Mechanical Testing of the Bone-Implant Interface

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***Abstract*—Orthopaedic implants are frequently used for both bone repair and joint replacement. In both cases the mechanical stability of the implant is essential to prevent fibrous encapsulation and the formation of wear debris. Critical instability may even cause the implant to fail, necessitating a costly and dangerous revision surgery.**

In this report we first review the complications arising during the mechanical testing of bone, explaining the typical physiological loading of bone, noting differences between cortical and cancellous bone, before highlighting the hierarchical and anisotropic nature of the tissue. We then assess the macro and micro factors that directly affect osteogenesis, considering implant geometry, surface finish and local stability. The report culminates by introducing specific testing methodologies - and their limitations - for some loading regimes.

CONTENTS

I. INTRODUCTION

Bone is biological material consisting of a series of hierarchically arranged, mineralised, collagen fibres [1], containing both organic and inorganic phases, with the organic phase being capable of mechanotransduction, renewal, repair, remodelling and even the formation of new bone [2]. This last process - known as osteogenesis - allows for bone to grow into implanted material in suitable conditions [3], vastly improving mechanical stability [4].

II. COMPLICATION IN THE MECHANICAL TESTING OF BONE

A. Physiological Loading of Bone and the Bone-Implant Interface

During any given day our bones can be expected to resist forces generated by gravity, our individual weights, contracting muscles and any external forces [5], however in healthy bone these forces are not sufficient to induce fracture, which is instead typically caused by trauma. In any given impact the forces acting on bone can be classified into a combination of five categories [6]: compression, tension, shear, torsion and bending. The unique micro and macro formulation of bone leads to different mechanical responses to each of the five loading regimes, and as such the nature of the fracture is determined by its loading [7].

Likewise, under physiological loading, the bone-implant interface experiences a combination of the five forces. Different mechanical testing methodologies try to isolate and test the implants response to a single force under conditions designed to replicate the *in vivo* environment. How different testing modalities go about achieving this will be discussed in detail later in the report.

B. Cortical vs Cancellous

Bone can be subdivided into two regions: cortical and cancellous [8]. Around 80% of all bone tissue in the human body is cortical, although the exact ratio can vary between different bones - for example the radial diaphysis contains 95% cortical bone while the vertebra contains only 25% [9]. Cortical bone is dense, and has anywhere between 8 and 115 times the compressive strength of cancellous bone [10]. It is also much stiffer, with a Young's modulus of 7-30 GPa [10]. The material properties of cortical bone make it an ideal material to bear and resist mechanical loads, and unsurprisingly it's purpose is to do just that: bone must be able to support the body (e.g. the spine), protect organs (e.g. the rib cage) and act as a lever arm (e.g. the radius and ulna) [8].

Cancellous bone totals only 20% of all bone found in the human body [9]. Mechanically, it is far weaker than cortical bone, both in compression (compressive strength of 2-12 MPa [10]) and tension (tensile strength of 10-20 MPa [10]). However the surface area to volume ratio of cancellous bone is far greater than cortical bone (20 vs 2.5 mm^2/mm^3 [11]), making it more suitable for metabolic activity and ion exchange [8].

TABLE I: A quantitative comparison between cortical and cancellous bone.

	Cortical Bone	Cancellous Bone
Percentage of Bone (%)	80 [9]	20 [9]
Density (g/cm^3)	1.9 [12]	0.05 to 1.1 [13]
Volume Fraction (mm^3/mm^3)	0.85 - 0.95 [11]	0.05 - 0.60 [11]
Surface/Volume (mm^2/mm^3)	2.5 [11]	20 [11]
Compressive Strength (MPa)	100-230 [10]	2-12 [10]
Tensile Strength (MPa)	50-150 [10]	10-20 [10]
Strain to Failure (%)	1-3 [10]	5-7 [10]
Young's Modulus (GPa)	7-30 [10]	0.005-0.05 [10]

C. The Hierarchical Structure of Bone

Like with most biological tissue, the formation of bone creates an inherently hierarchical structure, knowledge of which is essential to understand the results of any mechanical test. We will begin by considering cortical bone, which is formed from multiple functional units called osteons. Each osteon is formed through the motion of an osteoclasts, cells able to resorb bone. The osteoclast cleaves through either precursor tissue (e.g. cartilage or woven bone) or existing bone, as osteoblasts trail behind, slowly depositing bone in a lamellar morphology [14], forming what Giraud-Guille calls the *"twisted plywood"* structure of bone [15].

