13

Joint replacement

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13.1 Introduction

Replacement of damaged articular joints with prosthetic implants has brought relief to millions of patients who would otherwise have been severely limited in mobility and doomed to a life in pain. It is estimated that more than 30 million people in the world are affected by osteoarthritis, one of the various conditions that may cause joint degeneration and lead a patient to a total joint replacement. Trauma and fracture related to osteoporosis also lead to a need for joint replacement.

Metallic devices for orthopaedic applications have been very successful with hundreds of thousands being implanted annually. Original applications were as removable devices, such as those for stabilisation of fractures (Chapter 12). Permanent joint replacements began in the 1960s with Professor Charnley's use of self-curing polymethylmethacrylate (PMMA) 'bone cement', which provided a stable mechanical anchor for a metallic prosthesis in its bony bed. This type of anchoring of implants to bone is called 'cement fixation' if PMMA cement is used (Table 13.1). Clinical success of cemented orthopaedic implants has led to rapid growth in use of implants, especially for hip replacements, called total hip arthroplasty (THA) and knee replacements.

The increase in number of implants coincides with an increase in life expectancy of patients and a decrease in average age of patients receiving an implant. This means that a growing proportion of patients will outlive the expected lifetimes of their prostheses. When an implant fails, revision surgery is required. The patient, now 5 to 25 years older, has an increased probability of operative and postoperative complications. This chapter describes the biomaterials and devices used in joint replacements, their survivability and reasons for failure.

Total joint replacements are permanent implants, unlike those used to treat fractures (Chapter 12). The extensive bone and cartilage removed during implantation makes the procedure irreversible. The design of an implant for joint replacement is based on the kinematics and dynamic load transfer

Table 13.1 Applications and properties of common bioinert implant materials

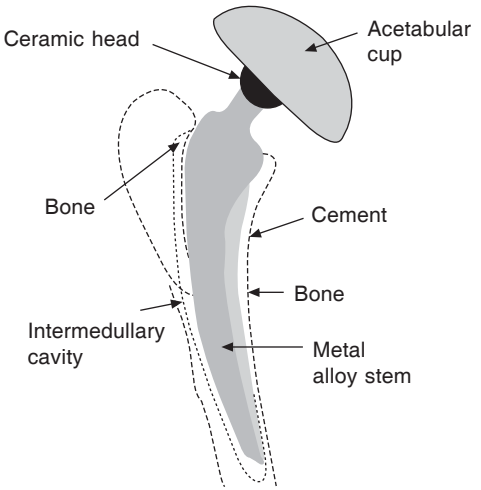
Materials	Applications	Properties
Co–Cr alloy	Stem, head (ball), cup, porous coating	Heavy, hard, stiff, high wear resistance
Ti alloy	Stem, porous coating, metal backing for UHMWPE	Low stiffness
Pure titanium	Porous coating	Excellent osseointegration
Calcium hydroxyapatite	Surface coating	Fast osseointegration, long-term degradation
Alumina	Head, cup	Hard, brittle, high wear resistance
Zirconia	Head	Heavy and high toughness, high wear resistance
UHMWPE	Cup, tibial plateau	
PMMA	Bone cement fixation	Brittle, weak in tension, low fatigue strength

Note: Stem: femoral hip stem/chondylar knee stem; head: femoral head of the hip stem; cup: acetabular cup of the hip.

characteristic of the joint. The material properties, shape, and methods used for fixation of the implant to the patient determine the load transfer characteristics. This is an important element that determines long-term survival of the implant, since bone responds to changes in load transfer with remodelling, e.g. Wolff's law (Chapter 8). Overloading the implant–bone interface or shielding it from load transfer results in bone resorption and subsequent loosening of the implant (stress shielding). The articulating surfaces of the joint must function with minimum friction and produce the least amount of wear products. The implant should be securely fixed to the body as early as possible, ideally during implantation. Removal of the implant should not require destruction of a large amount of surrounding tissues since loss of tissue, especially bone, makes reimplantation difficult and shortens the lifespan of the second joint replacement, called revision surgery.

