Before the advent of total joint replacement, patients who had end-stage arthritis of the lower extremities had unremitting pain and a greatly decreased functional capacity. In addition, they often were confined to a wheelchair and were dependent on the care of others. Today, the outcomes of primary total hip and knee replacement are predictable and usually excellent. Prosthetic joint replacement has dramatically improved the lives of millions of people worldwide.

As the fixation of total joint implants has become more reliable and durable and as the technology of total joint replacement has been applied to younger and more active patients, the current limitations of total joint arthroplasty are related to the wear of the components. Wear is the removal of material, with the generated wear particles that occur as a result of the relative motion between two opposing surfaces under load. In complex mechanical-biological systems such as total hip and knee replacements, there can be many types of wear.

A thorough understanding of wear, such as progressive thinning of polyethylene components, can limit the functional life of a joint replacement. The clinical problems from wear are often due to the release of an excessive number of wear particles into a biological environment. When particles within a certain size-range are phagocytized in sufficient amounts, the macrophages enter into an activated state of metabolism, releasing substances that can result in periprosthetic bone resorption. Progressive loss of periprosthetic bone can necessitate a reoperation, which is the definitive measure of clinical failure of a joint arthroplasty.

**Wear Modes**

It is important to distinguish among the fundamental mechanisms of wear (adhesion, abrasion, and fatigue);

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The predominant type of wear of one prosthetic joint can differ from that of another. Furthermore, in a specific joint, there may be different types of wear occurring at different times over the service life of the implant. The damage to an implant is a result of all of the mechanisms of wear that have acted on it over its service life, with the most recent events having the greatest influence.

- **Mode-1** wear results from the motion that is intended to occur between the two primary bearing surfaces, such as the motion of the prosthetic femoral against the polyethylene acetabular bearing surface. **Mode-2** wear refers to the condition of a primary bearing surface that moves against a secondary surface that it is not intended to move against. For example, mode-2 wear occurs when a femoral component penetrates through a modular polyethylene bearing and rubs against the metallic tibial base-plate or acetabular shell. **Mode-3** wear refers to the condition of the primary surfaces as they move against each other but with the interposition of third-body particles. In mode-3 wear, the contaminant particles directly abrade one or both of the primary bearing surfaces. This type of wear is known as three-body abrasion or three-body wear. The primary bearing surfaces may be transiently or permanently roughened by this interaction.

- **Mode-4** wear refers to two secondary (nonprimary) surfaces rubbing together. Examples of mode-4 wear include impingement of the prosthetic femoral neck on the rim of the acetabular component; motion at
the stem-cement or bone-cement interface or relative motion of a porous coating, or other metallic surface, against bone230,232,235,236; relative motion of the external surface of a modular polyethylene component against the metal support (so-called backside wear); fretting between a metallic substrate and a fixation screw104,123,142,166; and fretting and corrosion of modular taper connections201 as well as that of extra-articular sources104,133,218. The particles that are produced by these types of wear may be composed of bone, polymethylmethacrylate, metal alloys, metallic corrosion products, or hydroxyapatite.232,234,235 Wear particles produced by mode-4 wear can induce an inflammatory reaction and can be transported to the bearing surfaces and induce three-body wear (mode-3 wear)104,181,232.

Friction is the resistance to movement between two surfaces in contact. The degree of resistance is proportional to the load. The ratio between frictional force and load (friction/load) is the coefficient of friction (μ)13. Frictional torque is the force created as a result of the friction of bearing Charnley et al. initially selected a stainless steel-on-polytetrafluoroethylene bearing couple because of a low coefficient of friction19,20. The small, 22.25-millimeter-diameter head (subsequently referred to simply as a twenty-two-millimeter head) was selected to minimize the moment arm of the frictional forces and, thus, to minimize the frictional torque (coefficient of friction [μ] × τ2). Thick walls of polyethylene increased the outer diameter of the Charnley socket, which distributed the frictional torque over a large fixation area. Unfortunately, Charnley hip components with the polytetrafluoroethylene bearing uniformly failed because of rapid wear with the release of polytetrafluoroethylene wear particles, formation of granulomas, and loosening of the component.218,219. Thus, despite a sound theoretical premise supported by laboratory investigations, this early clinical experience demonstrated the difficulty of predicting wear resistance in vivo as well as the adverse effects of wear particles.

Contrary to theoretical considerations, frictional torque has not been demonstrated to be important in the initiation of aseptic loosening of either femoral or acetabular components200,202. A accumulating evidence indicates that wear particles have a far greater effect on the durability of the fixation of the implant than does frictional torque. From this perspective, the subsequent success of the Charnley low-friction arthroplasty with a polyethylene acetabular component is primarily a function of the low volumetric wear of the twenty-two-millimeter metal bearing, not of low frictional torque. Bearings with a larger diameter can be successful if the rate of wear is low21,22. This is an important consideration as methods to reduce polyethylene wear, such as extensive cross-linking of polyethylene or utilization of hard-on-hard bearings (such as ceramic-on-ceramic and metal-on-metal), are being investigated. The coefficient of friction for the metal-on-metal bearing of the McKee-Farrar hip prosthesis is roughly two to three times greater than that of the Charnley prosthesis. The larger diameter of the McKee-Farrar bearing couple, which is approximately forty millimeters, amplifies this difference, and as a result the frictional torque is as much as ten times greater than that of the Charnley bearing couple22. This value is still only about one-tenth the static torque to failure of an acetabular component immediately after it has been freshly implanted with cement211,212,213.

**Polyethylene Wear**

Most total hip replacements and effectively all total knee replacements have one primary bearing surface made of ultra-high molecular weight polyethylene (referred to simply as polyethylene). Although there are many potential sources of wear particles, in most total hip and knee replacements the greatest contribution is from mode-1 wear — that is, polyethylene particles generated by the intended motion of the joint at the primary bearing surfaces. For practical purposes, wear of...
the much harder bearing surface of the femoral component is negligible in mode 1. Polyethylene wear is distinct from creep, which is plastic deformation due to loading. Creep contributes to the deformation of a polyethylene bearing but does not produce wear particles. The rate of creep decreases rapidly over time, becoming negligible by the first twelve to eighteen months after implantation\(^\text{113,253}\). Wear accounts for most of the change in the surface of a polyethylene bearing over the longer term\(^\text{113,253}\).

