How Does Lubricant Viscosity Affect the Wear Behaviour of VitE-XLPE Articulated Against CoCr?

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Abstract

Using a 50-station pin-on-disc (SuperCTPOD) machine, the influence of lubricant viscosity on the wear of vitamin E blended crosslinked polyethylene was investigated. Five different test lubricants were prepared by mixing different concentrations of carboxymethyl cellulose powder with deionised water. The viscosity range of the lubricants was 0.002-0.155 Pas, a range that represents the viscosities of diseased and healthy synovial fluids. Five groups of pins (10 pins in each group) were articulated against cobalt chromium discs. Wear was measured in terms of weight loss from each pin and disc for every group. Every 500,000 cycles the experiment was stopped to take gravimetric measurements along with roughness measurements of the articulating surfaces. The test discs did not show a significant change in weight after 2.5 million cycles of testing ($p > 0.05$). For the pins, the group tested with the lowest viscosity (0.002 Pas) produced the highest wear rate, namely 0.931 mg/million cycles, and the wear rates of the other groups were 0.074, 0.027, 0.034 and 0.021 mg/million cycles respectively. The wear rates calculated for the five groups were all lower than the wear rates recorded for ultrahigh molecular weight polyethylene and not significantly different to crosslinked polyethylene. In addition, apart from group 1 pins (tested with the lowest lubricant viscosity (0.002 Pas)), the machining marks on the other pins were still present after 2.5 million cycles of testing, indicating low wear.

Introduction

Lubricant viscosity is an important property that influences surface interactions in natural [1] and manmade joints [2]. It has a major effect on the lubrication of articulating surfaces of orthopaedic implants especially in the potential of forming a sufficiently thick layer to fully separate surface asperities to reduce wear [3]. Theoretically, surfaces that articulate with viscous fluids tend to produce less wear due to having thick lubricant films in between the surfaces [4]. In human joints, synovial fluid is a viscous lubricant that reduces friction and wear between articulating surfaces [5]. Although the volume of synovial fluid present within the synovial joints is small, it is sufficient to separate and lubricate the articulating surfaces [1]. Synovial fluid viscosity is found to be influenced by joint diseases [5]. Compared to healthy synovial fluid, patients with osteoarthritis or rheumatoid arthritis tend to have lower viscosity synovial fluids [1, 2, 5]. A viscosity range of 0.01-0.4 Pas and 0.0025-0.2 Pas was recorded for non-diseased and osteoarthritic synovial fluids respectively at a shear rate of 300 s^{-1} [5]. Cooke et al. [1] recorded viscosity values of 0.1 Pas, 0.01 Pas and 0.005 Pas for non-diseased, osteoarthritis and rheumatoid arthritis synovial fluids respectively at 1000 s^{-1} shear rate. This reduction in viscosity in arthritic joints is one reason why visco-supplements have been offered [6]. However, their value has been debated [7, 8]. In biotribological tests, various types of lubricants such as water, serum-based lubricants, saline solution or Ringer solution, were assessed in terms of generating clinically relevant results to find the optimum test lubricant [9-12]. Both the type of the lubricant and the lubricant's protein concentration have been shown to affect the wear generated from the articulating surface of polyethylene used in joint replacements [9-12].

Several international standards recommend using diluted calf or bovine serums in biotribology tests, which resemble human synovial fluid in terms of composition. Diluting these serums to reduce the protein concentration to the level of synovial fluid protein concentration level is common. Yet, there has been no consensus reached on what is the optimum dilution ratio of the test lubricant should be. ASTM International F732 [13] and ISO 14242-1 [14] suggests 75 % (volume fraction) and 30 g/l respectively.

When a human joint fails due to a disease, such as osteoarthritis, it can be replaced with an artificial implant. Joint replacement has become an accepted and generally successful solution for diseased natural joints [15]. Metal-on-polymer (MoP) articulation is the most common biomaterial combination utilised in orthopaedics [15]. As the polymer component, ultrahigh molecular weight polyethylene (UHMWPE) was widely used as a bearing surface, however, its service life was said to be around 15 years due to its relatively low wear resistance against metals [4]. Wear debris from UHMWPE was associated with osteolysis and aseptic loosening and it was found to be the major factor that limits the service life of MoP prostheses [16, 17]. The increase in demand for arthroplasty from younger and more active patients was among the reasons that initiated research for finding alternative polymers to be used as bearing surfaces. One of these polymers is crosslinked polyethylene (XLPE). XLPE showed a significant reduction in wear rates compared to UHMWPE [16] so that it has become the material of choice for acetabular liners in contemporary total hip replacement [18].

