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Orthopedic Joint Implants

20.1 INTRODUCTION

Orthopedic joint implants are widely used in orthopedic surgery, particularly for the hip joint. Each year, more than 250,000 of orthopedic hip joints are implanted in the United States alone to treat severe hip joint disease, and this number is increasing every year. Although much research work has been devoted to various aspects of this topic, there are still several important problems. In the past, most of the research was conducted by bioengineering and medical scientists, and participation by the tribology community was limited. In fact, in the past decade there has been a significant improvement in bearings in machinery, but the design of the hip replacement joint remains basically the same. This is an example where engineering design and the science of tribology can be very helpful in actual bioengineering problems.

The common design of a hip replacement joint is shown in Fig. 20-1. The acetabular cup (socket) is made of ultrahigh-molecular-weight polyethylene (UHMWPE), while the femur head replacement is commonly made of titanium or cobalt alloys. The early designs used metal-on-metal joints in which both the femoral head and socket were made of stainless steel. In 1961, Dr. Sir John Charnley in England introduced the UHMWPE socket design. A short review of the history of artificial joints is included in Sec. 20.3.

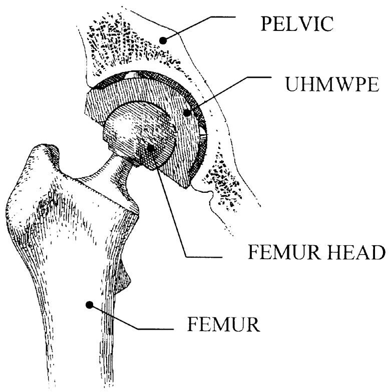

FIG. 20-1 Hip replacement joint.

The combinations with UHMWPE have relatively low friction and wear in comparison to earlier designs with metal sockets. Later, the stainless steel femur was replaced with titanium or cobalt alloys for better compatibility with the body. It proved to be a good design and material combination, with a life expectancy of 10–15 years. This basic hip joint design is still commonly used today.

For comparison with the implant joint, an example of a natural joint (synovial joint) is shown in Fig. 20-2. The cartilage is a soft, compliant material, and together with the synovial fluid as a lubricant, it is considered to be superior in performance to any manmade bearing (Dowson and Jin, 1986, Cooke et al., 1978, and Higginson, 1978).

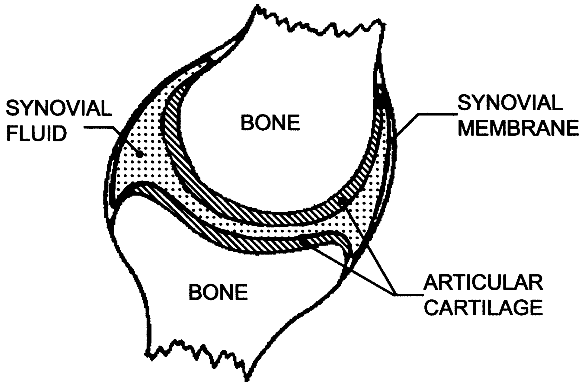

FIG. 20-2 Example of a natural joint.

Although significant progress has been made, there are still two major problems in the current design that justify further research in this area. The most important problems are

1. A major problem is that particulate wear debris is undesirable in the body.
2. A life of 10–15 years is not completely satisfactory, particularly for young people. It would offer a significant benefit to patients if the average life could be extended.

Previous studies, such as those by Willert et al. 1976, 1977, Mirra et al. 1976, Nolan and Phillips, 1996, and Pappas et al., 1996, indicate that small-size wear debris of UHMWPE is rejected by the body. Furthermore, there are indications that the wear debris contributes to undesirable separation of the metal from the bone. There is no doubt that any improvement in the life of the implant would be of great benefit.

