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Wear in Total Hip Prostheses

An Experimental Evaluation of Candidate Materials

by

JORGE O. GALANTE, M.D.
Professor and Chairman, Department of Orthopaedic Surgery, Rush Medical College, 1753 W. Congress Parkway, Chicago, Illinois 60612

and

WILLIAM ROSTOKER, PH.D.
Professor of Metallurgy, University of Illinois, Circle Campus, P. O. Box 4348, Chicago, Illinois 60680

From the Department of Orthopaedic Surgery, Rush Medical College, Rush-Presbyterian-St. Luke's Medical Center and Department of Materials Engineering, University of Illinois, Circle Campus, Chicago, Illinois, U.S.A.

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INTRODUCTION

The successful clinical use of total hip joint prostheses has been one of the most exciting developments in orthopaedic surgery in the past decade. The results in properly selected cases have been, to date, extremely gratifying to a degree unknown with other forms of hip reconstruction. A total hip prosthesis, however, is a complex engineering product, and its use in the human body poses a number of basic problems, some of which have not been satisfactorily solved as yet and which could result in eventual failure of the arthroplasty. A prosthesis of this type is a permanent implant. It must perform without failure for a lifetime of thirty to fifty years. Predictions of service for this length of time are not possible, given the current state of the art, and consequently an arbitrary age limit has been set in the selection of patients. There are many young individuals suffering severe handicap due to hip disease, who cannot at this time benefit from a total joint replacement due to the uncertainty surrounding the long-term performance of these prostheses.

One of the basic problems encountered in this context, is the wear of opposing articulating surfaces. Wear can result in failure of the arthroplasty for two reasons: 1) mechanical failure and 2) tissue response to particulate material (wear products). Past experience with polytetrafluorethylene has shown the effect of both of these mechanisms.

In the late 1950's, Charnley (1961) implanted total hip prostheses with a femoral component made of stainless steel and an acetabulum of polytetrafluorethylene (P.T.F.E.). Failure occurred in all cases. In some instances it was related to penetration of the femoral head through the plastic socket. In most cases failure resulted when the plastic socket loosened in the bony bed and caused pain in moving against the bone. Loosening resulted from the tissue reaction produced by particles of abraded P.T.F.E., (Charnley, Kamangar and Longfield, 1969).

There are two bearing systems in clinical use at present in total hip joint arthroplasty: a metal-on-plastic and a metal-on-metal bearing.

The Charnley low-friction arthroplasty (1969) and the Charnley-Mueller prosthesis, Mueller (1970), are the two more commonly used artificial joints incorporating a metal-on-plastic bearing. In the Charnley prosthesis the femoral component is made of stainless steel and the acetabulum of ultra high molecular weight polyethylene (RCH 1000). In the Charnley-Mueller prosthesis the femoral head is made of cobalt-chromium-molybdenum alloy and the acetabulum of ultra high molecular weight polyethylene. In the metal-on-metal prosthesis, both articulating surfaces are made of cobalt-chromium alloy, (McKee (1970)).

Clinical experience of up to 10 years has shown small amounts of wear with either one of the two systems. Charnley (1970) has reported a wear rate of 0.2 mm per year in the ultra high molecular weight polyethylene socket from measurements made on patients' X-rays. A great variability was found, however, with no detectable wear in some instances. There is overall uncertainty surrounding the long-term performance of these systems as there are no quantitative wear rates available at present on which to base an extrapolation of thirty or forty years of service.

This investigation was directed to provide answers to some of the problems related to wear in total joint prostheses. The specific objectives were the following:

1) To develop an objective laboratory method for evaluating and predicting rates of wear in total hip joint prostheses.
2) To evaluate the wear behaviour of a wide range of materials, metals, polymers and ceramics, including those currently in use.
3) To identify or develop a material with superior wear resistance and thus, to provide a solution to the problem of wear in total hip joint prostheses.

REVIEW OF THE LITERATURE

Various investigators have attempted to evaluate the wear characteristics of different prosthetic materials in the laboratory.

Duff-Barclay and Spillman (1966—67), described a hip joint simulator designed to test friction and wear in total joint prostheses. The simulator evaluated the performance of prostheses of various designs and materials under movement and pressure conditions similar to those pertaining to normal hip joints during walking. Tests were reported under dry conditions, in saline solution and in blood plasma. Titanium 160 and EN 58J stainless steel both exhibited high wear rates and were thought to be impractical in metal-to-metal prostheses. The lowest wear rate was elicited with a metal-on-metal cobalt-chromium prosthesis, with a horseshoe-shaped acetabulum bearing. A decreased coefficient of friction was observed using plasma, and was attributed to the presence of metal-protein complexes on the metallic surface. Tests with a Delrin (polyacetal) acetabular component showed production of particles attributed to a surface-fatigue type of failure. With RCH 1000 no true signs of wear were evident in the acetabulum although a significant deformation of the acetabulum was seen.

Charnley (1966–67), in a discussion on bearing materials, called attention to the poor wear resistance of P.T.F.E. articulating against metal with 7 to 10 mm of wear in the plastic within a $2^1/_2$ year period. He found that for a 41 mm head, linear wear was 3 mm; for a 22 mm head, linear wear was 10 mm. The volume of wear particles was lower with the smaller head than with the large one. Trials with filled P.T.F.E. appeared to be satisfactory in the laboratory but not in the human. Laboratory tests showed that high molecular weight polyethylene had resistance to wear superior to P.T.F.E. by a factor of 500, although *in vivo* measurements indicated only a factor of 16.5 (Charnley 1970).

Walker, Dowson, Longfield and Wright (1966–67), performed studies of friction and wear of artificial joint materials using a pin-on-disc machine. The specimens were tested both dry and lubricated with synovial fluid. Constant sliding speed and a nominal contact pressure of 19.6×10^6 nt/m^2 were used. The disc was made from EN 58J stainless steel and the pins from Perspex (polymethylmethacrylate), acetal copolymer, neoprene rubber, Viton rubber and cellular polyurethane. Nylon 66 exhibited the lowest wear rate, followed by high density polyethylene (from weight measurements). The lowest coefficient of friction in lubricated tests was exhibited by high density polyethylene.

Ahier and Ginsburg (1966–67), reported studies of wear in Vitallium (cobalt-chromium alloy) using a pin-on-disc machine. Cast Vitallium and a solution-treated (carbide-free) form were studied. For cast Vitallium, two regions of wear were defined: a mild wear region below a stress of 6.5×10^5 nt/m^2 and a severe region above. For the solution-treated alloy, the resistance to wear was lower than with cast Vitallium. Two regions of wear were also defined, with a transition at 3.4×10^5 nt/m^2. Tests were performed at speeds in the range of 20–2000 cm/sec. and wear rates appeared to be affected at the higher speeds.

Poli (1966–67), designed a simple hip simulator device to test wear in total hip prostheses. Evaluation of Nylonplast A, Teflon and Delrin showed that Delrin exhibited the best performance.

Scales, Kelley and Goddard (1969), discussed wear and friction studies of prosthetic materials. A pin-and-disc machine was used for preliminary studies. The conclusion was reached that cobalt-chromium-molybdenum alloy on itself was the most likely long-term bearing combination. Further studies were performed using the hip simulator described by Duff-Barclay and Spillman, (1966–67). Reconstituted human blood plasma was used as a physiological lubricant. The importance of design and finish and their effect on frictional values on metal-on-metal prosthesis was emphasized. A Charnley-type stainless steel/RCH 1000 prosthesis and a prosthesis with a cobalt-chromium-molybdenum alloy femoral head bearing on an RCH 1000 cup were also evaluated. These metal-on-plastic bearings exhibited lower frictional torque and lower wear volumes than metal-on-metal in a 500 hour period. The importance of symmetrical geometry and a mirror-like finish on the femoral head was emphasized.

Amstutz (1969), described friction and wear studies on polymeric materials. The tests were performed with an LFW-1 wear tester using mineral oil as a lubricant and with contact pressures of 6.4×10^6 nt/m^2 (932 psi). Ten different thermoplastic polymers bearing against carbon steel were evaluated. Polymides and ultra high molecular weight polyethylene exhibited the highest resistance to wear. The coefficient of friction appeared to decrease with sliding distance and a close correlation with wear rate was claimed to exist.

Theoretical calculation of the forces acting on the hip joint have been reported by many investigators leading to figures in the range of 2–2.5 times the body weight during one-leg standing, as reported by Inman (1947), and Williams and Lissner, (1962). During gait, values of 4.5 times body weight have been calculated as reported by Pauwels, (1935).

Rydell (1966), performed intravital measurements of forces acting on a femoral head prosthesis in two subjects. With one-leg stance, he recorded force values of 2.3 times body weight in one patient and 2.8 times body weight in the other. During gait, joint forces increased with walking speed, with maximum values of 1.76 times body weight in one subject and 2.7 times body weight in the other. During running, values of 4 times body weight were recorded.

J. P. Paul (1966–67 and 1969), calculated from experimental observations external to the body, including measurements of body movement and ground-to-ground force, the magnitude of the forces transmitted by the hip joint. Peak values occurred at the time of transfer of support from one foot to the other. Average values from 16 subjects of 3.88 times body weight shortly before toe-off with individual peak high values of 6.4 times body weight were reported. The values of joint force depended on a function of body weight, stride length and walking speed. Body weight and stride length were particularly important in increasing joint force. Relative peripheral joint velocities of 7.6 cm/sec. were reported.

Grieve (1966–67), evaluated hip joint velocities during gait in young adults and found that peak values varied between 2.5 and 7.6 cm/sec.

The measurements of wear recorded in the literature thus far do not provide a basis for design of human joint replacements to last the lifetime of the patient. Furthermore, there are material options and functional factors which have not been investigated. The present work was set out to provide a replicative methodology and data which was both comparative and useful for design objectives.

METHODOLOGY

Conditions of Wear Testing

The mechanisms of wear are not well understood and the functional relationships between wear rate and factors influencing them have limited application and cannot be used with confidence for design purposes. For that reason, the conditions for evaluating the relative wear rates of candidate materials must simulate wherever possible the actual service conditions. Wherever actual simulation is not possible, the conditions for laboratory testing should lean to the conservative side.

Wear in a human joint involves oscillatory motion, certain lineal sliding velocities, a range of forces and a fluid environment under closely controlled temperature. It is necessary to acquire measurements of wear that will be meaningful for the lifetime of an artificial human joint which may well exceed 30 years.

Translated into laboratory test terms, long wear experience signifies very large surface displacements. Since the character of sliding surfaces changes as wear progresses, short tests run the risk of not establishing typical surface conditions. This dilemma must be faced in the choice of wear test systems; in particular, on whether or not to incorporate oscillatory motion. Incorporation of the latter substantially reduces the allowable lineal surface translation distances per unit time. Thus, to produce large amounts of wear in a given test run, very long test times are required.

