CURRENT ORTHOPAEDICS

Failed Joint Replacement

Causes and prevention—(i) Biomechanics

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Twenty-five years ago John Charnley proclaimed the three basic conceptions for successful hip replacement arthroplasty[1]. His recognition that low friction at the prosthetic bearing surface could greatly decrease the chances of component loosening from bone, that 316L stainless steel, ultra high molecular weight polyethylene and methylmethacrylate self curing bone cement were biocompatible and caused little or no tissue inflammation in either bulk of particulate form, and that methylmethacrylate bone cement could durably fix prosthetic components to bone have been foundational to the development of joint replacement techniques that bring comfort and increased independence to well over 250,000 patients each year. While it is worthwile to reflect on our failures and to rectify our design and usage short comings, we should maintain confidence and be pleased about the great majority of long term durable successes.

Loosening with subsequent progressive development of increasing pain, disability, and deformity is the single most frequent cause of failure of total joint replacement arthroplasty. Accordingly, most attention will be paid to biomechanical factors in prosthetic component loosening. It is of importance to generalise from experiences with a group of patients treated with primary arthroplasty since this is a group that should expect good results. The others: the young with monarticular arthritis; those who have had prior failed arthroplasties; those with prior infection or with unusual pathologies such as neuropathic or Charcot joints each represent very special cases with far greater risks of failure. Every one of these patients is unique, and while many of the generalisations reviewed here may be of value in management of these patients, they must in addition, have individual consideration by a surgeon with more than ordinary experience and education in this field.

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Several causes of failure play additive roles to biomechanical factors. The patient's understanding of the serious nature and limitations of the surgery is important. A patient that returns to racket games is likely to repeatedly overload the joint, the cement interface, and the host bone itself. Failure in this case should be the patient's responsibility and not that of biomechanics of design or application. The very obese patient presents a similar problem. Those patients with severe systemic disease resulting in progressive bone weakening or loss, hyperparathyroidism as an example, or patients with widespread osteopoenia or osteonecrosis secondary to long term steroid usage have primary systemic failures, not biomechanical failures of total joint replacement.

Similarly, modern well trained surgeons are expected to perform at a skill level that reflects not only an understanding of the procedure, but also of physiological and biomechanical principles, concepts of modern cement usage, and patient management. Repeated failures due to improper patient selection or grossly inadequate surgical technique are outside that scope of a general review of the biomechanics of total joint failure.

Biomechanical aspects of failed joint replacement do however relate to the pre-operative systemic and local condition of the patient, the mechanical quality of the patient's trabecular bone, the quality of the bone cement interface that can be surgically accomplished, the design and metallurgy of the prosthetic components, and the techniques of surgical implantation of the devices. Many of these factors apply to a consideration of bone ingrowth cementless fixation as well. This usage in addition, demands a serious concern for the provision of immediate mechanical stability and for successful biological production of the ingrowth of bone of sufficient quality for durable component fixation.

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Trabecular bone considerations

A general biomechanical principle applies to our usage of the patient's trabecular bone in total joint replacement surgery. All living bone responds to changes in its environment, to changes of the stresses to which it is subjected. If a prosthesis is designed in such a way that the patient's bone is underutilised, atrophy will occur and component fixation failure will likely result (Fig. 1). The best example of atrophy resulting from insufficient loadings is seen with severe stress shielding. A securely fixed hip joint femoral component can transmit forces from the ball and socket articulation down the prosthetic stem to near the tip of the prosthesis where the load is transferred to the femoral shaft. Under these circumstances and considering the relative stiffness of the metal implant, the bone of the proximal femur is bypassed and shielded from the anticipated normal stress. Atrophy will rapidly occur with consequent loss of proximal support to the femoral component. With the loss of proximal metaphyseal support the stress on the limited remaining area of prosthesis may become excessive and failure may occur. If on the other hand the supporting bone is grossly overloaded, immediate trabecular or cortical fractures may occur (Fig. 2). If the fractures of supporting trabecular bone do not heal with the prosthetic component in an acceptable position arthoplasty failure will rapidly result. Such an overload is frequently related to the weight of the patient or to an unusual activity demand or an accident.