20.2 ARTIFICIAL HIP JOINT AS A BEARING

The artificial hip joint is a heavily loaded bearing operating at low speed and with an oscillating motion. The maximum dynamic load on a hip joint can reach five times the weight of an active person. During walking or running, the hip joint bearing is subjected to a dynamic friction in which the velocity as well as load periodically oscillate with time. The oscillations involve start-ups from zero velocity. The joint is considered a lubricated bearing in the presence of body fluids, although the lubricant is of low viscosity and inferior to the natural synovial fluid.

For a lubricated sleeve or socket bearings, a certain minimum product of viscosity and speed, μU, is required to generate a full or partial fluid film that can reduce friction and wear. At very low speed, there is only boundary lubrication with direct contact between the asperities of the sliding surfaces.

Dry friction of polymers (such as UHMWPE) against hard metals is unique, because the friction coefficient increases with sliding velocity (Fig. 16-4). Friction of metals against metals has an opposite trend of a negative slope of friction versus sliding velocity. For polymers against metals, the start-up dry friction is the minimum friction, whereas it is the maximum friction in metal against metal. However, for lubricated surfaces, there is always a negative slope of friction versus sliding velocity, and the start-up friction is the maximum friction for polymers against metals as well as metals against metals.

From a tribological perspective, the performance of artificial joints is inferior to that of synovial joints. The reciprocating swinging motion of the hip joint means that the velocity will be passing through zero, where friction is highest, with each cycle. In its present design, the maximum velocity reached in an artificial joint is not sufficient (or sustained long enough) to generate full hydrodynamic lubrication. Under normal activity, much of the motion associated with joints is of low velocity and frequency. In artificial joints this means that lubrication is characterized by boundary lubrication, or at best mixed lubrication. In contrast, natural, synovial joints are characterized by a mixture of a full fluid film and mixed lubrication. Experiments by Unsworth et al. (1974, 1988) and O'Kelly et al. (1977) suggest that hip and knee synovial joints operate with an average friction coefficient of 0.02. In comparison, the friction coefficient measured in artificial joints ranges from 0.02–0.25. High friction causes the loosening of the implant. In addition, wear rate of artificial joints is much higher than in synovial joints.

The synovial fluid provides lubrication in the natural joint. It is highly non-Newtonian, exhibiting very high viscosity at low shear rates; however, it is only slightly more viscous than water at high shear rates. Dowson and Jin (1986, 1992) have attempted to analyze the lubrication of natural joints. In their work, they couple overall elastohydrodynamic analysis with a study of the local, micro-elastohydrodynamic action associated with surface asperities. Their analysis indicates that microelastohydrodynamic action smooths out the initial roughness of cartilage surfaces in the loaded junctions in articulating synovial joints.

In natural joints a cartilage is attached to the bone surfaces. This cartilage is elastic and porous. The elastic properties of the cartilage allow for some compliance that extends the fluid film region. This is in contrast to artificial joints, which are relatively rigid and consequently exhibit poor lubrication in which ideal separation of the surfaces does not occur. Contact between the plastic and metal surfaces increases the friction and leads to wear. The problem is compounded due to the fact that synovial fluid in implants is much less viscous than that in natural joints (Cooke et al. 1978). Therefore, any future improvement in design which extends the fluid film regime would be very beneficial in reducing friction and minimizing wear in artificial joints.

20.3 HISTORY OF THE HIP REPLACEMENT JOINT

Dowson (1992, 1998) reviewed the history of artificial joint implants. The following is a summary of major developments of interest to design engineers.

Unsuccessful attempts at joint replacement were performed* as early as 1891. These attempts failed due to incompatible materials, and infections. In 1938, Phillip Wiles designed and introduced the first stainless steel artificial hip

*The German surgeon Gluck (1891) replaced a diseased hip joint with an ivory ball and socket held in place with cement and screws. Two years later, a French surgeon, Emile Pean replaced a shoulder joint with an artificial joint made of platinum rods joined by a hard rubber ball.

joint (see Wiles, 1957). The prosthesis consisted of an acetebular cup and femoral head (both made of stainless steel held in place by screws). The matching surfaces of the cup and femoral head were ground and fitted together accurately. The basic design of Phillip Wiles was successful and did not change much over time; however, the steel-on-steel combination lacks tribological compatibility (see Chapter 11), resulting in high friction and wear. The high friction caused the implants to fail by loosening of the cup that had been connected to the pelvis by screws. Failure occurred mostly within one year; therefore, only six joints were implanted.

