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LECTURE NOTES 2

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Biomechanical Techniques for Pre-clinical Testing of Prostheses and Implants

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These Lecture Notes summarize and extend the lectures delivered by the author as Visiting Professor at the Institute of Fundamental Technological Research, Polish Academy of Sciences, Warsaw, Poland.

Contents

Preface

Prostheses and other medical devices can be subjected to a wide range of biomechanical tests before they are implanted. The primary purpose of *pre-clinical* tests is to protect patients against ineffective devices, but they can also reduce the number of animals used in animal experiments and provide strong evidence for the approval of clinical trials. Biomechanical tests include both experimental and computational analyses of implants for use in orthopaedics, cardiology, otology, dental implantology, and maxillofacial and other applications in reconstructive surgery.

Pre-clinical biomechanical tests must ascertain the initial (or primary) stability – by this it is meant the post-operative durability in the absence of biological adaptations – and the secondary stability – which means the durability after biological reactions of the tissues to the presence of the implant. Experimental techniques are well-suited to the evaluation of initial stability but they are not so well-suited to the analysis of biological adaptations. Computational techniques, on the other hand, may be more suited to the analysis of secondary stability. The application of computational techniques to determine secondary stability requires mechanobiological models to simulate tissue adaptation to mechanical stimuli.

In these Lecture Notes, a methodology for the development of biomechanical tests for various types of prostheses is described, and detailed examples are given.

Chapter 1

Introduction

Biomechanics may be defined as the science that applies the principles of mechanics to living systems. Of the multitude of living systems, the human body has been studied most. In particular, the development of prostheses, orthoses, and implants to restore function to diseased or injured limbs has received much attention from antiquity to the present day [1, 2]. For example, biomechanical engineering has been applied with great success to hip replacement and, more recently, to such complex technical tasks as replacement of the whole heart. The subject of mechanobiology is also part of biomechanics – it is concerned with discovering the relationship between tissue morphology and mechanical and physical stimuli. These Lecture Notes present a course of study that aims to provide the reader with a knowledge of the design of pre-clinical tests for various implant types. The application of mechanobiology in implant testing is also described.

The introductory part of these Lecture Notes are relevant to testing all types of load-bearing biomechanical implants for use in reconstructive surgery. Examples of tests for orthopaedic implants (hip and shoulder) are provided, the "micro-orthopaedic" system of the middle-ear implants is also considered.

The notes can be read in conjunction with the author's chapters in the *Bone Mechanics Handbook* [3, 4]. The author aims to provide the reader with the following:

- a rationale for pre-clinical testing of implants,
- an understanding of the relationship between prosthesis design and failure mechanics, using the hip prosthesis as an example,

- a methodology for the development of an experimental pre-clinical test (the hip prosthesis is used as an example),
- examples of computational methods for testing of implants, focusing particularly on those aspects that can be collaborated with experimental results,
- an idea of the role of mechanobiology in pre-clinical testing,
- a concluding discussion of the present state and future prospects of pre-clinical testing and its role in the implant innovation process.

Texts which the reader may find useful in complementing these Lecture Notes are the book by Buchhorn and Willert [5] which presents many technical chapters on various orthopaedic implants and on implant testing procedures, the book edited by Mow and Hayes [6], particularly the final two chapters on hip and knee replacement, the textbook by Będziński [7] dealing with the skeletal biomechanics and testing, particularly the chapters dealing with photoelastic methods. Books dealing specifically with strain measurement (Miles and Tanner [8]) and optical methods (Orr and Shelton [9]) have also been published. State of the art investigations are presented at the European Society of Biomechanics biennial meetings [10].

Chapter 2

The Pre-clinical Testing Imperative[1)]

2.1. General

Pre-clinical tests are those which could be carried out before the implant is put into clinical use. Because of implications for public health, most countries have regulations governing what can be implanted in a person. Prior to 1996, the countries of the European Union had widely differing rules but since then a harmonized system has been put in place. In the United States, the Food and Drug Administration (FDA) regulates the medical device market based on the

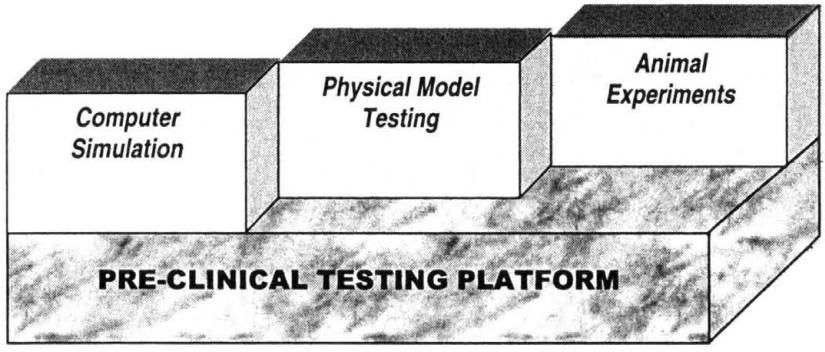


FIGURE 2.1. The three components of a pre-clinical biomechanical test.

[1)]This chapter is dealt with in greater detail in the author's paper titled *Issues in Pre-clinical Testing of Human Implants* co-authored with Dr Suzanne Maher and published in the *Journal of Materials Processing Technology*, 2001.

1976 Medical Devices Amendments Acts. Although there are fundamental differences between the European and American systems, both require *standard* tests to be carried out to ensure that a device is safe. Determining whether a device is efficacious (i.e. is capable of producing the intended results), and quantifying such efficacy, is the primary purpose of a *pre-clinical* test. One can imagine a pre-clinical testing platform consisting of three types of tests: experimental tests, computational tests, and animal trials (Fig. 2.1).

2.2. Purpose of pre-clinical tests

Pre-clinical tests can serve the following purposes:

Ensure the mechanical strength of new designs. Mechanical strength of implants must be sufficient so that they do not fail catastrophically in the body. Problems of implant durability are now quite rare, but some high profile cases have come to public attention in recent years. Three examples will suffice to illustrate this.

- The Capital hip prosthesis [11]. In this case, several hospitals in the UK reported implant failure of up to 21% of cases by five years. The femoral component appeared to be loosening early and there may have been extensive bone loss in the proximal femur.
- Silicone breast implants [12]. In this case, the silicone shell of the implants burst or begun to leak. The effects were particularly severe in the case where the fluid was silicone gel.
- The Bjork-Shiley heart valve [13]. In this case, one of the struts supporting the leaflets in the valve underwent fatigue failure due to a metallurgical defect in a particular batch of the implants.

Demonstrate the superiority of new designs over existing devices. It can take many years of clinical trials to demonstrate the superiority of one design of implant over another because several years of clinical trials are required before a statistical difference is seen regarding time to a revision operation. This creates an innovation barrier preventing new designs from coming onto the market. Pre-clinical tests allow evidence of implant superiority to be documented at an early stage in the implant innovation process, and this information can be used in making a decision to invest in further development of the device.

Reduce reliance on animal experimentation. Certain animals are suitable "animal models" for testing particular designs of implant; e.g. dogs and sheep for hip implants in orthopaedics, pigs for gastroenterological implants, cats for research on the ear, or rabbit femoral arteries to replicate human coronary arteries in tests of cardiovascular stents. However valuable animal experiments are assumed to be for toxicological or pharmacological experiments (i.e. testing cosmetics and drugs), they have often little relevance to biomechanical implants because these implants achieve their functionality by mechanical means. Furthermore, testing in animals is often criticized as being unnecessary and ethically unacceptable. Methods to minimize the use of animals come under three headings – the 3 Rs [14]:

- replacement (by non-sentient material),
- reduction (in the numbers of animals needed to obtain required data),
- and refinement (to decrease severity of inhumane procedures).

The relationship between pre-clinical testing and clinical testing is illustrated in Fig. 2.2. Pre-clinical testing is the first step in the procedure of releasing new implants onto the market. It is followed by prospective randomised studies (or randomised clinical trials, RCT) and multicentre studies. Finally, when the implant is released onto the market, an implant registry may be set up which tracks the survival of each implant.

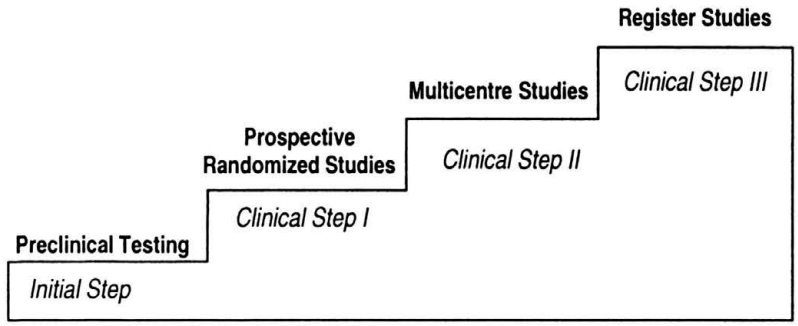


FIGURE 2.2. The place of pre-clinical testing relative to clinical tests for the introduction of implants onto the market, after Malchau [16].

There are difficulties with RCT in surgery which place even further importance on the pre-clinical testing step. Some of the problems with RCT in surgery may be listed as:

- the patient's refusal to accept the random group to which they have been allocated,
- the difficulty or impossibility of blinding the patient (and the operating surgeon) to which treatment has been used (i.e. to which prosthesis has been implanted, or to which treatment if a surgical and non-surgical treatment are to be compared). Sham surgical procedures can overcome the problem of blinding the patient but they are considered unethical by many,
- there is a learning-curve to an operative technique which will confound the results (no such learning-curve exists with administering a drug),
- results from specific validation centres are often not representative of the results from routine health care hospitals [15].

Chapter 3

Hip implants

3.1. Background

Because total hip arthroplasty is a well-known surgical procedure and the prosthetic components are probably familiar to most readers, it will be used as the main example in these Lecture Notes. The development of hip prostheses began in the late 19th Century and continues to be an active area of bioengineering innovation at the present time, see Fig. 3.1 (overleaf). Early success was prevented by infections and lack of biocompatibility of the materials; these problems were largely overcome by the innovations of the British Orthopaedic Surgeon John Charnley in the 1960s – he advocated polymethylmethacrylate as a bone cement and polyethylene for the acetabular socket. Total hip replacement involves the replacement of both the socket in the pelvis and the head of the femur to form a new ball-and-socket joint at the hip, see Fig. 3.2 for a schematic illustration.

