The Tribological Significance of the Joint Fluid Analog in a Hip Joint Simulator

by

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ABSTRACT

Wear is the number one concern with regards to the longevity of THR (total hip replacement). Therefore, reliable in-vitro prediction of wear is necessary. Thus, the laboratory should first validate their hip simulators with known clinical materials. The limiting factor in hip wear simulation has been the joint fluid analog. Using 100% bovine serum as the joint fluid analog, UHMWPE (ultra-high molecular weight polyethylene) wear-rates have been continually underestimated and PTFE (polytetrafluoroethylene) wear has been overestimated. Therefore, this work investigated the effect of protein concentration in bovine serum on the wear of PTFE and UHMWPE in a biaxial hip joint simulator. Validation criteria were developed based on the clinical findings of: ball size effect of increased wear with increased head size, 6% increase in wear for each millimeter of increased head diameter, clinical wear magnitudes, PTFE/UHWMPE wear-rate ratio and debris morphology. Both materials duplicated the clinical criteria using bovine serum with 10mg/ml of protein concentration. As protein concentration went from 0 to 10mg/ml, wear of both materials increased, however with greater than 10mg/ml protein; a) the rate of increase for PTFE was reduced by 80% and b) the wear of UHMWPE reversed, thus, showing that proteins cause wear. Additionally as the volume of fluid was increased, wear increased. This change in wear with protein concentration and volume was due to a protection of protein precipitate. As protein concentration increased protein precipitation increased and wear was decreased due to a protective layer of precipitates. Furthermore, wear protection was dependent on the amount of protein precipitation which was in turn, dependent on the initial concentration, volume of fluid and time. Therefore, wear in-vitro was dependent on the joint fluid analog. This work proved that the laboratory could duplicate clinical findings using bovine serum with 10mg/ml of protein concentration as the joint fluid analog and thus increase confidence in wear evaluation; taking the first steps to showing reliability of in-vitro THR wear studies.

To

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who brings all good things to my life
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Table of Contents

Abbreviations

ACS	alpha calf serum
Al	aluminum
Al_2O_3	aluminum oxide (alumina)
Anat	anatomical
Anat/horiz	anatomical horizontal
Anat/obliq	anatomical oblique
BCS	bovine calf serum
C	cemented
CL	confidence limit
Clin	clinical
CoCr	cobalt chromium alloy
D	dynamic soak
D_a	Diameter of femoral head ìAî
D_c	Diameter of reference head (Charnley 22.25 mm)
DI	de-ionized water
EDTA	ethylenediamine tetraacetic acid
γ	gamma irradiation
HSS	Hospital for Special Surgery (New York, USA)
Inv	inverted
JFA	join fluid analog
LFA	low friction arthroplasty
LST	linear surface tracking
MB	metal backed
Mc	million cycles
M-dST	multi-directional surface tracking
NA	not applicable
NaH_3	sodium azide
NC	non-cemented
NMB	no metal back
NR	Not Reported
PCA	porous coated anatomic total hip replacement
PE	polyethylene (all references are to ultra high molecular weigh)
PMMA	polymethyl methacrylate
PTFE	polytetrafluoroethylene
R_a	parameter of surface roughness
S	static soak
SS	stainless steel
THR	Total Hip Replacement
Ti	titanium
UHMWPE	ultra high molecular weight polyethylene
V_a	Volumetric wear-rate of femoral head ìAî
V_c	Volumetric wear-rate of reference head (Charnley 22.25 mm)
VWI	volumetric wear index – the ratio of wear-rate for any size implant to the wear-rate of the Charnley 22.25mm implant
Zr	zirconia

CHAPTER 1. INTRODUCTION

The demand for total hip arthroplasty (the replacement of the natural hip joint with an artificial joint) has grown in the last 30 to 40 years of widespread general practice. This surgery was once only considered for older patients, who probably would not outlive the 10 year lifetime of the implant. However, younger more active patients who want their quality of life improved are now demanding this surgery (Black, 1997, Schmalzried et al, 1998 and Weightman et al, 1991). Over a five year period from 1990 to 1995, the total number of implants in the US alone had increased by 44% and of these almost one quarter were implanted in "young" patients with a life expectancy of greater than 25 years (Black, 1997). This has created a need for new implants with expected lifetimes greater than that of the younger patient. In order to validate these new implants they must be thoroughly tested in the research laboratory before implantation in the patient. Therefore, the reliance on the laboratory to pretest for expected clinical outcome has become more important.

The most widely used total hip replacement materials have been a socket or cup made of ultra-high molecular weight polyethylene (UHMWPE or PE) and a head made of metal (stainless steel, SS or cobalt chromium alloy, CoCr). The metal on plastic combination was developed by Sir John Charnley in the 1960's along with the introduction of cement to fix the cup and stem to the bone interface. The Charnley hip today consists of a 22mm UHMWPE acetabular cup with a CoCr or high strength steel head. Other materials have been used with varying degrees of success and failure but the Charnley hip remains the "Gold" standard. However, this 'Gold" standard and other implant materials have limitations on the lifetime of the

prosthesis. Many studies agree that the survivorship for the Charnley hip was greater than 90% at 10 years (Alsema et al, 1994, Malchau et al, 1993 Smith et al, 1998). For long-term this survivorship can continue to be very good, greater than 90% at 25 years if the wear-rate is very low (<0.1mm/y). However, with greater wear-rate survivorship deteriorated rapidly to only 30% at 20 years with none predicted to survive more than 25 years (Sochart, 1999). The survivorship of other recent UHMWPE's has not been as good. Hylamer™, an UHMWPE with increased crystallinity, was expected to reduce wear and increase the lifetime of the prosthesis. But this was not the case and Chmell et al, (1996) reported that survivorship at 4 years was only 86%. The UHMWPE, Porous-Coated Anatomic hip (PCA), showed a survivorship of only 44% at 9 years due to design problems (Owen et al, 1994). Whether these materials were tested before clinical use is unknown. The only laboratory publications for these two materials were after clinical release (DePuy Brochure, 1994, McKellop et al, 1992 and Saikko, 1994). Therefore it is not clear whether these materials had undergone adequate testing before implantation.

The greatest factor affecting survivorship is the wear of the implant (Charnley and Halley, 1975, Clarke, 1981, Cooper et al, 1992, Griggs et al, 1996, Livermore et al, 1990, Ohlin and Persson, 1989 and Weightman et al, 1991). Wear is not the only limitation for implant longevity, but for a well-fixed, surgically competent procedure, wear becomes the most important factor. Wear of an UHMWPE socket produces 75% small round micron to sub-micron size particles ranging from 0.07 to 6.3μm in diameter and 25% elongated fibrils and shreds which range from 0.5 to 12 μm in length and 0.2 to 0.3μm wide (Campbell et al, 1996, McKellop et al, 1995 and Shanbhag et al, 1993). These particles number in the

millions. The foreign body response to these particles has been associated with loosening and bone resorbtion or osteolysis and ultimate failure of the hip implant necessitating revision (Campbell et al, 1996, Cooper et al, 1992, Howie, 1990, Sochart and Porter, 1998, Willert et al, 1981 and Xenos et al, 1995). The osteolytic response and subsequent failure is directly proportional to the amount of wear, therefore reduction in wear would reduce this response and in turn reduce the number of failures and revisions. Consequently, with the concern over wear of implants and the risk of failure and revision to the patient it is imperative that adequate laboratory testing be accomplished before new implants go to market.

Historically, the limitation in many laboratory studies was their use of a simple linear surface tracking (LST; linear refers to no crossing path on the wear surface) machine (Charnley, 1976, Dowson, 1978, Homsy and King, 1969, McKellop et al, 1977 and Seedhom et al, 1973) or single station simulators (Buchholz & Strickle, 1972, Duff-Barclay and Spillman, 1967, Ungethum, 1976, Walker et al, 1968 and Wright & Scales, 1977). LST devices did not use an actual prosthesis so the effect of varying ball sizes could not be reproduced. These machines have also shown instances of incorrect ranking as in gamma sterilized versus non-gamma sterilized PE (Fisher, 1995). In contrast, simulators used an actual implant and could test for variables such as the ball size effect. However, simulators were more complicated compared to the LST devices and their use was limited by testing capacity, i.e. most simulators only accommodated one specimen, whereas the LST devices could generally test 3 or more. The laboratory had either to invest in more machines or endure longer test duration in order to test more than one THR combination. For example a test of 2 specimens (control and experimental) for

a total of 3 million cycles each would take from 3 to 6 months, depending on the speed of the simulator. Beginning in 1980 at the Vernon Luck Laboratories (Orthopaedic Hospital, Los Angeles, California) multi-station, computer controlled simulators were introduced with the capacity to test greater numbers of THR combinations. This meant tests of multiple designs of specimens could be compared simultaneously. It also meant that tests could be completed within shorter amounts of time, which was important to manufacturers desiring to market a new device. Therefore, the laboratory was capable of wear testing more hip implants and at a faster rate than had been previously possible.

The major limitation for adequate simulation was that of the human joint fluid. The amounts of fluid in the joint are minute with the average healthy knee having only 0.2 to 2 ml of fluid (Binette and Schmid, 1965, Geigy Tables, 1981 and Yehia and Duncan, 1975). In joints suffering from disease, the fluid amount increases and can be as much as 30ml in an inflamed joint (Geigy Scientific Tables, 1981 and Rippey, 1979). But even this amount would not be enough to fill a multi-channel hip simulator. Chamber sizes in hip simulators can vary from 20 to 600ml per specimen. Therefore, one wear test would require 2,400 to 72,000ml of synovial fluid to fill 10 chambers (3 million cycles test duration). So, for one 3 million-cycle study 80 to 1400 donations from patients would be needed. Therefore, the laboratory has had to use alternative joint fluid analogs (JFA) such as water, saline or bovine serum. However, these are very different fluids, two without proteins and one with proteins that have produced contrary wear-rate results in hip simulators (Clarke et al, 1995, Dowson and Jobbins, 1988, Duff-Barclay and Spillman, 1967 and Wright and Scales, 1980).

Thus, the validity of simulators has been in question. Studies have typically underestimated UHMWPE and overestimated polytetrafluoroethylene (PTFE) wear-rates in both water and bovine serum (Charnley, 1976, Duff-Barclay and Spillman, 1967, Dumbleton et al, 1974 and McKellop et al, 1983). In addition the PTFE/PE ratio based on Charnley's clinical data, has been greatly overestimated. In-vitro studies have produced conflicting results with respect to increased wear-rate with increased head size (ball size effect; Clarke et al, 1996, McKellop et al, 1995, Pappas et al, 1995 and Wright and Scales, 1980) and debris from tests conducted in water for either PTFE or UHMWPE has been shown to be totally inaccurate (Dowson and Jobbins, 1988, Gold and Walker, 1974, McKellop et al, 1978 and Wang et al, 1995). This discrepancy between laboratory and clinical results was noticed as early as 1969 when Charnley realized that the laboratory PTFE/PE ratio was off by a factor of 12 fold (Charnley, 1976). Then, in 1977, Swanson reported on simulators and their significance to clinical materials and stated, "...no significant feature of current practice has been based on results obtained in simulators." Therefore, until the laboratory can reproduce known clinical values, the reliability for predicting wear-rates of new materials will be suspect.

It would, therefore, appear that the laboratories have not been diligent enough in validating the in-vitro model against known clinical criteria. It has been suggested that the laboratory evaluate their wear testing machines based on known clinical materials such as UHMWPE and PTFE (Clarke, 1981 and Unsworth, 1995). These two materials were on the opposite side of the polymer wear spectrum with catastrophic wear found for PTFE and UHMWPE producing at least 20 times less wear. Therefore, these materials test both upper and lower polymer wear limits of

the simulator. There has been ample in-vivo wear-rate magnitude data on UHMWPE and PTFE was well documented by Charnley. Thus these 2 materials provide the laboratory with excellent wear-rate models with which to validate hip simulators. In addition to clinical magnitudes, four more clinical criteria can be established with which to validate simulators, using PTFE and UHMWPE. Charnley called attention to the fact that the wear-rate ratio for PTFE/PE was too high in the laboratory (Charnley, 1976). He also noted that the wear-rate of PTFE increased linearly with respect to head size and at a rate of 6% per millimeter of head size (Charnley et al, 1969). This same finding has been shown for UHMWPE (Livermore et al, 1990). Finally, the debris from both materials can be compared to debris from the simulator (Campbell et al, 1995, Charnley et al, 1969, Howie, 1990 and Willert et al, 1981). Thus, establishing 5 clinical tools with which to validate hip simulators:

1) wear-rate magnitude,
2) PTFE/PE wear-rate ratio,
3) ball size effect
4) 6% wear-rate increase per millimeter of head size
5) debris characteristics.

Charnley was the first to attempt validation of his pin on disk machine using UHMWPE and PTFE but was unsuccessful (Charnley, 1976). Other studies have also used these materials for validation but none has reproduced the clinical magnitudes or the PTFE/PE wear-rate ratio (Clarke and McKellop, 1980, Gold and Walker, 1974, Homsy and King, 1969, McKellop, 1981, McKellop et al, 1977, McKellop et al, 1978 McKellop et al, 1981). Duplication of all 5 criteria has not

been reported in the laboratory using any simulator or with water, saline or bovine serum as the JFA.

CHAPTER 2. LITERATURE REVIEW

2.1 ULTRA HIGH MOLECULAR WEIGHT POLYETHYLENE (UHMWPE)

The beginning of polymer use for hip implants gained notoriety with the PTFE cup put into service by Sir John Charnley in the 1950's (Waugh, 1990). Within a few years after Charnley first implanted PTFE hip cups it became very clear that this material was wearing so badly that failure was imminent. Instead of giving up he searched for another material with which to replace the failing PTFE and was soon convinced that ultra-high molecular weight polyethylene (UHMWPE) was the answer.

Charnley began using non-gamma sterilized UHMWPE (RCH1000) articulating against a stainless steel head in November 1962 (Waugh, 1990). These implants were made in the machine shop at the Wrightington Hospital where Charnley worked and the cups were sterilized in formaldehyde. After March 1967 Thackray Ltd. began manufacturing the implants and supplied them sterilized with gamma irradiation in air (Wroblewski, 2000). Thus, the era of gamma sterilized UHMWPE cups began in March of 1967.

From Charnley's previous work with PTFE he developed the concept of the low friction arthroplasty to reduce cup loosening. This concept utilized reducing the inner diameter of the cup to 22mm in order to decrease the frictional torque between the cup and bone interface. This design had an additional benefit of decreasing the UHMWPE wear-rate. This concept has proved to be successful throughout the last 35 years with wear-rates of approximately $50mm^3/y$ and 90% survivorship at 10 years (Charnley, 1973, Griffith et al, 1978, Hamilton and Joyce, 1986, Malchau et al, 1993 and Sochart and Porter, 1998). Thus, the Charnley Low Friction Arthroplasty

(22mm RCH1000 against stainless steel or CoCr) has become the "gold" standard for hip implants.

With the success of the Charnley hip came an increased demand for hip implants. In Sweden the number of implants increased at a rate of 26% per year from 1970 to 1991 (Malchau et al, 1993). From 1990 to 1995 in the US alone the number of primary hip implants increased by 44% (Black, 1997). The number of younger patients also increased and with this new demand came the necessity for longer lasting implants (Black, 1997, Schmalzried et al, 1998 and Weightman et al, 1991). Therefore, the need for improved implants, which wore less, became imperative. Since Charnley first implanted UHMWPE new designs or new materials have been introduced with the hope of improving the wear resistance of UHMWPE. These changes have included size, shape, material and fixation. Some of the resulting designs included the 25, 29 and 35mm Stanmore Prosthesis; 28mm Trapezoidal-28; 26 and 32mm PCA; 28 and 32mm Hylamer; 32mm Mueller and the 38mm Buchholz (Amstutz and Clarke, 1991, Chmell et al, 1996, Cooper et al, 1992, Newman, 1971, Owen et al, 1994, Scales and Lowe, 1971 and Thomas et al, 1996). In addition new methods of fixation were applied such as non-cemented porous coatings, meshes, beads and press fit designs (Amstutz and Clarke, 1991). None of these designs has proved to be better than the 22mm Charnley cemented low friction arthroplasty (table 2.1), and some such as PCA and Hylamer™ have proved to be much worse (Chmell et al, 1996, Cooper et al, 1992, Owen et al, 1994 and Thomas et al, 1996).

Table 2.1 In-vivo UHMWPE Studies

Average age refers to age at implantation. SS = Stainless Steel, Al = Alumina, Ti = Titanium, LFA = Charnley Low Friction Arthroplasty, C = cemented, NC = non-cemented, MB = metal backed, NMB = no metal back

Author & Year	*Dia. (mm) & Head Material*	*No. of Hips*	*In-vivo (y)*	*Avg. Age (y)*	*Implant Type and Fixation*	*Vol. Wear-rate (mm^3/y)*	*Comments*
Charnley, 1972	22/SS	7	5.4	NR	LFA/C	50	From autopsy
Charnley & Cupic, 1973	22/SS	72	9-10	73	LFA/C	47	86% avg. 0.09mm/y, 14% avg. 0.3mm/y
Charnley & Halley, 1975	22/SS	59 4	3 5	<30 <50	LFA/C	70 39	young/limited activity young no limitations
Scheier & Sandel, 1976	28/CoCr	55	2.5			61	
Griffith et al, 1978	22/SS	491	8.3	60-69	LFA/C	27	62% <0.06mm/y
Atkinson et al, 1985	22/SS	25	8.9	45	LFA/C McKee Howse	74	25 out of 82 hips analyzed doesn't state which are used in analysis
Isaac et al,1989	22/SS	50 Rt 50 Lt	9.1	50.4	LFA/ C	49 63	Bilateral hip replacement on 50 patients
Livermore et al, 1990	22/CoCr 28/SS 32/CoCr	227 98 60	>9.5		LFA/C Trapezoidal/C Muller /C	47.5 48.4 84	All PE was compression molded from same manufacturer
Weightman et al,1991	22/?	12	8.5	59	LFA/C	93	Retrievals
Oonishi et al, 1992	28/SS 28/Al	15 73	>6	40-70	Trapezoidal/C Kyocera/C	154 60	
Cates et al, 1993	28/Ti6Al4V	134 99	6 6.8	69 67	C/MB C/NMB	66 48	
Kabo et al, 1993	22/? 26/? 28/? 32/?	5 3 23 9	14 13 10 10	52 44 50 55	Mixed	26 63 76 89	Retrievals from revision
Hamilton & Gorczyca, 1995	22/SS	195	>10	58	LFA/C	50	
Sugano et al, 1995	28/Al	57	11.1	53	Bioceram 4& 5/C/NMB	62	
Devane et al, 1995	26/CoCr 32/CoCr	85 56	5.6	61	PCA/NC/MB	67 99	
Chmell et al, 1996	28/CoCr 28/CoCr	3 3	3.1 2.9	30 62	Hylamer™ NC/MB	250 338	Revision
Callaghan et al, 1995	22/SS 22/SS 28/CoCr 28/CoCr 28/CoCr	23 61 20 43 63	20+ 15+ 10+ 7-8 5-7	57 60 67 66 69	LFA/C LFA/C Iowa/C Iowa/C/MB Iowa/NC/MB	41 35 71 66 41	Machined PE Molded PE Molded PE Molded PE Milled PE
Sychterz et al, 1996	32/Cer 32/CoCr 32/CoCr	5 13 8	5.5 5.8 11.5	65 70 72	Mixed/MB Mixed/MB Mixed/C	53 46 22	Post Mortem Well functioning
Kesteris et al, 1996	22/CoCr 32/CoCr	33 34	7.5 8.0	59 61	C/NMB	57 148	Scanhip™ Head
Hall et al, 1996	22/SS	129	10.7	56	LFA	51	explants
Wroblewski et al, 1996	22/Al	19	8.3	51	XLP/C	22	

Many of the early designs of THR, such as the PTFE hip, were implanted without any laboratory testing and it was uncertain whether some of the later THR such as PCA had been tested before clinical use (Sauer and Anthony 1998). Wright and Scales, (1980) commented, "Although it is not yet mandatory it is inexcusable to use total hip prostheses in man that have not been adequately tested in the laboratory." However, these pre-tests would not help the patient if the results from the testing devices were unreliable. Therefore, in-vitro duplication of known clinical results was the first step to predicting better reliability of the testing device. The long history of UHMWPE with many published clinical studies on wear-rates (table 2.1) and debris characteristics has made it an obvious choice as an in-vitro validation tool. Therefore, UHMWPE has been shown to be one of the premier implant materials and serves as an excellent clinical validation model for the laboratory.

In-vitro studies of UHMWPE have typically underestimated clinical wear-rates (Duff-Barclay and Spillman, 1967, Dowson and Jobbins, 1988, Dumbleton et al, 1974, Fisher et al, 1995, McKellop and Clarke, 1985 and Saikko et al, 1993). It has been suggested that the "clean" environment of the laboratory versus the 3rd body wear that can occur in the body was the reason for these lower wear-rates (Bigsby et al, 1997 and Schmalzried et al, 1999). Third-body wear due to cement, bone, cup or head particles entering the THR interface was attributed to greater surface roughness of the femoral component. Many in-vitro studies have been conducted to elucidate the subject of 3rd body wear (table 2.2). Studies have added bone cement (polymethyl methacrylate or PMMA) and particles of bone to the JFA. Other studies have roughened the femoral head to simulate 3rd body damage. Conclusions have been contradictory.

Table 2.2 In-vitro and In-vivo Studies on the Effect of 3rd Body Wear and Surface Roughness of the Femoral Head

Clin = clinical, R_a = Parameter of Surface Roughness, PMMA = polymethyl methacrylate

Type	*Author*	*Year*	*Machine Type*	*Treatment*	*Result with respect to control*	*Wear Comparison*
3rd Body	Taylor et al	1999	Bi-axial simulator	PMMA	Wear < clinical	<Clin
3rd Body	Polineni et al	1999	Bi-axial simulator	PMMA	Wear b3-fold	>Clin
3rd Body	Essner et al	2000	Bi-axial simulator	1mg/ml PMMA 5mg/ml PMMA 10mg/ml PMMA	Wear c2-fold Wear No corb Wear b 2-fold	<Clin =Clin >Clin
3rd Body	Laurent et al	2000	Bi-axial simulator	PMMA Alumina particles	Low wear High wear	<Clin >Clin
3rd Body	Murtaloglu et al	2000	3-axis simulator	Alumina particles PMMA	Wearb 7-11-fold & R_a b 12-17-fold Wear b 0-2-fold & R_a b 2-fold	=Clin <Clin
3rd Body	Caravia et al	1990	Pin on disk	Various particles	Most damage PMMA/ Zr	
3rd Body	Que & Topoleski	1997	Pin on plate	Bone and Bone cement	R_a b 3 to 4-fold	
3rd Body	Minakawa et al	1998	Pin on plate	Various particles PMMA PMMA with Zr	No change b 2-fold	
3rd Body	McNie et al	1999	Pin on plate	Various sizes of SS particles	b damage with particles >150μm	
Rough Heads	McKellop et al	1995	Bi-axial simulator	Retrieved Heads R_a b 3 to 10-fold	Wear b 1.3-fold & R_ac40 to 80%	<Clin
Rough Heads	Wang et al	1998	Bi-axial simulator	R_a b 3-fold R_a b 8-fold	Wear b 1.5-fold Wear b 6-fold	<Clin =Clin
Rough Heads	Essner et al	1998	Bi-axial simulator	R_a b 10-fold	Wear b 2-fold	No Clin value
Rough Heads	Polineni et al	1999	Bi-axial simulator	R_a b 13-fold	Wear b 2-fold	<Clin
Rough Heads	McKellop et al	2000	Bi-axial simulator	R_a b 10-fold	Wear b 0.9-fold	<Clin
Rough Heads	Cooper et al	1993	Pin on plate	R_a b 3-fold	Wear b 40-fold	<Clin
Rough Heads	Fisher et al	1994	Pin on plate	R_a b 1.4-fold	Wear b 20-fold	<Clin
Rough Heads	Besong et al	1997	Pin on plate	R_a b 10-fold	Wear b 45 to 242-fold	<Clin
Rough Heads	Wang et al	1998	Pin on plate	R_a b 10-fold R_a b 69-fold	Wear b 60-fold Wear b 1000-fold	<Clin =Clin
Clinical	Cooper et al	1993	Retrievals	R_a b 2-fold	Wear b 1.8-fold	
Clinical	Hailey et al	1997	Retrievals	R_a b <4-fold R_a b >4-fold	No change Wear b 2-fold	
Clinical	Schmalzried et al	1997	Retrievals	R_a b 4-fold	Wear b 1.5-fold	
Clinical	Morsher et al	1998	Retrievals	particles 100–200μm embedded in PE	Severe osteolysis however tine in-situ was 9-14years	

With regard to the addition of 3rd body particles, simulators have shown improved, equal and decreased wear resistance (Essner et al, 2000, Laurent et al, 2000, Murtaloglu et al, 2000, Polineni et al, 1999 and Taylor et al, 1999). Pin on plate studies have shown increased roughness and damage with the addition of 3rd body particles (Caravia et al, 1990, McNie et al, 1999, Minakawa et al, 1998 and Que and Topoleski, 1997). Simulator studies consistently showed higher wear-rates with alumina particles added to the JFA (Laurent et al, 2000 and Murtaloglu et al, 2000). However, with PMMA particles the wear-rates varied. Essner et al (2000) added 1, 5 and 10mg/ml of PMMA particles to the JFA in wear studies of UHMWPE articulating on ceramics and metal heads. The heads of ceramic showed a consistent decrease in UHMWPE wear-rates, regardless of the amount of PMMA. However, the metal head showed varied results going from decreasing, equal and increasing wear-rates of UHMWPE for the three increasing amounts of PMMA respectively. The effect of PMMA was also varied for different treatments of UHMWPE materials. Studies of highly crosslinked UHMWPE showed little or no difference to wear-rates with the addition of PMMA particles (Laurent et al, 2000, Murtaloglu et al, 2000 and Taylor et al 1999). There were no consistent findings between methods applied to these studies. Orbital and 3-axis simulators were used, the JFA was varied and the THR configuration was anatomical (Anat) with one exception. Thus, the addition of 3rd body particles was equivocal in the simulator and caused damage to the counterface in pin on plate studies.

Studies have also examined wear-rates after artificially roughening the head to simulate 3rd body damage. The simulator and pin on plate studies showed various degrees of decreased wear resistance with changes in the surface roughness.

Simulator studies averaged a 2 times decrease in wear resistance with a 10 times increase in R_a (Essner et al, 1998, McKellop et al, 2000, Polineni et al, 1999 and Wang et al, 1998). McKellop et al (1995) tested retrieved femoral heads with R_a 3 to 10times greater than the initial R_a at implantation. The wear-rates of the UHMWPE articulating on the retrieved heads was only 1.3 times greater than the control and the R_a values decreased by 40 to 80% after testing. Additionally, the UHMWPE wear-rates for both sets were lower than clinical averages. On the other hand, pin on plate studies with roughened heads showed very large increases in wear with small increases in R_a (Besong et al, 1997, Cooper et al, 1993, Fisher et al, 1994 and Wang et al, 1998). Thus, the in-vitro studies with roughened heads showed increased wear with increased R_a, but varied by type of machine.

The average increase in roughness (R_a) found clinically was 0.20μm, a 4 times increase from initial implant quality (Elfick et al, 1999, Polineni et al, 1999 and Schmalzried et al, 1997). Clinically there has been little evidence to support the idea of large differences in wear-rates with rougher heads versus smooth heads (table 2.2). A 4 times increase in R_a has shown an average increase in clinical wear-rates of 10%. The study by Hailey et al (1997) showed that with a 24 times increase in R_a there was only a 2 times increase in wear-rate. Consequently, it was shown clinically that 3rd body damage in the form of increased surface roughness did not show an appreciable change in wear. It was also shown that the in-vitro wear-rates with 3rd body particles and roughened heads generally continued to underestimate clinical wear-rates (table 2.2). Therefore, the reason for the disparity between in-vitro and in-vivo wear-rates was not 3rd body wear.

It remains to be determined why UHMWPE wear-rates have been underestimated in-vitro. Duplication of wear-rates and other clinical criteria were the focus of this thesis. This clinically relevant material remains an excellent target for validation of wear testing.

2.2 CLINICAL BALL SIZE EFFECT

Charnley's first PTFE acetabular cups were mated with a 41.5mm diameter head because this was similar in size to the human femoral head (Charnley et al, 1969). He reasoned that the frictional torque created by the large inner diameter of the cups was causing loosening of the prosthesis. Therefore to reduce the frictional torque, he used a smaller inner diameter while maintaining the same outer diameter of the cup. Thus, Charnley's Low Friction Arthroplasty (LFA) design was created (Waugh, 1990). The LFA reduced the inner diameter of the acetabular cup from 41.5 to 28mm and to eventually 22.25mm. This reduction in inner diameter also created less wear, although for PTFE this was a moot point. He first reported this decreased wear in 1969 after he had revised most of the PTFE hip replacements. Analysis of his retrieved data showed that as the size of head increased the volumetric wear-rate increased by $55mm^3/y$ or 6% per millimeter of head diameter (Clarke et al, 1997; fig 2.1). Thus, the ball size effect of increased wear-rate with increased head size was shown for PTFE.

This ball size effect was also demonstrated for UHMWPE (Hall et al, 1988, Kabo et al, 1993 and Livermore et al, 1990). Unlike PTFE, the UHMWPE THR has undergone many variations in design, conditioning and fixation along with various mating materials. Clinical reports of wear-rates vary from investigator to investigator depending on the aforementioned variations (table 2.1). However, it was

evident from the UHMWPE clinical results that the 22mm cups had lower wear-rates than the larger 28 and 32mm cup (table 2.3). The rate of increase for UHMWPE wear-rates was 7% per millimeter of head diameter (Clarke et al, 1993; fig 2.2). This was in good agreement with the PTFE data. Therefore, two polymers showed that wear-rate increased with respect to head diameter and this rate of increase was 6 to 7% per millimeter of head diameter.

This relationship provided an additional validation tool for the laboratory. Thus, the ball size effect coupled with the UHMWPE wear-rates and debris characteristics demonstrated increased reliability of in-vitro studies.

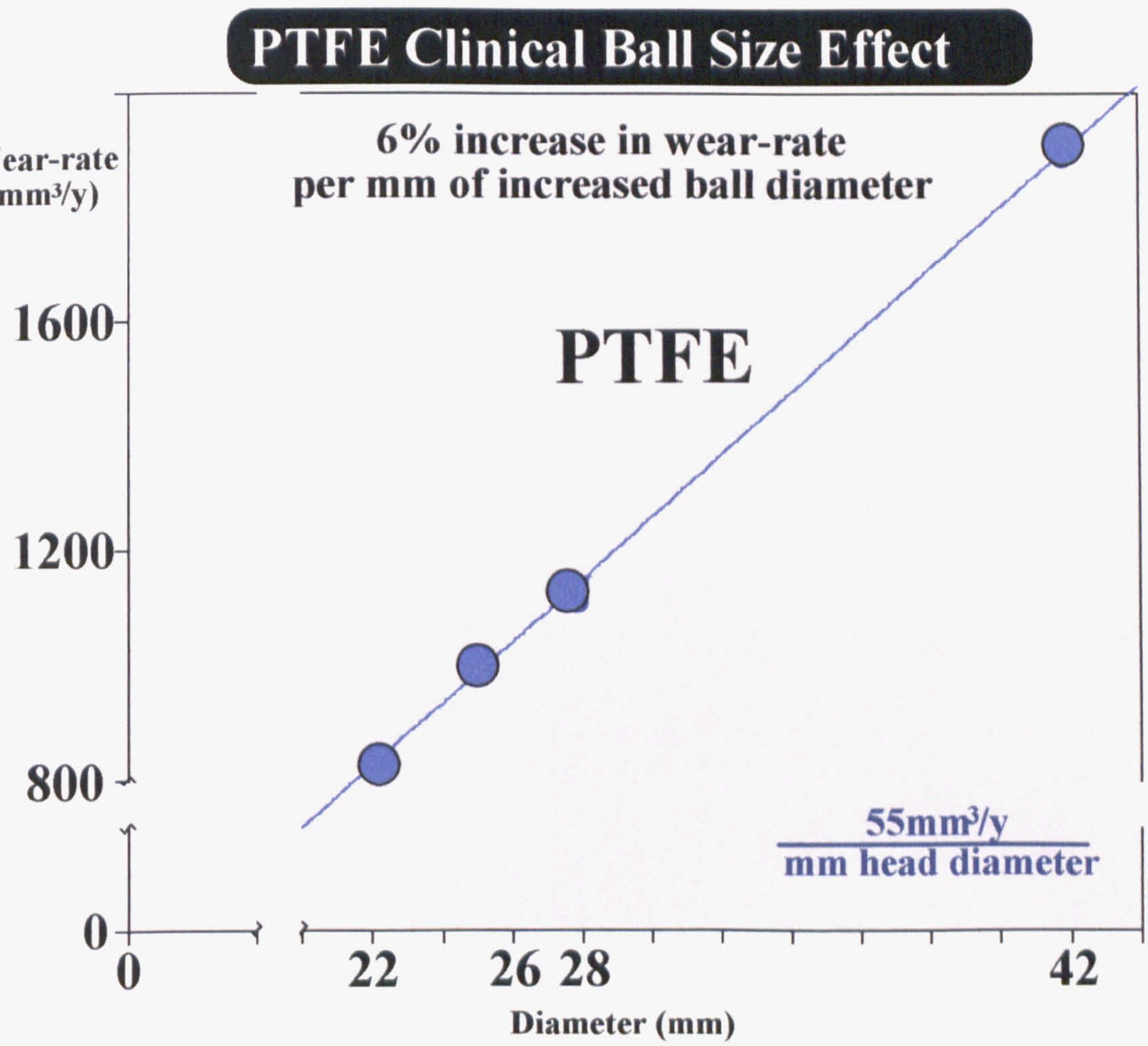


Figure 2.1 Clinical ball size effect for PTFE acetabular cup showing increased wear-rate with increased ball diameter

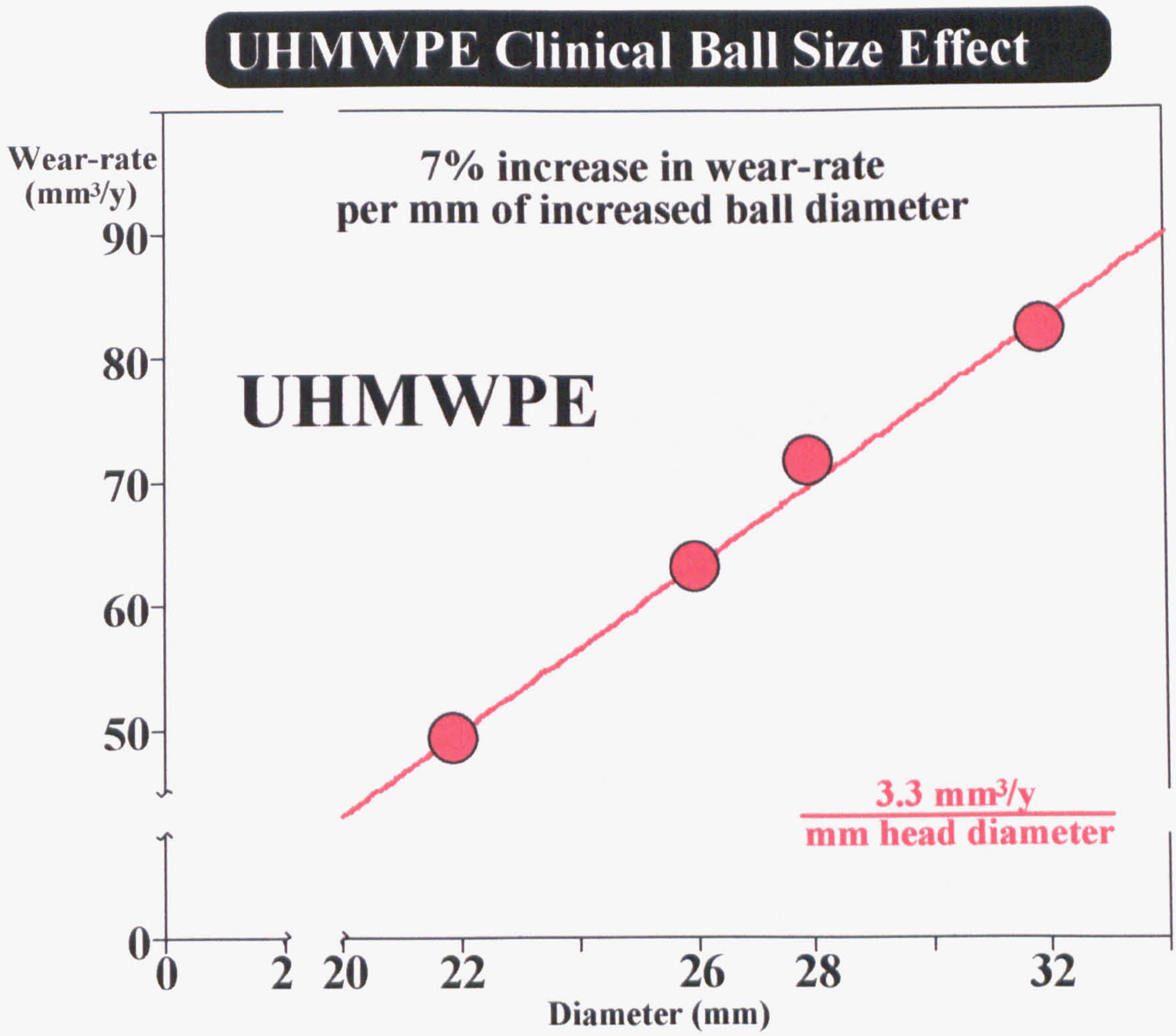


Figure 2.2 Clinical ball size effect for UHMWPE acetabular cup showing increased wear-rate with increased diameter

Table 2.3 Clinical UHMWPE Acetabular Cup Wear-rates (mm^3/y)
Femoral Head Materials of: Stainless Steel, Ceramic, CoCr, Ti or Not Specified

Author	*22mm*	*26mm*	*28mm*	*32mm*
Charnley, 1972	50			
Charnley & Cupic, 1973	47			
Charnley & Halley, 1975	70 39			
Scheier & Sandel, 1976			61	
Griffith et al, 1978	27			
Atkinson et al, 1985	74			
Isaac et al, 1989	49 63			
Livermore et al, 1990	47.5		48	84
Weightman et al, 1991	93			
Oonishi et al, 1992			154 60	
Cates et al, 1993			66 48	
Kabo et al, 1993	26	63	76	89
Alsema et al, 1994		<65 (25mm)		
Devane et al, 1995		67	99	
Hamilton & Gorczyca, 1995	50			
Sugano et al, 1995			62	
Callaghan et al, 1995	41 35		71 66	
Sychterz et al, 1996				53 37
Kesteris et al, 1996	57			148
Hall et al, 1996	51			
Wroblewski et al, 1996	22			
Average (unweighted)	49	65	74	82

2.3 IN-VIVO AND IN-VITRO FAILURES

2.3.1 Polytetrafluoroethylene (PTFE)

From 1959 to 1962 Professor Sir John Charnley implanted over 300 PTFE hips. He selected PTFE because of its low coefficient of friction and first used PTFE to resurface femoral heads and acetabular sockets. He then went on to use it as an acetabular replacement (Waugh, 1990). His success with PTFE was short lived; he began seeing evidence of high wear within the first 2 to 3 years and ultimately revised all cases. The PTFE hip fills a unique position in clinical history; a single surgeon implanted, revised and documented all recorded cases (Waugh, 1990). As a consummate scientist Charnley saw the value of documenting the wear from these retrievals, therefore, producing a database for future research.

Professor Sir John Charnley found that the femoral head tunneled into the PTFE acetabular cup, making "a cylindrical pathway of the same diameter as the steel head" (fig 2.3; Charnley et al, 1969). Using this observation he converted the linear wear-rate into a volumetric wear-rate. He found that the wear-rates for PTFE were 835, 1006, 1134 and 1905mm^3/y for the 22.25, 25.25, 28.5 and 41.5mm head sizes respectively. From this he discovered that the wear-rate volume increased linearly as the head diameter increased. However, his wear-rates were based on only 58% of the 100 hips evaluated in his study (Charnley et al, 1969). This set of 58 hips excluded extremely high wearing and worn through cups (fig 2.3). Thus, his study only reported on moderately worn cups artificially skewing the data on the low side of wear-rates. From Charnley's graph it could be seen that wear-rates were as high as 6mm/y for the highly worn 22mm diameter cups (Charnley et al, 1969). This was a 2.7 times increase from the reported average of 2.26mm/y. Thus, wear-rates of

PTFE cups were underestimated using only the moderate wear cups and volumetric wear-rates for the 22mm were actually as high as 2331mm^3/y. Consequently, the wear-rates reported by Charnley were adjusted up by a factor of 2 to reflect the moderate, highly worn and worn through PTFE cups. The adjusted wear-rates were 1670, 2012, 2268 and 3810mm^3/y for the 22.25, 25.25, 28.5 and 41.5mm head sizes respectively (Oparaugo et al, 1998 and Wang et al, 1999).