There are a myriad of variables that affect the wear of a polyethylene bearing in vivo, including the wear resistance of the materials as well as the loads, lubrication, sliding distance, motion pattern, specifics of the design and manufacturing of the polyethylene component, implantation techniques, type of wear, and amount and type of use of the joint. The wear resistance of polyethylene is a function of the base resin, the manufacturing, and the method of sterilization of the polyethylene component\(^\text{46-48}\). Oxidation reduces the static strength and elongation properties of polyethylene and decreases the resistance of polyethylene bearings to fatigue, leading to higher in vivo rates of wear. A time-dependent increase in the amount of oxidation can result from gamma irradiation in air, which until recently was the most common method of sterilization in the orthopaedic industry. Gamma irradiation breaks molecular bonds in the long polyethylene chains, giving rise to free radicals. When present, oxygen can combine with these free radicals. Peak levels of oxidation typically occur about one to two millimeters below the surface of a polyethylene component. As the degree of this subsurface oxidation increases, so does the occurrence of fatigue cracking and delamination, as has been observed in retrieved tibial components\(^\text{44}\). Components that have been sterilized less than one year before implantation exhibit lower in vivo oxidation and better in vivo performance than components with a longer so-called shelf life before implantation\(^\text{42}\). In laboratory wear tests, polyethylene that had been sterilized with gamma irradiation in air and had been aged exhibited a higher rate of wear than material that had not been irradiated\(^\text{113,253}\).

When free radicals are formed in polyethylene, such as by gamma irradiation, cross-linking of polyethylene molecules is a competing reaction to oxidation. Cross-linking also changes the material properties of polyethylene, but it can improve the resistance to wear. The relative amount of oxidation and cross-linking varies with the depth from the surface of the component and results in a corresponding variation in the resistance of the material to wear as a function of the depth from the surface. In general, as oxidation increases, cross-linking decreases and vice versa\(^\text{44}\). Methods for controlled cross-linking include the use of chemicals (peroxide), variable-dose gamma irradiation, and electron-beam irradiation. Clinical studies have indicated a substantial reduction in wear associated with cross-linked polyethylene\(^\text{15,253}\). Recent studies with use of a wear simulator have indicated that, with optimum cross-linking, the type of wear that occurs in acetabular cups can be reduced by more than 95 percent\(^\text{15,253}\). However, cross-linking may not have the same degree of benefit with regard to the types of wear that occur in total knee replacement\(^\text{21}\).

Polyethylene contact stresses are a function of the thickness of the component as well as the load and the contact area\(^\text{11}\). A minimum thickness of six millimeters is recommended for a conforming articulation such as a total hip replacement. In general, from a materials standpoint, a thickness of more than six millimeters is preferred for bearings with less conformity\(^\text{21}\). Polyethylene wear is also a function of the motion pattern. In wear tests that use a linear motion path, such as a reciprocating pin-on-disk, the rate of polyethylene wear for a given set of test conditions is ten to one hundred times lower than that in wear tests that use crossing motion paths\(^\text{21}\). Wear tests that use crossing motion paths have been shown to produce more closely the type and amount of wear occurring in vivo in total hip replacements\(^\text{15,151}\).

The clinical manifestations of polyethylene wear in total hip and total knee replacements include the removal of material, which results in progressive penetration of the femoral head into the polyethylene acetabular component and a reduction in the thickness of the polyethylene tibial bearing. A reoperation may be necessary because of thinning or wear-through of a polyethylene bearing or because of frank failure of the component\(^\text{113,253}\). Progressive penetration of the femoral head into the polyethylene acetabular component can lead to impingement between the femoral neck and the socket, which may contribute to loosening of the acetabular component\(^\text{12,254}\). A symmetrical wear of a tibial polyethylene bearing can alter the mechanical axis of the knee and thereby increase the rate of wear in that compartment because of the increased load\(^\text{15,30,257}\).

**Roughness of the Countersurface**

When a joint prosthesis has one polyethylene bearing surface, the other bearing surface is commonly referred to as the countersurface. The base material and the specifics of manufacturing, such as the method of polishing, determine the initial (as-manufactured) surface characteristics of a femoral head or those of a femoral component of a total knee prosthesis. The microtopography of the surface determines its roughness. A contact or noncontact (laser) stylus can be used to scan the surface and make an analog recording of the peaks and valleys over a specified length of the surface. As a result of mode-3 wear, the original surface can be damaged in vivo, resulting in a rougher surface. Increased roughness of the femoral countersurface...
may dramatically accelerate two-body abrasive wear of polyethylene. Experimental studies have indicated that a threefold increase in the roughness of the femoral countersurface can cause at least a tenfold increase in the rate of polyethylene wear. Polyethylene wear is sensitive to the specific type of surface damage. A scratch of two micrometers in depth with an average lip height of one micrometer may not substantially increase the measured average roughness of the surface, but in laboratory wear tests such scratches increased polyethylene wear thirty to seventyfold, depending on the motion pattern. The combined effects of increased roughness of the countersurface and increased oxidation on wear may be synergistic rather than simply additive.

Damage to the countersurface is common and was found on 89 percent of fifty-four cobalt-chromium-based metal alloys and are therefore more resistant to damage by third-body particles than are metal counterparts. However, in clinical comparisons, laboratory tests did not demonstrate a substantial advantage in a ceramic material. Ceramic materials are advantageous. Ceramic materials are harder than stainless-steel and cobalt-based metal alloys, and therefore are more resistant to damage by third-body particles than are metal counterparts. For this reason, the increased hardness of the material. The decreased hardness of titanium alloy, compared with stainless-steel and cobalt-based metal alloys, results in decreased resistance to abrasion. Although the initial surface roughness of a titanium-alloy femoral head may be equivalent to that of other bearing materials, there is a greater potential for its surface roughness to increase in vivo. In an environment with few or no hard third bodies, the wear performance of titanium alloy against polyethylene can be comparable with that of other metals. However, the presence of hard third bodies, such as particles of cement or metal, has a greater adverse effect on the performance of titanium alloy against polyethylene.

Ceramics are harder than stainless-steel and cobalt-based metal alloys and are therefore more resistant to damage by third-body particles than are metal counterparts. For this reason, the increased hardness of ceramic materials is advantageous. Ceramic materials did not demonstrate a substantial advantage in a laboratory joint simulator in which there were few or no third bodies. However, in clinical comparisons in which the operating environment of the articulation is more variable, ceramic heads have demonstrated rates of wear that are lower than those of metallic heads.

**Polyethylene Wear Particles**

Not until the 1990s was it recognized that most polyethylene wear particles are less than one micrometer in size and are produced in very large numbers, even by well-functioning joints. Because the resolution of light microscopy is limited by the wavelength of visible light (0.4 to 0.7 micrometer), objects that are of submicrometer size cannot be clearly seen with a light microscope. With polarized light microscopy, submicrometer polyethylene particles appear as a fine, diffuse, background birefringence in the cytoplasm of macrophages and giant cells. For this reason, polyethylene wear particles may have been present but not appreciated in some early studies of tissue from osteolytic lesions. Techniques have been developed to analyze wear particles generated in vivo by retrieving them from peri-prosthetic tissues. The concentration of wear particles from prosthetic joints can extend into the billions per gram of tissue. A suggested by light microscopic observations, most of these are polyethylene wear particles that have linear dimensions that are less than one micrometer. Wear particles are a function of the type of wear that produces them. For example, a smooth, polished femoral component moving against polyethylene in the absence of third bodies, as in a laboratory joint simulator or in an optimally functioning total joint replacement, produces a visually smooth, so-called burnished or polished-appearing polyethylene surface. This intended type of wear is the source of numerous submicrometer polyethylene wear particles. There are case-to-case differences in the mode and the frequency distribution of these small particles in total hip replacements, and some data have indicated that the distribution is influenced by the type and amount of surface damage of the femoral head.