Charnley also proposed the "low friction" concept, which used a small head on the prosthesis to keep torques to a minimum thereby minimizing the propensity for loosening. The disadvantage of this, of course, is that the stress is higher between the head and the cup leading to higher wear rates. Figure 3.2 shows a cementless fixation – in cementless designs the prosthetic components (cup and stem) are attached by osseointegration (bone ingrowth) to the surrounding bone. The alternative method, and the one that is most popular in many countries, is cemented fixation whereby a cement (polymethylmethacrylate with antibiotic and radio pacifier additives) is used to mechanically interlock the components

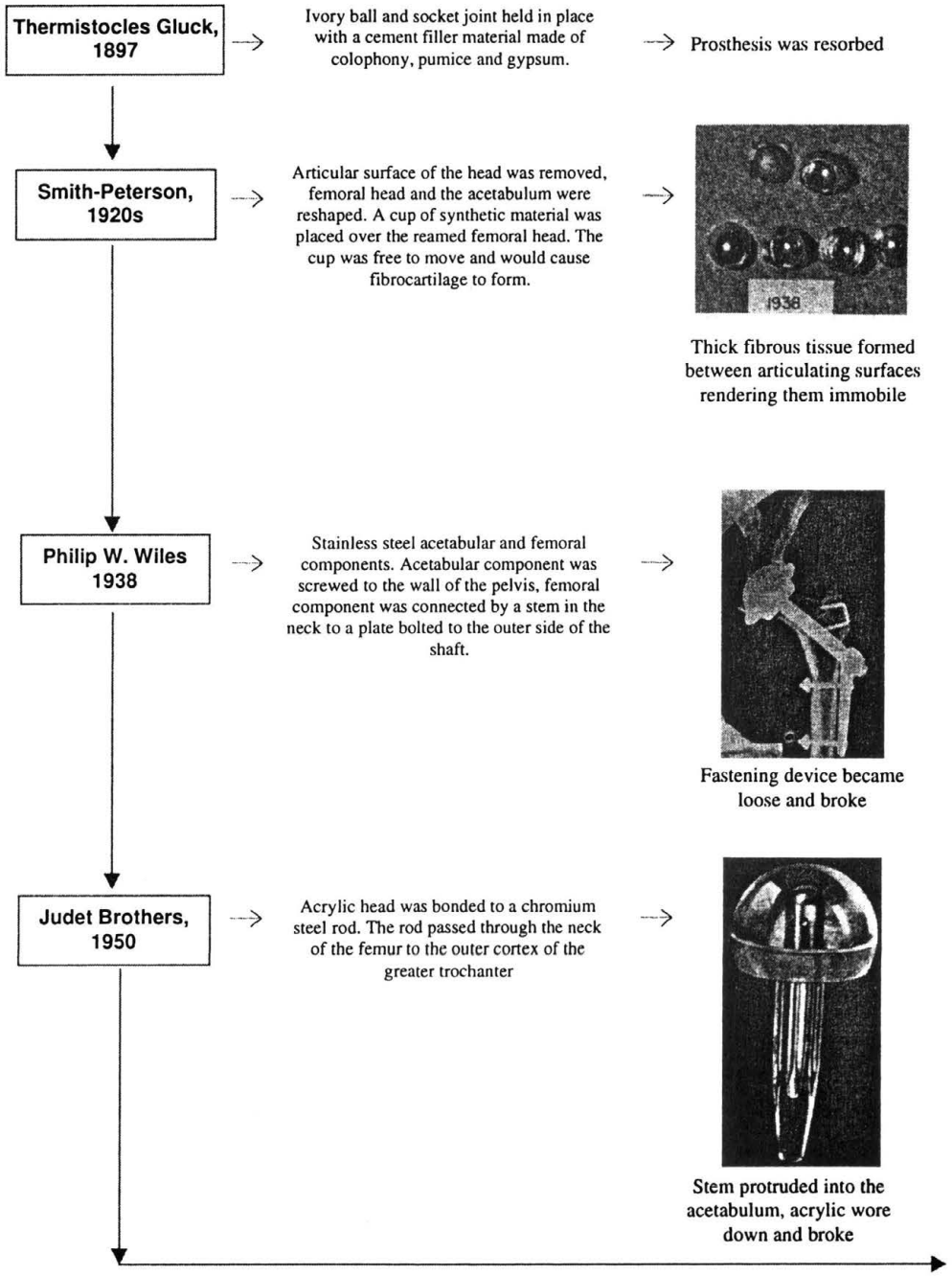


FIGURE 3.1. Historical chart describing the milestones in the development of total hip arthroplasty. Taken (with permission) from: S.A. Maher, *Design and Development of a Pre-clinical Test for Cemented Hip Replacements*, Ph.D. Thesis, University of Dublin, 2000.

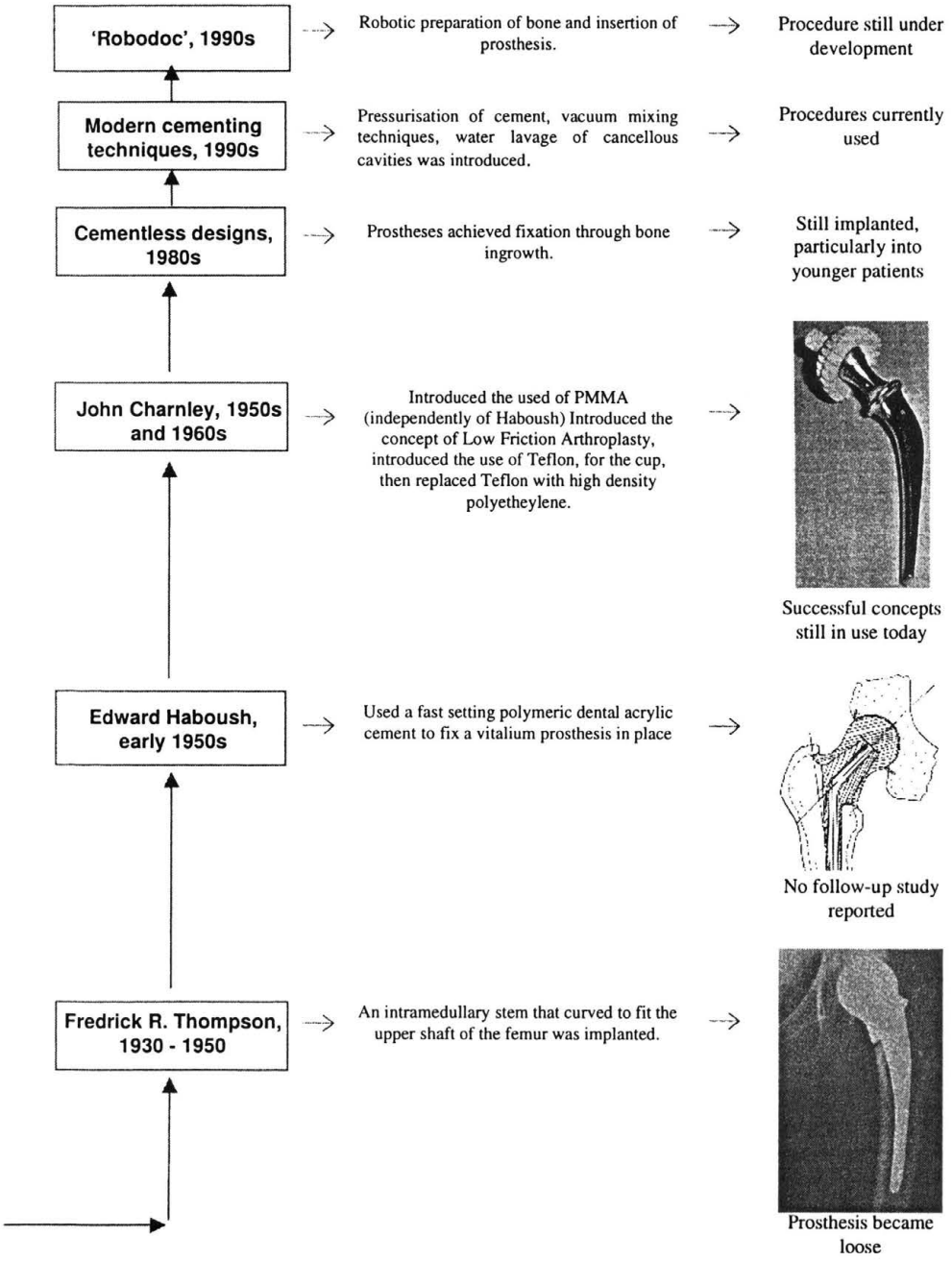


FIGURE 3.1. [cont.]

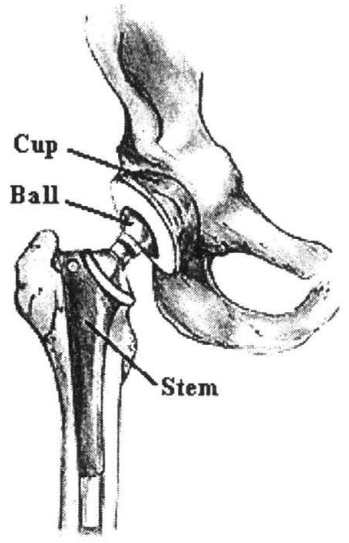


FIGURE 3.2. Total Hip Arthroplasty.

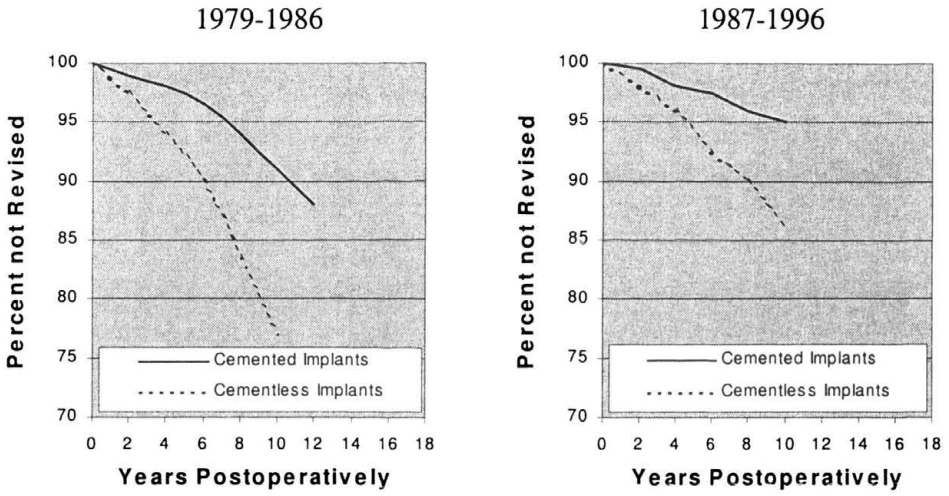


FIGURE 3.3. Survival curves for cemented and cementless implants for two time periods, taken from the Swedish hip register. Note how the survival of the cemented prostheses is improving significantly. Taken (with permission) from: B.P. Murphy, *Aspects of the Fatigue Behaviour of Acrylic Bone Cement*, Ph.D. Thesis, University of Dublin, 2001.

with the bone. There are a variety of surgical opinions on which fixation method is best – "to cement or not to cement, that is the question". Recent results from registry studies in Sweden suggest that cemented fixation has improved (with the advent of so-called third generation cementing techniques) to be superior to cementless fixation, see Fig. 3.3.