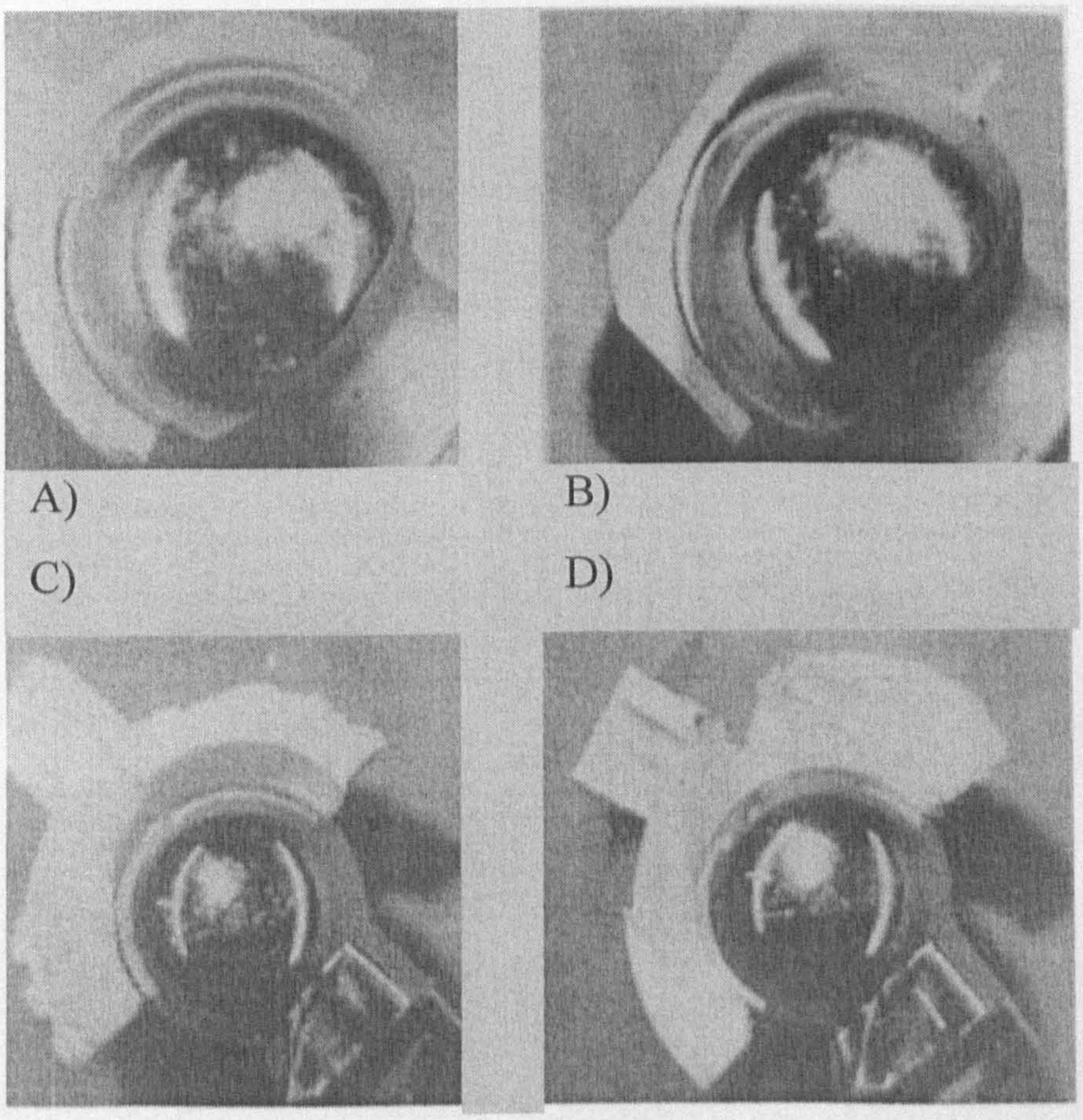


Figure 2.3 Clinical retrievals of Charnley PTFE THR A) 41.5 mm cup B) 41.5 mm cup worn through C) 28.5 mm cup showing wear tunneling D) 28.5 mm cup worn through

PTFE, like UHMWPE, provided an excellent clinical validation tool for the laboratory. The addition of PTFE combined with UHMWPE created another clinical validation tool of wear-rate ratio. Charnley was the first to use this tool in-vitro, finding a 250:1 PTFE/PE ratio. He voiced concern that this in-vitro ratio was too high because clinically this ratio was only around 17:1 (Charnley, 1976). With the adjusted wear-rates this clinical ratio would double but Charnley's laboratory ratio was still too high. Subsequent in-vitro studies of PTFE showed even higher PTFE/PE ratios (Clarke et al, 1980; Clarke et al, 1995; Dowson et al, 1985; McKellop et al, 1977; McKellop et al, 1978; McKellop et al, 1981 and McKellop et al, 1983; table 2.4). One study duplicated the clinical PTFE/PE ratio but wear-rates were extremely high, with a PTFE wear-rate of 33mm/y (Homsey et al, 1969). This study showed the importance of matching more than one clinical criterion. Wear-rates of in-vitro studies consistently overestimated clinical wear-rates of PTFE using bovine serum as the JFA and underestimated the clinical wear-rates using water as the JFA (table 2.4). This trend was true for both LST and multi-directional surface tracking (M-dST; multi-directional refers to a crossing path on the wear surface) testing devices. Interestingly, one M-dST study in water did produce clinically relevant wear-rates but produced uncharacteristic debris (Clarke et al, 1995), and they reported that the debris was very different from their test in serum. In water the debris floated to the top in large flakes. Charnley had reported that the size of the debris in-vivo ranged from 5 to 50μm and that there were never large sheets of debris visible to the naked eye (Charnley et al, 1969) thus, emphasizing again that duplicating only one criterion was not enough. Therefore, in-vitro studies have not duplicated the clinical PTFE results, but because of Charnley's excellent PTFE wear

Table 2.4 PTFE In-vitro Studies

M-dST = multi-directional surface tracking, LST = linear surface tracking, SS = Stainless Steel, Mc = million cycle, NA = Not Applicable

Author	*Year*	*Machine Type*	*Head Material*	*Fluid*	*PTFE Wear-rate*	*In-vitro PTFE/PE*	*Notes*
Homsy et al	1969	LST Ring on Ring	SS	Pseudo synovial fluid	33mm/y	29	
Gold et al	1974	LST Simulator	SS CoCr	Water	223.1* 28.5*	NA	*No units given
Charnley et al	1976	M-dST Pin on Plate	SS	Water Bovine SF Plasma 0.5% gelatin	25.7 193 200 260	174 1200 1500 1400	
McKellop et al	1977	LST Pin on Plate	SS	Bovine serum	2.4mm/Mc	2448	
McKellop et al	1978	LST Pin on Plate	SS	Bovine serum	5.0mm/Mc	1666	
Clarke et al	1980	LST Pin on Plate	SS	Bovine serum	5.0mm/Mc	1662	
McKellop et al	1981	LST Pin on Plate	SS	Bovine serum	$322mm^3/Mc$	5366	
McKellop et al	1983	M-dST Simulator	28mm SS	Bovine serum	$3000mm^3/Mc$	71	
Dowson et al	1985	LST Pin on Disk	SS	Water Bovine SF	$1.4x\ 10^{-5}\ mm^3/Nm$ $5.7x\ 10^{-5}\ mm^3/Nm$	NA	In-vivo PTFE wear = $3.7x\ 10^{-5}\ mm^3/Nm$
Clarke et al	1995	M-dST Simulator	22mm CoCr	Water Bovine serum	$1104mm^3Mc$ $2572mm^3/Mc$	NA	

documentation the laboratory had another model with which to validate in-vitro testing devices and parameters.

2.3.2 Polyester

Polyethylene terephthalate (PET or Polyester) was first used as the femoral head of a trunion bearing system with an acetabular cup made of CoCr (Weber and Stuhmer, 1976). The trunion-bearing concept was designed by B. G. Weber in 1961 (Weber and Semlitsch, 1972). The trunion bearing was a sleeve of plastic that fitted over the femoral neck and between the head. The concept utilized the three movements in the human hip to produce two movements in the prostheses; a) flexion/extension causing rotation of the head on the trunion and b) adduction/abduction and internal/external rotation; rotating the cup and head. The design was to produce wear only at the plastic component, which if heavily worn, could be replaced easily, leaving the cemented components intact. This THR was put into clinical practice utilizing a proprietary polyester, AP3 and AP4, for the femoral head (Weber and Semlitsch, 1972). Weber chose polyester because it was hard, tough and resistant to wear and cold flow, under pressure (Weber and Semlitsch, 1972). He also reported that polyester showed good tissue tolerance in animals and tissue cultures and that when tested in the simulator the prostheses "behaved most satisfactorily" (Weber and Stuhmer, 1976). Polyester was also used as an acetabular cup mated with a CoCr femoral head (Scheier and Sandel, 1976 and Willert et al, 1981)

Four years after the original implantation of the polyester trunion bearings, Weber reported favorable clinical results with minimal wear on the polyester and metal components that had been removed for late loosening (Weber and Semlitsch,

1972). What was not known then was that the late loosening was caused by large quantities of debris, which eventually led to failure. By 1976 it was shown that in as little as 3 years the polyester was wearing enough to cause a dramatic foreign body reaction resulting in loosening of the prostheses and subsequent revision (Weber and Stuhmer, 1976; table 2.5).

The in-vitro polyester pin on disk studies conducted before clinical applications agreed with Weber's initial observations that polyester wore less than UHMWPE (table 2.6). Not until reports of clinical failure (table 2.5) did laboratory studies begin to report that polyester showed greater wear than PE. According to Dumbleton et al, (1974) and Clarke and McKellop, (1982) there was an earlier pre-clinical report, which was an internal study conducted at Sulzer (manufacturer of the polyester bearing). This study using a "simulator" which in fact was a LST device, showed that the wear-rate of untreated AP4 was much higher than PE when used as either the head or the cup. However the tests of radiation treated AP4 (AP4*) showed wear-rates that were half that of UHMWPE in the same simulator. This test should have sent alarm bells ringing because their wear-rate of UHMWPE (178 mm^3/million cycles) was at least twice that of clinical wear-rates (Charnley, 1972; table 2.3) and historically UHMWPE in-vitro studies had produced lower than clinical wear-rates (Duff-Barclay and Spillman, 1967; Seedhom et al, 1973; Dumbleton et al, 1974 and Walker and Erkman, 1974). Therefore, laboratory studies failed to predict the true wear-rate of polyester and when studies showed irregular results they were ignored.

Table 2.5 Polyester In-vivo Studies

NR = Not Reported

Author	*Year*	*Polyester Wear-rate (mm/y)*	*Notes*
Weber	1972	NR	Only "pin hole" of the head showed wear
Weber & Stuhmer	1976	NR	Approx. 30% had deteriorated within 3-4 years
Scheier & Sandel	1976	0.2	Wear slightly > PE
Willert et al	1981	NR	Wear > PE with instances of socket fracture
Semlitsch	1974	NR	Unsuitable due to material and clinical trials
Dumbleton	1983	NR	Failure due to tissue reaction

Table 2.6 Polyester In-vitro Studies

(* treated with radiation) Mc = million cycles, LST = linear surface tracking, NR = Not Reported

Author	*Year*	*Machine Type*	*Head Material*	*Fluid*	*Polyester Wear-rate (mm^3/Mc)*	*Notes*
Scales & Lowe	1971	LST Pin on disk	CoCr	Bovine serum	5 to 20 times < PE	
Walker & Erkman	1974	LST Pin on disk	SS	Water	36 22	AP4 Polyterephthalate
Dumbleton (Reports on internal Sulzer study of 1972)	1981	LST Simulator	NR	saline	2000 178 267 96	Head of AP4 Head of AP4* Cup of AP4 Cup of AP4*
Weber & Semlitsch	1972	LST Simulator	NR	NR	0.09mm/Mc 0.004mm/Mc	AP4 AP4*
Walker & Salvati	1973	LST Ball on flat	CoCr	Water	Very low Very high	AP4 Polyterephthalate
Dumbleton et al	1974	LST Annulus on flat	SS	plasma	5×10^{-11} mm^3/Nm	
Galante & Rostoker	1976	LST Disk on flat	CoCr	water	> UHMWPE/CoCr	Polyester filled and unfilled and AP5
McKellop et al	1977	LST Pin on flat	SS	Bovine serum	1.35mm/Mc	
McKellop et al	1978	LST Pin on flat	SS	Bovine serum	2.6mm/Mc	
McKellop et al	1981	LST Pin on flat	SS SS	Bovine serum	160 11	Load of 445N Load of 223N

2.3.3 Polyacetal (Delrin™)

In 1970, Christiansen introduced a hip system which incorporated a Delrin™ (polyacetal) acetabular cup with a 37mm CoCr femoral head fitted over a Delrin™ trunion bearing. Approximately 8,000 THR's of Delrin™ were implanted in the 1970's and 1980's. By 1986 Delrin™ was clearly a clinical disaster (Ohlin & Persson, 1989). Wear-rates ranged from 15 to 967mm^3/y with over 5,800 revisions reported from 1979 to 1990 (table 2.7: Havelin et al, 1986; Ohlin and Persson, 1989; Mathiesen et al, 1986; Malchau et al, 1993) and only a 50% survivorship predicted at 10 years (Ohlin, 1990).

Table 2.7 Delrin™ In-vivo Studies

Author	*Year*	*Implant Design*	*No. of retrievals examined*	*Range of Wear-rates (mm^3/y)*
Havelin et al	1986	Christiansen	39 explants	36-600 (avg. 136)
Mathiesen et al	1986	Christiansen	12 explants	15-669 (avg. 240)
Ohlin & Persson	1989	Christiansen	22 explants	140-967 (avg. 418)

There were two laboratory wear studies of Delrin™ before it was used clinically, a simulator study from Duff-Barclay and Spillman, (1967) and a ring on ring study by Homsy and King, (1969). Both studies reported that Delrin™ was more wear resistant than UHMWPE. The ring on ring and simulator studies showed a 1.6 and 5000 times improvement over UHMWPE respectively (Table 2.8). However, three years later after Delrin™ was being used clinically, Homsy and King, (1972) again reported on Delrin™ but this time showed that it had 3.5 times more wear than UHMWPE. Four more laboratory studies from UCLA followed that study. Their studies using LST devices also reported that Delrin™ had more wear

than UHMWPE, up to 200 times (McKellop and Clarke, 1988; McKellop et al, 1977; McKellop et al, 1978; Clarke and McKellop, 1980; Table 2.8). Therefore, it would seem that two laboratories had finally predicted the high wear of Delrin™ but too late to avert the clinical disaster.

In a surprising *reversal* Edidin and Kurtz, (1999) reported that Delrin™ wore 50% less than PE in their new simulator. They stated, "In conclusion, the wear simulation data using a variety of previously clinically applied materials coupled with a determination of the large scale mechanical response curves suggest that the modern hip simulator properly ranks materials and process changes in a manner than can be transferred to the clinical environment." To support their argument that Delrin™ was more wear resistant than UHMWPE they stated that a) only the worse clinical cases of Delrin™ wear were reported and b) the 37mm head size of Delrin™ would produce wear-rates comparable to that of a 37mm UHMWPE cup. As shown by table 2.7, the range of clinical wear-rates included low wear-rates therefore, the first argument was invalid and using the ball size effect of 6% increase in wear-rate per mm of head diameter (Clarke et al, 1997) refutes the second argument. The wear-rate of a 37mm UHMWPE cup was 95 mm^3/y, which was 30 to 80% less than the wear-rate of Delrin™. Therefore, this data did not help validate their simulator.

It was interesting to note the results using two different testing devices. The six studies using LST devices were from two laboratories. Homsey and King's two studies were contradictory and the four studies from UCLA were consistent and clinically realistic. However, as will be demonstrated in the following sections, only simulators reproduce the multi-directional wear paths of the THR components and without these wear results were not reliable. The two simulator studies were from

different laboratories with both reporting similar yet clinically incorrect results. These studies used different parameters, i.e. number of motion axis, JFA, speed, and cup size. Therefore, there was no commonality with which to explain the incorrect rankings and the only conclusion may be that Delrin™ was an enigma in the laboratory.

Table 2.8 Delrin™ In-vitro Studies

M-dST = multi-directional surface tracking, LST = linear surface tracking

Author	*Year*	*Head Material*	*Machine Type*	*Fluid*	*Wear ratio Delrin™/PE*
Duff-Barclay & Spillman	1967	CoCr	M-dST Simulator	Plasma dry	~ 0.0002 ~ 0.025
Homsy & King	1969	SS	LST Ring on Ring	Pseudo Synovial Fluid	0.62
Homsy & King	1972	CoCr	LST Ring on Ring	Pseudo Synovial Fluid	3.5
McKellop et al	1977	SS	LST Pin on flat	Bovine serum	56
McKellop et al	1978	SS	LST Pin on flat	Bovine serum	200
Clarke & McKellop	1980	SS	LST Pin on flat	Bovine serum	60
McKellop & Clarke	1988	SS	LST Pin on flat	Bovine serum	28
Edidin & Kurtz	1999	CoCr	M-dST Simulator	70% Bovine serum	0.5

2.3.4 Porous-Coated Anatomic Total Hip Replacement (PCA)

The PCA hip was introduced in 1983 as an alternative to cementing. The head was pre-assembled and made of CoCr. The acetabular cup was UHMWPE, which fit into a metal backing that had sintered beads for bony in-growth. The backing also had 2 pegs for fixation with a hole in the central dome. The UHMWPE acetabular cup had a matching peg to fit into the central dome hole, a notch to

receive an anti-rotation peg from the metal backing and a rim along the top edge of the cup which extended beyond the metal backing (Aston et al, 1996). The reason for failure of the PCA hip was severe osteolysis causing fracture and migration (table 2.9). Three of the possible contributors to this severe osteolysis were design, size and patient age.

Table 2.9 PCA In-vivo Studies

NR = Not Reported

Author	*Year*	*PCA Wear-rate (mm^3/y)*	*Notes*
Cooper et al	1992	339	5% revision at 6 years all patients aged < 50years
Owen et al	1994	NR	50% osteolysis > 5years 44% survivorship at 9years
Devane et al	1995	67 99	26mm 32mm
Learmonth & Hussell	1995	NR	25.7% osteolytic lesions in 32mm group none in 26mm group
Xenos et al	1995	NR	15% osteolysis at 7 years
Aston et al	1996	NR	31% osteolysis at 7years 12% failure at 7years
Thanner et al	1997	NR	9% revised at 7years
Tollund et al	1998	NR	4% revised at 6.6years & 18% excessive PE wear

Design issues were the anti-rotation notch, the central peg and the extended rim. In a study examining 13 revised PCA hips, Aston et al, (1996) reported that the cups could freely rotate within the metal backing. This rotation was caused by gross distortion of the anti-rotation notch. The distortion of the anti-rotation notch was caused by wear and as the cup became loose, it would cause the cup to wear against the backing, resulting in more UHMWPE wear. Aston et al, (1996) also reported on distortion of the central peg, producing yet another wear zone. In a study by Xenos et al, (1995) the wear of the central peg was severe enough to cause the peg to fracture. Another design flaw was in the extension of the rim beyond the metal backing. This caused stress along the line of contact between the UHMWPE rim and

the metal backing. In their study of revised PCA's, Aston et al, (1996) reported cracks and evidence of fatigue and wear in the UHMWPE rims.

The 32mm diameter size of the PCA was another reason for failure. It has been shown clinically that larger diameter THR's produce greater wear-rates than smaller THR (Charnley et al, 1969 and Livermore et al, 1990; fig. 2.2 and 2.3). Learmonth and Hussell, (1995) compared 26mm to 32mm PCA's and found that the 26mm group had no osteolysis where 1/4 of the 32mm group did exhibit osteolysis.

The third possible explanation for failure of the PCA was the patient age. The PCA hip was seen as an alternative for younger patients because it eliminated the use of cement, which was thought to be a major contributor to wear. The PCA hip was subsequently implanted in many patients under the age of 50 (Cooper et al, 1992; Owen et al, 1994; Xenos et al, 1995 and Aston et al, 1996). Unfortunately, the problems of design and head size coupled, with the greater activity of the patients, exacerbated the wear, increasing the debris induced osteolytic response and subsequent failure of the PCA hip.

There were no published pre-clinical laboratory tests of the PCA hip. However, the design issues would have only been seen if the laboratory tested the cup with the metal backing. In the present laboratory this is not done routinely, therefore this problem might not have been discovered. Regardless of the testing protocols, there was no published data to recommend or warn against this prosthesis. This finding shows that either the manufacturer did not test the hip or did test but opted not to publish. Either way these options were not encouraging, proving that new hips should be adequately studied and these studies should be made available before clinical implantation.

2.3.5 Hylamer™

Hylamer™ (DePuy DuPont Orthopaedics) was made from ram extruded GUR415 UHMWPE which undergoes a proprietary heat and pressure treatment. "This process converts the crystalline regions from 'short folded chains' to a long 'extended chain' configuration resulting in a polymer with a higher degree of crystallinity and significantly larger crystalline regions. The enhanced polymer has improved strength, creep resistance, crack growth resistance and oxidation resistance as compared to conventional medical UHMWPE." (DePuy, 1994). This increased crystallinity was verified by McKellop et al, (1992) reporting that the crystallinity for enhanced Hylamer™ was 1.75 times greater than conventional GUR 415 UHMWPE.

The clinical findings for Hylamer™ indicated that despite the "enhancement" wear-rates were much greater than for conventional UHMWPE. In fact, five of the eight clinical studies compared Hylamer™ to other UHMWPE hip systems and all but one showed Hylamer™ to have higher wear-rates ranging from 6 to 108% more than the comparison UHMWPE (table 2.10). However, Thomas et al, (1996) reasoned that the increased wear-rates for Hylamer™ were due to the younger age of the patients receiving Hylamer™ hips in comparison to the older patient age receiving conventional UHMWPE (54.2 years versus 70.5 years). Yet, two limited studies by Chmell et al, (1996) and Scott et al, (2000) showed older patients had a 1.4 and 1.1 times increase in wear-rate respectively in comparison to younger patients, thus showing that the age of the patient did not necessarily indicate activity level or wear-rate level. In addition, Chmell et al, (1996) predicted only an 86% survivorship at 4 years, which is in stark contrast to the 90% survivorship of

Table 2.10 Hylamer™ In-vivo Studies

NR = Not Reported, NA = Not Applicable

Author	*Year*	*Average Time (Follow-up or Revision)*	*# Hylamer™ Hips/ Patient age*	*# Comparison Hips/ Patient age*	*Comparison Hip material*	*Hylamer™ Wear-rate (mm^3/y)*	*Hylamer™ /UHMWPE Ratio*	*Notes*
Thomas et al	1996	Follow-up 3.45y	41 54.2ys	47 70.5y	UHMWPE	133.75	1.1	
Chmell et al	1996	Revision 3y	3 - 30.6y 3 - 62.3y	NA	NA	250 338	NA	Broken down into age groups
Muratoglu et al	1997	NR	8	NA	NA	139.6	NA	
Sychterz et al	1997	Follow-up 3y	84 54.2	138 64.3	Enduron™	73.8	0.48	
Livingston et al	1997	Follow-up 3.3y	45	54 50	Osteonics	153	1.6 2.1	
Schmalzried et al	1998	Follow-up 3y	347<60y 333>60y	50<60y 349>60y	DePuy GUR415 DePuy heads	169 141	1.3 1.1	Hylamer™ with various heads
Akisue et al	1999	Revision 33mos	22	12	Enduron™	242	1.7	
Scott et al	2000	Revision 50mos	4<60y 8>60y	NA	NA	165 182	NA	

Charnley hips at 10 years (Sochart, 1999). Thus, these studies showed quite clearly that Hylamer™ was not an improvement over conventional UHMWPE.

Hylamer™ was yet another acetabular cup, which had no published data prior to 1990 and implantation. DePuy published a brochure in 1994 (table 2.11) reporting on 2 in-vitro studies, one pin on disk (assumed to be conducted before clinical release) and 1 simulator study (published in 1992). The pin on disk study was performed by DePuy and showed that Hylamer™ was 32% more wear resistant than the same non-enhanced, conventional UHMWPE (DePuy, 1994). The simulator study was performed at an independent laboratory but contrary to the pin on disk study the wear resistance of Hylamer™ was the same as that of the non-enhanced GUR 415 UHMWPE (McKellop et al, 1992). Similar to Delrin™ and polyester, in-vitro studies of Hylamer™ showed poor wear resistance only after clinical data was available (Essner et al, 1998; Huber et al, 1996; McKellop et al, 1997 and Sanford et al, 1997). So, again, laboratory studies were only able to "predict" clinical failure after the fact.

Table 2.11 Hylamer™ In-vitro Studies

M-dST = multi-directional surface tracking, LST = linear surface tracking, γ = gamma irradiation, NR = Not Reported

Author	*Year*	*Machine Type*	*Head Materia*	*Fluid*	*Comparison Material*	*Hylamer™ Wear-rate (mm^3/Mc)*	*Hylamer™ Comparison*	*Notes*
McKellop et al	1992	M-dST Simulator	CoCr	Bovine serum	GUR 415	92.6 27.3	1.0 0.9	0-2.5Mc 2.5-5Mc
DePuy Brochure	1994	LST Pin on disk	CoCr	Bovine serum	GUR 415 γN_2	2.42mm^3/Nm	0.7	
Huber et al	1996	LST Annulus on flat	Al_2O_3	NR	Chirulen™ Enduron™	2.1mm^3	1.1 0.9	
Sanford et al	1997	M-dST Simulator	NR	Bovine serum	GUR 4150 γ N_2	0.035 0.03	2.2 0.3	non-aged aged
McKellop et al	1997	M-dST Simulator	NR	Bovine serum	2.7 γin air 2.7γN_2	32.9	1.1 1.9	
Essner et al	1998	M-dST Simulator	CoCr	Bovine serum	PCA	*220 ^133	2.3 1.4	*γ/ air ^ γ/vac

Table 2.12 LST Device Studies

SS = Stainless Steel, Al = Alumina, Zr = Zirconia, NA = Not Applicable

Author	*Year*	*Head Material*	*Machine Type*	*Fluid*	*Wear-rate*	*In-vivo/In-vitro wear-rates*
Homsey & King	1969	SS	Ring on ring	Pseudo Synovial Fluid	1.14mm/y	0.13
Homsey & King	1972	CoCr	Thrust washer	Pseudo Synovial Fluid	1.4mm/y	0.15
Seedhom et al	1973	SS	Tri pin on disk	Bovine Synovial Fluid	10^{-9} to 10^{-10} mm^3/Nm	3,000 to 4,000
Dumbleton et al	1974	SS	Annulus on flat	Plasma	10^{-10} mm^3/Nm	4,000
Charnley	1976	SS	Pin on flat/random motion	Varied	7×10^{-8} mm^3/Nm	2,000
McKellop et al	1977	SS CoCr	Pin on flat	Serum	1.6µm/Mc 0.98µm/Mc	93 to 153
McKellop et al	1978	SS CoCr	Pin on disc	Serum	3.1µm/Mc 2.6µm/Mc	48 to 57
Mustafaev et al	1978	SS	Pin on disk	Mineral Oil Dry	10^{-7} mm^3/Nm 5×10^{-17} mm^3/Nm	2,000 to 3,000
McKellop et al	1978	SS	Pin on flat	Serum	3 µm/Mc	150
Dowson	1978	SS	Pin on disc	Dry Water	10^{-8} mm^3/Nm 10^{-9} mm^3/Nm	2,000 to 3,000
Atkinson et al	1978	SS	Pin on disk	Dry	10^{-7} mm^3/Nm	1,000
McKellop et al	1979	SS/Al	Pin on plate	Serum	<1 µm/Mc	150
Clarke & McKellop	1980	SS	Pin on flat	Serum	1 to 3 µm/Mc	50 to 150
Dowson & Harding	1982	Al	Pin on disc	Dry Water	10^{-9} mm^3/Nm 10^{-8} mm^3/Nm	2,000 to 3,000
Dowson et al	1985	SS	Pin on disk	Water	10^{-7} mm^3/Nm	1,000
McKellop & Clarke	1988	SS CoCr Al	Pin on flat	Serum	0.1 mm^3/Mc 0.09 mm^3/Mc 0.06 mm^3/Mc	500 to 1,000
Streicher et al	1991	Al Zr	Pin on disc	30% Serum	2.1×10^{-7} mm^3/Nm 3.2×10^{-7} mm^3/Nm	1,000
Cooper et al	1993	SS	Pin on disc/plate	Water Serum	10^{-7} to 10^{-9} mm^3/Nm 10^{-8} mm^3/Nm	1,000 to 3,000
Saikko	1993	CoCr Al Zr	Pin on plate	Water	1.12 mm^3/Mc 0.04 mm^3/Mc 0.03 mm^3/Mc	50 to 1,000
Derbyshire et al	1994	SS Zr	Pin on disc	Water Serum	Zr ~ SS = 10^{-8} mm^3/Nm	2,000
Fisher et al	1994	SS	Pin on disc	Serum	9×10^{-9} mm^3/Nm	3,000

Table 2.12 LST Device Studies Continued

Author	*Year*	*Head Material*	*Machine Type*	*Fluid*	*Wear-rate*	*In-vivo/In-vitro wear-rates*
Fisher et al	1995	SS	Pin on disc	Serum	10^{-8} - 19^{-9} mm^3/Nm	2,000 to 3,000
Derbyshire et al	1995	SS	Pin on plate	Water Serum	10^{-9} mm^3/Nm 10^{-8} mm^3/Nm	2,000 to 3,000
Bragdon et al	1996	CoCr	flexion/extension	Serum	no measurable wear	NA
Barbour et al	1996	NR	Pin on plate	Serum	10^{-8} mm^3/Nm	2,000
Wang et al	1997	CoCr	Pin on flat	Alpha Calf Serum	γ N_2 = 0.75 mm^3/Mc ETO = 0.49mm^3/Mc	73 to 112
Alhroos & Saikko	1997	SS	Pin on plate	Synovial Fluid Serum	Negligible	NA
Saikko & Alhroos	1997	CoCr	Pin on plate	DPPC	Negligible	NA
Atkinson et al	1985	SS	Clinical	NA	2.9 x 10^{-6} mm^3/Nm	NA

2.4 IN-VITRO HIP JOINT WEAR MACHINES

2.4.1 History of Wear Testing Devices with Linear Surface Tracking (LST)

Testing devices that produce a linear surface tracking on the polymer (LST) have been extensively used over the past 3 decades (table 2.12). The simple geometry of these devices eliminated the need and cost of testing the actual THR combination and with multiple test stations tests could be conducted in a very short time, months versus years. Some of the more widely used machines utilized a simple pin, roller or annulus moving on a flat disc or plate. Motion was unidirectional as in the case of a pin moving in a circular path around a disc or reciprocating as in a pin moving back and forth on a plate (fig 2.4). Generally the pin was polymer representing the acetabular cup and the flat consisted of the harder counterface material but this was interchangeable. Parameters such as pressure, sliding velocity and fluid could be varied and some of the later machines incorporated multiple test stations. Several laboratories introduced machines, which could test an actual prosthesis (table 2.13). Although these machines were characterized as simulators, the path created on the THR surface was linear and these machines were categorized as LST devices.

Tests of UHMWPE on LST devices have typically exhibited wear-rates 1 to 3 orders of magnitude lower than clinical values (table 2.12). In addition LST devices reported disproportionate increases in wear-rate from small increases in surface roughness. Pin on disk LST machines showed that a 2, 2 and 3 times increase in the surface roughness resulted in a 2, 7 and 40 times increase in wear-rate magnitude respectively (Atkinson et al, 1978 and Cooper et al, 1993). Fisher et al, (1994) also

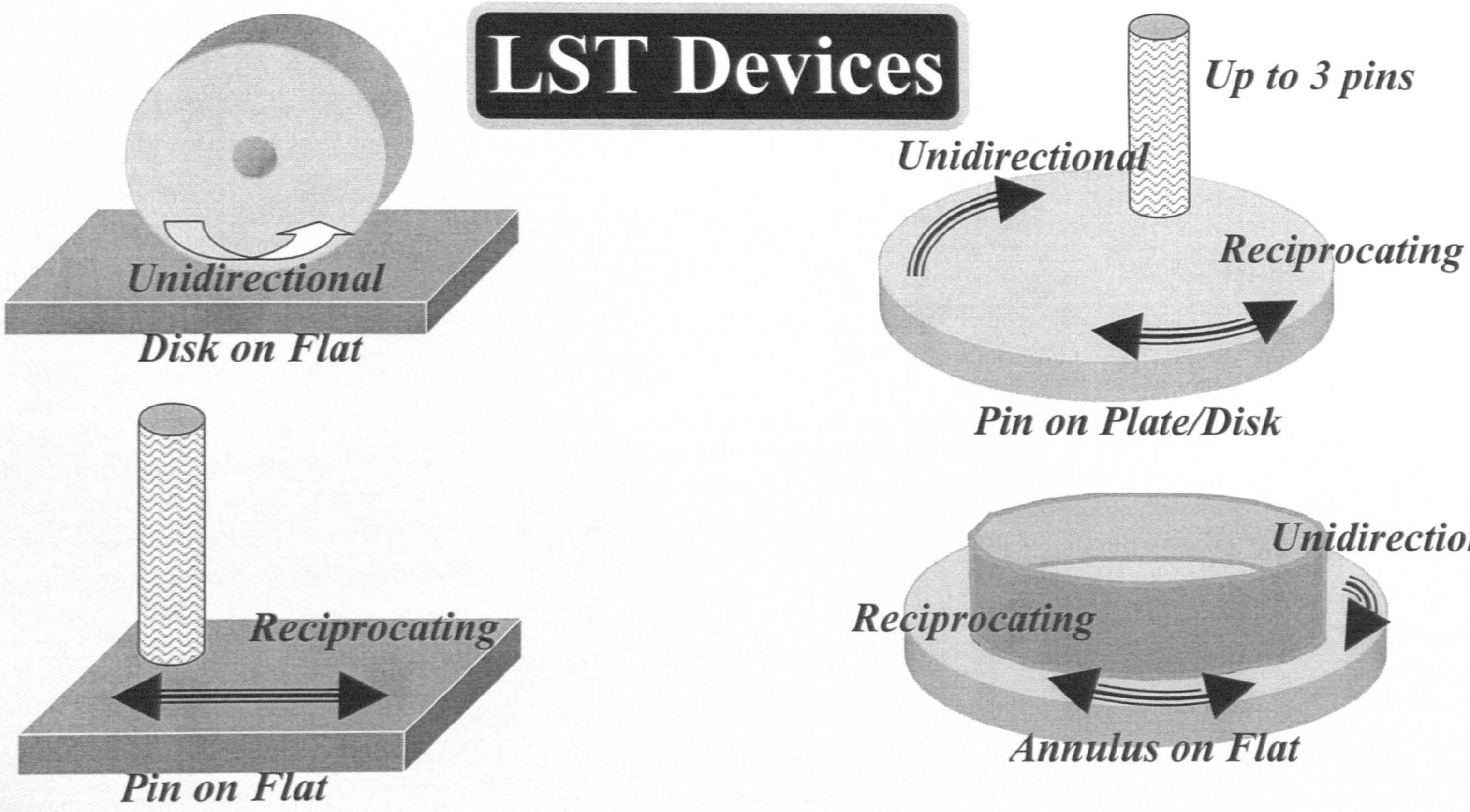


Figure 2.4 Various types of LST Devices

with pin on disk, reported a 20 times increase in wear-rate with only 1.1 times more surface roughness. However, three studies showed clinical wear-rates did not increase as dramatically with surface roughness (Cooper et al, 1993, Hailey et al, 1997 and Schmalzried et al, 1997). These clinical studies showed that as surface roughness increased by 2, 4 and 24 times, wear-rates only increased by 1.7, 1.5 and 2 times respectively (table 2.2).

Table 2.13 LST Simulators

Author	*Year*	*Description of Motion*
Weber & Semlitsch,	1972	Flexion/extension 30°
Weightman et al	1972	Flexion/extension
Gold & Walker	1974	Flexion/extension "swinging cradle"
Semlitsch et al	1977	Flexion/extension
Rose et al	1980	Flexion/extension "unidirectional and reciprocating"
Saikko et al	1992	Flexion/extension "uni-axial"
Pappas et al	1995	Flexion/extension movement in sagital plane

Further evidence of incorrect ranking using LST devices was shown with sterilization treatments of UHMWPE. LST studies of gamma sterilized or crosslinked UHMWPE exhibited less wear resistance than non-crosslinked UHMWPE. Fisher et al, (1995) reported that non-irradiated UHMWPE was 4 times more wear resistant than UHMWPE gamma sterilized in air using their pin on plate machine. Wang et al, (1997) also found this same discrepancy. Their test showed a 1.5 times increase in wear-rate for gamma sterilized UHMWPE in inert gas (crosslinked) when compared to EtO sterilized (non-crosslinked) UHMWPE. However, clinically this was not the case as seen with Hylamer™. Hylamer™ was EtO sterilized and showed at least a 1.3 times increase in wear-rate over conventionally gamma sterilized UHMWPE (table 2.10). Thus, LST devices did not

duplicate the clinical wear-rate magnitudes or clinical rankings of UHMWPE wear-rates with respect to surface roughness and crosslinking.

To better understand the discrepancies between results from LST devices and clinical values, Wang et al, (1997) and Bragdon et al, (1996) used computer modeling to map the surface of a femoral head as it traced a path on the acetabular cup during gait. Wang et al, (1997) reported that the motion of the hip joint during gait was primarily flexion/extension with a secondary cross shear pattern imposed by abduction/adduction. This resulted in a multi-directional crossing pattern rather than a linear path as produced by a LST device (fig. 2.5). This explained why the LST devices ranked non-crosslinked and crosslinked UHMWPE opposite to clinical findings. Both studies showed that linear tracking aligns the molecular chains of non-crosslinked UHMWPE in the direction of sliding. This alignment resulted in strengthening of the surface and more wear resistance in the direction of sliding, accounting for the lower wear-rate magnitudes of non-crosslinked UHMWPE. However, the multi-directional or cross patterns produced in-vivo would oppose alignment of the molecular chains. Thus, the non-crosslinked UHMWPE would experience alignment and then breaking of these molecular chains and more wear with multi-directional motion. On the other hand, crosslinking the UHMWPE with gamma irradiation increased the random patterns of the molecular chains. The randomization of the UHMWPE molecular chains would favor multi-direction motion but not linear alignment. Therefore, the crosslinked UHMWPE would have less wear with multi-directional motion.

Studies using LST devices have shown contradictory results with regards to the clinical findings of:

A) wear-rate magnitudes, from1 to 3 orders of magnitude difference

B) surface roughness, caused up to 6 times greater wear-rates than in-vivo

C) UHMWPE ranking, decreased wear resistance of crosslinked UHMWPE and increased wear resistance of non-crosslinked UHMWPE

D) motions produced on the surface of the THR

Thus wear studies using LST devices did not produce valid data and predictions of wear results from these machines could not be trusted.

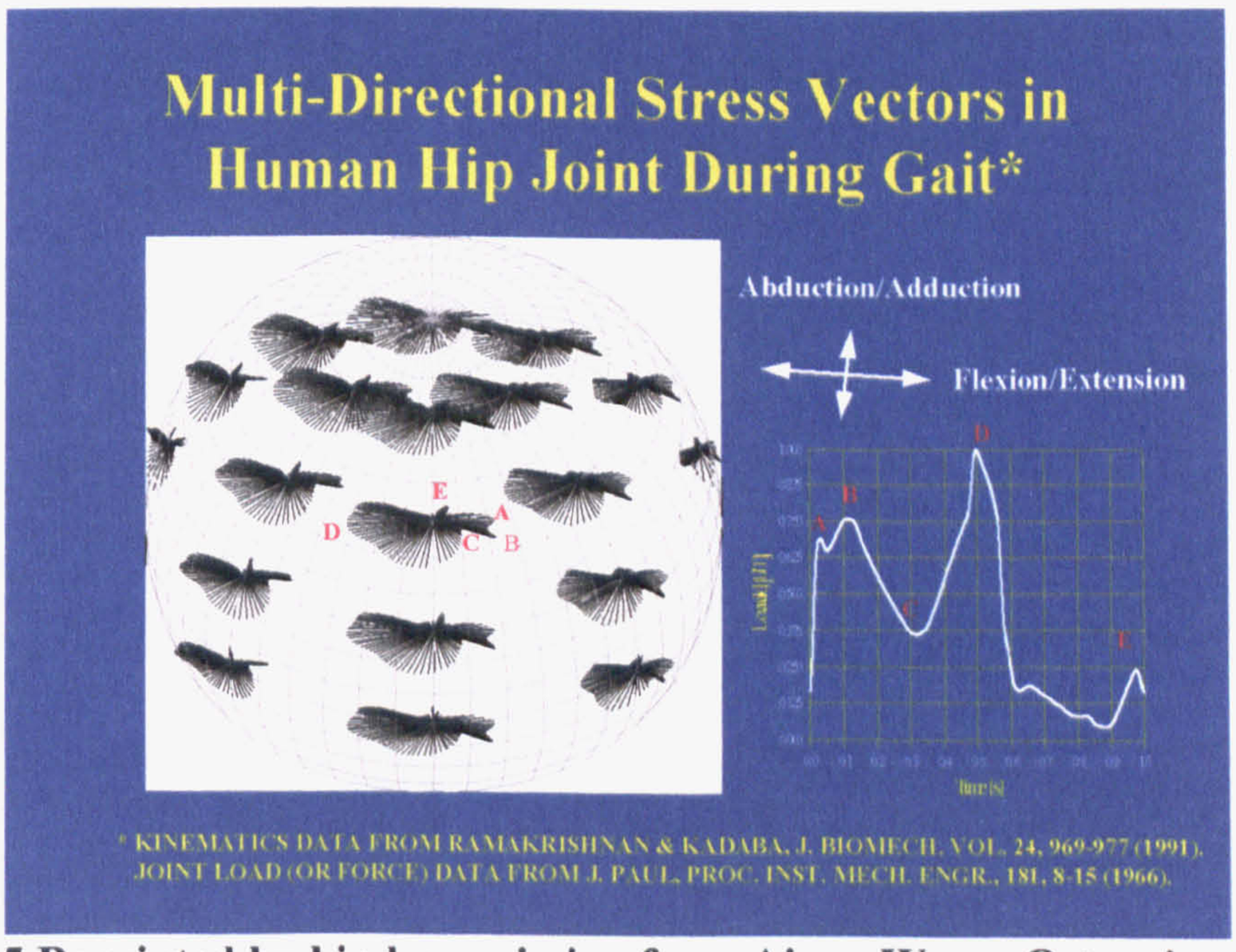


Figure 2.5 Reprinted by kind permission from Aiguo Wang, Osteonics-Howmedica (Wang et al, 1999)

2.4.2 History of Hip Joint Simulators with Multi-directional Surface Tracking (M-dST)

The definition of a simulator states, "an apparatus that generates test conditions approximating actual or operational conditions" (The American Heritage Dictionary of the English Language, 1975). To measure wear of an actual implant, it was not necessary to duplicate all conditions in the body. The real necessity was to

approximate those conditions such as force and motion, which in turn would reproduce the wear of a THR. The minimum requirements for the simulator to satisfy THR wear were wear-rate magnitudes and debris size, shape and distribution. Thus, it was not necessary to have an exact duplicate of the body but to have only the requirements necessary to test an actual implant and produce wear-rate magnitudes and similar debris. For clarification only machines that reproduce M-dST and accommodate an actual implant have been classified as simulators.

The multitude of forces imposed on the hip joint have been simplified into 3 orthogonal axis corresponding to posterior/anterior (X), medial/lateral (Z) and superior/inferior (Y) axis of the body (fig. 2.6; Paul, 1966). Gait analysis studies using telemetry and mathematical analysis have reported the magnitudes of these forces for various activities. By far the largest force reported was in the Y-axis (fig 2.6). For normal walking the vertical force magnitudes have been up to 4 times greater than body weight (table 2.14).