Differences in the articulating surfaces and motion patterns of total knee replacements as compared with those of total hip replacements have important effects on the wear of the polyethylene. Decreased conformity can result in substantially increased contact stresses that can exceed the yield strength of polyethylene. Furthermore, in a total knee replacement, the motion pattern can include rolling, sliding, and rotation on the same surface. The combination of these factors results in differences in the mechanisms of wear for total knee replacement compared with those for total hip replacement. In total hip replacement, the predominant wear mechanisms appear to involve microadhesion and microrosion with the generation of many polyethylene particles less than one micrometer in length. In contrast, subsurface delamination, pitting, and fatigue cracking, with the release of much larger particles of polyethylene, have been recognized as important mech-
anisms of wear in total knee replacement. These mechanisms result in the visually striking surface damage of some retrieved tibial polyethylene bearings. More variety is seen in the size, shape, and texture of polyethylene particles associated with posterior-cruciate-retaining total knee replacements compared with those associated with total hip replacements. Submicrometer granules are less prevalent in specimens obtained from the tissues around total hip replacements. Larger flake-shaped particles, measuring several micrometers in length and width, are relatively common in association with total knee replacements but are not common in association with total hip replacements. The overall average area of particles generated by total knee replacements has been reported to be about twice that of particles generated by total hip replacements. There are data indicating that specific features of the type of total knee prosthesis can influence the size of the polyethylene particles that are produced. Furthermore, modular total hip and knee components have several surfaces, such as the so-called backside of the polyethylene, which can contribute polyethylene wear particles to the periprosthetic tissues. In one study, tissues adjacent to failed total knee replacements had more particles per gram than those adjacent to failed total hip replacements. However, the knee replacements had a longer service life and there was no difference in the apparent rate of particle deposition.

Polyethylene Wear in Vivo

The clinical assessment of the wear of a polyethylene bearing has traditionally been based on radiographic studies. A's measured on standard radiographs, the degree of penetration of the femoral component into the polyethylene component is the linear wear of the bearing. Because of practical issues related to the alignment and reproduction of x-ray-beam projection, it is more difficult to assess wear routinely on radiographs of total knee replacements than it is on radiographs of total hip replacements. For this reason, there is less information in the literature on the radiographic assessment of wear of total knee replacements.

For total hip replacements, the method of radiographic measurement that is most commonly referenced is a variation of the duoradiographic technique originally described by Charnley and H alley and reported by Livermore et al. On standard anteroposterior radiographs of the pelvis, a compass is used to identify the shortest distance from the center of the femoral head to a reference point on the acetabular cup on the follow-up radiographs. A measurement is then made between the same reference points on the initial postoperative radiograph. After correction for magnification, the difference between the measurements on the initial postoperative and follow-up radiographs is the linear wear, which is conventionally expressed in millimeters. The linear wear rate is then calculated by dividing the linear wear by the duration of time after implantation. The linear wear rate is conventionally expressed as millimeters per year. This method only measures wear that occurs in the plane of the radiograph, and it cannot detect any component of the wear vector that occurs out of the plane of the radiograph. The maximum variation in the measurements made by a single observer with use of the Livermore technique was initially reported to be 0.1 millimeter. However, other investigators have found the variability to be between three and four millimeters.

Electronic calipers can measure to the nearest 0.05 millimeter, and the mean intraobserver error with this device is reported to about 0.08 millimeter.

A more accurate term for this measurement is linear penetration. The term linear wear has historically been used synonymously with the term linear penetration in discussions of polyethylene bearings. However, the measurement of linear penetration includes several factors in addition to wear, which is the removal of material, with the generation of wear particles, from the articulation of the femoral and acetabular components. Linear penetration includes creep (plastic deformation) and wear. The contribution of creep to linear penetration is greatest in the early postoperative period and decreases with time, becoming negligible by twelve to eighteen months. Aitionally, there is an initial running-in of the bearing, which results in better conformity, lower contact stresses, and a lower rate of wear. In longer-term studies, decreasing patient activity over time can result in decreased wear; also, there may be a survival selection for the better-functioning implants (those with lower wear rates). For these reasons, short-term rates of linear penetration are higher than long-term rates. Modularity is another factor that can contribute to linear penetration. Because of issues related to creep, manufacturing tolerances, and the in vivo assembly of modular acetabular components, the acetabular liner may change position relative to the acetabular shell, resulting in a change in the position of the femoral head and an increase in the short-term rate of linear penetration. So-called backside wear of the modular polyethylene liner could also contribute to higher rates of linear penetration.

Volumetric wear is a measure of the amount of material removed from the bearing surface. The simple formula \( v = \pi r^2 w \), in which \( v \) is the volume of the polyethylene bearing, \( r \) is the radius of the femoral head, and \( w \) is the measured linear wear, has commonly been used to calculate volumetric wear. This formula assumes a single, cylindrical wear track, but this assumption is not supported by some retrieval studies.

In the 1990s, techniques for computer-assisted measurement of wear have been developed. Standard radiographs can be digitized to create a computer model of the femoral head and the acetabular component. Use of both anteroposterior and lateral ra-
diographs allows construction of a three-dimensional model. Comparison of serial radiographs gives both the magnitude and the direction of the femoral head displacement. Such computer-assisted techniques can reduce measurement variability that is due to difficulties related to the identification of a single reference point, the angle of the radiographic beam, and positioning of the patient. Edge-detection techniques that infer the margins of the components by evaluating gray-scale intensity on digitized images of radiographs have been developed. These techniques minimize the potential for intraobserver and interobserver variability and should enable more accurate comparisons of results between institutions.

A assuming a negligible contribution by creep in the long term, the wear of retrieved polyethylene acetabular components has commonly been measured with variations of the so-called shadowgraph technique. With this technique, a cast of the acetabular bearing surface is made and the profile (the so-called shadow) of the cast is used to measure the wear track. This method allows determination of the angle of wear relative to the mouth of the component so that a wear vector (magnitude and direction) can be determined. Single-plane radiographic measurements have slightly underestimated the linear wear measured on the corresponding retrieved implants. This may be due, at least in part, to the fact that the wear vector is not consistently in the plane of the radiographs. Additionally, the wear track is not a complete cylinder. Volumetric wear calculated from linear wear with use of a formula for a hemicylinder with a correction factor for the direction of wear is about 47 percent less than that based on the simple cylindrical formula. Multilayered vectors have been identified in 30 percent of one type of polyethylene acetabular component. The vectors do not appear to be a result of loosening of the cup, but impingement between the edge of the cup and the neck of the femoral component may be a factor.

Fluid-displacement methods have also been used to measure the wear of retrieved polyethylene acetabular components. A femoral head of appropriate size is placed into the original articulating contour, and the volume of fluid that is required to fill the remaining contour (the worn area) is measured. The interobserver variability is reportedly within five cubic millimeters, and the accuracy is reportedly within fifteen cubic millimeters.

