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Alnaimat, Feras Adnan; Shepherd, Duncan; Dearn, Karl

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The effect of synthetic polymer lubricants on the friction between common arthroplasty bearing biomaterials for encapsulated spinal implants

F.A. Alnaimat, D.E.T. Shepherd* and K.D. Dearn

Department of Mechanical Engineering, School of Engineering, University of Birmingham, Edgbaston, Birmingham, B15 2TT, UK

Abstract

There are two major problems with ball and socket artificial joints; the migration of wear particles inducing an inflammatory response, causing toxicity, osteolysis and subsequent implant loosening; and the poor tribology between interstitial or synovial fluid and device's articulation surfaces. Experiments have been conducted to assess the potential of the promising bio-lubricant polyvinyl alcohol (PVA) at different concentrations with a range of materials (combinations of CoCr, UHMWPE and PEEK). Tests were conducted on a pin-on-disc tribometer and results were compared against Ringer's solution and bovine calf serum. The highest friction coefficient was for CoCr/CoCr for all lubricants. The lowest, and superlubricity was measured for UHMWPE/CoCr (a friction coefficient of 0.009) with 20 g/100ml PVA (PVA-C).

KEYWORDS

Friction; Polyvinyl alcohol; Spine; Synthetic lubricant

1. Introduction

Ball and socket articulations are widely used for joint replacement implants such as those for the spine and hip. A significant problem that can arise in joint replacement implants is the generation and subsequent migration of wear particles resulting from the combined rolling and sliding between the socket against the ball [1-3]. In this case, when the wear debris reaches the surrounding tissues it can cause problems such as inflammation, toxicity, osteolysis and then implant loosening [4, 5]. The size and shape of generated debris are also likely to change throughout the life of an implant, further complicating these problems [6].

An improved lubrication regime can help to minimise the generation of these wear particles. It has been proposed that joint replacement implants might benefit from a capsule which can seal an artificial lubricant within the joint to reduce friction and wear, whilst simultaneously preventing debris from migrating [7, 8]. This study is the first part to developing an encapsulated disc replacement (figure 1) where the friction of various biomaterials and lubricants was investigated. A commercially available example of this is the Bryan disc (Medtronic Sofamor Danek, Inc., Memphis, TN), which relies on the idea of using encapsulation with saline solution as a lubricant. The disc has shown promising clinical results [9-11]. It seems reasonable therefore

that a biocompatible polymer based lubricant could be used with joint implant replacements. Synthetic polymer lubricants have been used since the 1960s [12]. More recently, Kobayashi et al. [13] have used polyethylene glycol (PEG) as a synthetic polymer lubricant with synovial fluid to lubricate knee replacements. In natural synovial joints, the lubricant is synovial fluid and the lubricity of this fluid is similar to water due to the high shear rates inside these prostheses [14]. In disc replacements, the implants are likely to be lubricated with interstitial fluid which has low viscosity and a value for lubricity between Ringer's solution and bovine calf serum, both of which are lower than synovial fluid [15, 16]. This relatively low viscosity is likely to lead to challenging tribological conditions for any spinal implant device that relies on bearing surfaces. If encapsulation is to be successful in spinal implant devices, there will be a need to find an appropriate synthetic lubricant to reduce friction and wear, and increase implant durability.

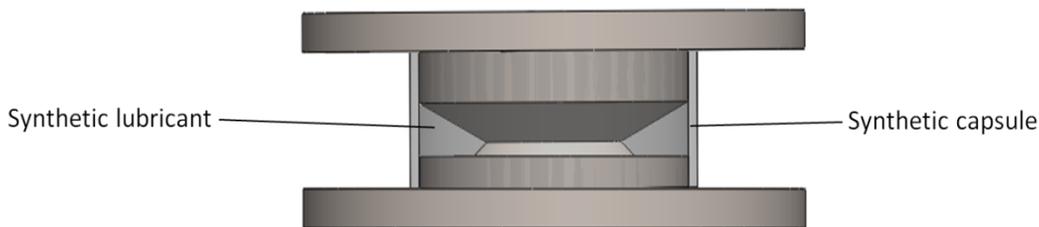


Figure 1: Artificial disc prosthesis with synthetic capsule and lubricant.