13.2 Hip joint replacement

The prosthesis for total hip replacement consists of a femoral component and an acetabular component (Fig. 13.1 (CD Fig. 13.1(a)). The femoral stem is divided into head, neck, and shaft. The femoral stem is made of Ti alloy or Co–Cr alloy (less expensive 316L stainless steel is still used in some countries) and is fixed into a reamed medullary canal by cementation or press fitting. Femoral heads are made of Co–Cr alloy, alumina, or zirconia. Although Ti alloy heads function well under clean articulating conditions, they are less used because of their low resistance to third-body wear.



13.1 Schematic of a cemented Charnley hip prosthesis (CD Fig. 13.1(a)).

Monolithic hip prostheses consist of one part with the advantage of being less expensive and less prone to corrosion or disassembly. Modular devices consist of two or more parts and require assembly during surgery. Modular components allow customising of the implant intra-operatively and during revision surgeries. The length of an extremity can be modified by using a different femoral neck length after the stem has been cemented in place. A worn polyethylene-bearing surface can be exchanged for a new one without removing the metallic part of the prosthesis from the bone. In modular implants, the femoral head is fitted to the femoral neck with a Morse taper, which allows changes in head material and diameter and neck length. Table 13.1, summarises the most frequently used combinations of materials in total hip replacement. Compositions and properties of metals used are given in Tables 12.3 and 12.4.

A monolithic acetabular component is made of ultra-high molecular weight polyethylene (UHMWPE) (Chapters 4 and 10). A modular acetabular component consists of a metallic shell and an UHMWPE insert. The metallic shell minimises microdeformation of the UHMWPE and provides a porous surface for biological fixation of the cup. The metallic shell allows worn polyethylene liners to be exchanged. If a patient has repetitive dislocation of the hip after surgery, the metallic shell allows replacement of the liner with a more constrained one to provide additional stability The metallic shell and liner must fit precisely because dislodgement of the insert results in dislocation of the hip and damage of the femoral head, since it is in direct contact with the metallic shell. Micromotion between insert and shell produces additional polyethylene debris, which contributes to bone loss.

The hip joint is a ball-and-socket joint. Stability comes from congruity of the implants, pelvic muscles and capsule. Total hip components are optimised to provide a wide range of motion of the prosthetic joint without impingement of the neck of the prosthesis on the rim of the acetabular cup. Designs have evolved to enable implants to support loads that may reach eight times body weight. Proper femoral neck length and correct restoration of the centre of motion and femoral offset decrease the bending stress on the prosthesis–bone interface.

High stress concentrations or stress shielding may result in bone resorption around the implant. The origin of stress shielding is the very large elastic modulus (stiffness) of orthopaedic metals, especially Co–Cr–Mo alloys (10 to 15 times that of cortical bone), as summarised in Chapter 12.

13.3 Failure mechanisms

There are several causes of failure of total joint replacements described in *Clinical Performance of Skeletal Prostheses* (see Reading list) and summarised in Table 13.2.

Table 13.2 Major causes of failure of total joint replacements

Aseptic loosening
(a) Biological factors including host reaction
(b) Material properties of the implant component
Infection
Surgical techniques/mechanical causes
Dislocation
Implant fracture
Bone fracture

Aseptic loosening of prosthetic components is the most common cause of failure. Aseptic loosening refers to the failure of joint prostheses without the presence of mechanical cause or infection. It is often associated with osteolysis (bone resorption) and an inflammatory cellular response within the joint. Clinically, aseptic loosening is defined on the basis of the radiographic evidence of the presence of lucent lines at the interface between the bone cement and the implant. Migration of the implant component may also cause significant osteolysis. Osteolysis has also been identified around well-fixed implants. Complications in patients with joint prostheses are often related to the release of particulate wear debris. Particulate material from total joint replacements falls into three categories:

1. ultra-high molecular weight polyethylene (UHMWPE) from the acetabular component or tibial tray;

2. metal wear debris from acetabular, femoral, or tibial components made from titanium alloy or cobalt chrome alloy;
3. polymethylmethacrylate bone cement.