The amount of wear of hard-on-hard bearing surfaces, such as metal-on-metal or ceramic-on-ceramic, is typically so small that it cannot be measured on routine clinical radiographs. Furthermore, the amount of wear can also be too small to be measured on retrieved specimens with use of the shadowgraph technique. Consequently, computerized coordinate measuring devices have been employed to quantify the amount of wear. These devices can be used to assess the sphericity of the bearing surfaces by comparing the measured dimensions in multiple planes with the best-fit sphere. The volume of wear can be calculated by integrating the depth of multiple individual wear points on the worn surface.

Studies of Wear In Vivo

The rates of polyethylene wear in total hip replacements have been reported in a number of previous studies. The findings of those studies were summarized in table form by one of us [T. P. S.] and colleagues in a previous issue of the Journal of Bone and Joint Surgery. Many variables influence polyethylene wear in vivo and, consequently, the rates are highly variable. These studies have demonstrated substantial variability not only with regard to the average rates of linear wear but also with regard to the ranges (when reported). Regardless of the duration of follow-up, a number of hip replacements have no radiographically measurable wear and some demonstrate wear that is several times the average for the study. These large patient-to-patient variations in rates of wear have not been explained by differences in the wear resistance of the polyethylene. This is not surprising considering the number of variables that contribute to wear in vivo.

Patient-related variables include age, gender, weight, general health, and activity as it relates to the use of the hip prosthesis. Variables related to the hip reconstruction include the implanted materials (including, but not limited to, the polyethylene bearing material); the design and manufacturing of the prosthesis; and the characteristics of the implantation procedure, such as the operative techniques, biomechanical considerations, and the initial as well as the long-term fixation of the implants. These variables are important with respect to wear as they can affect the loads and the motions of the bearing and the degree of three-body-wear mechanisms. There is also variability in wear measurements because of differences in the methods of assessment and limitations of the measurement techniques, as already discussed. For these reasons, the strength of comparisons made between different studies is limited. Furthermore, in cohort studies, the rates of wear do not follow a Gaussian distribution, and appropriate methods should be used in statistical analysis to account for these distributions.

In one study, in which serial radiographs of each hip replacement were analyzed and compared over a minimum of five years, the rate of penetration of the femoral head into the polyethylene liner tended to decrease with time, reaching a steady state after the sixth postoperative year. Despite substantial differences in the rates of penetration over the first three years in subsets of hip replacements with high and low rates of penetration, the rates became similar over time. These results indicate that, for individual patients and prosthetic hip systems, multiple assessments of wear over time are
more valuable than a single measurement\textsuperscript{226} and that caution should be exercised when comparing rates of penetration in hip replacements after different durations of follow-up\textsuperscript{131}.

Theoretical models and retrieval analyses have shown that the rate of volumetric wear of polyethylene components increases with an increase in the diameter of the femoral head. This is because of an increase in the contact surface and because, during the same gait cycle, the sliding distance (the motion of the surface of the femoral head relative to the surface of the socket) increases with an increase in the size of the bearing\textsuperscript{21}. With use of the simple cylindrical formula, $v = \pi r^2 w$, it is shown that for any given amount of linear wear the volumetric wear increases exponentially with increases in the radius of the bearing. In one retrieval study, for each millimeter increase in the diameter of the head, there was an increase of 6.3 cubic millimeters per year in volumetric wear\textsuperscript{21,121}. In another report, the rate of volumetric wear increased from 7.5 to 10 percent for each millimeter increase in the diameter of the head\textsuperscript{21}.

In clinical studies, the linear wear rates of polyethylene against thirty-two-millimeter-diameter femoral heads have been equivalent to or greater than those of polyethylene against heads with a smaller diameter\textsuperscript{115,144}, and they have been associated with bone resorption and loosening of the acetabular component\textsuperscript{175,201,204}. Strictly with respect to wear, heads that have a thirty-two-millimeter diameter do not compare favorably with heads that have a smaller diameter. Because of their large diameters, surface-replacement components have rates of volumetric polyethylene wear that are four to ten times higher than those of conventional total hip replacements with twenty-eight-millimeter heads\textsuperscript{198}. It had been hoped that the reduction in polyethylene stresses due to the larger contact area would result in reduced linear wear, but this has not been the case. Within the range of contact areas for metal-on-polyethylene prosthetic hip bearings, it appears that the stresses in these conforming bearings are all so low that any differences do not appreciably affect wear in vivo. The increased contact area and sliding distance of larger heads result in increased volumetric wear. The relationship between the size of the head and the rate of wear may be confounded by the use of a relatively thin polyethylene component with some thirty-two-millimeter bearings and in surface replacements.

The association between volumetric wear and peri-prosthetic bone resorption is related to the number of polyethylene wear particles that have been released into the so-called effective joint space, which includes all periprosthetic spaces and tissues in communication with joint fluid\textsuperscript{18}. A twenty-eight-millimeter-diameter bearing with a conservative linear wear rate of 0.05 millimeter per year (a volumetric wear rate of about thirty cubic millimeters), with individual wear particles equal in volume to a 0.5-micrometer-diameter sphere, generates a total of 500 billion particles. Assuming that a patient takes one million steps per year, this translates to 500,000 particles per step. Such estimates are very sensitive to particle size. The number of particles produced by a given volume of wear varies with the cube of the diameter of the particle. A single ten-micrometer-diameter spherical particle has the same volume as 8000 0.5-micrometer-diameter particles. There would be only about sixty-three million ten-micrometer-diameter particles for the same amount of volumetric wear (thirty cubic millimeters). At one million steps per year, this translates to only sixty-three wear particles per step\textsuperscript{216}.

Studies of wear particles retrieved from periprosthetic tissues and analyses of worn polyethylene surfaces have demonstrated findings that are consistent with an average particle size in the 0.5-micrometer-diameter range\textsuperscript{18,25,61,48,61,22,214,242}. If the size of the wear particles is constant, then increases in volumetric wear lead to increased numbers of polyethylene wear particles. From a combined mechanical and biological perspective, optimization of in vivo wear requires not only a reduction of wear volume but also a reduction in the generation of the most biologically active wear particles. A lower rate of wear may not necessarily be preferred clinically if a higher number of biologically active wear particles is generated.

Compared with the assessment of wear on the hemispherical bearing of a total hip replacement, the assessment of wear in a total knee replacement is more complex because of differences in the geometry of the numerous femoral and tibial components and because of differences in kinematics. Studies have indicated that the rates of polyethylene wear in total knee replacements are even more variable than the rates of wear in total hip replacements and that they are a function of design, including the conformity of the articulation\textsuperscript{2,13,245,246}, the operative technique, such as the mechanical alignment of the knee and the fixation of the components\textsuperscript{13,22,222}; the presence of abrasive third bodies\textsuperscript{211,146}; the thickness of the polyethylene component; and the polyethylene manufacturing process\textsuperscript{210,12,67,143,245,246}.