In this investigation the choice was made to use a non-oscillatory test system with single direction sliding motion at lineal speeds of 635–8900 cm per minute. These limits of test speed represent about 2–50 times, as reported by Grieve (1966–67), and J. P. Paul (1966–67), the lineal speeds expected in an artificial hip joint at realistic rates of walking. Obviously the influence of higher rates of testing must be evaluated in judging the significance of the test results.

Wear rates are known to depend on the apparent contact loads or stresses exerted perpendicular to the sliding surfaces. It is, therefore, important to estimate the range that is likely to be operative in a human hip joint. The pressure term is made up of two variables: the diameter of the femoral head and the dynamic equivalent load. The femoral head diameter in various designs in use varies between 22 and 37 mm. The dynamic equivalent load has been variously estimated as up to five times the body weight, as described by J. P. Paul, (1966–67 and 1969). Conservatively, therefore, we may use about 450 kg as the operative force. This leads to contact stresses in the range of 2.1 to 6.9×10^6 nt/m^2 (300–1000 psi) for wear studies applicable to the present problem.

It is not practical to use synovial fluid as the environment medium for a large program of wear study. It is necessary, therefore, to choose a fluid medium which will essentially perform the same function in a wear situation. A lubricant serves two purposes. It reduces the incidence of real contact between the asperities of the matching surfaces and it acts as a coolant or heat transfer medium. In synovial fluid, protein-polysaccharide complexes including hyaluronic acid provide a low shear strength film lying between the surfaces of articulating cartilage. There is, however, no evidence that the lubricating factors in synovial fluid continue to be available when a human joint is totally replaced. Nor is it evident that they would be an effective wear-suppressing agent in the presence of metallic, ceramic and/or polymeric materials. Accordingly, it appeared conservative to evaluate wear in an aqueous media in the absence of special lubricating conditions.

Amstutz (1969), has argued that laboratory wear studies should attempt to reproduce frictional conditions in the human joint. The implication being that this is essential to duplicate the wear mechanism. As it will be shown later, this may not be true. Actually, high wear rates can be associated with low friction and low wear rates can be associated with quite high friction. Therefore, the use of a lubricant such as mineral oil as the fluid medium for wear testing in the present context did not appear justified.

The remaining attribute of synovial fluid is the heat transfer or cooling capabilities. In this respect, it is no different than a saline solution. A saline solution would be a suitable fluid medium except that it is highly corrosive, requiring that all parts of a wear tester be made of materials at least as corrosion resistant as stainless steel. It is very difficult to engineer these materials into the shafts and bearings of precision machinery. Keeping these considerations in mind, the decision was made to conduct wear studies for the bulk of the program with recirculated 37° C water. Charnley (1970), came to the same decision. As will be reported a number of preliminary runs were made with Dulbecco solution (a balanced, buffered tissue culture media).

Wear Test System

Potential candidates for total human joint replacement include metallic alloys, high-density hard ceramics, a wide variety of polymers and combinations of these. One of the considerations in the choice of configuration of test specimens, particularly in the case of hard ceramics, is that precision dimensions can be really attained only in reasonably simple shapes. Furthermore, it is preferable to perform a large number of tests on a single specimen rather than be constrained to acquire a large number of specimens.

A disc-on-plate configuration was selected for the wear test system. The wear specimen was chosen to be a disc 80 mm in diameter by approximately 7.6 mm thick with two flat faces ground or machined parallel. This wear specimen could serve interchangeably, both as a rotating or a stationary member (configuration of rotating disc-on-plate). A pair of specimens allowed as many as 32 individual wear spots and each wear spot allowed several consecutive tests. Both, rotating and stationary specimens were

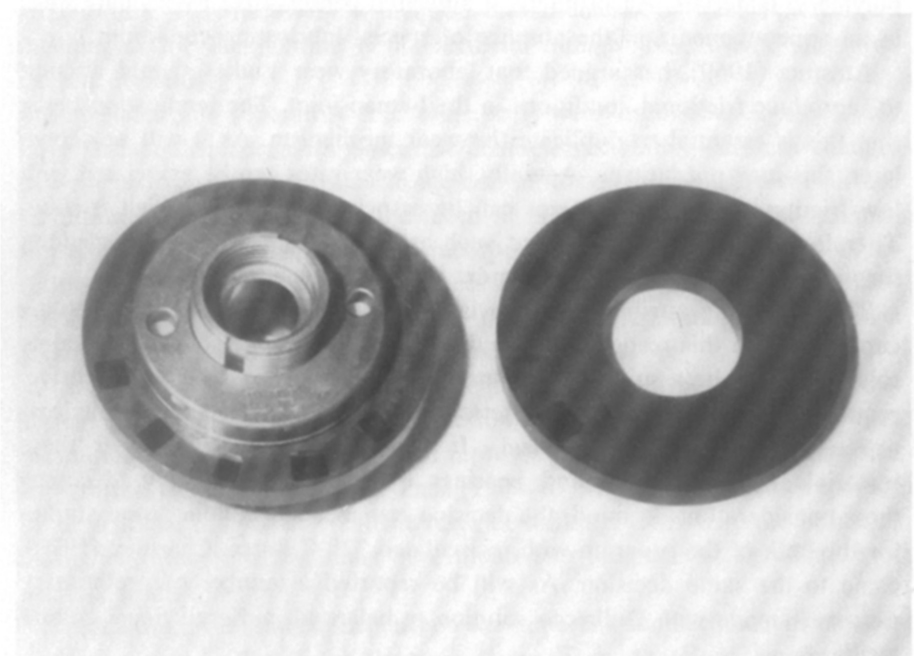

Figure 1. A specimen mounted on a collet and an unmounted specimen showing the characteristic "wear spots".

mounted in collets and the assembly fixed tightly to conically ended arbors. The specimen, and the wear pattern produced, are illustrated in Figure 1.

In the case of metals and polymers, provision was made on the test facility to dress the rotating specimen in situ. This minimized the vibrational action between the rotating and stationary specimens. The ceramic specimens were produced by suppliers in their facilities. These specimens were finished on collets supplied by us and maintained integrally with the specimen so that they could be mounted directly on the test machine.

The peripheral surface of the rotating specimen was carefully polished while rotating on the arbor of the shaft of the test machine. A high degree of polish is essential to low wear rates and must be incorporated, both, in the wear studies and in the ultimate application. The mode of polishing was such that residual surface markings were oriented circumferentially so that they had a minimal influence on wear. The quality of the finish approached that necessary for metallographic examination, i. e., better than 2 micro-inches RMS.

For this work, a test machine was specially designed and constructed* having the following attributes:

(a) A very heavy shaft and bearing system with alignment-concentricity characteristics of a precision surface grinder.
(b) Fixture for dressing specimens in place on the shaft. Items (a) and (b) minimized specimen chatter and impacting to reduce spurious wear effects and to enhance the precision of measurement of wear depth.
(c) Direct load range of 0–270 kg; sliding speed continuously variable between 500–20,000 cm per minute, power capacity of 5 HP.
(d) Immersion of the sliding interface in circulating solution at a controlled temperature.
(e) Direct and continuous monitoring of wear depth with an accuracy of 2×10^{-4} mm in the range of $0–4 \times 10^{-2}$ mm and 2×10^{-3} mm in the range of $4–20 \times 10^{-2}$ mm.
(f) Simultaneous and continuous monitoring of the coefficient of friction by means of a torque-actuated load cell on the stationary specimen.

Photographs of the instruments and machine assembly are shown in Figures 2 and 3.

* Designed and constructed by Materials Research Laboratory, Inc., Glenwood, Ill.

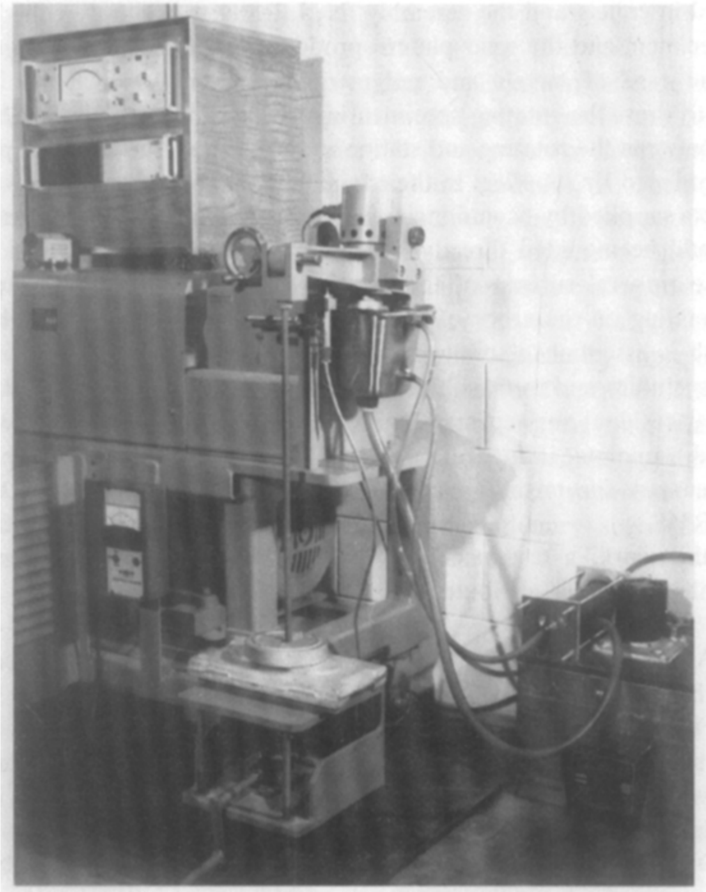

Figure 2. A view of the wear tester.

Measurements

The progress of wear is usually characterized either in terms of wear volume or wear depth. Wear volume as a measurement is more common but is meaningful only if the geometry of the wear zone is unchanged during a test and from test to test. In the present system the dimensions of the wear spot are continuously changing so that measurement of wear depth is not only the most direct measurement but also the most meaningful. The two alternative designations are interconvertible provided the geometry of the wear zone is specified.