Other differences in design include the following: shape of the stem, presence of a collar, diameter of the ball, prosthesis material (e.g. chromium cobalt, titanium, stainless steel), ball material (metallic, ceramic), use of metal backing or not on the acetabular cup. There are many hundreds of designs of hip prosthesis on the market. Murray *et al.* [17] counted 62 separate designs on the British market alone in 1995. The differences in design are known to affect the outcome of the surgery, and different designs are continuously becoming available with varying amounts of pre-clinical testing having been performed. In a paper on *An Analysis of Theories in Biomechanics* [18], the present author argues that each new prosthesis is, or should be, considered as a hypothesis – the hypothesis being that the new implant is superior to one already in clinical use. The testability of this hypothesis is low because the pre-clinical tests applied to the problem cannot easily discriminate between good and bad prostheses.

3.2. Failure scenarios

When conducting a pre-clinical test on an implant of any sort, it is first necessary to determine the predominant failure mode of the implant. This can be done based on follow-up studies, retrieval studies, or by *in vivo* diagnostic imaging (radiographic assessment, DEXA scanning, MRI, etc.). Analysing the wealth of information on the failure of orthopaedic implants, Huiskes [19] proposed six failure scenarios. These failure scenarios combine the various observed failure phenomena into a sequence of events. The failure scenarios can proceed simultaneously and the one that causes failure is the one that ultimately becomes the precipitate cause of the need for a revision operation.

The six biomechanical failure scenarios identified are:

1. *The damage accumulation failure scenario*: Accumulation of damage in prosthesis materials occurs due to cyclic loading and creep. Proof of damage accumulation in bone cement of hip replacements was provided by Jasty *et al.* [20], who found partial cracking in autopsy-retrieved specimens. Experimental confirmation that the cracking process is gradual and continuous under bending loads [21] and torsional loads [22] has been pro-

vided. Damage may also accumulate within the prosthesis (leading to stem breakage) or on the interfaces [23, 24]. Verdonschot and Huiskes [25] carried out a computer simulation of damage accumulation in the cement mantle of a hip replacement and found that debonding very much accelerates the damage accumulation rate. Pre-coating of the stem with a PMMA layer to reduce interfacial porosity may slow the rate of damage accumulation.

2. *The particulate reaction failure scenario*: There are three possible sources of particles in joint replacement: wear of the articulating surfaces, abrasion of the PMMA/prosthesis/bone interfaces, and fretting between metal parts in modular prostheses. For example, polyethylene particles have been found at the cement/bone interface in acetabular cups where they initiate a macrophage inflammatory response and subsequent formation of a fibrous tissue layer between the implant and the bone. The next event in this failure scenario is increased interfacial micromotion and mechanical loosening.
3. *The failed ingrowth failure scenario*: Failure of osseointegration to occur can be caused by large unbridgeable gaps between prosthesis and bone, or by excessive micromotion.
4. *The stress shielding failure scenario*: The implant takes load formerly transferred to the bone, thereby shielding the bone from the load and causing bone resorption. This process is most probably dependent on mechanical factors, and has been observed in the proximal medial bone after hip replacement, and under the tibial component of knee replacements. Whilst more flexible components have been proposed to prevent stress-shielding in the hip, finite element models predict that too flexible a stem creates unsustainable bone/prosthesis or bone/cement interface stresses, which may lead to the failure scenario of damage accumulation (cemented implants) or failed ingrowth (cementless implants).
5. *The stress bypass failure scenario*: A prosthesis which is badly designed for a particular bone, or malsized, may transfer the loads in such a way as to by-pass part of the bone entirely. Stress bypass may also result from localised osseointegration in uncemented devices – such "spot welding" causes the load to bypass whatever bone tissue lies between the "weld" and the articulation.

6. *The destructive wear failure scenario*: Wear of the bearing surfaces may proceed to such an extent that the polyethylene component "wears out". For example, the head may penetrate the acetabular cup in a hip replacement.

Perhaps another scenario should be added to this one for completeness:

7. *Kinematic constraint failure scenario*: The prosthesis components can dislocate, or can give insufficient range of motion. This could be the result of excessive wear and it may lead to high stresses on the fixation leading to the damage accumulation failure scenario.

Chapter 4

Development of pre-clinical tests

4.1. General

The progression of the dominant failure scenario should be monitored in an ideal pre-clinical test. To be capable of "monitoring the progression" of failure, it is necessary that the parameter which is measured should vary continuously during the test. The elucidation of a suitable parameter to measure requires, as mentioned above, an extensive examination of follow-up studies, retrievals, and diagnostic images.

Consider the example of total hip arthroplasty. Failure (need for a revision operation[1)]) is caused by pain felt by the patient. Malchau and Herberts [26] and others have shown that, in most cases, it is loosening of the components (failure of the fixation) that causes pain. The main reason for fixation failure in cemented implants is that damage accumulates in the polymethylmethacrylate layer that fixates the stem into the medullary cavity of the femur – i.e. failure by the *damage accumulation failure scenario*. The problem with measuring that failure scenario is that it is difficult to measure damage accumulation in the cement mantle because it is not visible since it is encased in the bone (difficult, but not impossible if acoustic damage detection methods are used [27]). That is why, traditionally, the damage accumulation failure scenario is not measured.

Next failure can be often due to *particulate reaction*; this scenario can be simulated, along with the *destructive wear failure scenario*, in a wear test. The

[1)] Note that there are problems with using time-to-revision as a failure point because different definitions of when to revise are possible. Furthermore a revision may not be done for some patients, even if it is necessary.

size and number of wear particles can be counted and the wear rate can be measured for different head/cup materials and configurations. These tests are commonly done.

The most common failure scenario measured experimentally is the *stress shielding failure scenario*. Since the earliest designs, the strain in the femur has been measured with strain gauges [28] and recently these tests have become more sophisticated using complex physiological loading apparatus [29] and optical measurement methods [9]. However this failure scenario is not commonly the

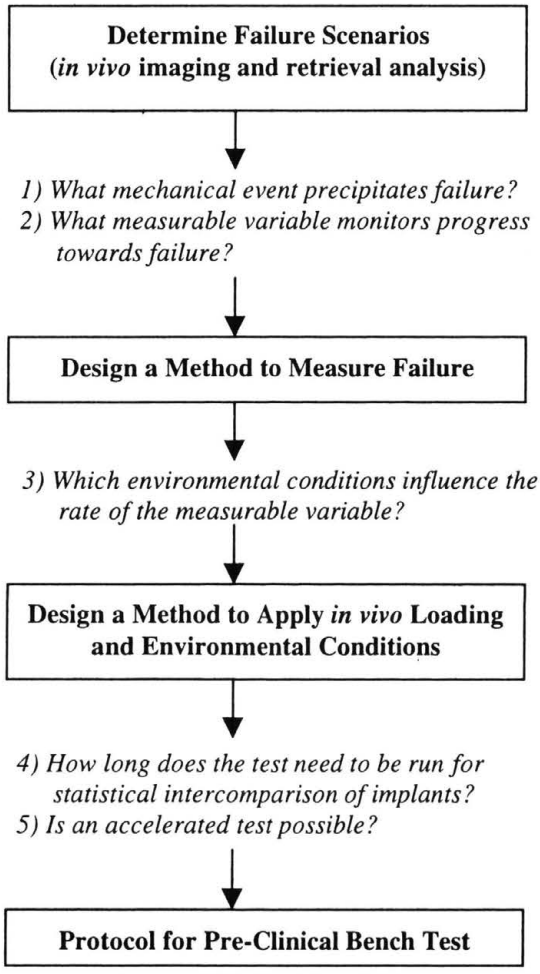


FIGURE 4.1. Creating an experimental pre-clinical test.

cause of failure, although it may have an influence in accelerating the damage accumulation failure scenario.

When the parameter to measure the required failure scenario has been selected, a method to measure it needs to be designed. Figure 4.1 shows an algorithmic illustration of the further steps to be followed to establish the pre-clinical test.

4.2. Experimental vs. computational (numerical) tests

The relative value of experimental tests using physical models versus computational tests using finite element models is discussed in this section. The advantage of the experimental model is that a test on the real implant can be carried out. This lends credibility to the test for satisfying regulatory authorities (and enhances the usefulness of the test if approval for animal or clinical testing is to be sought). However, from the scientific point of view, both types of test have their strengths as follows:

- The complex loading conditions observed *in vivo* can be more readily applied in computational tests – see Stolk *et al.* [30] for the case of loading of the proximal femur. As more complex loading datasets become available for various musculoskeletal structures, it is certain that this advantage will make computational models more attractive in certain applications because complicated apparatus is required to apply muscle loading experimentally, see, for example, Britton *et al.* [31].
- Adequate representation of biological tissue can be difficult in experimental models for many reasons, including the difficulty of obtaining and preserving human tissues. Furthermore tissues dry out; for example bones become brittle and change their elastic and failure properties.
- Human musculoskeletal structures show significant anthropometric variability. This introduces variability into the experimental test if cadaveric material is to be used. This can be overcome using bone analogues, such as the composite femur, see Cristofolini *et al.* [32].
- Experimental models can, at the present time, give a more valid description of the post-operative behaviour of the interfaces between implant and tissue. Developments are being made to better model the interface, including numerical methods allowing maintenance of tensile forces across the interface due to adhesion, see Rojek and Telega [33].

- Computational models offer the possibility of simulating the adaptive behaviour of the tissues surrounding the implant. For orthopaedic implants, this refers either to bone remodelling which occurs as bone loss in the proximal region or bone formation in the region near the prosthesis tip, or soft-tissue formation at the interface between implant and bone caused by the relative motion of implant and bone. For both of these behaviours, pre-clinical simulations require *mechanobiological* models that relate the mechanical environment in the tissue to biological changes of the tissue – such models are described in the next section.

  It might be worth noting that such mechanobiological phenomena also occur in non-orthopaedic situations: for example in cardiovascular tissues where a stent is used to hold open a stenosed (occluded) blood vessel. In such cases the stress induced by the stent in the vessel wall creates a reaction causing the vessel to remodel [34]. Another example of tissue adaptation to an implant is that when a grommet is placed in the tympanic membrane (ear drum) a process of tympanosclerosis occurs (calcification of the drum) [35].