Table 2.14 Force Analysis from Gait Studies Performed during Normal Walking

All results are in multiples of body weight NR = Not Reported

Author	*Year*	*Superior/ Inferior*	*Anterior/ Posterior*	*Medial/ Lateral*	*Resultant*	*Comments*
Paul	1966	3.9	1.0	1.3	2.8	
Brown et al	1984	NR	NR	NR	3.8 2.8 3.5	Muller Hip Charnley Hip Normal Hip
Bergmann et al	1993	3.75 3.5 3.0	0.20 0.50 0.25	1.5 1.5 1.25	3.07 3.24 2.25	Patient 1 Patient 2 Left Patient 2 Right
Paul	1993	NR	NR	NR	2.4	
Brand et al	1994	NR	NR	NR	2.5	
Average		**3.5**	**0.4**	**1.4**	**2.9**	

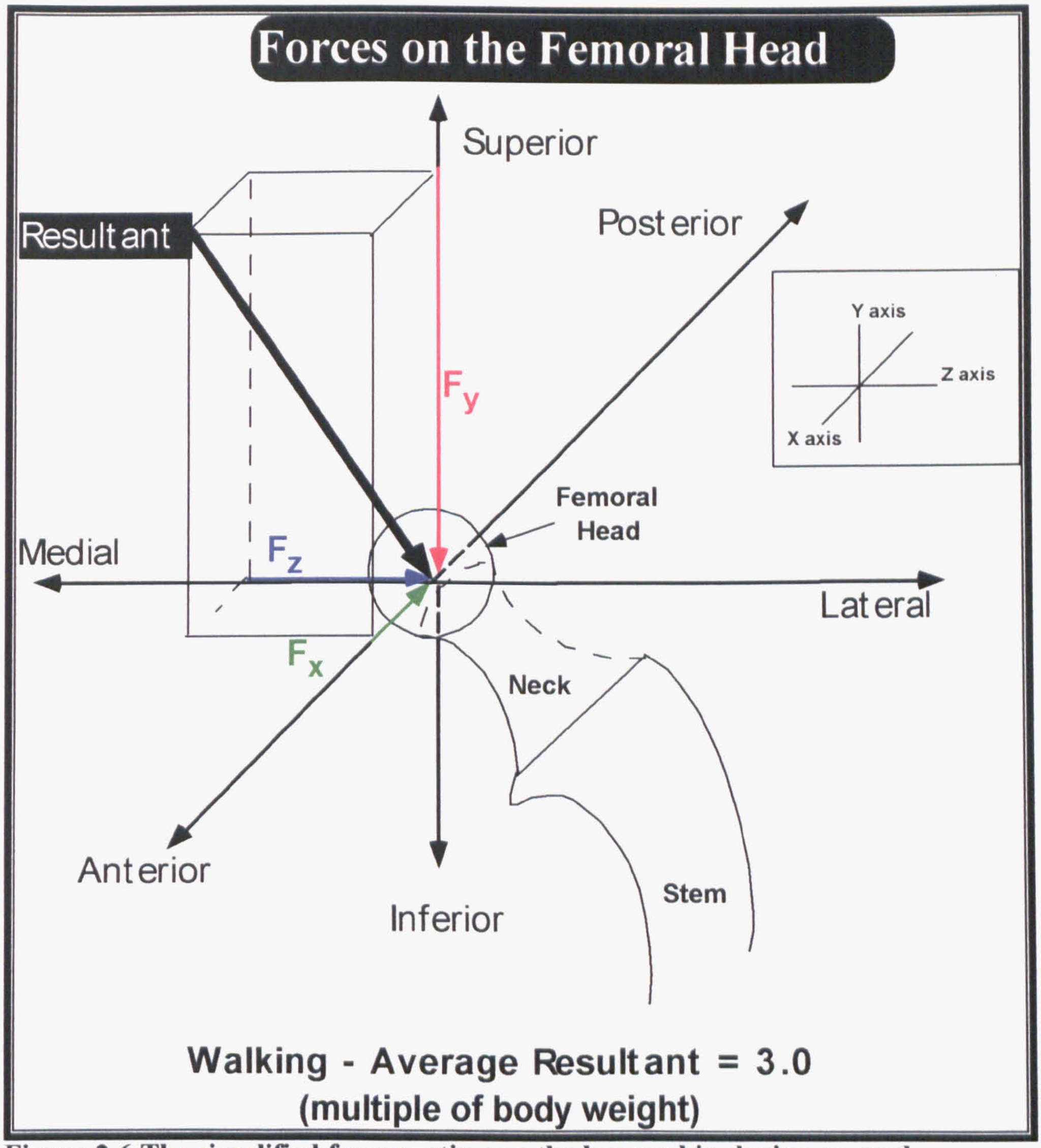


Figure 2.6 The simplified forces acting on the human hip during normal walking. The average resultant force was reported as a multiple of body weight (Bergmann et al, 1993, Brand et al, 1994, Brown et al 1984, Paul, 1966 and 1993).

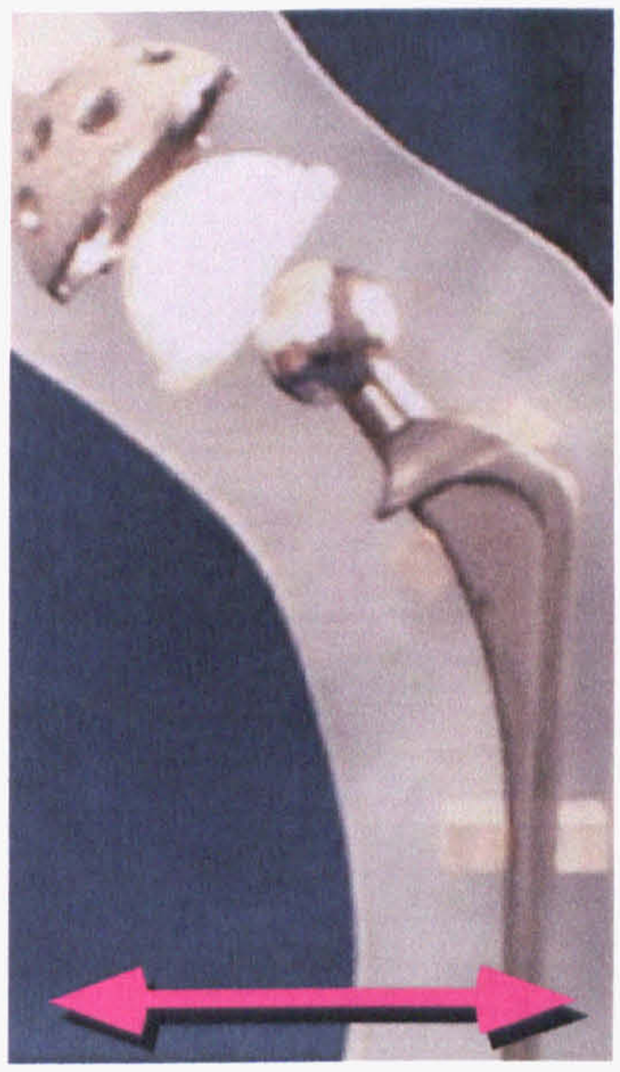
The motions of the hip have also been simplified into three components representing flexion/extension (about the Z-axis), abduction/adduction (about the X-axis) and internal/external rotation (about the Y-axis; fig 2.7). After THR, most people experience decreased range of motion from the normal healthy person. Flexion/extension about the Z-axis averaged $\pm 20^{\circ}$, abduction/adduction about the X-axis averaged $\pm 10^{\circ}$ and internal/external rotation about the Y-axis averaged $\pm 10^{\circ}$ during normal walking (Paul and Poulson, 1974).

It has been found that one of the most important aspects of wear simulation has been the multi-directional surface tracking (M-dST) produced on the THR interface. This multi-directional or crossing path has been found to be a necessity when testing polymers (Bragdon et al, 1996, Ramamurti et al, 1998 and Wang et al, 1997).

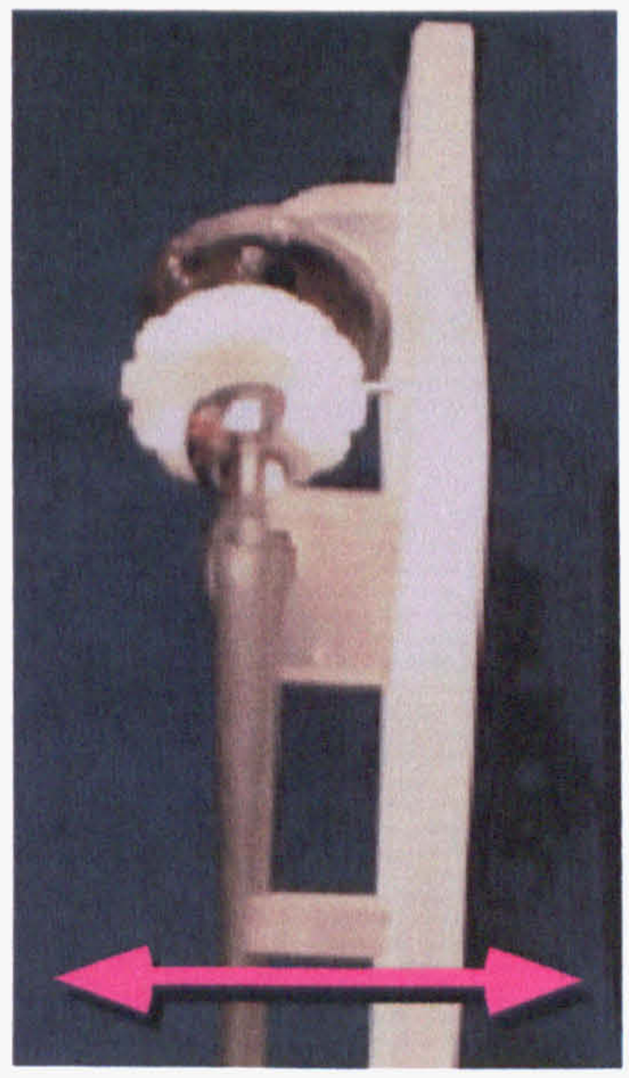
Hip simulators have varied in their duplication of the motions and forces produced on the THR. For ease of interpretation this author has classified the existing simulators into three categories, orbital, 2-axis and 3-axis machines (table 2.15). The orbital simulators rocked about the horizontal axis at $\pm 23^{\circ}$, and this horizontal axis also rotates about the vertical axis at the inputted frequency, thus incorporating both flexion/extension and abduction/adduction motions. The kinematics of these simulators were fixed. The 2-axis simulators and 3-axis simulators were capable of varying their respective angles (Dowson and Jobbins, 1988, Dumbleton, 1981, and Wright and Scales, 1977; fig 2.7).

With the exception of two simulators (Dowson and Jobbins, 1988 and Walker et al, 1968) all studies report the use of only one axis of loading (table 2.16). However studies using simulators with multiple load inputs have not reported on use

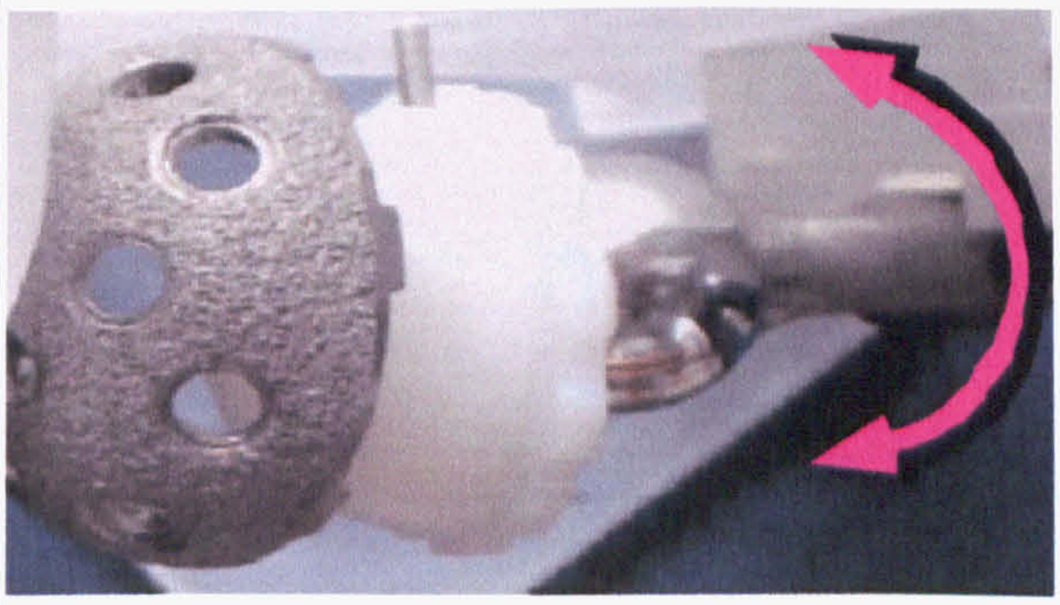
Motions of the Hip



A) adduction/abduction



B) extension/flexion



C) internal/external rotation

The average human hip experiences the following motions during normal walking

A) adduction/abduction - side to side swing of the leg in the frontal plane

B) flexion/extension - front to back swing of the leg in the sagittal plane

C) internal/external - inward and outward rotation of the leg in the horizontal plane

Figure 2.7 The three motions of the human hip during normal walking

Table 2.15 M-dST Simulator Comparisons

Anat = anatomical, Inv = inverted, NR = Not Reported

Author & Year	*Simulator Classification*	*No. Of Motions*	*No. of Forces*	*Motion Input*	*Variable Load Magnitude and Profile*	*Variable Frequency*	*THR Configuration*	*No. of Stations*	*UHMWPE Wear Magnitude*
Duff-Barclay & Spillman, 1967	3-axis (Stanmore MKI)	3	1	Variable	Yes	No	Anat	1	Low
Ungethum, 1976	3-axis (Munchen)	3	1	NR	NR	NR	Anat	1	NR
Wright & Scales. 1977	3-axis (Stanmore MKII)	3	1	Yes	Yes	No	Anat	1	Low
Dowson & Jobbins, 1978	3-axis (Leeds)	3	3	Yes	Yes	Yes	Anat	2	Low
Streicher et al, 1990	3-axis	3	1	Yes	Yes	No	Inv	2	Low
Saikko, 1996	3-axis	3	1	Yes	Yes	Yes	Inv	1	Low
Bragdon et al, 1996	3-axis (AMTI)	3	1	Yes	Yes	Yes	Anat	12	Low
Walker et al, 1968	2-axis	2	3	Yes	Yes	Yes	Anat	1	NR
Buchholz & Strickle, 1972	2-axis	2	1	Yes	Yes	Yes	Anat	1	NR
Geiger et al, 1981	2-axis	2	1	Yes	Yes	Yes	Anat	1	High
Smith et al, 2000	2-axis (Durham)	2	1	Yes	Yes	Yes	Anat	5	Low
Clarke, 1981	Orbital (Matco)	2	1	Fixed	Yes	Yes	Inv	10	NR
Clarke et al, 1993	Orbital (SWM)	2	1	Fixed	Yes	Yes	Inv	9	Low
Polineni et al, 1995	Orbital (MTS)	2	1	Fixed	Yes	Yes	Anat	8	Low
McKellop et al, 1997	Orbital (SWM)	2	1	Fixed	Yes	Yes	Inv	12	Low

of more than one load. (Derbyshire et al, 1994, Dowson and Jobbins, 1988 and Walker et al, 1968). All classifications of simulators applied the resultant force which could be varied in magnitude and dynamic load profile. The direction of the resultant force was maintained by either inclining the components or inclining the load cell to match to the loads resultant direction in the body (fig 2.6).

In addition to motions and angle input there were two other differences among the three classifications of simulators. These were the number of test stations and the tracking distance produced by the wear path. The number of test stations was limited to one or two in the 3-axis and 2-axis simulators before 1996. After that time only 2 machines have had more than 3 test stations, the AMTI and the Durham simulators (table 2.15). On the other hand, the orbital simulators have had at least 8 test stations (table 2.15). The orbital simulators, with multiple specimens, could report on repeatability within one simulator run and could test more than one parameter at a time. The second difference involved the distance traveled for a point on the femoral head. A study by Ramamurti et al (1998) showed that the distance was greater for the SWM orbital simulator than for the Leeds and AMTI 3-axis simulators. They also reported that the SWM simulator distance was approximately 1.5 times greater than clinical (Ramamurti et al, 1998). It has been suggested that this led to a more aggressive wear situation with more wear for the orbital machines (Lerdahl and Vasquez, 2000, Ramamurti et al, 1998 and Schmidt and Lunn, 1998). This would be a benefit for the orbital simulators. In wear testing, laboratories should be examining the more aggressive parameters in order to identify implants that can hold up to more active as well as less active patients.

Table 2.16 M-dST Hip Joint Simulator Studies

SS = Stainless Steel, Al = Alumina, Ti = Titanium, Zr = Zirconia, Anat = anatomical, Inv = inverted, orbital = fixed 2-axis, NR = Not Reported

Author	*Year*	*Head Dia (mm)*	*Material*	*Fluid*	*Million Cycles*	*Configuration*	*Wear-rate (mm^3/Mc)*	*Comments*
Duff-Barclay & Spillman	1967	22 28 28	Charnley PE/SS CoCr/CoCr of: McKee-Farrar Stanmore	Plasma	1.62	MKI - 3 axis, 1 load Anat 1 wear station	14.7 13.4 10.8	These were the only wear-rates not subject to interpretation
Walker et al	1968	22	Charnley UHMWPE	NR	NR	2 axis, 3 load 1 wear station	"satisfactory"	
Buchholz & Strickle	1972	38	HDPE/CoCr	*Custom	NR	2 axis, 1 load Anat 1 wear station	136	*Electrolytes in fluid
Ungethum	1976	NR	NR	NR	NR	3 axis, 1 Anat 1 wear station	NR	Munchen 1 No tests reported
Wright & Scales	1977	25	RCH1000/CoCr	Bovine serum	8.7 5.2	MKI & II - 3 axis, 1 load, Anat 1 wear station	^55 (MKII) ^6.25 (MKI)	Authors reported 1.74Mc = 1year ^Converted to 1Mc = 1year
Wright & Scales	1980	25 32 32	RCH1000/CoCr RCH1000/Al RCH1000/CoCr	Bovine serum	5.2	MKII - 3 axis, 1 load, Anat 1 wear station	101 32 67	
Clarke	1981	28	PTFE/CoCr	Bovine serum	0.06	orbital, 1 load Inv 10 wear stations	2100	Calibration tests
Geiger et al	1981	42	UHMWPE/SS	Mineral oil	0.7	2 axis Inv 1 wear station	166 - 415	
McKellop & Clarke	1983	28 28 32	1900/SS PTFE/SS RCH1000/CoCr	Bovine Serum	1	orbital, 1 load Inv 10 wear stations	42 3000 79	
McKellop & Clarke	1985	28	UHMWPE/SS UHMWPE/CoCr UHMWPE/Ti	Bovine serum		orbital, 1 load Inv 10 wear stations	44 29-115 60-145	Cups and heads from 3 Orthopaedic Companies All PE 2.5Mrad
Dowson & Jobbins	1988	22	Charnley/SS Charnley/Al	Water	3	3 axis, ~3 load Anat 2 wear stations	8.7 2.49	Leeds Simulator ~ only vertical load used

Table 2.16 M-dST Hip Joint Simulator Studies Continued

Author	*Year*	*Head Dia. (mm)*	*Material*	*Fluid*	*Million Cycles*	*Configuration*	*Wear-rate (mm^3/Mc)*	*Comments*
Streicher et al	1990	28	CoCr/CoCr	30% serum	2.5	3 axis, 1 load, Inv 3 wear stations	[a]12 to 24 [b]1.2 to 2.4	[a] Run-in [b]Steady-state MKIII is capable of 0.5Hz and a max load of 1550N
McKellop et al	1992	32	Enhanced (Enh) PE Enh Hylamer Conventional PE	Bovine serum	4	orbital, 1 load Inv 10 wear stations	31.7 30.4 31.6	
McKellop et al	1992	32	UHMWPE/CoCr UHMWPE/Al UHMWPE/Zr	varied solutions	2	orbital, l load Inv 10 wear stations	[c]22, [d]33, [e]80 [c]33, [d]50, [e]105 [c]36, [d]14, [e]70	[c] Serum [d] Serum/EDTA [e] Saline/Albumin
Mejia & Brierley	1993	32	UHMWPE/CoCr	Bovine serum	5	orbital, 1 load Inv 8 wear stations	NR	Wear was "comparable" to other simulator data
Derbyshire et al	1994	22 26 32	GUR 412/SS GUR 412/Zr	Water	2	3 axis, ~3 load Anat 2 wear stations	Unknown	Leeds Simulator ~only vertical load used Wear assessed dimensionally only 2 data pts. Creep was confounding factor
Wang et al	1995	32	GUR415/CoCr	Bovine serum Water		orbital ,1 load Anat 8 wear stations	NR	Debris morphology studied
Clarke et al	1995	22	PTFE/CoCr	Bovine serum water	0.137 0.233	orbital, 1 load Inv 9 wear stations	2572 1140	

The three classifications of simulators had similarities as well. Machines from each of the three categories were capable of varying force magnitude, load profile inputs, frequency and were synchronized for motion and load. Wear path studies showed that these machines produced M-dST (Bragdon et al, 1996. Ramamurti et, 1998, Smith et al, 2000 and Wang et al, 1997). Additionally, machines from these groups could configure the THR in the anatomical (Anat; cup on top) or inverted (Inv; cup on bottom) position.

There were distinct advantages to the different simulators. The standardized kinematics of the orbital simulator produced consistency for multiple wear tests. In addition, its simpler mechanics reduced the odds of mechanical failure. The 3-axis and 2-axis simulators, on the other hand, had the advantage of variable motion inputs. These machines could generate information on variable range of motion studies.

The history of simulators began with the STANMORE MKI developed in 1962 (Duff-Barclay and Spillman, 1967 and Wright and Scales, 1977). The STANMORE MKI incorporated the 3 axes of motion: flexion/extension, adduction/abduction and internal/external motion (fig 2.7) and 1 force component up to 2.5 times body weight (fig 2.6). The cup and ball were mounted Anat and only 1 THR combination could be run at a time (table 2.16). The first reported use of this simulator was from Duff-Barclay and Spillman, (1967). They studied various material combinations such as; metal on same metal of CoCr and SS and plastic combinations of UHMWPE and Delrin™ mated with CoCr. These materials were run dry or with saline or plasma as the JFA. Much of the data from this study was presented in graphs and wear-rate magnitudes were subject to interpretation.

However they did report numerical wear-rates for three of the studied combinations, 22mm Charnley UHMWPE/SS, 28mm McKee-Farrar CoCr/CoCr and Stanmore CoCr/CoCr. The wear-rate magnitudes for the three materials varied by less than 30% with the Charnley having the greatest wear. These results were contrary to the clinical findings. The clinical wear-rate averages for metal on metal have been found to be approximately 2 orders of magnitude less than UHMWPE (McKellop et al, 1996). In addition, the wear-rate magnitude of the Charnley was approximately 70% less than the clinical wear-rate reported 6 years later by Charnley and Cupic, (1973). Thus, despite the physiological nature of the simulator, it did not duplicate clinically relevant wear-rates.

By far the most limiting factor to early simulators was that only one THR set could be tested at a time. Running one THR implant at 1 Hz for a duration of 3 million cycles (Mc) would take approximately 6 weeks. For the MKII, which ran at 0.5 Hz, this time was doubled. This drawback was partially responsible for the lack of interest in these machines. The simple LST devices could test more than one specimen, were less expensive to build and didn't have the added expense of requiring an actual THR.

In 1981 Clarke introduced the first multi-station (10 THR sets) orbital simulator (Clarke, 1981). The M-dST simulator combined the motions of abduction/adduction and flexion/extension rocking the THR $\pm 23^{\circ}$ about the horizontal axis and rotating this axis about the vertical axis. The vertical load applied represented the resultant force. The load profile could be varied by computer control and the maximum dynamic force was 2kN. This simulator was much simpler than the 3-axis simulator and reproduced the M-dST that was absent in the LST

machines. The increased THR capacity meant that a study of 10 THR sets for 3 Mc could be completed within 6 weeks. This meant an experiment on the orbital simulator would take only a 10^{th} of the time that a single station 3-axis simulator needed for the same study.

The THR was mounted with the cup below the head (Inv) in the orbital machine. The Inv configuration was at less risk for JFA starvation. Two factors contribute to JFA starvation a) the fluid level dropping below the interface and b) cavitation. Evaporation, either from the normal loss of fluid over time or from a change in conditions (e.g. large increase in friction and heat) could cause fluid levels to drop below the THR interface and cause the interface to become dry. In the Inv configuration the THR interface was the lowest point and was constantly wet as long as there was fluid in the chamber. Cavitation was the second concern for JFA starvation. Unsworth et al (1971) showed that cavitation was produced by dynamic loading of the finger joints. Cavitation from the dynamic loading of the THR would cause gas bubbles to form. These bubbles rise to the top and gather at the THR interface in the Anat configuration. With changes in pressure these bubbles would collapse temporarily leaving the interface dry with subsequent intermittent JFA starvation. However, in the Inv configuration the gas bubbles would escape from the THR interface. Thus, the Anat configuration was at a greater risk for JFA starvation.

UHMWPE wear-rates from the 3 classifications of simulators have generally underestimated the clinical wear-rates (table 2.15 and 2.16). Regardless of machine type, the in-vitro wear-rate of UHMWPE has averaged 2 to 4 times less than clinical in physiological fluids. In a study by Schmidt and Lunn (1998) an orbital and 3-axis machine were compared. They found that the orbital simulator produced 10 to 30%

more wear than the 3-axis machine. Wang (1998) reported that the difference in wear-rate of UHMWPE in the Leeds and MTS simulator was proportional to the difference in sliding distance per cycle. Therefore, the wear-rate difference in the study by Schmidt and Lunn (1998) could be due to the difference in sliding distance for the Matco and AMTI simulators. However, the largest wear-rates for the orbital simulator were at least 2 times less than clinical results.

The size and morphology of UHMWPE debris from both the 3-axis and orbital simulators has been found to be similar to that found clinically when using bovine serum as the JFA (Bragdon et al, 1997, McKellop et al, 1995 and Wang et al, 1995). However, a greater percentage of round versus elongated particles have been found in-vitro. On the other hand, when using water as the JFA the debris from both UHMWPE and PTFE has been wholly uncharacteristic (large flakes visible to the naked eye) of clinical findings (Clarke et al, 1995, Dowson and Jobbins, 1988 and Saikko, 1996). Thus, there has been some doubt as to whether simulators have reproduced all of the in-vivo wear debris characteristics.

These studies have shown that the three classifications of M-dST simulators have produced similar UHMWPE wear, regardless of the kinematics. The M-dST simulators duplicated the necessary motions needed for polymer wear testing, whereas the LST devices did not. Despite this and the better approximation of clinical wear, the M-dST simulators still had not duplicated UHMWPE wear-rates and debris. UHMWPE wear-rate magnitudes from simulators were an underestimation of clinical wear-rates and the debris round/elongated ratio was overestimated, thus showing that the model approximation of clinical wear involved more than the kinematics of the simulator.

2.5 FLUIDS IN HIP WEAR TESTING

2.5.1 History of Non-physiological JFA

There have been two types of fluids used in hip joint testing, non-physiological (without proteins) such as water, saline, mineral oil or Ringers solution and physiological (with proteins) such as bovine serum or plasma. Many reasons for using non-physiological fluids have been proposed. Dumbleton, (1978) wrote, "If it is desired to duplicate the actual environment of the joint then some type of synovial fluid should be used; however, it is not clear that this is an essential requirement and distilled water may serve just as well". Dowson, (1978) states that synovial fluid is basically a dialyzate of blood and a watery substance therefore, it would be reasonable to test with water. Another consideration for using non-physiological fluids was the cost and characteristics. Water was by far much less costly than bovine serum, more readily available and there was no worry about bacterial growth or degradation. Therefore, many studies have been conducted in non-physiological fluids throughout the 30-year history of implant testing and most of these used distilled water (table 2.16). Over 75% of these studies were on LST devices that have been shown to produce no reliable hip wear data (table 2.12) therefore, only studies using simulators were examined.

The results from studies conducted on M-dST devices were varied and inconsistent. Charnley's results for UHMWPE/CoCr in water were one tenth of the clinical findings (Charnley, 1976). In two other studies, one in mineral oil and the other in saline with albumin, wear-rates were higher or ranked incorrectly to clinical (Geiger, 1981, McKellop, 1992). Duff-Barclay and Spillman, (1967) also showed

that wear-rates of CoCr/CoCr were higher than clinical or in the case of stainless steel (SS) and

titanium (Ti) produced seizure of the components. A study by Clarke et al, (1995) tested 22mm PTFE in water and produced clinically relevant wear-rates but large flakes of wear debris were visible to the naked eye. These large flakes were contrary to the clinical data presented by Charnley, (1969). He reported that the debris in-vivo was 5 to 50μm long and that no large particles were found. Of interest was that without exception and regardless of machine type or polymer material all studies with mention of debris reported large particles in the form of streamers or flakes (table 2.17). This was in direct contrast to in-vivo UHMWPE debris found in the tissues around THR and the PTFE data reported by Charnley. UHMWPE particles have been shown to be sub-micron in size, circular in shape with approximately 25% fibrils (Campbell et al, 1995). Thus, simulator studies using non-physiological fluids showed high and low wear-rates along with seizure of some metals, incorrect clinical ranking and uncharacteristic debris. Therefore, these studies did not duplicate any reliable hip wear data.

2.5.2 History of Physiological JFA

Laboratories were not able to use the human hip joints' own synovial fluid in wear testing because of the amounts necessary to run a test. The amount of synovial fluid in human joints varies between 0.2 and 10ml in healthy (non-diseased) and up to 50ml in diseased (Altman and Gray, 1984, Binette and Schmid, 1965, Geigy Tables, 1981, Kitridou, 1972, McCarty, 1979, Rippey, 1979, Shanfield et al, 1988 and Yehia and Duncan, 1975). Test chambers vary in volume from 40 to 600ml, which for a multi-chamber test of 9 THR components would require between 4320 to

Table 2.17 Non Physiological Fluids in Laboratory Hip Wear Studies

SS = Stainless Steel, Al = Alumina, Ti = Titanium, Zr = Zirconia, M-dST = multi-directional surface tracking, LST = linear surface tracking, NR = Not Reported

Author	*Year*	*Head Dia. (mm)*	*Material*	*Fluid*	*Testing Device*	*Debris*	*Wear-rate (mm^3/Mc)*	*Comments*
Duff-Barclay & Spillman	1967	NR	SS/SS & Ti-160/ Ti-160 CoCr/CoCr	Saline	M-dST	NR	seized ~50	Had to stop tests of SS and Ti CoCr showed high amounts of burnishing
Homsey & King	1969	NR	PTFE/SS Delrin™/SS UHMWPE/SS	NaCMC in PECF	LST	NR	33mm/y 0.71mm/y 1.14mm/y	NaCMC (Sodium carboxymethylecellulose) with PECF (inorganic ionic constituents)
Galante & Rostoker	1973	NR	RCH1000/CoCr RCH1000/Ti Delrin™ CoCr Al/Al	Water	LST	NR	2×10^{-9}in/in 1.5×10^{-9} in/in 2 to 5×10^{-9} in/in 1.2×10^{-7} in/in	
Walker & Erkman	1974	NR	UHMWPE AP4	Water	LST	NR	AP4>UHMWPE	Wear by debris collection Size of debris not reported
Gold & Walker	1974	22& 28	UHMWPE/SS UHMWPE/CoCr PTFE/SS PTFE/CoCr	Water	LST	PE debris size several times > clinical	PTFE>>UHMWPE	No units on wear numbers
Dowson et al	1974	NR	RCH1000/SS	Water	LST	NR	8.3×10^{-9} & 2.2×10^{-8} mm^3/Nm	
Gallante & Rostoker	1976	NR	UHMWPE/CoCr AP5/CoCr	Water	LST	NR	AP5>UHMWPE	
Charnley	1976	NR	PTFE/SS RCH1000/SS	Water	M-dST	NR	25.69 0.147 mi/in	PE wear 10x < clinical
Semlitsch et al	1977	32	UHMWPE/CoCr UHMWPE/Al	Water	LST	NR	140 7	
Mustafaev et al	1978	NR	RCH1000/SS	Mineral oil	LST	NR	1.38×10^{-7} mm^3/Nm	
Dowson	1978	NR	UHMWPE/SS	Water	LST	NR	10^{-9} mm^3/Nm	
McKellop et al	1978	NR	UHMWPE/SS	Water Saline	LST	large flakes	0.08mg/Mc 4.76mg/Mc	
Geiger et al	1981	42	UHMWPE/SS	Mineral oil	M-dST	NR	415 166	
Dowson & Harding	1982	NR	RCH1000/Al	Water	LST	NR	10^{-8} to 10^{-9} mm^3/Nm	
Dowson & Wallbridge,	1985		PTFE/SS	Water Ringers	LST		1.4×10^{-5} mm^3/Nm 4.4×10^{-5} mm^3/Nm	

Table 2.17 Non Physiological Fluids in Laboratory Hip Wear Studies Continued

Author	*Year*	*Head Dia. (mm)*	*Material*	*Fluid*	*Testing Device*	*Debris*	*Wear-rate*	*Comments*
Dowson & Jobbins	1988	NR	UHMWPE/SS UHMWPE/Cer	Water	M-dST	Several μm long	SS>ceramic	Cer = "new" ceramic
McKellop et al	1992	32	UHMWPE/CoCr UHMWPE/Al UHMWPE/Zr	Saline with albumin	M-dST	NR	80 105 70	
Saikko	1993	NR	UHMWPE/CoCr UHMWPE/Al UHMWPE/Zr	Water	LST	NR	1.05mg/Mc 0.04mg/Mc 0.03mg/Mc	
Cooper et al	1993	NR	UHMWPE/SS	Water	LST	NR	2.8×10^{-7} to 6.0×10^{-9} mm^3/Nm	
Saikko et al	1993	32	UHMWPE/SS UHMWPE/CoCr UHMWPE/Al	Water	LST	cm long streamers	53 & 178mg/Mc 4 & 37mg/Mc 0.03 to 6 mg/Mc	Wear varied by manufacturer
Cooper et al	1993	NR	UHMWPE/?	Water	LST	long streamers due to transfer film	3.5 to 7.8×10^{-9} mm^3/Nm	
Derbyshire et al	1994	22 26 32	GUR 412/SS GUR 412/Zr	Water	M-dST	NR	unknown	Wear assessed dimensionally with only 2 data pts. Creep confounding factor
Saikko	1994	32	UHMWPE/Zr	Water	LST	2cm long streamers	4.4	
Derbyshire et al	1995	NR	UHMWPE/SS	Water	LST	NR	4.2×10^{-9} mm^3/Nm	
Pappas et al	1995	47	UHMWPE/Ti	Water	LST	NR	1.04mg/Mc	
Saikko	1995	28	UHMWPE/Zr UHMWPE/CoCr UHMWPE/Zr	Water	LST	NR	21.8mg/Mc 83mg/Mc no wear	7.0mm thick 10.9mmthick 10.9mmthick
Clarke et al	1995	22	PTFE/CoCr	Water	M-dST	large flakes	1140	
Saikko	1996	32	UHMWPE/CoCr	Water	M-dST	20μm thick Several mm diameter	44	
Wroblewski et al	1996	22	UHMWPE/Al	Water	M-dST	NR	[a]46.6 [b]6.2	[a]Run-in [b]Steady-state

64,800ml per three million cycle test (average UHMWPE test duration). This would necessitate 86 to 1,296 (diseased) or 2,160 to 32,400 (healthy) patient donations per experiment (fig 2.8). Thus, laboratories had to adopt another type of fluid for use as a JFA, which would duplicate the wear behavior of THR combinations in a similar way to the clinical data.

Synovial fluid is a dialyzate of blood serum (Freemont and Denton, 1991, Rippey, 1979, Swann, 1978, Yehia and Duncan, 1975). Healthy synovial fluid contains about 1/3 of the proteins in blood serum. However, the ratio of small molecular weight proteins to large molecular weight proteins was greater in synovial fluid. The albumin (small molecular weight) to globulin (large molecular weight) ratio was approximately 1.6:1 versus 1.0:1 for synovial fluid and blood serum respectively (table 2.18). The diseased joint exhibits increased inflammation and an increased volume of synovial fluid. As the amount of inflammation increased, the concentration of proteins increased (McCarty, 1979, Pruzanski et al, 1973, Rippey, 1979, Yehia and Duncan, 1975). For osteoarthritis and rheumatoid arthritis the synovial fluid protein concentration increased by 1.8 to 2.4 times respectively (table 2.18). However, the total amount of protein was still at least 1.5 times lower than serum. Healthy synovial fluid is also very viscous, however in disease the viscosity decreases and in rheumatoid arthritis viscosity is very low similar to serum (Cracchiolo, 1971, Kitridou, 1972, Swann et al, 1974 Yehia and Duncan, 1975) Bovine synovial fluid has been used as a JFA, however, the protein concentration was 2.5 to 3 times lower than human synovial fluid (Andersson and Liberg, 1980 and Ropes et al, 1940). Thus, proteins would have to be added to simulate human synovial fluid. Bovine serum has also been used as a JFA and to examine its

Figure 2.6 The human joint contains on average 2ml of synovial fluid. A 12-station hip joint simulator requires 250 to 500 ml of JFA per station and event. Therefore, it would require a minimum of 22,500 donations to fill the simulator with synovial fluid for one UHMWPE experiment.

Table 2.18 Human Synovial Fluid and Serum Protein Concentrations

Healthy = Non-diseased, OA = Osteoarthritis, RA = Rheumatoid Arthritis

Author	*Year*	*Synovial Fluid Total Protein (mg/ml)*			*Synovial Fluid Albumin/Globulin Ratio*			*Serum Total Protein (mg/ml)*			*Serum Albumin/Globulin Ratio*		
		Healthy	OA	RA	Healthy	OA	RA	Healthy	OA	RA	Healthy	OA	RA
Ropes et al	1940	25.7											
Schmid and MacNair	1958	15			1.6								
Decker et al	1959	21.5	29.9	41.9	0.3	0.4	0.6	63	67	66	0.62	1	0.78
Nettelbladt and Sunbladt	1959					1.1	1.5					1.5	1
Wilkinson and Jones	1962		39.9	49		1.4	0.8		71	70.3		1.1	0.8
Schur & Sandson	1963	20.2		44.9									
Kushner and Somerville	1971	19	43	58									
Geigy Tables	1981	18		45	1.5		0.8	70			1		
Walker	1972	17.2			1.4								
Willumsen and Friis	1975		32.6	44.1			0.6		68	71.1		1.1	2
McCarty	1979	18											
Simkin	1979	20											
Dumbleton	1981	20			4			70			1.5		
Altman and Gray	1984	17	31										
Wallis et al	1987		30.6	34.6	0.5	0.5			61.6	64.1			
Berntzen et al	1991												
Sawada et al	1991												
Saari et al	1993		31.8										
Streicher et al	1996	20.3											
Noordin et al	1997		33										
Reiker et al	1998	17											
Average		**19.1**	**34.0**	**45.4**	**1.6**	**0.9**	**0.9**	**67.7**	**66.9**	**67.9**	**1.0**	**1.2**	**1.1**

relationship to human serum, six lots of bovine serum (all from the same manufacturer , Hyclone, Inc. Logan, Utah) were examined for their protein concentration and albumin/globulin ratio. The average protein concentration and albumin/globulin ratio was 68mg/ml and 1.2:1 respectively (table 2.19). Human blood serum averaged similar protein concentrations regardless of health condition and the albumin/globulin ratio for bovine serum was similar to the ostoearthritic patient's blood serum (table 2.20). Thus, with regard to protein concentration and the albumin/globulin ratio, bovine serum was very similar to human serum, which was the basis of synovial fluid.