In some wear test systems the geometries of the rotating or sliding specimen and the stationary specimen are conforming curved surfaces. Despite

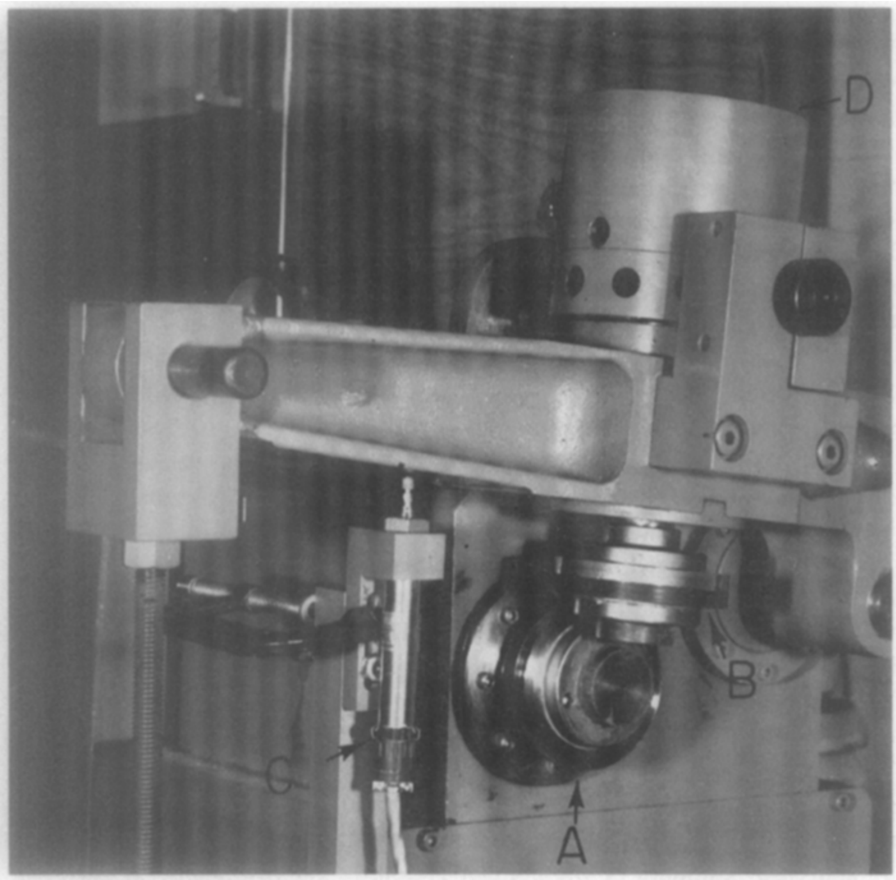

Figure 3. Close-up view of the testing assembly. The fluid circulating container has been removed. Note the rotating specimen (A), the stationary specimen (B), the differential transducer (C) and the torque actuated load cell (D).

the most careful machining and grinding some lack of conformity exists initially and wear measurement at a given contact stress can only begin after a wear-in-period. In the present system the rotating disc was allowed to wear a spot of finite magnitude before actual tests began. In this way, the wear specimens established their own conformity. From the wear depth and the simple geometry involved, the area of the wear spot could be directly computed. Wear tests were run between limits of initial and final depth from which initial and final wear spot areas were computed. An increment of wear, therefore, corresponded to an established stress range.

In this work, the wear rate d_R, for a given stress range was computed as the ratio of the depth increment divided by the total sliding distance which, in turn, was the product of the total number of revolutions and the circumference of the rotating specimen. The depth increment was assumed to be, with a high degree of accuracy, the wear of the stationary specimen only. This is reasonable since any wear on the rotating member would be distributed over the entire circumference where it otherwise would be confined to the spot on the stationary specimen.

The depth increment involved, both, wear and an elastic distortion component, especially in the case of wear testing of polymers (usually the stationary specimen). To establish that part which was due solely to wear it was necessary to make initial and final depth measurements after the load had been removed and after elastic recovery was substantially complete. It was found that a relaxation period of 60 minutes was sufficient for this purpose. In all cases, wear depths were measured in the relaxed state although the relaxation times for metal/metal and ceramic/ceramic combinations were only a few minutes at most.

As will be discussed in more detail later, the dependence of wear rate on contact pressure is a major consideration in the choice of materials and in the design of prosthetic human joints. The present wear test system was ideally suited to deriving such relationships. With a given load and a preformed wear spot it was possible to run a test for a predetermined span of contact stresses (dynamic depth measurements converted to wear spot areas). The run was stopped, a relaxed wear spot depth was taken and then the system was reloaded on the same wear spot and run for another fixed decrement of contact stress. This procedure when repeated several times provided a "staircase" estimate of the contact pressure-wear rate curve. It was also possible to change loading and arbitrarily run up or down on the "staircase", to check reproducibility. Data shown in subsequent figures are representative of both test procedures.

The data for a single wear run are presented graphically as two identical symbols joined by a line. This is meant to indicate that a certain average wear rate was computed for an experiment in which there was an initial and a final contact pressure. The span of the line between the two symbol points indicates the change in contact pressure during the term of the experiment. In most cases, this was a small span but in cases of higher wear rates the span became quite large. Some data show arrows pointing down. This indicates that the wear depth was less than the smallest resolvable reading. The arrow simply indicates that the wear rate must be less than the magnitude indicated.

Wear is a sequence of intermittent events on a microscale, each of which results in the dislodgement of a wear particle. The measurement of wear depth after a protracted wear experiment becomes a statistical summation of these events. Correlations of average wear rates have been presented graphically in the form of bands. The upper line of each band generally represents an upper bound of wear rates and the other lines represent a lower bound. This is meant to signify that the wear rates cannot be higher than or lower than the indicated values. Basically this is an extremely conservative statistical treatment. It serves the purposes of deciding on the relative merits of candidate materials Only when the upper bound of one material is below the lower bound of another material can one say, with confidence, that the former is of superior wear resistance.

Materials

The following materials were studied:

A. Commercially available materials.

Metals:
1. Stainless steel 316 L
2. Vitallium cast (Howmedica, Inc.)
3. Ti-6% Al-4% V (Titanium Metals Corporation of America)

Ceramics:
1. Pyroceram 9608, glass-ceramic (Corning Glass Works)
2. Pyroceram 0000, glass-ceramic, experimental grade (Corning Glass Works)
3. Aluminum Oxide, high density grade, A.D. 995 (Coors Porcelain Company)
4. Boron Carbide, high density grade (Carborundum Corp.)
5. Silicon Carbide, KT grade (Carborundum Corp.)

Polymers:
1. TFE (Chicago Molded Products)
2. Polypropylene
3. Polymethylmethacrylate
4. Polyallomer, Tennite 5321A (Eastman Chemical Products Co.)
5. Polyacetal, Delrin 550 (DuPont)
6. Polystyrene
7. Polycarbonate (General Electric Co.)

8. TFE + 10% graphite (Chicago Molded Products)
9. Polyimide-Vespel SP 1 (DuPont)
10. Polyimide + 15% graphite, Vespel SP 2 (DuPont)
11. Polyimide + 15% MoS_2, Vespel 3 (DuPont)
12. Polyimide + 10% TFE + 15% graphite, Vespel SP 211 (DuPont)
13. High density polyethylene (Phillips 5095)
14. Ultra-high molecular weight polyethylene (Hercules)
15. Ultra-high molecular weight polyethylene, medical grade RCH 1000 (Howmedica)

B. Laboratory produced specimens.
1. High density polyethylene (Phillips)
2. High density polyethylene-graphite composite
3. Ultra high molecular weight polyethylene (RCH 1000)
4. Ultra high molecular weight polyethylene (RCH 1000)/graphite composite

EXPERIMENTS AND RESULTS

Saline Solution as a Wear Medium

A limited number of tests were performed for comparative purposes in a standard buffered saline solution used in tissue cultures. The results are summarized in Table 1. Neither large nor consistent differences between results in water and saline solutions appeared to exist. Accordingly, all further tests were conducted using water as the fluid medium.

Preliminary Evaluation of Wear Candidates

It will be appreciated that among the possible choices of metals, ceramics, polymers, composites and combinations of these, there is a large number of options. It would be prohibitive and perhaps wasteful to obtain detailed information on the wear rates of each of these. In any search for improved materials it is appropriate to select a preliminary test or series of tests which are not very time-consuming and which give a clear indication that further study is justified.

A series of tests were run on a large number of wear couples using contact stresses of about 2 to 5.5×10^6 nt/m^2 (300–800 psi) and a sliding velocity of 8900 cpm.

A comparison of wear rates at 8900 cpm for a large number of candidate wear couple is assembled in Table 2. They are ranked in order of decreasing wear rate. Some attempt has been made to indicate the range of wear rates encountered for a range of contact stresses. Of course, the higher wear rates correspond to the higher contact stresses. The complete trends for some of these are presented in the various graphs where specific comparisons are to be made. While the low wear rates turned out to be too optimistic, the high wear rates encountered served to eliminate many candidates from further testing.

Despite their very high hardness ceramic materials do not appear to offer useful wear resistance. The Pyroceram specimens were supplied by the Corning Glass works. The material designated as 9608 represents a commercially produced composition of high hardness. The undesignated grade

TABLE 1

*Comprison of Wear rates in Water and Dulbecco Solution**

Rotating Member	Stationary Member	Medium	Wear Rate d_R, mm/mm	Sliding Velocity, cpm
316 L Stainless Steel	high density polyethylene	Water	4.8×10^{-9}	8900
		Dulbecco Solution	8.4×10^{-9}	8900
Vitallium	high density polyethylene	Water	4.9×10^{-9}	8900
		Dulbecco Solution	6.8×10^{-9}	8900
Boron Carbide	Boron Carbide	Water	3.7×10^{-9}	8900
		Dulbecco Solution	1.5×10^{-9}	8900

* per 1000 cc of diluted solution: 8 g NaCl; 0.2 g Kcl; 2.16 g $Na_2HPO_47H_2O$; 0.2 g KH_2PO_4; 0.1 g $CaCl_2$; 0.1 g $MgCl_26H_2O$. Diluted 10:1.

TABLE 2

Comparison of Wear Rates at 8900 cpm

Rotating Member	Stationary Member	Contact Stress $\times 10^6$ nt/m²	Wear Rate d_R, mm/mm
Pyroceram 9608	Pyroceram 9608	0.83–1.2	6.8×10^{-6}
Pyroceram (expt'l)	Pyroceram (expt'l)	0.83–0.92	5.1×10^{-7}
Pyroceram 9608	Vitallium	3.5–3.7	1.3×10^{-7}
High Density Alumina	High Density Alumina	3.5–3.7	1.2×10^{-7}
Vitallium	Polystyrene	3.4–5.0	1.05×10^{-7}
Vitallium	tetrafluorethylene (TFE)	2.1–4.8	6×10^{-8}
Vitallium	'polycarbonate (Lexan)	3.6–4.7	3×10^{-8}
Vitallium	polymethylmethacrylate	3.4–4.8	2×10^{-8}
Vitallium	polypropylene	3.9–4.8	5×10^{-9}
Vitallium	TFE + 10 % graphite	2.8–3.4	2.5×10^{-9}
Vitallium	polyacetal (Delrin 550)	2.1–5.5	2.5×10^{-9}
Vitallium	Vitallium	0.69–4.1	$2–10 \times 10^{-9}$
Vitallium	polyallomer (Tenite)	4.0–4.3	1.5×10^{-9}
Vitallium	Polyimide (SP-1)	4.5–5.9	$1–10 \times 10^{-9}$
Silicon Carbide (SiC)	Silicon Carbide	4.1–5.9	1×10^{-10} to 2×10^{-8}
Vitallium	UHMW polyethylene	2.8–6.2	2×10^{-11} to 2×10^{-9}
Boron Carbide (B_4C)	Boron Carbide	6.9–9.7	3×10^{-11} to 6×10^{-9}
Vitallium	Polyimide (SP-21)	6.2–9.0	3.5×10^{-11} to 2×10^{-8}
Polyimide (SP-21)	Polyimide (SP-21)	~ 9.0*	2×10^{-11}

* Contact stress uncertain.

