383		Canine Femoral Fracture-Gap Model. Vet Surg 2016;45:319–26.
384	[10]	Rowe-Guthrie KM, Markel MD, Bleedorn JA. Mechanical Evaluation of
385		Locking, Nonlocking, and Hybrid Plating Constructs Using a Locking
386		Compression Plate in a Canine Synthetic Bone Model. Vet Surg 2015;44:838-
387		42.
388	[11]	Donaldson FE, Pankaj P, Simpson AHRW. Bone properties affect loosening of
389		half-pin external fixators at the pin-bone interface. Injury 2012;43:1764–70.
390	[12]	Taylor M, Tanner KE. Fatigue failure of cancellous bone: a possible cause of
391		implant migration and loosening. J Bone Joint Surg Br 1997;79–B:181–2.
392	[13]	Xie S, Manda K, Wallace RJ, Levrero-Florencio F, Simpson AHRW, Pankaj P.
393		Time Dependent Behaviour of Trabecular Bone at Multiple Load Levels. Ann
394		Biomed Eng 2017;45:1219–26.
395	[14]	Manda K, Xie S, Wallace RJ, Levrero-Florencio F, Pankaj P. Linear
396		viscoelasticity - bone volume fraction relationships of bovine trabecular bone.
397		Biomech Model Mechanobiol 2016;15:1631–40.
398	[15]	Manda K, Wallace RJ, Xie S, Levrero-Florencio F, Pankaj P. Nonlinear
399		viscoelastic characterization of bovine trabecular bone. Biomech Model
400		Mechanobiol 2016;16:173–89.
401	[16]	Lujan TJ, Henderson CE, Madey SM, Fitzpatrick DC, Marsh JL, Bottlang M.
402		Locked plating of distal femur fractures leads to inconsistent and asymmetric
403		callus formation. J Orthop Trauma 2010;24:156–62.

- 404 [17] Henderson CE, Lujan TJ, Kuhl LL, Bottlang M, Fitzpatrick DC, Marsh JL. 2010
 405 Mid-America Orthopaedic Association Physician in Training Award: Healing
- 406 complications are common after locked plating for distal femur fractures. Clin
 407 Orthop Relat Res 2011;469:1757–65.
- 408 [18] Gautier E, Sommer C. Guidelines for the clinical application of the LCP. Injury
 409 2003;34:B63-76.
- 410 [19] MacLeod AR, Pankaj P, Simpson AHRW. Does screw-bone interface
- 411 modelling matter in finite element analyses? J Biomech 2012;45:1712–6.
- [20] Nalla RK, Kinney JH, Ritchie RO. Mechanistic fracture criteria for the failure of
 human cortical bone. Nat Mater 2003;2:164.
- [21] Oni OOA, Capper M, Soutis C. A finite element analysis of the effect of pin
 distribution on the rigidity of a unilateral external fixation system. Injury
 1993;24:525–7.
- 417 [22] Ali AM, Burton M, Hashmi M, Saleh M. Treatment of displaced bicondylar tibial
- 418 plateau fractures (OTA-41C2&3) in patients older than 60 years of age. J
- 419 Orthop Trauma 2003;17:346–52.
- 420 [23] Board TN, Yang L, Saleh M. Why fine-wire fixators work: an analysis of
- 421 pressure distribution at the wire-bone interface. J Biomech 2007;40:20–5.
- [24] Steiner JA, Ferguson SJ, van Lenthe GH. Screw insertion in trabecular bone
 causes peri-implant bone damage. Med Eng Phys 2016;38:417–22.
- 424 [25] Steiner JA, Ferguson SJ, van Lenthe GH. Computational analysis of primary
- 425 implant stability in trabecular bone. J Biomech 2015;48:807–15.
- 426 doi:10.1016/j.jbiomech.2014.12.008.
- [26] Pankaj P. Patient-specific modelling of bone and bone-implant systems: the
- 428 challenges. Int j Numer Method Biomed Eng 2013;29:233–49.

- [27] Bowman SM, Keaveny TM, Gibson LJ, Hayes WC, McMahon TA.
- 430 Compressive creep behavior of bovine trabecular bone. J Biomech
 431 1994;27:301–10.
- [28] Fondrk M, Bahniuk E, Davy DT, Michaels C. Some viscoplastic characteristics
 of bovine and human cortical bone. J Biomech 1988;21:623–30.
- 434 [29] Pankaj P. Computational biomechanics of bone. In: Simpson H, Peter A,

435 editors. Exp. Res. Methods Orthop. Trauma, 2015.

- 436 [30] Goffin JM, Pankaj P, Simpson AHRW, Seil R, Gerich TG. Does bone
- 437 compaction around the helical bladeof a proximal femoral nail anti-rotation
- 438 (PFNA) decrease the riskof cut-out?: A subject-specific computational study.
- 439 Bone Jt Res 2013;2:79–83.
- 440 [31] Taddei F, Schileo E, Helgason B, Cristofolini L, Viceconti M. The material
- 441 mapping strategy influences the accuracy of CT-based finite element models
- of bones: An evaluation against experimental measurements. Med Eng Phys
 2007;29:973–9.
- 444 [32] Cowin SC, Mehrabadi MM. Identification of the elastic symmetry of bone and
 445 other materials. J Biomech 1989;22:503–15.
- [33] Rho JY. An ultrasonic method for measuring the elastic properties of human
 tibial cortical and cancellous bone. Ultrasonics 1996;34:777–83.

448 [34] Zysset PK. A review of morphology-elasticity relationships in human trabecular
bone: Theories and experiments. J Biomech 2003;36:1469–85.