Polyvinyl Alcohol (PVA) has been shown to be a promising lubricant because of its physical properties, the lubricity of the solution and its biocompatibility [17, 18]. PVA is one of the water soluble polymers which comes as solid white granules or as a powder. The solubility of PVA depends on the degree of hydrolysis in water [19]. PVAW40/140 with 5% and 10% was studied as a synthetic lubricant for synovial joints [18]. PVA and polyvinylpyrrolidone (PVP) have been used in artificial tear fluids with appropriate viscosity to lubricate dry eyes [20]. PVA hydrogels have been developed as an artificial membrane in contact lenses [21]. The material has also been used as a hydrogel membrane for the encapsulation of implanted Langerhans islets cells in the pancreas to immunoisolate them from the immune response [22, 23]. It has also been used by Jiang et al. [24] as a one piece tricuspid heart valve made entirely from PVA hydrogel. PVA was also been used as an artificial articular cartilage to repair joint surfaces [25], as well as the film coatings for pharmaceutical and dietary supplement tablets where a barrier is required to protect tablets from moisture and other contaminants. PVA is not considered as carcinogenic and there are no reports related to the chronic toxicity and carcinogenicity when it is taken orally [26].

The aim of this study was to compare the frictional behaviour of the most widely used materials in artificial discs with potential synthetic lubricants for use in encapsulated implants. PVA of different viscosities, Ringer's solution and bovine calf serum were used as the test fluids. In addition, lubricant viscosity, modified by PVA concentration was also assessed.

2. Materials and Methodology

2.1. Tribological specimens

The test samples comprised an upper pin and a lower disc. The upper pin samples were machined to 8 mm diameter and 15 mm length, the lower disc samples were machined to 79 mm diameter and were 5 mm in thickness. The tested materials were cobalt chrome (CoCr) alloy Haynes 25/L605 [27] supplied by (Dynamic Metals Ltd, Hemel Hempstead, UK), ultra high molecular weight polyethylene 1000 (UHMWPE), unreinforced polyether ether ketone (PEEK) 450G (supplied by Direct Plastics Ltd, Sheffield, UK). These materials have been previously tested for spinal implants [28, 29].

All of the tribological specimens were polished using a Buehler Alpha Grinder Polisher (Buehler Ltd, Illinois, USA). The surface roughness of each pin and disc was measured three times using an Alicona G4 InfiniteFocus (Alicona, Raaba, Austria). The results of the surface roughness measurements are shown in table 1.

Table 1: Surface roughness of test specimens.

Material	Mean Roughness \pm Standard Deviation (μm)
CoCr Disc	0.92 \pm 0.05
CoCr Pin	1.11 \pm 0.02
UHMWPE Pin	0.82 \pm 0.11
PEEK Disc	0.87 \pm 0.06
PEEK Pin	1.26 \pm 0.04

2.2. Lubricants

Three types of lubricant were used in this study, although it should be noted that several concentrations of polyvinyl alcohol (PVA) were tested. They were:

- Ringer's solution – prepared with a 1.2 g Ringer's solution tablet (Oxoid Ltd., Hampshire, UK) in 500 mL of distilled water.
- Bovine calf serum – prepared by defrosting the serum (SeraLab, West Sussex, UK) over 24 hours at 5 °C before diluting with 20 g/L deionized water and adding 0.3 g of sodium azide powder to minimize bacterial growth (Sigma-Aldrich, MO, USA) .
- PVA – prepared from powder with a molecular weight of 31,000-50,000 g/mol, 98-99% hydrolysis. The PVA was prepared at 80 °C and the concentrations of PVA were:
 - PVA A: 4 g per 100 mL of distilled water.
 - PVA B: 10 g per 100 mL of distilled water.
 - PVA C: 20 g per 100 mL of distilled water.
 - PVA D: 30 g per 100 mL of distilled water.