(experimental) is a substantially higher hardness formulation which is not yet produced commercially. Pyrocerams, in general, are partially crystallized with low thermal coefficients of expansion and high moduli of rupture. The high density alumina was supplied by the Coors Porcelain Company and is representative of their AD-995 grade which is nominally 99.5 per cent of theoretical density. The specimens of high density silicon carbide (SiC) and high density boron carbide (B_4C) were supplied by the Carborundum Corp. and are representative of present commercial quality.

The importance of eccentricity control in the rotating specimen can be appreciated from the test results shown in Figure 4. One set of data corresponds to the condition where the fluctuation of the displacement indicator was as much as 1.2×10^{-2} mm; the other set corresponds to 2×10^{-3} mm of

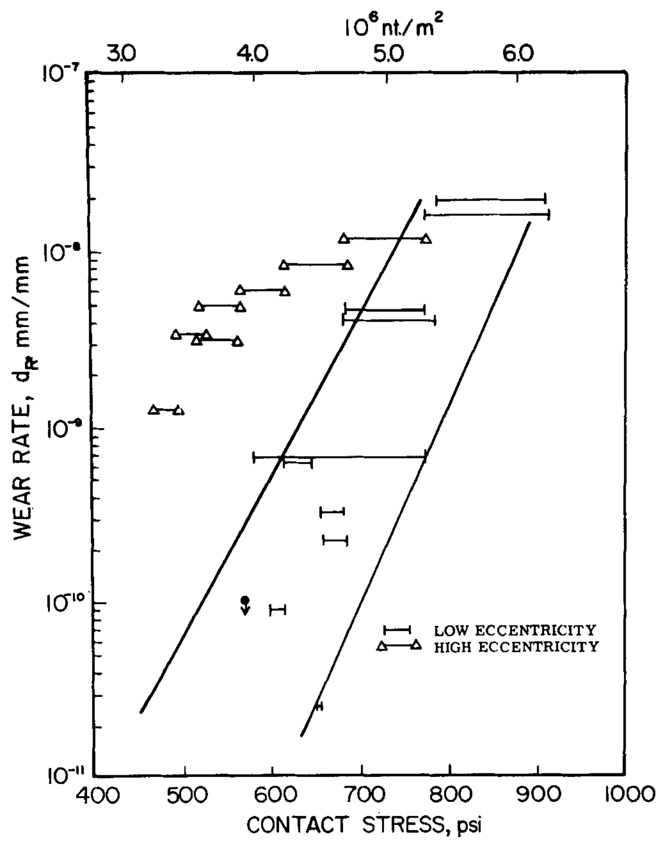

Figure 4. Wear rate vs. contact stress, 8900 cpm, silicon carbide (SiC) on silicon carbide.

TABLE 3
Comparison of Wear Rates at 635 cpm

Rotating Member	Stationary Member	Contact Stress $\times 10^6$ nt/m^2	Wear Rate d_R mm/mm
Vitallium	polypropylene	2.6–3.6	7.8×10^{-9}
Boron Carbide (B_4C)	Boron Carbide	1.4–2.1	$6–9 \times 10^{-9}$
Vitallium	polyallomer (Tenite)	2.2–2.5	5×10^{-9}
Vitallium	polyacetal (Delrin 550)	2.1–5.2	$1.5–5.5 \times 10^{-9}$
Vitallium	Vitallium	2.1–5.5	$2–4 \times 10^{-9}$
Vitallium	Polyimide (SP-21)	1.6–5.5	$1–6.5 \times 10^{-9}$
Vitallium	UHMW polyethylene	1.6–5.5	$1–2 \times 10^{-9}$
Polyimide (SP-21)	Polyimide (SP-21)	1.8–4.0	1.2×10^{-9}

fluctuation. Clearly for SiC which is a very brittle material, the impacting factor leads to exaggerated wear rates.

Elimination of noncompetitive candidates reduced substantially the number of tests to be re-run at 635 cpm. These results are presented in summary form in Table 3. At this point it seems clear that no commercial product is superior to the presently used system of Vitallium on medical grade UHMW polyethylene (RCH 1000) although several have comparable wear characteristics.

Influence of Pressure and Velocity on Wear Rate

The study of wear is not a simple subject partly because of the large number of potential mating materials and partly because of the difficulty in characterizing such important factors as surface profile, true areas of contact, local stresses, local temperatures and the nature and distribution of intervening films. There is no single mechanism that can describe the physical events involved in the generation of wear particles. Indeed, more than one mechanism can be assumed as operating simultaneously. The processes of forming wear particles must be multiple and diverse in the present context of nonprejudiced consideration of combinations of metals, ceramics, polymers and composites.

It is generally agreed that wear rates fall into two regimes which perhaps can best be termed tolerable and intolerable. The literature on the subject deals largely with two topics: (1) the factors which govern the transition from tolerable to intolerable wear rates and (2) the factors which govern the wear rates in the tolerable wear rate regimes.

The analysis by Archard (1953), is most commonly accepted as descriptive of the behavior of metals rubbing on metals. The functional relationship proposed is: $V_R = K' \cdot L$, where V_R is the volume of wear per unit sliding distance, L is the force perpendicular to the plane of sliding surfaces, K′ is a combination of terms including the probability of forming a wear fragment and the hardness of the softest metal involved.

Since we shall characterize wear for present purposes in terms of d_R, the depth of wear per unit sliding distance, Archards' function converts to: $d_R = K' \cdot P$—where both sides have been divided by the apparent contact stress $= L/A$.

Sliding velocity is not construed to be a major factor although the work of Lancaster (1963), indicates that high velocity can lead to a transition from mild to severe wear rates. Presumably the increased surface temperatures resulting from increased sliding velocity can act either to increase the rate of oxide film formation (decreasing wear rate) or to increase the efficiency of adhesion (increasing wear rate).

The Archard model and most subsequent related studies have dealt largely with the condition of dry wear, that is, in the absence of any fluid which could provide both cooling and hydrodynamic lubrication effects. Since an aqueous medium was used in this study it is to be expected that under these circumstances sliding velocity can be more influential in the determination of wear rates.

The mild wear of polymers in dry rubbing contact with metallic surfaces has been cast by Lewis (1963), in similar functional form:

$$V = K \cdot L \cdot v \cdot t$$

where
V = wear volume
L = load supported
v = sliding velocity
t = time
K = wear factor

By recognizing vt as total sliding distance, this relation converts to the Archard form: $V_R = \frac{V}{vt} = K \cdot L$—further, by dividing both sides by the apparent area of contact: $d_R = K \cdot P$—thus, in both cases, wear rate in terms of the depth of wear per unit sliding distance is linearly dependent on the apparent contact stress and seemingly independent of the sliding velocity.

Lewis (1963), implicates sliding velocity in the conditions for the transition from mild to severe wear. A Pv term is defined as a limit above which wear rates increase sharply. The Pv term becomes a property characteristic of each polymer. The implication of the Pv limit is that it defines conditions for the generation of abnormal surface temperatures with consequent degra-

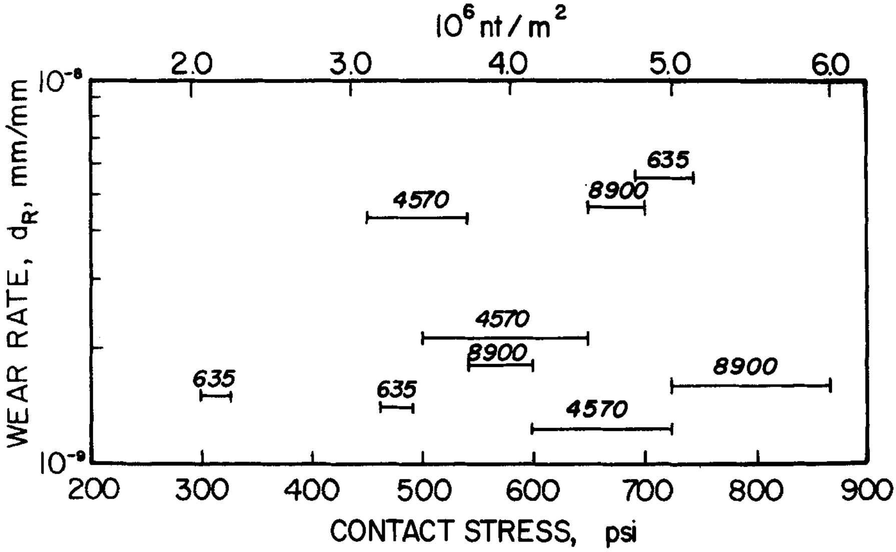

Figure 5. Vitallium (rotating) on polyacetal (Delrin 550). Comparison at various sliding speeds (cpm).

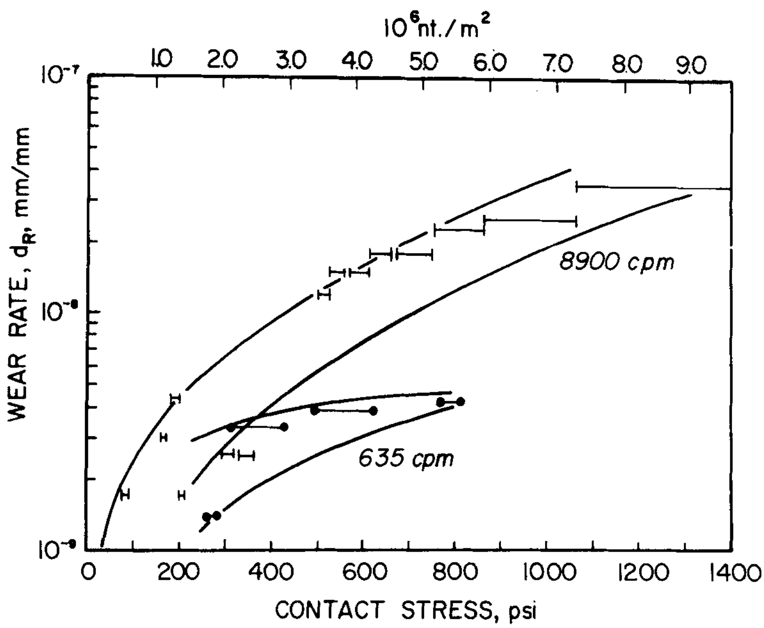

Figure 6. Wear rate vs. contact stress. Vitallium on Vitallium.