The most frequent disadvantageous interaction with the supporting bone is long term cyclic overload with fatigue fracture of supporting trabecular bone, pseudarthrosis of the trabecular fractures seen as a radiolucent line and finally component loosening and arthroplasty failure (Fig. 3).

Total joints should thus be designed and installed such that the supporting bone is stressed sufficiently to stimulate maintenance or hypertrophy of the all important trabecular bone[2] (Fig. 4).

The cement bone interface

With cement usage for fixation, the cement-bone interface becomes of greatest interest[3]. Our design and usage of cemented components assumes an intimate, completely connected, firmly fixed boundary between bone and cement. This connected interface should be capable of directly transmitting normal compressive forces to supporting trabecular bone. Likewise interdigitation of cement into the trabecular bone provides an effective boundary that resists shear or torsional disrupting forces. For the tibial resurfacing component interface at the knee this is particularly relevant. In this application the cemented interface can even to some extent resist the tensile disrupting loads that occur with assymetric loading (Fig. 5).

At times a complete direct interdigitation with trabecular bone may be impossible or undesirable. The best example is the cemented acetabulum where it is possibly desirable to retain subchondral bone.

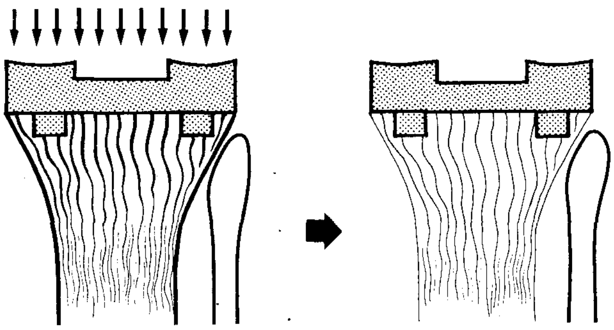

Fig. 1 Underutilised bone atrophies and provides poor support for prosthetic components

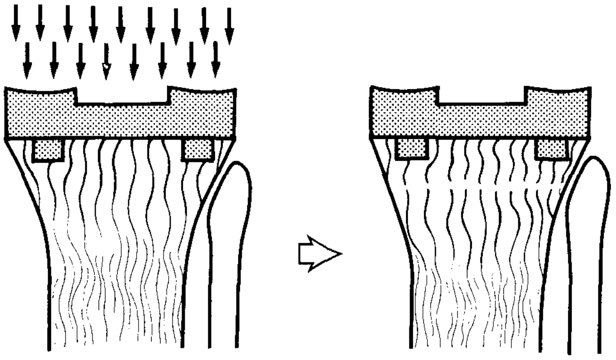

Fig. 3 Fatigue failure of supporting trabecular bone can result in trabecular nonunion, trabecular pseudarthrosis formation, and the appearance of a radiolucent line followed by loosening of the component.

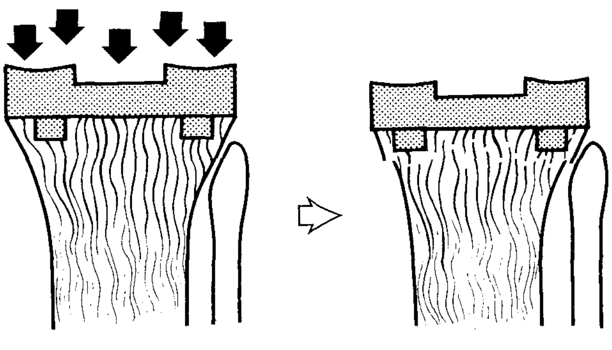

Fig. 2 Overloaded trabecular bone sustains fracture failure. Fractures of supporting trabecular bone can heal and ultimately support a prosthetic component.

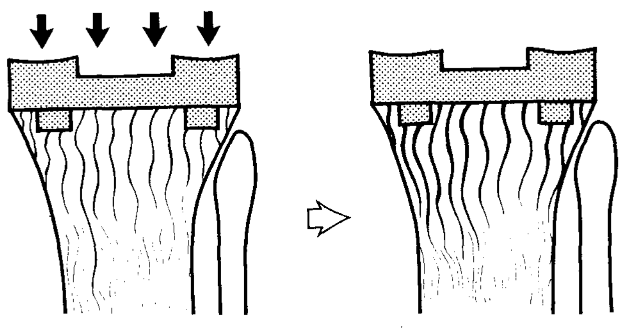

Fig. 4 Appropriate cyclic loading of the trabecular bone supporting a prosthetic component results in maintenance of trabecular strength or in hypertrophy.