2.3. Viscosity Measurements

The viscosities of the different PVA lubricants were measured using an AR-G2 cone on plate rheometer (TA Instruments Ltd., West Sussex, UK). A 60 mm diameter standard steel cone with a 2° angle was used to measure the viscosity of the lubricants.

A lubricant volume of 8 ml was placed between the cone and the plate of the rheometer using a plastic pipette. The cone was then lowered until there was direct contact between the cone, plate and lubricant. The cone was positioned with a truncation gap of $54\ \mu\text{m}$ [30].

The experiment was performed at 22°C for shear rates from $0.1\ \text{s}^{-1}$ to $1000\ \text{s}^{-1}$. Three tests were undertaken for each lubricant and the mean viscosity with 95% confidence intervals calculated.

2.4. Tribological Experiment

Tribological tests were conducted using a bespoke pin on disc tribometer as used in previous studies for biomedical materials for implants [31-34], according to the standard ASTM G99-05 [35] at room temperature (figure 2). The experimental approach was to use a pin on disc tribometer to screen the different materials and lubricants before deciding which combinations would be fully tested using a spine simulator in a future study. The apparatus was calibrated against a variety of reference materials including steel against steel and steel against ceramic, both dry and lubricated. The rotational speed of the disc, as recommended by the standard, was 60 RPM ($6.3\ \text{rad/s}$) and the tangential velocity was $160\ \text{mm/s}$. The normal force was $29.4\ \text{N}$ applied vertically on the pin similar to previous studies [31-34]. Each experiment was run for 60 minutes and a sliding distance of $560\ \text{m}$.

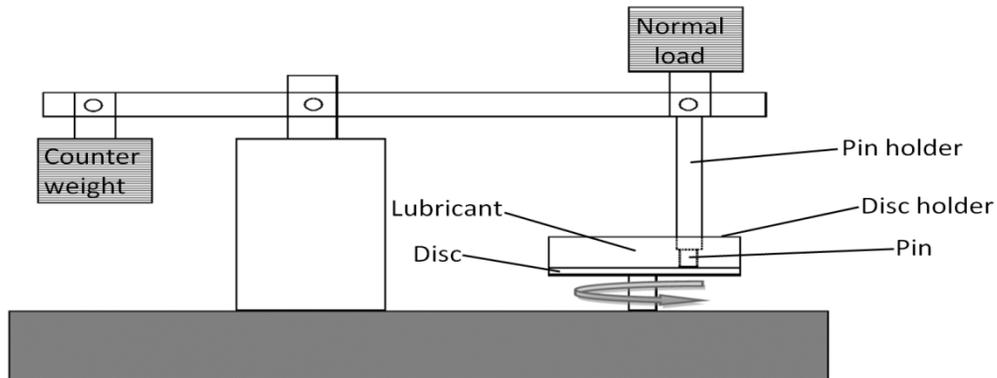


Figure 2: A schematic representation of the pin on disc tribometer.

Friction force data was acquired using a SoMat eDAQ mobile data acquisition system (Hottinger Baldwin Messtechnik GmbH, Germany) at a rate of 1 Hz. For each experiment 10 ml of lubricant was added to the disc holder. The mean steady-state friction coefficient and 95% confidence intervals were calculated from the last ten values of each experiment. Table 2 lists the materials and lubricants used in each test. A moving average trendline with a period of 25 readings was used to smooth out fluctuations in friction coefficient data. The friction force was calculated from the strain gauge by measuring the strain from standard weights according to the calibration method described in the Somat edaq manual (HBM Inc., Marlborough, USA). A half bridge strain gauge was used comprising two strain gauges each of them with gauge factor of