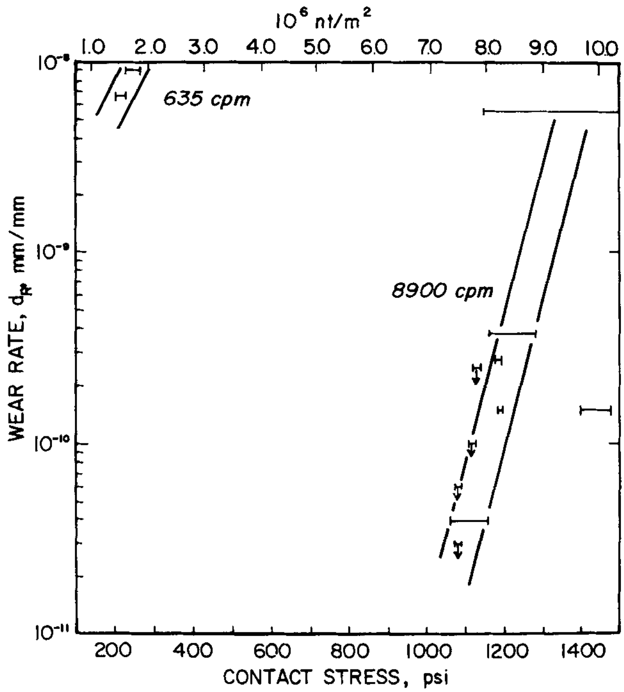

Figure 7. Boron carbide on boron carbide. Comparison of sliding speeds.

dation of mechanical properties and to the extreme, melting. In the present context of wear testing in an aqueous medium, published values of Pv limit cannot be used since they pertain only to dry wear conditions.

To begin with, the conclusion had to be drawn that existing studies and theories of wear were not particularly useful or directly applicable to derive meaningful results in the design of human joint prostheses. Accordingly, the campaign of wear studies embraced contact stresses over the entire range of likely interest and utilized two extremes of sliding velocity, 635 and 8900 cm per minute (cpm).

Figures 5 through 8 are presented as experimental wear rate data and trends illustrating the influence of contact stress and sliding velocity for three classes of wear couples—metal/metal, metal/polymer and ceramic/ceramic. Several observations can be made:

(a) Within the range of contact stresses of interest in the design of human joint replacements, very large changes in wear rate can occur. This must be a matter of great importance in the choice of the femoral head diameter for a total hip joint prosthesis.

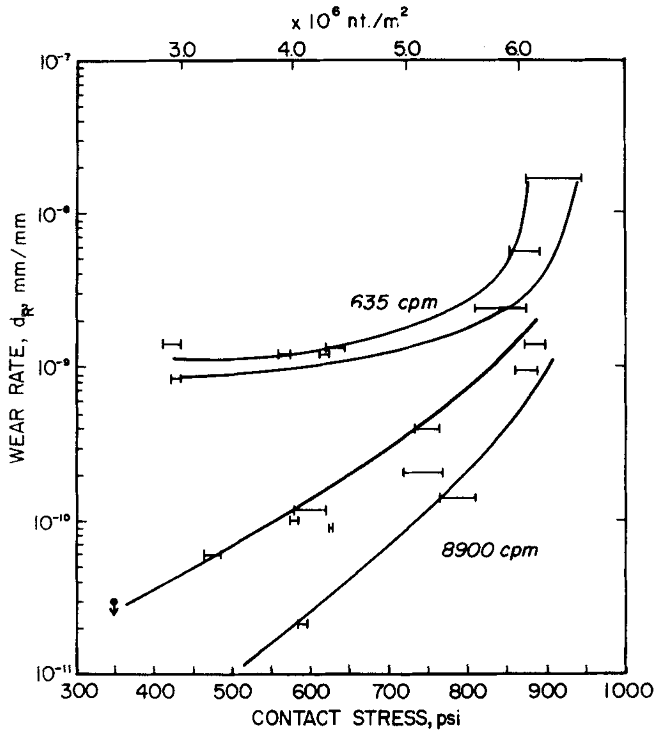

Figure 8. Wear rate vs. contact stress for Vitallium on ultra-high molecular weight polyethylene (RCH 1000) and industrial grade at 8900 cpm and 635 cpm.

(b) Wear rates can range over as much as three orders of magnitude.

(c) In most cases, reducing the sliding velocity increases the wear rate, particularly at low contact stresses. There are two notable exceptions.

The wear rate of polyacetal seems substantially unaffected by contact stress and sliding speed. (Figure 5). The wear rate of Vitallium on Vitallium seems to be decreased somewhat by low sliding velocities, (Figure 6).

The wear behavior of boron carbide (B_4C) on boron carbide, (Figure 7) is most remarkable. At 8900 cpm, the wear rate drops with contact stress to an almost immeasurable magnitude but decreasing the sliding velocity to 635 cpm increases the wear rate by several orders of magnitude.

A linear relationship between wear rate and contact pressure could not be verified in any of the cases examined. In fact, a logarithmic relationship seemed more applicable. This probably signifies that the transition from mild to severe wear lies within the zone of contact stresses and sliding velocities investigated.

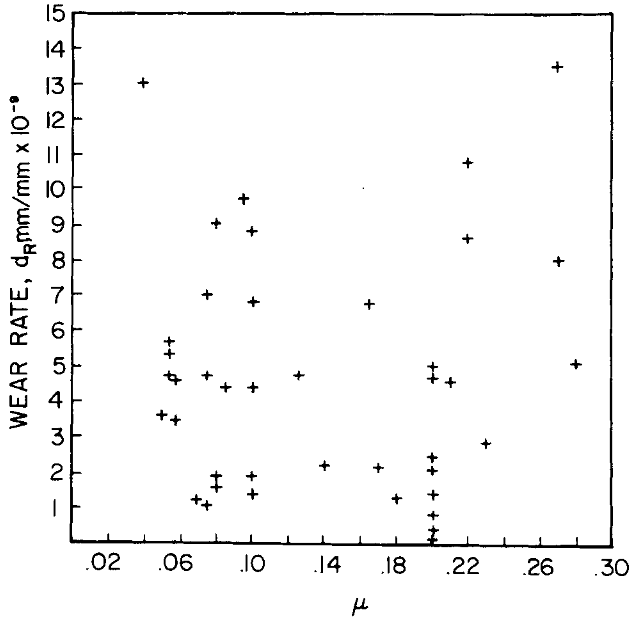

Figure 9. Correlation of coefficient of friction μ with wear rate. Vitallium (rotating) against polymers (stationary). 8900 cpm.

It seems likely that the sliding velocity effects observed relate to the immersion of the wear system in water. One can conclude that hydrodynamically entrained water between the mating wear surfaces act effectively to reduce the true area of contact. Increased sliding speed would then simply imply a thicker and more complete film of water between the opposing surfaces. For that reason high sliding velocities 8900 cpm were used only in preliminary experiments to illustrate trends. All the significant tests were performed at 635 cpm which represents approximately twice the sliding velocity encountered in the human hip joint.

The measured wear rate of Vitallium on Vitallium is of special concern because of its present use in the McKee-Farrar prosthesis. At low contact stresses of about 1.4×10^6 nt/m^2 (200 psi) the measured wear rates were comparable to those found for ultra high molecular weight (UHMW) polyethylene over a larger range of contact stresses. However, in wear testing with metal on metal, galling began immediately. This has not been observed in prostheses recovered after some years of use. These wear rates, therefore, must be regarded as the upper limit of expectation. Presumably in the absence of galling the wear rates would be substantially less.

A series of tests with Vitallium on Vitallium were run using human plasma instead of water as the fluid environment. A very significant decrease in wear rate was observed from 2×10^{-9} mm/mm to 5×10^{-10} mm/mm. It thus appeared that the protein components of plasma successfully lubricated the metallic surfaces.

The wear behavior of RCH 1000* and a comparable industrial grade of UHMW polyethylene is shown in Figure 8. The data for both fall in the same band and are not distinguished. The trend of wear vs. contact pressure at 635 cpm for RCH 1000 was used as a basis for comparison in subsequent evaluation of wear couples.

Wear Rate vs. Coefficient of Friction

Wear rates for a variety of different materials wearing against Vitallium have been plotted against the coefficient of friction measured during the test in Figure 9.

It is obvious that no correlation exists between these two parameters. A high wear rate can be associated with a low coefficient of friction and vice versa.

Choice of Metal in Metal/Polymer Wear Systems

Vitallium and 316L stainless steel are presently used in hip joint prostheses. The use of titanium for this purpose has been proposed. Although in the present work, Vitallium was primarily used in metal/polymer wear studies, some limited tests were conducted using rotary wear members of 316L stainless steel and a common titanium alloy containing 6 per cent Al and 4 per cent V.

The wear behavior of RCH 1000 in the presence of 316L stainless steel seemed unusual. Testing began at about 4.8×10^6 nt/m^2 (700 psi) and was then dropped in succeeding tests to 4.3, 3.5, 3.2 and 2.1×10^6 nt/m^2 (630, 510, 460 and 300 psi) respectively. The wear rate dropped with a decrease in contact stress from 4.8 to 4.35×10^6 nt/m^2 (700 to 630 psi) as was usual but then increased again at 3.5×10^6 nt/m^2 (510 psi) and remained at the increased level despite a reduction in the contact stress. These results are shown in Figure 10 and compared to the wear behavior with a Vitallium rotating member. Apart from the one low wear rate, the wear of RCH 1000 was greater with the stainless steel. Examination of the rubbing surface of

* Commercial designation for a medical grade of ultra high molecular weight polyethylene.

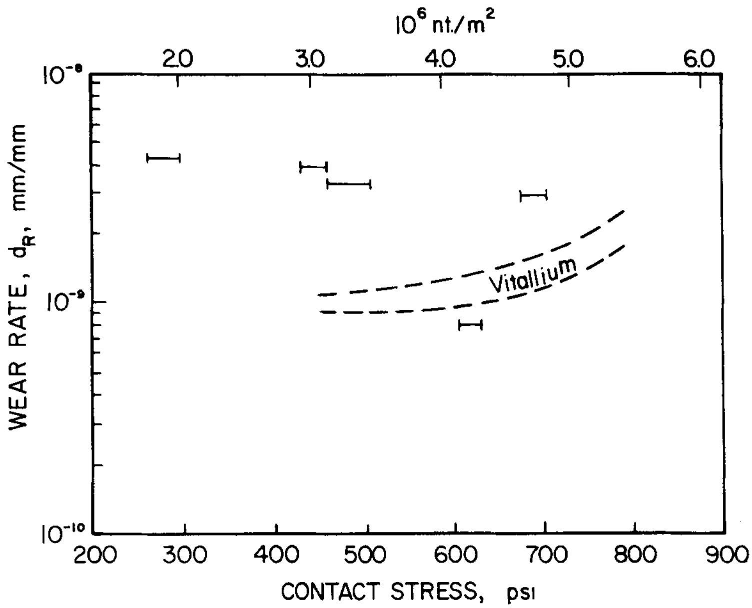

Figure 10. Wear rate vs. contact stress for 316L stainless steel on ultra-high molecular weight polyethylene (RCH 1000). 635 cpm. Comparison with Vitallium/polyethylene dotted.

the stainless steel disc after testing under a stereoptic microscope at X50 magnification revealed a faint orange cast. It is possible that the passive film on stainless steel is weakened by the rubbing action leading to some corrosion (rusting) which in turn would exert some abrasive action on the polymer. The average wear rate of RCH 1000 developed by stainless steel is quite close to the in-service wear rates quoted by Charnley (1970), with a femoral prosthesis manufactured in Great Britain from 316L stainless steel.