In the 1950s, there were several interesting attempts to improve the femoral head material. For example, the Judet brothers in Paris used acrylic for femoral head replacement in 1946 (Judet and Judet, 1950); however, there were many failures due to fractures and abrasion of the acrylic head. In 1950, Austin Moore in the United States used *Vitallium*, a cobalt–chromium–molybdenum alloy, for femoral head replacement (see Moore, 1959).

Between 1956 and 1960, the surgeon G.K. McKee replaced the stainless steel with Vitallium; in addition, McKee and Watson-Farrar introduced the use of methyl-methacrylate as a cement to replace the screws. The design consisted of relatively large-diameter femoral head, and the outer surface of the cup had studs to improve the bonding of the cup to the bone by cement (see McKee and Watson-Farrar, 1966, and McKee, 1967). They used a metal-on-metal, closely fitted femoral head and acetabular cup. These improvements significantly improved the success rate to about 50%. However, the metal-on-metal combination loosened due to fast wear, and it was recognized that there is a need for more compatible materials.

Dr. Sir John Charnley developed the successful modern replacement joint (see Charnley, 1979). Charnley conducted research on the lubrication of natural and artificial joints, and realized that the synovial fluid in natural joints is a remarkable lubricant, but the body fluid is not as effective in the metal-on-metal artificial joint. He concluded that a self-lubricating material would be beneficial in this case. In 1969, Dr. Charnley replaced the metal cup with a polytetraflouro-ethelyne (PTFE) cup against a stainless steel femoral head. The design consisted of a stainless steel, small-diameter femoral head and a PTFE acetebular cup. The PTFE has self-lubricating characteristics, and very low friction against steel. However, the PTFE proved to have poor wear resistance and lacked the desired compatibility with the body (implant's life was only 2–3 years).

In 1961, Dr. Charnley replaced the PTFE cup with UHMWPE, which was introduced at that time. The wear rate of this combination was 500 to 1800 times lower than for PTFE cup. In addition, he replaced the screws and bolts with methyl-methacrylate cement (similar to the technique of McKee and Watson-Farrar). A study that followed 106 cases for ten years, and ended in 1973, showed a success rate of 92%. This design remains (with only a few improvements) the most commonly used artificial joint today.

The use of cement in place of screws, UHMWPE, ceramics, and metal alloys with super fine surface finish has led to the remarkable success of orthopedic joint implants; this is one of the important medical achievements.

However, there are still a few problems. Wear debris generated by the rubbing motion is released into the area surrounding the implant. Although UHMWPE is compatible with the body, a severe foreign-body response against the small wear debris has been observed in some patients. Awakened by the presence of the debris, the body begins to attack the cement, resulting in loosening of the joint. Recently, complications resulting from UHMWPE wear debris have renewed some interest in metal-on-metal articulating designs (Nolan and Phillips, 1996).

Wear is still a major problem limiting the life of joint implants. With the current design and materials, young recipients outlive the implant. With the average life span increasing, recipients will outlive the life of the joint. Unlike natural joints, the implants are rigid, the lubrication is inferior, and there is no soft layer to cushion impact. Further improvements are expected in the future; new implants will likely be more similar to natural joints.

20.4 MATERIALS FOR JOINT IMPLANTS

The materials in the prostheses must be compatible with the body. They must not induce tumors or inflammation, and must not activate the immune system. The materials must have excellent corrosion resistance and, ideally, high wear resistance and low coefficient of friction against the mating material. Publications by Sharma and Szycher (1991) and Williams (1982) deal with materials compatible with the body.

