CHAPTER ELEVEN

Friction and wear

When two bodies are placed in contact and then caused to move relative to each other, we find that a force is required to maintain a constant velocity. This is contrary to our ideas of equilibrium: we expect that, once set in motion, the bodies will continue to move with respect to each other at a constant velocity. To preserve this idea, a new force concept, called the frictional resultant or restraining force, was created. As we will see, this force results from interactions between the bodies.

Friction dissipates energy, primarily as heat but sometimes as noise (“squeaking”), and the deficit must be made up continually to maintain motion. Sometimes friction is desirable, as in producing “traction” during walking or retention of bone screws. Sometimes it is necessary to regulate frictional forces, as in braking an automobile when too large a force would result in a precipitous stop, whereas too small a force would produce inefficient stopping and perhaps result in a collision. In other cases, as in designing total joint replacements, it is necessary to reduce friction to as low a value as possible, since such forces might contribute to bearing surface damage and loosening.

Accompanying the presence of frictional forces between two sliding surfaces are the processes of wear. Material interactions between contacting surfaces may produce transfer of matter from one body to another or create small particles called wear debris. It is usually desirable to keep wear at an absolute minimum, both to preserve the properties of the surfaces involved and to minimize the production of wear debris. This is the case in total joint prostheses. However, there are isolated examples in which controlled wear acts to preserve properties, as in the self-sharpening of teeth of the beaver.

Friction and wear may be controlled by introducing additional materials to change the nature of the interface between sliding bodies. These materials, which may be solids, liquids, or gases, are collectively termed lubricants. In the natural joint, components of synovial fluid act as lubricants. The exact nature of the effects of lubricants on friction and on wear is not clear and remains an active target of research in biologic and engineering systems.
Friction

If two bodies are placed in contact under a normal force $W$, both chemical and mechanical interactions may occur. This produces a net attraction between the surfaces. If we now try to slide one surface over the other, we find that there is an initial force, $F_i$, needed to start motion, and then to maintain a constant velocity, a smaller steady-state force, $F_s$, is required.

Both of these forces are related to the interfacial load:

$$F_i = \mu_S \cdot W$$

$$F_s = \mu_D \cdot W$$

where $\mu_S$ and $\mu_D$ are called the static and dynamic coefficients of friction. It is usual for $\mu_D < \mu_S$; thus, it is easier to maintain motion in the presence of friction than to initiate it.

PROBLEM 11.1
Find the values of $\mu_S$ and $\mu_D$ from Figure 11.1.

ANSWER:
The force perpendicular to the interface is the gravitational force acting on 5 kg of mass, which equals $9.8 \times 5$ or 49 N. Thus, $\mu_S = 15/49$ or 0.31, whereas $\mu_D = 10/49$ or 0.20.

Thorough investigations of frictional phenomena have resulted in the recognition of more or less two universal rules:

1. The frictional coefficients depend on load, $W$, rather than upon stress; that is, they are independent of the area of the interface (this statement is sometimes referred to as the “law of friction”).
2. Over broad ranges of interfacial loads, the coefficients of friction are linear; that is, they are independent of $W$.

Origin of frictional forces
Frictional forces depend on both the roughness of the opposed surfaces and their chemical composition. Of these two effects, the role of roughness is the easier to understand. Surfaces show a wide range of roughnesses, defined as the average height of local elevations or “asperities”
Friction and Wear

above the average level. Table 11.1 gives some typical ranges of values for surfaces of manufactured parts, in comparison with a natural bearing surface, normal articular cartilage.

Frictional forces in everyday applications arise directly from surface roughness. The existence of these asperities prevents surfaces from actually coming into contact. When it is believed that contact has occurred, what has actually happened is that only some asperities on each surface have come into contact (Figure 11.2). The true contact area, $A_T$, is between 0.001% and 1% of the overall surface of the interface ($A$) (this percentage has been exaggerated in Figure 11.2 for clarity of illustration). As a result, modest forces can easily produce at the tips of asperities stresses that exceed yield and ultimate strengths. Bonding, secondary to diffusion, encouraged by high local stresses, occurs at these locations. Thus, the initial force exerted to produce relative motion must act to disrupt physical junctions between the surfaces.