First-generation XLPE is obtained by irradiating UHMWPE, followed by heat treatment. During the irradiation process, free radicals are produced which makes the material less stable. Therefore, heat treatment is needed to eliminate the free radicals generated during the irradiation process; however, melting reduces the fatigue strength of the irradiated UHMWPE [18]. In addition, oxidation was observed for XLPE components produced with this technique (irradiation followed by heat treatment) in vivo [19]. Recent studies showed that the addition of vitamin E (α -tocopherol) to XLPE (VitEXLPE) results in a material with high wear resistance and superior mechanical properties compared to UHMWPE and without the need for the melting process [20]. This approach, producing a stable, oxidation resistant XLPE by adding vitamin E, resulted in second-generation XLPE. These second-generation XLPEs can be obtained in two ways: blending vitamin E with UHMWPE resin and then consolidating the mixture by compression molding followed by an irradiation process. Alternatively, vitamin E can be diffused into an already irradiated and crosslinked UHMWPE. With the first approach (blending vitamin E with UHMWPE resin followed by compression molding and irradiation processes) however, the crosslinking efficiency of UHMWPE may be reduced because vitamin E can act as a free radical scavenger during the irradiation process. This may result in a relatively less wear resistant material (due to less crosslinking) compared to VitEXLPE produced with vitamin E diffusion [18].

Given its relatively recent introduction, there are limited in vitro studies which have analysed the wear behaviour of VitEXLPE, and the summary of the results of these studies are discussed below.

Table 1 Studies tested the wear behaviour of VitEXLPE against CoCr using serum-based lubricants.

Author	Materials Tested (pin material named first)	Type of wear testing	Test Machine	Test Lubricant	Test Lubricant Temperature (°C)	Contact stress (MPa)	Wear Factor $(\frac{mm^3}{Nm}) \times 10^{-6}$
Saikko [21]	Vitamin E (0.1%) stabilised UHMWPE (gamma- irradiated (100 kGy) GUR 1020- E) vs CoCr	Flat-on-flat	Computer-controlled 16-station multidirectional RandomPOD	HyClone Alpha Calf Fraction serum (dilution ratio: 1:1 with deionised water, protein concentration: 20 mg/ml)	20	0-2.2 (a load ranging from 0- 142 N was applied to 9 mm diameter pins)	1.016 ^a
						0.65-10 ^b	1.260 ^b
Saikko [19]	Vitamin E (0.1%) stabilised UHMWPE (gamma- irradiated (100 kGy) GUR 1020- E) vs CoCr	Flat-on-flat	Computer-controlled 16-station multidirectional RandomPOD	HyClone Alpha Calf Fraction serum diluted with 1) phosphate buffer saline, 2) phosphate buffer, 3) saline solution and 4) deionised water (dilution ratio: 1:1, protein concentration: 20 mg/ml, same for all lubricants tested)	20	0-2.2 (a proportional- pneumatic load ranging from 0- 142 N was applied to 9 mm diameter pins, sliding distance:55km)	1) 0.025 2) 0.11 3) 0.56 4) 1.19

^a the relation between the wear factor and disc surface roughness $k = 1.27 \times 10^{-5} (Sa)^{0.56}$ was found by the author and the value provided in the table was calculated for Sa=0.011 µm.

^b the relation between the wear factor and disc surface area $k = -0.012A + 2.023$ was found by the author and the value provided in the table was calculated for A=63.62 mm².

Encouraging results were reported by Saikko [19, 21] (table 1) in terms of wear and oxidation resistance for VitEXLPE compared to conventional irradiated and heat treated UHMWPE. Interestingly, the wear rate of VitEXLPE was found to decrease with an increase in the contact area which is in contrast with what was observed for UHMWPE and XLPE [22, 23].