4.3. Mechanobiological phenomena and pre-clinical testing

4.3.1. Bone remodelling

Adaptation of bone to a change in loading is known as Wolff's law. Mathematical models have been developed to describe it, see the papers by Telega and Lekszycki [36] and Lekszycki and Telega [37] for a recent review. As regards the application of these models to simulation of peri-prosthetic bone remodelling, the first simulation was presented by Huiskes *et al.* [38] for a simple axisymmetric finite element model representing a generalised intramedullary fixation. This work showed that, in principle, design features could be related to change in bone remodelling around an implant. Later simulations were carried out on three-dimensional finite element models of hip and knee reconstructions, see Prendergast [39]. These models made the following predictions:

1. Lower Young's modulus hip prosthesis stems increase the stress in the proximal bone which reduces, but not altogether prevents, proximal bone loss in the femur. However, lower Young's modulus prostheses also generate higher interface shear stresses indicating that the propensity for interfacial failure would be higher with such stems. Therefore there is a design conflict with respect to Young's modulus of a hip prosthesis stem.

2. The "fit" of the implant within the medullary cavity affects the long term remodelling. Proximal over-reaming will cause a stress-bypass leading to, perhaps, massive proximal bone loss whereas no such effect is predicted with distal over-reaming [40].
3. Partial areas of osseointegration caused by partially bonded prostheses can succeed in balancing, to some degree, the trade-off required for (1.) above.

The models employed by Huiskes and co-workers to simulate peri-prosthetic remodelling employ strain energy density as a stimulus and predict the remodelling process. Other stimuli, such as accumulated damage [41] or "effective strain' [42] may be used, but they give similar predictions for the particular application of remodelling around a hip implant.

4.3.2. Tissue differentiation at interfaces

The formation of a layer of soft tissue between an orthopaedic implant and the bone can cause implant loosening and the need for a revision operation. The soft tissue layer can manifest itself as a radiolucent line on a radiograph, and can be a cause/consequence of the failed ingrowth failure scenario in a cementless prosthesis – in such a prosthesis bone is expected to grow into the surface of the implant and fill whatever gaps there are between the implant and the bone. In many cases the implant is coated with a layer of beads (Fig. 4.2) or with a wire mesh to facilitate the development of a strong interface by osseointegration.

Soft tissue layers are a result of excessive micromotion between the implant and the bone – a micromotion threshold of 150 μm is often cited. An interesting

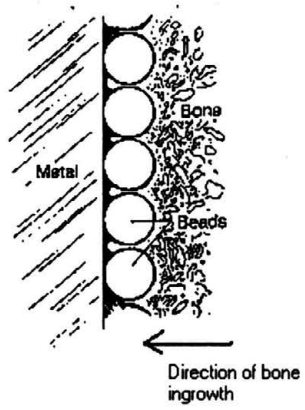


FIGURE 4.2. Osseointegration at a bead-coated implant surface.

study of gap healing around an implant under various levels of micromotion (and surface coating) was reported by Søballe [43] in a now classical experiment. He placed an implant into the condyle of dogs and subjected the implant to a displacement of either 0 μm, 150 μm or 500 μm. Depending on the coating and the degree of motion, different tissues were found in the gap region surrounding the implant, see Fig. 4.3. When the dog loads the knee, the polypropylene (PP) is pressed as the tibia and femur come together. This causes the implant (I) located on the piston (P) to move inwards thereby shearing the *gap tissue* (black region). When the implant is unloaded, the spring (S) pushes the implant out again.

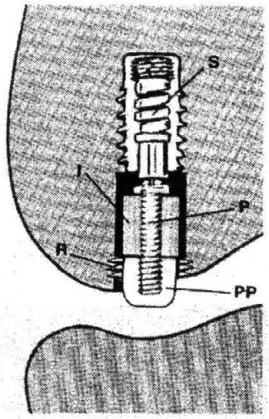


FIGURE 4.3. An implant in the knee condyle of dogs. (After Søballe [43]).

Tissues were found surrounding the implant: fibrous tissue, fibrocartilage, cartilage, and bone, see Fig. 4.4 below. Prendergast *et al.* [44, 45] analysed the change of stimuli in the gap tissue as tissue differentiation progressed and concluded that shear strain and fluid flow could combine to regulate tissue differentiation. This experiment indicates that a mechanobiological process of tissue differentiation in response to mechanical stimuli is ongoing around an implant. It was shown that the process of soft-tissue formation around an implant could be simulated using this concept [46] which opens the possibility for using such algorithms in pre-clinical tests.

Other algorithms which can predict tissue formation at interfaces have been presented; these are reviewed by Prendergast and van der Meulen [4]. Investigating the mechanics of endochondral ossification, Carter and colleagues [47] proposed that intermittent or cyclic mechanical loading occurring over a period of time (load history) stimulates tissue differentiation. The loading history was

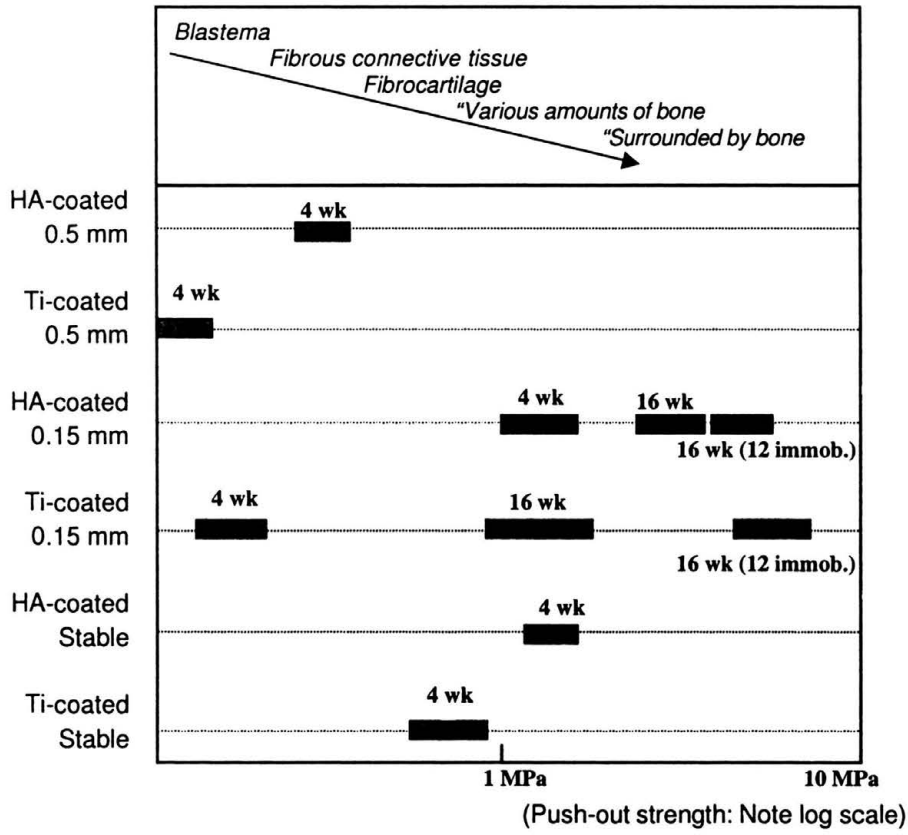


FIGURE 4.4. A graphical summary of the results of Søballe's experiment. The experiment shows that larger micromotions inhibits the process of interfacial ossification.

decomposed into discrete loading conditions, denoted c. Using the concepts proposed by Pauwels [48], the stress acting on the regenerating tissue was described as a combination of two scalar quantities:

- Hydrostatic stress (related to the dilatational strain) denoted D_i and
- Octahedral shear stress also called the deviatoric stress (related to the distortional strain) denoted S_i where the subscript denotes the i^{th} load case, and $i = 1, 2, 3, \ldots c$.

Carter and colleagues proposed that cyclic octahedral shear stress encourages cartilage ossification whereas the action of a cyclic hydrostatic stress inhibits ossification. The driving force for bone formation may be described by a linear combination of the stress invariants. This is termed the Osteogenic Index (OI) given as

$$OI = \sum_{i=1}^{c} n_i(S_i + kD_i) \tag{4.1}$$

where n_i is the number of loading cycles for the i^{th} load case. The value of k, the empirical constant, is determined by parametric variation in computer models of fracture healing [47, 49], joint formation [50], and endochondral ossification of the sternum [51]. High values of OI are caused by high shear stress or by tensile hydrostatic stresses – therefore, according to the osteogenic index theory, these stimuli favour endochondral bone formation. Bone formation is inhibited in the presence of large compressive hydrostatic stresses. This model has also been used to study tissue formation around implants, see Simmons and Pilliar [52].

In a finite element analysis of the mechanical stimuli on ossifying surfaces during fracture healing, Claes and colleagues [53, 54] hypothesized that magnitudes

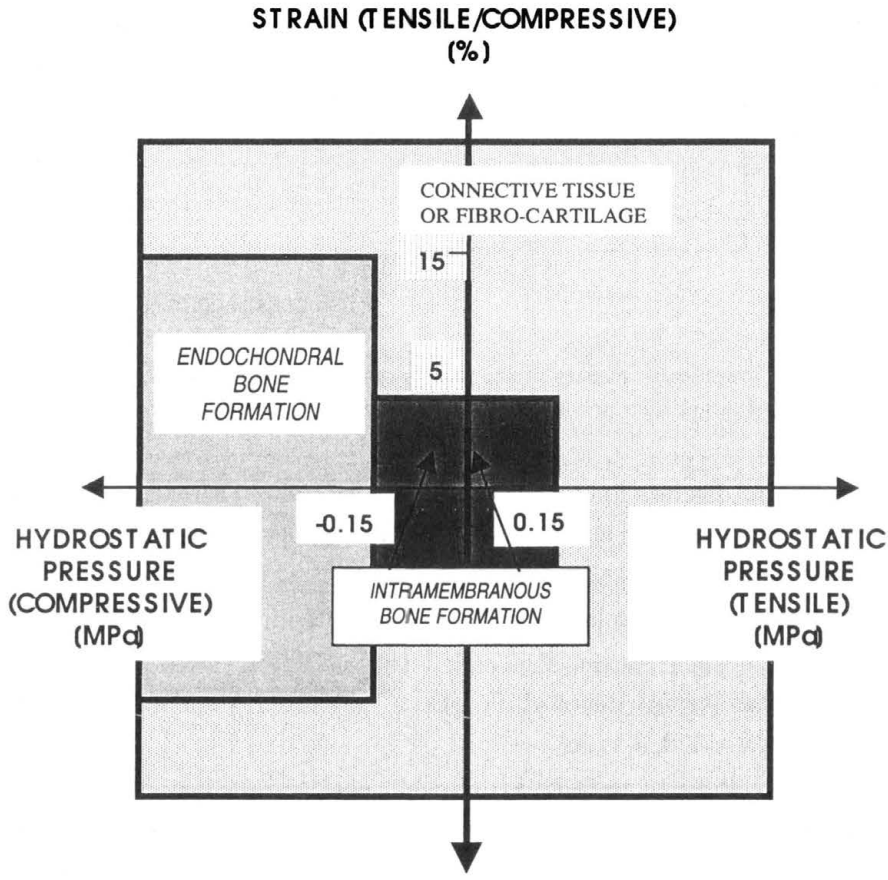


FIGURE 4.5.

of hydrostatic pressure and strain regulate the selection of either intramembranous or endochondral bone formation processes. By quantitative analysis of a real callus geometry, they found that if the compressive hydrostatic pressure (negative) exceeded 0.15 MPa at the regenerating bone surface then endochondral bone formation (i.e. prior formation of cartilage) occurred, whereas if the hydrostatic pressure was below this threshold then intramembranous bone formation proceeded, see Fig. 4.5.