<table>
<thead>
<tr>
<th>Type of surface</th>
<th>Surface roughness maximum asperity height (μm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Saw cut</td>
<td>3–25</td>
</tr>
<tr>
<td>Inside of drilled hole</td>
<td>1–12</td>
</tr>
<tr>
<td>Lathe turned surface</td>
<td>0.4–3</td>
</tr>
<tr>
<td>Reamed</td>
<td>0.75–3</td>
</tr>
<tr>
<td>Lapped (150 grit)</td>
<td>0.5</td>
</tr>
<tr>
<td>Lapped (600 grit)</td>
<td>0.1</td>
</tr>
<tr>
<td>Optically polished</td>
<td>0.005–0.1</td>
</tr>
<tr>
<td>Articular cartilage</td>
<td>0.02–0.2</td>
</tr>
<tr>
<td>Alumina(^a)</td>
<td>0.004</td>
</tr>
<tr>
<td>Polyethylene(^a)</td>
<td>1</td>
</tr>
<tr>
<td>Metal(^a)</td>
<td>0.02</td>
</tr>
</tbody>
</table>


**FIGURE 11.2** Effect of load on surface contact.
Chemical reaction between the two materials will tend to promote failure, during motion, through the weaker of the materials. A small increase in surface roughness can have a large effect on wear. Experiments have shown that increasing the surface roughness of a femoral component by a factor of 3 can lead to a 10-fold increase in polyethylene wear in a metal-on-polymer (MOP) total hip replacement (THR).

It can be shown that the value of \( A_T/A \) for materials that can undergo plastic deformation is given by the ratio of the load across the interface to the hardness, \( p \), of the softer of the two materials involved. As a result, the force, \( F_i \), required to initiate motion is simply the shear strength of either the interface or the weaker of the two materials, \( \sigma_{u(sh)} \), times the area of the true area of the interface. Thus, the static friction coefficient is given by

\[
\mu_s = \frac{F_i}{W} = \frac{\sigma_{u(sh)}}{p}
\]

The ratio of shear strength to hardness of materials (which depends on \( \sigma_{u(t)} \)) tends to vary only over a very narrow range; thus, static coefficients lie between 0.1 and 1 for the vast majority of material pairs. Dynamic frictional coefficients are smaller than static ones since less deformation of asperities can take place, thus producing a smaller value of \( A_T \). Table 11.2 gives values for some combinations of materials. Note that frictional coefficients depend on three factors: the two opposing materials and the nature of the lubricant (if present). Since frictional coefficients are ratios of forces, they are dimensionless.

<table>
<thead>
<tr>
<th>Material combination</th>
<th>Coefficients</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>( \mu_s )</td>
</tr>
<tr>
<td>Rubber tire/concrete (dry)</td>
<td>1.0</td>
</tr>
<tr>
<td>Rubber tire/concrete (water)</td>
<td>0.7</td>
</tr>
<tr>
<td>Leather/wood</td>
<td>0.5</td>
</tr>
<tr>
<td>Steel/steel</td>
<td></td>
</tr>
<tr>
<td>Co–Cr/Co–Cr (saline)</td>
<td></td>
</tr>
<tr>
<td>UHMWPE/steel (serum)</td>
<td>0.35</td>
</tr>
<tr>
<td>UHMWPE/steel (synovial fluid)</td>
<td>0.07</td>
</tr>
<tr>
<td>UHMWPE/Co–Cr (serum)</td>
<td></td>
</tr>
<tr>
<td>UHMWPE/Ti6Al4V (serum)</td>
<td></td>
</tr>
<tr>
<td>Al₂O₃/Al₂O₃ (saline)</td>
<td></td>
</tr>
<tr>
<td>UHMWPE/Al₂O₃ (saline)</td>
<td></td>
</tr>
<tr>
<td>Hip joint (natural) (saline)</td>
<td></td>
</tr>
<tr>
<td>Hip joint (natural) (synovial fluid)</td>
<td></td>
</tr>
</tbody>
</table>
Lubrication

In Table 11.2, we see the effects of lubrication. The pattern that can be seen here is that the introduction of a fluid between the sliding surfaces reduces the values of both \( \mu_S \) and \( \mu_D \). This is the basic idea of lubrication: to interpose a material between two solids to minimize interaction between them. It is desired that the lubricant have a low shear strength such that the shearing motion will take place within the lubricant film rather than through asperities on either surface. However, fluids with low shear strength (low viscosity) can be squeezed out from between the surfaces, so that lubrication becomes a very complex problem.

Recent research has shed more light on the role of proteins in the lubrication process. Historically, it is thought that lubrication theory is based on an understanding of simple continuum fluids and does not capture important mechanisms of protein film formation, encountered either in vivo or in vitro with biological lubricants such as serum. Synovial fluid is a viscoelastic shear-thinning fluid that is assumed to be Newtonian (constant viscosity) at physiologic shear rates, though its behavior is actually non-Newtonian over a broader range. In terms of composition, synovial fluid is composed of a mixture of large, surface active molecules, consisting mostly of proteins, hyaluronic acid, and phospholipids. Proteins in synovial fluid are thought to be subjected to shear-induced aggregation and surface deposition at the inlet region of contacting surfaces, participating as an active agent in boundary film formation.