It is important to note that there is no "guiding hand" arranging bone into this lamellar structure, the hierarchy seen in Figure 1 is simply a product of how bone is deposited: we can therefore say the structure of bone is formed through self-assembly [16].

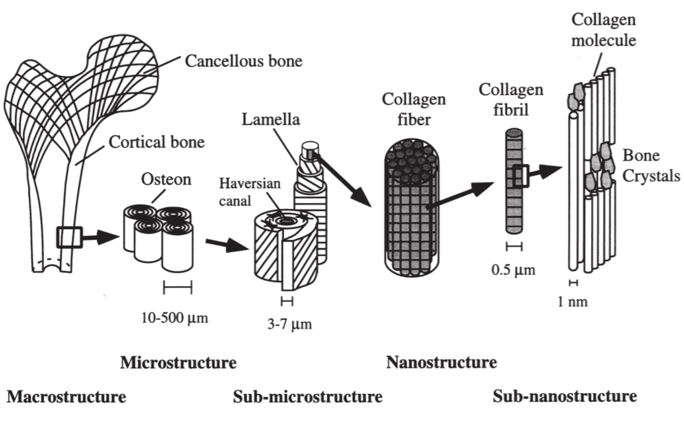


Fig. 1: The hierarchical structure of cortical bone. Image taken from Rho *et al.*, 1998 [17].

The hierarchical nature of bone complicates the process of mechanical testing. Are you interested in measuring macroscopic samples of bone [18], probing individual fibrils [19] or even studying nanoscale deformations [20]? Testing at each scale introduces different complications and necessitates unique testing methodologies.

For the purpose of this report we will consider the bone-implant interface in the macroscale, using conventional mechanical testing apparatus to measure micromotion, shear and tensile forces.

D. The Anisotropy of Bone

Both cortical and cancellous bones exhibit mechanical anisotropy, but for different reasons. The self-assembly process of cortical bone forms osteon channels that travel along the long axis of the bone. One would intuitively expect the collagen to also align in this direction, however research by Weiner *et al.* has identified that each osteon is composed of five distinct lamellar arrays, with the orientation of the collagen fibrils offset by 30° in each successive unit [21]. This can be seen in Figure 2. Note however that there is not an even number of collagen fibres aligned at each angle, the distribution is bimodal with peaks at 30° and 70° [21].

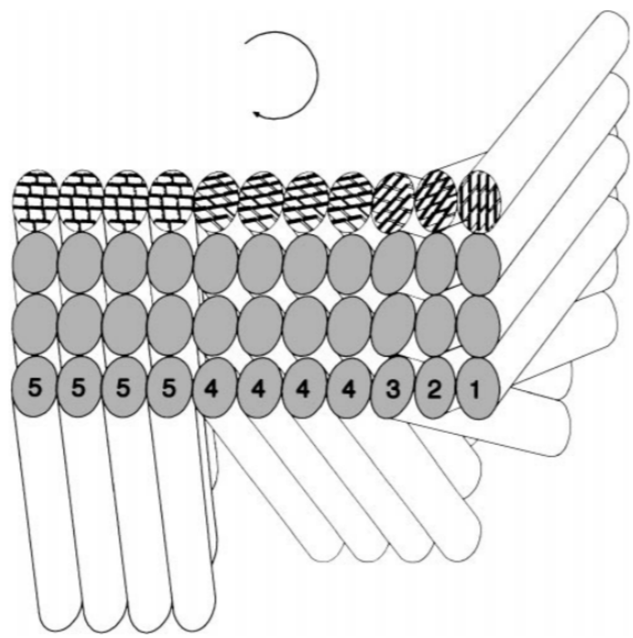


Fig. 2: Weiner's five layer model of lamellar bone. There is a greater frequency of collagen fibres orientated at approximately 30° and 70° from the osteon axis, matching experimental data [21]. The "brick wall" pattern in the first row depicts the orientation of mineral plates, which also vary in direction [22]. Image taken from Weiner *et al.*, 1999 [23].