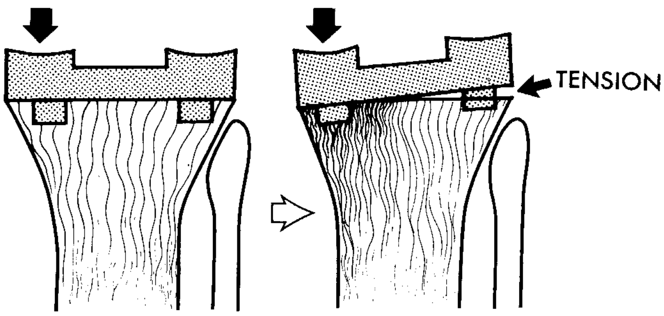

Fig. 5 Asymetric loading of a tibial component can cause tilting of the component and tension at the bone cement interface.

Cement used against cortical or naked subchondral bone can transmit compression loads durably and successfully. Such an interface is very subject to torsional or shear failure. Under these circumstances there is an increased requirement to drill holes throughout the acetabular subchondral shell so as to provide cement protrusion that can effectively resist shear or torsional forces. The interaction of a stemmed component cemented into the diaphysis of a long bone is a special case to be expanded in consideration of failures of femoral components.

Methylmethacrylate bone cement

Since its introduction as the major fixation material for total joint replacement arthroplasty methylmethacrylate bone cement has been repeatedly modified to change its viscosity, time course of polymerisation, and strength. Its essential biomechanical characteristics have, however, remained essentially the same. This material has an elastic modulus that is greater than polyethylene and trabecular bone, but less than cortical bone. Methylmethacrylate is remarkably resistant to compressive loading, but is relatively brittle and fails at relatively low levels when subjected to shear or bending forces. Its fatigue resistance is naturally poor[1]. The fatigue resistance as well as the observed resistance to tension, shear, and bending loads is greatly diminished by voids, cysts, and the like. The dominant role of failure of the methylmethacrylate bone cement in total joint replacement failure has led to recommendation for centrifugation[4] of the liquid mix to eliminate voids and to the use of vacuum mixing for the same purpose. Many current prosthetic component designs subject their cement mantle to concentrated tensile, bending, or shear loads. To a considerable extent a metal prosthetic surface that provides for intimate penetration and interdigiation[3] can, in a sense, reinforce the weaker material and protect it from otherwise detrimental bending or shear loadings.

The prosthetic components

The materials from which total joint implants are fabricated have greatly improved during the past 25 years. Stainless steel is less strong than cobalt-chromium-molybdenum (Co-Chro-Moly) alloy or titanium. Early hip femoral components made from these materials failed because the materials strength associated with specific designs was simply insufficient. Forged Co-Chro-Moly and the addition of hot isostatic pressing and powder metal technology has greatly increased the mechanical strength characteristics of Co-Chro-Moly and titanium alloys. These fabrication techniques together with advantageous design changes, the avoidance of stress raisers, and better interaction with and utilisation of cement have virtually eliminated component breakage for conventional prostheses.

The recent production of metal components with porous surface regions for bony ingrowth requires high temperature sintering techniques that to some extent negate the value of the described strengthening techniques. Still, implant failure has greatly decreased in frequency and importance as a cause of total joint replacement failure. Notches, holes, and surface irregularities in or on metal components greatly increase the local stress experienced by the component. This stress concentration effect may lead to rapid component fatigue failure. Positive surface irregularities of the prosthetic component cause a mated negative irregularity in the cement mantle which is even more stress raiser vulnerable than the metal part. Because the cement mantle is relatively thin, these component design irregularities may contribute to early cement failure and consequent joint failure.