This film is characterized by a multimolecular layer with complex time-dependent behavior. First, protein molecule adsorption is directly responsible for the formation of a thin adherent film of 10 to 20 \( \mu \)m on contacting surfaces. The initial solid film is augmented by a thicker, hydrodynamically generated film of high-viscosity “gel-like” material. Clearly, our understanding of wear is dependent on the nature of synovial fluid lubricant formed during gait process. At the start of gait, the multimolecular film thickness is increased in comparison to that suggested by classic elastrohydrodynamic theory. Film thickness drops at higher speeds, due to some breakdown of the high-viscosity material, though it is unclear at this point why this occurs. Practically, because of the high conformance of mating surfaces in orthopaedic applications such as hip replacement, it is suspected that the escape of protein agglomerations from entrainment will occur less than in laboratory studies. Osteoarthritis and rheumatoid arthritis affect the chemical and physical makeup of synovial fluid, generally decreasing its viscosity and altering the pH and protein content. Protein content may even increase in periprosthetic synovial fluid. The conclusion from these recent findings is that classic lubrication theory may underestimate the thickness of the hydrodynamic film and the subsequent local reduction of wear in vivo. It is noted that these films are sensitive to contact pressure, may be disrupted during initiation of motion, and may be destroyed if pressure during motion exceeds a certain tolerance.
Fluid entrainment occurs when articulating surfaces drag lubricant into the contact region.

The thickness of the film between the two surfaces, \( h \), is proportional to a parameter \( \phi \), sometimes referred to as the Sommerfeld number, which is given by

\[
\phi = \eta \cdot \frac{V}{P}
\]

where \( \eta \) is the viscosity of the lubricant, \( V \) is the relative velocity of the two surfaces, and \( P \) is the stress across the interface, seen as a pressure in the lubricant phase. (It should be remembered that \( \eta \) may be a function of \( V \); see Chapter 4) This parameter affects the mechanism of lubrication, through its effect on \( h \) (see below), and, indirectly, the dynamic coefficient of friction, as seen in Figure 11.3.

Fluid film thickness is also considered in relation to the respective mating surface roughnesses when incorporated into a \( \lambda \) ratio, which is the ratio of the minimum predicted film thickness to the combined surface roughness of the mating surfaces \( (R_a) \).

\[
\lambda = \frac{h_{\text{min}}}{\sqrt{(R_a)_{\text{head}}^2 + (R_a)_{\text{cup}}^2}}
\]

While this equation is tailored specifically for hip replacement components, the principle can be adapted for alternate mating surfaces as long as geometric details of the surfaces can be parameterized appropriately.

Figure 11.4 is a key to interpreting the relationship between friction and film thickness. When film is sufficiently thick \( (\lambda > 3) \), asperity contact will be prevented, whereas a thinner film \( (\lambda < 1) \) will lead to a boundary lubrication environment.

This analysis assumes that the surfaces remain in contact and only move parallel to each other, thus shearing the lubrication film.
Displacement perpendicular to the interface of as little as $h_{\text{min}}$ as possible in lateralization or microseparation can disrupt the lubricating film and introduce impact and fatigue wear, replacing the normal mechanisms of lubrication (discussed in the next section).

There are four principal lubrication mechanisms or regimes that depend on the nature of the relative motion and on the type of the lubricant used (Figure 11.5).

**Hydrodynamic or Fluid Film.** This is the lubrication mechanism present in most engineering applications, such as rotating bearings in electric motors, internal combustion engines, and turbines. As motion of the surfaces occurs, sufficient pressure results in the lubricant that the surfaces are completely isolated from each other and all shear deformation occurs in the film between them. If intermediate viscosity lubricants are used with very smooth (lapped, etc.) surfaces, the dynamic coefficient of friction and the wear rate are both

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**FIGURE 11.4** (See color insert.) Mechanisms of lubrication as dependent on film thickness and friction. (Adapted from Stewart, T.D., *Orthop Trauma* 26: 435–440, 2010.)

**FIGURE 11.5** (See color insert.) Principal mechanisms of lubrication.
small. Because of their extreme hardness, ceramic components can be polished to a fine surface finish that will facilitate their potential for fluid film lubrication during walking.

**Boundary.** The other extreme occurs at low values of $\phi$ that permit surface–surface contact. Lubricant films are very thin, on the order of the asperity height, and lubrication occurs not by lubricant shear but by modification of surface properties. For a material, either liquid or solid, to be a good boundary lubricant, it is necessary for it to interact with the surface. Long-chain molecules with a chemically active region, such as fatty acids, are highly effective in this respect, reacting with surfaces to form soaps. However, both frictional coefficients and wear rates are relatively high, compared with those in other lubrication regimes. Because of the relatively high roughness of polyethylene, traditional MOP arthroplasty contact articulations fall into the boundary lubrication regime.

**Elastohydrodynamic.** The transition from hydrodynamic to boundary lubrication is not abrupt but passes through two other regimes. In the first of these regimes, the elasto-hydrodynamic, the lubricant prevents the surfaces from interacting directly, but pressure waves conducted through it can produce elastic deformations in one surface opposite asperities in the other. This permits maintenance of a thicker lubricant film than would be otherwise expected. The result is excellent lubrication, with slightly higher $\mu_D$ than in pure hydrodynamic lubrication, and with relatively low wear rates.