Table 2.19 Bovine Serum Protein and Albumin/Globulin Ratio
(All bovine serum from Hyclone, Inc. Logan, Utah with lots as specified)

Bovine Serum	*Protein (mg/ml)*	*Albumin (mg/ml)*	*Globulin (mg/ml)*	*Albumin/Globulin Ratio*
Lot AFD5137	70.0	41.0	29.0	1.4
Lot 21112355	63.0	36.0	27.0	1.3
Lot 21112521	70.0	41.0	29.0	1.4
Lot AFK5740	69.0	41.0	28.0	1.5
Lot AGK7210	65.0	31.0	34.0	0.9
Lot AHL9372	69.0	32.0	37.00	0.9
Average	***68***	***37***	***31***	***1.2***

The use of physiological fluids (containing proteins) and in particular bovine serum has predominated M-dST simulator testing. With very few exceptions, UHMWPE wear-rates in bovine serum have been 1.2 to 10 times lower than clinical (table 2.1 and 2.20). In contrast 2 UHMWPE studies (Buchholz and Strickle, 1972 and Wright and Scales, 1980) had high wear-rates. However, at closer inspection it was found that the high-density polyethylene tested by Buchholz and Strickle, (1972)

Table 2.20 Physiological Fluids Used in Hip Wear Studies

SS = Stainless Steel, γ = gamma irradiation, Al = Alumina, Zr = Zirconia, Ti = Titanium, M-dST = multi-directional surface tracking, LST = linear surface tracking, NR = Not Reported

Author	*Year*	*Head Dia (mm)*	*Material*	*Fluid*	*Testing Device*	*Wear-rate (mm^3/Mc)*	*Comments*
Duff-Barclay & Spillman	1967	22 28 28	Charnley PE/SS CoCr/CoCr of: McKee-Farrar Stanmore	Plasma	M-dST	14.7 13.4 10.8	These were the only wear-rates not subject to interpretation Unknown if PE γ sterilized
Scales & Lowe	1971	NR	RCH1000/CoCr RCH1000 γ /CoCr	Bovine serum	LST	26 (mm^3/hourx10^{-3}) 58 (mm^3/hourx10^{-3})	
Buchholz & Strickle	1972	38	HDPE/CoCr	*Haemaccel	M-dST	136	*Electrolytes and proteins (polypeptides) in fluid
Seedholm et al	1973	NR	UHMWPE(γ air)/SS	Bovine Synovial Fluid	LST	2.7 x 10^{-9} to 4.3 x 10^{-10} mm^3/Nm	
Dumbleton et al	1974	NR	UHMWPE/SS UHMWPE 20Mrad/SS	Plasma	LST	10^{-12} mm^3/Nm 3.6 x 10^{-12} mm^3/Nm	
Charnley	1976	NR	RCH1000/SS PTFE/SS	Synovial Fluid Plasma 5% Gelatin	Random Motion Pin on flat	PTFE PE 193 0.160 200 0.130 260 0.180	All wear numbers are x10^{-4} in/mi Unknown if PE γ sterilized
McKellop et al	1977	NR	UHMWPE/SS UHMWPE/CoCr Delrin™/SS PTFE/SS Polyester/SS	Bovine serum	LST	1.6μm/Mc 0.98μm/Mc 90μm/Mc 2400μm/Mc 1350μm/Mc	
Wright & Scales	1977	25	2.5Mrad/air RCH1000/CoCr	Bovine serum	MKI & MKII M-dST	* 55 (MKII) *6.25 (MKI)	Authors reported 1.74Mc = 1y *Converted to 1Mc = 1y (the accepted standard)
McKellop et al	1980	NR	UHMWPE/SS PE 2.5Mrad air/SS PE 7.5Mrad air/SS	Bovine serum	LST	No measurable wear 0.11 0.31	
Wright & Scales	1980	25 32 32	RCH1000/CoCr RCH1000/Al RCH1000/CoCr	Bovine serum	M-dST	101 32 67	Unknown if γ sterilized
McKellop et al	1981	NR	UHMWPE/SS Carbon fiber PE/SS Polyacetal/SS Delrin™ 500/SS AP4/SS PTFE/SS	Bovine serum	LST	0.20 0.35 2.7 9.4 160 322	

Table 2.20 Physiological Fluids Used in Hip Wear Studies Continued

Author	*Year*	*Head Dia (mm)*	*Material*	*Fluid*	*Testing Device*	*Wear-rate (mm^3/Mc)*	*Comments*
Clarke	1981	28	PTFE/CoCr	Bovine serum	M-dST	2100	
McKellop & Clarke	1983	28 28 32	1900/SS PTFE/SS RCH1000/CoCr	Bovine serum	M-dST	42 3000 79	2.5 Mrad sterilization to PE
McKellop et al	1985	41	UHMWPE/Delrin™ UHMWPE/CoCr	Bovine serum	M-dST	*80 98	*PE Cup 61 mm^3/Mc and Delrin™ head 19 mm^3/Mc 2.5 Mrad sterilization to PE
McKellop & Clarke	1985	28	UHMWPE/SS UHMWPE/CoCr UHMWPE/Ti	Bovine serum	M-dST	44 29-115 60-145	2.5 Mrad sterilization
Dowson & Wallbridge	1985	NR	PTFE/SS	Bovine Synovial Fluid	LST	5.7×10^{-5} mm^3/Nm	
Weightman & Light	1986	NR	UHMWPE/SS UHMWPE/Al	Bovine serum	LST	0.28×10^{-5} mm^3/Nm 0.44×10^{-7} mm^3/Nm	
McKellop & Clarke	1988	NR	UHMWPE/SS UHMWPE/CoCr UHMWPE/Al	Bovine serum	LST	0.1 0.09 0.06	
Streicher et al	1990	28	CoCr/CoCr	30% Bovine serum	M-dST	[a]12 to 24 [b]1.2 to 2.4	[a]Run-in [b]Steady-state
Streicher et al	1991	NR	UHMWPE/Al UHMWPE/Zr	30% Bovine serum	LST	2.1×10^{-7} mm^3/Nm 3.2×10^{-7} mm^3/Nm	
Cooper et al	1993	NR	UHMWPE/SS	Bovine serum	LST	0.7 to 1.01×10^{-8} mm^3/Nm	
Mejia & Brierley,	1993	32	UHMWPE/CoCr	Bovine serum	M-dST	NR	Wear was "comparable" to other simulator data
Derbyshire et al	1994	NR	UHMWPE/SS UHMWPE/Zr	Bovine serum	LST	0.4×10^{-8} mm^3/Nm 3.5×10^{-8} mm^3/Nm	
Fisher et al	1994	NR	UHMWPE/SS	Bovine serum	LST	9×10^{-9} mm^3/Nm	
Fisher et al	1995	NR	UHMWPE/SS UHMWPE γ air/SS	Bovine serum	LST	6×10^{-9} mm^3/Nm 2.5×10^{-8} mm^3/Nm	
Derbyshire et al	1995	NR	UHMWPE/SS	Bovine serum	LST	1×10^{-8} mm^3/Nm	
Clarke et al	1995	22	PTFE/CoCr	Bovine serum	M-dST	2572	

was experimental, therefore it was not known if this was a high wear-rate for this material. The study by Wright and Scales, (1980) exhibited 1.5 times higher wear-rates for 25mm RCH1000/CoCr than for 32mm RCH1000/CoCr. This was in direct opposition to the clinical ball size effect of increased wear-rate with increased head size resulting in questionable findings from this study. Another study by Wright and Scales, (1977) was within 12% of the clinical average using their Stanmore MKII simulator. McKellop and Clarke, (1983) studied three materials and only one (32mm UHMWPE cup) reproduced the clinical average. Thus, of the 27 studies presented only two (or less than 10%) were able to duplicate clinical wear-rate magnitudes. Three studies of UHMWPE (Duff-Barclay and Spillman, 1967, Charnley, 1976 and McKellop et al, 1992) were in other physiological fluids and exhibited the same low wear-rates as in bovine serum. The PTFE wear-rates in simulator studies using physiological fluids were, without exception, at least 1.5 times greater than clinical (table 2.20).

The polymer studies in physiological fluids duplicated the clinical magnitudes of UHMWPE in 2 simulator studies. UHMWPE simulator studies using physiological fluids as the JFA exhibited wear-rates that were generally lower than clinical and wear-rates for PTFE in the simulator, using physiological fluids were higher than clinical. Therefore, wear-rates of polymers using physiological fluids were closer to matching the clinical data than non-physiological fluids.

2.6 LITERATURE OVERVIEW AND HYPOTHESES

Wear has been identified as the major obstacle to increasing the longevity of hip implants. The demand for new longer-lasting implants has increased the need for reliable in-vitro wear testing. Unfortunately the past in-vitro studies have not shown

reliability when predicting THR outcomes. Clinical disasters such as Hylamer™, Delrin™ and polyester were shown to have better wear resistance in the laboratory than actually found clinically. Some of these findings used LST devices, which have been shown to be without merit, and therefore, could be ruled out. However, both Hylamer™ and Delrin™ were tested in simulators with M-dST, which also did not predict clinical disaster. Therefore, reliability of in-vitro testing has been in question and in order to gain confidence laboratory studies must first show that clinical findings from known materials can be duplicated in order to gain reliability for wear testing of new materials.

The JFA was shown to be a limiting factor for in-vitro wear studies. The two predominating fluids were water (non-physiological) and bovine serum (physiological). The results from these two fluids were very different; generally water predicted much lower UHMWPE wear-rates than clinically shown with large flakes of debris. Many studies reported a transfer film that led to clinically uncharacteristicly roughened heads and increased wear-rates. The bovine serum results were mixed, with UHMWPE data still lower than clinical but generally higher than in water. However in contrast, studies of PTFE showed wear-rates to be greater than clinical and Delrin™ showed both low and high wear-rates. No transfer films or large flake type debris were reported in bovine serum. M-dST simulator studies in both water and serum showed greater wear-rates than the LST wear studies. However, the trend for lesser and greater than clinical wear-rates of UHMWPE and PTFE respectively continued. Thus, using bovine serum as the JFA in a M-dST simulator was closer to producing clinical wear-rates than studies in water or any studies using LST devices. Therefore it was hypothesized that by varying the in-

vitro JFA and using a M-dST simulator it would be possible to duplicate in-vivo wear-rates.

The specific aims were to test two known clinical materials using the baseline of water and 100% bovine serum and then adjust the protein concentration through dilution of the bovine serum to test the hypothesis. The goal of these studies was to duplicate the known clinical results and therefore validate in-vitro testing.

Two materials that were obvious choices for this validation were PTFE and UHMWPE. These two polymers have been studied clinically and have ample data showing the wear characteristics of both materials. Also, both were on opposite ends of the wear spectrum. UHMWPE has been considered the "gold" standard with low wear-rates and has 30-years of clinical results. PTFE on the other hand, has had the highest clinical wear-rates with clinical data provided by Charnley's retrieval studies. To validate the in-vitro studies it was necessary to determine what clinical criteria would be used.

The clinical wear data of PTFE and UHMWPE have shown that with increased head size came increased wear-rates and the rate of increase was linear (Clarke et al, 1996, Hall et al, 1998). Thus two of the in-vitro wear criteria were, the clinical ball size effect and the rate of increase known as the volumetric wear index or VWI (table 2.21, criteria 1 & 2). Target wear-rate magnitudes of 49, 72 and 82mm^3/y for 22, 28 and 32mm respectively (table 2.3) were used for UHMWPE. The PTFE wear-rate targets from chapter 2.3.1 were 1670, 2268, 2672 and 3810mm^3/y for the 22.25, 28.5, 32 and 41.5mm (results for 32mm were not reported by Charnley, however, the wear-rate target was computed using the 6% per millimeter algorithm; table 2.21 criterion #3). The PTFE/PE volumetric wear-rate

ratio and the debris morphology were the final clinical validation criteria (table 2.21, criteria 4 & 5). The debris morphology for PTFE has not been reported with the same quantitative analysis as UHMWPE. However, Charnley's qualitative description of the debris, "…'stringy' or 'hairy' appearance and they vary greatly in size ranging from 5 to 50 μm." and "In no case were large sheets of PTFE encountered …large enough to be visible to the naked eye." would serve as a simplified criterion. Two additional testing criteria were necessary for accurate simulator validation, precision among the wear specimens and degree of linearity. Therefore it was the goal of this work to duplicate all of the validation criteria by varying the protein concentration in the bovine serum JFA.

Table 2.21 Wear Validation Criteria

No.	*Wear Validation Criteria*
1	Ball Size Effect (increased wear-rate with increased head size)
2	VWI (6-10%/mm of ball dia)
3	Wear-rate Magnitude (±10%)
4	PTFE/PE (range 20-50:1)
5	Debris comparable to in-vivo
6	Precision (95% Confidence Limits (CL) of ±15%)
7	Linearity

CHAPTER 3. MATERIALS AND METHODS

The first four experiments studied the effect of the two most widely used JFAs of water and bovine serum on the wear of PTFE and UHMWPE (table 3.1, HE040, HE050, HE057 and HE071). The next ten experiments examined the effect of protein concentration in the JFA to wear of PTFE and UHMWPE (table 3.1, HE041, HE043, HE044, HE047, HE064, HE066, HE110, HE111, HE114, HE115). Following these experiments four studies of PTFE examined the effect of JFA volume (table 3.1, HE141, HE147, HE041, HE043) and three studies examined the wear response to protein concentration in larger volumes of JFA and three different THR configurations (table 3.1, HE140, HE141, HE147).

3.1 HIP SIMULATORS

All 17 studies were run on orbital type (biaxial) hip simulators (Shore Western Manufacturing Inc., Monrovia, CA) with either 9 or 12 channels of wear and soak. In the biaxial simulator identical vertical loads were applied from below by one hydraulic actuator for each wear station. The following description was for a THR combination mounted Inv (fig 3.1) and the only change for Anat was the reversal of the cup and ball (fig 3.2). The cup was mounted on a rotating swash block, which was inclined at 23° to the horizontal plane with the head mounted above and attached to a self- aligning fixture (fig 3.1). Both the head and cup were constrained from rotating about the vertical axis with anti-rotation pegs, which therefore allowed the biaxial rocking motion with an amplitude of $\pm 23°$.

Table 3.1 Experiment List

All experiments used CoCr Heads. BCS = Bovine Serum, ACS = Alpha Calf Serum, (S) = Static Soak no load in DI Water, (D) = Dynamic Soak, load no rotation, Inv = Inverted, Anat/Horiz = Anatomical Horizontal, Anat/Obliq = Anatomical Oblique (cup inclined at 23°), EDTA = ethylenediamine tetraacetic acid, NaH_3 = sodium azide, NA = Not Applicable

Expt. No.	*Fluid /Protein*	*Fluid ID*	*Volume (ml)*	*Dia. Wear (N)*	*Soak (N)*	*Position*	*Cup Material & Sterilization*	*Material Manufacturer Cup/Head*	*Load Profile*	*Duration (cycles)*	*Serum Additives*
HE040	Water	NA	250	22mm (3) 32mm (3) 42mm (3)	1 (S) 1 (S) 1 (S)	Inv	PTFE None	Coast Plastics /Protek	Sine	525,633	None
HE041	30% BCS 21mg/ml	BCS 3	250	22mm (3) 32mm (3) 42mm (3)	1 (S) 1 (S) 1 (S)	Inv	PTFE None	Coast Plastics /Protek	Sine	290,559	None
HE043	15% BCS 10mg/ml	BCS 3	250	22mm (3) 32mm (3) 42mm (3)	1 (S) 1 (S) 1 (S)	Inv	PTFE None	Coast Plastics /Protek	Sine	377,085	None
HE044	7.5% BCS 5mg/ml	BCS 3	250	22mm (3) 32mm (3) 42mm (3)	1 (S) 1 (S) 1 (S)	Inv	PTFE None	Coast Plastics /Protek	Sine	371,571	None
HE047	4% BCS 3mg/ml	BCS 2	250	22mm (3) 32mm (3) 42mm (3)	1 (S) 1 (S) 1 (S)	Inv	PTFE None	Coast Plastics /Protek	Sine	387,368	None
HE050	Water	NA	250	22mm (3) 32mm (3) 42mm (3)	1 (S) 1 (S) 0	Inv	GUR 4150 None	HSS/Protek	Paul	2,233,756	None
HE057	100% BCS 70mg/ml	BCS1	250	22mm (3) 28mm (3) 32mm (3)	1 (S) 1 (S) 1 (S)	Inv	PTFE None	Coast Plastics/Protek	Sine	506,314	None
HE064	30% BCS 21mg/ml	BCS4	250	22mm (2) 28mm (2) 32mm (2)	3(D) 3(D) 3(D)	Inv	GUR 4150 None	HSS/Protek	Sine	2,610,897	10% EDTA & NaH_3
HE066	50% BCS 34mg/ml	BCS4	250	22mm (2) 28mm (2) 32mm (2)	3(D) 3(D) 3(D)	Inv	GUR 4150 None	HSS/Protek	Sine	2,148,129	10% EDTA & NaH_3
HE071	90% BCS 62mg/ml	BCS 4	250	22mm (3) 28mm (3) 32mm (3)	3 (S) 3 (S) 3 (S)	Inv	GUR 4150 None	HSS/Protek	Paul	2,723,633	10% EDTA & NaH_3

Table 3.1 Experiment List Continued

Expt. No.	*Fluid /Protein*	*Fluid ID*	*Volume (ml)*	*Dia. Wear (N)*	*Soak (N)*	*Position*	*Cup Material & Sterilization*	*Material Manufacturer Cup/Head*	*Load Profile*	*Duration (cycles)*	*Serum Additives*
HE110	40% BCS 26mg/ml	BCS 5	250	22mm (3) 28mm (3) 32mm (3)	0	Inv	GUR 4150 None	HSS/Protek	Paul	2,116,442	10% EDTA & NaH_3
HE111	15% BCS 10mg/ml	BCS 5	250	22mm (2) 28mm (5) 32mm (3)	0	Inv	GUR 4150 None	HSS/Protek	Paul	2,140,139	10% EDTA & NaH_3
HE114	4% BCS 3mg/ml	BCS 5	250	22mm (3) 28mm (3) 32mm (3)	0	Inv	GUR 4150 None	HSS/Protek	Paul	1,035,051	10% EDTA & NaH_3
HE115	7.5% BCS 5mg/ml	BCS 5	250	22mm (3) 28mm (3) 32mm (3)	0	Inv	GUR 4150 None	HSS/Protek	Paul	1,645,229	10% EDTA & NaH_3
HE140	50% ACS 22mg/ml	ACS 1	450	32mm (3) 32mm (3) 32mm (3)	0	Anat/Horiz Anat/Obliq Inv	PTFE None	Coast Plastics /Biomet	Paul	400,409	5% EDTA
HE141	30% BCS 21mg/ml	BCS 6	450	32mm (3) 32mm (3) 32mm (3)	0	Anat Horiz Anat/Obliq Inv	PTFE None	Coast Plastics /Biomet	Paul	505,671	5% EDTA
HE147	15% BCS 10mg/ml 25% BCS 17mg/ml 50% BCS 34mg/ml 90% BCS 62mg/ml	BCS 6	450	32mm (3) 32mm (3) 32mm (3) 32mm (3)	0	Inv Anat/Obliq Anat/Obliq Anat/Obliq	PTFE None	Coast Plastics/ Howmedica	Paul	449,776	5% EDTA

Inverted THR Configuration on the Biaxial Simulator

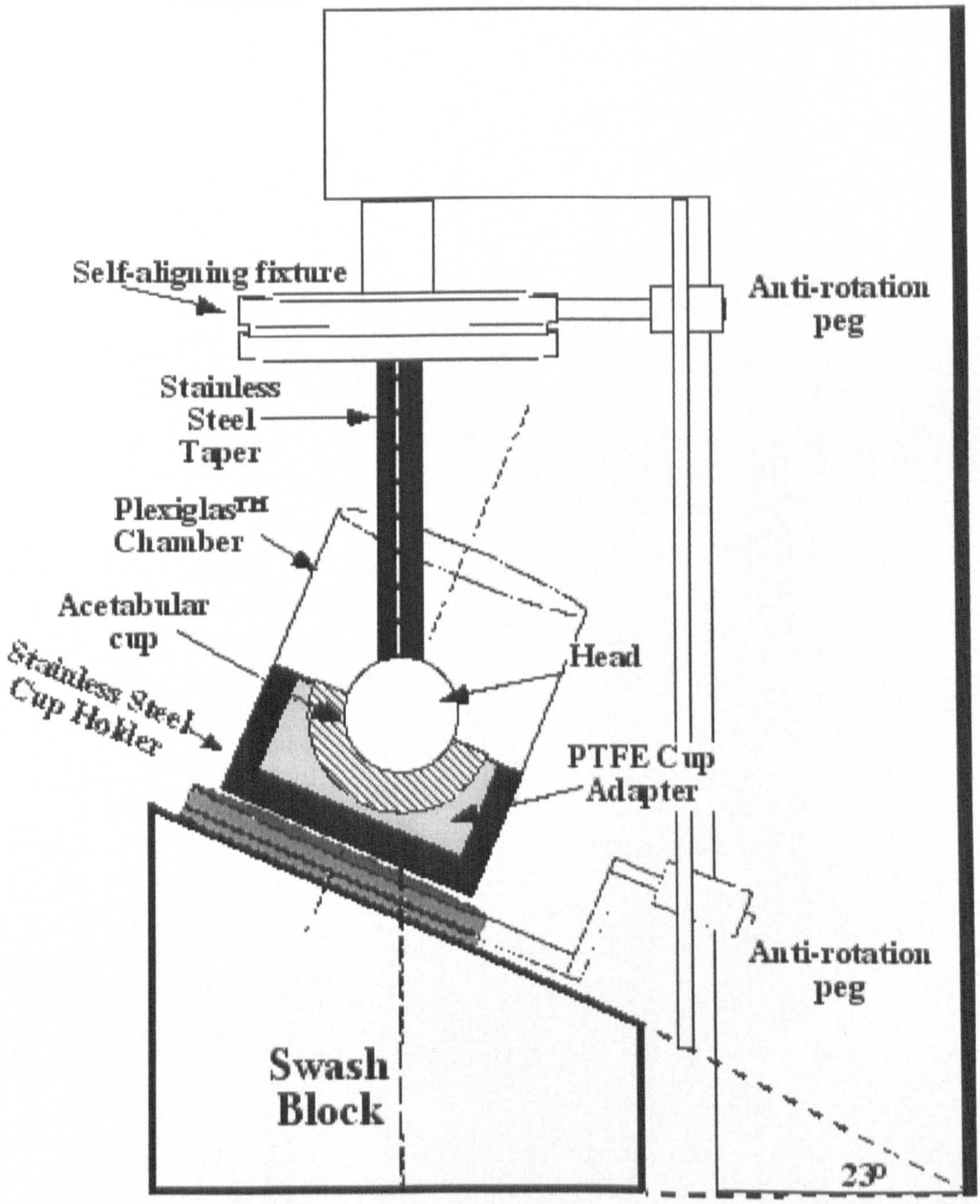


Figure 3.1 Inv Set up of Biaxial Simulator

Anatomical THR Configuration on the Biaxial Simulator

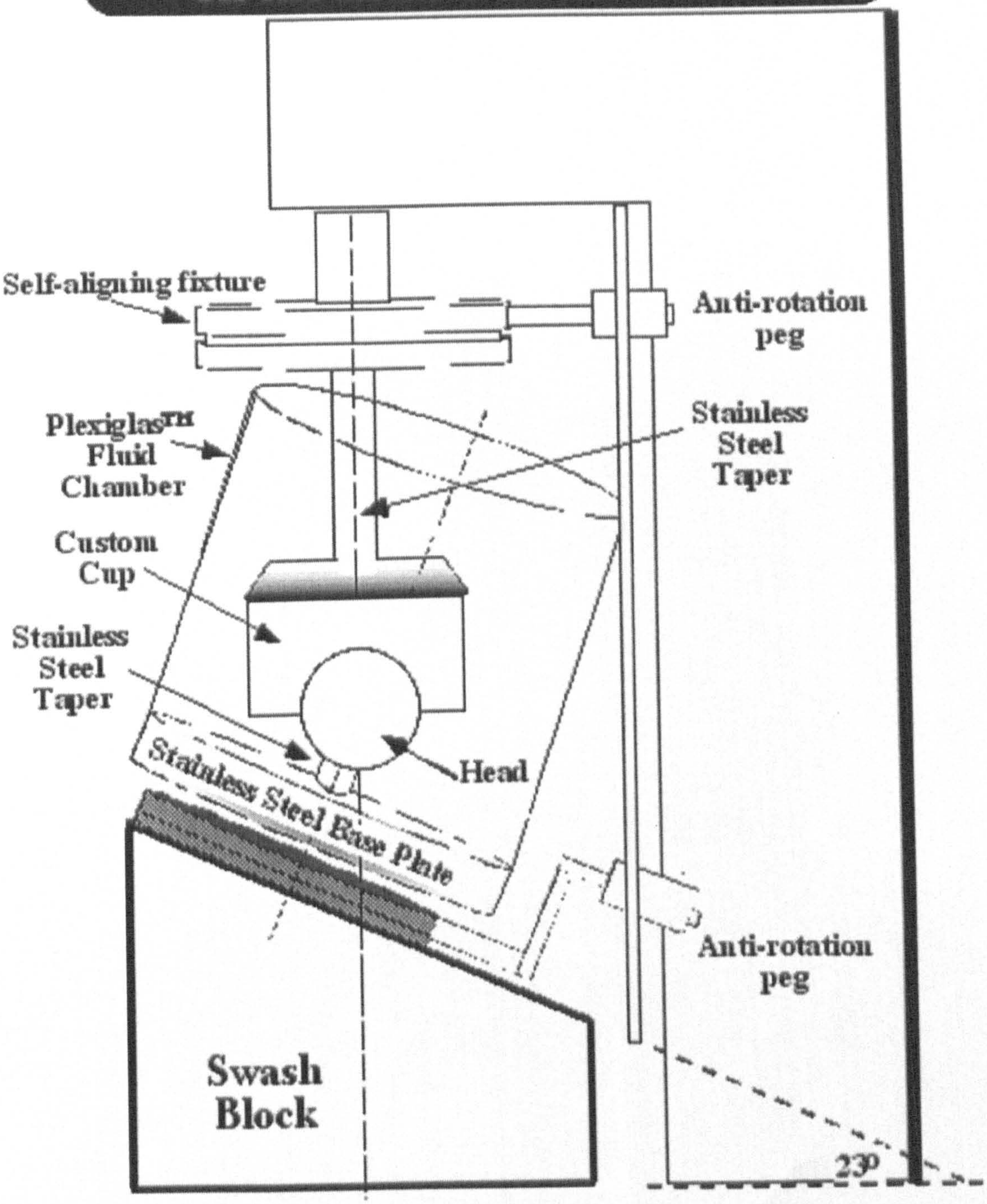


Figure 3.2 Anat Set up of Biaxial Simulator

3.2 HIP SIMULATOR LOAD AND LOAD PROFILE

In all studies the maximum load was 2000N and minimum was 200N. The load profile was either sinusoidal or physiological (Paul) and run at a frequency of 1 Hz. Historically the wear-rate of PTFE was very high and the effect of load profile was not thought to be significant to such high wear-rates, therefore initial studies of PTFE used a simplified sinusoidal load curve. However, subsequent to this work, two repeated studies from this laboratory were run to compare these two load curves and found that the wear-rate was 17% lower using a Paul load profile compared to the sine load profile. Thus the wear-rates for all PTFE studies using a sine curve were decreased by 17%.

3.3 THR CONFIGURATION

THR combinations were run either Inv (cup on bottom; fig 3.1) or Anat (cup on top; fig. 3.2). In the Anat THR configuration there were 2 options, anatomical horizontal (Anat/horiz) or anatomical oblique (Anat/obliq; fig 3.3). The Anat/horiz configuration mounted the cup face parallel to the horizontal axis and the Anat/obliq configuration mounted the cup face at an angle of 23^{o} to the horizontal axis. In all Anat configurations the fluid volume was 450ml and in the Inv configuration volumes were either 250ml or 450ml (Table 3.1). Acetabular cups were mounted in PTFE adapters, which were backed by stainless steel fixtures (fig. 3.1 and 3.2). PTFE adapters have an elastic modulus between 0.5 – 4.0 GPa, which was similar to the elastic modulus of cancellous bone of 0.1 – 3.3 GPa (Markolf, 1991), therefore, simulating the bony bed of the pelvis.

Two Types of Anatomical THR Configuration on the Biaxial Simulator

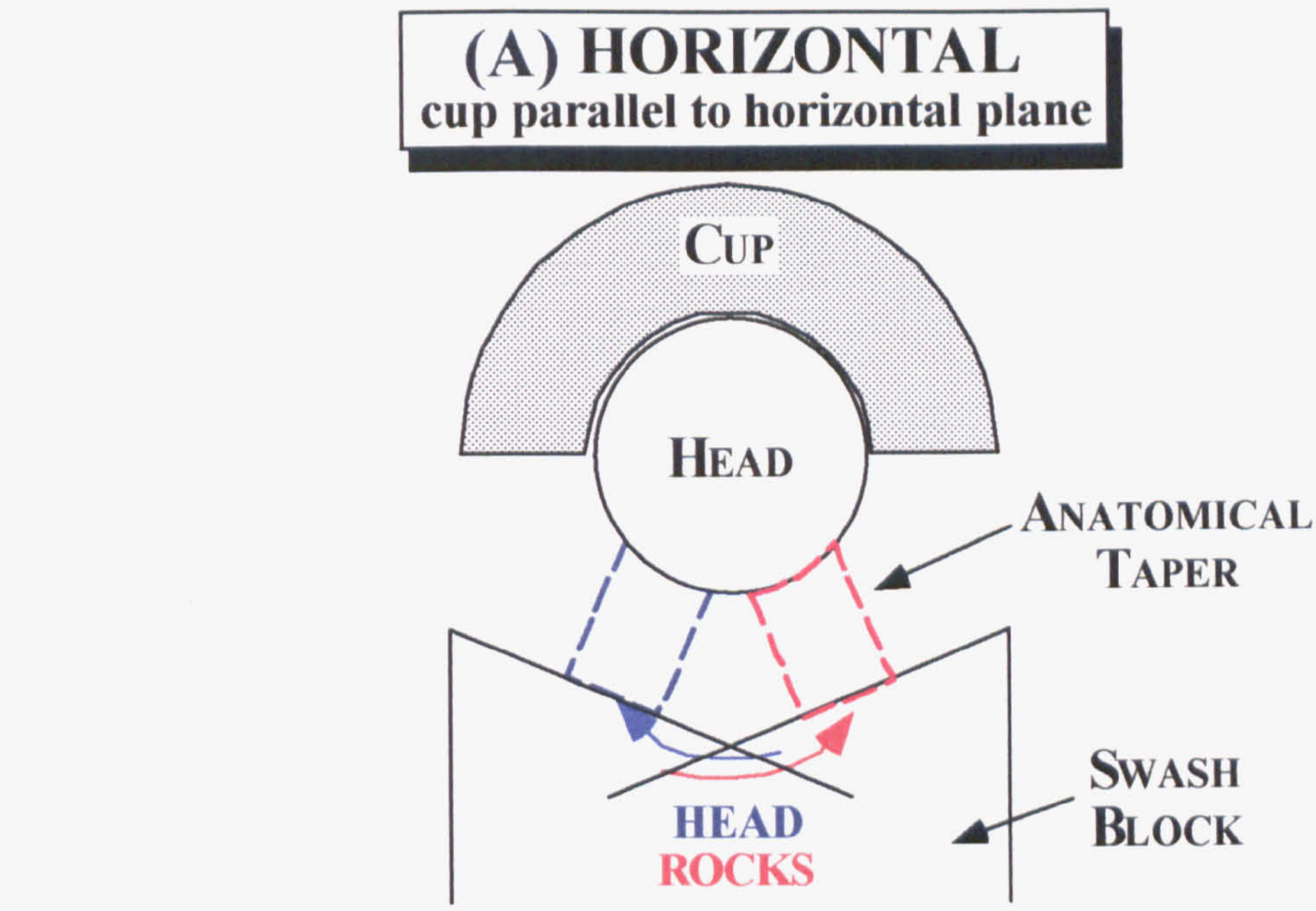


Figure 3.3 In the Anat THR configuration the cup was stationary about the vertical load axis and the head rocked at $\pm 23^{o}$ to the horizontal plane. Anat THR configurations, A) horizontal - the cup is positioned parallel to the horizontal plane and B) oblique - inclined 23^{o} to the horizontal plane.

3.4 THR MATERIAL

All PTFE (virgin extruded bar-stock ASTM D-1710, Coast Plastics, Inc.) acetabular cups were machined by Moran Innovations from 2 lots (Lot 1,all experiments before HE140; Lot 2 HE140 and HE147). The acetabular cups were not gamma irradiated so as to reproduce the same conditions used by Charnley (Waugh 1990). All tests of PTFE ran for 60,000 to 80,000 cycles per event (20 – 24 hours continuous running) with a minimum of 4 events per experiment. At the end of each event a sample of the fluid was stored for future debris analysis, all cups were cleaned and weighed and prepared for the next event. New fluid was added to the fluid chambers at the beginning of each event. Anti-microbials were not added to any of the PTFE tests due to short duration.

The UHMWPE from Hospital for Special Surgery (HSS) was extruded GUR 4150 supplied and characterized as a standard material (Bennett et al, 1996). These cups were machined from a single rod (same lot) by Moran Innovations and left non-irradiated as were Charnley's PE cups up until March 1967 (Wroblewski, 2000). Experiments with UHMWPE ran for approximately 250,000 cycles (3 days of continuous running) per event with a minimum of 9 events. Because of the longer duration and evaporation, fluid replenishment was necessary twice a day. Any test using bovine serum was replenished with like serum and tests with alpha calf serum were replenished with de-ionized water. As in the PTFE experiments, fluid samples were collected for further debris analysis and new fluid was added at the beginning of each event.

CoCr heads were used for all experiments. These heads were supplied sterile and implant ready from three manufacturers, Sulzer (Protek , Berne, Switzerland), Biomet (Warsaw, Indiana) and Howmedica (Rutherford, New Jersey).

3.5 JOINT FLUID ANALOG (JFA)

All serum was supplied and characterized by Hyclone Inc., Logan Utah (table 3.2). Deionized water was used for the water studies and dilution of bovine serum and alpha calf serum for the JFAs of various protein percentages. All serum was filtered according to the serum protocol (Appendix I). A ten percent solution of sodium azide, (NaH_3 0.05% for bacterial retardation) and 20mM ethylenediamine tetraacetic acid (EDTA to reduce calcium phosphate formation) was added to all UHMWPE tests in bovine serum. UHMWPE studies ran 24 hours for approximately 3 days and the JFA would evaporate during that time. Originally the evaporated fluid was replenished with like fluid. Subsequent to the dose response studies it was found that for bovine serum fluids this added more proteins, so after experiment HE115 fluid evaporation was replenished with de-ionized water. Fluid temperatures were monitored but not controlled. Room temperature was maintained at 70°F $\pm 5^{o}$ and after experiment HE124 humidity was maintained between 45 and 50%.

3.6 WEAR MEASUREMENTS

Wear was determined by the weight-loss method using a Sartorius Microbalance (Watson Bros., Model RC #167). Wear and soak specimens and a 32mm CoCr control femoral head were cleaned and dehydrated together using the standard cleaning procedure (Appendix II). All specimens and the CoCr femoral head control were weighed four times in sequence. A second control (1cm disk of

stainless steel), which did not go through the cleaning process, was added to these weight measurements after experiment HE115.

Wear-rates were based on 1 million cycles equal to one year of walking. This number was recently verified by Schmalzried et al (1998) using a pedometer. Their studies showed a wide variance (up to 45 fold) in walking activity. This variance was due to the high (former Olympic gymnast) and low (patient taking less than 1.1 steps a day) however; the average for all patients was just under the one million cycles per year.

Linear regression analysis was used to determine the wear-rate (weight loss per million cycles). Volumetric wear-rates were determined using specific density (PTFE = 2.16 mg/mm^3, HSS = 0.94 mg/mm^3) as certified by the supplier. The soak-rate due to fluid-sorption (soak specimens) was subtracted from the specimen wear-rate to determine net wear-rate. The precision among the sets of wear cups for each variable was determined using 95% confidence limits. Precision was reported as a percentage of the experimental scatter, the equation was:

$$\text{Percent Precision} = 100*(W_{cl} - W)/W$$

where W_{cl}= wear-rate of upper 95% confidence level, and W = average wear-rate. The Volumetric Wear Index (VWI) was used to relate the increase in wear-rate of any size femoral head relative to the Charnley 22.25 mm size (Clarke et al, 1997). To find the VWI ratio for femoral head size "A", the equation was:

$$\text{VWI (\% per millimeter)} = \frac{100 *(V_a - V_c)/V_c}{(D_a - D_c)}$$

Where V_a = Volumetric wear-rate of femoral head "A", V_c = Volumetric wear-rate of reference head (Charnley 22.25 mm), D_a = Diameter of femoral head "A" and D_c = Diameter of reference head (Charnley 22.25 mm)

3.7 PROCEDURE FOR UHMWPE AND PTFE: JFA WATER VERSUS BOVINE SERUM

The 4 studies (HE0040, HE050, HE057 and HE071) were run Inv with 250ml chambers on a 9 channel biaxial simulator (table 3.1). All studies used CoCr heads supplied implant ready with surface finishes as designated by the manufacturer (Protasul, Protek Inc., Berne, Switzerland). De-ionized water was used in the water studies. The bovine serum was made according to serum protocol (Appendix I). PTFE studies did not use additives, however UHMWPE studies ran for 3 days and therefore a 10% solution of EDTA (eliminate calcium phosphate layer) and NaH_3 (antibacterial) was added to the bovine serum. Assembly of the cups and heads was according to Assembly Protocol A (Appendix III). The nine THR components were mounted on the simulator using Simulator Protocol 9 and 12B Channel (Appendix IV; 9 channel). At the end of the event specimens were removed for cleaning (Appendix II) and weighing (Appendix V). The JFA from each size of specimen (e.g. all 22mm) was stirred and poured into beakers and a 40ml sample from each beaker was collected for debris analysis. All of the above procedures applied to all four experiments with the only differences due to duration (table 3.3)

3.8 PROCEDURE FOR UHMWPE AND PTFE: JFA BOVINE SERUM DOSE RESPONSE

Studies followed the same procedure as 3.7 with bovine serum dilutions as listed in table 3.4. All tests were run on either the 9 or 12B channel biaxial simulator.

Table 3.2 Bovine Serum Constituents

Components listed from Hyclone, Inc. (biochemical assay) BCS = Bovine calf serum, ACS = Alpha calf serum

Component	*Alpha Calf Serum Lot #AHJ8879 ACS 1*	*Bovine Serum Lot AFD5137 BCS 1*	*Bovine Serum Lot 21112355 BCS 2*	*Bovine Serum Lot 21112521 BCS 3*	*Bovine Serum Lot AFK5740 BCS 4*	*Bovine Serum Lot AGK7210 BCS 5*	*Bovine Serum Lot AHL9372 BCS 6*
Protein (mg/ml)	45.0	70.0	63.0	70.0	69.0	65.0	69.0
Albumin (mg/ml)	23.0	41.0	36.0	41.0	41.0	31.0	32.0
Globulin (mg/ml)	22.0	29.0	27.0	29.0	28.0	34.0	37.00
pH	7.66	7.50	7.71	7.51	7.71	7.56	7.51
Iron (µg/dl)	45.00	21.00	58.00	47.00	45.00	43.00	29.00
Calcium (mg/ml)	0.006	0.118	0.103	0.11	0.116	0.111	0.111
Sodium (meq/L)	156.00	145.00	141.00	144.00	143.00	143.00	144.00
Potassium (meq/L)	>14.00	6.10	5.50	5.80	5.80	5.80	6.00
Chloride (meq/L)	136.00	101.00	100.00	100.00	101.00	100.00	102.00
Glucose (mg/ml)	0.23	1.36	1.18	1.17	1.11	0.91	0.97

Table 3.3 PTFE versus UHMWPE Event Differences

Parameter	*PTFE*	*UHMWPE*
No. of cycles per event	~80,000 cycles (24 hours)	~250,000 cycles (72 hours)
Bovine Serum	100% No additives	90% serum, 10% additives EDTA and sodium azide (to eliminate calcium phosphate and bacteria)
Replenishment due to evaporation	None	Twice daily with new 'like' fluid

Table 3.4 Serum Dilution

BCS = Bovine calf serum, DI = De-ionized water

Experiment #	*PTFE* *JFA per 1000ml*	*UHMWPE* *JFA per 1000ml*
HE040, HE050	DI water	DI water
HE047, HE114	40ml BCS 2 960ml DI water	40ml BCS 5 100ml Additives 860ml DI water
HE044, HE115	75ml BCS 3 925ml DI water	75ml BCS 5 100ml Additives 825ml DI water
HE043, HE111	150ml BCS 3 850ml DI water	150ml BCS 5 100ml Additives 750ml DI water
HE041, HE064	300ml BCS 3 700ml DI water	300ml BCS 5 100ml Additives 600ml DI water
HE110	-	400ml BCS 5 100ml Additives 500ml DI water
HE066	-	20ml BCS 4 100ml Additives 400ml DI water
HE057, HE071	1000ml BCS 1	900ml BCS 4 100ml Additives

3.9 PROCEDURE FOR PTFE: JFA VOLUME AND THR CONFIGURATION

The PTFE studies (HE140, HE141 and HE147) were run on the 12B simulator in 450ml of JFA. HE140 and HE141 THR configurations were identical, 3 THR each in the Inv, Anat/horiz and Anat/obliq configurations and assembled according to Assembly Protocol B and C (Appendix III). Experiment HE147 used the Anat/obliq configuration and was assembled according to Assembly Protocol C (Appendix III). All studies used CoCr heads supplied implant ready with surface finished as designated by the manufacturer (Howmedica/Osteonics, Rutherford, NJ).

The bovine and alpha calf serum were diluted with de-ionized water according to Table 3.5 with no additives and made according to the serum protocol (Appendix I). The twelve THR components were mounted on the simulator and the study was begun using Simulator Protocol 9 and 12B Channel (Appendix IV). At the end of the event specimens were removed for cleaning (Appendix II) and weighing (Appendix V). The JFA from each size of specimen (e. g. all 22mm) was combined and a 40ml sample was collected for debris analysis.

Table 3.5 Constituents of JFA for PTFE/CoCr Large Fluid Volume Studies

ACS = Alpha calf serum, BCS = Bovine calf serum, DI = De-ionized water

Experiment #	*PTFE JFA per 1000ml*
HE140	500ml ACS 1 500ml DI Water
HE141	300ml BCS 6 700ml DI Water
HE147	150ml BCS 6, 850ml DI Water 250ml BCS 6, 750ml DI Water 500ml BCS 6, 500ml DI Water 900ml BCS 6, 100ml DI Water

3.10 PROCEDURE FOR DEBRIS ANALYSIS

After each event all JFA for each head size was mixed and 40ml samples were collected for both PTFE and UHMWPE experiments. The fluid was stored in a freezer until debris analysis was begun. PTFE debris was processed according to Appendix V and UHMWPE debris was processed according to Appendix VI. Samples were examined using a scanning electron microscope (Philips XL30 FEG) after processing. Images were processed with commercial software (ProImage).

4.1 PTFE WEAR (THR INV): 250ml JFA OF WATER AND BOVINE SERUM WITH PROTEIN CONCENTRATION > 63 mg/ml

In water the data for PTFE appeared inconsistent as the test progressed (fig 4.1). In order to evaluate the data consistently the test was separated into phases. Wear-rate of the 22mm head size dropped sharply and then leveled off during phase A for an average wear-rate of 333 mm^3/Mc. Wear-rate decreased during phase B to 246 mm^3/Mc followed by an increase during phase C of 338 mm^3/Mc. Wear-rate of the 32mm femoral head size was slightly more consistent with wear-rates of 206 mm^3/Mc and 234 mm^3/Mc for phase A and B respectively, then abruptly doubled for phase C to 488 mm^3/Mc. The wear-rate for the 42mm size increased continuously throughout the test. The rates increased from 290 mm^3/Mc in phase A to 577 mm^3/Mc in phase B and then doubling as did the 32mm to 1097 mm^3/Mc in phase C. The precision varied from phase to phase with a range of $\pm$26 to $\pm$72% (table 4.1).

The weight changes in the PTFE fluid-sorption control cups were less than 1% of the weight changes in the wear cups. Averaging and correcting for fluid-sorption of the 3 phases for each size resulted in wear-rates of 307, 311 and 659 mm^3/Mc for 22, 32, and 42mm respectively. The equal wear-rate of the 22mm and 32mm was inconsistent with clinical data showing increased wear-rate with increased head diameter (Callaghan et al, 1995, Charnley 1969, Kabo et al, 1993, Livermore et al, 1990). Wear-rates were 5 times lower than clinical. Debris was only visible after 3,000 cycles and appeared from the edge of the cup in ribbons sometimes 2 to 3cm long before breaking off and floating to the surface. The debris size ranged from 5 μm to 800 μm in length and 0.5 μm to 50 μm in width. When

viewed with the SEM, the debris exhibited long rod shapes along with flakes with smooth, leaf-like surfaces (fig. 4.2 A and B).