Specific biomechanical aspects of total hip arthroplasty failure

While, in principle, a femoral component could be designed to work in the desirable compressive mode, in fact these implants are nearly always subjected to cantilever bending. Body weight and muscle forces act at the prosthetic femoral head. The medial calcar fulcrum resists varus component displacements as does the lateral wall of the femur near the tip of the prosthesis stem. The characteristic modes of failure for cemented femoral components have been outlined in some detail[5].

Some general biomechanical design factors must also be emphasised. The greater the neck length of the femoral component and the lesser the neck-shaft angle, the greater the cantilever bending load that must be sustained by the now most vulnerable mantle of cement. A prosthesis implanted in relative varus increases the cantilever bending loads experienced by the prosthesis, the cement, and the supporting bone. A varus surgical implantation thus subjects the patient to a greatly increased risk of failure. This same implantation error causes the inner ring of the cement mantle at the calcar and the cement mantle at the lateral distal cortex to become areas of great stress concentration and thus they assume maximum vulnerability to cement failure. This same varus misplacement of the femoral component causes the cement at the calcar and at the lateral cortex at the tip of the stem of the prosthesis to be quite thin and again the cement strength is compromised.

The cement that surrounds the proximal stem region of a typical tapered stem component is subjected to a complex loading pattern. It is easy to accept that the cement between the calcar region of the femoral neck and the proximal stem is loaded in compression. As, however, this region is, in principle, forced away from the trochanter supported lateral cement, the 'circumference' of the cement band is increased. The so called 'hoop' stresses result in a true tensile loading mode to which the cement is particularly vulnerable (Fig. 6). The basic features of a femoral component, long neck, small neck stem angles, all increase these cement damaging loads. Should the generally wedge shaped stem component subside after cement breakage at the tip again generalised 'hoop' stresses would be generated in the cement surrounding the prosthetic stem.

The effects of stem length have been studied in some detail[6]. The longer the stem, the lesser the stresses experienced by the bone and cement at the calcar and at the distal lateral stem. On the other hand, the longer the stem, assuming an excellent job of cement fixation, the greater is the extent and importance of stress shielding. With proximal stress shielding the patient's bone can atrophy and the proximal support of the prosthesis become compromised.

The function and possible value of a 'collar' remains uncertain. While in principle, the collar should load the femoral neck in a more natural compressive way[7], strain gauge studies of the proximal femur have failed to generally document this effect[8]. It appears very difficult to achieve a truly precise fit between the prosthetic collar and the remaining femoral calcar. Post-operative radiograph often demonstrate resorption of bone in just this region partially validating the supposition that effective 'natural' load transfer rarely takes place.

With failure of cement and consequent failure to support the prosthesis at the proximal stem, the fulcrum for cantilever bending moves more distally.

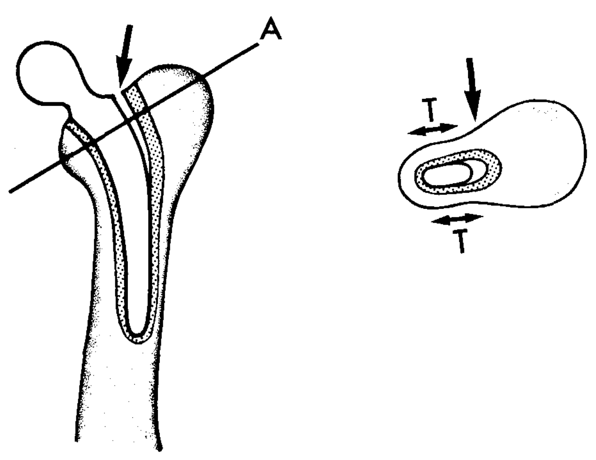

Fig. 6 Varus placement and further varus deformation of a femoral component causes "hoop stresses in the proximal cement mantle. "Hoop" stresses are actually tensile stresses in the "ring" of cement at the calcar.

The distance between the points of maximum stress decreases and the loads supported by the cement and the bone become much greater for the same individual performing the same tasks. These same circumstances increase the length of the unsupported prosthesis which tends increasingly to be bent around the fulcrum. Most reports of stem failure or breakage[8,9] emphasise this mechanism as well as poor prosthetic stem design.