Weiner's bimodal model [21] matched existing experimental data found by Bonfield and Grynpas, who measured the Young's modulus of bovine femur samples, cut at different angles from the long axis [24]. Unsurprisingly bone is strongest when loaded parallel to the axis of the femur, but there exists two peaks at 30° and 70° [24]. They concluded that cortical bone cannot be modelled as a unidirectional fibre reinforced composite and that *"an alternative model is required to account for the dependence of Young's modulus on orientation"* [24]

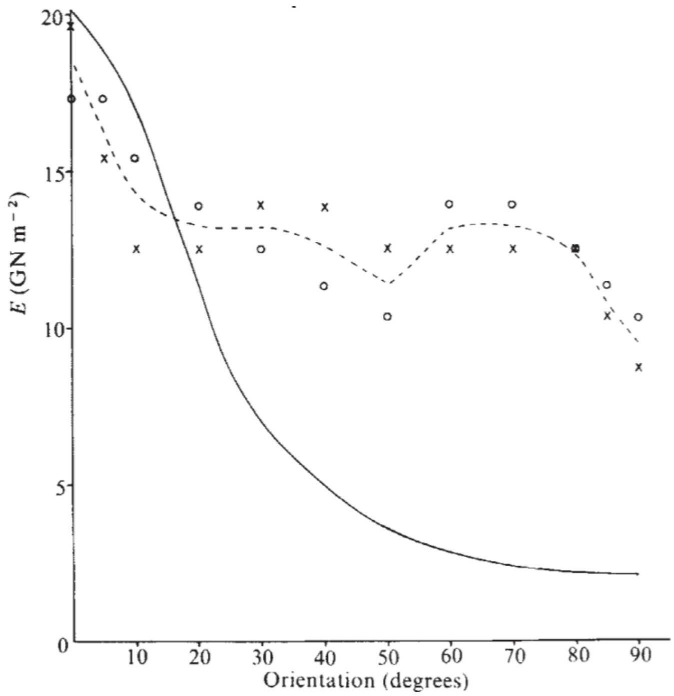


Fig. 3: Bonfield and Grynpas measured the Young's modulus of bovine femur bone at different angles relative to the long axis. $\circ$ represents measurements for wet bone and $\times$ for dry. The solid line is the theoretical curve for a unidirectional fibre reinforced composite. Their results support Weiner's bimodal model in which the orientation of collagen fibres peaks at 30° and 70°. Note that the minimum value of E for bone is significantly higher that that the theoretical composite. Image taken from Bonfield and Grynpas, 1977 [24].

This multi-directional collagen fibre orientation has significant implications for implant stability. Bony ingrowth into implants happens transversely to the plane of the bone, but the five layer structure of cortical bone provides a greater stiffness when loaded at 90° as compared to a simple unidirectional fibre composite (see Figure 3).

Cancellous bone also exhibits anisotropic properties through obeying Wolff's law, which states that bone is capable of remodelling in response to the mechanical stresses imposed on it [25]. As such the trabeculae arrange themselves parallel to direction of the mechanical load [26], resulting in differing strengths when stressed in varying directions [27], [28].

III. Factors Affecting the Bone Implant Interface

A. Geometry

There are two main ways in which the macroscopic geometry of an implant can affect the bone-implant interface, and both require maximising the contact area with bone. To illustrate this we consider how the geometry of a bone screw affects pullout strength.

Implants are often affixed into bone using specialized screws, the threads of these screws are designed to mechanically interlock with bone, preventing pullout through physical obstruction of motion. Therefore we can intuitively predict that by maximising the screw-bone bearing surface we can increase pullout strength. Research by Chapman *et al.* has modelled the mathematical theory behind screw pullout to create Equation 1 [29]:

$$F_s = S \times (L \times \pi \times D_{major}) \times TSF \tag{1}$$

where:

$$\begin{aligned} F_s &= \text{predicted shear failure force } [N] \\ S &= \text{material ultimate shear stress } [MPa] \\ L &= \text{thread length } [mm] \\ D_{major} &= \text{major diameter } [mm] \\ TSF &= \text{dimensionless thread shape factor } [N] \end{aligned}$$

Where the thread shape factor (TSF) has been experimentally found to be:

$$TSF = \left(\frac{1}{2} + 0.57735\frac{d}{p}\right) \tag{2}$$

where d is the thread depth and p is the thread pitch.