2.05±1% and resistance of 120.4Ω ±0.35% (Omega, Cheltenham, UK). The procedure used to calibrate the gauges is fully described in the Somat edaq manual (HBM Inc., Marlborough, USA) but is summarised below. The summary for computing the strains from gauge and bridge factors using Somat edaq, comprises a tool that defines the calibration using selected shunt resistors for the Somat edaq software along with a corresponding equivalent strain based on gauge and bridge factors of the bridge configuration. Using the Somat software the strain is calculated from Eq. [1]:

$$E_s = \frac{1000000 \times R_g}{B_F \times G_F (R_g + R_s)} \quad [1]$$

where E_s is the equivalent strain, R_g is the nominal gage resistance, R_s is the shunt resistance, G_F is the gage factor and B_F is the bridge factor. Standard masses were added and the strain recorded.

This process was repeated six times for each standard mass. The friction coefficient was calculated from Eq. [2]:

$$\mu = \frac{F}{N} \quad [2]$$

where F is the friction force and N is the normal force.

The friction force was calculated from the strain gauge by measuring the strain from standard weight and measuring the gauge factor.

Table 2: Materials and lubricants used in each test.

Experiment	Pin	Disc	Lubricants
CoCr/CoCr	Cobalt chrome alloy	Cobalt chrome alloy	Ringer's solution, calf serum and PVA A, B, C and D
CoCr/PEEK	Cobalt chrome alloy	PEEK	Ringer's solution, calf serum and PVA A, B, C and D
PEEK/PEEK	PEEK	PEEK	Ringer's solution, calf serum and PVA A, B, C and D
UHMWPE/PEEK	UHMWPE	PEEK	Ringer's solution, calf serum and PVA A, B, C and D
UHMWPE/CoCr	UHMWPE	Cobalt chrome alloy	Ringer's solution, calf serum and PVA A, B, C and D

2.5.Data Analysis

The confidence interval is a measure to estimate the true mean value from a range around the mean value [36]. A 95% confidence interval was used in this study. The result was considered to be significant ($p < 0.05$) when the intervals did not overlap.

3. Results

3.1. Viscosity Results

Results were taken at a shear rate of 1000 s^{-1} which was judged to be the point at which the measured viscosity became stable. The viscosity of Ringer's solution was $2.25 \times 10^{-3} \pm 1.3 \times 10^{-5}$ Pa.s which was the lowest viscosity for all of the tested lubricants. Bovine calf serum followed with a viscosity of $2.35 \times 10^{-3} \pm 1.14 \times 10^{-4}$ Pa.s. The viscosity of PVA A was $2.53 \times 10^{-3} \pm 0.23 \times 10^{-3}$ Pa.s, PVA B was 0.094 ± 0.01 Pa.s, PVA C was 1.650 ± 0.01 Pa.s and the viscosity of PVA D was 2.08 ± 0.077 Pa.s. All viscosity measurements are shown in figure 3.