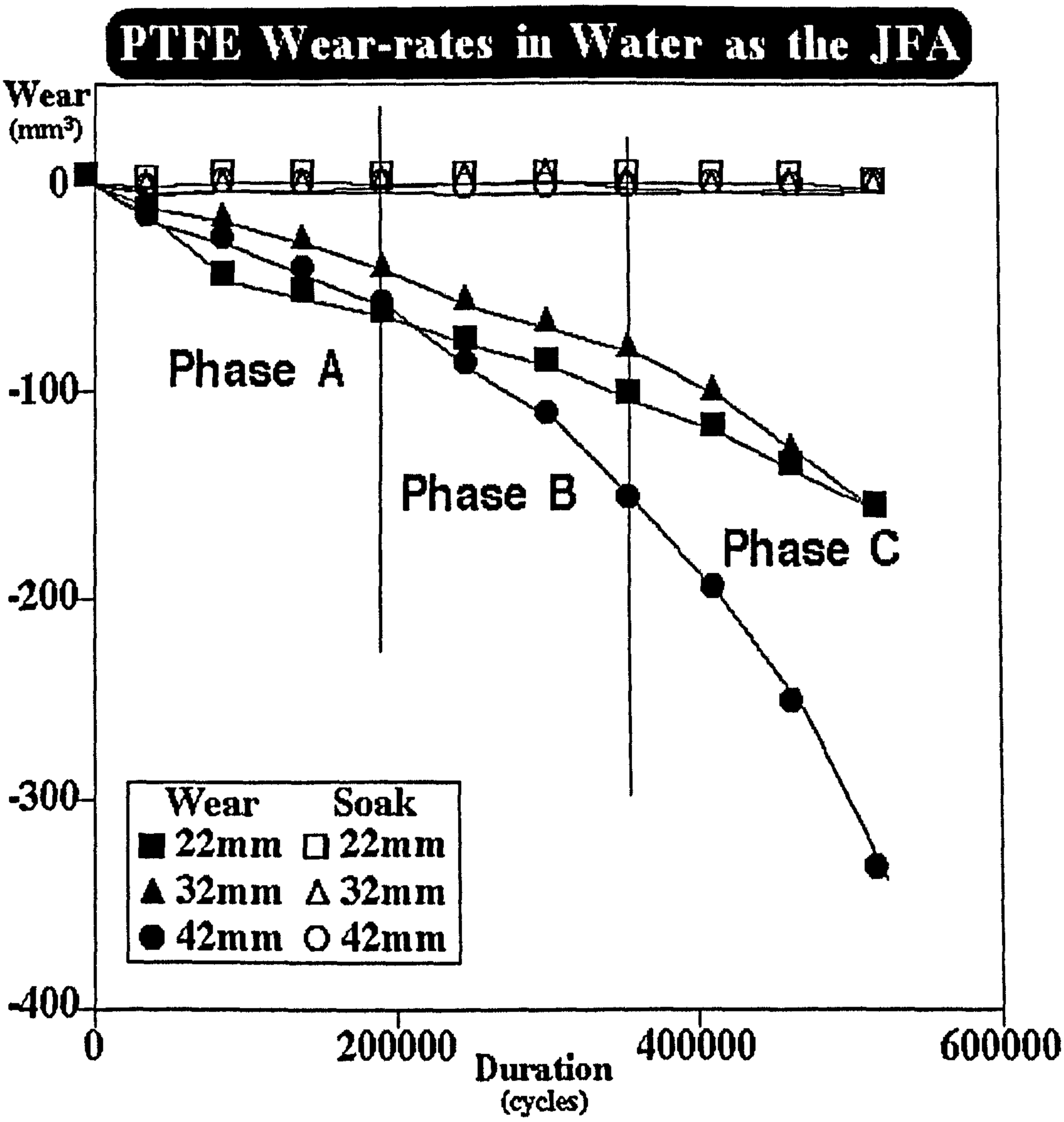


Figure 4.1. PTFE/CoCr in water with volumetric wear-rates broken down into 3 phases for clarity

Table 4.1 Wear-rates of PTFE/CoCr in water and 100% serum.

VWI = Volumetric Wear Index, S = static soak (no load in a container of de-ionized water)

Expt. No.	*Fluid*	*Dia (mm)*	*Wear-rate (mm^3/Mc)*	*Soak (mm^3/Mc)*	*Net Wear-rate (mm^3/Mc)*	*Precision & (R^2)*	*VWI*
HE040 Overall	Water	22	305	2 (S)	307	72% (0.984)	Reference
		32	309	2 (S)	311	58% (0.964)	0.1%
		42	655	4 (S)	659	59% (0.924)	6%
Phase A HE040	Water	22	333			26% (0.92)	Reference
		32	206			26% (0.990)	-4%
		42	290			41%(0.998)	-1%
Phase B HE040	Water	22	246			72% (0.996)	Reference
		32	234			52% (0.974)	-0.5%
		42	577			38% (0.986)	7%
Phase C HE040	Water	22	338			33% (0.994)	Reference
		32	488			58% (0.993)	4%
		42	1097			59% (0.978)	11%
HE057	100% Bovine Serum 70mg/ml protein	22	3417	1 (S)	3418	4% (0.992)	Reference
		28	4967	0 (S)	4967	2% (0.998)	8%
		32	6830	1 (S)	6831	2% (0.998)	10%

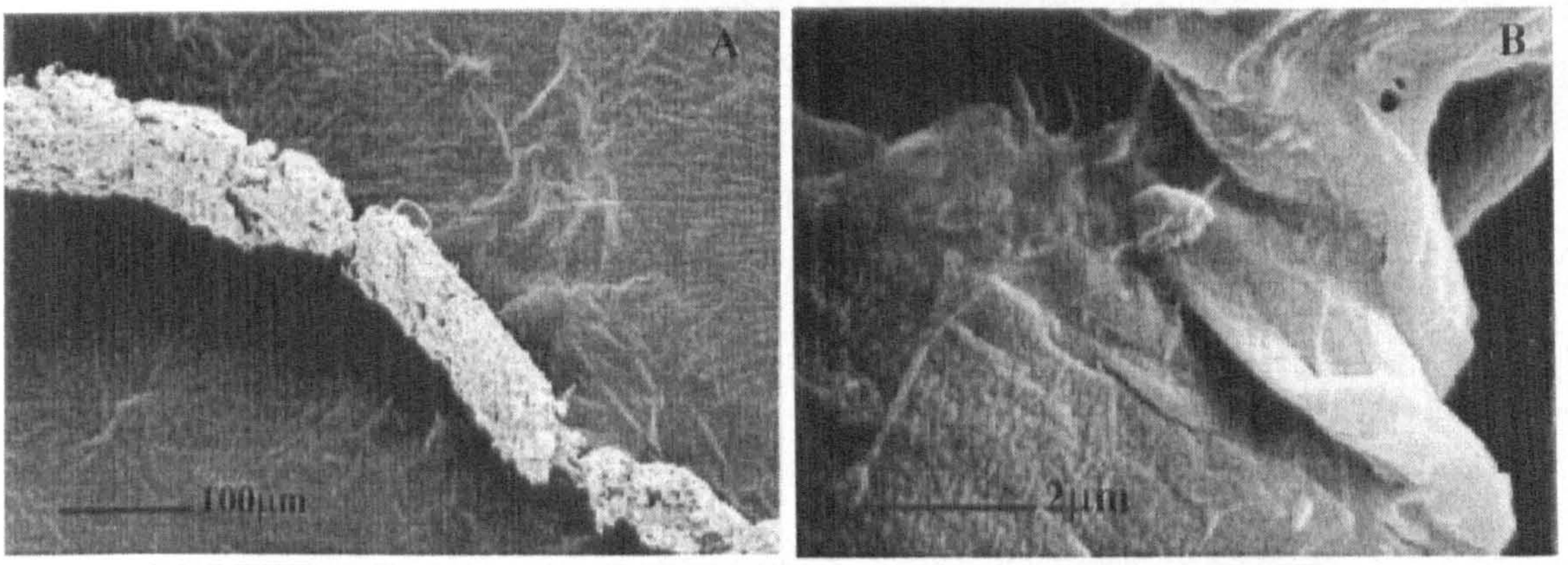


Figure 4.2 PTFE debris from PTFE/CoCr study using water as the JFA (experiment HE040) A) debris in the shape of a long rod 156x magnification B) smooth leaf like flake of debris 10,255x magnification

The PTFE in 100% serum (70mg/ml of protein) wear data were predictably linear throughout the study (fig 4.3). The weight changes in the PTFE fluid-sorption control cups were less than 1% of the weight changes in the wear cups and wear-rate ranged from 3418 mm^3/Mc for the 22mm to 6831 mm^3/Mc for the 32mm (table 4.1). Precision among the three specimens was better than 4% and wear-rate of the 28 and 32mm cups increased at a rate of 8% and 10% per millimeter with respect to the 22mm cup. Wear-rates were 2 times higher than clinical. PTFE wear debris

appeared after only a few hundred cycles. Evidence of debris was easily visible to the naked eye and created a 'snow storm' effect with much of it sinking to the bottom of the specimen container. The debris consisted mainly of elongated rods with some as long 1mm and many irregular particle conglomerations formed from these rods. The round particles averaged 25 μm in diameter (fig.4.4 A and B).

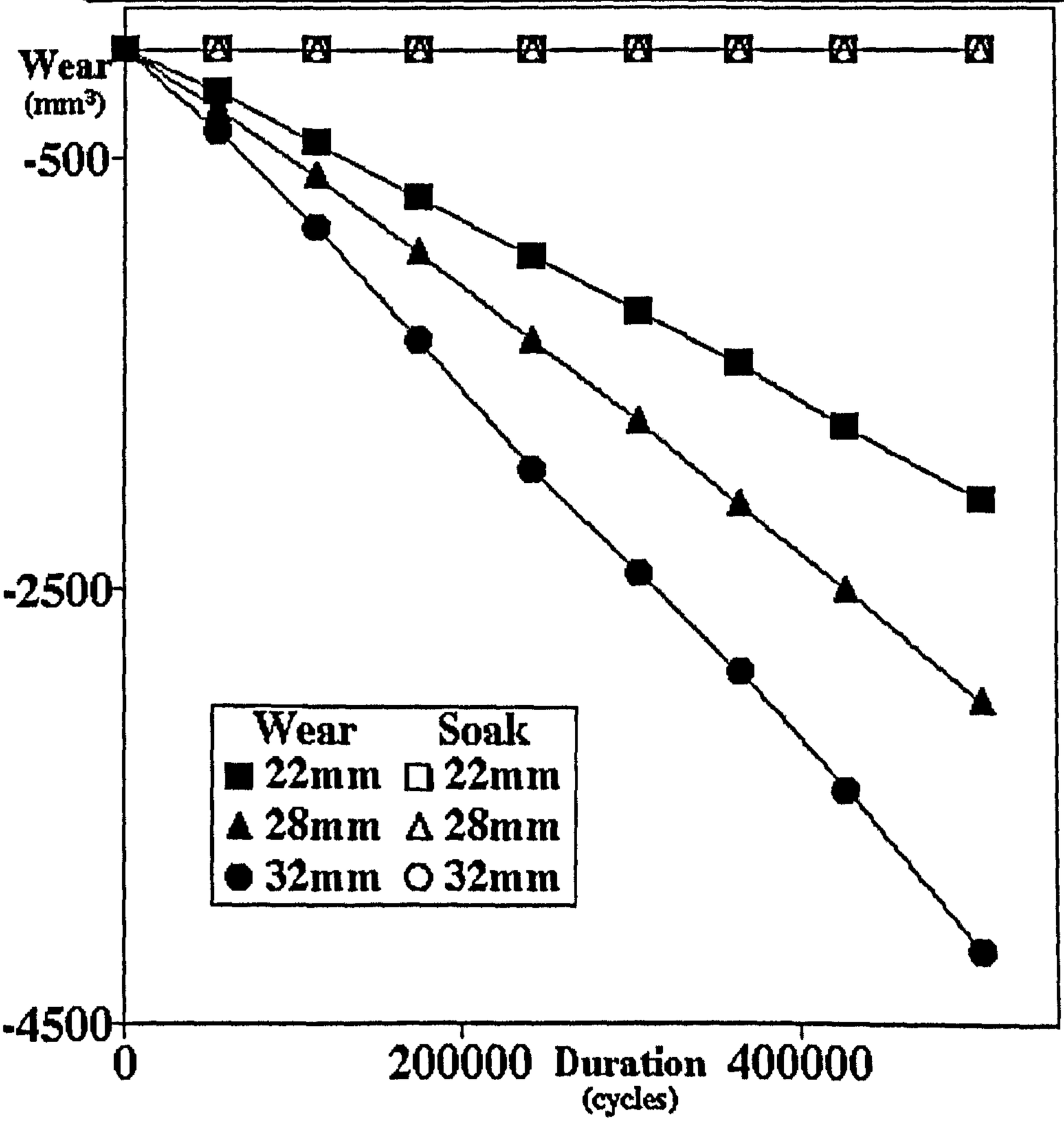
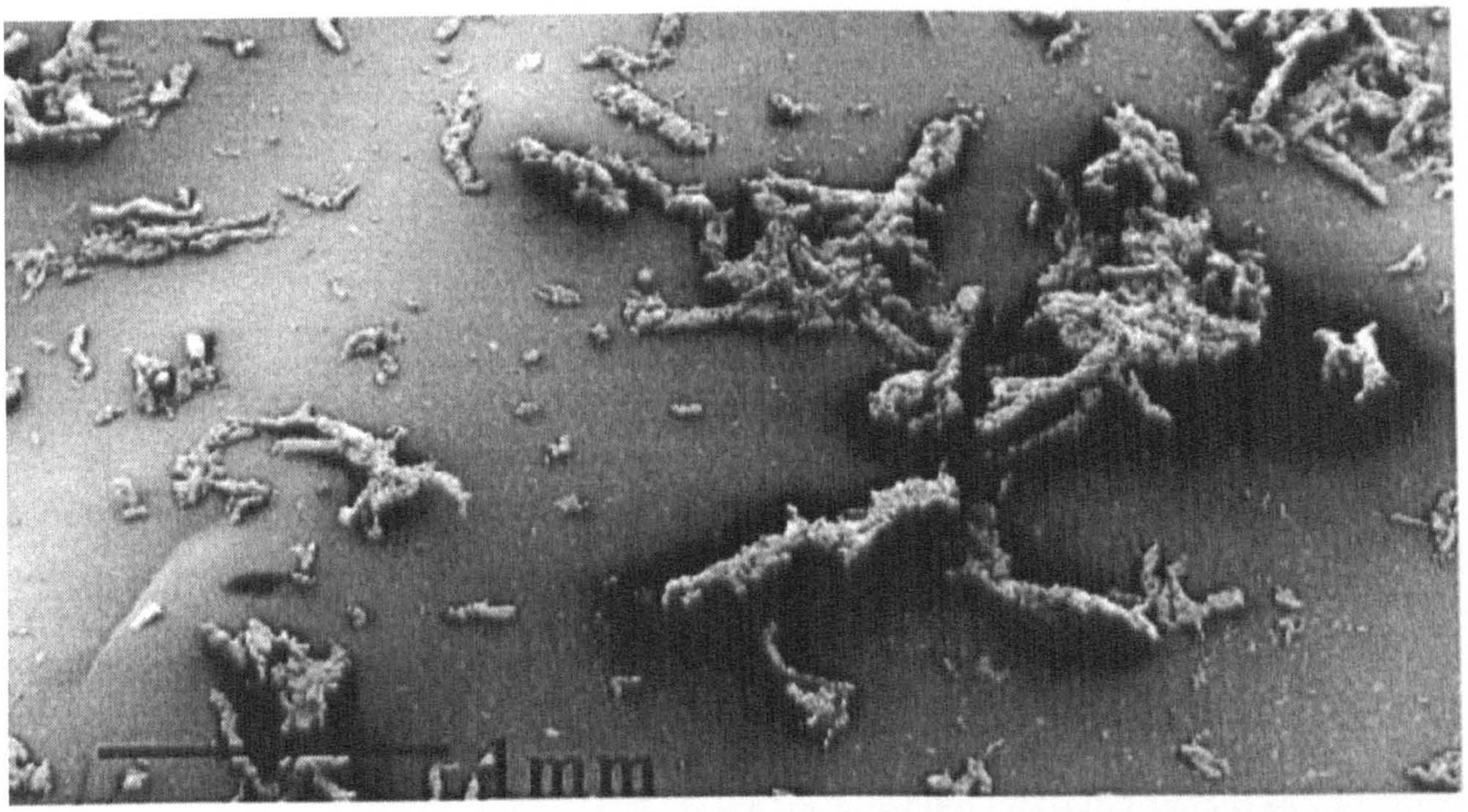


Figure 4.3 Volumetric wear-rates of PTFE/CoCr in 100% serum with 70mg/ml of protein



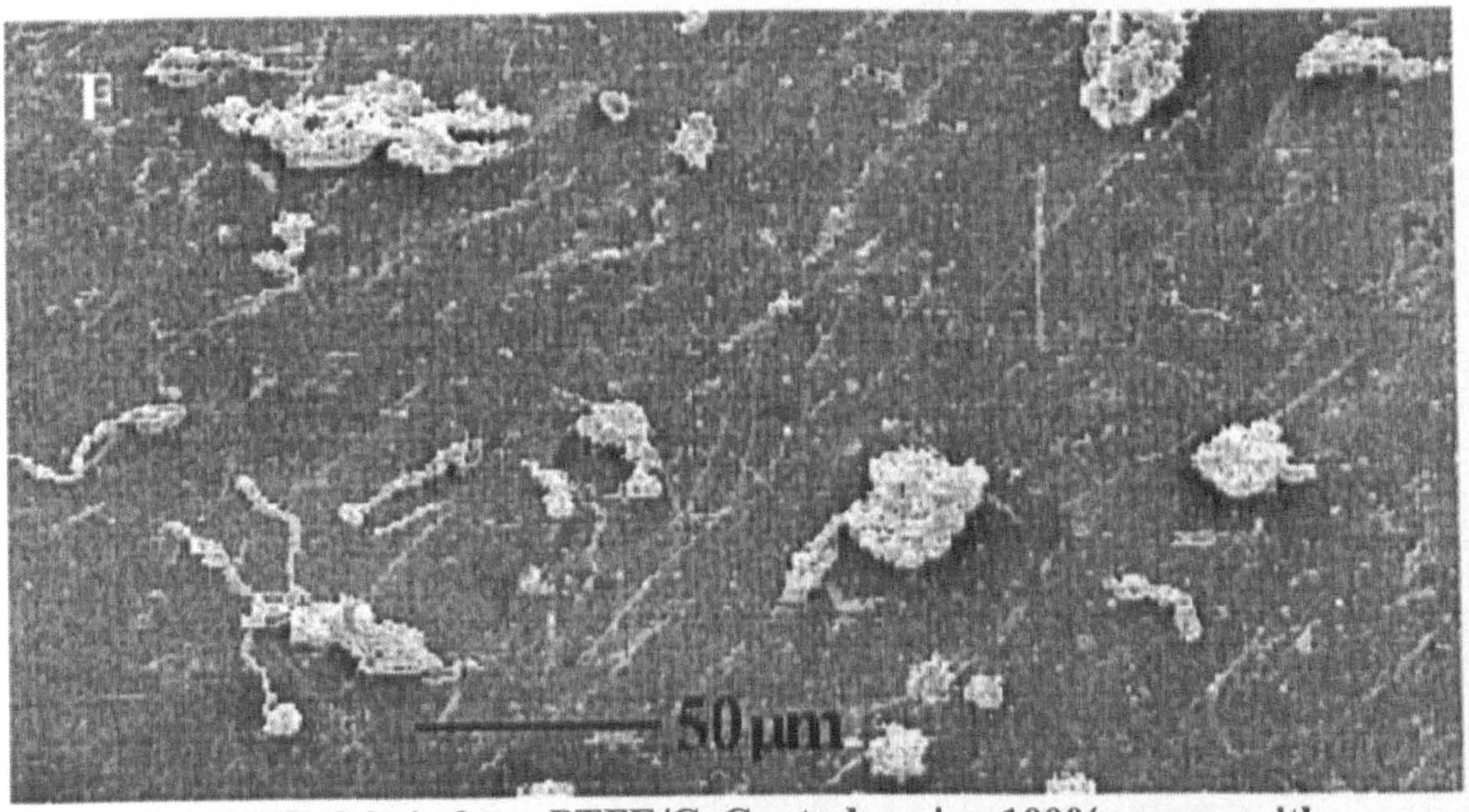


Figure 4.4 PTFE debris from PTFE/CoCr study using 100% serum with 70mg/ml protein as JFA showing the stringy appearance noted by Charnley. A) Long rod shapes, which can clump to form irregular shapes with sizes ranging from 50 µm to over 1mm. B) Higher magnification showing smaller particle conglomerations and rods.

4.2 UHMWPE WEAR (THR INV): 250ml JFA OF WATER AND BOVINE SERUM WITH PROTEIN CONCENTRATION > 60 mg/ml

The wear-rate of non-sterilized GUR 4150 UHMWPE in water was very low, lacked precision (scatter) and soak-rates were 18 to 50% of wear-rates (fig 4.5). Net wear-rates averaged 1.9, 1.7 and 2.8 mm^3/Mc for 22, 32 and 42mm respectively and precision was poor from 49 to 102% (table 4.2). As in the PTFE in water study, the wear-rates of the 22 and 32mm cups were equal and did not show the characteristic increase in wear-rate as diameter increased. Debris was also similar to PTFE in water with ribbons forming at the edge of the cup and then detaching and floating to the top. When examined in the SEM the debris looked shredded without evidence of rounded particles and fibrils. Debris size averaged 100μm long by 25μm wide (fig 4.6 A and B).

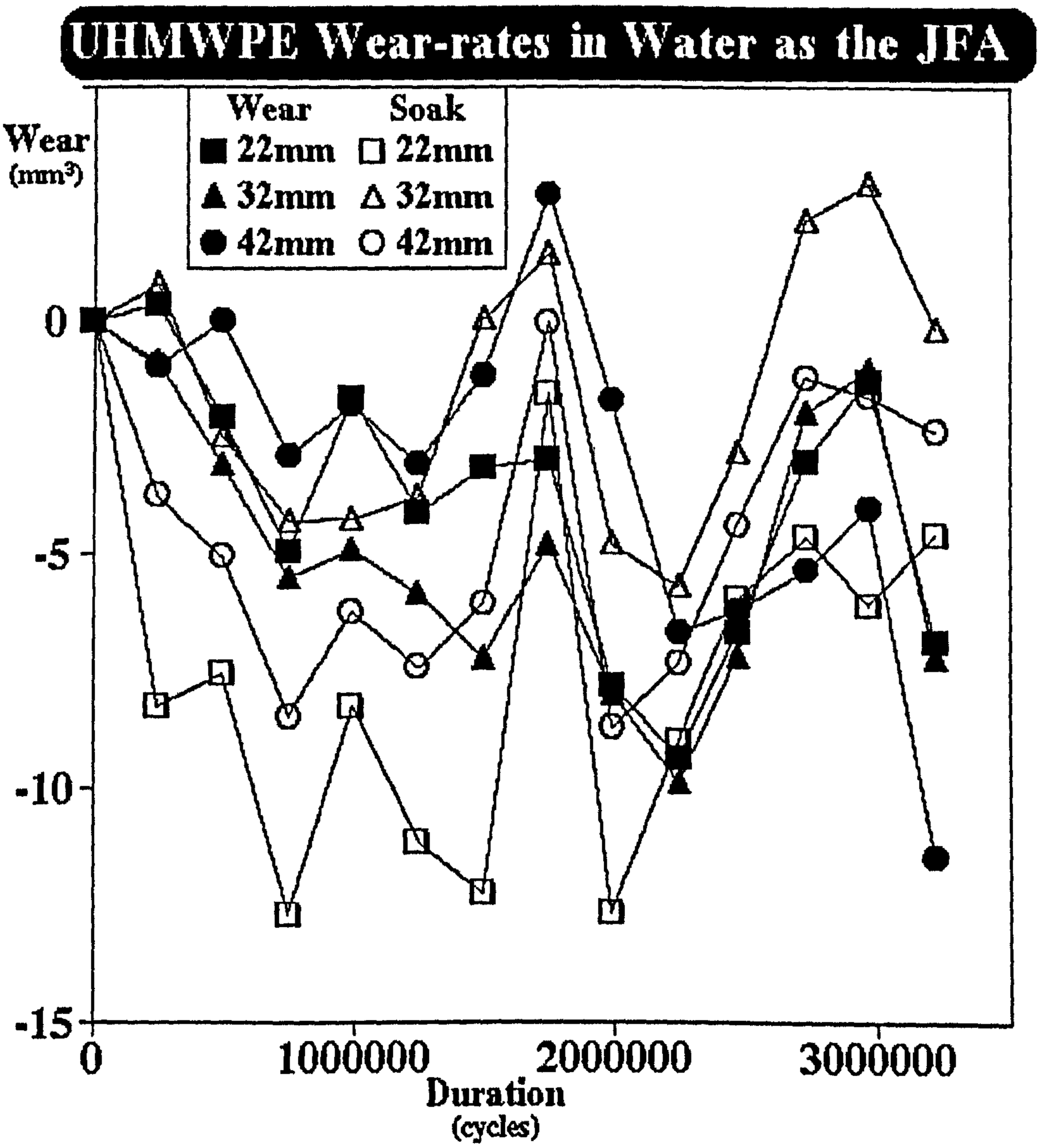
Table 4.2 Volumetric Wear-rates of UHMWPE/CoCr in Water and 90% Serum.

VWI = Volumetric Wear Index. S = static soak (no load in a container of de-ionized water)

Expt. No.	*Fluid*	*Dia (mm)*	*Wear-rate (mm^3/Mc)*	*Soak (mm^3/Mc)*	*Net Wear-rate (mm^3/Mc)*	*Precision & (R^2)*	*VWI*
HE050	Water	22	1.6	0.4 (S)	2.0	61% (0.309)	Reference
		32	1.1	0.6 (S)	1.7	102% (0.490)	-2%
		42	2.4		2.4	49% (0.151)	1%
HE071	90% Bovine Serum 62mg/ml protein	22	28.0	3.0 (S)	31.0	9% (0.985)	Reference
		32	30.2	4.0 (S)	34.2	8% (0.990)	1%
		42	31.3	4.9 (S)	36.2	12% (0.992)	1%

In the 90% bovine serum with 62mg/ml of protein the non-sterilized GUR 4150 showed a trend for linear wear-rates (fig 4.7). Wear-rates were at least 14 times higher than in water at 28, 34.2 and 36.2 mm^3/Mc for 22, 32 and 42mm respectively and soak-rates were 15% of the wear-rates (Table 4.2). Precision was better than 12% but the wear-rate penalty (VWI) was low at 1% for both the 32 and 42mm diameters. The debris from UHMWPE was not readily visible as in the PTFE

and what was produced sank to the bottom of the chamber. There was little evidence of fibrils and greater than 95% of the particles were round averaging 0.7μm in diameter (fig 4.8). Wear-rates of UHMWPE in both water and 62mg/ml of protein in bovine serum were lower than the clinical average by 30 and 2 times respectively



(table 2.2).

Figure 4.5 Volumetric wear-rates of non-sterilized GUR4150 UHMWPE/CoCr in water.

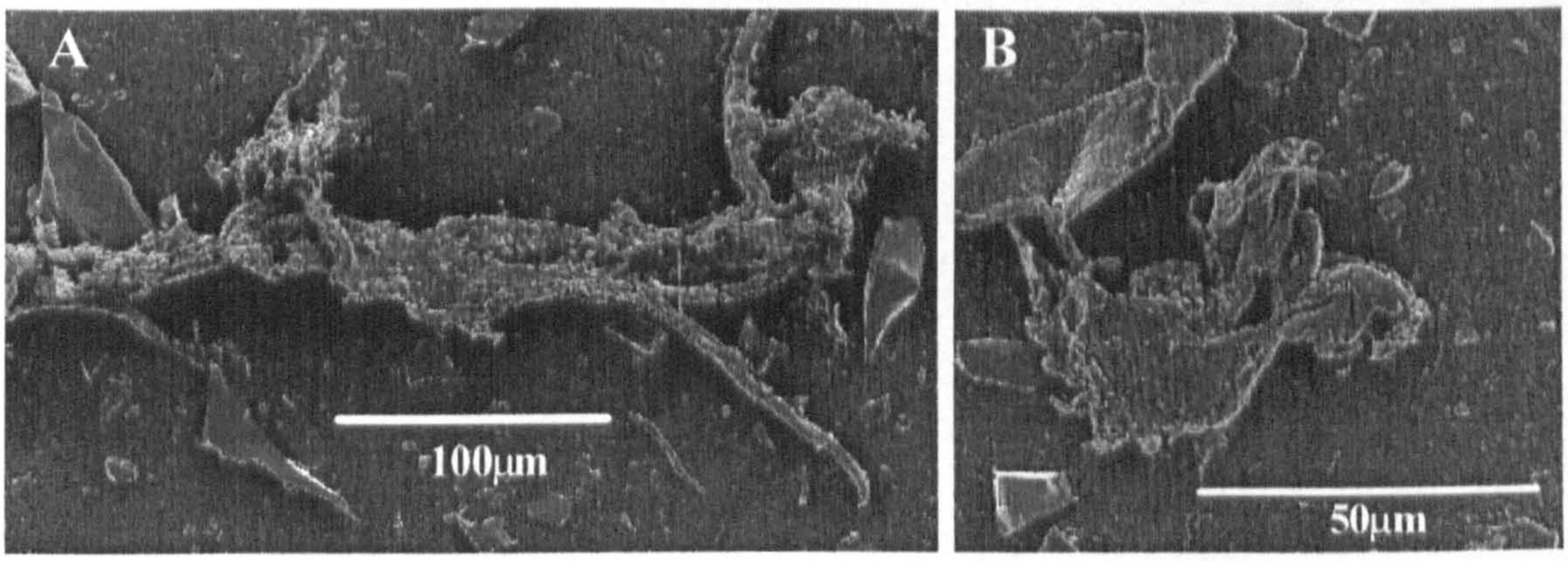


Figure 4.6 UHMWPE debris from water. A) Long shred B) rounded shred

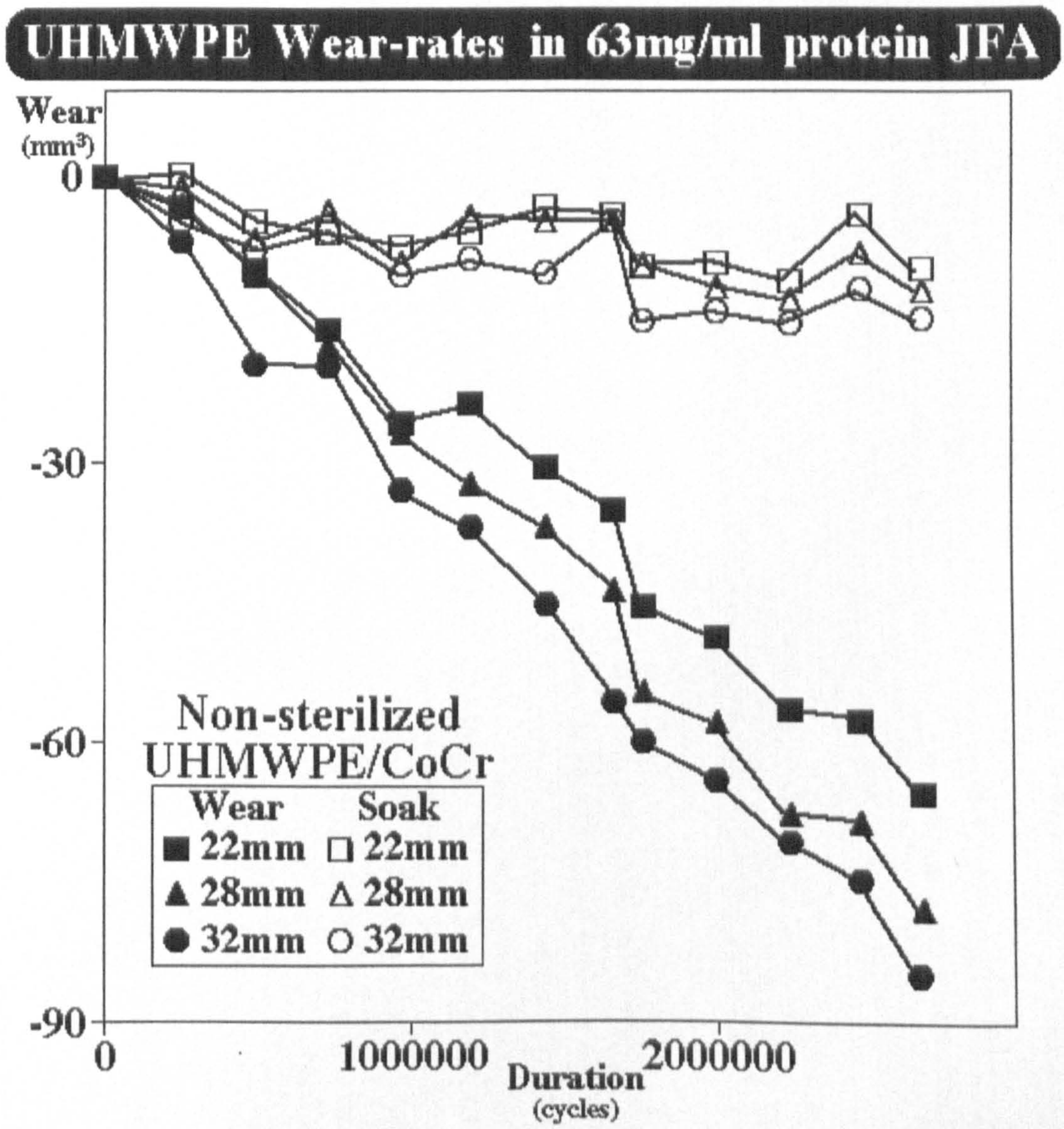


Figure 4.7 Volumetric wear-rates of non-sterilized GUR 4150 UHMWPE/CoCr in 90% serum (62mg/ml protein).

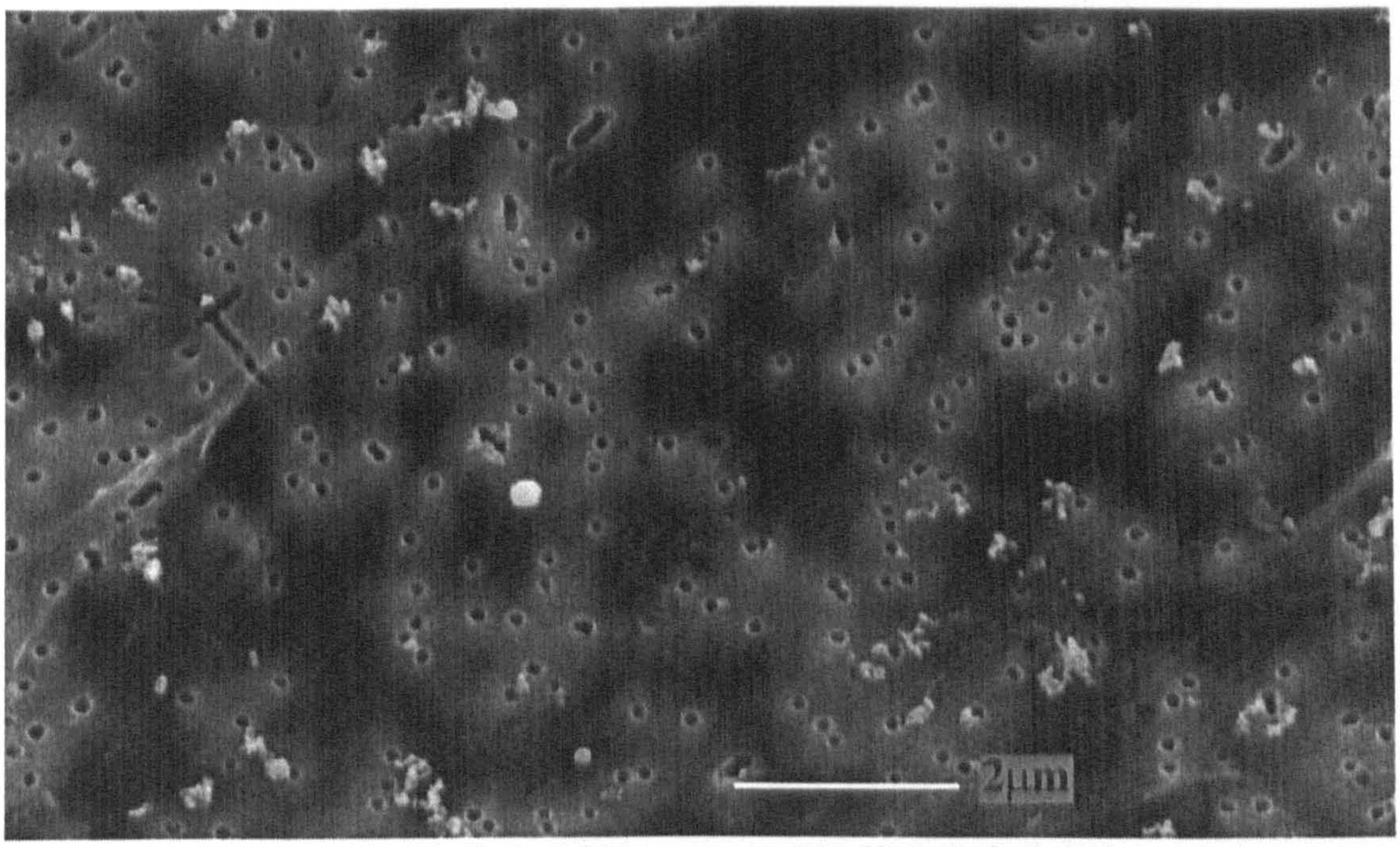


Figure 4.8 UHMWPE debris in 90% serum with 62mg/ml protein concentration

4.3 PTFE WEAR (THR INV): JFA PROTEIN CONCENTRATION DOSE RESPONSE

Bovine serum was diluted with de-ionized water to produce the 4 protein concentrations (3, 5, 10 and 21mg/ml). PTFE wear-rates in all 4 protein concentrations were linear and precision was better than 4% for the sets of three in each size (fig 4.9 - 4.12, table 4.3). Soak-rates continued to be less than 1% of the wear-rates. With the addition of the water and 100% bovine serum (70mg/ml protein) results it could be seen that there was a clear trend for increased wear-rate with increased protein concentration and the ball-size effect of increased wear-rate with increased head size was maintained (fig 4.13, table 4.3). From 3 mg/ml protein concentration to 70mg/ml protein concentration wear-rates increased by 3.8 and 4.5 times for the 22 and 32mm diameters respectively. Debris from 21mg/ml and 10mg/ml of protein was examined in the SEM. The PTFE debris in 21mg/ml

continued to show large conglomerations of particles up to 1mm in length and the average particles were short rods approximately 100μm long by 30μm wide and rounded conglomerates averaging 50μm diameter (fig 4.14). The debris morphology in 10mg/ml was similar to the debris morphology in 21mg/ml but averaging 75μm long by 25 to 30μm wide (fig 4.15).

Table 4.3 Volumetric Wear-rates of PTFE/CoCr using various Bovine Serum JFA concentrations

PTFE/CoCr in bovine serum diluted with de-ionized water to protein concentrations of 21, 10, 5 and 2mg/ml. VWI = Volumetric Wear Index.