**Mixed.** At still lower values of $\phi$, the lubricant film becomes discontinuous and a mixed regimen of elasto-hydrodynamic and boundary lubrication occurs. This regimen is not terribly efficient and is usually accompanied by higher wear rates than the other three. Traditional metal-on-metal bearings typically fall into the mixed lubrication category.

In addition to these four classical lubrication regimes, there are additional important lubrication processes that may occur in special cases (Figure 11.6).

**Hydrostatic.** In some engineering applications, an external pressure source is provided to maintain $h$ above some critical limit, thus producing lower values of $\mu_D$.

**Weeping.** If the bearing surfaces are porous and deformable, relative motion may squeeze additional lubricant out of the surfaces into the separating film. This is a highly efficient mechanism and is believed to be a contributing factor to the low coefficients of friction observed in articular cartilage-lined joints.*

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*The mechanisms of lubrication in natural joints are still a matter of some debate. The present consensus is that it is a combination of boundary, elasto-hydrodynamic, and weeping lubrication. Lubrication in replaced joints is primarily elasto-hydrodynamic, with the lubricant, produced by regenerating synovial tissues, resembling normal joint fluid.
Squeeze film. Structured surfaces may be used to hinder the lateral flow of lubricant out from the sliding interface under the influence of the applied pressure. This produces a dynamic version of hydrostatic lubrication, without the need for an external pressure source. Squeeze film lubrication may be assisted in the natural joint by deformation of the cartilage, which will restrict fluid movement out of the contact region.

Wear

Normal wear

Because of the contact of surface asperities during relative motion, it is usual for one or more of the materials to be eroded or worn away. This has unfortunate consequences both to the material, since its original shape and size are changed, and to the surrounding biologic system through the production of wear debris.

Such debris are produced by all material pairs used in total joint replacement (TJR) articulations; they may also be produced in other multipart devices. After an initial high wear rate period, called the wearing in (or running in) period, the quantity of wear debris produced, usually measured as a volume, is given by the relationship (Figure 11.7)

\[ V = k \cdot F \cdot x \]

where \( k \) is a constant characteristic of the material pair making up the articulation and its local environment, \( F \) is the force across the articulation, and \( x \) is the distance of relative travel between the articulating faces. It should be noted with care that wear depends on interfacial force, not on interfacial stress; this is a result of the true contact area increasing with load while the true stress (greater than the geometric stress) actually stays nearly constant (≈\( \sigma_y \)).
PROBLEM 11.2

If a Co–Cr coupled with ultrahigh-molecular-weight polyethylene (UHMWPE) THR (28 mm head diameter) retrieved after 5 years in vivo shows a total surface recession of 0.75 mm (Figure 11.8), find (A) the annual volume of wear debris, (B) the number of wear particles produced per year (assume spherical particles 5 μm in diameter), and (C) the appropriate value of the wear constant ($k$). (Neglect creep in this calculation.)

ANSWER:

A. Wear volume

$$V = 2\pi r^2 \times \frac{0.75}{5}$$

$$= 185 \text{ mm}^3 = 1.85 \times 10^{-7} \text{ m}^3$$
B. Volume of particle

\[ V_p = \frac{4}{3} \pi r_p^3 \]

\[ = \frac{4}{3} \pi (2.5 \times 10^{-6})^3 \]

\[ = 6.54 \times 10^{-17} \text{ m}^3 \]

No. of particles

\[ \frac{V}{V_p} = 2.8 \times 10^9 \text{ or } 2,800,000,000 \text{ (2.8 billion!)} \]

C. Assume: 1 million paces per year, total angular motion per pace (hip extension–flexion–extension) of 80°.

\[ x = 10^6 \times (80/360) \times 2\pi r = 1.94 \times 10^4 \text{ m/year} \]

\[ V = kFx \rightarrow k = \frac{V}{Fx} \]

F: Assume 70 kg body weight, average load (over gait cycle) = 2.5 \times body weight = 1715 N

\[ k = 1.85 \times 10^{-7} \frac{V}{(1.94 \times 10^4 \times 1715)} = 5.56 \times 10^{-15} \text{ m}^2/\text{N} \]

Wear mechanisms

Since both coefficients of friction and wear rates depend on interfacial load rather than stress, there is a natural tendency to draw the conclusion that good lubrication implies low wear rates. Unfortunately, that is not the general case, owing to the radical differences between the processes that generate frictional force and those that produce wear debris. For a particular wear process, improved lubrication may radically reduce the wear rate; however, different wear processes have widely different intrinsic rates. Thus, there is no correlation between \( \mu \) and \( k \).

When two surfaces are in contact and in relative motion, there are three primary wear processes that may result (Figure 11.9).