The wear rate of polyimide SP-21 in sliding contact with titanium was enormously increased over that generated by Vitallium. The comparison is shown graphically in Figure 11. The same trend is produced on RCH 1000. At a contact stress of only 1.6 to 1.9×10^6 nt/m² (238–277 psi) and a sliding velocity of 635 cpm, the wear rate, d_R rose to 15×10^{-9} mm/mm. This is more than ten times the wear rate produced by sliding contact with Vitallium (1×10^{-9} mm/mm).

The titanium discs were severely scored after these tests, although contact was only with polyethylene. It was also observed that the wear spot in the polyethylene was filled with black powder. This seems to be another and more emphatic example of rubbing action damaging the passive film of the metal, and a series of events involving accelerated corrosion and runaway

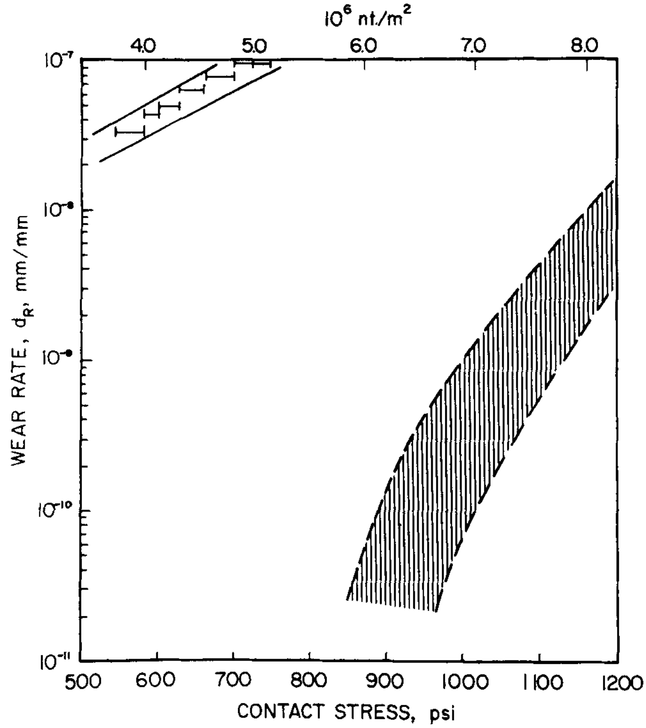

Figure 11. Wear rate vs. contact stress; 8900 cpm. Polyimide SP-21 (stationary) Titanium (rotating) compared to Vitallium (dashed).

wear produced by the abrasive action of wear products. Clearly, titanium is unsuitable as the material for the femoral head in a human hip joint prosthesis. Considering this restriction, the use of titanium in such devices may have little justification.

Wear Rates in Vitallium/Polyimide Systems

Polyimides are a fairly recent development. They are reputed to have excellent wear resistance. In the present work, the unfilled polymer SP-1* did not show wear resistance superior to RCH 1000 (see Table 2). Even more recently a number of polyimide composites have become available. These include:

SP-21*	15 weight per cent graphite,	SP-22*	40% graphite
SP-211*	10%, TFE, 15% graphite,	SP-3*	15%, MoS_2

* Du Pont products.

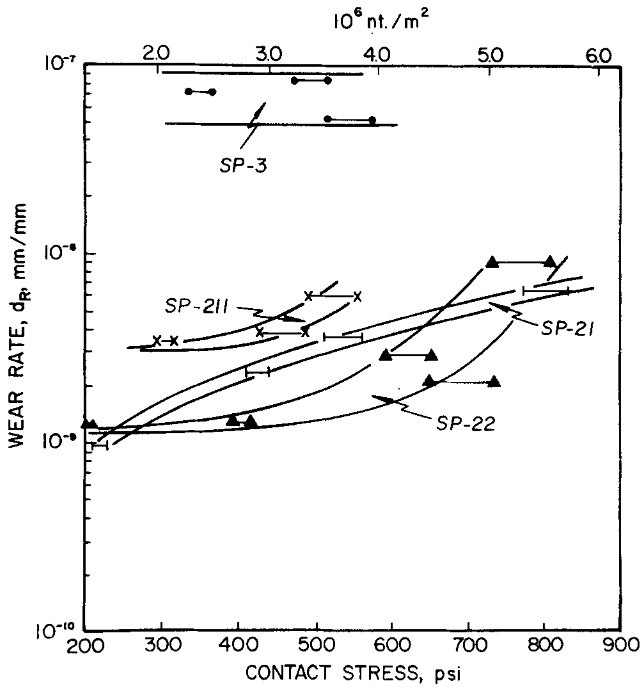

Figure 12. Wear rate vs. contact stress for Vitallium on filled polyimides. 635 cpm.

In general, these fillers were intended to reduce the coefficient friction and perhaps increase their wear resistance.

The results of wear rate measurements at 635 cpm sliding velocity are summarized in Figure 12. It is concluded that none of the polyimides provide wear resistance superior to RCH 1000.

Wear Rates of Vitallium—Polyethylene Systems

Polyethylene is not one polymeric material but a class of materials. The various grades or numbers of the class can be distinguished in terms of mean molecular weights, densities, branching frequency and capacity for crystallization. The most common commercial products are designed as follows:

low density	(0.912–0.925 g/cc)
intermediate density	(0.926–0.940 g/cc)
high density	(0.941–0.965 g/cc)

The density variations are being governed largely by the branching frequency of the chain-like molecules, the low density products exhibiting the highest branching frequency. Mean molecular weight is also a factor and in these products it ranges from 1 to 5×10^5 mer weights. Another product termed ultra high molecular weight (UHMW) polyethylene (density about 0.939 g/cc) has mean molecular chain lengths in the range of 1 to 3×10^6 mers.

High density polyethylene and UHMW polyethylene, both, are capable of a high degree of crystallization provided that the cooling rate from the molding temperature is adequately slow or the product is subsequently given an appropriate crystallization anneal; i. e., heating to a specific temperature (150°–250° C) and holding for a number of hours. Crystallization of polyethylene involves a decrease in specific volume of as much as 10 per cent. This shrinkage appears in the product as voids, unless molding pressure is sustained during the cooling cycle.

In the present work, specimens of polyethylene were both procured from commercial sources and molded in the laboratory from commercially produced powder. These specimens gave a measured density of 0.9665 g/cc.

The Charnley hip prosthesis in common use today has as one component a polyethylene cup cemented into the acetabulum. The material is a medical grade designated as RCH 1000, an ultra high molecular weight polyethylene.

Using specimens machined from industrially produced block, preliminary wear studies of high density polyethylene were made. These are summarized in Figure 13 and compared to UHMW polyethylene. Since RCH 1000 gave substantially lower rates under the same conditions and over the same range of contact stresses it was concluded that high density polyethylene was inferior for the intended application and no further tests were performed.

The test results shown in Figure 13 represent a composite of data taken from RCH 1000 block obtained from Great Britain (Thackray Ltd.) and from the United States (Howmedica Inc.) as well as from a non-medical grade (Hercules, USA). The test data were indistinguishable. However, all block materials contained random imperfections.

UHMW polyethylene is a slightly translucent material. During the machining of test specimens it was frequently observed that there were small, black opaque specks below the surface. Sometimes these particles were surrounded by a yellowish halo. Since inorganic salts of metals are frequently used as catalysts and other process control modifiers in the polymerization stage of manufacture, these specks could be such residues occluded in the final product. As such they represent a hazard to the use of this material because of their abrasive character.

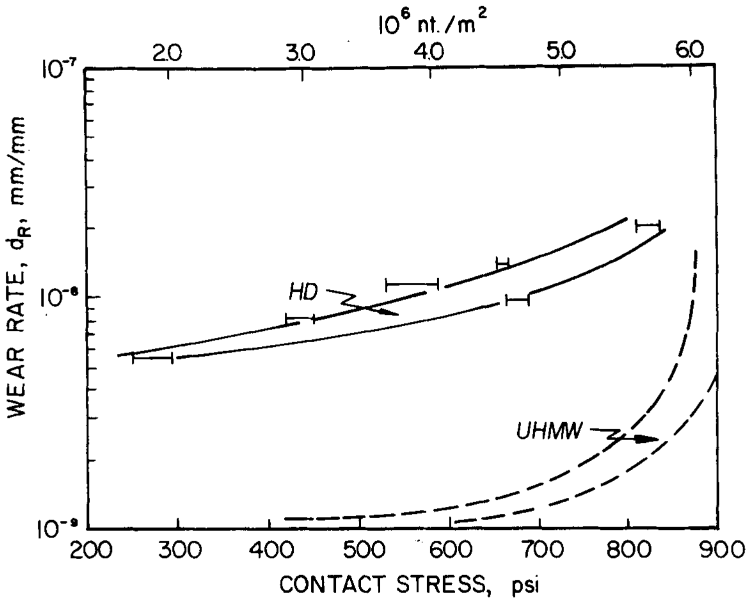

Figure 13. Wear rate vs. contact stress at 635 cpm. Vitallium rotating on high density polyethylene (HD) compared to ultra-high molecular weight polyethylene (UHMW).

One instance was encountered which may represent such a situation. Normally Vitallium retains its highly polished aspect when used as the rotating member against polymers. In one case, an abnormally high (d_R = $4–9\times10^{-9}$ mm/mm) wear rate was observed when wear testing a specimen of RCH 1000. On stopping the test and examining the surfaces involved, it was observed that the Vitallium was scored and the RCH 1000 wear spot was impregnated with a black wear product.

It seems reasonable to deduce that one of the aforementioned particles existed in the wear zone. Being abrasive it penetrated the passive film that normally protects Vitallium effectively against corrosion. A corrosion product was formed which is itself a more abrasive material. By a progression of such events the passive film was increasingly penetrated, more corrosion product was formed and wear of the polyethylene was substantially accelerated. Thus, while the speck itself may have been eliminated in the wear product at an early stage, it set off a train of events which led to abnormal wear behavior. It would seem that careful study of the manufacturing process is warranted to eliminate the possibility of this occurring in an implanted prosthesis.