Load bearing and motion of the prosthesis produces wear debris from the articulating surfaces and from interfaces where there is micromotion. The principal source of wear under normal conditions is the UHMWPE (ultra-high molecular weight polyethylene)-bearing surface in the cup. Many particles are generated with each step, and a large proportion of these particles are smaller than one micrometre in diameter. Cells from the immune system of the patient respond to the polyethylene particles as foreign material and initiate a complex inflammatory response. This response leads to focal bone loss (osteolysis), bone resorption, loosening and/or fracture of the bone. Numerous efforts are under way to modify the material properties of UHMWPE, to harden and improve the surface finish of the femoral head, and to develop other bearing couples, for example, ceramic-to-ceramic and metal-to-metal (Section 13.5).

13.4 Survivability of total hip replacements

The cemented low friction (Charnley-type) total hip arthroplasty (THA), using a metallic femoral component and UHMWPE cup, has the highest level of clinical success, as illustrated in Fig. 13.2. The Kaplan–Meier survivability curve shown in Fig. 13.2 is based upon 42 published clinical studies involving a total of 15 051 patients for time periods as long as 25 years (see *Clinical Performance of Skeletal Prostheses* in the Reading list). The predicted survival rates for the cemented low friction type of THA are:

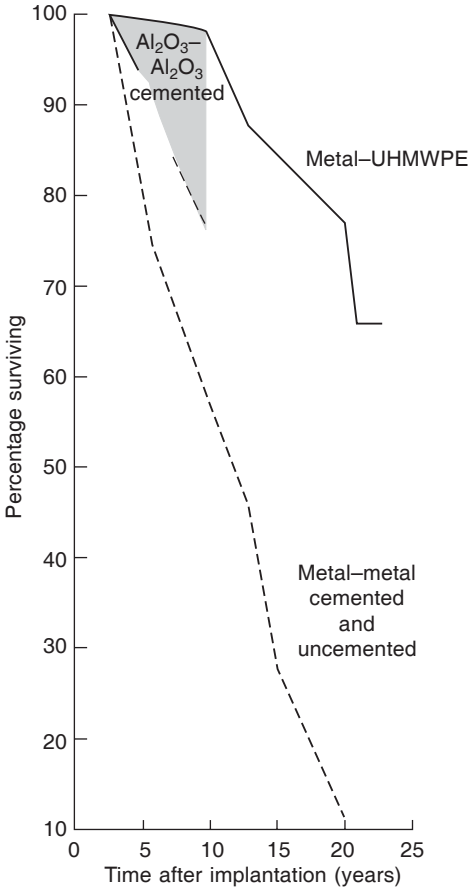
- 5 years 99.41 ± 0.02%;
- 10 years 95.48 ± 0.04%;
- 15 years 83.12 ± 0.18%;
- 20 years 66.53 ± 0.35%.

A lifetime of 19 years is expected for the average total hip replacement using the cemented low friction prosthesis.

It is uncertain whether the fall-off in survivability between 15 and 20 years, shown in Fig. 13.2, will continue. This is because of the relatively small number of long-term studies reported and the significant improvement in clinical techniques developed during the last 15 years. Advances in surgical procedures should lead to improved 15–25 year lifetimes for cemented THAs.

Life expectancy of THAs depends upon the following factors:

- gender (males lower than females);
- weight (overweight lower than normal);
- age (younger patients with high activity lower than older).



13.2 Schematic graph of the survivability of total hip replacements.

Low friction THA has a low risk of failure for patients > 65 years of age and is generally considered to be the standard for hip replacement.

Other types of THA systems generally show lower survival rates than the cemented low friction type. The alumina–alumina cemented prosthesis is the exception (see below). However, the distribution of results and relatively low number of clinical studies reported for alumina–alumina prostheses currently makes it difficult to establish statistical evidence that this system improves THA survivability by eliminating UHMWPE wear debris.