- 450 [35] Donaldson FE, Pankaj P, Cooper DML, Thomas CDL, Clement JG, Simpson
- 451 AHRW. Relating age and micro-architecture with apparent-level elastic
- 452 constants: a micro-finite element study of female cortical bone from the
- 453 anterior femoral midshaft. Proc Inst Mech Eng Part H-Journal Eng Med

454 2011;225:585–96.

- [36] Goffin JM, Pankaj P, Simpson AH. The importance of lag screw position for the
 stabilization of trochanteric fractures with a sliding hip screw: A subject-specific
 finite element study. J Orthop Res 2013;31:596–600.
- [37] Schileo E, Taddei F, Cristofolini L, Viceconti M. Subject-specific finite element
 models implementing a maximum principal strain criterion are able to estimate
 failure risk and fracture location on human femurs tested in vitro. J Biomech
 2008:356–67.
- 462 [38] Goffin JM, Pankaj P, Simpson AH. Are plasticity models required to predict
- relative risk of lag screw cut-out in finite element models of trochanteric
 fracture fixation? J Biomech 2014:323–8.
- [39] Bayraktar HH, Keaveny TM. Mechanisms of uniformity of yield strains for
 trabecular bone. J Biomech 2004;37:1671–8.
- 467 [40] Levrero-Florencio F, Margetts L, Sales E, Xie S, Manda K, Pankaj P.
- 468 Evaluating the macroscopic yield behaviour of trabecular bone using a
- 469 nonlinear homogenisation approach. J Mech Behav Biomed Mater
- 470 2016;61:384–96.
- 471 [41] Naghdi PM, Trapp JA. The significance of formulating plasticity theory with
- reference to loading surfaces in strain space. Int J Eng Sci 1975;13:785–97.
- 473 [42] Pankaj P, Donaldson FE. Algorithms for a strain-based plasticity criterion for
- bone. Int j Numer Method Biomed Eng 2013;29:40–61.
- 475 [43] Basler SE, Traxler J, Müller R, van Lenthe GH. Peri-implant bone
- 476 microstructure determines dynamic implant cut-out. Med Eng Phys

477 2013;35:1442–9.

478 [44] Bianco R-J, Aubin C-E, Mac-Thiong J-M, Wagnac E, Arnoux P-J. Pedicle

- 479 screw fixation under nonaxial loads: A cadaveric study. Spine (Phila Pa 1976)
 480 2016;41:124–30.
- [45] Born CT, Karich B, Bauer C, Von Oldenburg G, Augat P. Hip screw migration
 testing: First results for hip screws and helical blades utilizing a new oscillating
 test method. J Orthop Res 2011;29:760–6.
- [46] Xie S, Manda K, Pankaj P. Bone's time-dependent behaviour accentuates
 loosening in fracture fixation using bone-screw systems. Bone Jt Res 2018, (in
 press).
- 487 [47] Schoenfeld CM, Lautenschlager EP, Meyer PR. Mechanical properties of
- 488 human cancellous bone in the femoral head. Med Biol Eng 1974;12:313–7.
- [48] [48] Zilch H, Rohlmann A, Bergmann G, Kölbel R. Material properties of femoral
- 490 cancellous bone in axial loading. Part II: Time dependent properties. Arch
 491 Orthop Trauma Surg 1980;97:257–62.
- 492 [49] Bowman SM, Guo XE, Cheng DW, Keaveny TM, Gibson LJ, Hayes WC, et al.
- 493 Creep contributes to the fatigue behavior of bovine trabecular bone. J Biomech
 494 Eng 1998;120:647–54.
- 495 [50] Guedes RM, Simões JA, Morais JL. Viscoelastic behaviour and failure of
- 496 bovine cancellous bone under constant strain rate. J Biomech 2006;39:49–60.
- 497 [51] Xie S. Characterisation of time-dependent mechanical behaviour of trabecular
- 498 bone and its constituents. PhD Thesis, The University of Edinburgh, 2018.

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	• Fundamental mechanics of the bone-fixator construct with focus on interfacial
	strains that result in loosening are discussed
•	 Bone models as time-independent and time-dependent material that have
	been used to simulate and predict loosening are reviewed
•	 Capability of time-dependent models to capture cyclic accumulated
	deformation at bone-pin/ interface is highlighted

- Figure 1 Locking plate used for mid-shaft fracture fixation: prior to load application (a) and after load application (b); pattern of large strains at the bone screw interface for screws 2 and 3 (c); compressive and tensile strain distributions for the near cortex for screw 2 (d). Unilateral fixators present similar strain patterns.
- Figure 2 Ilizarov ring-wire external fixator construct (a); the deformed shape of bone-wire system with regions of large interfacial bone strains (b).
- Figure 3 Predicted volumes of bone above 0.02% equivalent strain (EqEV) for different working lengths. (a) Screw arrangements C123; C234; and C345. EqEV values at different screw locations for (b) healthy bone and (c) osteoporotic bone. Load of 250N is applied to the bone-fixator construct. Reproduced from MacLeod et al. [4] (open access)
- Figure 4 Geometry of the bone-screw system showing symmetry surface with location of load application (a); section A-A (b); load application - each model was subjected to 500 cycles of triangular load of 300 N amplitude followed by 1000 s of recovery (c). From Xie et al. [46] (open access)
- Figure 5 Compressive (a, b and e) and tensile (c, d and f) strain (%) contours from the symmetry surface and Section A-A. Three representative cycles were selected to show the strain accumulation with increasing cycle number when load is at its peak (a and c); at the time points when load is zero (b and d); and recovery after 1000 s (c and f). Redrawn from Xie et al. [46] (open access)





Figure 3 Click here to download high resolution image





Figure 5 Click here to download high resolution image



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