Equation 1 shows us we can maximise pullout force through maximising the bearing surface of the screw. We can do this through increasing the length of the threaded region, L, the screw diameter, D, and the thread shape factor, TSF. Chapman's equation also notes that selecting a material with a higher shear failure force can increase pullout resistance [29].

Looking at Equation 2 we can note that pullout strength increases as the thread depth of a screw increases, and decreases as the pitch of a screw increases [29].

The design of bone screws can be explained using Chapman's equations. For example, an analysis by Asnis and Kyle has found that screws designed to be implanted into cancellous bone have a larger TSF to compensate for the lower pullout strength as compared to cortical bone [30].

However Equation 1 does not tell a complete story of the stresses involved in screw design: increasing the major diameter of the screw, D_{major}, will increase pullout strength, but it also increases stress concentration in the surrounding bone. Assuming the screw diameter is much smaller than the surrounding bone, such that we can model the bone surface as an infinite element, we can calculate the stresses around the screw using Equation 3 [31]:

$$\begin{aligned} \sigma_r &= \frac{1}{2}\sigma\left(1 - \frac{a^2}{r^2}\right) + \frac{1}{2}\sigma\left(1 - \frac{4a^2}{r^2} + \frac{3a^4}{r^4}\right)cos2\theta \\ \sigma_\theta &= \frac{1}{2}\sigma\left(1 + \frac{a^2}{r^2}\right) - \frac{1}{2}\sigma\left(1 + \frac{3a^4}{r^4}\right)cos2\theta \\ \tau_{r\theta} &= -\frac{1}{2}\sigma\left(1 + \frac{2a^2}{r^2} - \frac{3a^4}{r^4}\right)sin2\theta \end{aligned} \tag{3}$$

where:

$a =$ the hole radius

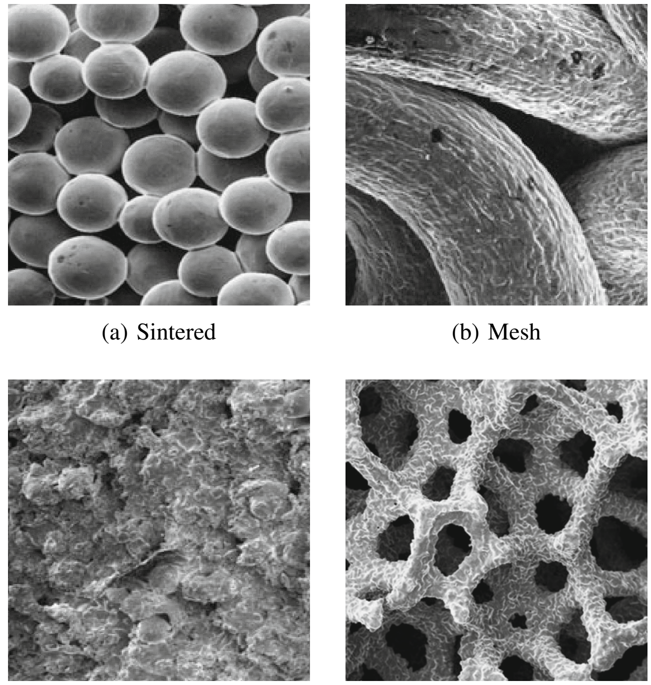
$r =$ radius to the point of interest

$\theta =$ polar angle

The most striking observation from Equation 3 is that the stresses in the surrounding bone are non-linearly related to the diameter of the screw (represented in this equation by a). Designers must strike a balance between maximising pullout strength and reducing stress concentration, particularly in weak cancellous bone [32].

The second mechanism by which implant geometry affects the bone-implant interface is much easier to understand. Bony ingrowth happens as osteoblasts migrate into the implanted material and begin to deposit bone [33], it stands to reason that an increase in bone-implant contact area would increase osteogenesis.

However, larger implants require more invasive operations, and necessitate the destruction of a greater area of periosteum. As such some researchers have proposed small footprint, periosteum preserving implant designs (see Perren's low contact dynamic compression plate [34]).



(a) Sintered (b) Mesh (c) Plasma (d) Porous

Fig. 4: Electron microscopy of different surface treatments used to promote bony ingrowth. Images taken from Karachalios *et al.*, 2014 [36].