Saikko [19] found that the wear behaviour of UHMWPE and VitEXLPE were highly dependent on serum diluent and dilution ratio [19]. Therefore, the lubricant content was one of the main features that was responsible for the wear particles generated from the articulating surfaces, as can be seen from the different wear factors calculated using different serum-based lubricants (table 1). Compared to the wear generated from UHMWPE pins tested with the same machine with HyClone Alpha Calf Fraction serum (diluted with distilled water with a ratio of 1:1), Saikko and Kostamo [24] recorded a wear factor of $2.55 \times 10^{-6} \frac{\text{mm}^3}{\text{Nm}}$ which is approximately twice as high of that recorded for VitEXLPE pins [19, 21].

Popoola et al. [25] compared the wear behaviours of two different types of elbow implants. The first design generated contacts of CoCr/UHMWPE, titanium alloy (Ti6Al4V)/UHMWPE and UHMWPE/UHMWPE between the polymer bushing and metal pins, and the second one generated contacts of CoCr/VitEXLPE, Ti6Al4V/VitEXLPE and VitEXLPE/VitEXLPE [25]. They found that after 3 million cycles of testing, wear rates of UHMWPE and VitEXLPE bushings were $9.3 \pm 2.8 \frac{\text{mm}^3}{\text{Nm}}$ and $0.8 \pm 0.2 \frac{\text{mm}^3}{\text{Nm}}$ respectively [25]. Popoola et al. [26] also tested the wear behaviour of VitEXLPE acetabular cups (diameter: 40mm) against CoCr femoral heads with water diluted bovine calf serum and recorded a wear rate of $1.93 \pm 0.25 \text{ mm}^3/\text{million cycles}$ and $38.08 \pm 0.75 \text{ mm}^3/$ for VitEXLPE and UHMWPE respectively. These results indicated that the wear rate of VitEXLPE was at least one order less than its conventional equivalent.

Rowell and Muratoglu [27] investigated 12 explanted VitEXLPE acetabular liners (gamma irradiated (100 kGy), GUR 1050) and three explanted VitEXLPE tibial bearings (gamma irradiated (100 kGy), GUR 1020) having in vivo times ranging from 0.1 to 37 months (with a mean of 10 months). They observed visible machining marks on the bearing surfaces of both types of explants which was in contrast with the burnished appearance recorded for the explanted UHMWPE liners [28]. This short-term analysis implies VitEXLPE's superior wear and deformation resistance compared to UHMWPE. In a systematic review conducted by Wyatt et al. [29] VitEXLPE was found to have significantly better results compared to XLPE liners in terms of reduced femoral head penetration for a minimum of 3 years follow-up.

In a hip simulator study, Affatato et al. [30] showed that vitamin E (0.1%) blended XLPE (electron beam irradiated (70kGy) and heat treated at 135°C, GUR 1020) acetabular liners (inner diameter: 28 mm, outer diameter: 44mm) wore more than XLPE (electron beam irradiated (70 kGy) and heat treated at 135°C, GUR 1020) liners of the same size in diluted bovine calf serum against 28 mm CoCr femoral heads, namely 129 mg and 68 mg after 5 million cycles of testing, but the recorded wear rates were significantly lower than conventional UHMWPE (which wore 163 mg after 5 million cycles) [30]. In contrast, Oral and Muratoglu [31] showed that vitamin E diffused 100 kGy gamma irradiated UHMWPE acetabular liners (inner diameter: 28 mm, outer diameter: 40mm) had lower wear rates than 100 kGy irradiated and melted XLPE liners (inner diameter: 28 mm, outer diameter: 40mm) against CoCr femoral heads, both tested in bovine serum. Therefore, it appears that the manufacturing technique to obtain VitEXLPE can affect its wear behaviour against metals in vitro.

The aforementioned studies all investigated the wear behaviour of MoP implants with a serum-based lubricant, without considering the effect of lubricant viscosity on wear. As discussed previously,

lubricant viscosity can have a major effect on the wear mechanisms taking place between the articulating surfaces. Therefore, it is essential that the prostheses used in joint replacement operate successfully in the relatively lower viscosities that are seen with diseased synovial fluids. Therefore, the wear behaviour of VitEXLPE versus CoCr using carboxymethyl cellulose (CMC) fluids with different viscosities was studied. To our best knowledge, the effect of lubricant viscosity on the wear behaviour of VitEXLPE against CoCr has not been investigated before.