Chapter 5

Example of an experimental test – the hip prosthesis implant[1)]

The objective of this section is to show how the methodology presented in Section 4.1 above can be applied in the case of the hip prosthesis. To do this we rely on the clinical observation (made using stereophotogrammetric methods [55] and enhanced radiographic techniques [56, 57]) that hip prosthesis *subsidence* exceeding two millimeters after two years correlates with implant loosening. This suggests that, if prosthesis migration could be measured during a test that emulates two years of use, a preclinical assessment about the risk of loosening can be made.

Using this variable to monitor the progression towards loosening requires a technical system to measure bone/prosthesis micromotion throughout several million cycles of loading. The migration measurement device was based on the concept employed by Berzins *et al.* [58] to measure the motion of cementless prostheses for a small number of loading cycles. The design developed in our laboratory comprised of four main components:

- a *target device*,

[1)] Published in three papers as follows: S.A. Maher, P.J. Prendergast & C.G. Lyons, *Measurement of the migration of a cemented hip prosthesis in an in vitro test*, Clinical Biomechanics, Vol.16, pp.307-314, 2001; S.A. Maher, P.J. Prendergast, A.J. Reid, D.V. Waide & A. Toni, *Design and validation of a machine for reproducible precision insertion of femoral prostheses for preclinical testing*, Journal of Biomechanical Engineering, Vol.122, pp.203-207, 2000; and S.A. Maher & P.J. Prendergast, *Discriminating cemented hip prostheses based on migration and inducible displacement*, Journal of Biomechanics (in press).

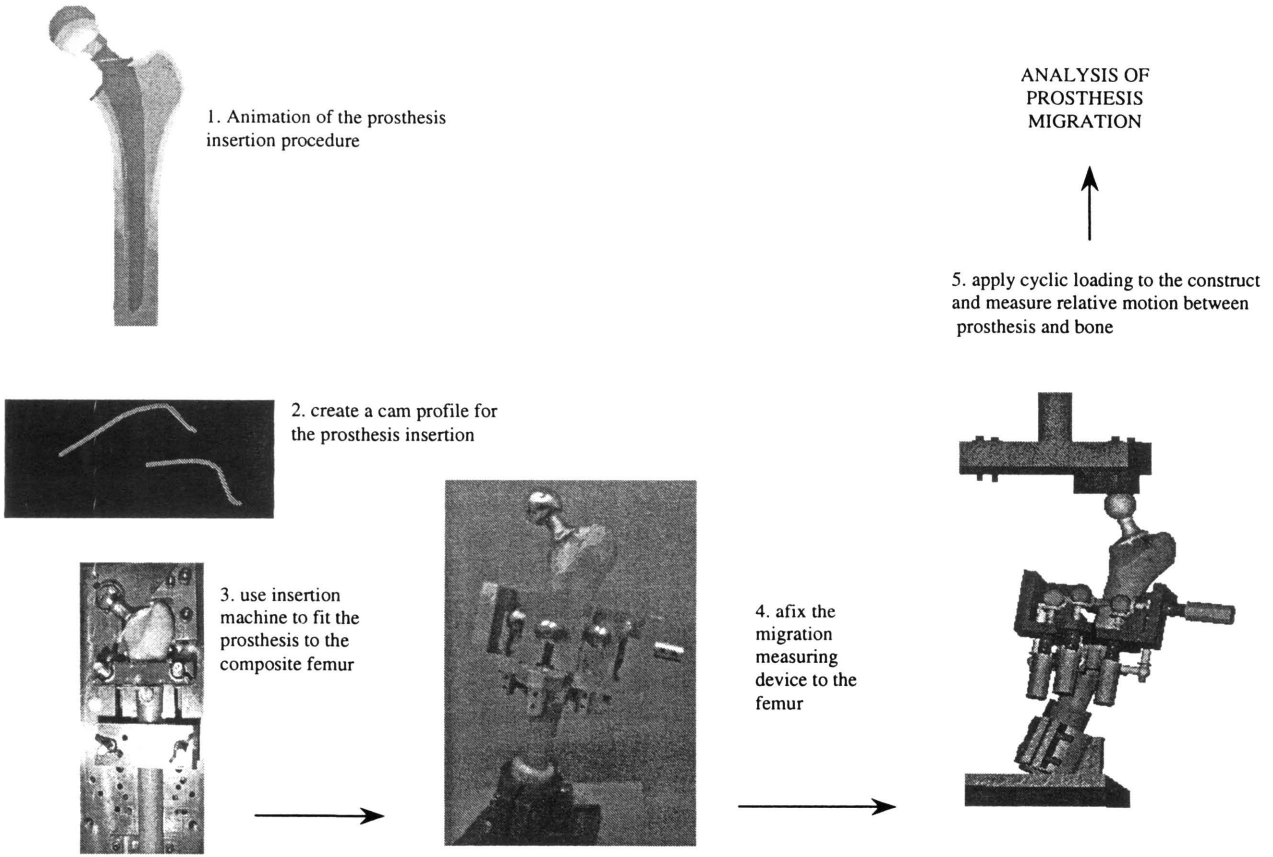


FIGURE 5.1. The summary of the protocol for the experimental pre-clinical test to discriminate hip prostheses designs. Taken (with permission) from: S.A. Maher, *Design and Development of a Pre-clinical Test for Cemented Hip Replacements*, Ph.D. Thesis, University of Dublin, 2000.

- an *LVDT holder* with six LVDTs (Linear Variable Displacement Transducers),
- a *femoral ring*, and
- *adjustable linkers* that attached the LVDT holder to the femoral ring.

A protocol for assembly of the micromotion measurement device was required to ensure exact alignment between the LVDTs and the prosthesis, and to ensure no elastic deformations were *locked-in* to the components during assembly. The six LVDTs measurements could be used to determine the relative motion between the prosthesis and the implant. Two types of relative motion occur; the first is called *inducible displacement* and it is the recoverable elastic displacement of the prosthesis relative to the bone whereas the second is called *migration* and is the permanent subsidence of the prosthesis due to irreversible effects, such as cement creep or interface slippage.

The complete protocol for the test is shown in Fig. 5.1. The first step in this procedure was the preparation of the implanted composite femur. An animation of the insertion procedure is first carried out (Step 1, Fig. 5.1). This animation is used to design a cam which physically guides the insertion of the prosthesis in a purpose-designed prosthesis insertion machine [59] – see Step 3, Fig. 5.1. Next the migration measurement device, described above, is attached to the composite femur (Step 4, Fig. 5.1). The whole construct is then subjected to cyclic loading and the migrations and inducible displacements are measured. To test the protocol, it was decided to attempt to discriminate two prostheses: the Lubinus SPII prosthesis (Fig. 5.2) and the Müller Curved Stem (MCS) prosthesis (Fig. 5.3).

The procedure used to measure the subsidence of the prosthetic stem is described in detail in Maher *et al.* [60]. When the migrations are measured they may be plotted for each of the six degrees of freedom (three translations and three rotations). Results comparing the two prosthesis over two million cycles are shown in Figure 5.4 for five of each prosthesis design. It can be seen first of all that there is great variation of prosthesis subsidence rates. However, on average the Müller prosthesis subsides at a faster rate than the Lubinus. Figure 5.4 shows the permanent migrations, but the experiment also shows that there are different changes in the inducible displacements over time. In the case of the majority of the Lubinus SPII prostheses the inducible displacement reduced as the test progressed whereas in the majority of the Müller prostheses the inducible displacement increased over time. Maher and Prendergast [61] show that there

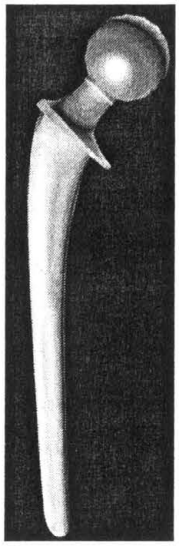


FIGURE 5.2. The Lubinus SPII hip prosthesis.

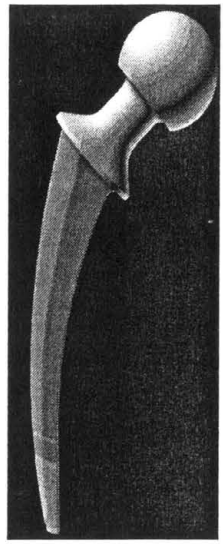


FIGURE 5.3. The Müller hip prosthesis.

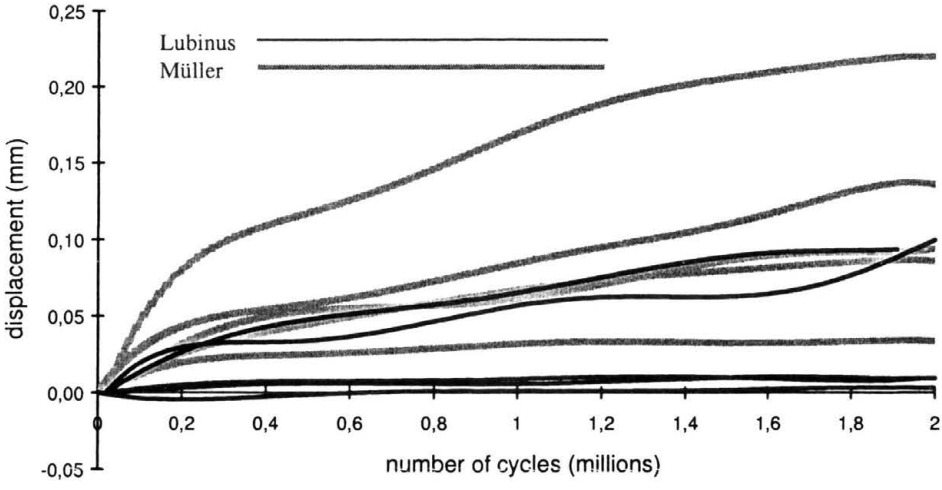


FIGURE 5.4. Migration in the distal direction (subsidence) of 5 Lubinus SPII prostheses and 5 Müller prostheses. Results for the other degrees-of-freedom are shown in Maher and Prendergast [61].

is a correlation between inducible displacement and migration. Further tests on a range of prostheses in the Swedish hip register are to be carried out to verify whether or not this preparation and testing protocol can rank the performance of various prosthesis designs according to their observed revision rate. One issue to be addressed is that polished stems which facilitate stem/cement debonding (e.g. the Exeter prosthesis [62]) have increased subsidence compared to stems designed to maintain cement/stem bonding [55], but nonetheless perform well in the Swedish hip register [15]. Although absolute migration is a good indicator of failure for stems designed to stay bonded with the cement, we feel that the change in inducible displacement may be more broadly applicable to include both designs that aim to maintain the bond between the cement and those designs which facilitate de-bonding from the cement – called shape-closed and force-closed designs respectively by Huiskes *et al.* [63]. Although potentially more difficult to measure clinically, the change in inducible displacement may be a more design-independent measure of prosthesis loosening than absolute migration.