The geometry of an articulation can be generally described as a convex surface on a concave surface. In total knee replacement, the degree of conformity between the two surfaces can be described as the ratio of the radius of curvature of the tibial component ($R_t$) to the radius of curvature of the femoral component ($R_f$) or $R_t/R_f$. Such an analysis can be done for both the sagittal and the coronal plane geometry. As the value of this ratio approaches one, the conformity of the articulation increases. Thus, the most conforming articulation (such as in a total hip replacement or a flat-on-flat surface) would have matched radii and a ratio of one. As $R_t$ becomes larger than $R_f$, the conformity decreases and contact stress increases. For a metal-on-polyethylene articulation, the contact stress is roughly doubled when this ratio increases from one to five\textsuperscript{216}. Because the ra-
dius of curvature of a femoral component may not be constant over the entire sagittal profile, the contact stresses for the same load may vary throughout the range of motion of the knee. Constraint is distinct from conformity. Constraint is the restriction of motion. A flat-on-flat articulation is completely conforming and has no constraint to motion in any plane, whereas a dished articulation of matched radii is completely conforming but motion is constrained to one plane.

Condylar designs with a conforming tibiofemoral articulation have large contact areas, lower contact stresses, and more favorable wear characteristics, but they may not allow physiological translational and rotational movements. Relatively flat tibial articulations can accommodate such motions, but they have smaller contact areas, higher contact stresses, and higher rates of wear. Experience indicates that, when an attempt is made to maximize function through design innovations, priority must be given to the material limitations of the prosthetic implants. The apparent dilemma of balancing the goals of conformity and multidirectional motion is addressed by designs with mobile tibial polyethylene bearings. These designs have a high degree of tibiofemoral conformity and allow for rotation (a so-called rotating platform) or rotation combined with anterior-posterior translation (so-called meniscal bearings). Dislocation of the mobile bearings was reported in 2 percent (three of 123 knees) and was associated with instability in flexion and with revision total knee replacement.

Rather than quantifying a change in dimensions, as is done in assessments of wear in total hip replacements, most investigators who have performed retrieval studies of total knee components have analyzed the relative amounts of several types of surface damage, including pitting, scratching, abrasion, burnishing, delamination, and embedded third-body abrasive particles. The degree of surface damage is related to the weight of the patient and the duration of time after implantation. In total knee replacements, peak stresses occur at about one to two millimeters below the surface. The combination of high subsurface stresses and decreased mechanical properties due to oxidation at roughly this same depth predisposes to delamination, which is a frequent type of surface damage in tibial bearings with lower conformity and is associated with rapid rates of wear.

Pitting was the most common type of surface damage seen in association with a conforming total condylar design; only two of forty-eight components showed evidence of delamination. The highest scores for damage were in the central, weight-bearing portions of the medial and lateral tibial concavities. There was greater damage on the anterior aspect than on the posterior aspect, which is probably due to the kinematics of this design. No delamination was seen in the standard tibial polyethylene components of twenty retrieved posterior cruciate-substituting total knee replacements. Burnishing was the most common type of surface damage reported for patellar components. It was roughly equal on the medial and lateral aspects of the patellar component, where contact occurred with the femoral component, and the severity of the damage was related to the weight of the patient. In metal-backed patellar components, the polyethylene may be quite thin and demonstrate rapid wear, fracture, or dissociation from the metal backing.

In a retrieval study of a posterior cruciate ligament-sacrificing total knee replacement with a high degree of tibiofemoral conformity in the sagittal plane and with polyethylene that had a minimum thickness of six millimeters, the rate of linear penetration of the femoral component averaged 0.025 millimeter per year, which is considerably lower than that reported for the Charnley total hip prosthesis. The calculated rate of volumetric wear was sixteen cubic millimeters per year, which is also lower than that reported for the Charnley total hip prosthesis. The relative amount of penetration on the medial side was increased in varus knees and decreased in valgus knees. Symmetrical wear was most likely to occur when the knee was in about 5 degrees of tibiofemoral valgus. In a study of one design of a so-called meniscal-bearing knee replacement, which has a high degree of tibiofemoral conformity in the sagittal plane, bearings retrieved at reoperation because of loosening or dislocation had an average rate of penetration of only 0.026 millimeter per year, which is quite low. The contact area available for load transmission in each tibiofemoral compartment was 5.7 square centimeters, which is larger than that of the Charnley hip prosthesis. Such low rates of polyethylene wear are consistent with a linear motion path.

Higher rates of wear in total knee replacements have been associated with some designs characterized by lower conformity, especially when the tibial polyethylene is less than six millimeters thick and when so-called heat-pressing of the tibial articular surface has been used. Younger, more active, and larger (predominantly male) patients are at risk for accelerated wear in association with these prosthetic variables. This type of rapid wear tends to involve only the medial or lateral compartment, suggesting a role of asymmetrical loading and high localized stresses from a suboptimum mechanical environment. The rate of progression of the femoral component into the polyethylene can be dramatic, as much as one millimeter per year, resulting in wear-through of a four-millimeter-thick component in four years. The type of wear that occurs is predominantly due to subsurface fatigue resulting in delamination (and the production of larger particles of polyethylene). Wear-through or gross mechanical failure of the polyethylene component can result in mode-2 wear with abrasive damage to the femoral component and substantial generation of metal particles.
The clinical triad of effusion, pain, and progressive change in the coronal alignment of the knee (most commonly into varus alignment) is characteristic of accelerated polyethylene wear. The joint fluid is laden with polyethylene particles of various sizes, and aspiration can confirm the diagnosis of polyethylene-induced synovitis. A rhthroscopy can be helpful in assessing the degree of polyethylene wear and the damage of the femoral component as well as in planning a revision operation. A rhthrosopic débridement may provide temporary relief, but most patients who have this triad need a revision.

In clinical studies, wear rates generally have been retrospectively compared on the basis of a specific variable, such as the type of femoral component or the type of fixation (with or without cement). The tremendous number of potentially confounding variables in such clinical studies is a fundamental limitation to this type of comparison. Furthermore, great care should be taken when extrapolating the results to other reconstructions with the same generic variable. An example is the issue of metal-backed acetabular components. One specific type of metal-backed acetabular component inserted with cement has been reported to have an increased rate of polyethylene wear compared with an all-polyethylene component inserted with cement. However, a different type of metal-backed acetabular component has been reported to be associated with a lower rate of polyethylene wear in other studies.

The issue cannot be as simple as the presence or absence of metal backing. Specific details of the design, materials, and manufacturing of the components as well as any differences in the populations of patients, durations of follow-up, measurement techniques, and statistical analyses all must be carefully considered when evaluating and comparing rates of polyethylene wear in vivo.