For the femoral head, low density is desirable, and high yield strength is very important. Common materials used are cobalt-chromium-molybdenum alloys and titanium alloy (6Al-4V). Cobalt alloys have excellent corrosion resistance (much better than stainless steel 316). The titanium alloy has high strength and low density but it is relatively expensive. Titanium alloys have a low toxicity and a strong resistance to pitting corrosion, but its wear resistance is somewhat inferior to cobalt alloys. Titanium alloy is considered a good choice for patients with sensitivity to cobalt debris. Aluminum oxide ceramic is also used in the manufacture of femoral heads. It has excellent corrosion resistance and compatibility with the body.

20.4.1 Ceramics

Aluminum oxide ceramic femoral heads in combination with UHMWPE cups have increasing use in prosthetic implants. Fine grain, high density aluminum oxide has the required strength for use in the heavily loaded femoral heads, high corrosion resistance, and wear resistance, and it has the advantage of self-anchoring to the human body through bone ingrowth. Most important, femoral ball heads with fine surface-finish ceramics reduce the wear rate of UHMWPE cups. Dowson and Linnett (1980) reported a reduction of 50% in the wear rate of UHMWPE against aluminum-oxide ceramic, in comparison to UHMWPE against steel (observed in laboratory and in vivo tests).

The apparent success of the ceramic femoral head design led to experiments with ceramic-on-ceramic joint (the UHMWPE cup is replaced with a ceramic cup). However, the results showed early failure due to fatigue and surface fracture. Ceramic-on-ceramic designs require an exceptional surface finish and precise manufacturing to secure close fit. Surgical implantation of the all-ceramic joint is made more difficult by the necessity to maintain precise alignment. In addition, the strength requirements must be carefully considered during the design (Mahoney and Dimon, 1990, Walter and Plitz, 1984, and McKellop et al., 1981).

20.5 DYNAMIC FRICTION

Most of the previous research on friction and wear of UHMWPE against metals was conducted under steady conditions. It was realized, however, that friction characteristics under dynamic conditions (oscillating sliding speed) are different from those under static conditions (steady speed).

Under dynamic condition, the friction is a function of the instantaneous sliding speed as well as a memory function of the history of the speed. It would benefit the design engineers to have an insight into the dynamic friction characteristics of UHMWPE used in implant joints. During walking, the hip joint is subjected to oscillating sliding velocity. Dynamic friction experiments were conducted at New Jersey Institute of Technology, Bearing and Bearing Lubrication Laboratory. The testing apparatus is similar to that shown in Fig. 14-7, and the test bearing is UHMWPE journal bearing against stainless steel shaft. The oscillation sliding in the hip joint is approximated by sinusoidal motion, obtained by a computer controlled DC servomotor. The friction and sliding velocity are measured versus time, and the readings are fed on-line into a computer with a data acquisition system, where the data is stored, analyzed and plotted.

Figures 20-3 and 20-4 are examples of measured f–U curves for simulation of the walking velocity and frequency. The frequency of normal walking is approximated by sinusoidal sliding velocity $\omega = 4\,\mathrm{rad/s}$, and a maximum sliding velocity of $\pm 0.07\,\mathrm{m/s}$. The shaft diameter is 25 mm, with $L/D = 0.75$ and a constant load of 215 N.

For dry friction (Fig. 20-3), the friction increases with sliding velocity. At the start-up (acceleration) the friction is higher than for stopping (deceleration).