B. Surface

For bony ingrowth to occur there must be surface for the osteoblasts to proliferate into and deposit new bone. A perfectly polished implant cannot accommodate this, and as such various processing methods are used to increase the porosity of the implant surface. This report chooses to neglect chemical processing methods such as acid etching [35], instead briefly highlighting four popular mechanical surface treatments: sintering beads, fibre meshing, plasma spraying and use of a porous metal (see Figure 4.)

Sintered beads are structural microspheres deposited and attached onto the surface of an implant through, unsurprisingly, sintering. The porosity of a sintered bead surface can be controlled by sintering time, temperature and even initial bead diameter, which can be used to optimise implant integration with native bone [37]. The primary advantage of a sintered bead surface is their excellent bond strength and high abrasion resistance [36], important for reducing fretting corrosion.

Titanium fibres are woven into meshes and adhere to the surface of the implant via diffusion bonding [36]. Similar to sintering, meshing is a additive process, no material is removed from the underlying implant. Meshes have been integrated with bone morphogenetic proteins to aid with bone growth [38].

For plasma spraying metal powders are heated until they ionise, they are then mixed with an inert gas and sprayed at high pressure onto the implant surface. They have weaker mechanical bonding, abrasion and wear as compared to sintered beads and fibre meshes, but retain over 90% of the implants fatigue strength (compared to 50% for the previous procedures [39]).

Porous metals are the most recent development in implant design [36] and are unique in that the porous surface is not actually a surface, it extends throughout the implant, allowing for a greater degree of bony ingrowth as compared to all other procedures. The open cell structure of metal foams also enable osteoconductive promoters to be found in greater numbers and deeper in the implant, patents exist to manufacture medicated porous metal prosthetics [40]. The advent of metal foam implants have also facilitated the development of novel implant fabrication procedures, such as additive manufacturing [41]. Additionally, it is not necessarily disadvantageous that metal foams are mechanically weaker than traditional implants, stress shielding is a serious problem in implant design caused by the modulus mismatch between implant and underlying bone [42]. Researches have managed to optimise the porosity of metal foam implants to match cortical bone [43].

This is a brief summary of the most popular surface finish treatments for implant design, however researchers continue to develop novel procedures to optimise implant topography. Consider a simple titanium implant, Brammer *et al.* have used electrochemical anodization to grow TiO_2 nanotubes from the surface [44], Götz *et al.* have used lasers to ablate their ideal topography [45] and Munir *et al.* have instead used template-assisted electrohydrodynamic atomization to deposit nanometer-scaled SiHA on the titanium [46]. Additionally it is worth considering how different surface finish treatments can be combined in tandem to create implants with a bimodal distribution of porosity [47]. The differences in production time, cost, toughness, abrasion resistance and scalability prevent any one treatment being singularly dominant.

C. Stability

There are two mode of fracture healing in bone, primary (direct) and secondary (indirect). The degree of strain determines which mode occurs, with primary healing happening when $0\% < \epsilon < 2\%$ and secondary when $2\% < \epsilon < 10\%$ [48]. Primary bone healing proceeds through immediate Haversian remodelling of the fracture [49], while secondary bone healing is a four stage process of immediate inflammation, soft callus formation, woven bone creation and final remodelling [50].

Likewise a high degree of stability is necessary for osteointegration of an implant, with micromotions between 40 and 150μm causing the growth of fibrocartilage [33] and micromotions above 150μm leading to fibrous encapsulation [51].

Primary bone healing is not necessarily better than secondary bone healing, the process is much slower [52] and the lack of callus formation makes it *"impossible to assess the state of healing of the fracture"* [53] In fact, some degree of micromotion has been found to promote bone growth [54]. For any given fracture orthopaedic surgeons must decide on using an implant designed for primary or secondary healing. While secondary bone healing implants (such as intramedullary nails or Kirschner wires [14]) have faster healing time and the benefits of micromotion, primary healing fixations (such as bone plates and lag screws [55]) have greater mechanical stability and the option of osteointegration.

Note however that it is much easier to achieve secondary bone healing [14] and that researchers have been able to adapt rigid fixation devices to allow for some degree of beneficial micromotion: Foux *et al.* have developed a bone plate with polymeric "*cushions*" around the screw to allow for controlled lateral motion [56].