1. In the simplest case, asperities on the harder surface wear groove in the softer surface. This is termed plowing or abrasive wear and produces the lowest wear rate of these processes. However, resulting stress concentrations may result in accelerated component failure, by single-cycle or fatigue fracture. Chemical bonding between surfaces, especially in the absence of adequate lubrication, may increase abrasive wear. During abrasion, debris size is of a similar size to the roughness of the harder surface.

2. It is possible for the softer material to adhere to the harder surface, filling in the spaces between the asperities, thus forming a smooth transfer film. If the film is fairly strong and adherent, a stable intermediate wear rate occurs.
3. However, it is possible for either abrasive debris or portions of transfer film to be trapped between the surfaces. These debris produce extremely local, very high stresses, resulting in rapid (high-stress, low-cycle) localized fatigue failure. This process, called third-body wear, is generally discontinuous, characterized by rapid variations in wear rate even after wearing-in, rather than by the constant rate shown in Figure 11.7.

In third-body wear, fatigue plays a major role, especially in the cases of rough initial surfaces. This commonly occurs when surfaces adhere prior to eventual asperity fatigue owing to shear stress at the adhered asperity. In certain cases, delamination can occur when cyclic shear stress from asperity adherence exceeds the fatigue limits of the bulk material. Because fatigue processes depend on stress rather than on load, some consideration must be given to an aspect of design of articulating implants that may produce a more general fatigue wear process.

In a spherical ball-in-socket joint, such as a THR, when the radius of the socket is essentially that of the ball, the interfacial compressive or contact stress is given by

$$\sigma_0 = \frac{W}{\pi r^2}$$

where $W$ is the normal load (assuming that $\mu_s$ and $\mu_D$ are very small) and $r$ is the common radius ($= r_1 = r_2$) (Figure 11.10). This is called a congruent design.

However, two factors can contribute to higher surface stresses:

1. If the surfaces are incongruent, that is, $r_1/r_2 > 1$, there is an increase in contact stress ($\sigma_c$) above $\sigma_0$. This would be the case in a poorly

**FIGURE 11.9** Primary wear processes.
fitting THR combination or in the articulation of a total knee replacement (TKR) or, for that matter, in most natural joints.

2. If the softer material, usually the socket because of creep considerations, is too thin (“t” too small), then the contact stress is additionally increased. This occurs since the “substrate” or supporting material, whether in a natural joint or in an artificial replacement, is much stiffer than either articular cartilage or a typical bearing polymer, respectively. One may easily appreciate this effect by comparing the feeling of standing barefoot on concrete or on soft earth; the relative sensations are a direct consequence of the difference in average contact stress imposed by the metatarsals on the soft tissues of the mid- and forefoot. (This is, unfortunately, not a pure example since contact areas may also differ.)

These effects may combine to produce increases of up to 10 times normal in contact stress in noncongruent thin polymeric components such as thin metal-backed tibial plateaus in TKRs. For the typical radial dimensions encountered in THR and TKR designs, these effects diminish and approach a lower limit for polymer (UHMWPE) thicknesses of 7–10 mm and are unimportant at greater thicknesses.

The consequences of such high local contact stresses in polymers are first a localized surface cracking called “mud-caking” because of the similarity of appearance to the surface of dried mud. As the process progresses, fatigue cracks extend parallel to the surface, releasing fragments and producing accelerated wear, through both material loss and third-body wear (Figure 11.11). This has been observed in late failure of early metal-backed acetabular components with thin polymer shells; evidence of the early stages has also been found in more conventional implants retrieved after long periods (7–10 years). The similarity between these defects in polymeric components and those observed in articular cartilage in osteoarthritis (OA) has led to suggestions that fatigue
processes also play a role in OA, secondary to stiffening of subchondral bone (owing to sclerosis or microfracture) and thinning of cartilage.

Small degrees of incongruence in hard-on-hard bearings such as metal-on-ceramic (MOC) (e.g., CoCr–Al₂O₃), ceramic-on-ceramic, and metal-on-metal (MOM) bearing pairs can produce extreme elevations of local stress leading to high wear and failure. Thus, very precise tolerances are necessary for successful manufacture of such component pairs.

**Wear** is a complex process, with many features of specific materials pairs and designs affecting the outcome. Some important additional processes, which may occur independently or together, are as follows:

- **Corrosive wear.** When sliding takes place in a corrosive environment, wear processes may disrupt the passivation layer on the metal part of the pair, producing accelerated material loss. This is discussed further in Chapter 12.

- **Fretting.** When relative motion is possible, but over a very restricted range, such as between screw head and hole in a screw plate internal fracture fixation device, significant localized wear, called fretting, can occur. The amount of debris produced is often amazing, producing extensive tissue discoloration *in vivo* even under conditions that do not favor corrosion processes. *In vitro* studies demonstrate that serum proteins act as lubricants, radically reducing the effects seen in saline solutions.