Uncemented THA systems, regardless of means of fixation, generally show lower survivability results, however, improvements in hydroxyapatite (HA) coatings may lead to longer survivability during the next 10 years. The difference appears by 5 years and expands with time. The difference between cemented and uncemented THA shown in Fig. 13.2 is confirmed by the Havelin *et al*. study of 14 000 patients which shows two times the number of

failures of uncemented hips by 5 years (see *Clinical Performance of Skeletal Prostheses* in the Reading list).

13.5 New developments to improve survivability

High-density, high-purity (> 99.5%) Al_2O_3 (alumina) is used as the femoral head in load-bearing hip prostheses for younger patients because of its combination of excellent corrosion resistance, good biocompatibility, high wear resistance and high strength. Most Al_2O_3 devices are very fine grained polycrystalline α-Al_2O_3. Al_2O_3 with an average grain size of <4 μm and >99.7% purity exhibits good flexural strength and excellent compressive strength. These and other physical properties are summarised in Table 13.3 for a commercially available implant material, along with the International Standards Organisation (ISO) requirements. Alumina implants that meet or exceed ISO standards have excellent resistance to dynamic and static fatigue, and resist sub-critical crack growth and implant failure. Load-bearing lifetimes of 30 years to 12 000 N loads or 200 MPa stresses have been predicted and are confirmed by long-term clinical success, with survivabilities of > 80% after 10 years. It is estimated that more than one million hip prostheses to date have been implanted with an alumina ball for the femoral head component and that the number is growing by at least 100 000 per year. Femoral components of total hip prostheses with alumina balls are shown in the CD lecture for this chapter, as discussed above. Clinical studies indicate that alumina/UHMWPE wear rates are approximately ten times lower than Co–Cr–Mo alloy/UHMWPE wear rates of 200 μm per year. Alumina–alumina wear rates are only about 2 μm per year.

Table 13.3 Properties of alumina and zirconia used in surgical implants

Property	Unit	Al_2O_3	TZP	Mg–PSZ
Purity	%	>99.7	97	96.5
Y_2O_3 / MgO	%	<0.3	3 mol	304 wt
Density	g/cm^3	3.98	6.05	5.72
Grain size (average)	μm	3.6	0.2–0.4	0.42
Bending strength	MPa	595	1000	800
Compressive strength	MPa	4250	2000	1850
Young's modulus	GPa	400	150	208
Hardness	HV	2400	1200	1120
Fracture toughness K_{1C}	MPa/m$^{1/2}$	5	7	8

Due to moderate flexural strength and toughness, the size of alumina femoral heads are limited to 32 mm or greater. A higher fracture toughness, flexural strength and lower Young's modulus make medical grade zirconia (ZrO_2) an attractive option for producing smaller femoral heads. A comparison

of physical properties of both ceramics is given in Table 13.3. Concerns about long-term survivability of zirconia in the physiological environment compared with 30 years of successful clinical use of alumina makes it difficult to judge the merits of the two materials at this time. A third alternative to either a metal or ceramic femoral head is to produce a coating of ceramic on the surface of a specially designed metal alloy. This is accomplished by converting a zirconium alloy head (97.5% Zr–2.5% Nb) into a ceramic by heating in an oxidising atmosphere to grow a very thin, tough, wear-resistant oxide ceramic layer. The manufacturer reports the oxide coating is nearly 5000 times more resistant to abrasions and scratches than a metallic head and should produce substantially fewer UHMWPE wear debris particles since the coating is 160 times smoother than metal heads.

13.6 Knee joint replacements

The prosthesis for total knee replacement consists of a femoral, a tibial, and/or a patellar component. Compared with the hip, the knee has a more complicated geometry and movement biomechanics and it is not intrinsically stable. In a normal knee the centre of movement is controlled by the geometry of the ligaments. As the knee moves, the ligaments rotate on their bony attachments and the centre of movement also moves. The eccentric movement of the knee helps distribute the load throughout the entire joint surface.

The prostheses for total knee replacement (see CD Figs. 13.8–13.10) can be divided according to the extent to which they rely on the ligaments for stability. Constrained implants have a hinge articulation, with a fixed axis of rotation, and are indicated when all the ligaments are absent, for example, in reconstructive procedures for tumour surgery. Semi-constrained prostheses control posterior displacement of the tibia on the femur and medial–lateral angulation of the knee but rely on remaining ligaments and joint capsule to provide the rest of the constraint.