IV. Testing of the Bone-Implant Interface

A. Measuring Micromotion

Recall that the type of tissue that forms around an implant is directly dependent on the micromotion of said implant [33]. Therefore, researchers need a method of quantifying the physiological micromotion of any given implant design. Unfortunately the testing method is dependent on the type of implant being studied, an experiment to measure micromotion in a dental implant cannot then be used to test the micromotion of a bone plate. However similarities can be drawn between different testing modalities: they all apply physiological loading to the bone-implant interface and attempt to measure deflection in a *in vitro* environment. See Table II for a brief critical review of existing testing modalities for measuring micromotion in an implanted femoral stem.

B. Sensing Shear

The pullout and pushout tests measure an implant's resistance to shear, and are the most popular methods for testing the bone-implant interface due to fact that the method only requires a uniaxial testing machine and accompanying jig [57]. Pullout and pushout testing has been used to assess the effect different materials [58], surface finishes [59] and biochemical coatings [60] [61] have on osteoinduction.

While many different testing methods have been invented to measure micromotion, most pullout and pushout tests share a similar procedure: a defect is created on an animal model, a rod-like implant is inserted and allowed to osteointegrate with the bone. After a set amount of time the animal is sacrificed and destructive testing of the bone-implant interface can begin. A schematic diagram of a standard pushout test is shown in Figure 5

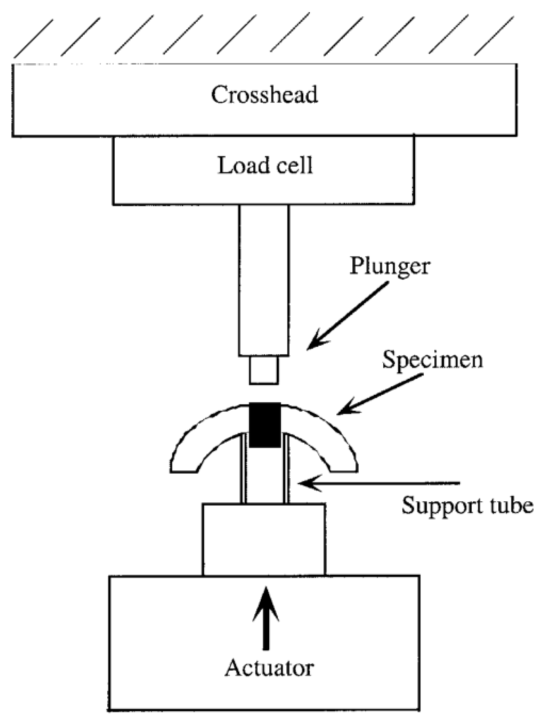


Fig. 5: Schematic diagram of a pushout test. Image reproduced from Berzins and Sumner [57] which was modified from Cook *et al.* [62].

Typically a cylindrical geometry is used for pullout and pushout tests, both because the antisymmetric shape prevents stress concentration and it is trivial to preform surface area calculations [57] (shear stress is directly proportional to surface area).

The use of a uniaxial testing machine results in pullout and pushout tests sharing a source of systematic error with standard tensile testing procedures: the displacement of the crosshead may not be the displacement of the specimen. Fortunately the problem can be easily resolved by attaching strain transducers directly to the sample, and using them to measure displacement [63].

Bone is a viscoelastic material [64], and theoretically the mechanical response of bone should be rate dependent *et al.*. Experimentally there is some disagreement on whether the viscosity of bone is negligible during pullout or pushout tests [65], or whether viscosity has a significant effect as high displacement rates [66].

The site of implant fixation dramatically influences the results of the pullout or pushout test. Research by Stone *et al.* has identified a non-linear relationship between shear strength and density [67]:

$$S = 21.6\rho^{1.65}$$

As such we can predict that cancellous bone will have a significantly lower pullout strength (density of 0.05 to 1.1 g/cm^3 [13]) as compared to cortical bone (density of 1.9 g/cm^3 [12]). Researchers must pick a suitable bone to fix their implant into, for example: pedicle screws should be tested in cancellous bone [68] while bone plate lag screw should be set in corticle bone [69].

The results of a typical pullout/pushout test can seen in Figure 6. Note the similarities to a standard tensile stress strain curve. The implant interface first undergoes recoverable elastic deformation (represented by the linear region), before suffering permanent plastic deformation and rapid failure after yielding. A well integrated implant would have a tough interface with the bone, allowing for more energy to be absorbed before fracture.