Chapter 6

Examples of computational pre-clinical tests

6.1. Orthopaedic implants

As described in Section 4.2 above, computational tests have certain advantages over experimental tests; among these are a more accurate representation of *in vivo* loading and a more consistent representation of the geometry, and the possibility of investigating tissue adaptations using mechanobiological models. The major disadvantage is that constitutive models for the tissues and for the tissue implant interfaces are not yet fully developed; parametric analysis can overcome this deficiency but, as one authority puts it "to survey parameters forever is very tiring, and cannot come to grips with real problems' [64].

One of the most common uses of finite element analysis has been to analyse the stress distribution in cemented orthopaedic implants [39]. In the first part of this section, I will present some of our recent research regarding the testing of cemented hip and shoulder prostheses.

One of the problems in the analysis of cemented fixation is that great variability is found in experimental tests of polymethylmethacrylate bone cement. Our studies of the fatigue strength of polymethylacrylate bone cement [65, 66] have shown that the variability is related to porosity and the number of shrinkage cracks. These entities, which are essentially random, determine the rate of the *damage accumulation failure scenario*. Recognising this variability as a clinical attribute of orthopaedic bone cement, we have carried out finite element analyses for both total hip arthroplasty and total shoulder arthroplasty using the probability-of-survival as a function of stress in where the probability-of-survival is denoted P_s and stress σ,

$$P_s = A\sigma^3 + B\sigma^2 + C\sigma + D, \tag{6.1}$$

where the co-efficients are given in Table 6.1 below for both the hand-mixed and the vacuum-mixed bone cements. (Hand mixing occurs when the monomer and powder of the PMMA are mixed together by hand in a mixing bowl whereas with vacuum mixing the cement is mixed in with air pumped out).

TABLE 6.1. The co-efficients in the relationship between probability-of-survival P_s and stress σ.

	A	B	C	D
Hand-mixed	-0.0005	+0.0202	-0.3304	1.8365
Vacuum-mixed	0.003	-0.1154	1.3427	-3.9564

Equation (4.1) allows the probability of survival of the element of the cement to be determined as a function of the cement preparation method. In this way the stress calculated in a finite element model can be reduced to one easily-understood variable – this should help facilitate precise inter-comparison of implants, as shown in the two examples below.

6.1.1. Femoral side replacement in the hip

As described in Section 3.1 above, hip prostheses come in two broad classes: cemented and uncemented. There is a further division within the class of cemented prostheses: those prosthesis which have a matt surface and those which have a polished surface. The prostheses with a polished surface are expected to debond from the cement immediately post-operatively and subsequently to subside so as to tighten within the cement mantle.

In what follows, a paper by Lennon and Prendergast [67] will be summarized – in that paper we report an analysis of a polished cemented femoral prosthesis under both bonded and debonded conditions. The debonded case was also compared to the case with the cement removed beneath the prosthesis tip. The stress in each element of the finite element model was used to compute the probability-of-survival (Eq. 4.1) and hence the probability-of-failure for that element. In this way the probability-of-survival of the cement in the mantle can be plotted and the regions where the probability-of-survival is low can be identified; these are the proximo-medial region and the cement distal to the prosthesis tip – see Fig.6.1.

Since the volume of the element can be calculated, a plot of the probability-of-failure against the percentage of the cement mantle having that probability-of-

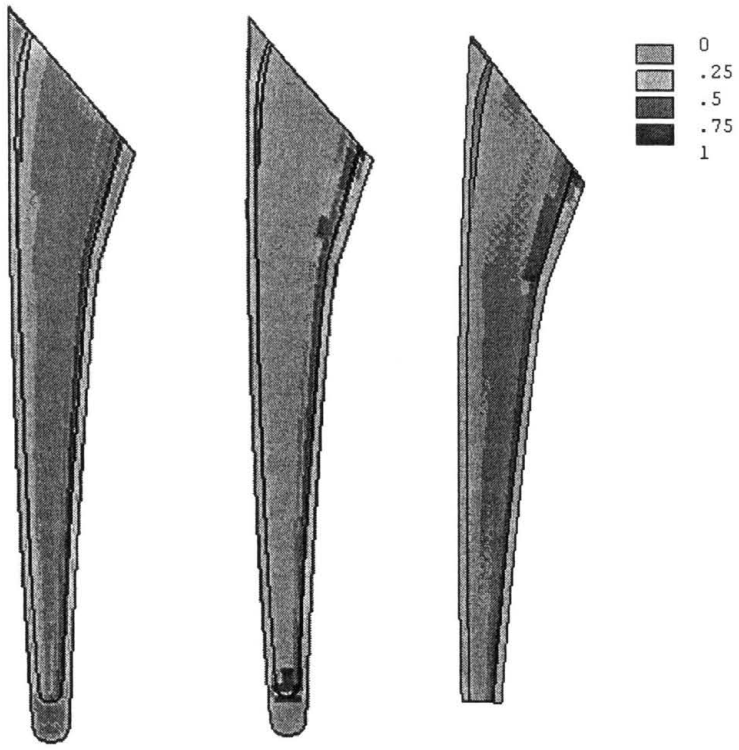


FIGURE 6.1. Probability-of-failure of cement in (a) a bonded prosthesis, (b) a debonded prosthesis, and (c) a debonded prosthesis with the distal cement removed. Courtesy of A.B. Lennon, Trinity College, Dublin.

failure, or a lower one, can be generated. For example, referring to Fig. 6.2, 99% of the mantle around the bonded stem has a probability-of-failure of less than 0.1 and none of the cement has a probability-of-failure of greater than 0.4. On the other hand, when debonding occurs, a small amount of the cement volume has a large probability-of-failure.

6.1.2. Glenoid replacement prostheses in the shoulder

The majority of total shoulder replacements are performed to relieve pain caused by rheumatoid arthritis (RA). This disease causes extensive changes in bone shape and material properties of the scapula. A total shoulder replacement is done by replacing the humeral head with a humeral component fixated into

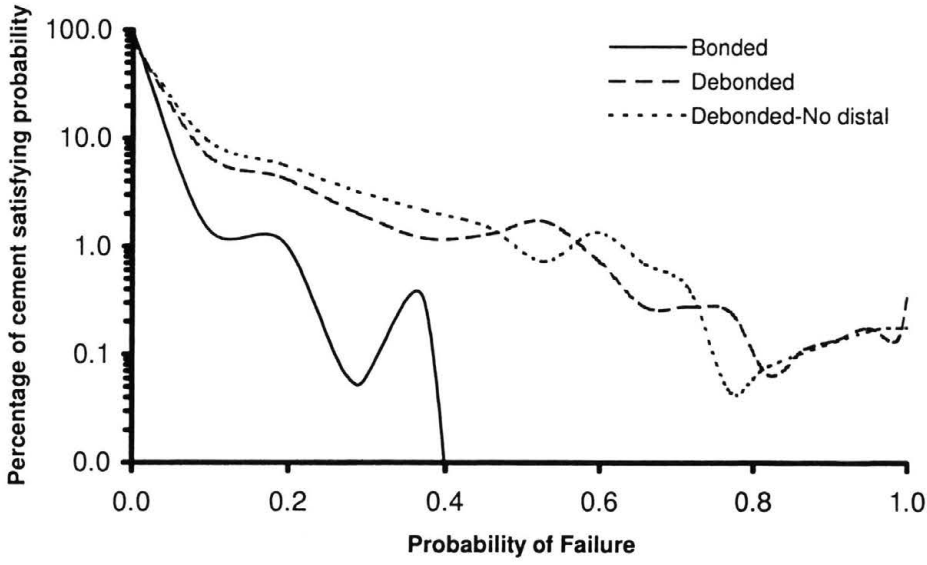


FIGURE 6.2. Percentage volume of cement satisfying probability-of-survival at 10 million cycles ($P_F = 1$ predicts failure within 10 million cycles and $P_F = 0$ predicts survival). From Lennon and Prendergast [67].

the medullary cavity of the humerus and a glenoid component fixated into the scapula. It is this latter component that loosens, presumably because the cement there is overstressed. Since the stress on the cement/bone interface, and also the *strength* of this interface, depends on the material properties of the underlying bone, we hypothesise that different prosthesis designs should be used for different levels of bone degeneration. If this were true then bone densitometry could be used to select the best prosthesis as part of a pre-operative planning procedure.

Murphy *et al.* [68] present a review of nine two-dimensional and three-dimensional finite element analyses of glenoid replacement components. These studies show that the different design concepts create different stresses in the fixation, but that bone stresses are not so much affected by implant design. Gupta *et al.* [69] present an analysis of both cemented and uncemented designs and conclude that, because cemented prostheses have high cement fixation stresses, cementless prostheses have better possibilities than cemented prosthesis. Only three studies, the first by Dalstra *et al.* [70], the second by Lacroix *et al.* [71], and the third by Murphy *et al.* [68] analyse the consequences of reduced bone quality for prosthesis fixation. In this research, the stresses in the cement layers

around three prosthesis types are compared, for both normal bone and RA bone. These are:

- centre-keel glenoid prosthesis,
- anterior offset-keel prosthesis,
- a pegged prosthesis.