Semi-constrained knee prostheses are often used in patients with severe angular deformities of the extremities or in those that require revision surgery, when moderate ligamentous instability has developed. Non-constrained implants provide minimal or no constraint. The prostheses that provide minimal constraint require resection of the posterior cruciate ligament during implantation, and the prosthetic constraint reproduces that normally provided by this ligament.

The prostheses that provide no constraint spare the posterior cruciate ligament. These implants are indicated in patients who have joint degeneration with minimal or no ligamentous instability. As the degree of constraint increases with knee replacements, the need to use femoral and tibial intramedullary extensions of the prosthesis is greater, since the loads normally shared with the ligaments are then transferred to the prosthesis–bone interface.

Table 13.4 Most commonly used knee prostheses

Type I Hinged	Type III Unicompartmental	
Waldius Shiers St Georg Stanmore Guepar	McKeever MacIntosh Porous coated anatomic Marmor Blauth St Georg Sledge Modular Geometric	Oxford Lotus Polycentric/Gunston Huit Unicondylar Robert Brigham Mark I Mark II

Type II Constrained	Type IV Meniscal bearing
Total condylar III Kinematic rotating hinge Sheehan Attenborough Spherocentric	N. J. Low contact stress Accord Oxford Meniscal

Type V Condylar resurfacing

Posterior cruciate ligament retaining	Posterior cruciate ligament sacrificing	Posterior cruciate ligament substituting
Duocondylar Duopatellar Posterior cruciate condylar Kinematic I Kinemax Robert Brigham Townley/anatomic Cloutier AGC Miller–Galante I Anatomic modular knee Press-fit condylar	Total condylar Total condylar II Freeman–swanson ICLH Porous coated anatomic	Insall–Burstein Kinematic stabiliser I Kinematic stabiliser II Kinematic II

Total knee replacements can be implanted with, or without, PMMA cement, the latter relying on porous coating for fixation. The femoral components are typically made of Co–Cr or titanium alloy and the tibial components of UHMWPE. In modular components, the tibial polyethylene component assembles onto a metallic tibial tray. The patellar component is made of UHMWPE, and a metal back is added to components designed for uncemented use.

The wear characteristic of the surface of tibial plateaux differs from acetabular components of total hip replacements. The point contact stress and sliding motion of the knee components may result in delamination and fatigue wear of the UHMWPE. Presumably because of the relatively larger

particle size of polyethylene debris, osteolysis around a total knee joint is less frequent than in a total hip replacement. The relatively small size of the patellar component compared with the forces that travel through the extensor mechanism, and the small area of bone available for anchorage of the prosthesis, make the patella vulnerable. The most commonly used knee prostheses are shown in Table 13.4. All total knee arthroplasty (TKA) systems exhibit generally similar survivability results of 82–90% at 10 years.

There is no statistical difference between cemented and uncemented TKAs.

13.7 Ankle joint replacements

Total ankle replacements have not met with as much success as total hip and knee replacements, and they may loosen within a few years of service. This is mainly due to the high load transfer demand over the relatively small ankle surface area and the need to replace three articulating surfaces (tibial, talar and fibular). The joint configurations that have been used are cylindrical, reverse cylindrical and spherical. The materials used to construct ankle joints are usually Co–Cr alloy and UHMWPE. Degeneration of the ankle joint is currently treated with fusion of the joint, since prostheses for total ankle replacement are considered to be still under initial development. CD Fig. 13.13 shows ankle and other total joint replacements.

13.8 Shoulder joint replacements

The prostheses for total shoulder replacement consist of a humeral and a glenoid component. Like the femoral stem, the humeral component can be divided into head, neck and shaft. Variations in the length of the neck result in changes in the length of the extremity. However, since the patient's perception of length of the upper extremity is not as accurate as that of the lower, the various neck lengths are used to fine-tune the tension of the soft tissues, to obtain maximal stability and range of motion.