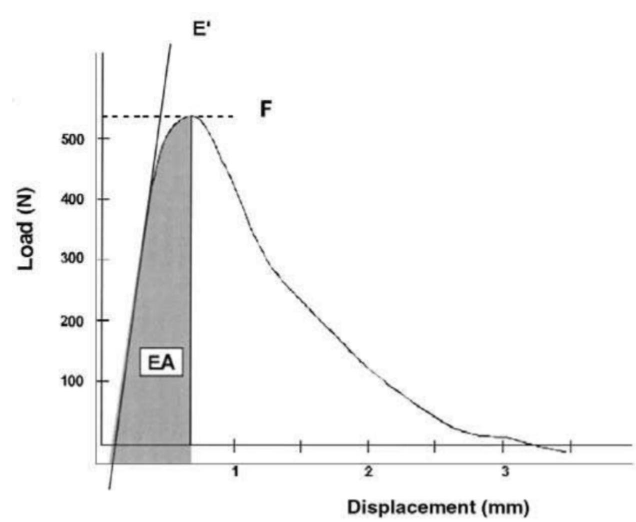


Fig. 6: Typical load-displacement from a pullout/pushout test. F = the maximum force applied to the implant, E' = the shear stiffness and the region EA = the energy absorbed before failure. Image reproduced from Berzins and Sumner [57]

C. Testing Tension

Despite being ubiquitous in materials science, tensile testing of the bone implant interface has not been explored nearly as thoroughly as other testing modalities. Tensile testing uniquely is able to measure the biochemical adhesion of the implant, neglecting the *"resistive force due to surface roughness of the biomaterial"* [70]. This makes it ideal for testing the effect of different surface coatings or osteopromotive factors.

There is no standard implant tensile test used by researchers, instead we highlight a proposal by Nakamura *et al.* in Figure 7. This method has successfully been used on titanium [71], tantalum [72] and PMMA [73] implants.

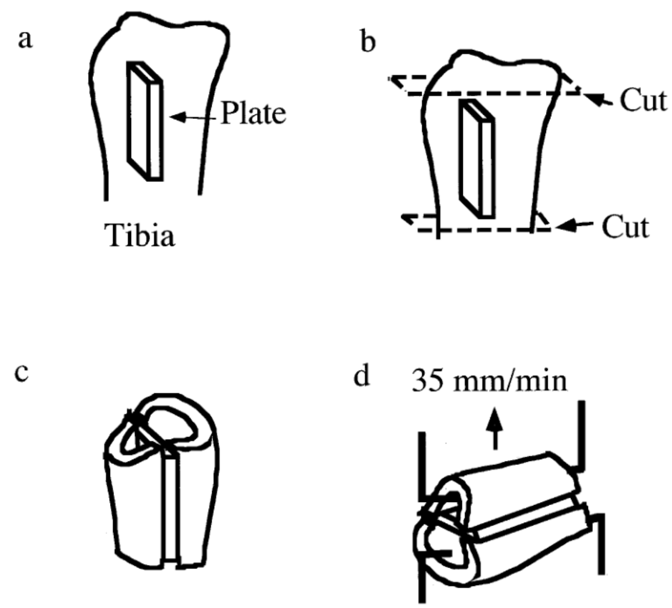


Fig. 7: Nakamura's method for tensile testing of the bone implant interface. Subfigure **a** shows a rectangular plate (15 x 10 x 2 mm) implanted into the tibia of a rabbit. The implant material, porosity and coating are determined by the goals of the experiment. **b** depicts preparation of the sample for tensile testing following sacrifice of the rabbit after a predetermined period of time. **c** is the prepared sample ready for testing, while subfigure **d** shows the tibia bone clamped into a standard universal testing machine retracting at a rate of 35 mm/min (recall that bone is viscoelastic and as such the mechanical response is strain rate dependent). Image reproduced from Nakamura and Nishiguchi [70]

TABLE II: Methods for Measuring Micromotion in the Bone-Implant Interface of a Femoral Stem