Expt. No.	*Protein Concentration*	*Dia (mm)*	*Wear-rate (mm^3/Mc)*	*Precision & (R^2)*	*VWI*
HE040	0	22	307	72% (0.984)	Reference
		32	311	58% (0.964)	0.1%
		42	659	59% (0.924)	6%
HE041	21mg/ml	22	1669	3% (0.999)	Reference
		32	3363	2% (0.999)	10%
		42	4460	3% (0.999)	8%
HE043	10mg/ml	22	1381	3% (0.999)	Reference
		32	2558	4% (0.999)	9%
		42	3811	2% (0.999)	9%
HE044	5mg/ml	22	1173	3% (0.999)	Reference
		32	2146	3% (0.999)	8%
		42	3010	4% (0.999)	8%
HE047	3mg/ml	22	877	4% (0.998)	Reference
		32	1167	4% (0.998)	3%
		42	1502	4%(0.999)	4%
HE057	70mg/ml	22	3418	4% (0.992)	Reference
		28	4967	2% (0.998)	8%
		32	6831	2% (0.998)	10%

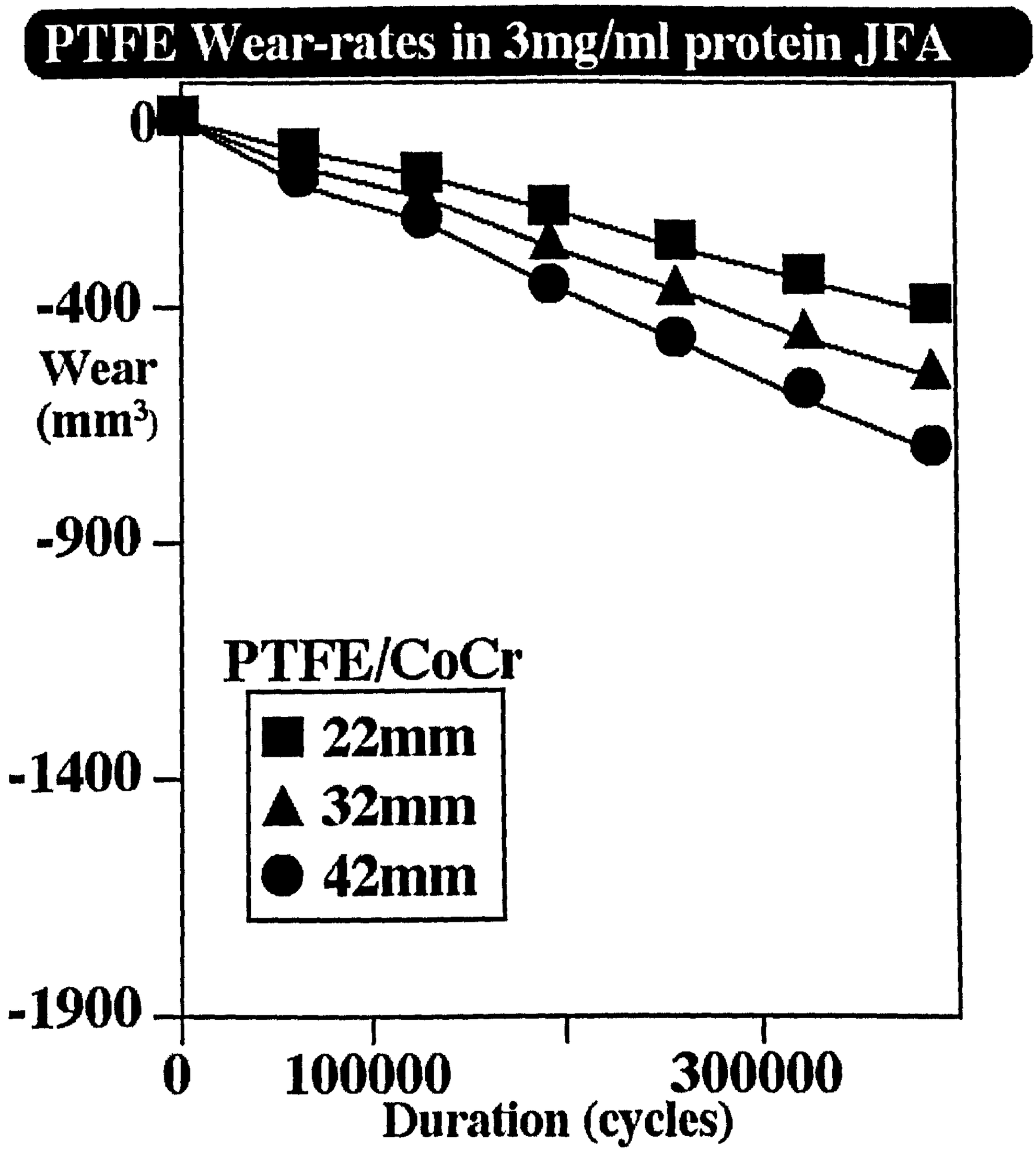


Figure 4.9 Volumetric wear of non-sterilized PTFE using bovine serum with 3mg/ml protein concentration as the JFA

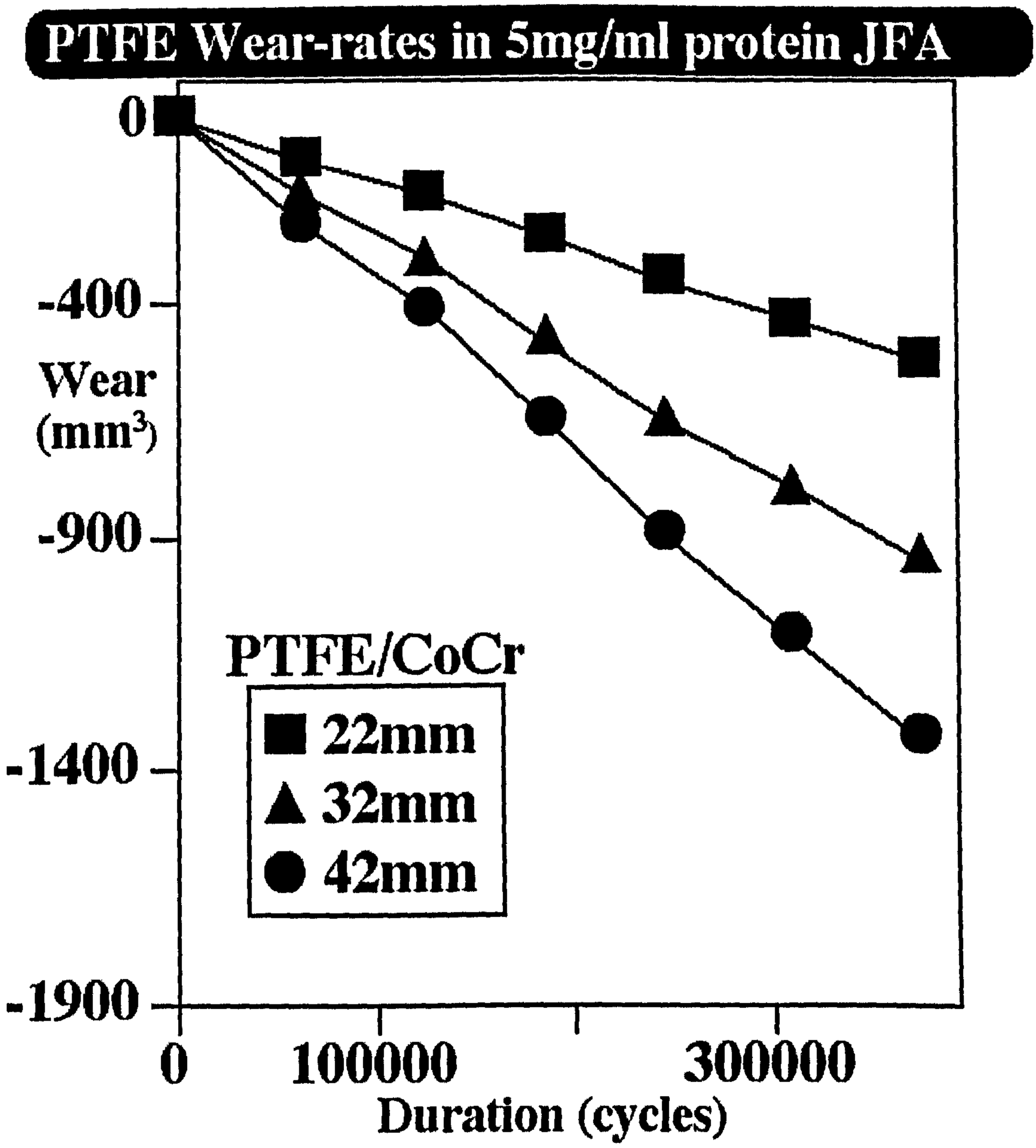


Figure 4.10 Volumetric wear of non-sterilized PTFE/CoCr using bovine serum with 5mg/ml protein concentration as the JFA

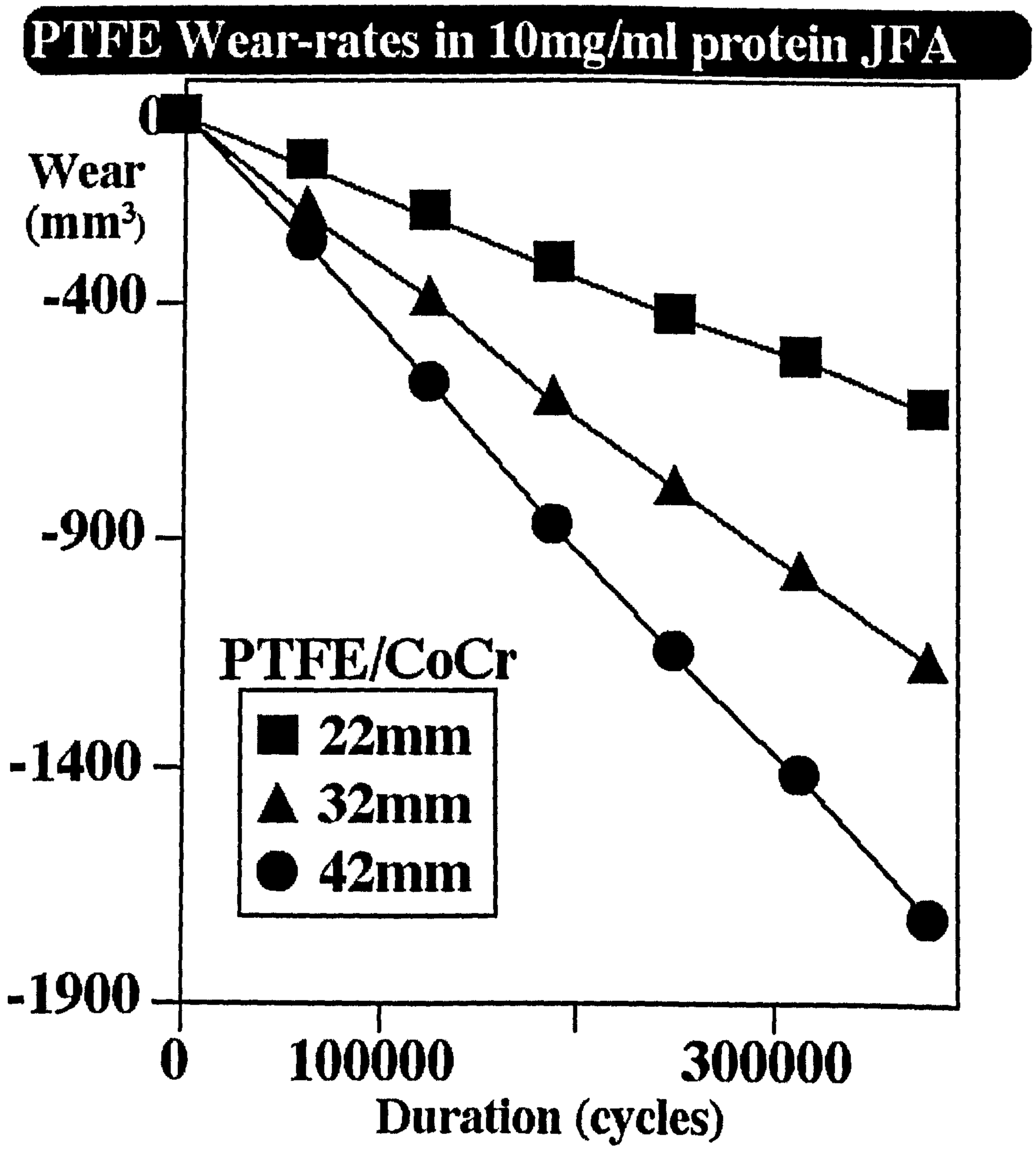


Figure 4.11 Volumetric wear-rates of non-sterilized PTFE/CoCr using bovine serum with 10mg/ml protein concentration as the JFA

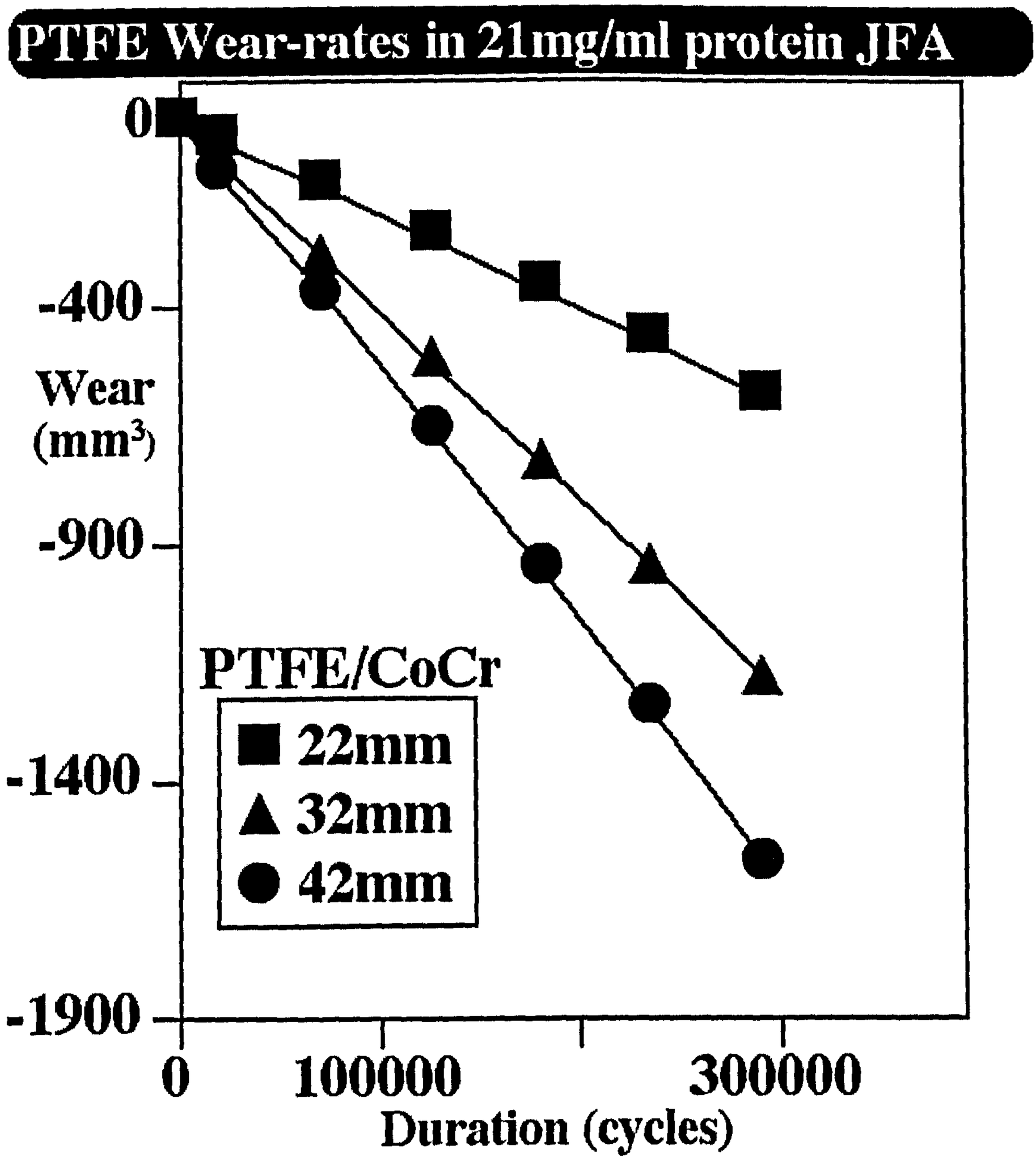


Figure 4.12 Volumetric wear of non-sterilized PTFE/CoCr using bovine serum with 21mg/ml protein concentration as the JFA

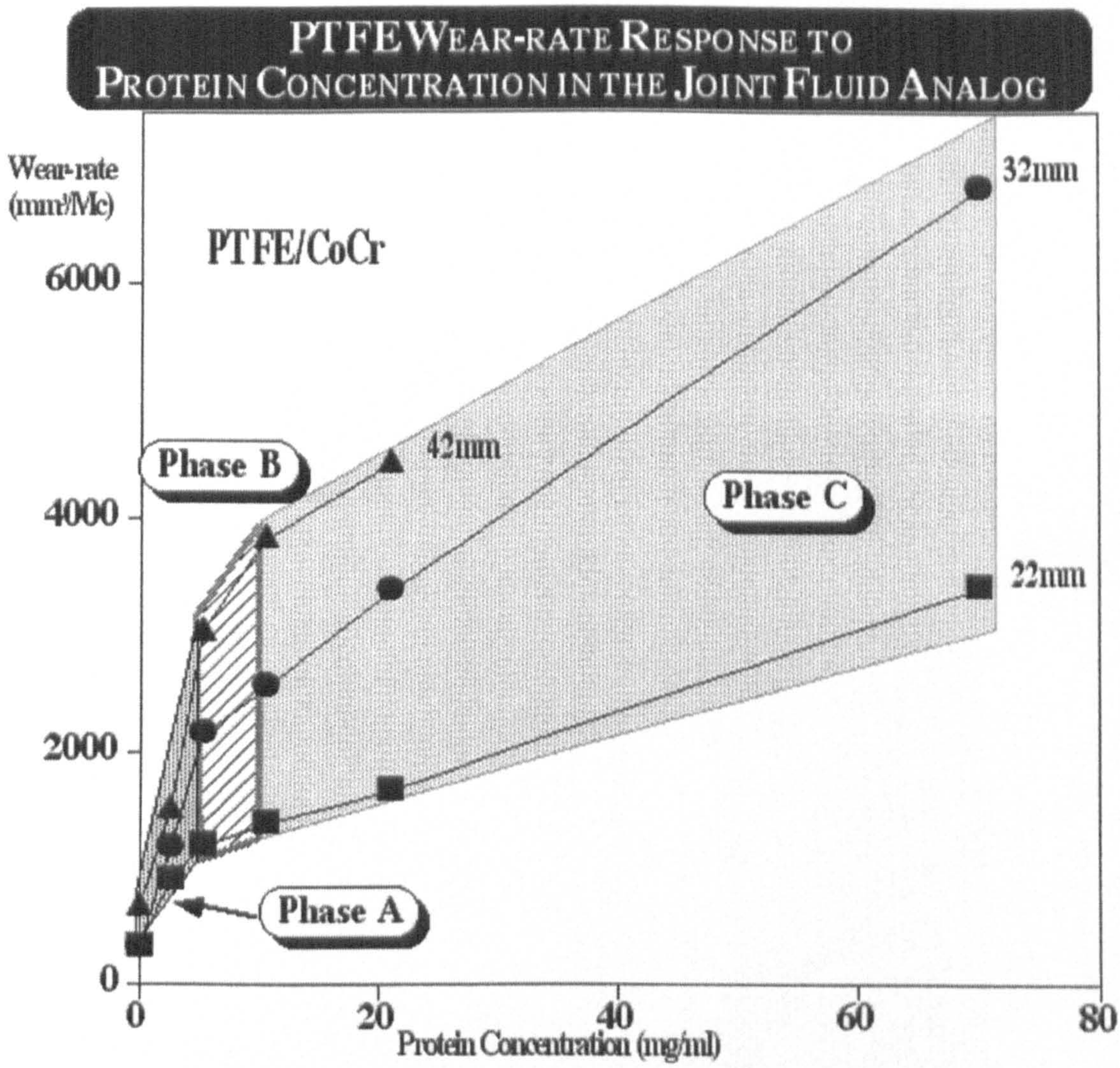


Figure 4.13 Volumetric wear-rates of PTFE/CoCr with respect to protein concentration in bovine serum with 3 distinct phases of wear.

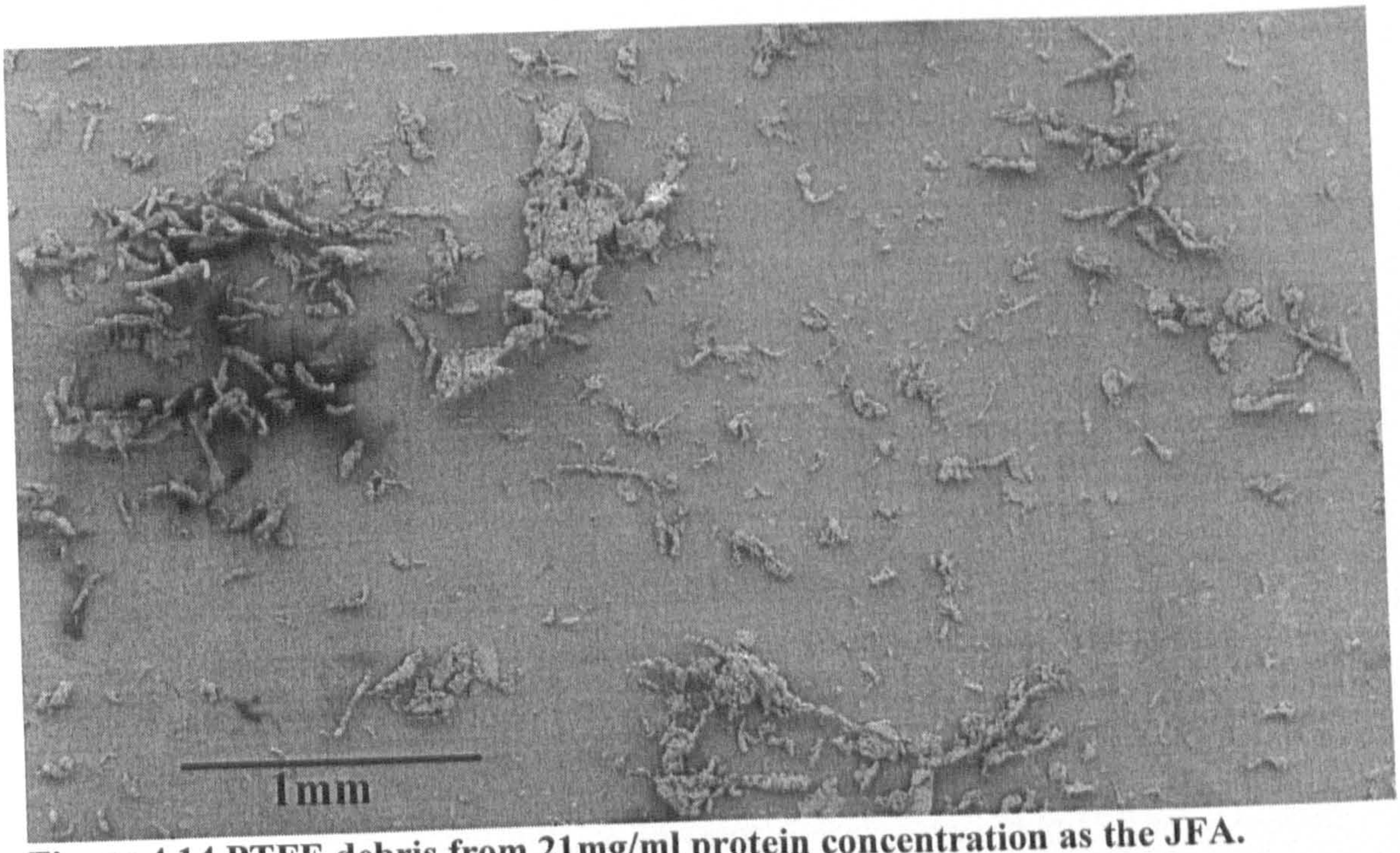


Figure 4.14 PTFE debris from 21mg/ml protein concentration as the JFA.

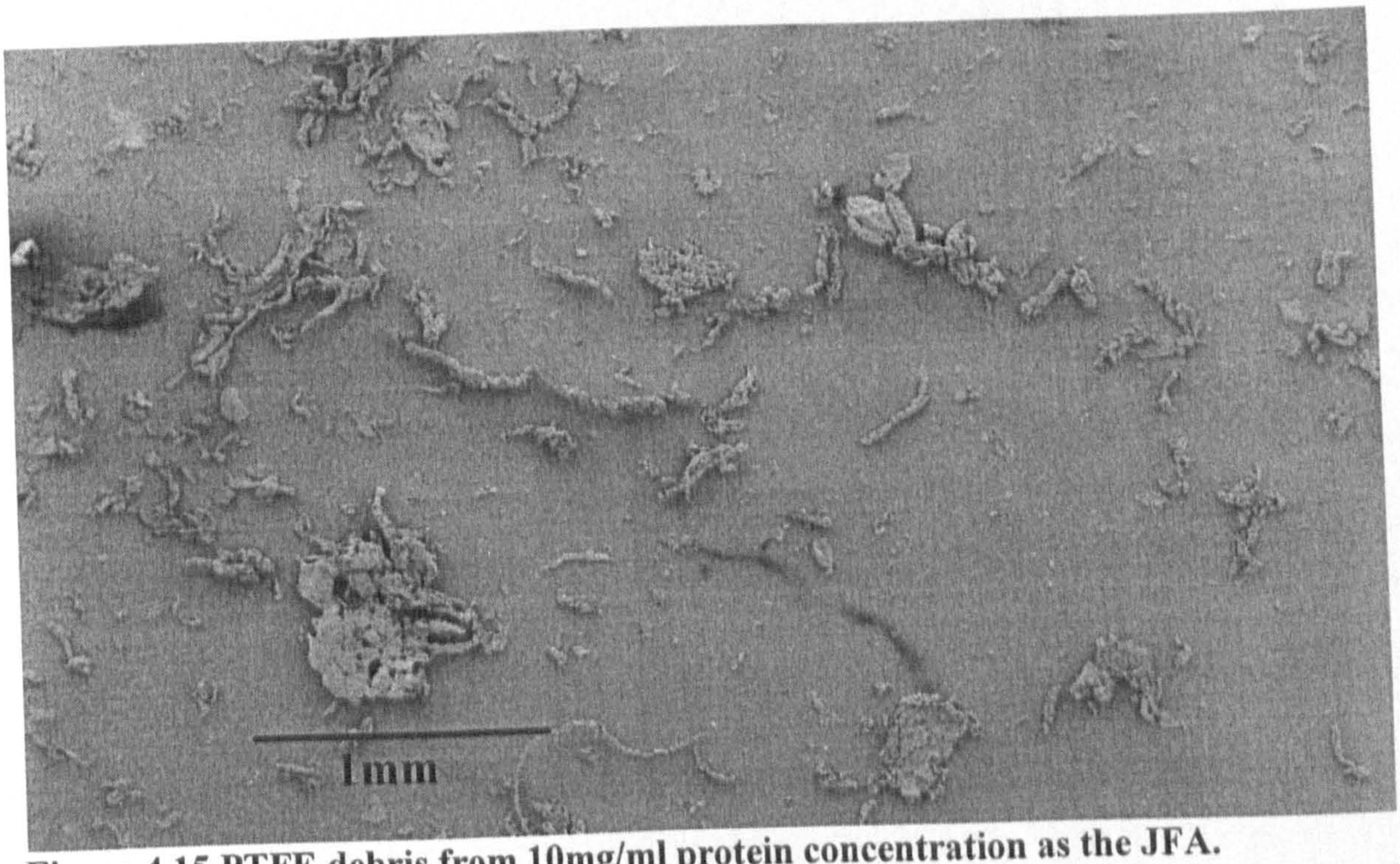


Figure 4.15 PTFE debris from 10mg/ml protein concentration as the JFA.

4.4 UHMWPE WEAR (THR INV): JFA PROTEIN CONCENTRATION DOSE RESPONSE

Wear-rates of the non-sterilized GUR 4150 UHMWPE were linear with test duration for the six protein dose response experiments (fig 4.16 - 4.21) and soak-rates were between 1 and 4% of the wear-rates. Wear-rates were highest, 46.1, 62.1 and 74.7 mm^3/Mc for 22, 28 and 32mm respectively, at 10 mg/ml of protein (fig 4.22, table 4.4). At this protein concentration (10mg/ml) there was a clear ball size effect with a 6% wear-rate penalty per millimeter of head size as shown clinically (fig 4.18). The wear-rates in protein concentrations below 10 mg/ml were at least 1.2 times lower and the ball size effect was evident but the VWI varied from 2 to 6% (fig 4.16 and 4.17). In protein concentrations above 10 mg/ml wear-rates were again lower but the 28mm wear-rate was generally indistinguishable from the 22mm (fig 4.19 - 4.21)

There were more fibrils (fig. 4.23 and 4.24) in the debris from 21mg/ml and 10mg/ml of protein concentration than in 62mg/ml protein concentration. The fibrils from 10mg/ml protein concentration were approximately 3μm long by 0.5μm wide whereas the fibrils from 21mg/ml were similar in width but slightly shorter at 2μm. Round particles from both protein concentrations were similar averaging 0.8 to 1μm in diameter.

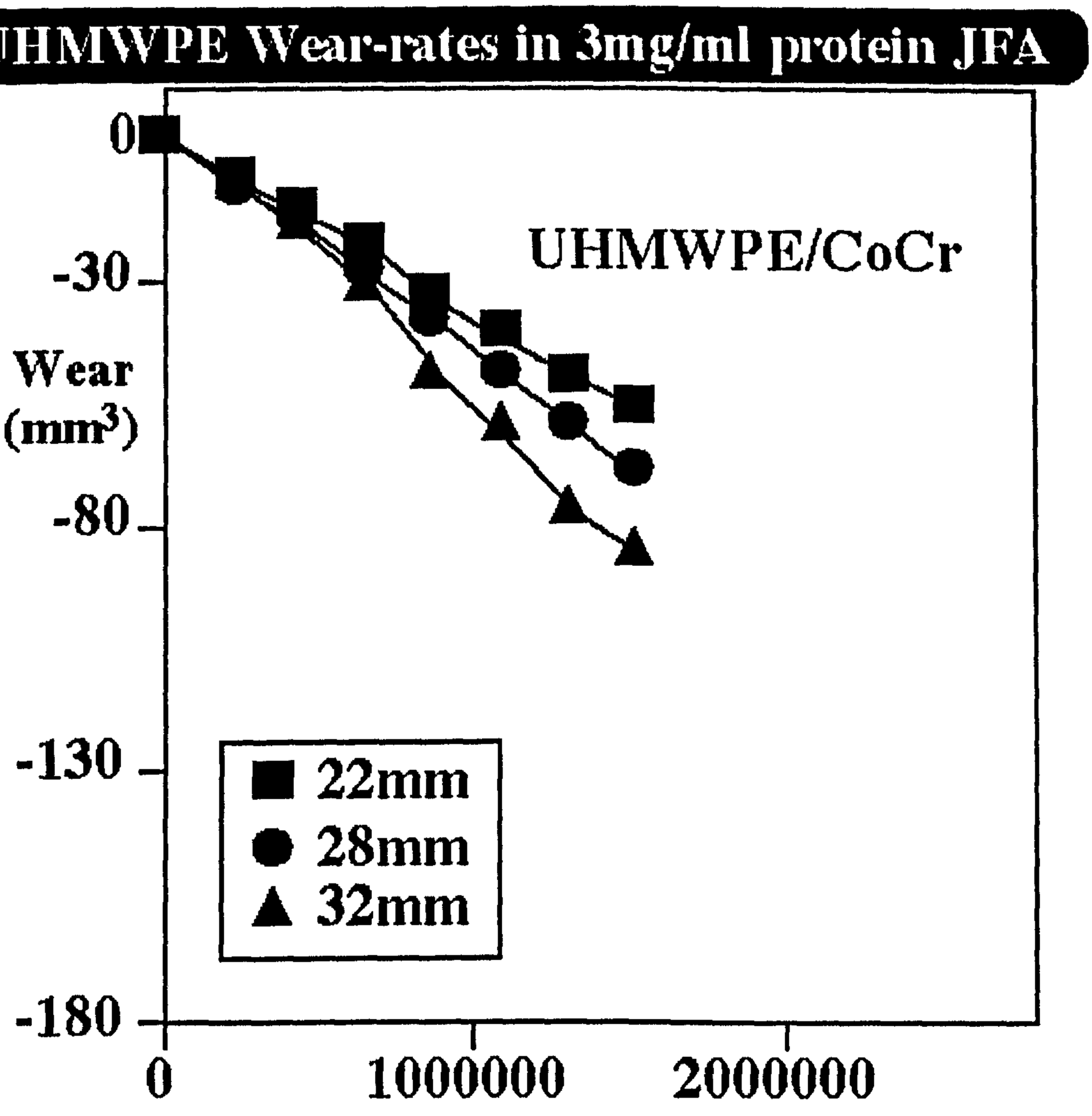


Figure 4.16 Volumetric wear-rates of non-sterilized GUR 4150 UHMWPE/CoCr using bovine serum with 3mg/ml protein concentration as the JFA.

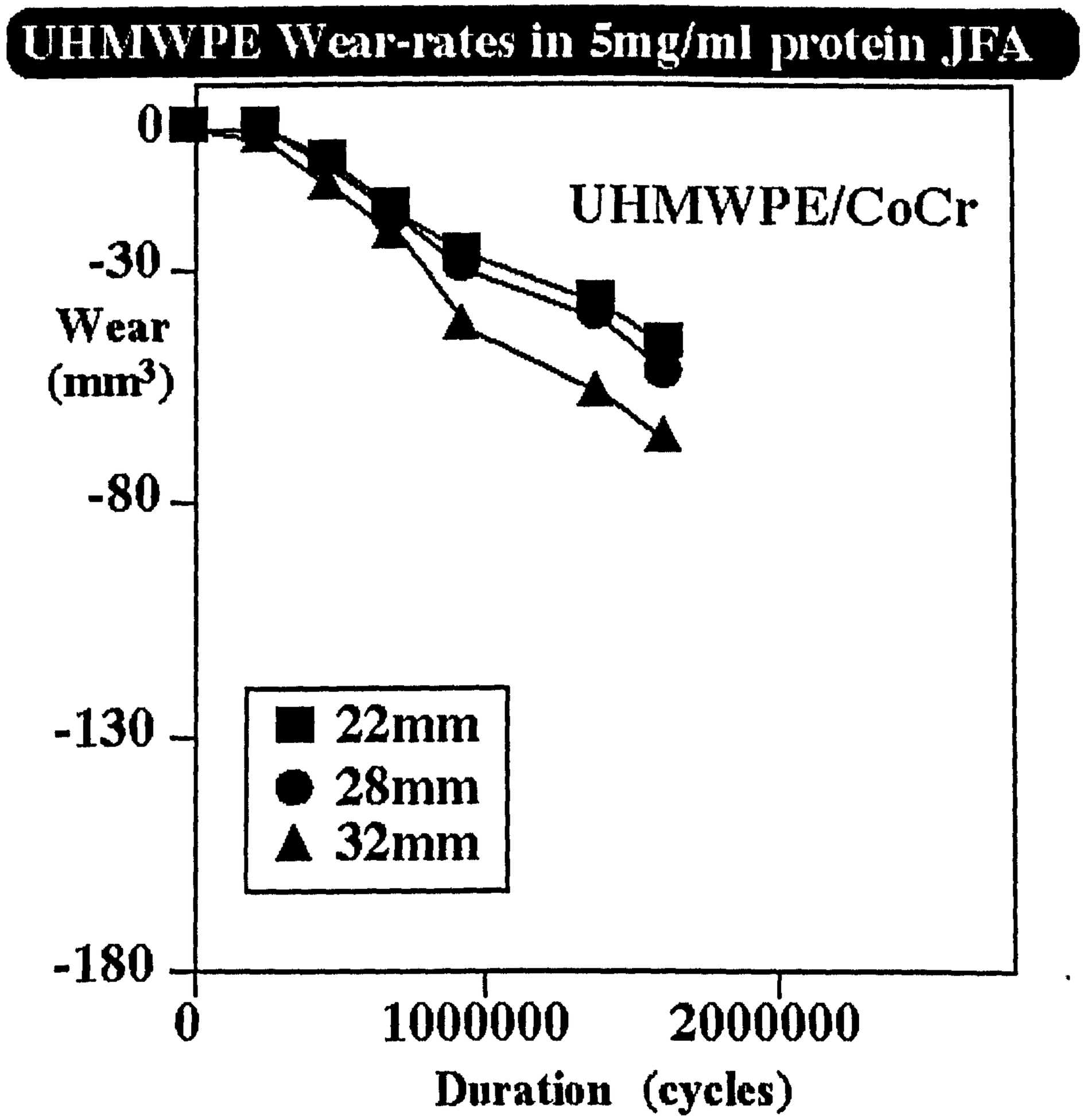


Figure 4.17 Volumetric wear-rates of non-sterilized GUR 4150 UHMWPE/CoCr using bovine serum with 5mg/ml protein concentration as the JFA.

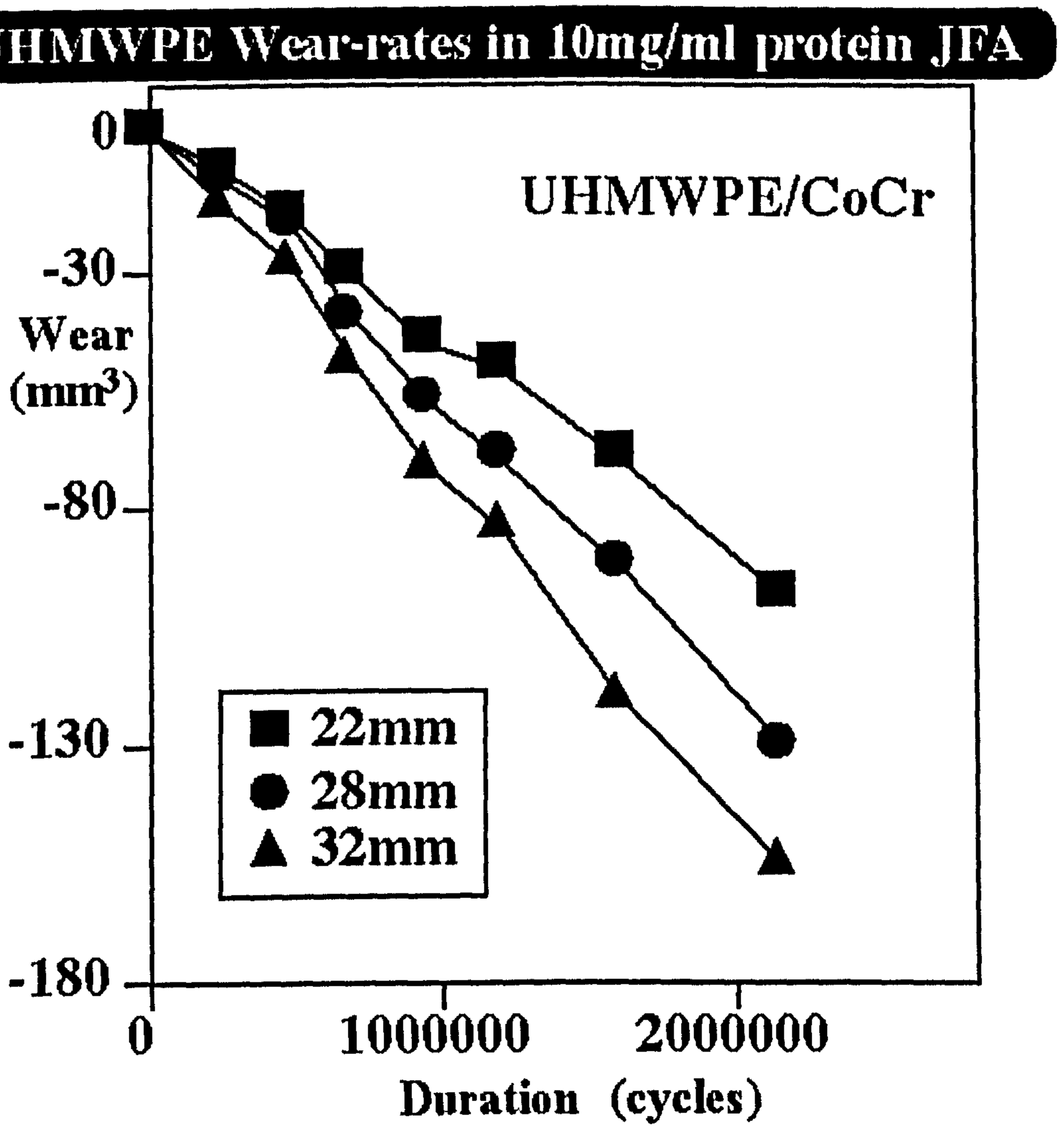


Figure 4.18 Volumetric wear-rates of non-sterilized GUR 4150 UHMWPE/CoCr using bovine serum with 10mg/ml protein concentration as the JFA.

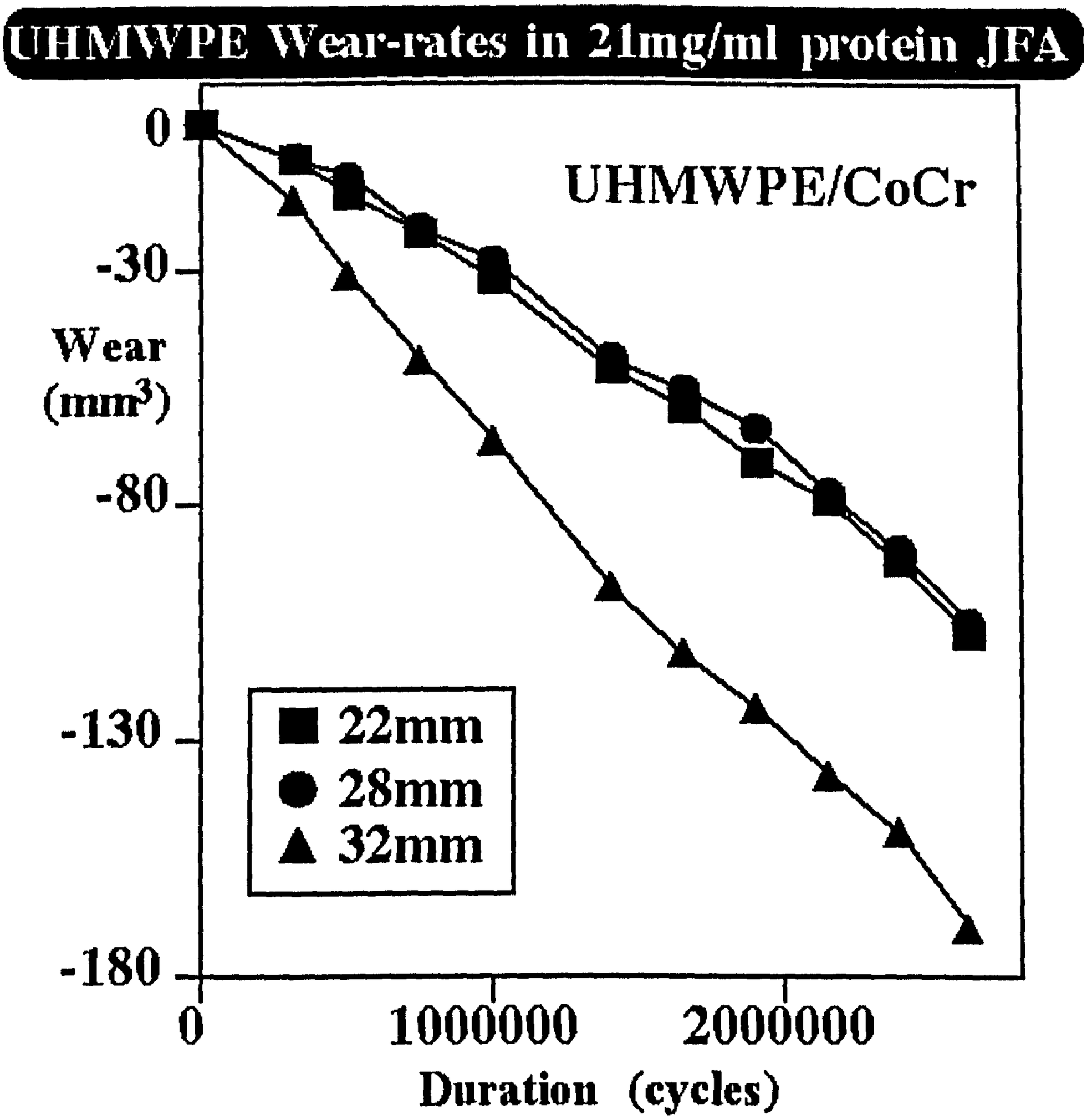


Figure 4.19 Volumetric wear-rates of non-sterilized GUR 4150 UHMWPE/CoCr using bovine serum with 21mg/ml protein concentration as the JFA.