**Wear rates**

There have been many efforts made to measure the rate of wear debris production in the laboratory. In general, the results depend on the geometry of the test, on the lubricant selected to simulate synovial fluid, and, to some degree, on the experimenter. There have been great difficulties encountered in reproducing *in vitro* experimental results. However, all of these studies agree in that both the size of the individual particle and the number of particles produced in a given situation vary directly with the ability of the surfaces to “stick” and inversely with the elastic modulus,
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$E_\text{r}$ of the material in question. Since like materials are able to bond easily, the former factor explains why like pairs of materials tend to show higher wear rates than unlike pairs. However, the range of moduli is far greater than the range of surface associations (“stickiness”) in the presence of lubrication, so modulus effects dominate production of wear debris.

Thus, in vivo, we expect that wear will occur preferentially to the less stiff (possessing lower modulus) member of the articulating pair. In a MOP pair, the polymer wears almost exclusively, and in a MOC pair, the metal will wear but will produce much smaller particles than in the previous case (<0.01–1.0 μm for metal [against ceramic] vs. 1–100 μm for UHMWPE [against metal]).

These general relationships have been verified clinically, at least with respect to typical wear debris particle size, through a number of studies of tissue retrieved from patients during THR arthroplasty revision. However, such studies have failed to provide reasonable estimates of either relative wear rates or total volume of wear debris, owing to the high solubility of fine metallic debris and the increase in phagocytosis and transport rates with decreasing particle size.

The limit on radiographic determination of dimensional changes associated with wear in vivo is probably ±0.25 mm/year, although this number has been criticized as being too small. Thus, most reliable data on wear–creep rates are obtained from prosthetic components recovered during revision surgery and, occasionally, at autopsy. The best estimates for in vivo volumetric wear rates for typical THR designs are given in Table 11.3. There is general agreement that the rate of 0.15 mm/year (expressed as surface recession rates rather than volume of wear debris) is a fair estimate of the true wear rate of conventional UHMWPE/cobalt-base alloy pair in THRs.

Wear-induced osteolysis continues to be a salient concern in large joint arthroplasty. In general, reducing the volume of wear particles associated with arthroplasty components should reduce the adverse biologic responses to them. Concerns over polyethylene wear-induced osteolysis have spurred interest in alternative bearing surfaces.

### Table 11.3 Estimated in vivo wear rates

<table>
<thead>
<tr>
<th>Material combination</th>
<th>Material worn</th>
<th>Rate (mm$^3$/year)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Metal/polymer (Co–Cr/UHMWPE)</td>
<td>Polymer</td>
<td>40.8</td>
</tr>
<tr>
<td>Metal/metal (Co–Cr)</td>
<td>Metal</td>
<td>0.023–6.3</td>
</tr>
<tr>
<td>Ceramic (alumina)/polymer (Al$_2$O$_3$/UHMWPE)</td>
<td>Polymer</td>
<td>51</td>
</tr>
<tr>
<td>Ceramic (alumina)/ceramic (alumina) (Al$_2$O$_3$/Al$_2$O$_3$)</td>
<td>Ceramic</td>
<td>0.004–0.04</td>
</tr>
<tr>
<td>Ceramic (zirconia)/ceramic (alumina) (ZrO$_2$/Al$_2$O$_3$)</td>
<td>Ceramic</td>
<td>0.1</td>
</tr>
</tbody>
</table>

of wear debris in hard-on-hard joints is one to two orders of magnitude lower than traditional metal-on-polyethylene bearings. (Also of note: dimensions of hard bearing wear particles are also much smaller.) The past 15 years have seen further development of hard-on-hard bearings to directly address the issue of implant wear.

Wear in newer generations of dense, polycrystalline alumina is on the order of a few microns per year, though there are exceptions. Some alumina bearings show a patch of localized surface damage, including loss of grains, commonly termed striped wear. The initiation of striped wear is suspected to occur when the femoral head dislocates slightly from the bearing and impinges upon the edge at the beginning of the gait cycle. This process is variously termed lateralization or microseparation. Zirconia has improved fracture toughness properties compared to alumina, but the amount of wear in zirconia ceramic on ceramic bearings is somewhat controversial. Some studies have reported severe wear, while others report wear to be very low. Still, because of the typically low wear rate of ceramic bearings, osteolysis is very rare.

MOM bearings have also gained significant market share, though the tenure of this bearing has been more controversial. MOM bearings exhibit a linear wear rate that is on the order of a few microns per year, though wear tends to be very dependent on finding an optimum clearance. A smaller clearance in MOM bearings increases the capability to generate fluid film thickness. Unfortunately, if the tolerance of the sphericity is larger than that of the clearance, binding or excessive wear may occur. On the other hand, too large a clearance will not facilitate fluid film lubrication, lead to a smaller contact area, and ultimately result in higher contact stresses. Carbon content and heat treatment are also important and can affect this distribution of surface carbides, which directly affect wear. Also notable, it has been proposed that MOM bearings can self-polish, moderating surface scratches in vivo.