Title	Author	Year	Method	Advantages	Disadvantages	Source
Strains and micromotions of press-fit femoral stem prostheses.	Walker *et al.*	1987	First a designed femoral stem was fixed into four fresh human cadaveric bones, before rosette strain gauges were attached (using cyanoacrylic glue) to measure displacement. To measure micromotion steel pins are inserted into the implant through the femur and eddy current transducers were used to measure relative motion.	Eddy current transducers are a non-contact method of measuring displacement as compared to LVDTs and so cause minimal disturbance to bone samples.	This is an old study, written when no standard stem design was dominant. As such they actually designed their own stem, which (with the benefit of hindsight) we now know to be an inferior design. Furthermore the strain gauges used only have a precision of 5µm.	[74]
Methods for quantitative analysis of the primary stability in uncemented hip prostheses.	Monti *et al.*	1999	Four linear variable displacement transducers (LVDT) were inserted into a synthetic femur specimen which had been which had just undergone a simulated total hip arthroplasty (THA). The distal femur was then secured to a hinge connected to a biaxial load cell. The femur construct was subjected to cyclic load approximating stair climbing (high torque is the most common mode of failure following THA [75] and torques are highest during stair climbing [76]).	This was the first paper to comment on - and measure - the influence of loading frequency. They also measure the deformation of cortical bone important as that factors that in to micromotion calculations.	They used a synthetic femur for validation. They then talked about how repeatable alignment is necessary for accurate measurements, a problem that is clearly trivial to overcome when identical synthetic femurs are used. Finally, the LVDTs are drilled into bone, which may crack in osteoporotic patients (the people actually most likely to get the THA).	[77]
Simultaneous and multisite measure of micromotion, subsidence and gap to evaluate femoral stem stability.	Gortcha-cow *et al.*	2012	A cadaveric human femur is reamed and 500 stainless steel radio-opaque beads are placed along the length of the medullary cavity. 12 tantalum markers were printed onto a femoral stem which was then fixed into the bone. A load of 2000N is applied to the head of the stem and the deflection is recorded using a CT scan.	The CT scan data can be digitalised for use in computer modelling, this would not be possible if only four LVDTs were used.	The distal end of the femur was cemented, in reality this would move on a hinge (knee joint) and the viscoelastic cartilage would be able to absorb some stress. Additionally, there is uniaxial application of force on the femur, physiologically the direction and magnitude of force changes during the gait cycle.	[78]
A biomechanical testing system to determine micromotion between hip implant and femur accounting for deformation of the hip implant: Assessment of the influence of rigid body assumptions on micromotions measurements.	Leuri-dan *et al.*	2017	A femoral stem was implanted into either synthetic or cadaveric bones. A frame is fixed around the bone to act as a template for drilling and securing the LVDTs. The LVDTs have a unique design with pins being unconstrained against their brushing surface.	This is the only design that did not have to make rigid body assumptions for bone. Additionally, they used both synthetic and cadaveric femurs for validation.	The distal end of the femur has been secured in a fibreglass putty with no ability for motion (knee) and no compensation representing the dampening forces of the cartilage. Finally, the frame adds weight onto the bone so that it is already loaded before testing.	[79]

V. Conclusion

Accurate assessment of the bone-implant interface is a difficult task complicated further by the complex nature of biological tissue. This report ends with ten key conclusions:

1) The physiological loading of bone is too complex to be entirely captured by any one testing modality.
2) Researchers must have a proper understanding of fracture mechanics when mechanically evaluating a chosen implant design (it would be disingenuous to test a new femoral stem in simple compression when most failure arise due to torsional loading [75]).
3) Consideration must be given to the physiological bone environment when designing *in vivo* tests (e.g. pedicle screws would not be implanted into a femur).
4) Bone is hierarchical and not all tests have to be preformed on the macro scale (histology, staining and immunofluorescence have all been used to assess the adhesion of osteoblastic cells on implant surfaces [80]).
5) Bone is anisotropic, but the *"twisted plywood"* [15] arrangement of collagen makes it stronger when loaded transversely than otherwise expected.
6) Bone is viscoelastic and strain rate must be specified in all mechanical tests.
7) Implant design is often a compromise (increasing screw diameter increases pullout strength [29] but also stress concentration [31]).
8) Osteointegration is dependant on surface finish but is completely impossible on an unstable implant, therefore proper assessment of implant micromotion is essential.
9) Pullout and pushout testing is a reliable, popular and simple assessment of implant-interface shear strength.
10) Conversely tensile testing is more complicated and specialised, but is able to neglect the effects of surface roughness.

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