Clinical rates of wear traditionally have been expressed with use of a denominator of time because of convenience, not accuracy. More appropriately, investigators performing in vitro studies involving laboratory wear simulators have always used the number of loading cycles as a denominator. Similar to the use of a set of automobile tires, the wear of a prosthetic hip or knee is a function of use or the number of cycles and not a function of time in situ. The assumption made in clinical studies is that all patients with a joint replacement have about the same level of activity — that is, the actual use of or the number of cycles on the bearing is about the same — or that any differences average out over a large sample size. The limitations of this assumption must be recognized.

One of us (T. P. S.) and colleagues studied, with use of an electronic digital pedometer, the walking activity of 111 patients who had a total joint replacement. We calculated an average of about 0.9 million cycles for each joint in the lower extremity per year. This number is close to the average of 1.0 million cycles per year proposed by Seedhom et al. on the basis of their study of the gait cycles of nine elderly individuals, on vacation, who did not have a joint prosthesis. The most important finding of our study, however, was not the average but the fact that there was a forty-fivefold difference in the range of gait cycles between the least active and the most active individual. The most active individual averaged 3.2 million cycles per year, about 3.6 times higher than the average. These data indicate that individual differences in activity are a substantial source of variability in rates of wear. A forty-fivefold difference in rates of wear as well as rates that are more than 3.5 times the average can be accounted for by differences in an individual's activity.

A ge was associated with daily walking activity but with a high degree of variability (the standard deviation was 3040 steps per day). Individuals who were less than sixty years of age walked about 30 percent more on the average than those who were sixty years of age or older (p = 0.023). The men walked 28 percent more on the average than the women (p = 0.037), and the men who were less than sixty years old walked 40 percent more on the average than the other individuals (p = 0.011). Thus, the variation in an individual's activity contributes to the variability in rates of wear that is consistently seen in in vivo studies. These results should be considered when analyzing studies of in vivo wear rates that have a denominator of time in situ.

Periprosthetic Bone Loss

Periprosthetic bone loss can occur as a result of a reduction in the load transmitted to bone, so-called stress-shielding. Periprosthetic bone loss also occurs as a result of an inflammatory reaction to small particles, such as those produced by the various wear modes. To varying degrees, both processes occur simultaneously in complex mechanical-biological systems such as joint replacements, and the adverse effects can be additive. Bone with decreased density secondary to stress-shielding may be more susceptible to osteolysis. Relative motion between an implant and bone can cause bone loss through both mechanical and biological mechanisms.

Cellular Mechanisms

The tissue adjacent to total hip and knee prostheses consists of synovial tissue, variably organized and variably vascularized fibrous tissue, lymphocytes (occasionally), and foreign-body inflammatory cells (macrophages and giant cells) that are present roughly in proportion to the number of small particles. Prosthetic particles elicit a cascade of responses at the cellular and tissue levels. The cell whose function is central to the biological reaction to prosthetic wear particles appears to be the macrophage. The mononuclear stem cell, which originates in the bone marrow, is the progenitor for both mononuclear macrophages and osteoclasts. M acrophages phagocytose small wear particles and may...
fuse to form foreign-body multinucleated giant cells, usually in association with larger particles. Osteoblasts and fibroblasts also may be important in the response to wear particles resulting in altered formation of bone and connective tissue. A lthough lymphocytes are occasionally present, their role in the inflammatory reaction is unclear.

Most periprosthetic bone resorption is effected by osteoclasts, but there is evidence that macrophages and foreign-body giant cells are capable of direct, low-grade bone resorption. In vitro studies have indicated that activated macrophages release cytokines, including interleukins and prostaglandins, which play a role in the recruitment and differentiation of cells and stimulate bone resorption, but the specificity of the cytokine response and the regulatory mechanisms have not been defined. Under certain conditions, macrophages appear to directly release interleukin-1 beta and tumor necrosis factor. Several cytokines, including interleukin-1 beta, stimulate osteoclast maturation. A lthough cytokines released by macrophages may directly stimulate bone resorption by osteoclasts, other effects may be mediated by intermediary cells such as fibroblasts or osteoblasts. Matrix metalloproteinases (collagenase, gelatinase, and stromelysin), which are capable of effecting bone resorption, are also produced by interfacial membrane tissue around failed total hip and knee replacements. On the basis of the knowledge of such cellular and biochemical mechanisms of bone resorption, there has been increasing interest in and investigation of pharmacological agents that may modify these cellular responses. Other work indicates that, in addition to bone resorption, there is also a decrease in bone formation in association with periprosthetic osteolysis.

It appears that all of the materials used in total joint replacement are capable of inducing an inflammatory foreign-body reaction if the particles are within a certain size-range and there are enough of them. The bone-resorbing ability of macrophages in vitro is a function of the size, shape, and composition of the particles and it is dose-dependent. It has been previously recognized that there is an upper size limit for particle reactivity, but there may also be a lower size limit. For a given concentration of particles, the stimulatory effect of polyethylene particles in vitro decreases when the particles are larger than about seven micrometers or smaller than about 0.2 micrometer. This information suggests that polyethylene wear particles generated by current de-
with particles of polymethylmethacrylate. This appears to be the origin of the concept of so-called cement disease\textsuperscript{\textsuperscript{106,126}}. Jasty et al. subsequently reported osteolysis in association with polymethylmethacrylate particles adjacent to well fixed total hip replacements that had been inserted with cement\textsuperscript{174}. Polyethylene wear particles were not identified. It was not until endosteal osteolysis was identified around femoral components inserted without cement\textsuperscript{107} that polyethylene, metals, and other materials in particulate form were seriously considered as potential contributors to the inflammatory reaction and associated bone resorption.

In 1990, Anthony et al.\textsuperscript{16} demonstrated a communication between the articulation and the endosteal surface of the femur, through a space between the stem and the cement (as described by Fornasier and Cameron\textsuperscript{64}), and then through a defect in the cement mantle. Particles of metal, cement, and polyethylene were identified in macrophages in these osteolytic lesions. A rhematography of the hip, performed after the patients had exercised the hip joint, demonstrated transfer of contrast material from the femoral-acetabular articular space to the area of the osteolysis. Studies\textsuperscript{15,113,126} of well functioning total hip replacements retrieved at autopsy have indicated that some degree of separation of the femoral stem from the cement mantle, so-called debonding, occurs frequently and as early as two weeks after implantation. Fractures of the cement mantle were seen in association with debonding and had originated at the metal-cement interface, most often at the corners of the mantle or in association with pores in the cement. Cement fractures most frequently occurred in areas where the cement was thin or in association with frank defects in the mantle.

A common finding in patients who have femoral endosteal osteolysis around a cemented component appears to be a defect in the cement mantle. The risk of osteolysis can be decreased when technical errors are avoided\textsuperscript{114}. Separation of the stem from the cement mantle does not lead to localized endosteal osteolysis if the cement mantle remains intact, and such separation appears to be compatible with satisfactory long-term function of certain types of femoral components\textsuperscript{15,126} but not others\textsuperscript{15,114}. The relationships between osteolysis and the geometry of the stem, the surface finish of the stem, and the operative techniques have continued to be investigated\textsuperscript{81,149,174,241,244}.