The cross sectional shape of the prosthetic stem is of major importance. Diamond or wedge stem cross sectional shapes cause thin cement mantles at regions of greatest vulnerability. These designs have been implicated unusually frequently in proximal cement and support failure. Relatively large, smooth rounded cross sectional contours are to be recommended. Any shape design feature which imprints sharp edges or corners into the cement mantle or any prosthetic stem design which leads to irregularity in the thickness of the cement mantle leads to a greater incidence of loosening and total hip failure.

Because secure distal fixation coupled with very stiff prosthetic stems can result in the more proximal bone of the femur being subjected to unnaturally low loads and since these pathologic low loads can lead to bone resorption, atrophy, and eventual failure titanium and its alloys have been recommended as more appropriate materials from which to fabricate femoral components. The elastic modulus of titanium and its alloys, $0.10 \times 10^6 N/mm^2$, is approximately one half that of Co-Chro-Moly alloys, $0.20 \times 10^6 N/mm^2$. Both of these orthpaedically useful metals have moduli which are magnitudes greater than bone, $0.01 \times 10^6 N/mm^2$. While titanium as a prosthetic stem material may have a minimal advantage in this respect, the differences are so small that a recommendation for the selection of one or the other metals cannot be strongly supported.

Solid biomechanical data on the mechanics of bony ingrowth fixation of the femoral components for total hip surgery is presently not available. The extent of expected ingrowth in humans, the strength of the interface, and the necessary areas of union are presently unknown.

For more than a decade the femoral stem-shaft interface appeared to be the most vulnerable to loosening, but with improved design and materials together with better use of acrylic cement[3,4,7,10] and emphasis on consistent proper operative placement, first failures have now begun to occur on the acetabular side of the joint.

Charnley's early selection of teflon, which he thought to be a biocompatible material, resulted in an important disaster. In fact, while bulk teflon was non-reactive, the omnipresent wear debris was not well tolerated. Hundreds of revision procedures with debridement and a change to polyethylene actabular components for failed teflon parts were performed by Charnley himself before going on to de novo polyethylene cups. Charnley also demonstrated that a 22 mm femoral head was associated with less friction at the bearing surface under load, and thus should further limit the torsional and frictional forces being transmitted to the acetabulum-bone interface. The advantages of a 22 mm as opposed to a 32 mm one have never been unequivocally demonstrated under clinical circumstances. Certainly the smaller head size allows for greater wall thickness for the acetabulum. The disadvantage of thin, less than 6 mm unsupported polytheylene for knee tibial components has become evident. Similarly the thin acetabular components used with femoral head resurfacing arthroplasty were associated with frequent rapid loosening of these pelvic components lending credance to Charnley's contention.

Finite element analysis studies have clearly indicated[11,12,13] that both the muscular and gravitational forces transmitted across the head-socket bearing interface are best distributed to the supporting pelvic bone if the actetabular polyethylene component is supported by a strong metal shell. Careful clinical studies have demonstrated[14] an advantage for metal support, particularly for younger active patients. The question regarding whether to ream through the subchondral plate to the underlying trabecular bone or to clean but preserve the subchondral plate and to use multiple penetrations of the shell into the pelvic trabecular bone for torsional control is complex and has not been resolved.

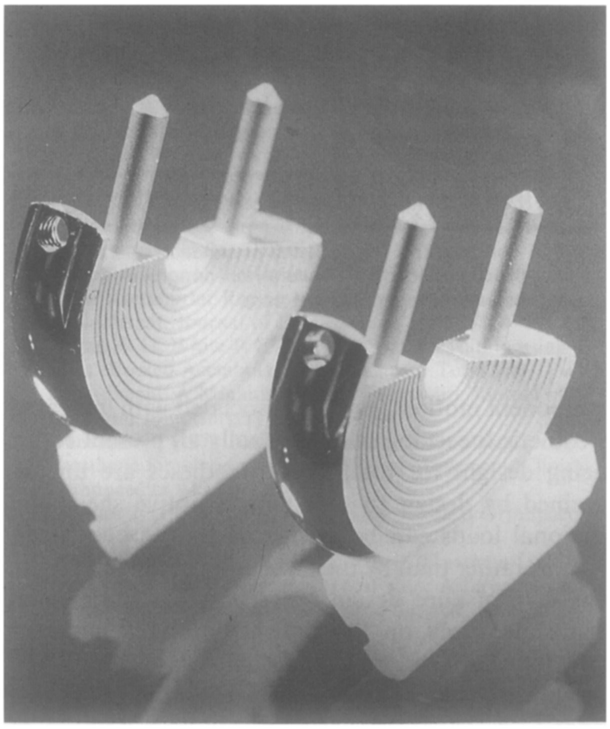

Fig. 7 The polycentric knee prosthesis, a four part knee with little constraint.