A three dimensional model of the scapula was generated using CT-data for geometric and material property definition [72], see Fig. 6.3. For the comparative analysis of prostheses under abduction and flexion loading, only the lateral segment of the scapula was modelled; i.e. the part of the scapula medial to a plane intersecting the scapular notch and parallel to the glenoid surface. The medial border of the model was restrained as described by Murphy *et al.* [68]. Models of a centre-keeled prosthesis, an anterior offset-keeled prosthesis, and a pegged prosthesis were inserted into the glenoid, each surrounded by a 1 mm layer of bone cement. The prostheses are shown in Fig. 6.4. Bone Young's moduli in the glenoid vault were reduced to simulate bone destruction due to rheumatoid arthritis (RA): by 50% for the cortical bone and by 90% for the cancellous bone, according to the data for Larsen Grade IV rheumatoid arthritis as determined experimentally by Frich [73]. The prostheses were then subjected to a variable

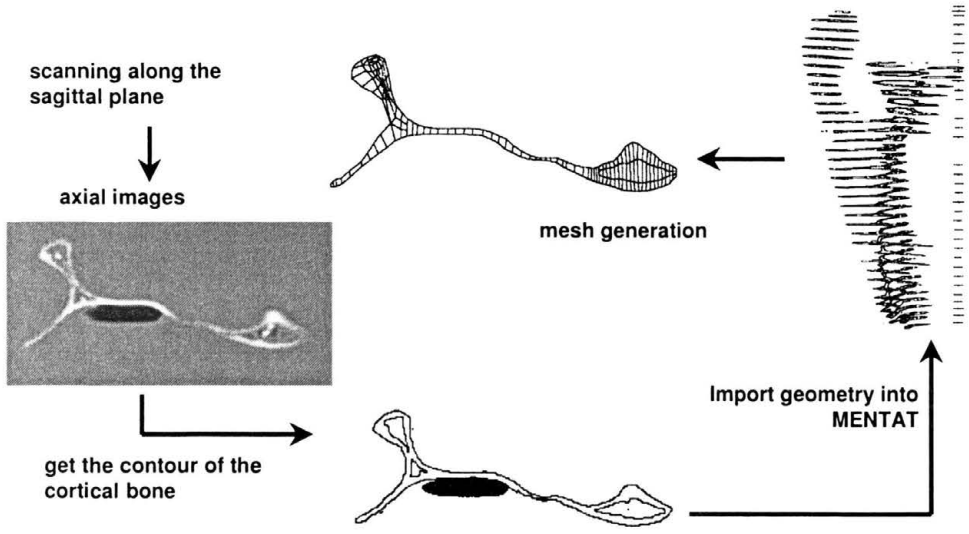
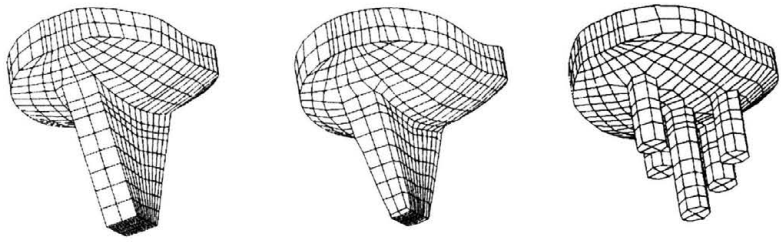


FIGURE 6.3. Using CT scans to generate a geometric model of a scapula bone. From: D. Lacroix, *Finite element analysis of the scapula and design criteria for glenoid prostheses*, M.Sc. Thesis, University of Dublin, 1997.



(a) centre keel conventional component (b) offset keel design (c) pegged design

FIGURE 6.4. Finite element models of three types of glenoid component for shoulder arthroplasty.

joint load (taken from van der Helm [74]) to compare the resulting durability. These flexion and abduction loads were applied with a parabolically distributed load giving precise loads at the various arm positions, see Murphy *et al.* [68]. Resulting stresses in the polyethylene prosthesis, cement layer and glenoid bone were calculated for each load case in healthy and RA bone.

What these results show is that the pegged prosthesis has the lower cement stresses in normal bone whereas the keeled prostheses have the lowest cement

TABLE 6.2. Stresses in the cement mantle around a glenoid component. From Prendergast and Murphy [75].

90 DEGREES *ABDUCTION* LOADING						
	Normal healthy bone			Rheumatoid arthritic bone		
Stress range	Centre	Offset	5-pegged	Centre	Offset	5-pegged
0 to 1 MPa	5%	63%	72%	65%	82%	50%
1 to 2 MPa	15%	22%	15%	25%	18%	22%
2 to 3.3 MPa*	20%	10%	7%	10%	0%	14%
above 3.3 MPa*	60%	5%	6%	0%	0%	14%
90 DEGREES *FLEXION* LOADING						
	Normal healthy bone			Rheumatoid arthritic bone		
Stress range	Centre	Offset	5-pegged	Centre	Offset	5-pegged
0 to 1 MPa	35%	40%	–	44%	100%	–
1 to 2 MPa	40%	45%	-	39%	0%	–
2 to 3.3 MPa*	19%	13%	-	12%	0%	–
above 3.3 MPa*	6%	2%	-	5%	0%	–

* The value of 3.3 MPa is used because there is a 95% probability-of-survival to 1 m cycles [75].

stresses in RA bone. This is because the support offered in RA bone is low and the pegs are pushed into the cement causing high stresses. With the keeled prostheses, much of the week bone is removed when making place for the cemented keel and the stresses that result are much lower because the remaining cancellous bone can offer more support. The offset keel performs better than the centre keel in abduction because the keel is offset in the posterior direction so that it is more directly under the load in abduction meaning that there is less bending stress in the cement. One might expect the reverse to be the case in flexion; however the offset keel does not perform appreciably differently under flexion – Murphy *et al.* [68] propose that this is because there is stronger bone in that part of the glenoid.

6.2. Middle-ear prostheses

The three smallest bones of the human body are the malleus, incus, and stapes. These bones connect the tympanic membrane (ear drum) to the oval window (which separates the middle ear from the inner ear) thereby creating the mechanical linkage to transfer the acoustic vibrations of the air to the fluids

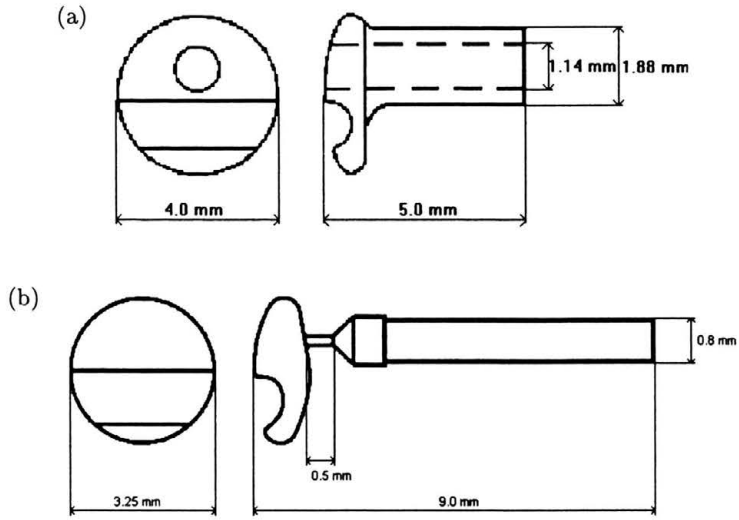


FIGURE 6.5. XoMed (Jacksonville, FL, USA) (a) partial ossicular replacement prosthesis and (b) total ossicular replacement prosthesis; the 0.5 mm titanium link connects the head of the prosthesis to the shaft. The groove on the head of the prosthesis fits onto the manubrium of the malleus.

of the inner ear. If these bones degenerate, or are not present due to congenital abnormality, then a person will not be able to hear normally. Prostheses have been designed to replace these bones if they degenerate. The prosthesis for middle-ear reconstruction are of two classes:

- partial ossicular replacement prostheses (PORPs) which connect the tympanic membrane to the head of the stapes (Fig. 6.5a),
- total ossicular replacement prostheses (TORPs) which connect the tympanic membrane to the footplate of the stapes (Fig. 6.5b),

– any anatomy textbook, such as *Gray's Anatomy*, will give a precise description of the anatomy of the ear where these terms are explained.

To test the differences between these two design concepts of middle-ear prostheses, we generated a finite element model of the outer ear canal and the middle ear. The mesh of the model is shown in Fig. 6.6 – the details of the material properties and the modelling of the damping are to be found in Ferris and Prendergast [76] and Kelly [77]; suffice it to say here that MRI imaging was used to determine the geometry of the ear canal and microCT scanning (courtesy of the Institute of Biomedical Engineering, University of Zürich Switzerland) was used to determine the geometry of the ossicles. The ligaments and muscles of

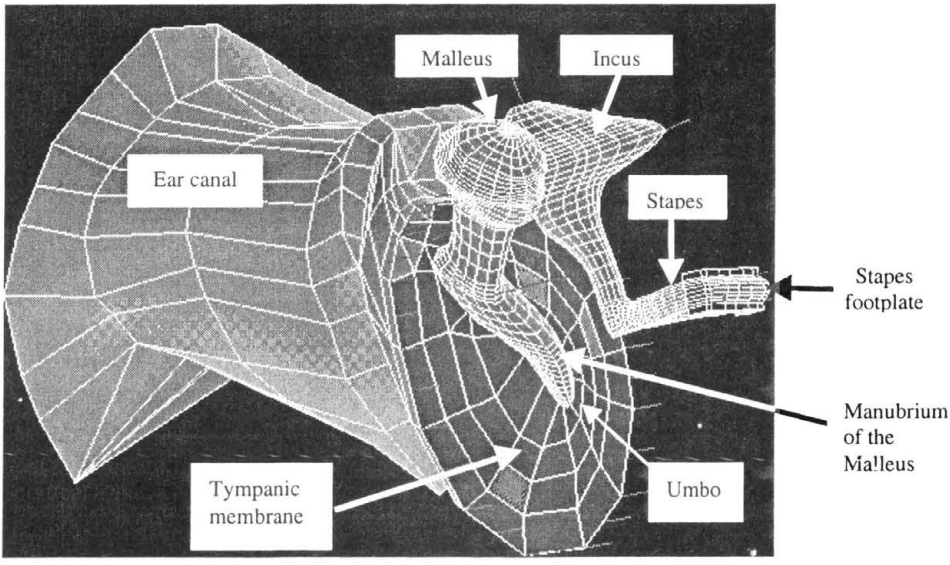


FIGURE 6.6. A finite element model of the outer and middle-ear. MicroCT scanning was used to generate the mesh of the three ossicle bones.

the middle-ear were also included, as shown in Fig. 6.6. It should be noted that work similar to ours has been carried out first by Wada and colleagues [78].

In the past the biocompatibility of the prosthesis materials received most attention, along with, from the mechanical design point-of-view, design for ease-of-insertion (handleability) during the operation. Since these objectives have been more or less achieved, attention has begun to focus on bio-functional aspects of implant design. Biofunctional criteria are:

- reconstruction of the lever ratio [79] (the lever ratio is hypothesised to act due to the fact that the distance from the umbo to the centre of rotation of the malleus/incus is greater than the distance from the centre of rotation to the head of the stapes),
- obtaining the same impedance for the normal and reconstructed ears [77].