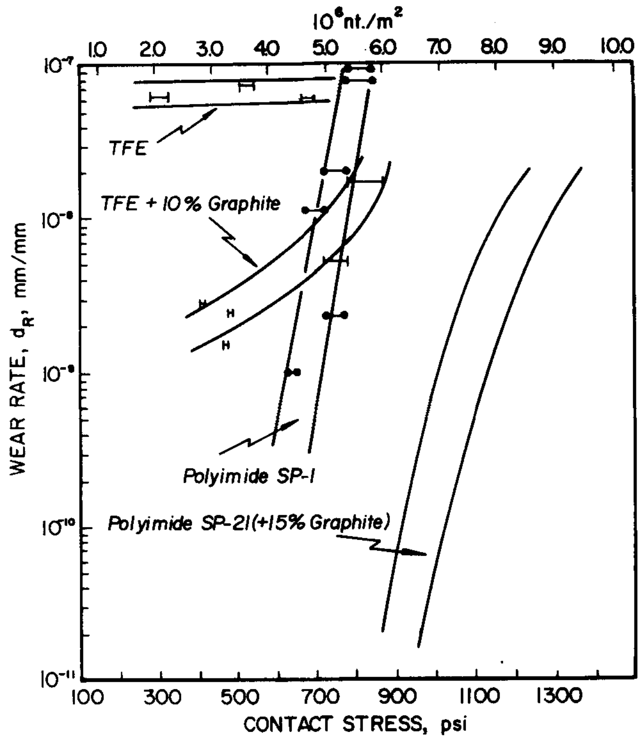

Figure 14. Vitallium (rotating) on TFE and polyimide showing the effect of graphite filling. 8900 cpm.

Wear Rates of Graphite-Filled Polymers

It has been known for some time that graphite interposed between bearing surfaces reduces wear rate. The occlusion of graphite between such surfaces has normally been accomplished by synthesizing composites with graphite contents up to 50 per cent. Thus, gray cast-iron which is a steel/graphite composite produced by the freezing process and graphite bronze produced by the powder metallurgy process have been widely used as bearing materials.

Two classes of graphite-filled polymers are commercially available: tetrafluorethylene (TFE) and polyimide. The former contains 10 weight per cent graphite and the latter can contain 15 per cent (SP-21) and 40 per cent (SP-22). Figure 14 shows comparisons of the wear rate dependencies on contact stress for two unfilled and graphite-filled materials. In both cases it is quite clear that with sufficient graphite there is a remarkable increase in wear resistance.

Wear Rates of Graphite-Filled Polyethylene

Summarizing an extensive campaign of evaluation of the wear behavior of combinations of commercial materials it must be concluded that no combination has been discovered which is superior to Vitallium interfacing with ultra high molecular weight polyethylene.

The only significant clue pointing to the direction of an improved material is the observation that graphite-filled polymers have substantially higher wear resistance than the unfilled equivalents. The possibility that graphite-filled polyethylene could provide some measure of improvement over the unfilled polymer was obvious. However, this is not a commercial product, and it was necessary to manufacture such a composite in the laboratory.

In the beginning composites were made using high density polyethylene. The polyethylene powder size was less than 500 microns. A laboratory grade of graphite powder (–38 microns) was blended with the polyethylene powder by tumbling in a glass jar for about 2 hours. Graphite powder readily coats the polyethylene particles. Cylindrical blocks 80 mm in diameter by 18 mm in thickness were molded in the laboratory. Wear specimens were machined from these molded blocks.

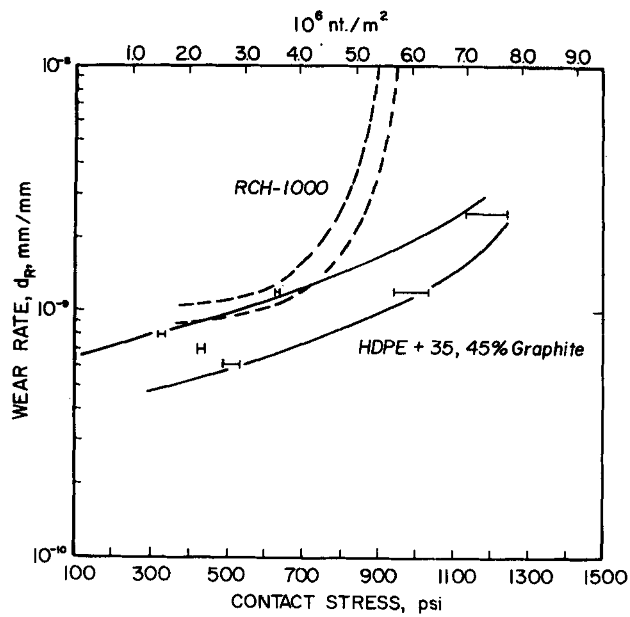

Figure 15. Vitallium (rotating) on polyethylene (stationary). Comparison of RCH-1000 with graphite-filled polyethylene. 635 cpm.

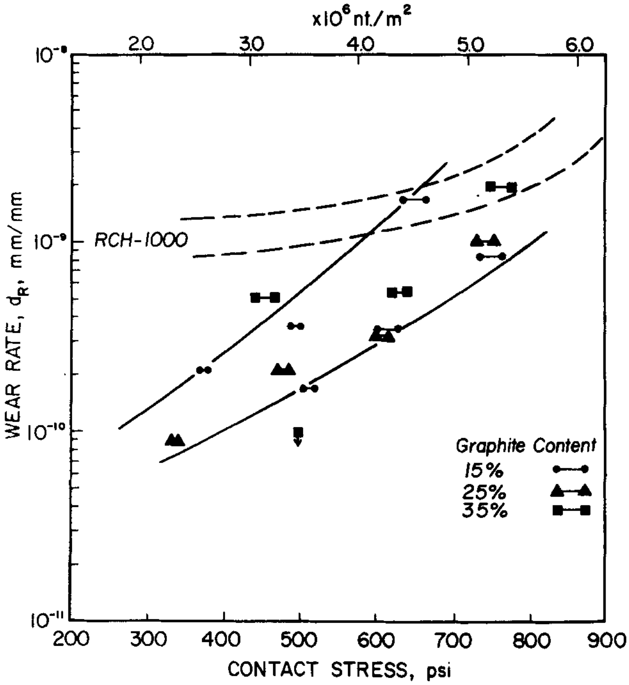

Figure 16. Wear rates of graphite-filled polyethylene (medical grade); Vitallium rotating disc. 635 cpm. Comparison with RCH-1000.

Specimens were produced containing, by weight, 15, 25, 35 and 45 per cent graphite. There was a progressive improvement in wear resistance as graphite content was increased up to 35 per cent. There was no distinguishable difference in the wear rates of 35 and 45 per cent graphite. These latter results are presented in Figure 15 where it may be seen that although high density polyethylene is inferior to RCH 1000, the graphite filled high density polyethylenes were significantly superior in wear resistance.

Subsequently RCH 1000 was used. The particle size was –200 microns. A similar series of graphite-filled composites were prepared in the laboratory.

The wear resistance of these new products appears to be outstanding. The test results are shown in Figure 16 and compared to those of RCH 1000 which is the reference for improvement. A band was drawn to include eleven out of thirteen data points irrespective of graphite content to show the influence of contact stress. Clearly there is convergence with the performance of RCH 1000 at contact stresses greater than about 5.5×10^6 nt/m^2

(800 psi) and a progressive divergence at lower contact stresses. The coefficient of friction in the contact stress range of $2.1–3.4 \times 10^6$ nt/m^2 (500 psi) varied between 0.04–0.10 for the graphite-filled grades which may be compared to 0.04–0.07 for the unfilled material.

The wear rate of graphite-filled UHMW polyethylene can be conservatively estimated as between 1/7th and in the limit of 1/30th of the wear rate of RCH 1000 at a contact pressure of 2.1×10^6 nt/m^2 (300 psi). The preferred level of graphite content is probably 25 per cent based on the data taken. This conclusion reflects the lack of scatter in the data of 25 per cent graphite composition as compared to some scatter in the data of the other two compositions. This preferred graphite content probably reflects the ability to produce and maintain a homogenous mixture during the powder handling and die filling operations. With the graphite content too low there is the possibility of clusters of polyethylene particles free of graphite and with the graphite level too high there is the possibility of graphite clusters. Both possibilities would act detrimentally on the wear behavior.

Medical grade ultra high molecular weight polyethylene filled with 25 weight per cent grade graphite satisfies the goal of a wear system that will serve in a human joint for some 30 years or more. Moreover, the only addition agent—graphite—is well tolerated in the human body so that given a graphite grade produced in a particle size and purity consistent with *in vivo* usage, there should be no serious obstacle to biological tissue acceptance.

Of great importance is the recognition that the advantage in wear resistance connotes an upper limit of operating contact stresses. This, in turn, relates to the ball size in the hip joint prosthesis and comparable radial dimensions associated with other human joint prostheses.

DISCUSSION

One of the key factors in the meaningful evaluation of a large number of materials is compression of the time scale. The system used in this investigation allowed the study of long-term performance of many different materials in a reasonable period of time. An exact simulation of *in vivo* conditions such as that provided by a "hip simulator" may at first glance appear desirable. However, that type of test is only of practical use in the final evaluation of devices or prototypes. Substantial test acceleration is not practical with it.

The conditions of testing chosen in this investigation were, in general, conservative; that is to say, they were chosen to be more severe than the actual service conditions. Consequently, some of the obvious trends observed in the laboratory may not be evident in the patient. Take as an example, the results related to the choice of metal in the metal-on-plastic type of bearing. Increased rates of wear were observed when stainless steel was used instead of Vitallium articulating with ultra high molecular weight polyethylene. This was thought to be due to a mechanical loss of the passivation layer under conditions of low oxygen concentration. With oscillating motion in an implanted total hip prosthesis this phenomena may not occur or its effect may be minimal.

It is of interest to compare the information obtained in this study with data from other investigators. Table 4 takes data from Amstutz's (1969) paper on the wear of polymers in a reciprocating wear test system and compares them to the present work. Taking into account the geometry of the test specimens, his cycling rate calculates to be about 488 cpm average sliding velocity. This is to be compared to the present results obtained at 635 cpm. His tests were performed at 6.4×10^6 nt/m^2 (932 psi) contact stress which is rather high. Our data for polyacetal reached only to 5.2×10^6 nt/m^2 (750 psi). With that exception the comparison is quite good, considering the difference in test conditions (mineral oil medium, carbon steel rotating member, oscillatory motion).