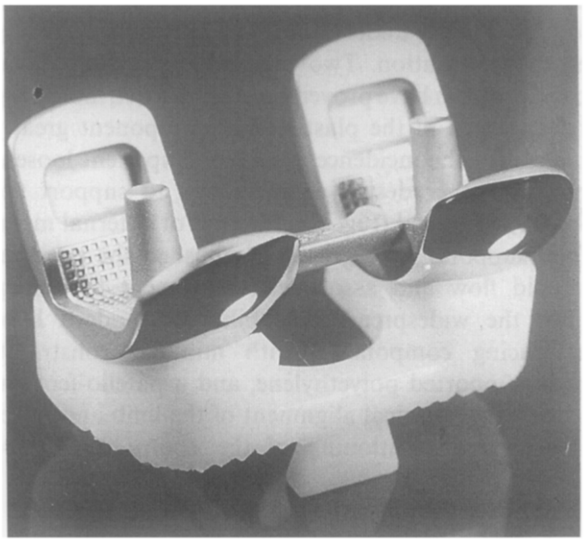

Fig. 8 The Geometric knee prosthesis, A two part resurfacing prosthesis with considerable constraint because of the congruent fit of the parts.

Some biomechanical aspects of total knee arthroplasty failure

Clinical problems associated with the use of early total knee prostheses rapidly identified the many biomechanical design and utilization shortcomings that limited the general applicability of the operation. These shortcomings have been subsequently addressed by design and procedural improvements such that now total knee arthroplasty can be just as successful as total hip arthroplasty.

The first modern total knee prosthesis, the Polycentric, designed by Frank Gunston[15] was composed of two femoral half disks and two unsupported tibial tracks (Fig. 7). Each of the four components was implanted separately. The three rotational and three translational degrees of freedom associated with the implantation of each piece made the prosthesis a surgeon's nightmare. Optimally each piece should be level, face forward, be parallel with all other implanted pieces, and be implanted to the proper depth. Considerable surgical experience, attention to detail, and artistry were necessary to accomplish an optimal installation. This total knee design and that of Marmor[16] emphasised the need for linkage of the medial and lateral femoral and medial and lateral tibial component parts and of instrumentation to facilitate installation.

The next popular prosthesis model, the Geometric has linked medial and lateral femoral and tibial components (Fig. 8). The curvature of the convex prosthetic femoral condyles matches that of the concave, mated tibial tracks. These design changes made installation much more reproducable and successful[17]. Unfortunately the congruent fit between the convex femoral and the concave tibial components under even minimal muscular and weight compressive loading caused any shear or tibial axial torional loadings imposed on the joint to be transmitted directly to the bone-cement interfaces (Fig. 9). For the knee as well as the hip, bone cement is capable of resisting high compressive loading but is very vulnerable to shear or torsional disruptive loads. This constraint to prosthetic

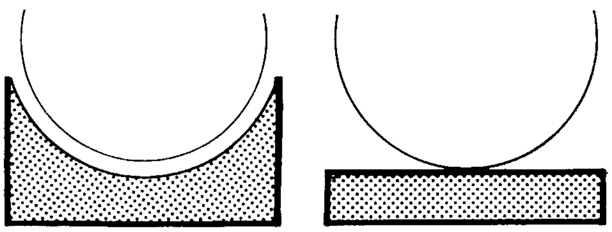

Fig. 9 A congruent fit between the prosthetic parts transfers shear and torsional loads, A while a non-congruent fit allows slippage between the component parts.

component motion has been associated with increased prosthetic loosening rates. Virtually all present resurfacing designs for total knee prostheses are unconstrained by design so that the disruptive shear and torsional loads can be dissipated by the knee's soft tissues rather than by the mechanics of the joint itself. The Oxford knee design[18] has polyethylene menisei that slide freely on flat polished tibial component superior surfaces. This design was specifically developed to manage the described torsional and shear loads.