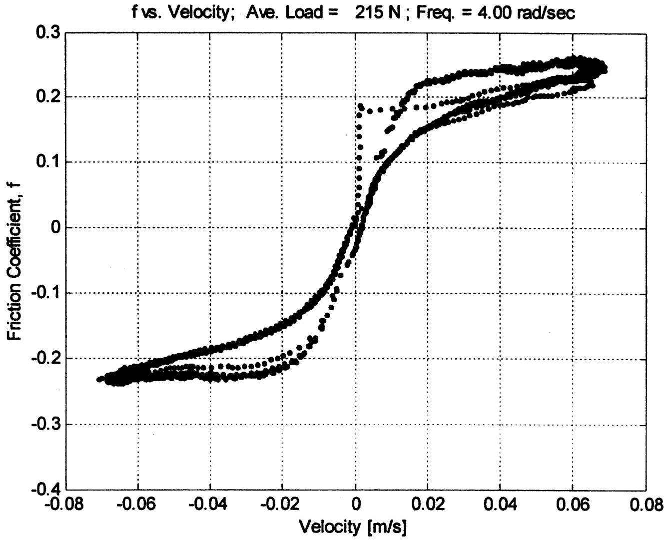

FIG. 20-3 Friction–velocity curve for dry friction, UHMWPE against stainless steel, frequency = 4 rad/s, maximum velocity = ± 0.07 m/s, load = 215 N (Bearing and Bearing Lubrication Laboratory, NJIT).

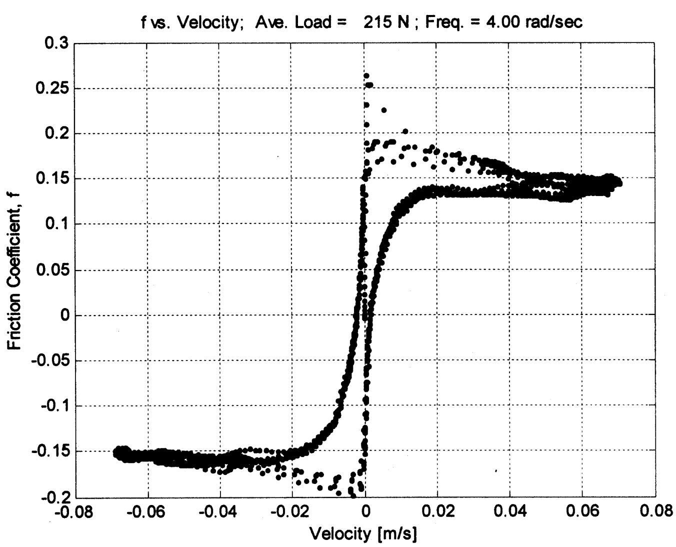

FIG. 20-4 Friction–velocity curve, lubrication with low viscosity oil, $\mu = 0.001$ N-s/m^2, UHMWPE against stainless steel, frequency = 4 rad/s, maximum velocity = ± 0.07 m/s, load = 215 N (Bearing and Bearing Lubrication Laboratory, NJIT).

Unlike the metal-on-metal curve, the dynamic f–U curve with UHMWPE has considerable hysteresis for dry friction. This effect suggests that the friction of polymers involves viscous friction.

Several cycles are measured, and the curve shows a good repeatability of the cycles—except for the first cycle (dotted line), which has a higher stiction force. Unlike what we see with metal-on-metal testing, the friction coefficient increases with the velocity, reaching a maximum of $f = 0.26$. In this case, the breakaway friction at each cycle approaches zero. However, at the first cycle of the experiment (dotted line), there is a higher stiction force, and the breakaway friction coefficient is near 0.2.

Figure 20-4 is for identical conditions, but lubrication is provided with a very light (low viscosity) oil, $\mu = 0.001$ N-s/m^2. This curve simulates the friction in an actual joint implant. The curve indicates that even for a low viscosity and speed, the bearing operates in the boundary and mixed lubrication regime, and the friction decreases versus sliding velocity. This curve also shows a considerable hysteresis. For lubricated surfaces, the first cycle (dotted line in Fig. 20-4) also demonstrates a higher stiction force of $f = 0.25$ while the following cycles have a reduced maximum breakaway coefficient of $f = 0.2$.