Reports of results are in the form of the amplitude of vibration of the stapes footplate as a function of frequency. As close a correspondence as possible between the healthy ears and the reconstructed ears is the objective. What we find is that the XoMED PORP transfers more vibration to the inner ear across a broad range of frequencies than the TORP because it is stiffer. The TORP

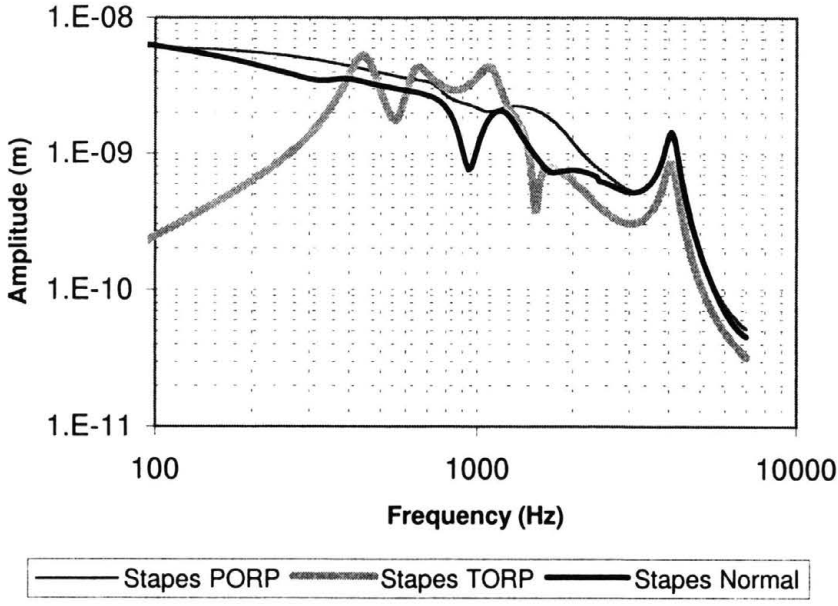


FIGURE 6.7. Amplitude of vibration at the stapes footplate as a function of frequency, for the normal ear and the ear reconstructed with partial and total ossicular replacement prostheses.

furthermore brings various natural frequencies into the hearing range because it is not so stiff. These results are shown in Fig. 6.7, and are taken from Ferris and Prendergast [76] – that paper should be consulted for further discussion of pre-clinical testing of ossicular replacement prostheses.

The results show that, at low frequencies, the TORP performs worse because it is not as stiff (the little titanium link shown in Fig. 6.5b is responsible for the low stiffness of this prosthesis). At high frequencies, the resonance peaks occur and, since the constraining ligaments will have been removed due to the surgical intervention, significant motion of the prosthesis is created. This could be the reason why the TORP prosthesis has a higher recorded loosening rate than the PORP prosthesis.

Furthermore, detailed analysis of the lever ratio concept shows that force amplification by the lever does not really occur, and anyway it makes little sense to talk of a lever since the bones move in a complex motion in three dimensions. Therefore we have concluded that ossicular replacement prosthesis should attempt to match the stiffness of the natural ossicular chain. This result has been further generalized by Kelly [77].

Chapter 7

Discussion

The purpose of these Lecture Notes has been to introduce the subject of pre-clinical testing of prosthetic implants, to outline the importance of the subject, and show how it can be done by presenting a methodology backed-up by concrete examples. In many respects, only the *potential* of pre-clinical testing can be described because there is so much still to be done to establish the "pre-clinical testing platform" illustrated in Fig. 2.1. Nonetheless, it is hoped that the importance of pre-clinical testing is clearer after the description given in Chapter 2 above, and that the methodology presented in Chapter 4 for the development of pre-clinical tests will prove to be a useful basis for designing new pre-clinical tests for prostheses and implants. In particular the part of Chapter 4 dealing with the relationship between mechanobiological phenomena and pre-clinical testing shows, or so the author hopes, that an important and challenging body of research needs to be completed if mechanobiological models are to be used for computer simulation of implant performance. Perhaps it is worth noting that testing of tissue-engineered implants will also rely on the theoretical development of biomechanical regulatory models of tissue differentiation and bone remodelling.

To show the feasibility of pre-clinical testing in practice, the author has chosen examples he has worked on with the Bioengineering Group in Trinity College, Dublin, Ireland. These examples were chosen because the author can explain them best, for no other reason – many other, perhaps better, examples can be found in the literature, see for example Huiskes and Verdonschot [80], Walker and Blunn [81], Anglin *et al.* [82] and Viceconti *et al.* [83]. In Chapter 4 has

been shown how an experimental method has been designed and tested, which can successfully discriminate the performance of hip prosthesis.[1)] The computational pre-clinical testing examples chosen here are from our own work on hip stems and shoulder prostheses. In Chapter 6 it was shown that, for both hips and shoulders, computational models based on finite element analysis can give valuable information that predicts the possible differences of performance between different designs. The underlying trend with respect to computational pre-clinical testing is the generation of visually-accurate finite element models using CT scans which may be used to compare the performance of prostheses (usually a proposed design is compared to one for which there is clinical follow-up) rather than quantitative analysis of a particular design. Our work as presented in Chapter 6 goes further in attempting to quantify performance in terms of probability-of-survival of the fixation. This, we believe, is superior to attempting to qualitatively compare contour plots of the stress data. More research is required on the failure mechanics of the fixation if this line of enquiry is to be pursued (for example the effect of multiaxial stress on bone cement failure, interface failure mechanisms, loading datasets). Finally, to show that the methods extend beyond orthopaedics, an example relevant to ear surgery is presented in Section 6.2.

All this work on pre-clinical testing can be thought of as improving the *testability* of the hypothesis that one implant is superior to another. For many years this hypothesis has been difficult, if not impossible, to test because of the lack of the discriminatory power of experimental, computational, and animal experiments. Even randomised controlled trials have proven problematic, as explained at the end of Section 2.2 above. The various Scandinavian implant registries have changed the environment for orthopaedic implant innovation because register studies can avoid many of the problems with RCT. However, these studies require the implant to be put into clinical use, and many years of follow-up are required before the results become available – we might say that the patient is the experimental model and the operating theatre is the laboratory. Therefore, if we agree that implant registries provide the final conclusive test, implants should

[1)] **This test was developed under the Standards, Measurement and Testing programme of the European Commission in collaboration with the Universities of Nijmegen, Bologna, Göteburg, Berlin and with further sponsorship from Aesculap AG, Tuttlingen, W. Link, Hamburg, Tecres, Verona, Mitab, Sweden, and Sulzer Orthopaedics, Winterthur, Switzerland. (A computational modelling project was carried out in parallel by Ir. J. Stolk under the supervision of Dr. ir. Nico Verdonschot and Prof. dr. ir. Rik Huiskes).**

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situation and will have high variability between specimens. Furthermore only one loading condition is applied whereas, in reality, the joint is exposed to a sequence of various loads depending on patient related factors and daily activity level. Therefore pre-clinical test described in Chapter 5 may be represented by Eq. (7.2) as

$$x\big(<\Omega_I>^{\text{Reduced}}, <\Omega_B>^{\text{Composite-femur}}, <\Gamma_{IB}>, L\big) > x^{\text{norm}}\big(<\Omega_I>^{\text{Reduced}}, <\Omega_B>^{\text{Composite-femur}}, <\Gamma_{IB}>, L\big). \quad (7.2)$$

Great difficulty is caused by comparing the variable of the test to that measured clinically as shown in Eq. (7.3),

$$x\big(<\Omega_I>^{\text{Reduced}}, \Omega_B^{\text{Specific}}, <\Gamma_{IB}>, L\big) > x^{\text{norm}}(<\Omega_I>, <\Omega_B>, <\Gamma_{IB}>, L_1, L_2, \ldots, L_n). \quad (7.3)$$

The sensitivity of the stem to the implantation position is known to be an issue, and may be one of the reasons for the long-term results often found with the Müller prosthesis. Likewise the loading applied is for gait whereas some evidence suggests that stair-climbing, which causes torsion of the implant, initiates the loosening process [24]. This emphasises that pre-clinical tests for inter-comparison of implants (Eq.(7.2)) are more likely to be useful than those comparing to clinical data (Eq.(7.3)) – as far as pre-clinical testing is concerned the use of the clinical data is primarily useful in selecting the monitoring variable x.

Considering the analysis of the hip prosthesis stem and the shoulder glenoid component given in Section 6.1. above; x could be either the volume of cement stressed above a certain threshold, or it could be the probability-of-failure. In the finite element analysis, a specific bone geometry $\Omega_B$is obtained from an averaged bone (there is no variation, either real or reduced), the prosthesis is implanted into some ideal position, the interface conditions are reduced to two (bonded and frictionless debonded), and the loading is reduced to one load case, i.e. the test is,

$$x\big(\Omega_I^{\text{Ideal}}, \Omega_B^{\text{Specific}}, \Gamma_{IB}^{1 \text{ or } 2}, L\big) > x^{\text{norm}}\big(\Omega_I^{\text{Ideal}}, \Omega_B^{\text{Specific}}, \Gamma_{IB}^{1 \text{ or } 2}, L\big). \quad (7.4)$$

It can be seen that Eq. (7.4) is some considerable simplification of Eq. (7.1) which shows the challenges inherent in designing pre-clinical tests.

Passing to a consideration of *secondary stability* where mechanical deteriorations and biological adaptations are included, the problem becomes even more

complex. The mechanobiological problem is one of calculating the rate of change of Ω_B and Γ_{IB} with time. Bone remodelling means that the shape and elasticity of the bone will change over time. Remodelling is usually simulated on a specific bone using finite element analysis:

$$\Omega_B^{\text{Specific}}(t) = \Omega_B^{\text{Specific}}(\psi), \tag{7.5}$$

where ψ denotes a mechanical stimulus for remodelling (such as strain energy density [38] or damage [41]). Similarly the interface will change over time, either becoming stronger in the case of osseointegration or becoming weaker if there is damage accumulation at the interface.

$$\Gamma_{IB}(t) = \Gamma_{IB}(\psi). \tag{7.6}$$

With both Eq. (7.5) and Eq. (7.6) mechanobiological models are required for the solution of the pre-clinical testing problem. Development and confirmation of such bone remodelling models has achieved some considerable success [80] and models to simulate interface osseointegration have been proposed [44, 52], but yet it appears that much research is still to be done.

Chapter 8

Concluding remarks

In conclusion, the problem of pre-clinical testing of implants is highly complex and involves many aspects of biomechanics. It is probably true that most implants can be subjected to some kind of test that can help inform the design process, and can help eliminate inferior design concepts at an early stage. Finite element modelling is widely used and has many advantages [39] and future possibilities; advantages include the possibility of representing complex loading and accurately depicting the geometry based on digital imaging. Disadvantages include lack of constitutive models for interfacial failure and lack of ability to capture the results of variation inherent in the clinical situation. Physical model testing can present many advantages over finite element modelling, but tests are difficult to set up and very time-consuming.

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