By the early 70s it was obvious that unsupported thin polyethylene tibial components loosened from the bone far more frequently than did the metal femoral components. The plastic was noted to cold flow or deform under load. Any assymetric loading led to deformation. Two biomechanical approaches to the problem have proved successful. First, increasing the thickness of the plastic tibial component greatly diminished the incidence of tibial component loosening[19]. Another design solution was to support the plastic with metal (Fig. 10). Internal or external metal tibial supports have greatly diminished the problem of cold flow and associated component loosening. Since the widespread availability of modern knee resurfacing components with minimal constraint, metal supported polyethylene, and a patello-femoral articulation, surgical alignment of the limb and selection of and positioning of the components have become the most important biomechanical factors in preventing total knee failure.

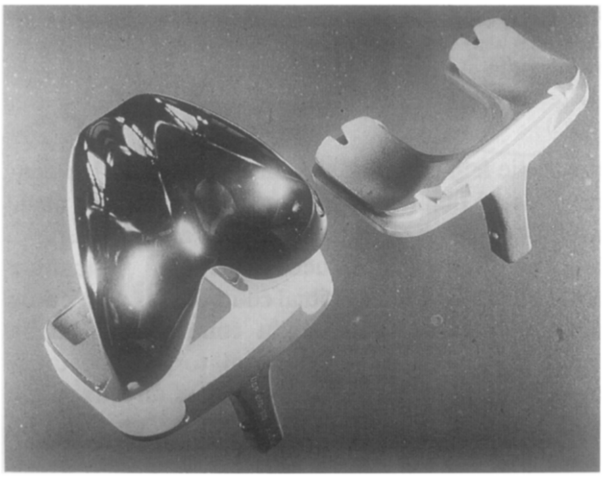

Fig. 10 The polyethylene of this tibial component is supported by an endoskeleton of strong metal.

An arthritic knee with a pre-operative varus deformity which is not corrected at prosthetic arthroplasty will rapidly fail. Several new instrumentation sets have been developed which assure anatomical and correct alignment of the limb segments at the time of total knee arthroplasty. One such system uses a radiograph to determine the centre of the femoral head. A cord is tensed from the centre of the femoral head to the centre of the ankle. Limb alignment is not accepted until the cord passes over the centre of the knee. This assures 5–7° of anatomical valgus. Today limb alignment is the single most important controllable biomechanical factor in total knee arthroplasty success or failure.

It is advantageous to accomplish limb alignment with a combination of femoral and tibial osteotomies guided by appropriate instrumentation. Soft tissue balancing procedures are also essential to obtain a knee that will function well with its prosthetic components. Harmony between the prosthetic components and the restored ligaments of the knee is essential. This envolves selection of correctly sized components. Femoral component placement is such that the axis of rotation of the prosthetic knee approximates the anatomic axis. If for instance the femoral component is installed too far posteriorly, the fully extended knee could appear to have excellent alignment and good tension, but with flexion the collateral ligaments would become excessively tense and fail or they would cause bone compression. This would be very apt to cause total knee failure. Provision of nearly normal, mutual harmonious kinematics of the prosthesis treated knee together with its ligamentous structures is of major importance in any effort to improve durability and success of total knee arthroplasty.

Bone ingrowth fixation of prosthetic components

Because component loosening has been the greatest problem of total joint replacement arthroplasty and because failure seems associated with the qualities and use of bone cement many scientists have tried to develop and design components which can utilise the ingrowth of bone into 100–400 m pores in the surface of orthopaedic prosthetic components. There is little clinical information regarding success of these techniques at the present time. Ingrowth has been successfully demonstrated for total knee femoral components. Unfortunately there is very little direct human data which we can use to validate this usage for tibial component fixation. Likewise the extent of ingrowth into experimental femoral components at total hip arthroplasty has been disappointing. A few more years are necessary for ingrowth fixation to assume its proper role.

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