As a gross approximation, total volume of wear is equal to the wear depth multiplied by the contact area. Increasing the diameter of the head will increase the contact area and the sliding distance. In metal-on-polyethylene total hip arthroplasty articulations, the volume of wear per step has been noted to increase with increasing ball diameter. This relationship does not hold for metal-on-metal bearings. This is likely due to the ability of larger-diameter metal articulations to achieve fluid film lubrication. Because of this, larger-diameter MOM components were considered advantageous in that they could reduce the risk for dislocation, and the thinner acetabular cup component facilitated the conservation of more bone stock in the acetabulum. There have been noted potential biological adverse effects of metal ions, which is discussed in detail in Chapter 14.

Wear rates for any material pair increase with higher patient weight and activity. Larger femoral head sizes tend to result in lower surface recession rates due to the lower contact stress but may be associated with greater volumes of wear debris due to the greater sliding distance. This may be particularly obvious with surface replacement devices for the hip, for which it has been estimated that wear rates (as measured by rate of debris production) are 1.5 to 2 times those for conventional THR designs.
Materials subject to creep, such as UHMWPE, may show local deformation that may be mistaken for wear. This is easily appreciated in polymeric prosthetic components for tibial plateau replacement, in which peak stresses may exceed 10 MPa, and has come to be called “cold flow” in the clinical literature. At more moderate stresses, such as the 3–4 MPa experienced by the typical acetabular cup, it may not be obvious but has been estimated to account for as much as one-fourth to one-half of the apparent wear observed in retrieved UHMWPE cups. This type of deformation depends on geometric stress, so that heavier patients or those whose femoral prostheses have relatively smaller heads will experience greater creep. In addition, there is believed to be some slight increase in creep rate over time in vivo, owing to absorption of low-molecular-weight species that can act as plasticizers, making creep easier.

The limiting creep rate for a typical patient for an acetabular cup of at least 10 mm wall and 32 mm internal diameter is believed to be between 0.03 and 0.1 mm/year. Composite materials, such as carbon fiber–reinforced UHMWPE, may be expected to show lower creep rates in vivo. Metals and structural ceramics essentially do not creep at body temperature.

Examples of wear

Wear produces changes in surface appearance, particularly since bearing surfaces are usually highly polished. The most frequent change in appearance of metals associated with wear is a “frosted” or matte appearance. This is due to an increase in surface roughness, secondary to the inhomogeneous nature of wear. When relative motion is oriented, parallel lines or score marks occur, as seen in the hinge pin of a fully constrained stainless steel TKR retrieved after 7.5 years (Figure 11.12). Evidence of wear of metal components in MOP devices is usually less obvious. However, exceptional cases may occur, as seen in the Ti6Al4V femoral head retrieved after 15 months shown in Figure 11.13. The origin of such extreme wear in titanium-base alloy components is not known.

**FIGURE 11.12** Wear of hinge pin of TKR (stainless steel).
Although it apparently occurs quite rarely, it produces sufficient volumes of wear debris to require revision surgery for pain relief.

Polymers wear at much higher rates than metals (see Table 11.3), and large material removal may occur, as seen in the UHMWPE tibial liner shown in Figure 11.14. This damage reflects both wear and creep. Therefore, in such cases of polymeric wear, care must be taken to distinguish between wear and creep; microstructural examination of the material below the surface may be necessary to resolve the relative contributions when such questions arise.

When third-body wear occurs in a joint replacement, it is frequently accompanied by the presence of embedded debris in the polymeric component (metal, poly(methyl methacrylate), bone chips, etc.) and associated score marks on the metal component. Figure 11.15 shows the appearance of a UHMWPE acetabular liner cup removed after 12 years. The darker spot near the top is a fragment of bone, while wear, delamination, and local fatigue breakdown are all visible as well.

**FIGURE 11.13**  Wear of femoral head of THR (Ti6Al4V).

**FIGURE 11.14**  (See color insert.) Wear of tibial plateau (UHMWPE).
Wear continues to be a challenge in implant surgery. As will be discussed at greater length in Chapter 14, the biologic response to wear debris may prove to be a limiting factor in TJR arthroplasty life. Although it will take very long development periods, new materials and possibly newer designs offer future hope of performance improvement in this area.

**Additional problems**

**PROBLEM 11.3**

Refer to Figure 11.16 and observe that TKR A has bearing surfaces that fit closely, whereas TKR B has a distinctly smaller radius of curvature for the femoral component than for the tibial component.

Considering the bearing surfaces, select the true statements from the following:

1. Because type A prostheses allow no entrance for lubricating fluids, they wear out faster.
2. The bearing contact stress for prosthesis B is greater than that for A.
3. Considering the coefficient of friction between Co–Cr alloys and UHMWPE, prosthesis A should significantly resist flexion–extension more than B.