Charnley (1969), reported from measurements of radiographs of patients that the wear rate of RCH 1000 was in many instances about 0.22 mm per

TABLE 4
Comparison with Wear Rates from Amstutz (1969)

Polymer	d_R, mm/mm (Amstutz)	d_R mm/mm (Present)
UHMW polyethylene	22.5×10^{-9}	18×10^{-9}
polyimide SP-21	$2.12\text{–}10^{-9}$	7×10^{-9}
polyacetal (Delrin)	26.8×10^{-9}	5.5×10^{-9}
polycarbonate (Lexan)	7×10^{-9} 27×10^{-9}	300×10^{-9}

year. Assuming a walking activity of 9.3 km/day, 1.14 meters per step cycle, and 1.32 cm of ball sliding motion per step cycle, this wear rate converts to the method of designation used here—$d_R \cong 6\times10^{-9}$ mm/mm. In Figure 11 the wear rate of RCH 1000 against stainless steel at 635 cpm in the contact stress range of $2.8\text{–}5.5\times10^{6}$ nt/m² (400–800 psi) was given as $\cong$ 3 to 4×10^{-9} mm/mm. Thus, the laboratory wear rates compare closely to reported *in vivo* wear behavior.

One of the interesting principles of wear behavior outlined in this investigation is its close relation to contact stresses. This is obviously important in choosing materials for given applications. Wear rates that may be intolerable in total hip joint protheses may be acceptable in other joints where the contact stresses are much lower. More important are the design applications of this principle. In Figure 17 contact stresses have been plotted against femoral head diameter for a dynamic equivalent load of 454 kg. It can be appreciated that contact stresses can vary anywhere between 5.5×10^{6} nt/m² (800 psi) for a 22 mm diameter femoral head (as in the Charnley prosthesis) to around 2.1×10^{6} nt/m² (300 psi) for the 35 to 39.5 mm femoral head in the McKee-Farrar prosthesis. It should be emphazied that these are apparent contact stresses and in the implanted prosthesis the areas of contact may be smaller, particularly in metal-on-metal bearings.

One of the important considerations in the choice of the size of the femoral head is the material used. Charnley (1969), felt that the clinical success of the McKee-Farrar prosthesis was possibly related to the large size of the femoral head. Our findings support this conclusion as they indicate that wear rates are quite low for Vitallium articulating against Vitallium. 1×10^{-9} mm/mm at 2.1×10^{6} nt/m² (300 psi). The wear rates would increase significantly with femoral heads of smaller diameter. The importance of design appears to be very significant in the success or failure of these metal-on-metal prostheses. Failure can occur by the loosening of the acetabular component due to high frictional torque, and this was found to be very

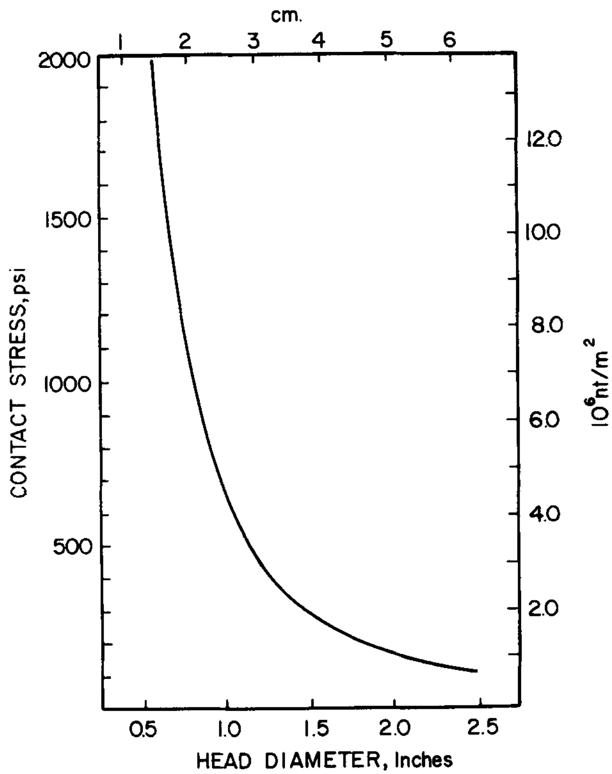

Figure 17. Contact stress vs. femoral head diameter for a dynamic equivalent load of 454 kg.

sensitive to the type of contact existing between the femoral head and the acetabulum, as reported by Walker, Gold and Wilson, Jr., (1971).

For metal-on-plastic bearings Charnley maintained that contact pressures were not important and outlined some of the advantages of a small femoral head as follows:

1) it allows a thick plastic socket decreasing the possibility of disturbing the underlying cement bond by elastic distortion of the artificial acetabulum;
2) it lowers the frictional torque;
3) it allows further medial displacement of the center of rotation of the femoral head.

Actually the relation between wear rate and contact stress depends on the polymer used. Unfilled polyimides exhibited a sharp stress dependency throughout the stress range tested. Ultra high molecular weight polyethylene showed a region between 2.1 to 5×10^6 nt/m^2 (300—700 psi) where wear rate was almost unaffected by increasing contact stress; above 5×10^6 nt/m^2 (700 psi), the wear rate increased rapidly. It is feasible that in a young, active individual contact stresses above that range with correspondingly high wear rates might be observed.

The real issue in the tissue acceptance of a bearing system is the volume of wear products produced. Unpublished evidence from this laboratory indicates that ultra high molecular weight polyethylene implanted in particulate form will induce granulomatous reactions in the tissues of rabbits.

The problem of a choice in femoral head sizes, thus resolves itself into the design that will lead to the lowest volume of wear products. The wear measurements presented in this investigation related depth change per unit sliding distance to the contact stress. For a given dynamic load equivalent (454 kg in the present calculations) the contact stress can be related to a supporting area and hence, to a femoral head diameter. Wear volumes per unit sliding distance can, therefore, be obtained by multiplying the depth change by the supporting area. The wear volumes per unit walking activity must take into account the sliding distance dependence on femoral head diameter (sliding distance will be smaller with a small femoral head). Thus, a number directly proportional to the wear volume per unit walking activity is given by the function:

$$d_R \times \frac{\pi D^2}{2} \times \frac{\pi D}{n}$$

Where d_R is the depth wear rate in mm/mm for a given contact stress, D is femoral head diameter equivalent to that contact stress for a dynamic equivalent load of 454 kg (Figure 18) and n is a numerical term involving arc of motion. For comparative purposes a walking wear rate factor can be taken as:

$$d_R \times D^3$$

This factor was plotted against femoral head diameter in Figure 19 for Vitallium bearing against RCH 1000 and against 25 per cent graphite-filled RCH 1000.

It is obvious that different materials will behave in a different manner. The ideal femoral head size using RCH 1000 is between 22 and 28 mm with a slow increase in wear volumes above that figure and a very rapid increase below that figure. From a practical standpoint the differences above 28 mm may not be significant. The data, however, supports Charnley's contention that volume wear rates are not increased and may, indeed, be lower with a

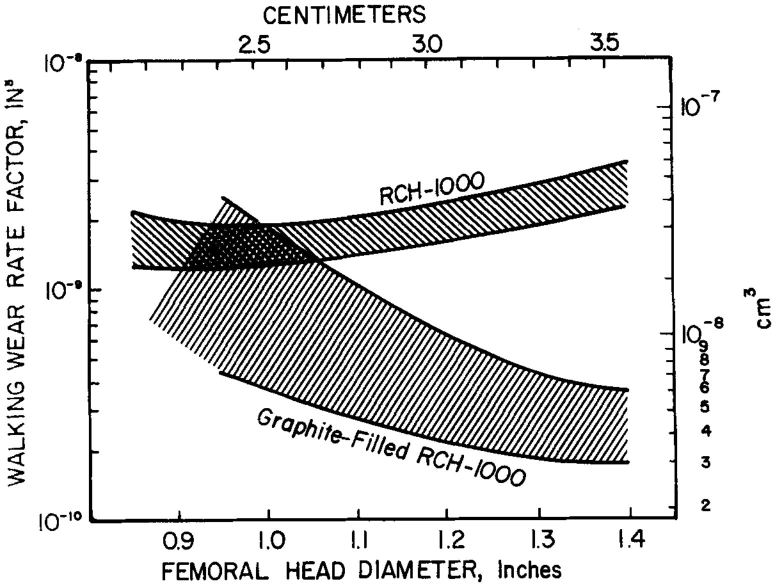

Figure 18. Wear measurements converted to a "walking wear rate factor" and correlated to a femoral head diameter.

22 mm femoral head than with a 32 mm head using RCH 1000. With graphite-filled RCH 1000 the situation is altogether different as there is a sharp dependence between volumetric wear rate and head diameter. The size of the femoral head is critical in this application. A prosthesis manufactured with this material should be designed with a femoral head between 32 and 35 mm in diameter to perform at its best.

RCH 1000 was clearly superior in its wear behavior to all the commercially available materials tested. The question is whether its wear rate is acceptable in terms of thirty or forty years of use as both, Charnley's *in vivo* results and the laboratory derived wear rates indicate that considerable wear volumes can be expected over that length of time. The behavior of the RCH 1000 — 25 percent graphite composite bearing against Vitallium is on the other hand promising. Laboratory wear rates are between 1/7th and 1/30th the ones obtained with RCH 1000. Graphite is one of the most inert materials available (National Aeronautics and Space Administration's, 1970), and although further compatibility studies are in progress, the indications are that the composite will be as well tolerated by human tissue as the unfilled polyethylene.

SUMMARY

A laboratory study of wear applied to human hip joint prosthesis was presented.

A wear testing machine was developed incorporating the following characteristics:

1) Disc-on-plate specimen configuration.
2) Alignment and concentricity characteristics of a precision surface grinder.
3) Fixture for dressing specimens in place.
4) Direct load range of 0–270 kg. Continuously variable speed between 500 and 20,000 cpm.
5) Immersion of sliding interface in circulating solution at controlled temperature.
6) Direct and continuous monitoring of wear depth and coefficient of friction.

The materials were studied at sliding speeds of 635 and 8900 cpm, in a stress range varying between 2.1 to 6.9×10^6 nt/m^2. They were immersed in water at 37° C.

Over 25 different materials and combinations of materials were evaluated including ceramics, polymers and metals. The following conclusions were derived:

1) For most materials wear rate is independent of coefficient of friction. High wear rates can be associated with low coefficient of friction and vice-versa.
2) The wear rates of ceramics including Pyroceram, Aluminum Oxide, Silicon Carbide and Boron Carbide were too high and not acceptable for use in total joint prostheses.
3) RCH 1000 (ultra high molecular weight polyethylene) bearing against Vitallium offered the lowest wear rates of the commercially available materials.
4) The choice of metal in a metal-on-plastic bearing is important in determining wear rates. Titanium and its alloys produced very high

wear rates. Stainless steel bearing against RCH 1000 produced higher wear rates than Vitallium. The process was thought to be related to a surface corrosion phenomenon.

5) Wear rates were found to be closely dependent on contact stress. This is an important consideration in prosthetic design where the size of the femoral head and hence, contact stresses can be varied over a wide range.
6) A material was developed consisting of ultra high molecular weight polyethylene with 25 per cent graphite powder. This material exhibited wear rates between 1/7th and 1/30th of those obtained with ultra high molecular weight polyethylene alone. It is suggested that its use in a total hip joint prosthesis would satisfy the requirements of a very successful and long service life.

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