Compared with femoral components that have been inserted with cement, femoral components that have been inserted without cement and that have a limited proximal porous coating can be associated with earlier, more frequent localized endosteal bone resorption in the femoral diaphysis\textsuperscript{106,126}. The size of the lesions can increase with time as can the prevalence, which has exceeded 30 percent in some reports\textsuperscript{126,142,127}. This appears to be due to the fact that a limited proximal porous coating, especially if it is not circumferential, allows joint fluid and wear particles relatively easy access to the diaphyseal endosteum\textsuperscript{116,212}. Diaphyseal endosteal osteolysis has not been reported in association with extensive, circumferential, porous coating even in the presence of substantial stress-shielding\textsuperscript{24,64}.

Inflammatory bone resorption (induced by small particles) also occurs in a more linear pattern, which may progress along the cement-bone interface and cause or contribute to loosening of the implant, especially in the acetabulum\textsuperscript{112,126,156,212}. There may be mechanical factors that contribute to interfacial particle transport. Once stability has been lost, additional motion can only be detrimental\textsuperscript{121}. Even with the relatively low rate of volumetric wear of the Charnley total hip prosthesis, accumulated generation of wear particles eventually decreases the stability of the acetabular implant. A highly significant association (p < 0.01) has been demonstrated between the depth or rate of polyethylene wear and loosening of the acetabular component\textsuperscript{15,112,212,252}. A high rate of polyethylene wear precludes a long service life for the implant\textsuperscript{121}.

A similar process of linear osteolysis can occur in association with acetabular components inserted without cement\textsuperscript{108,129,129,212,129,240,244}. The integrity of the peripheral implant-bone interface governs the ingress of joint fluid and wear particles. Progressive radiolucent lines are more likely to develop adjacent to components with initial peripheral interface gaps than they are likely to develop adjacent to components without such initial peripheral gaps\textsuperscript{15,126,129,156,212}. Tissues from the peripheral acetabular implant-bone interface contain macrophages laden with small polyethylene wear particles\textsuperscript{112}. Insertion of acetabular components with a tight peripheral press-fit has led to a reduced prevalence of initial peripheral interface gaps, a reduced prevalence of progressive peripheral interface radiolucency, and a higher prevalence of hip replacements without any interface radiolucency\textsuperscript{15,16}. Regardless of other design features, a similar process of inflammatory bone resorption can occur in the proximal part of the femur after hip replacement with or without cement\textsuperscript{108,129,129,156,212}. This process can cause or contribute to the loosening of femoral components inserted with cement\textsuperscript{14}. Inflammatory bone resorption may play a role in loosening after previous rigid fixation by bone ingrowth, of a component inserted without cement and with a limited proximal porous coating\textsuperscript{112}. This same process has not resulted in loosening of a component with an extensive porous coating\textsuperscript{7}.

A cementless component inserted without cement have a lower prevalence of interface radiolucency than components inserted with cement\textsuperscript{23,24,34}. However, the patterns of bone resorption around acetabular components inserted without cement are different than those seen around components inserted with cement. The bone loss associated with cementless components inserted with cement typically occurs predominantly along the interface, following the contours of the cement mantle. The bone loss associated with cementless components inserted without cement typically pro-
gresses away from the interface into the cancellous bone of the pelvis, resulting in localized bone resorption, the classic nonlinear or expansile form of osteolysis. A though less common, this form of pelvic osteolysis can also occur in association with acetabular components inserted with cement. Pelvic osteolysis is associated with a younger age, vertical positioning of the acetabular component, and high volumetric wear of the polyethylene. Substantial bone resorption around well fixed components may be asymptomatic until a pelvic fracture occurs. For this reason, annual evaluations with radiographs made in multiple planes are recommended.

Other concerns associated with acetabular components inserted without cement are wear of the convex surface of the modular polyethylene liner against the metal shell (so-called backside wear) and fretting of the fixation screws placed through the shell. On the basis of reports of several series of acetabular components inserted without cement, no consistent relationship has been demonstrated between the presence of screw-holes and the development of osteolysis behind the component. Furthermore, particle access by way of screw-holes cannot be the cause of isolated osteolysis in the ischium and pubis. It appears that an interplay of specific design and implantation variables is involved in the development of pelvic osteolysis.

Osteolysis in Association with Total Knee Replacement

Classic, expansile osteolysis associated with total knee replacements in which all components were inserted with cement has been generally unrecognized in the literature despite the fact that retrieval studies have demonstrated substantial wear of the tibial polyethylene. A though uncommon, when osteolysis has occurred in association with components inserted with cement, it has generally been associated with a flat tibial articulation. Long-term follow-up studies have not indicated that inflammatory bone resorption, either classic osteolysis or interfacial bone resorption, is a frequent cause of failure of cemented total knee replacements with a more conforming articulation.

Osteolysis necessitating reoperation has developed in association with several designs of total knee replacements without cement. Osteolysis has been reported in the proximal part of the tibia in as many as 16 percent (twenty-seven) of 174 patients, in the distal part of the femur in 11 percent (thirty) of 271 patients, and in the patella in 80 percent (twenty-four) of thirty patients who had a prosthesis implanted without cement. Radiographs may underrepresent the true extent of the bone loss. The development of osteolysis is not a direct function of the absence of cement but is related to other design, operative technique, and patient-related variables that are associated with so-called first-generation total knee replacements without cement. These variables include the mechanism of attachment of the modular polyethylene insert to the metal base-plate and the interface between the polyethylene and the metal base-plate as well as the design and location of holes in the base-plate and the presence of fixation screws. With limited distribution of porous coating, wear particles can enter the implant-bone interface and metaphyseal bone by way of the unbonded interface between the smooth metal and bone. Fully porous-coated components are associated with a lower prevalence of osteolysis than those with a less extensive or discontinuous porous coating. In addition, access to tibial metaphyseal bone may be gained by means of holes through the tibial base-plate or along the course of tibial fixation screws. Increased rates of polyethylene wear have been associated with fixation without cement due to the use of thin polyethylene components with low tibiofemoral conformity and metal-backed patellar components as well as with a younger age, male gender, and greater body weight. In aggregate, this information indicates that the fundamental variables for the long-term success of a total hip or knee arthroplasty include the attainment and maintenance of the integrity of the implant-bone interface, which provides not only mechanical stability but also a barrier to joint fluid and particulate debris, and a low rate of generation of biologically active particles from all sources. These findings illustrate the interplay between fixation and wear, and they show that many factors, including the materials and design of the implants, the operative techniques, and the amount and type of the patient's activity contribute to the long-term success (or failure) of a total joint replacement.