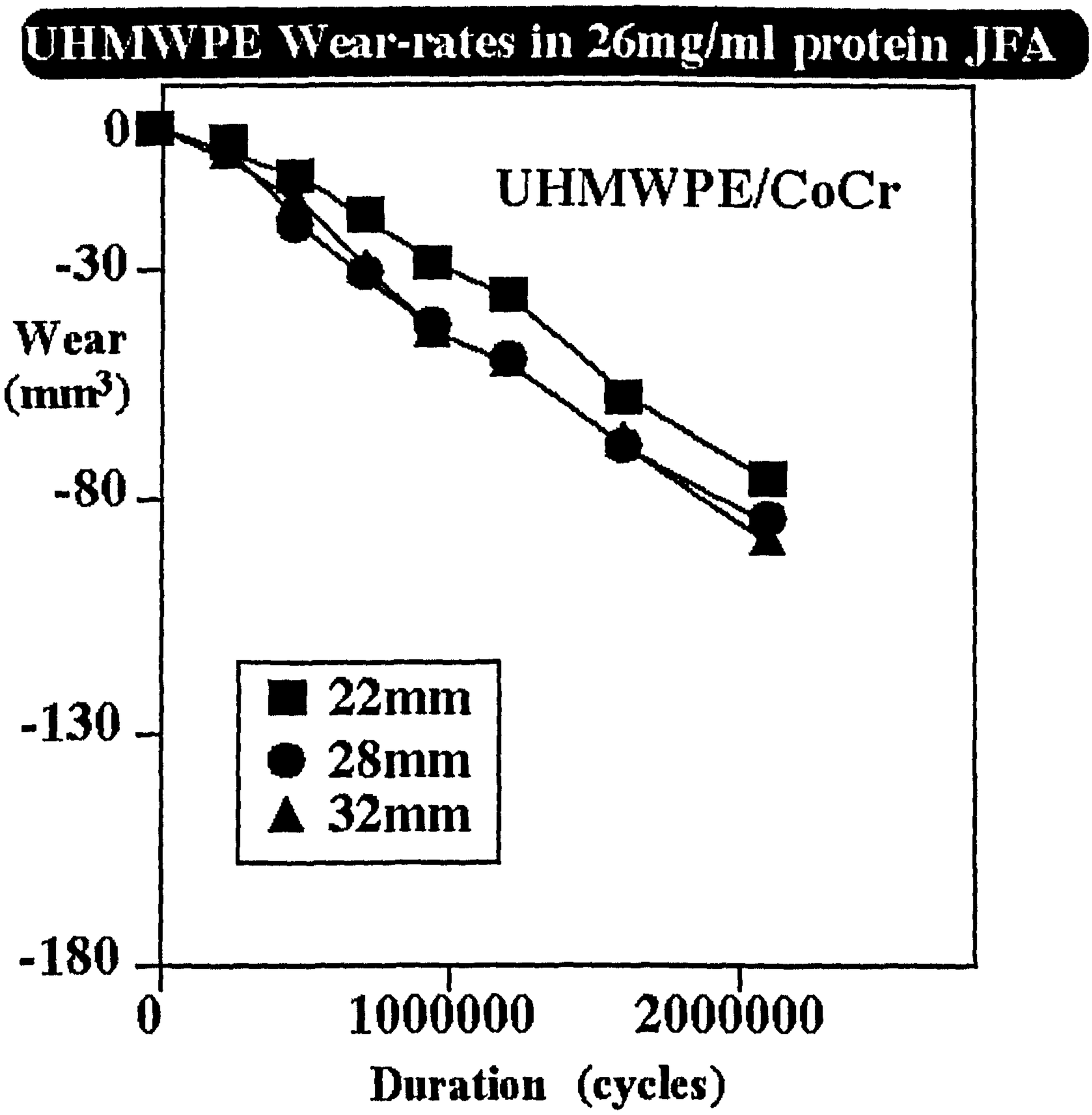


Figure 4.20 Volumetric wear-rates of non-sterilized GUR 4150 UHMWPE/CoCr using bovine serum with 26mg/ml protein concentration as the JFA.

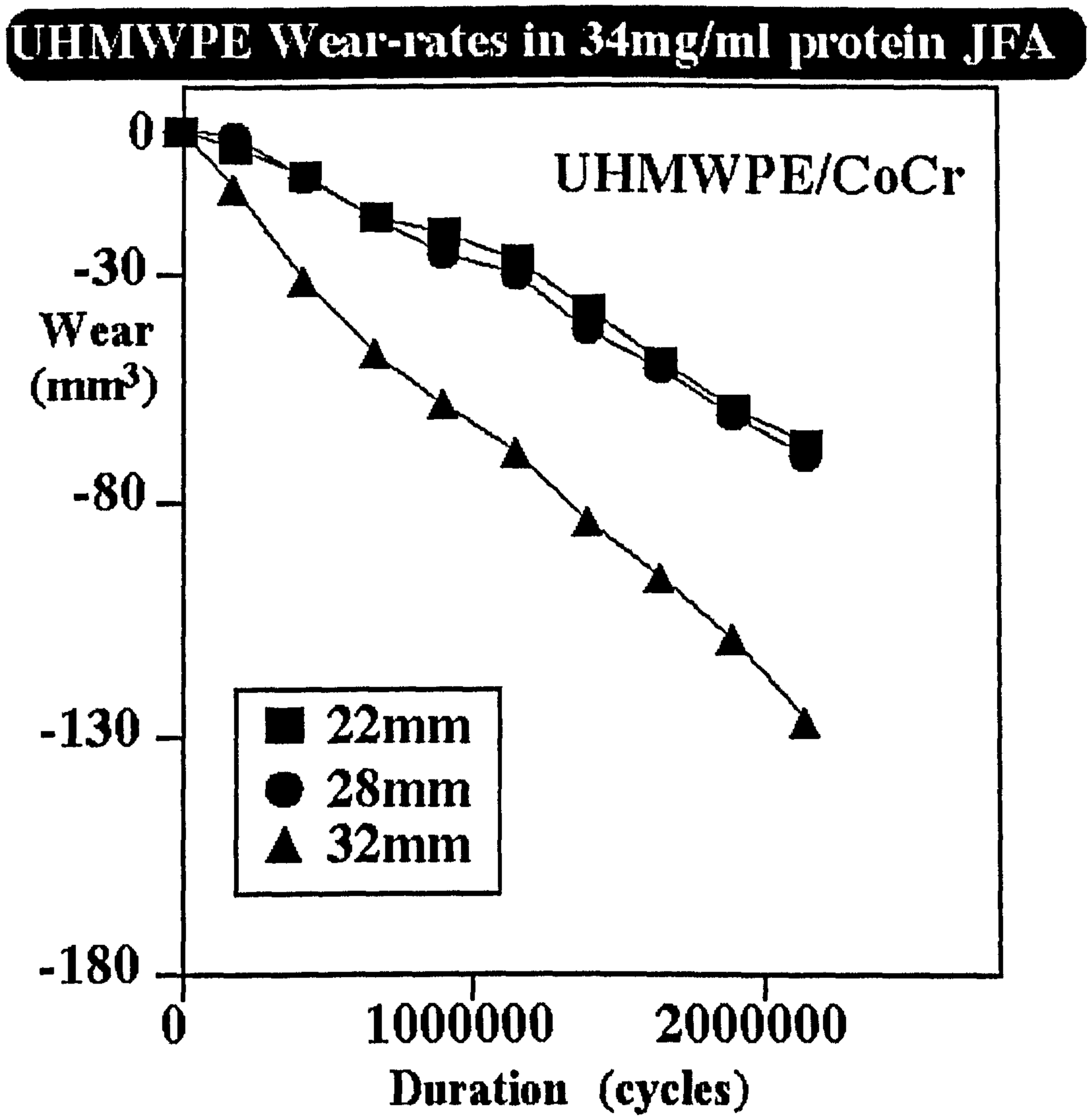


Figure 4.21 Volumetric wear-rates of non-sterilized GUR 4150 UHMWPE/CoCr using bovine serum with 34mg/ml protein concentration as the JFA.

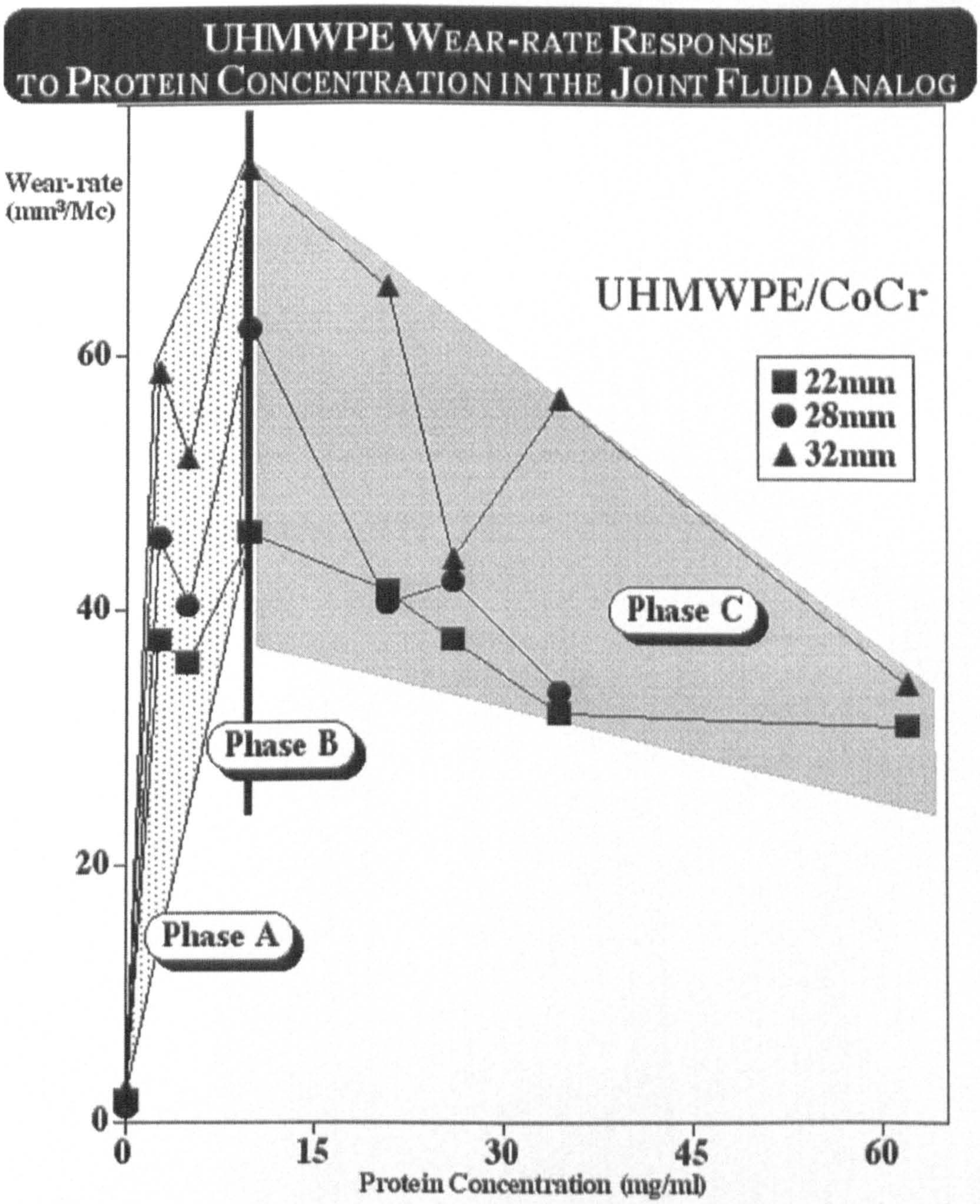


Figure 4.22 Volumetric wear-rates of non-sterilized GUR4150 UHMWPE/CoCr with respect to protein concentration in bovine serum with three distinct phases of wear-rates

Table 4.4 Volumetric wear-rates of non-sterilized GUR4150 UHMWPE/CoCr

UHMWPE/CoCr in bovine serum diluted with de-ionized water to protein concentrations of 34, 26, 21, 10, 5 and 3mg/ml. VWI = Volumetric Wear Index.

Expt. No.	*Protein Concentration*	*Dia (mm)*	*Wear-rate (mm^3/Mc)*	*Precision & (R^2)*	*VWI*
HE050	0	22	2.0	61% (0.309)	Reference
		32	1.7	102% (0.490)	-2%
		42	2.4	49% (0.151)	1%
HE066	34mg/ml	22	31.8	18% (0.984)	Reference
		28	33.5	19% (0.993)	1%
		32	56.4	12% (0.994)	8%
HE110	26mg/ml	22	37.6	9% (0.986)	Reference
		28	42.2	9% (0.991)	2%
		32	44.1	7% (0.994)	2%
HE064	21mg/ml	22	41.5	4% 90.993)	Reference
		28	40.5	6% (0.987)	0%
		32	65.4	8% (0.996)	6%
HE111	10mg/ml	22	46.1	12% (0.995)	Reference
		28	62.1	6% (0.994)	6%
		32	74.7	11% (0.996	6%
HE115	5mg/ml	22	35.8	11% (0.975)	Reference
		28	40.3	10% (0.974	2%
		32	51.9	13% (0.976)	4%
HE114	3mg/ml	22	37.7	3% (0.998)	Reference
		28	45.7	12% (0.998)	4%
		32	58.7	10% (0.992)	6%
HE071	62mg/ml	22	31.0	9% (0.985)	Reference
		32	34.2	8% (0.990)	1%
		42	36.2	12% (0.992)	1%

The PTFE/PE wear-rate ratios were highest (>100) in water and for protein levels above 62mg/ml of protein (table 4.5). The 22mm cups exhibited PTFE/PE ratios well within the clinical range for all other protein concentrations. For the 32mm cups the PTFE/PE ratio was high at 21mg/ml of protein but within the clinical range for the remaining 3 protein concentrations of 10, 5 and 3mg/ml (table 4.5).

Table 4.5 PTFE/UHMWPE Wear-rate Ratio

Volumetric wear-rate ratios of PTFE to non-sterilized GUR4150 UHMWPE in 0, 3, 5, 10 and 21 mg/ml of protein concentration in bovine serum.

	Protein Concentration (mg/ml)					
Dia (mm)	*0*	*3*	*5*	*10*	*21*	*>60*
22	190	23	33	30	40	>100
32	280	20	41	34	51	>200
42	273	NA	NA	NA	NA	>200

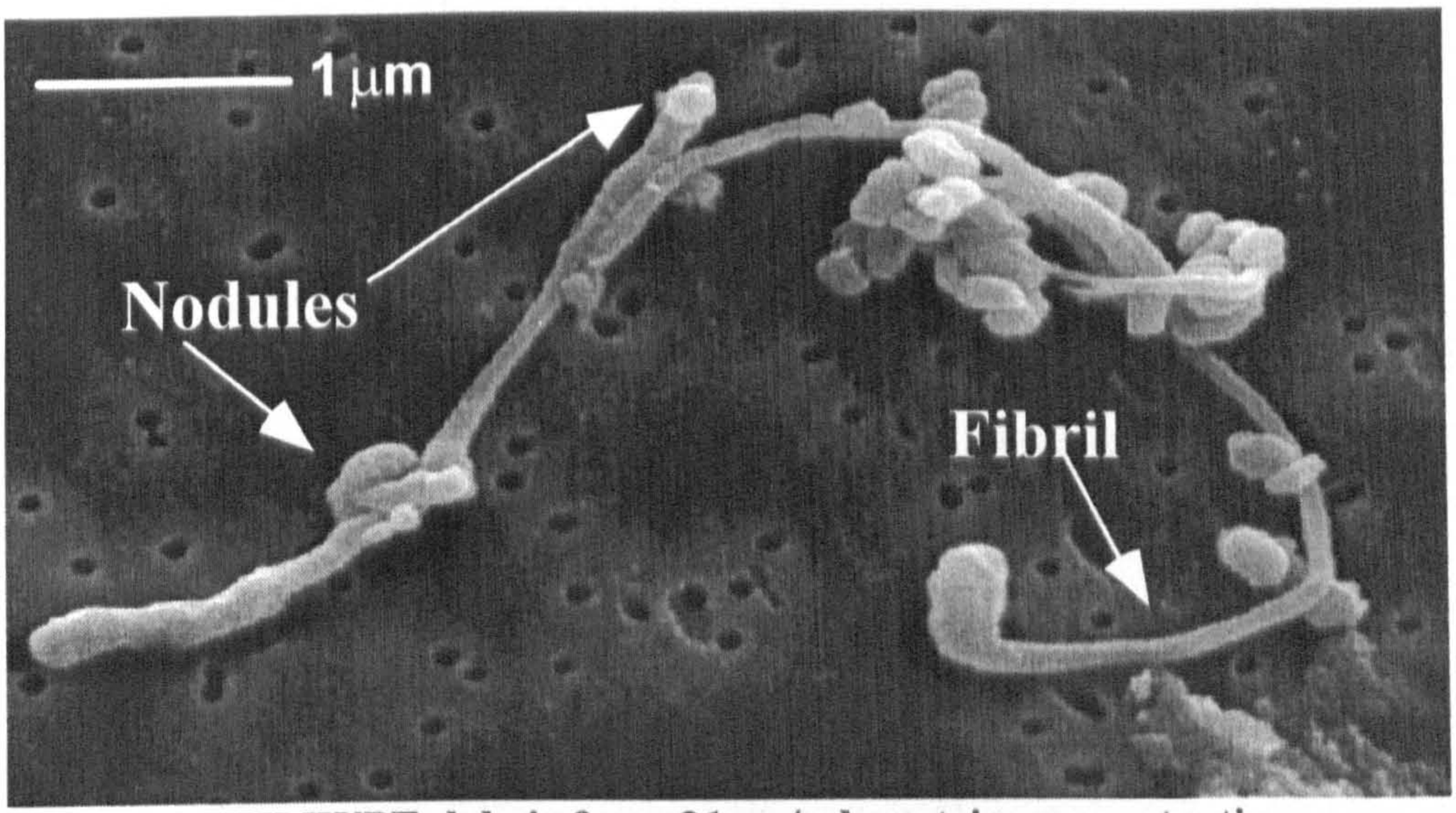


Figure 4.23 UHMWPE debris from 21mg/ml protein concentration

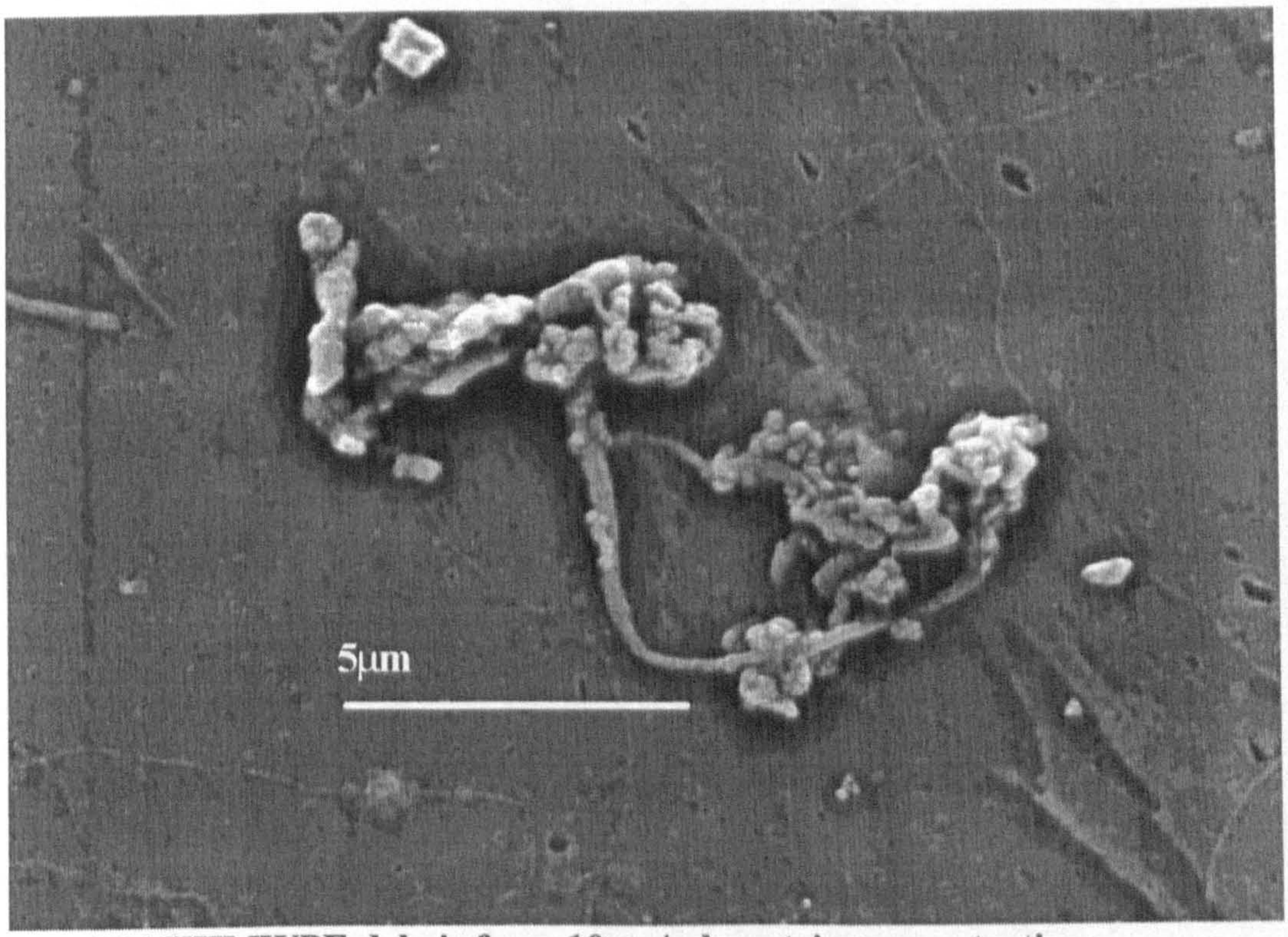


Figure 4.24 UHMWPE debris from 10mg/ml protein concentration

4.5 PTFE WEAR: EFFECT OF JFA VOLUME AND THR CONFIGURATION

Wear-rates of PTFE run Inv with 450ml of JFA were greatest using both bovine serum with 21 mg/ml of protein and alpha calf serum with 22 mg/ml of protein and lowest in the 10mg/ml of protein concentration (table 4.6). In the Anat THR configurations with 450ml of fluid wear-rates were greater for the Anat/horiz versus the Anat/obliq cups (fig 4.25). Both Anat configurations had a 12 to 14% increase in wear-rates from 21 mg/ml of protein to 22 mg/ml of protein. However, wear-rates in the Inv configuration showed less than 4% difference between the 2 types of serum. A comparison of amount of fluid showed that the wear-rates in the Inv configuration with 250ml of serum (10 and 21mg/ml protein) were 38% less than in 450ml of serum (fig 4.25).

The PTFE wear-rates in the Anat/obliq THR configuration were lower than the wear-rates in the Anat/horiz configuration (table 4.6 and fig 4.25). Wear-rates in the large volume Anat configuration continued to show the same trend for increased wear-rate with increased protein concentration as seen with the small volume Inv configuration studies (fig 4.26).

Table 4.6 Volumetric wear-rates of PTFE/CoCr

PTFE/CoCr in diluted (de-ionized water) alpha calf serum (ACS) and bovine calf serum (BCS). Cups were mounted Anat/obliq (cup on top with cup face inclined 23° to the horizontal plane), Anat/horiz (cup on top with cup face parallel to the horizontal plane) and Inv (cup on bottom) all in 450ml chambers.

Expt. No.	*Material*	*Proteins & Fluid Type*	*Dia (mm)*	*Wear-rate (mm^3/Mc)*	*Precision & (R^2)*	*THR Configuration*
HE141	PTFE	21mg/ml BCS	32	4442	11% (0.995)	Anat/obliq
				4867	18% (0.998)	Anat/horiz
				6576	14% (0.998)	Inv
HE140	PTFE	22mg/ml ACS	32	4843	4% (0.999)	Anat/obliq
				5270	8% (0.999)	Anat/horiz
				6331	5% (0.999)	Inv
HE147	PTFE	10mg/ml BCS	32	5790	7% (0.981)	Inv
		17mg/ml BCS	32	3978	4% (0.995)	Anat/obliq
		34mg/ml BCS	32	7513	7% (0.982	Anat/obliq
		62mg/ml BCS	32	8116	5% (0.990)	Anat/obliq

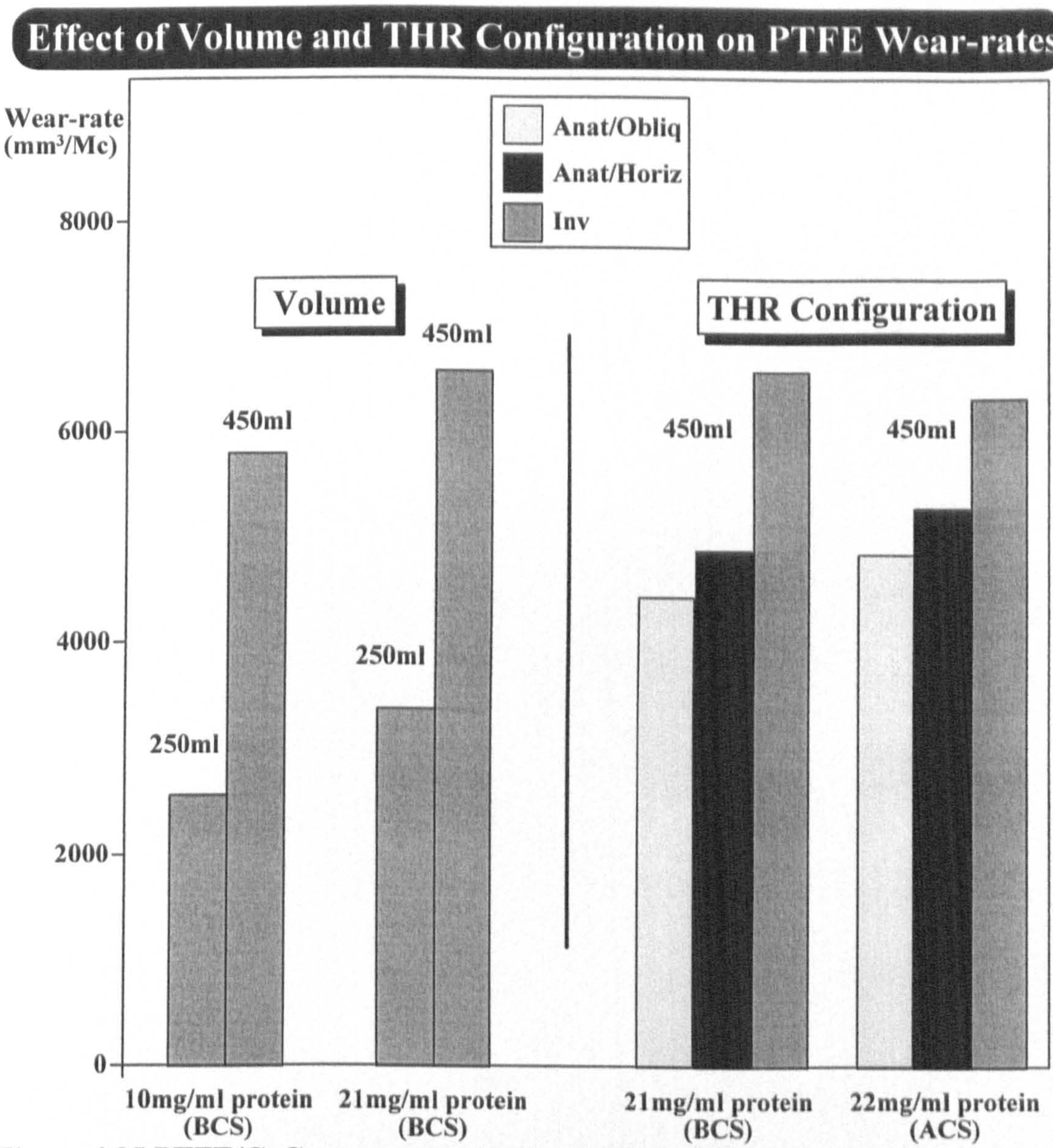


Figure 4.25 PTFE/CoCr wear-rates in 10mg/ml and 21mg/ml protein concentration (bovine calf serum: BCS) and 22 mg/ml protein concentration (alpha calf serum: ACS) with respect to THR configuration and volume. The 250ml Inv 21mg/ml protein results were from HE041 reported in Table 4.3

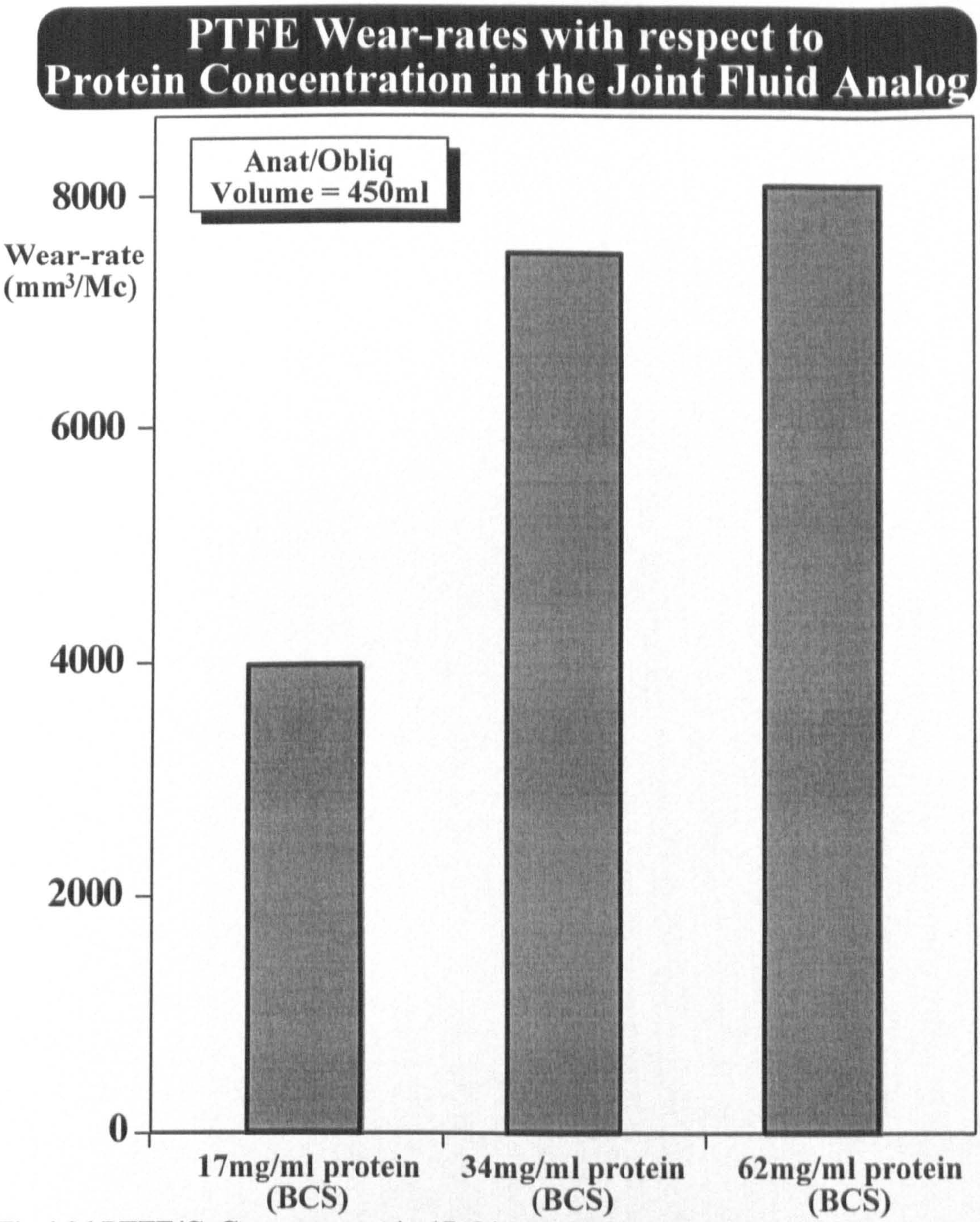


Fig 4.26 PTFE/CoCr wear-rates in 17, 34 and 62mg/ml protein concentration (bovine calf serum: BCS) THR configuration for the three studies was Anat/obliq and volume of JFA was 450ml

CHAPTER 5. DISCUSSION

5.1 PTFE AND UHMWPE WEAR: JFA OF WATER

The wear studies of PTFE and UHMWPE, using water as the JFA, did not duplicate any of the validation criteria (table 5.1). Both materials showed clinically conflicting results with regard to the five clinical criteria and the data was outside of the limits of the two definitions criteria of precision and linearity.

The current studies showed that smaller diameter cups for both PTFE and UHMWPE had more wear than the larger diameter cups, which was contrary to clinical findings. In a study by Pappas et al, (1995) they too found this same phenomenon using water as the JFA. Their study showed that 47mm UHMWPE cups wore less (<18 fold) than the 32mm UHMWPE cups. The VWI was also incorrect in the present studies due to the contrary ball size effect. The VWI was based on the clinical findings that both polymers showed an approximate 6% wear-rate increase per millimeter of head diameter. However, since the current studies in water showed wear-rates to decrease with increased diameter the VWI was negative.

The PTFE/PE ratio was also incorrect in the present studies due to the large discrepancy in wear-rates. The PTFE wear-rates were 5 times lower than clinical and UHMWPE was at least 25 times lower. This was not the first time low in-vitro wear-rates in water were reported. Wear tests using LST devices have consistently exhibited wear-rates 1 to 3 orders of magnitude lower than clinical for UHMWPE (Atkinson et al, 1978, Charnley, 1976, Derbyshire et al, 1994, Dowson, 1978, Dowson et al, 1974, Pappas et al, 1995, Saikko, 1993 and Seedhom et al, 1973). Wear-rates using M-dST devices have also been lower but with not as large a difference as LST devices (Dowson and Jobbins, 1988 and Saikko, 1996). Despite

the underestimation of wear-rates in the current studies, the magnitude difference between the two materials was large enough to produce a PTFE/PE wear-rate ratio at least 4 times greater than clinical.

The debris characteristics were also opposite to clinical. The large ribbons and flakes of PTFE and UHMWPE debris found in the present studies were also reported by laboratories using M-dST simulators (Clarke et al, 1995, Dowson and Jobbins, 1988, Saikko, 1996 and Wang et al, 1995). These flakes were different from the micron and sub-micron size particles of UHMWPE seen from retrieved tissues (Campbell et al, 1995, Maloney et al, 1995 and McKellop et al, 1995). Similarly, Charnley made a point of stating that no large flakes of PTFE were seen at revision surgery (Charnley et al, 1969).

The data definition criteria of precision and linearity were also not met in the current studies using water as the JFA. Precision among the wear-rates of PTFE and UHMWPE was very poor (>49%). This indicated that there were large discrepancies between the three specimens tested for each diameter. In addition, linearity of the UHMWPE wear-rates (R^2<0.5) was indiscernible. Although the linearity of the PTFE wear-rates was better than UHMWPE, it was still very poor for this material. In previous studies from this laboratory using bovine serum as the JFA, PTFE wear-rates have been shown to be very linear with R^2 >0.98.

Based on the discrepancies found in the current and past studies, water as the JFA did not duplicate the clinical findings. Therefore, it is this author's conclusion that water should not be used as a JFA in PTFE and UHMWPE wear studies.

5.2 PTFE AND UHMWPE WEAR: JFA OF BOVINE SERUM (>60mg/ml PROTEIN CONCENTRATION)

The current studies, using bovine serum with >60mg/ml protein concentration as the JFA, duplicated 5 of the validation criteria for PTFE and 3 for UHMWPE (table 5.1). The PTFE acetabular cups duplicated three clinical criteria, ball size effect, VWI and debris and the two data definition criteria of precision and linearity. On the other hand, the UHMWPE acetabular cups only duplicated one clinical criterion, ball size effect, and the two data definition criteria.

The current studies showed that both PTFE and UHMWPE duplicated the ball size effect, however only PTFE duplicated the clinical VWI. The use of multiple head diameters for hip wear testing has not been widespread, however there were five studies which examined UHMWPE with at least two head diameters and used bovine serum as the JFA (Bragdon et al, 1998, Clarke et al, 1993, Clarke et al, 1996, McKellop et al, 1995 and Wright and Scales, 1980). With only one exception (Wright and Scales, 1980) all found the same results as the current study, the smaller diameter cups had lower wear-rates than the larger diameter cups. The four separate studies, which duplicated the ball size effect also, had VWI within the clinical range, however the current study did not. The VWI for the current study of UHMWPE was 1%, due to the similar wear-rates of the three diameters. For PTFE there have been two studies using bovine serum and multiple head diameters (Clarke et al, 1997 and Phipatanakul et al 1997; both from the author's laboratory). These studies also duplicated the clinical findings of ball size effect using bovine serum as the JFA. These and the current study also duplicated the clinical VWI.

Most studies, using bovine serum as the JFA, have underestimated wear-rates of UHMWPE (table 2.17). LST devices showed the largest underestimation of wear-rate, which varied from 1 to 4 orders of magnitude less than clinical (Charnley,

1976, Cooper et al, 1993, Dumbleton et al, 1974, Fisher et al, 1995 and McKellop and Clarke, 1988). The high wear resistance of UHMWPE was attributed to the linear surface tracking of the testing device which was shown to be incorrect (Bragdon et al, 1996 and Wang et al, 1997). M-dST simulator studies using bovine serum also underestimated clinical wear-rates. The difference was less than for LST devices, however, at 1.5 to 50 times lower than clinical values the wear-rates were still not within 20% of the clinical range (Bigsby et al, 1997, Bragdon et al, 1996, Duff-Barclay and Spillman, 1967, Hamilton et al, 1996, McKellop et al, 1992 and Wright and Scales, 1977). The present simulator study was no exception; wear-rates of UHMWPE were 1.6 to 2.3 times lower than the clinical values.

Contrary to UHMWPE studies, PTFE studies have overestimated clinical wear-rates (Clarke and McKellop, 1980, McKellop, 1981, McKellop et al, 1977, Phipatanakul et al, 1997). Again the present study was no exception with wear-rates 2 to 3 times higher than clinical.

Based on the overestimated wear-rates of PTFE and the underestimated wear-rates of UHMWPE, the PTFE/PE wear-rate ratio was at least 2 times greater than clinical. Four additional studies comparing UHMWPE and PTFE also found the PTFE/PE ratio to be too high in comparison to clinical (Charnley, 1976, Clarke et al, 1997, McKellop and Clarke, 1983 and McKellop et al, 1981).

In a comparison of UHMWPE wear debris in-vivo to in-vitro, McKellop et al, (1995) showed that the in-vitro debris contained more round particles than found in-vivo. Using a M-dST simulator with 90% bovine serum as the JFA, UHMWPE debris consisted of approximately 95% round particles and 5% elongated fibrils and shreds whereas in-vivo the number of round particles was approximately 75% and

elongated fibrils 25%. They also showed that the rounded and elongated particles in the simulator were smaller (0.2 versus 0.2 to 0.8μm) and shorter (2 versus 4 μm) than in-vivo respectively. The debris from this study with 90% bovine serum showed comparable results to the simulator data from McKellop et al, (1995) with the rounded particles smaller, elongated particles shorter and a greater predominance of rounded (approximately 90 to 95%) particles than in-vivo (fig 4.8).

A literature survey showed no published quantitative analysis of PTFE debris only the qualitative description by Charnley et al, (1969). The validation of PTFE debris was restricted to Charnley's description of 'hairy' and 'stringy' along with size varying from 2 to 50 μm long. The debris from this study demonstrated the 'stringy ' appearance described by Charnley (fig 4.4). These strings were actually made up of many fibril type particles forming fibrous rods. However, the rods in the simulator could be as long as 1mm. It is not documented whether rods of this length were produced in-vivo; therefore, the debris characterization fit with Charnley's description with the possible exception of the length.

Both materials showed a high degree of precision among the three identical THR specimens. In addition, linearity was very good at R^2 greater than 0.98. Therefore, both PTFE and UHMWPE met the data definition criteria (linearity and precision) using bovine serum with protein concentration > 60mg/ml.

From the current studies it was shown that using bovine serum with > 60mg/ml of protein concentration did not duplicate all the validation criteria. In regards to the validation criteria, the current studies and past studies generally agreed for both PTFE and UHMWPE. This showed that bovine serum was closer to duplicating the clinical findings than water, however, it was this author's conclusion

that bovine serum with >60mg/mel protein concentration was not the correct JFA for hip wear testing of UHMWPE and PTFE.

Table 5.1 Wear Validation Criteria for PTFE and UHWPE

Studies in water and bovine serum with > 60mg/ml protein concentration

No.	*Wear Validation Criteria*	*PTFE Water*	*UHMWPE Water*	*PTFE Serum*	*UHMWPE Serum*
1	Ball Size Effect	NO	NO	YES	YES
2	VWI (6-10%/mm ball dia)	NO	NO	YES	NO
3	Wear-rate Magnitude (±10%)	NO	NO	NO	NO
4	PTFE/PE (range 20-50)	NO		NO	
5	Debris comparable to in-vivo	NO	NO	YES	NO
6	Precision (95% CL of ±15%)	NO	NO	YES	YES
7	Linearity	NO	NO	YES	YES

5.3 PTFE AND UHMWPE WEAR: SUMMARY OF WATER VERSUS BOVINE SERUM JFA

The present simulator wear studies of PTFE and UHMWPE found that using water as the JFA did not duplicate any validation criteria and was not a valid JFA for hip wear simulations. The studies using bovine serum with greater than 60mg/ml protein concentration as the joint fluid were only slightly better. The ball size effect was the only clinical criterion to be duplicated for both PTFE and UHMWPE. However, this one validation criterion did not prove that bovine serum was adequate for hip wear testing.

The ISO draft standard (ISO 14242 part 1) requires 25% calf serum but a literature survey did not turn up any articles showing scientific support for this standard. Therefore, it could only be speculated that the intention of the ISO standard was to approximate physiological protein levels. However, as seen from table 3.2 the amounts of proteins in calf serum can vary from lot to lot. Thus, this standard was vague, unsupported and would have been better to have specified the amount of protein rather than a percentage of serum. Therefore, as part of this thesis the effect

of protein concentration in bovine serum on the wear of both PTFE and UHMWPE was examined.