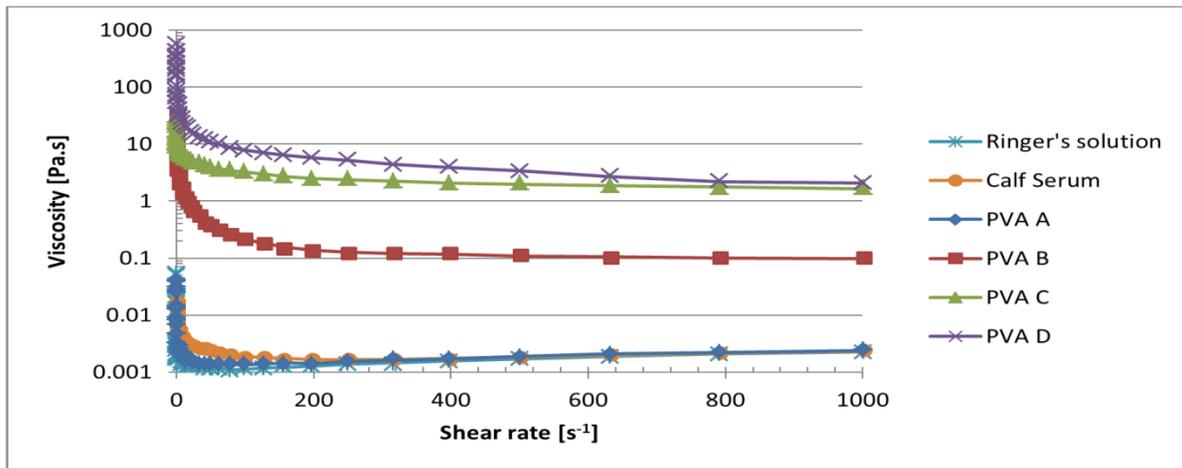


Figure 3: The viscosity of tested lubricants against shear rate at 22 °C; note that the viscosity axis is on a logarithmic scale, base 10.

3.2. Tribological Results

The UHMWPE/PEEK friction coefficient results are shown in figure 4 as an example of the friction coefficient against sliding distance data. The mean friction coefficient for Ringer's solution was 0.2 and for bovine calf serum it was 0.19. With this combination the difference between the Ringer's solution and calf serum was small. The mean friction coefficient for the PVA A was 0.014 and decreased for PVA B to become 0.079 and reached the lowest value of 0.02 for the PVA C for this pair. The value of the mean friction coefficient for this material combination increased to a value of 0.04 when PVA D was used.

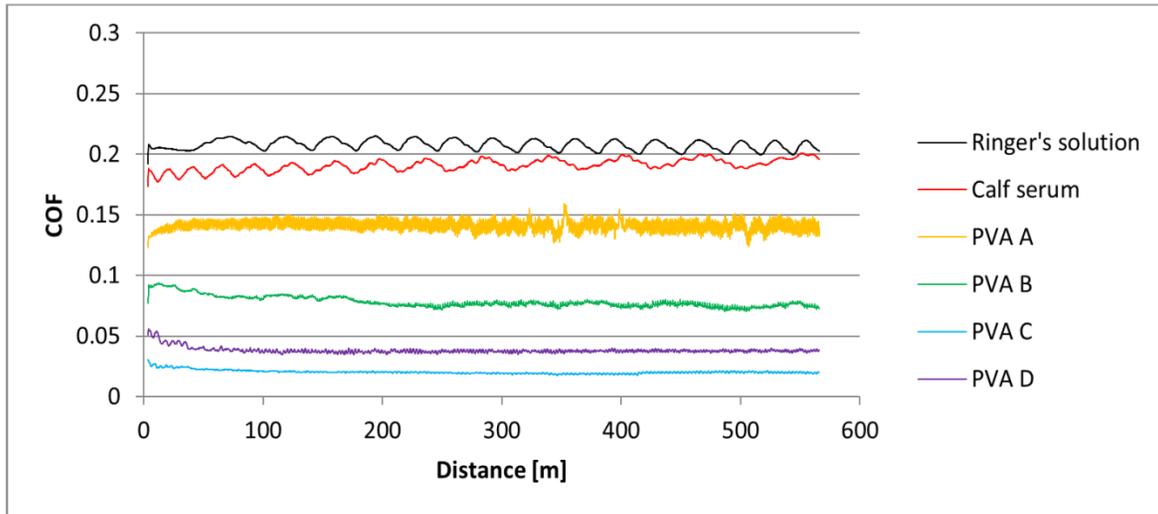


Figure 4: Transient UHMWPE/PEEK friction coefficient against sliding distance for test lubricants.

Figure 5 shows the summary of the friction coefficients for different bearing material combinations with different lubricants. For all materials combinations the highest friction coefficient was found using Ringer's solution as the lubricant (range from 0.1 to 0.34). The friction coefficient was less when bovine calf serum was used. The friction coefficient results for the PVA showed the same trend for all materials combinations with a decreasing friction as the PVA percentage was increased between PVA A and PVA C. With PVA D the friction was then found to increase. The lowest friction coefficient for all materials combinations was for PVA C. PVA C was found to have a significantly ($P < 0.05$) lower friction coefficient compared to all other lubricants for all material combinations. The lowest friction coefficient (0.009) was for the UHMWPE against CoCr with PVA C as the lubricant.