The shoulder has the largest range of motion in a body, which results from a shallow ball and socket joint that allows a combination of rotation and sliding motions between the joint surfaces. To compensate for the compromise in congruity, the shoulder has an elaborate capsular and ligamentous structure, which provides the basic stabilisation; however, the muscle girdle of the shoulder provides additional dynamic stability. A decrease in the radius of curvature of the implant to compensate for soft-tissue instability will result in a decrease in the range of motion. There is no statistically significant difference between cemented and uncemented shoulder prosthetic systems.

Prior condition of the patient and associated soft tissue defects are the primary factors for success of a shoulder prosthesis. If a shoulder prosthesis is essential, the results show that an unconstrained hemi-arthroplasty or unconstrained total arthroplasty is preferred.

13.9 Elbow joint replacements

The elbow joint is a hinge-type joint allowing mostly flexion and extension but having a polycentric motion. The elbow joint implants are hinged, semi-constrained or unconstrained. These implants, like those of the ankle, have a high failure rate and are not commonly used. The high loosening rate is the result of high rotational moments, limited bone stock for fixation and minimal ligamentous support. In contrast to fusions of the ankle, which function well, fusions of the elbow result in a moderate degree of incapacitation. All types of elbow prostheses provide marked improvement in quality of life, less pain and more range of motion.

Failures of elbow prostheses are uncommon (10–15%) for most types of implants but complications range from 30 to 80%. There are insufficient data to establish statistically significant differences between the types of prostheses used.

13.10 Finger joint replacements

Finger joint replacements are divided into three types: hinge, polycentric and space filler. The most widely used are the space-filler type. These are made of high-performance silicone rubber (polydimethylsiloxane) and are stabilised by a passive fixation method. This method depends on the development of a thin, fibrous membrane between implant and bone. This fixation can provide only minimal rigidity of the joint. Implant wear and cold flow associated with erosive cystic changes of adjacent bone have been reported with silicone implants. Pyrolytic carbon finger joints have been recently developed and show considerable promise for improved clinical success for patients with arthritic joints. See Chapter 14 in *Introduction to Bioceramics* in the Reading list for a description of the structure and properties of pyrolytic carbon.

13.11 Prosthetic intervertebral disc

Fusion of a spinal motion segment in degenerative disc disease increases the stiffness across the stabilised segment and stress on adjacent levels. The results of spinal fusion are unpredictable. It can lead to further degeneration of the adjacent spinal levels. In order to reduce the adverse effects of the fusion process, artificial disc prostheses, similar in concept to the total joint replacements above, have been developed. These designs, although in infancy, range from flexible polymer inserts to ball-and-socket or hinge-type designs.

13.12 Summary

Most of the articulating joints of the body are being replaced routinely by prostheses, the most common being the total hip replacement. Monolithic hip prostheses consist of a femoral stem, typically made from Co–Cr or Ti alloy and an acetabular component made of ultra-high molecular weight polyethylene. Polymethylmethacrylate (PMMA) is used for cement fixation and HA or porous metal coatings are used for uncemented fixation. The most common cause of failure is aseptic loosening caused by bone lysis due to wear debris. Alumina–alumina bearing surfaces minimise wear debris. Survivability of total hip prostheses are excellent; approximately 66% at 20 years. Other joint replacements include knees, shoulders, elbow, fingers and intervertebral discs.

13.13 Key definitions

Bone resorption: a type of bone loss due to the greater osteoclastic activity than osteogenic activity.

Callus: unorganised meshwork of woven bone, which is formed following fracture of bone to achieve an early stability of the fracture.

Fibrous membrane: thin layer of soft tissue that covers an implant to isolate it from the body.

Necrosis: cell death caused by enzymes or heat.

Osseointegration: direct contact of bone tissues to an implant surface without fibrous membrane.

Osteoclastic: activity of bone destruction or removal of old bone by bone cells called osteoclasts.

Osteogenic: activity of bone formation in growth or repair of bone. Bone cells for the osteogenic activity are called osteoblasts.

Primary healing: bone healing in which union occurs directly without forming a callus.

Secondary healing: bone union with a callus formation.

13.14 Reading list

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