The Effective Joint Space

In 1977, Willert reported his classic observations after reoperations for the removal of failed joint replacements. He found that capsular tissue has some capacity to transport particles through the lymphatic system by means of perivascular lymph spaces, leading to regional and systemic distribution. If the capacity for elimination by this mechanism is exceeded, then particles accumulate in the periaricular tissues. The pseudocapsule appears to be the primary location for phagocytosis of particles. This process results in the development of foreign-body granulomas with areas of necrosis and fibrosis that are somewhat proportional to the amount of particles. In principle, the whole environment of the implant could become involved. Extension of this foreign-body response into the cement-bone interface could cause loosening of the implant. The findings from subsequent experimental models and analyses of specimens from well-functioning implants retrieved at autopsy are in accord with this principle. A side from the observations that particles larger than about ten micrometers in linear dimension tend to remain localized and that rapid
wear with the production of an extreme amount of particles is associated with systemic distribution, the variables that influence local accumulation compared with systemic distribution of wear particles have not been defined.

Polyethylene wear particles are dispersed in the fluid around a prosthetic joint. Conceptually, the effective joint space includes all periprosthetic regions that are accessible to joint fluid. The operative implantation procedure alters the natural anatomy of a joint. In prosthetic total hip and knee replacement, some bone as well as the implant-bone interfaces are exposed within the new joint space. Contraction of muscles, such as the psoas and abductor muscles, and changes in joint position (flexion, extension, abduction, adduction, and rotation) can alter the volume of the effective joint space, resulting in changes in intracapsular joint-fluid pressure.

Variations in the pressure of the joint fluid play a role in transporting the joint fluid and wear particles around the effective joint space. Hendrix et al. reported large variations in the pressure of the intracapsu-

Diagram showing the effective joint space, which includes all periprosthetic regions that are accessible to joint fluid. The operative implantation procedure alters the natural anatomy of a joint. In prosthetic total hip and knee replacement, some bone as well as the implant-bone interfaces are exposed within the new joint space. Contraction of muscles, such as the psoas and abductor muscles, and changes in joint position (flexion, extension, abduction, adduction, and rotation) can alter the volume of the effective joint space, resulting in changes in intracapsular joint-fluid pressure.

Diagram showing wear and the release of wear particles into the effective joint space. Capsular tissue has some capacity to transport particles through the lymphatic system (large straight arrow). In the effective joint space, joint fluid and wear particles follow the path of least resistance, which is dependent on the prosthetic and anatomical details of each specific reconstruction. The pseudocapsule becomes thickened (small straight arrows) because of phagocytosis of wear particles and the development of foreign-body granulomas. The effective joint space can extend along interfacial planes, expand into bone, or expand into soft tissues or a variety of combinations is possible (curved arrows).
lar fluid around total hip replacements during activities of daily living, with peak pressures of more than 750 millimeters of mercury (100 kilopascals). During reoperations for osteolysis, intracapsular pressures of more than 500 millimeters of mercury (sixty-seven kilopascals) have been measured. Fluid pressure of 198 millimeters of mercury (twenty-six kilopascals) has been measured in a femoral diaphyseal lesion. In addition to moving joint fluid around the effective joint space, the fluid pressure may contribute to expansion of the effective joint space. The expansive and balloon-like nature of classic osteolytic lesions may be a consequence of fluid pressure. The recorded pressures are high enough to interfere with the normal perfusion and oxygenation of bone, and a physical effect of pressure may account for cases of osteolysis in which no prosthetic particles are identified in the lytic lesions.

In normal synovial joints, such as the hip or knee, bone is not exposed to joint fluid. The boundaries of the joint are defined by the capsule and, within the capsule, bone is covered by cartilage or synovial tissue. In disease processes such as osteoarthritis, and in total joint replacement, the normal anatomical and physiological compartmentalization of a synovial joint can be disrupted. Osteoarthritic cysts, or so-called geodes, are a form of periarticular bone resorption that occurs in the absence of prosthetic implants. Elevated intracapsular fluid pressure can lead to intrusion of joint fluid into cancellous bone through gaps in degenerated articular cartilage. Numerous investigations have shown evidence of a role for intrusion of joint fluid and fluid pressure in the enlargement of these cysts. Intra-capsular fluid pressure is a function of many variables, including capsular compliance, fluid compartmentalization, joint position, and muscle action. The compliance of the osteoarthritic joint capsule is reduced. Fluid pressures are increased in osteoarthritis and vary with use of the joint. In addition, these pressures can be transmitted to periarticular cysts. A difference in the etiology of geodes and that of osteolysis associated with total joint arthroplasty is activated macrophages; geodes develop without this foreign-body response to prosthetic particles but demonstrate a cytokine profile similar to that of the osteolytic lesions around total joint implants. The effective joint space can expand into soft tissue as well as bone. Increased intracapsular fluid pressures can be painful and may result in the formation of synovial cysts or rupture of the capsule. The presence of an effusion or synovial cysts may be apparent on examination of the knee because of the relatively limited amount of overlying soft tissues. Such evidence of fluid accumulation is less often appreciated in the hip. Synovial cysts may have a protective function by accommodating fluid volume and limiting the increase in intra-articular pressure, thus protecting bone from pressure damage. An inverse relationship has been demonstrated between the development of periarticular bone cysts and the development of synovial cysts. The natural hip joint can communicate with the iliopsoas bursa. Pressure-driven synovial fluid may be pumped into the bursa, causing distension that may be symptomatic. Similar events can occur after total hip and knee arthroplasty. In the effective joint space, joint fluid seeks the path of least resistance; the path is variable depending on the anatomical and physiological specifics of the reconstruction and includes soft tissue as well as bone.

Thus, after total hip or knee replacement, the fluid in the effective joint space has at least three components that can contribute to periprosthetic bone resorption: wear particles, soluble factors, and the physical effects of fluid pressure. Wear particles are released into joint fluid and are variably distributed throughout the effective joint space (Figs. 1-A and 1-B). These particles cause inflammation and contribute to the development of an effusion. Chronic inflammation causes fibrosis and decreased capsular compliance, which can result in elevations of intracapsular joint-fluid pressures. The flow of joint fluid is an active process driven by fluctuations in joint-fluid pressures that occur with activities of daily living and can contribute to progressive expansion of the effective joint space. This fluid also contains various soluble factors that are capable of effecting bone resorption at distant sites. Fluid pressure can cause bone erosion from a purely physical effect or necrosis due to circulatory compromise. In contrast to osteoarthritic joints, in which the progression of the osteoarthritis limits the use of the joint, a well fixed and otherwise well functioning prosthetic joint may be asymptomatic; the patient remains active and maintains the driving forces for osteolysis, which may account for the progression and extreme size of some osteolytic lesions associated with total joint replacement.

Overview

In summary, improving the durability of total hip and knee replacements requires a reduction in the total production and release of small particles into the biological environment. There is a need not only for more wear-resistant bearing materials to decrease mode-1 wear but also for concomitant improvements in the design and manufacturing of the implant and the operative techniques to minimize the occurrence of mode-2, 3, and 4 wear. A further limitation of current prosthetic joint arthroplasty is the variable disruption of the anatomy and physiology of the joint that occurs as a result of the prosthetic implantation and that may allow variability in the access of joint fluid (and wear particles) to bone. Although a reduction in the production of particles is desirable, limiting access of joint fluid to bone is necessary to decrease periarticular inflammatory bone resorption.
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