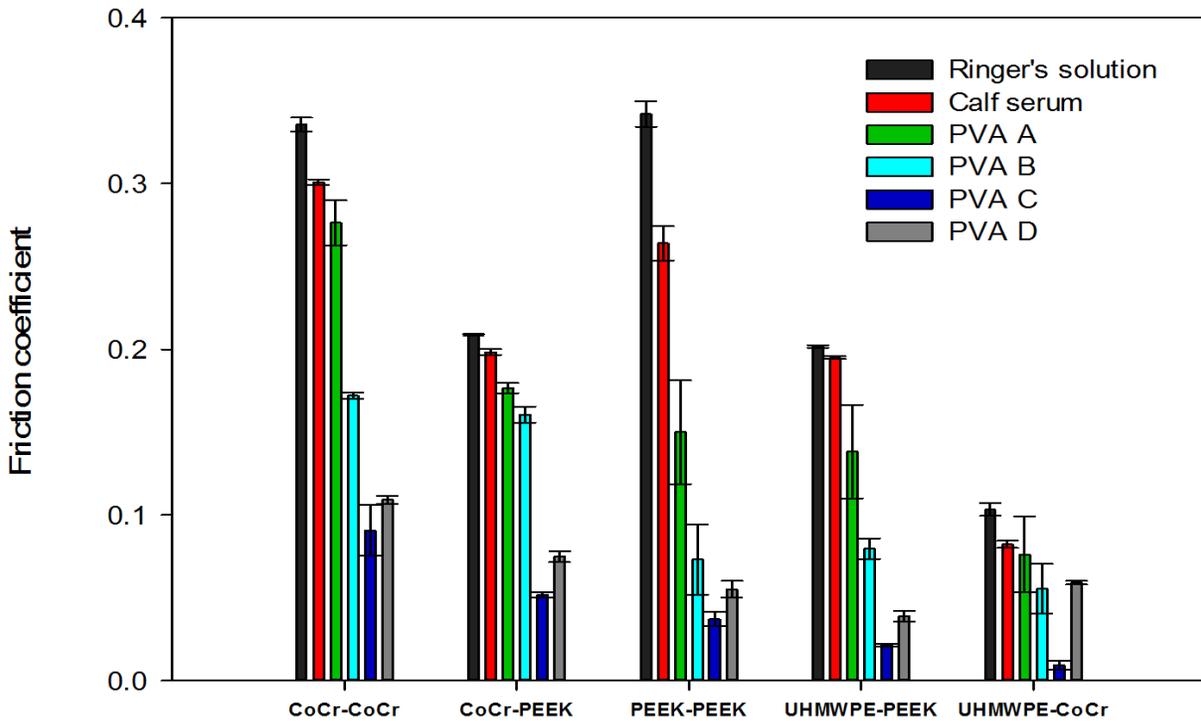


Figure 5: Friction coefficient for all material combinations with different lubricants. Error bars represent 95% confidence intervals.

4. Discussion

The aim of this study was to investigate the friction coefficient for different biomaterials for disc replacement implants using biocompatible lubricants. These experiments were undertaken to help determine suitable biomaterials and lubricants that could be used in an encapsulated disc replacement implant. All the lubricants have shown shear thinning with the increase of the shear rate as shown in figure 3. There was a slight increase in the viscosities of Ringer’s solution, calf serum and PVA A at high shear rates, and this phenomenon has been observed by others [16].

The Ringer’s solution and bovine calf serum gave the highest friction coefficient for all combinations. The diluted serum had lower friction compared to the Ringers solution and this may have been caused by the development of a protein layer. This layer, adsorbed on to the frictional surfaces, would have improved boundary lubrication conditions by providing a phase between the surface and lubricant, leading to reduced friction. This is a phenomenon that has been studied by Scholes and Unsworth [37].