**FIGURE 11.15** (See color insert.) Wear and delamination of an acetabular cup (UHMWPE).

**Final remarks**

Wear continues to be a challenge in implant surgery. As will be discussed at greater length in Chapter 14, the biologic response to wear debris may prove to be a limiting factor in TJR arthroplasty life. Although it will take very long development periods, new materials and possibly newer designs offer future hope of performance improvement in this area.
4. Little or no cold flow will occur for prosthesis B since there is less contact between the parts than for A.

5. The greater area of contact for prosthesis A will result in a greater volume of wear debris for the same usage period.

**ANSWER:**
F, T, F, F, F. 1 is incorrect since both designs allow easy entry of knee joint fluids. 2 is true since congruency and contact stress vary inversely. 3 is false since frictional forces are independent of contact area; in either case, such low-congruency designs, using modern materials, produce very low resistance to relative motion. 4 is false since, for a given joint force, “cold flow” (creep) increases with decreasing contact area. 5 is false since the rate of wear debris production is dependent on sliding distance rather than contact area.

**PROBLEM 11.4**
When two surfaces slide over each other (select the best answer[s]),

A. The surfaces wear
B. A frictional force resists motion
C. The surfaces become smoother
D. Some wear will occur and a friction force occurs
E. The surfaces become rougher

**ANSWER:**
The best answer is D. Wear always occurs; the consequence may be either smoother or rougher surfaces depending on the situation.

**PROBLEM 11.5**
“Stiction” friction is unimportant and disappears on motion (true or false).

**ANSWER:**
True. Stiction is an older concept that arose from the observation that static frictional coefficients are higher than dynamic ones. In reality, both static and dynamic frictional resultant forces for modern TJR designs are so low that it is not credible that they contribute to loosening of components. The actual effects previously ascribed to stiction were more likely related to deformation of polymeric components producing “pinching” effects and apparently large frictional resistive forces.

**PROBLEM 11.6**
When a smooth metal slides on a smooth polymer (select the best phrase[s]),
A. It is necessary to use a lubricant to prevent wear
B. Frictional forces will be very high
C. Wear will be abrasive in character
D. Fatigue wear will likely occur
E. There will be no wear at all

**ANSWER:**

D is the best answer.

**PROBLEM 11.7**

Wear in a metal–polymer total hip prosthesis (true or false)

1. May be easily detected *in vivo*
2. May be produced by PMMA and bone fragments
3. Is confined to the acetabular component
4. Is confined to the femoral component
5. Occurs on both femoral and acetabular components, but unequally

**ANSWER:**

F, T, F, F, T.

**PROBLEM 11.8**

Hydrodynamic lubrication (select the best phrase[s])

A. Occurs at high loads and low speeds
B. Is desirable because it reduces turbulence
C. Is the lubrication mechanism in THRs
D. Results in a complete separation of the two surfaces
E. Can only occur if the surfaces are very smooth

**ANSWER:**

D is the best answer.

**PROBLEM 11.9**

Wear debris (true or false)

1. Are spherical
2. Differ in size for different materials
3. May be retained between articulating surfaces
4. Are a rare occurrence
5. May produce third-body wear
ANSWER:
F, T, T, F, T.

PROBLEM 11.10
Rank the following wear pairs in decreasing order of (1) dynamic frictional coefficient and (2) wear debris production rate (assume saline lubrication):

A. Co–Cr/Co–Cr
B. Al₂O₃/UHMWPE
C. Co–Cr/UHMWPE
D. Stainless steel/UHMWPE
E. Ti6Al4V/UHMWPE

ANSWER:
1: A (C, D, E [equivalent]), B; 2: D (C, E [equivalent]), B, A.
Note: In 2, $D > C$ because of relative corrosion rate of stainless steel.

Annotated bibliography

   Includes the majority of Frank Philip Bowden’s extensive contributions to tribology, extended by the work of his student David Tabor. An engineering text but extremely clear and straightforward.

   A good synopsis of tribology in polyethylene, ceramic, and metal bearings.

   Discusses properties and test methods in great depth. Includes some of Dumbleton’s work on modification of properties of ultrahigh-molecular-weight polyethylene (UHMWPE).

   Research discussing the role of proteins in the lubrication process.

   Early effort to relate coefficients of friction to wear rates.

   Often-cited compilation of laboratory studies.

Review of clinical experience with a THR device including a ceramic-ceramic articulating pair. Includes reports of laboratory wear tests of \( \text{Al}_2\text{O}_3/\text{UHMWPE} \) and \( \text{Al}_2\text{O}_3/\text{Al}_2\text{O}_3 \) wear pairs.

   
   An intermediate level work on friction and wear, with the main emphasis on metals.

   
   A review of basic tribological principles for orthopaedics, including the relationship between lubrication and film thickness.

    
    This chapter deals with friction, wear, and lubrication in natural joints.

    
    Early discussion of friction and wear in total joint replacements, with comparison of metal-on-metal with metal-on-polymer devices.

    
    A concise review of the basic principles, with relevant *in vitro* test results and patient observations, related to wear in total joint replacements.