5.4 PTFE AND UHMWPE WEAR: RESPONSE TO JFA PROTEIN CONCENTRATION

This was the first time the effect of protein concentration on the wear of either PTFE or UHMWPE was examined. The dose response curve showed three distinct phases for each material (fig 4.13 and 4.22). Phase A went from 0 to 10 mg/ml of protein concentration in the JFA. In water (0 proteins) wear-rates for both materials were very low and with the addition of proteins wear-rates of PTFE increased by more than 100% and the wear-rates of UHMWPE increased by more than 1800%. Phase B was the transition zone where the rate at which PTFE wear-rates were increasing was reduced. The transition zone for UHMWPE represented the peak or maximum wear-rates for each head size. In phase C the rate at which PTFE wear-rates increased was reduced by 80%. For UHMWPE phase C represented a reversal in wear-rates where as wear-rates in phase A showed a positive trend, in phase C the trend was negative. These results were thought to be due to protein precipitates, which acted as an interposed solid film, protecting the cup from wear. It has been shown that the amount of protein precipitate increases as the amount of proteins in the JFA increase (Liao et al, 1999 and Wang et al, 1999). Similarly, the wear-rate decreases as the amount of precipitate increases therefore protecting the cup from wear (Wang et al, 1999). In addition, a study by Scholes and Unsworth (2000) showed that proteins (in particular serum albumin) adsorb to the UHMWPE surface. It has also been shown that the amount of protein precipitate was time dependent, i.e. precipitate increased with time (Liao et al, 1999). Therefore with time, this increased

amount of precipitate would create an additional barrier to wear and decrease wear-rates. Two factors of the UHMWPE studies contributed to the reversed rate of wear in phase C in comparison to the continued increase in wear-rates of the PTFE phase C. First, the present studies of UHMWPE were typically run for 3 days before the serum was replaced and the PTFE studies ran for 1 day before serum was replaced. This was important because a study by Liao et al, (1999) showed that for UHMWPE/CoCr the amount of protein precipitate increased from 8% after 24 hours to 21% after 72 hours. Thus, the bovine serum in the 72-hour UHMWPE studies would have 13% more protein precipitated than the bovine serum in the 24-hour PTFE studies. Second, fluid evaporation in the UHMWPE investigations was replaced with bovine serum (more proteins), however due to the short duration of the PTFE studies no replacement for evaporation was necessary. Consequently, the original 13% difference between protein precipitate in PTFE and UHMWPE would increase. Therefore, instead of UHMWPE wear-rates continuing to increase after 10mg/ml of protein concentration as with PTFE, the wear-rates decreased. Thus the UHMWPE was better protected than PTFE by protein precipitate resulting in a greater decrease in wear-rate for UHMWPE.

The laboratory at Howmedica has duplicated the present dose response studies and found similar results. Polineni et al, (1997) examined three protein concentrations of bovine serum using 32mm diameter EtO and gamma sterilized UHMWPE. Their study used bovine serum with 4.4, 44 and 71mg/ml protein concentration demonstrating that the bovine serum with 44mg/ml protein concentration had the highest wear-rate. Wang et al, (1998 and 1999) repeated the dose response of the present study with one head diameter. Their investigation used

32mm non-sterilized and gamma sterilized GUR 4150 UHMWPE in 0, 5, 10, 22, 30, 44, 66 and 75mg/ml protein concentration. Both UHMWPE studies (sterilized and non-sterilized) showed the same three phases as seen with the present studies. They found that the wear-rate of gamma sterilized UHMWPE reached a maximum or transition from 10 to 30mg/ml of protein concentration. The wear-rate of non-sterilized UHMWPE changed from 5 to 10 mg/ml protein concentration as in the present studies with non-sterilized UHMWPE. The second dose response study by Wang et al, (1999) examined 32mm PTFE acetabular cups in the various protein concentrations and their results were similar to the present studies. Their PTFE acetabular wear-rates continued to increase with increased protein concentration and as in the present studies wear-rates fell within the clinical range at 10mg/ml of protein concentration. The wear-rates from the Howmedica study were slightly larger than the present studies, however remained within $\pm$ 20% of the clinical target and the PTFE/PE wear-rate ratio was the same. The reason for the increased wear-rates of the Howmedica study were due to the larger volume of JFA (450ml versus 250ml) and THR configuration (Anat versus Inv). Therefore, two separate laboratories confirmed that the clinical wear-rate magnitudes along with the clinical PTFE/PE ratio could be duplicated in-vitro using bovine serum with 10mg/ml of protein concentration as the JFA.

Two more UHMWPE studies reported increased wear-rates with decreased protein concentrations, however they reported that they observed pitting of the acetabular cup surface (Liao et al, 1999 and Sauer and Salehi, 1998) in low protein concentrations. The study conducted by Liao et al (1999) used small volumes of joint fluid analog (40ml) and zirconia heads. These two combinations were very

likely to increase the interface temperature and therefore change the nature of the mechanical wear. In a communication from Salehi (2001) it was noted that their report of pitting was an error. After close examination of the cup it was seen that the pitting was actually UHMWPE transferred back on to the articular surface. The pitting reported by Liao et al (1999) might also have been back transfer. There was no forthcoming data producing verifiable evidence that it was pitting and not back transfer of the polymer or some other artifact. In both the study by Wang et al, (1999) and the present study pitting was not seen on any UHMWPE cup in any protein concentration. For further clarification Dr. Fisher at Leeds was contacted for information on their UHMWPE cups tested in low protein concentrations (25% serum). He reported no pitting in sterilized or non-sterilized UHMWPE (Fisher, 2001). To verify that no pitting was seen in the present study, an UHMWPE cup tested in bovine serum with 26mg/ml of protein concentration (same concentration as Liao et al, 1999) was analyzed using a scanning electron microscope (SEM). Analysis showed smooth burnished surfaces in the worn area of the transition zone and center of the wear area with no evidence of pitting (fig 5.1).

The PTFE acetabular cups duplicated the clinical wear criteria in more than one protein concentration as the JFA. The clinical ball size effect of increased wear-rate with increased head diameter was duplicated using all protein concentrations of bovine serum. With a lack of clear quantified clinical PTFE debris the only conclusion that could be drawn was that the protein containing JFAs produced debris similar in appearance to Charnley's clinical description (Charnley et al, 1969). Therefore, two of the clinical wear criteria were duplicated using any concentration of bovine serum. However, the clinical VWI (6 – 10%/mm) was only duplicated

using bovine serum with 5, 10 and 21mg/ml of protein concentration and the clinical wear-rate magnitudes were only duplicated using 5 and 10mg/ml of protein concentration. Thus, the only studies to duplicate all the clinical wear criteria used bovine serum with protein concentrations of 5 and 10 mg/ml as the JFA.

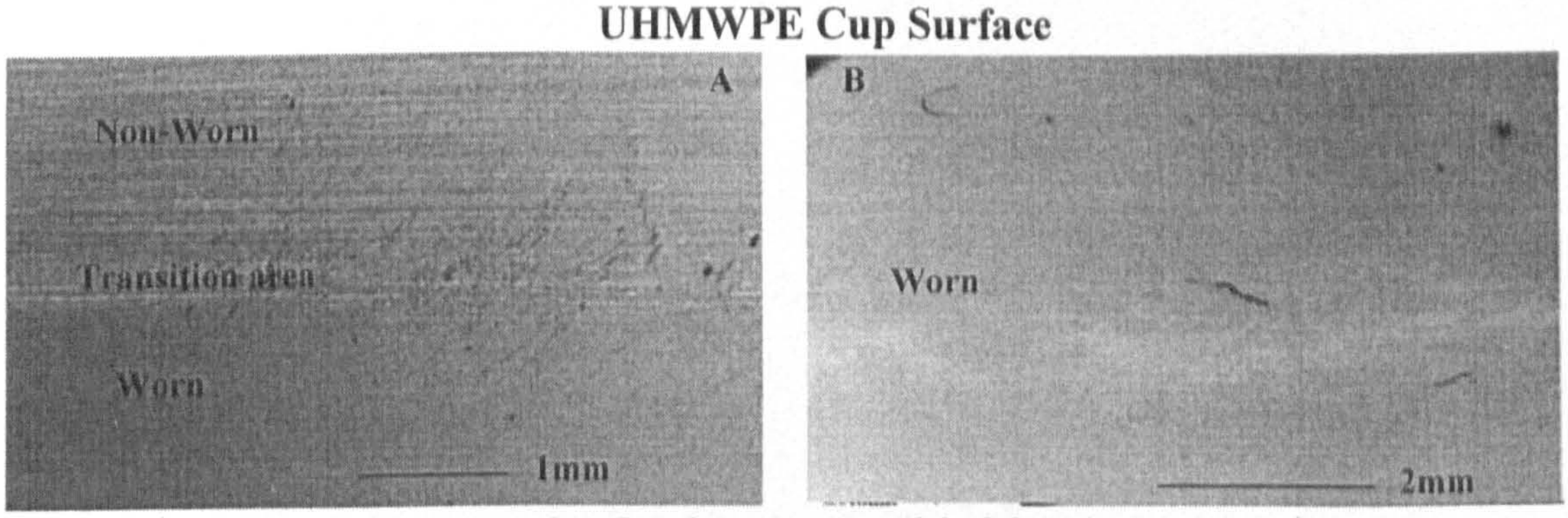


Figure 5.1 UHMWPE cup using bovine serum with 26mg/ml protein concentration with no evidence of pitting. A) transition zone showing non-worn (machine marks) and worn area (smooth burnished surface) B) high wear zone showing smooth burnished surface with no pitting.

The UHMWPE acetabular cups did not duplicate as many of the wear criteria in various protein concentrations as did the PTFE cups. The PTFE/PE wear-rate ratio was duplicated in the JFAs containing bovine serum with protein concentrations less than 21mg/ml (phase A, however no water and phase B). The ball size effect was duplicated in most of the protein containing fluids however, in phase C (> 10mg/ml of protein concentration) the distinction between wear-rates of the 22 and 28mm diameter cups was not as apparent. This was shown by the low VWI (<2%) in phase C for the 28mm diameter cups. The UHMWPE clinical VWI was accurately duplicated using 10mg/ml of protein concentration. In addition the wear-rate magnitudes for the three diameter cups and the debris characteristics were also duplicated using 10mg/ml protein concentration as the JFA. Thus, the only JFA to

duplicate all UHMWPE clinical criteria was bovine serum with 10mg/ml of protein concentration.

Table 5.2 Validation Criteria for Wear of PTFE dose response to protein concentration

*Within range for the 22mm but outside of the range of the 32mm diameter cups.

No.	*Wear Validation Criteria*	*Protein Concentration (mg/ml)*					
		0	*3*	*5*	*10*	*21*	*70*
1	Ball Size Effect	NO	YES	YES	YES	YES	YES
2	VWI (6-10%/mm ball dia)	NO	NO	YES	YES	YES	NO
3	Wear-rate Magnitude (±10%)	NO	NO	YES	YES	NO	NO
4	PTFE/PE (range 20-50)	NO	YES	YES	YES	*YES	NO
5	Debris comparable to in-vivo	NO	NA	NA	YES	YES	YES
6	Precision (95% CL of ±15%)	NO	YES	YES	YES	YES	YES
7	Linearity	NO	YES	YES	YES	YES	YES

Table 5.3 Validation Criteria for Wear of UHMWPE dose response to protein concentration

* Within range for the 32mm but not for the 28mm. ^ Within range for the 22mm but outside of the range for the 32mm. +Within range for the 32mm but outside of the range for the 22 and 28mm.

No.	*Wear Validation Criteria*	*Protein Concentration (mg/ml)*							
		0	*3*	*5*	*10*	*21*	*26*	*34*	*62*
1	Ball Size Effect	NO	YES	YES	YES	NO	YES	YES	YES
2	VWI (6-10%/mm ball dia)	NO	*	NO	YES	*	NO	*	NO
3	Wear-rate Magnitude (±10%)	NO	NO	NO	YES	NO	NO	NO	NO
4	PTFE/PE (range 20-50)	NO	YES	YES	YES	^	NA	NA	NO
5	Debris comparable to in-vivo	NO	NA	NA	YES	YES	NA	NA	YES
6	Precision (95% CL of ±15%)	NO	YES	YES	YES	YES	YES	+	YES
7	Linearity	NO	YES	YES	YES	YES	YES	YES	YES

5.5 PTFE AND UHMWPE WEAR: SUMMARY OF RESPONSE TO JFA PROTEIN CONCENTRATION

In bovine serum with a protein concentration of 10mg/ml this work was the first to duplicate the UHMWPE and PTFE clinical findings of:

1) wear-rate magnitude

2) ball size effect

3) volumetric wear index (VWI)

4) PTFE/PE wear-rate ratio and

5) debris morphology

As a result of this work the author proposed that the effects of increased protein concentration on the wear-rate of PTFE (wear-rate continues to increase) and UHMWPE (wear-rate decreases) was a result of protein precipitates forming an interposed solid film, thus protecting the acetabular cup from wear. The next studies were designed to examine the same wear response for PTFE with regard to JFA volume and THR configuration.

5.6 PTFE WEAR: EFFECT OF JFA VOLUME AND THR CONFIGURATION

This study showed that in addition to JFA protein concentration, the fluid volume affects PTFE wear-rates. This was in agreement with an UHMWPE study by Howmedica (Wang et al, 1999). The Howmedica UHMWPE study found a 1.3 fold wear-rate increase for 400ml volume compared to 200ml volume of JFA (65mg/ml of protein). They also found 70% less protein precipitation with the larger volume of fluid. The current PTFE studies showed wear-rates were approximately 2 times greater in 450ml versus 250ml of JFA (fig 5.2). This trend was consistent for JFA protein concentrations of 10 and 21mg/ml (fig 4.25). Thus, wear-rates increased with increased fluid volume.

THR configuration also affected PTFE wear-rates. Wear-rates in the Inv (cup on bottom) configuration were 17 to 26% greater than wear-rates in the Anat (cup on top) configuration (fig 4.25 and 5.2). This trend was consistent for JFA protein concentrations of 21 and 22mg/ml protein concentration. Bowsher and Shelton

(2000) found an 18% wear-rate increase in the Inv versus the Anat THR configuration in a study of gamma-sterilized UHMWPE. Their study used 600ml (Anat) and 400ml (Inv) of 25% bovine serum as the JFA. They also found twice the frictional torque and a 2°C rise in JFA temperature for Anat versus Inv THR configuration. This was possibly due to JFA starvation through cavitation or low fluid levels. Lu et al (1998) also found an increase in temperature at the UHMWPE/CoCr THR interface in the Anat configuration. It was of interest to note that the Anat configuration had a larger volume of JFA than the Inv THR configuration, similar to the study by Bowsher and Shelton (2000). The study by Lu et al (1998) also found patchy films of protein precipitate on the heads, however wear was not measured in this study. Thus, the increased temperature of the Anat configuration would lead to more precipitates and less wear.

A THR configuration study of UHMWPE by Wang et al (1997) showed contrary wear results to that of Bowsher and Shelton (2000). They found no discernable difference in wear for either THR configuration using gamma sterilized UHMWPE (similar UHMWPE as used by Bowsher and Shelton, 2000). However EtO sterilized UHMWPE wear-rates were 30% lower in the Inv versus Anat configuration (opposite to the current studies and the study by Bowsher and Shelton, 2000). Both laboratory studies of UHMWPE used a MTS simulator, similar fluid volumes and similar JFA protein concentrations, however Wang et al (1997) held the JFA temperature constant at 37°C whereas Bowsher and Shelton (2000) did not. The constant temperature would result in the same protein precipitation in both configurations and therefore similar wear as in the gamma sterilized UHMWPE. As for the EtO sterilized UHMWPE there was no explanation as to the difference.

Whether this was a material property was uncertain; therefore, it was unknown why the Inv showed less wear than the Anat configuration.

Two THR Anat configurations were also examined for their effect on wear of PTFE. The oblique angle (cup face inclined at 67° to the vertical load axis) was incorporated in the Anat configuration to reproduce the head and cup angles in the body. Wear-rates in the Anat/horiz THR configuration (cup face perpendicular to the vertical load axis) were 10% greater than in the Anat/obliq configuration. This difference was found for both JFAs of 21mg/ml and 22mg/ml protein concentration. However, this was not as large an effect as the Inv versus the Anat configurations (fig 4.17).

A second PTFE dose response study was initiated as part of an international multi-laboratory project. This study utilized one head size (32mm) in the Anat/obliq THR configuration with 450ml of JFA and tested protein concentrations of 17, 34 and 62mg/ml. These protein concentrations showed a similar dose response to wear-rates as the previous Inv, 250ml volume of JFA studies, i.e. increased wear-rates with increased protein concentration. However, the wear-rate magnitudes were much greater in the large volume Anat studies (fig. 5.2). Another participant of the multi-laboratory study used identical conditions and produced a similar wear-rate trend of 86mm^3/Mc per mg/ml of protein concentration (Clarke et al, in press). The wear-rate magnitude for 34mg/ml of protein concentration JFA (circled on fig 5.2) was omitted due to the unusually high wear-rate, 4 other participants did not show this large wear-rate for the same protein concentration regardless of configuration.

Thus, wear-rates of PTFE increased with increasing protein concentration, but wear-

rate magnitudes were greater in the Anat/obliq large volume than in the Inv small volume.

Only three of the validation criteria were applied to these volume and THR configuration PTFE studies (wear-rate magnitude, linearity and precision; table 5.4). Linearity and precision were very good as expected with PTFE in serum. However, the PTFE wear-rate magnitudes were at least 1.5 times greater than clinical in the 450ml volume of JFA in both Inv and Anat configurations (table 5.4). The PTFE/PE wear-rate ratio was not applicable since studies of UHMWPE were not conducted. Additionally, with only one head size, the ball size effect was not applicable. These studies will be initiated in the future. Wear debris was not analyzed at this time but was stored for future analysis.

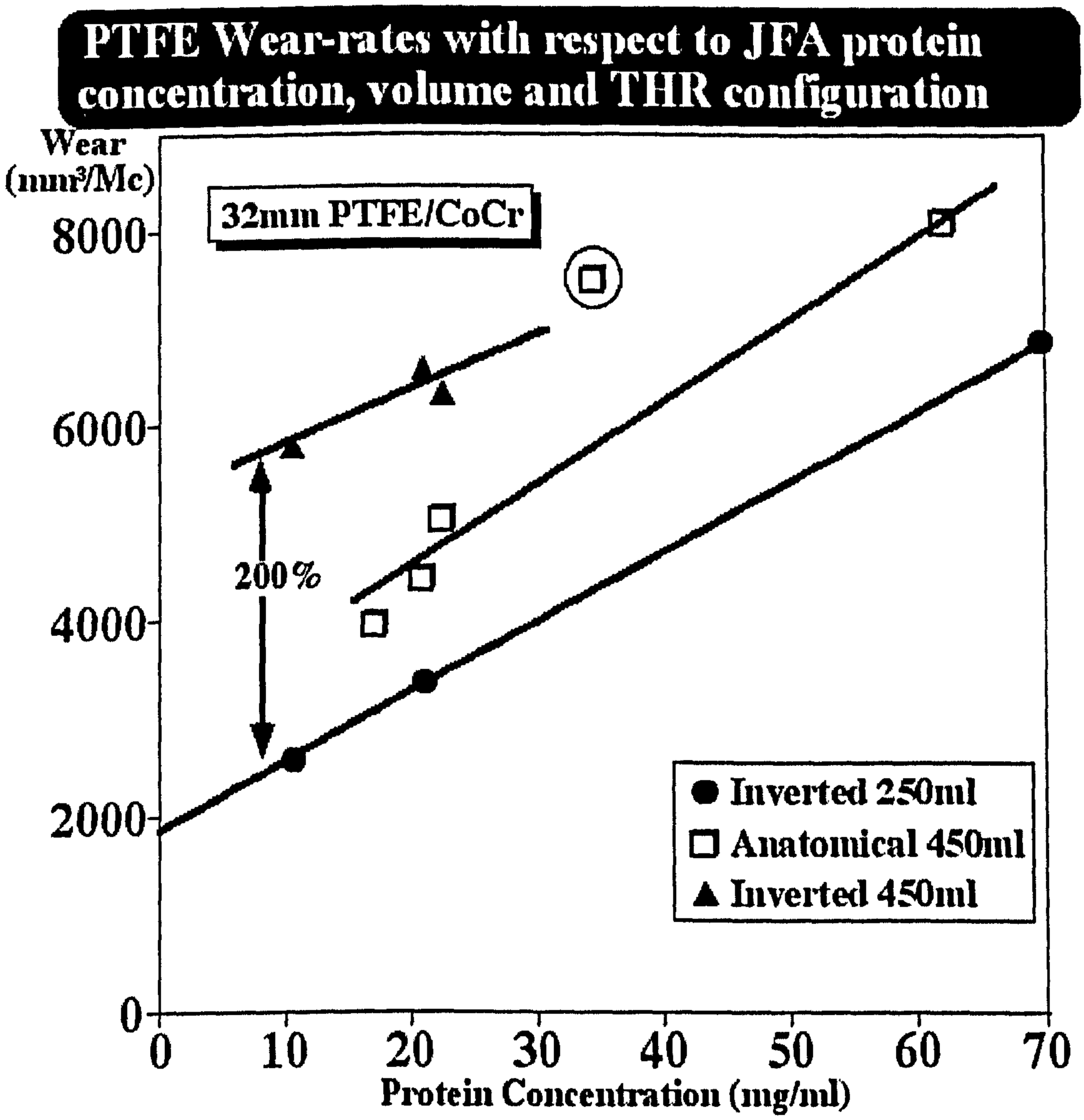


Figure 5.2 PTFE/CoCr wear-rates with respect to serum protein concentration of the JFA. In 34mg/ml of protein concentration the Anat TIIR configuration showed an abnormally high wear-rate (circled) in comparison to 4 studies using the same material and protein concentration conducted at outside laboratories. Therefore this point was omitted from the slope calculation.

Table 5.4 Wear Validation Criteria with respect to THR configuration

All studies used 450ml of JFA. Identical criteria were met for PTFE regardless of THR configuration or serum type. Debris analysis was not completed because of the lack of duplication of the wear-rate magnitude. BCS = bovine calf serum, ACS = alpha calf serum, NA = Not Applicable

No.	*Wear Validation Criteria*	*PTFE 62, 34, 21 and 10mg/ml Protein BCS*	*PTFE 22mg/ml Protein ACS*
1	Ball Size Effect	NA	NA
2	VWI (6-10%/mm ball dia)	NA	NA
3	Wear-rate Magnitude (±10%)	NO	NO
4	PTFE/PE (range 20-50)	NA	NA
5	Debris comparable to in-vivo	NA	NA
6	Precision (95% CL of ±15%)	YES	YES
7	Linearity	YES	YES

The amount of protein precipitate was the unifying concept from these findings. In all studies showing higher wear-rates there was a corresponding decrease in protein precipitate in comparison to the lower wear-rate studies. This affect was very distinct for the current PTFE studies of volume. The affect of THR configuration was clear with regards to Anat versus Inv but between the two Anat configurations the correlation was not as strong. This was true of the two UHMWPE studies by Wang et al (1999) and Bowsher and Shelton (2000). Thus, the effect of THR configuration for UHMWPE needs further study.

5.7 PTFE WEAR: SUMMARY OF AFFECT OF JFA VOLUME AND THR CONFIGURATION

The study of PTFE wear-rates in two volumes of JFA and three different THR configurations showed that:

1. Wear-rates were 200% greater using 450ml versus 250ml of JFA in the Inv THR configuration
2. Wear-rates were 17 to 26% greater in the Inv configuration versus the Anat configuration using the same volume of JFA.

3. Wear-rates in the Anat/horiz configuration were 10% more than in the Anat/obliq configuration (fig 4.25).

CHAPTER 6. CONCLUSIONS

This work was the first to duplicate five clinical wear criteria for both UHMWPE and PTFE acetabular cups in a hip wear simulator. This thesis examined the affect of protein concentration in the JFA on wear of the two polymers. This showed that bovine serum with 10mg/ml of protein concentration duplicated all five of the clinical wear criteria. The wear criteria were based on the clinical findings of: ball size effect, VWI of 5 to 10 percent increase in wear for each millimeter of increased head diameter, clinical wear magnitudes, PTFE/UHMWPE wear-rate ratio and debris morphology.

The clinical PTFE acetabular cup wear-rate criteria of ball size effect and VWI were duplicated in all experiments with bovine serum as the JFA. The UHMWPE acetabular cup wear-rates only duplicated both criteria using 10mg/ml of protein concentration as the JFA. In contrast, neither criterion was duplicated for PTFE or UHMWPE wear-rates when water (0 protein) was used as the JFA.

This thesis was the first to duplicate the clinically relevant PTFE/PE wear-rate ratio. Several protein concentrations of bovine serum duplicated the PTFE/PE ratio, however only the 10mg/ml of protein concentration duplicated the clinically relevant wear-rate magnitudes for PTFE and UHMWPE. The water JFA studies of both materials did not duplicate either clinical wear criteria.

The UHMWPE debris produced in any bovine serum concentration was within the clinical description. However the percentage of round particles to total amount of debris was reduced to the clinically relevant 75% at 10mg/ml of protein concentration. The PTFE debris produced in all percentages of bovine serum as the

JFA was similar to the clinical description. The debris produced from both PTFE and UHMWPE in water was not at all comparable to clinical debris.

This work proved that the JFA has a significant affect on THR wear in the hip joint simulator. The nature of the wear to dose response of protein concentration showed unequivocally that proteins cause wear in polymers. Wear-rates were lowest for both PTFE and UHMWPE in the non-protein JFA of water. Wear-rates of both materials sharply increased (>100% for PTFE and >1800% for UHMWPE) as the bovine serum to water dilution was increased. However, after reaching a transition point (10mg/ml protein concentration), the wear trends changed. The wear trend for PTFE remained positive but the slope was reduced by 80 percent. UHMWPE wear-rates decreased resulting in a negative wear trend. This change in wear trend was due to protein degradation. The protein precipitate acted as an interposed solid film thus, protecting the acetabular cup and decreasing wear-rates. The amount of protection was dependent on the amount of precipitate, and the amount of precipitate was dependent on the initial amounts of protein. The decreased wear-rates of UHMWPE suggested an additional protection in comparison to PTFE. Two factors contributed to this additional protection. First, the increased test duration between change of JFA (72 versus 24 hours for UHMWPE and PTFE respectively) and second, the addition of bovine serum (more proteins) to compensate for evaporation during the UHMWPE tests.

This thesis also showed that the volume of JFA influenced wear-rates of PTFE. The wear-rates of PTFE in 450ml of bovine serum as the JFA were 200% greater than in 250ml of the same JFA. This suggested that fewer proteins were degraded for the larger volume thus, allowing for the proteins to increase wear.

The THR configuration (Inv, Anat/horiz and Anat/obliq) of the PTFE acetabular cup in the simulator was also shown to be important with respect to wear-rates. PTFE wear-rates in the three THR configurations ranked Inv > Anat/horiz > Anat/obliq with 26% greater wear in the Inv configuration. It was suggested that the cavitation of the serum in the Anat configuration would cause intermittent periods of dry wear thereby increasing friction and heat at the THR interface. This in turn caused an increase in protein precipitation affording protection to the acetabular cup from wear. It was also speculated that the small difference in wear between Anat/obliq and Anat/horiz might be due to increased friction however, more study in this area is needed.

In conclusion this thesis has shown that the JFA has a significant tribological influence on the in-vitro wear of polymers. These findings were significant because they proved that the laboratory could duplicate the clinical outcome for two known materials. These studies have taken the first steps in the understanding of effect of proteins on the wear of polymers and have shown the need for further evaluation.

CHAPTER 7. NEED FOR FURTHER WORK

This work has shown that proteins influence the wear-rate of polymers in the hip simulator, however it was not known whether only one or many of the proteins in bovine serum were responsible for this wear. Also, the exact nature of the relationship between wear and proteins was not known. Why did the polymer material back transfer to the cup surface in water and when proteins were added this back transfer disappeared and wear was promoted? Was there a particular protein, which promoted this process or was it many or all of the proteins acting together? From the experiences in this laboratory and others there was evidence of serum coating on the debris, suggesting a possible interaction between the serum proteins and debris causing the back transfer not to form. The study by Scholes and Unsworth (2000) showed protein adsorbed to the UHMWPE surface. Their study did not show whether the protein adsorption was dependent on the protein concentration. Therefore further work should examine the dose response of protein adsorption to protein concentration and also examine the affect of wear-rates on the adsorption of proteins. In addition their finding that albumin was the most abundantly adsorbed protein shows that there may be an individual protein that can increase or decrease wear-rates. Therefore, further work such as a dose response to individual proteins should focus on this question in order to develop an understanding of the protein/wear-rate process. In addition, further work should focus on whether these same results can be found for other polymers, such as Delrin™ and polyester. There also should be an investigation on the wear of rigid on rigid THR and the protein affect on wear of these systems.

For the first time clinical results were duplicated in the laboratory for two known polymer materials using a biaxial wear simulator with 10mg/ml of protein concentration. The current ISO standards call for a 3-axis simulator and 25% bovine serum, however, it has not been shown that the same clinical results from this work have ever been duplicated using the ISO standards. A comparison study by Essner et al, (2000) examined the same UHMWPE on a 3-axis and 2-axis simulator and showed that the 3-axis produced up to 80% less wear-rate than the 2-axis and only the 2-axis simulator duplicated clinical wear-rate magnitudes. Unfortunately, the variables were not consistent between machines, i.e. differences included serum, load and speed, and therefore a true comparison was not possible. In order to evaluate machine similarities and differences an international multi-laboratory study has been initiated by academia and industry. The studies are using PTFE to evaluate machine differences using various serum concentrations including the 25% bovine serum concentration required by the ISO standard. These type of studies will help define whether there is a difference between biaxial and 3-axis simulators and whether protein effects are the same regardless of machine type.

The matter of protein precipitate was also of much interest. Two laboratories using UHMWPE have measured increased protein precipitate with increased protein concentration and smaller volumes. However the cause of the increased precipitate and the reason for the wear protection was unknown and has raised many questions. It has been speculated that protein precipitates form because of high temperatures at the cup/ball interface. It has been shown that wear-rates decreased with increased protein concentration, smaller volumes of JFA and when the THR was configured anatomically. Was this caused by increased temperature and more precipitate or was

there another phenomenon causing this decrease in wear-rates? Also, the time factor for UHMWPE was speculated to be further reason for decreased wear-rates. Would the wear-rate of UHMWPE increase with shorter time between replacement of the JFA? Therefore, further study should focus on the surface temperature of the THR combinations, amount of protein precipitate in PTFE, the volume and THR configuration affect on UHMWPE and the time dependence of protein precipitation formation.

Another area to be examined was the PTFE wear-rate dose response to protein concentration with respect to the larger volumes. This work showed a similar dose response to wear-rate in the higher protein concentrations using 450ml of JFA in the Anat configuration as in the first studies using 250ml and Inv. As part of an ongoing multi-laboratory study this protein dose response will be continued for the lower protein concentrations using the same fluid volume (450ml) and Anat/obliq THR configuration.

These studies have gained a new insight into the significance of the JFA on the wear of PTFE and UHMWPE and shown that clinical results can be duplicated in a hip joint simulator. Further work on these affects is necessary to fill in the unknowns but the present work has given the orthopaedic research community more definition to the questions that should be asked.

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Appendix I
Serum Protocol

ALWAYS WEAR GLOVES

NEVER ALLOW PRESSURE TO GO HIGHER THAN 20psi

Mix all batches according to the serum information sheet

1) Add appropriate number of bottles of serum, deionized water and additives if called for in 5 gallon pressure cooker
2) Stir vigorously for 1 minute
3) Put pressure cooker lid on and secure by turning till it locks
4) Attach 0.2μm filter to hose on pressure cooker
5) Attach air compressor line to pressure cooker
6) Turn valve till serum begins to flow
7) Label bottles according to type of serum and date
8) Store serum in refrigerator for no more that 6 weeks.
9) Forward and Back Flush filter with Deionized water
10) Store filter in refrigerator
11) Clean pressure cooker with standard dishwashing soap

Discard any unused and old serum in serum waste collection containers

Appendix II
Cleaning Protocol

ALWAYS WEAR GLOVES

1) Clean and rinse hip or knee liners with specially-designated nylon brush under running tap water, brushing vigorously to remove all attached debris.
2) Place liners in plastic container and cover with a solution of 2% Alqinox liquid in water - place in ultrasonic bath (filled with de-ionized water) and run for 10 min.
3) Rinse liners under running tap water and place in clean plastic container.
4) Cover liners with de-ionized water - place in ultrasonic bath and run for 10 min.
5) Place liners in a new clean plastic container and repeat #4 .
6) Dry liners using air jet to remove surface moisture.
7) Place liners in a new clean plastic container and cover with Ethyl Alcohol 200-proof - place in ultrasonic bath and run for 10 min.
8) Repeat #6.
9) Place liners in vacuum dessicator - vacuum for 30 min.
10) Follow weighing protocol.

Appendix III
Assembly Protocol for Polymers

ALWAYS WEAR GLOVES

A) Polymer Inverted 250ml Assembly

All parts are listed on station information sheet and should be put together accordingly

1) Mount adapter in the stainless steel cup holder
2) Insert polymer cup into adapter
3) Very lightly Vaseline outside O-rings on stainless steel cup holder.
4) Insert Plexiglas chamber on stainless steel cup holder
5) Insert screw into base of stainless steel cup holder
6) For first assembly mount head on taper cone, all subsequent events head remains on taper cone
7) Mount to simulator, see Simulator Protocol

B) Polymer Inverted 450ml Assembly

All parts are listed on station information sheet and should be put together accordingly

1) Screw stainless steel cup holder into stainless steel anatomical base plate
2) Insert polymer cup into stainless steel cup holder
3) Very lightly Vaseline outside O-rings on stainless steel anatomical base plate
4) Insert anatomical Plexiglas chamber on stainless steel anatomical base plate
5) For first assembly mount head on taper cone, all subsequent events head remains on taper cone
6) Mount to simulator, see Simulator Protocol

C) Polymer Anatomical 450ml Assembly

All parts are listed on station information sheet and should be put together accordingly

1) For first assembly mount head on anatomical taper cone, all subsequent events head remains on anatomical taper cone
2) Screw anatomical taper cone/head assembly into stainless steel anatomical base plate
3) If polymer cup is manufacturer designed prosthesis go to step 4, if cup is custom with 3 holes for taper go to step 6
4) Insert polymer acetabular cup into UHMWPE cup adapter, lining up numbers on cup and adapter.
5) Place hold down ring on adapter and screw into place
6) Line up arrow on polymer cup or UHMWPE adapter with circle on cup taper cone and Insert
7) Very lightly Vaseline outside O-rings on stainless steel anatomical base plate
8) Insert anatomical Plexiglas chamber on stainless steel anatomical base plate
9) Mount to simulator, see Simulator Protocol

Appendix IV
Simulator Protocol 9 and 12B Channel

Inverted 250ml/450ml and *Anatomical 450ml*

ALWAYS WEAR GLOVES

All specimens are pre-assembled according to the station information sheet.

1) Load stainless steel cups/*stainless steel anatomical base plates* onto the stations of the simulator according to the station information sheet.
2) The taper cones (heads attached)/*Cup tapers* (cups attached) are placed into the self-centering device and tightened down.
3) Each chamber is filled with approximately 100ml/*200ml* (enough to cover the cup/ball interface).
4) The head or cup is lowered onto the cup/*head* with an approximate 1 to 2mm gap to ensure that the loads do not bottom out.
5) Verify that loads are on (orange knobs pointing to on)
6) Push green start button (hydraulics) on front panel of simulator
7) Reset counters to 0
8) Verify DC motor switch is in 'on' position
9) Turn on computer
10) Type in hip at the prompt
11) Choose "Load Configuration"
12) Highlight "Paul.cfg" and enter
13) Choose "Test Parameters"
14) Verify that speed is set at 1Hz, max load is set to 2000N and min load is set to 200N
15) Enter number of cycles for event
16) ESC
17) Choose "Run Test"
18) Enter experiment number (3 digits)
19) Enter event number (3 digits)
20) Simulator will start
21) Verify output loads on screen match input loads, if not press F10 to redraw, if still not correct hit Esc and call Victoria Good, Dr. Clarke or Shore Western.
22) Examine height of swash blocks to make sure that none are more than 3cm from simulator base. If answer is yes continue, if answer is no go to step 27.
23) If yes then press F1 and turn off DC motor switch.
24) Turn the load control knob for that station off and readjust the lowered head or cup to within 2mm of corresponding cup or head.
25) Turn load knob on, press F3 and turn on DC motor switch
26) Repeat step 21 until answer is no
27) After approximately 2 minutes of smooth running pause test to fill chambers
28) Press F1 and turn off DC motor switch
29) Fill chambers to within 5mm of top of Plexiglas
30) Press F3 to resume and turn on DC motor switch
31) Enter new event information in simulator log book

Appendix V
Weighing Protocol
ALWAYS WEAR GLOVES

Turn on the Sartorius Microbalance at least 10 minutes before weighing specimens
While waiting for balance, turn on computer and open LLUMC database

1) Choose Events New
2) Choose Experiment from list
3) Choose event log sheet
4) Enter data from simulator log book in event log sheet
5) Print and save event log sheet
6) Choose event weight sheet
7) Enter name, room temperature and type of weigh information (initial wear, wear or soak)
8) Arrange specimens according to the order of the weigh sheet on lint free paper
9) Press door button to open glass door on balance
10) Place first specimen in the center of the plate
11) Close glass door
12) "Click" the appropriate bubble on the weigh sheet for the specimen being weighed
13) When the microbalance shows "g" press send/print button to send data to weigh sheet on computer
14) If it takes longer than 1 minute a timeout error will appear, click on ok and go to step 12.
15) Open glass door and replace specimen on lint free paper
16) Insert next specimen
17) Repeat until all specimens have been weighed 4 times sequentially
18) Remember to save data every 6 to 12 specimens.
19) When finished print data and save
20) Return specimens to dissector
21) Put event log sheet and weigh sheet in appropriate binder.
22) Turn off balance and cover
    Exit program and turn off computer

APPENDIX VI
PTFE Wear Debris Analysis
Sample Preparation, Processing and Filtration

NaOH Digestion

Application:

This digestion method is only utilized for the isolation of PE and PTFE wear samples which are in Bovine Serum Albumin (BSA) and Clinical retrievals when only PE debris is of interest. The NaOH digests the proteins allowing for subsequent separation and extraction of the debris from the organic material. The subsequent washings aid in the removal of NaOH, glycerol, and organic contaminants.

Procedures:

1. Add 5N NaOH to the debris sample in a ratio of 5 ml of NaOH to 1 ml of 100% BSA. If the BSA is less than 100% then add the equivalent proportion of NaOH to sufficient BSA to total 5 or 6 ml (table A-I and A-II)
2. Mix the debris sample and NaOH using a Vortex for 30 seconds.
3. Incubate sample and NaOH for 12 hours at 65°C in a shaker bath.

Washing

Procedures:

4. Siphon fluid from vial down to about 1 ml.
5. Refill vial with 50% Ethanol to 10 ml.
6. Place samples in shaker bath for 1 hour at 55°C.
7. Centrifuge at about 6,000rpm for 10-15 minutes.
8. Repeat steps 1 through 4 until the fluid is clear and appears clean.

APPENDIX VII
POLYETHYLENE DEBRIS - SAMPLE PROCESSING

Polyethylene debris processing is divided into the following five parts:

- Digestion
- Separation
- Extraction
- Washing
- Filtration

NaOH Digestion

1. Add 5N NaOH to the debris sample in a ratio of 5 ml of NaOH to 1 ml of 100% BSA. If the BSA is less than 100% then add the equivalent proportion of NaOH to sufficient BSA to total 5 or 6 ml (table A-I and A-II).
2. Mix the debris sample and NaOH using a Vortex for 30 seconds.
3. Incubate sample and NaOH for 12 hours at 65°C in a shaker bath.

Separation

4. Add 4 ml of Glycerol to the sample.
5. Vortex sample for 30 seconds to mix the sample and Glycerol.
6. Centrifuge at approximately 6,000 or 7,000 rpm for 8 hours or more.

Extraction

7. Triple rinse the appropriate number of vials with triple filtered distilled water (TFDW).
8. Add 1 to 2 ml of filtered 50% or 100% Ethanol to the top of the sample.
9. Remove the material at the Top of the Glycerol and place the sample to a clean vial.
10. Add 50% Ethanol so that the total fluid in the sample vial is 10 ml.
11. Place samples in shaker bath for 1 hour at 55°C.
12. Centrifuge at about 6,000rpm for 10-15 minutes.

Washing

The subsequent washings aid in the removal of NaOH, glycerol, and organic contaminants. Procedure 2 is a modified method allowing for supplemental digestion of proteins.

<u>Procedure 1</u>:

13. Siphon fluid from vial down to about 1 ml.
14. Refill vial with 50% Ethanol to 10 ml.

15. Place samples in shaker bath for 1 hour at 55°C.
16. Centrifuge at about 6,000rpm for 10-15 minutes.
17. Repeat steps 1 through 4 until the soaks are clean.

Procedure 2:

18. Siphon fluid from vial down to about 1 ml.
19. Add 5ml of 5N NaOH and 5ml of 100% Ethanol.
20. Mix the sample using a Vortex for 30 seconds.
21. Incubate sample for 4 hours at 65°C in a shaker bath.
22. Centrifuge at about 6,000rpm for 15 minutes.
23. Repeat steps 1 through 5 until the fluid is clear, then continue with washing procedure 1 as necessary.

Filtration

24. Mount the microanalysis filtration apparatus (MFA) on a flask or manifold assembly.
25. Insert a 25mm filter, of the pore diameter to be used for filtering the samples, into the MFA.
26. Run TFDI water through the MFA to clean it.
27. Repeat step 3 with 95% ethanol.
28. Remove and discard the filter.
29. Place the sample's pre-weighed 25mm filter in 95% ethanol for about 30 seconds or more and then insert the filter into the MFA.
30. Apply 2 to 5ml of 95% ethanol through the MFA to prime the filter.
31. After the ethanol has passed through the filter add about 5ml of the sample at a temperature of 55°C. Once this is filtered, add the remaining 5ml or add smaller amounts maintaining 5ml in the MFA until all of the sample is filtered.
32. After the sample is filtered three rinse cycles are performed as follows with all fluids maintained at 55°C:

- 10ml of TFDI water.
- 10ml of 50% ethanol.
- 10ml of 95% or 100% ethanol.

33. After the three rinses are finished, carefully and quickly remove the filter from the MFA and place it (filtrate side up) into a plastic weighing boat.
34. Place filters into a dessicator and desiccate for 12 hours or more.

The filters can be weighed and then prepared for SEM or they can be prepared for SEM without weighing. However, once the filters are mounted for SEM they cannot be weighed.

Table A-I: Sample and NaOH volumes for PE processing of different concentrations of serum (%serum) with dehydration of sample.

% Serum	*Sample Volume (ml)*	*NaOH Volume (ml)*
<5	5-10	5
5-10	4-8	5-6
10-20	6	6
20-50	2	5
>50	1	5-6

Table A-II: Sample and NaOH volumes for PE processing of different concentrations of serum (%serum) without dehydration of sample.

% Serum	Sample Volume (ml)	NaOH Volume (ml)
<5	5	1
5-10	4	2
10-20	3	3
20-50	2	4
>50	1	5-6