The PVA concentrations used in these experiments produced lubricants that induced lower frictional results even though viscosity of the PVA A was similar to the bovine calf serum. The results were smoother, reducing fluctuations which were the result of poor lubrication conditions, as shown in figure 4, for example. By increasing the concentration of PVA, the viscosity increased. When the viscosity of the fluid increased, the friction coefficient decreased to a certain

value and then increased as shown in figure 5. The incremental increase in the viscosity will lead to a thicker lubrication film which in turn will increase the squeeze time between the film and the two bearing surfaces [38, 39]. For all material combinations the lowest friction coefficient was obtained using PVA C where the lubricant was viscous enough to reduce the friction coefficient to the lowest value. In addition, the steady-state friction coefficient was smoother when PVA C was used for all material combinations. There was no overlapping of the 95% confidence interval between the PVA C results and the other lubricants and this indicated that the friction for PVA C was significantly lower than the other lubricants. The viscosity of PVA D increased the friction coefficient instead of decreasing it, as shown in figure 5. The reason for this increment was due to an increase in frictional forces caused by high viscosity shear; in addition, the fluid flow may be restricted, performing somewhat like a foam [40, 41]. This study has found that the polymer lubricant was more effective than Ringer's solution and calf serum in reducing the friction coefficient. This is supported by Wathier et al. [42] where they found that the polymer lubricants helped in reducing the friction coefficient more than synovial fluid and saline solutions and display a better compatibility with common biomaterials.

Metal sliding against metal showed in general the highest friction coefficient and this can be seen clearly in the CoCr/CoCr results for most lubricants in figure 5. The sliding of CoCr/PEEK shows a lower friction coefficient than the metal against metal. When UHMWPE was used with PEEK and CoCr, it showed lower friction. The lowest friction was for UHMWPE against CoCr.

Figure 5 shows that all the friction coefficient for PVA C fall within the range of 0.009 to 0.09, which according to Bhushan [43] falls within the mixed lubrication regime. By using UHMWPE/CoCr with PVA C as a lubricant the friction coefficient was 0.009 which is close to the hydrodynamic lubrication conditions. The CoCr/CoCr falls firmly within the boundary lubrication regime as shown in figure 5. Gispert et al. [34] found in their study that the friction coefficient of UHMWPE/CoCr lubricated using Hanks' balanced salt solution, 0.39 MPa contact stress, 46 mm/sec and surface roughness of (2.121, 0.317) μm was 0.085. The results from this study are in agreement with those of [Gispert et al. [34]].

5. Conclusion

PVA of different viscosities, Ringer's solution and bovine calf serum were used as the test fluids. In addition, lubricant viscosity, modified by PVA concentration has also been assessed. A comparison has been made between the bio-lubricant polyvinyl alcohol (PVA) with varying concentrations, and other commonly used lubricants in-vitro tribological tests, for potential use in encapsulated implants. Experiments measured viscosity and the frictional behaviour of the test fluids with a range of bearing material combinations. The conclusions from this study are:

- The PVA lubricant lowered the friction coefficient for CoCr/CoCr, CoCr/PEEK, PEEK/PEEK, UHMWPE/PEEK and UHMWPE/CoCr in comparison with Ringer's solution and bovine calf serum.
- Increasing the concentration of PVA gave a higher viscosity, which in most cases led to a lower friction coefficient. However, this effect was superseded by shear losses when the viscosity exceeded 2.08 Pa.s, as was observed for the PVA D (30 g per 100 mL of distilled water).

- PVA C (20 g per 100 mL of distilled water) with 1.650 ± 0.01 Pa.s viscosity showed the lowest friction coefficient for all combinations compared with other lubricants.
- UHMWPE/CoCr was the optimum material combination with the lowest friction coefficient for all lubricants.
- PVA C with an UHMWPE/CoCr material combination is suggested for encapsulated joint replacement implants.

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