Materials and Methods

Test Lubricants

Human synovial fluid is a non-Newtonian, shear rate dependent fluid [9]. It becomes thinner with an increase in shear rate. This behaviour is associated with the presence of hyaluronic acid (HA), which is one of the components of synovial fluid [9, 32]. The flow properties of synovial fluid are affected by HA concentration. When the HA concentration drops from its natural levels, the viscosity of the synovial fluid decreases, leading to a thinner lubricant film generated between articulating surfaces. This increases the possibility of wear and artificial joint failures [9].

For wear testing, synovial fluid cannot be used as the test lubricant even though it may be considered the ideal fluid to obtain clinically relevant results. There are several reasons for its unavailability. First, there are major ethical concerns in terms of taking synovial fluid from patients. Secondly, the volume of synovial fluid contained within the synovial membrane is very limited, a range of 0.2-0.4 ml [33], whereas in biotribology analysis conducted with a pin-on-disc machine, i.e. the SuperCTPOD (circularly translating pin-on-disc) used in this study which has 50 test stations, approximately 16 ml of test lubricant in each test station is required. For a test that runs 2.5 million cycles in which the test lubricant is needed to be replaced every 500,000 cycles, 4 litres of test lubricant is needed to complete the test.

In this study, CMC fluids with different viscosities was used to investigate the effect of viscosity on the wear behaviour of VitEXLPE test pins rubbing against CoCr discs. CMC fluids are also non-Newtonian and have very similar rheological properties compared to synovial fluid [1]. The test fluids were prepared by mixing low density CMC powder (Sigma-Aldrich, carboxymethylcellulose sodium salt low density) and deionized water. The viscosity values of the test lubricants were measured with a cone-on-plate Malvern KINEXUS pro+ rheometer at 20 °C. A special solvent trap system was used to prevent liquid evaporation. Three replicates of rheological tests were done before the biotribological experiment and every time it was stopped for measurements. The shear rate range was 0 – 10³ s⁻¹ (logarithmic scale), see figure 1.

Figure 1 Variation of the viscosity values of CMC fluids with different concentrations with shear rate. The viscosity values given in brackets corresponds to the average viscosity values measured at $1000 s^{-1}$ shear rate for each group respectively.

Two shear thinning behaviours were observed for the lubricant prepared for Group 1 (CMC fluid having a viscosity of 0.002 Pas). A similar rheological response was observed by Benchabane and Karim [34] for the CMC fluids prepared with concentrations (weight percentage) of 0.8% and lower. For higher viscosities (0.056 to 0.155 Pas), only a single shear thinning behaviour was observed for the shear rate range tested. This was also reported by Benchabane and Karim [34].

Five different viscosity CMC fluids were prepared. The viscosities of the lubricants ranged from 0.002-0.155 Pas, which is a range that includes the values of healthy and diseased synovial fluids [5]. At a shear rate of $1000 s^{-1}$, viscosities of 0.002, 0.056, 0.089, 0.122 and 0.155 Pas were measured for each CMC fluid prepared which were then used to wear test group 1, group 2, group 3, group 4 and group 5 of pins and discs, respectively.

Wear Test

The wear test used the 50-station multidirectional SuperCTPOD (figure 2). The machine and its working principle have been explained in detail elsewhere [23, 35-37]. Five groups of VitEXLPE pins were tested against 5 groups of CoCr discs using CMC fluids as test lubricants. A sample test pin and a disc used in the experiment are shown in figure 3.

Figure 2 a) Multi-station SuperCTPOD that consists of a loading module (1), pin carrier module (2), test bath (3), motion module (4) and control unit (5),

2 b) Schematic diagram of the pin-disc assembly showing (1) load module, (2) polyacetal pin holder, (3) PVC tube, (4) lubricant, (5) test pin, (6) silicone O-ring and (7) test disc.

Figure 3 Sample CoCr disc and VitEXLPE pin.

Pins were vitamin E (0.1%) stabilised, gamma irradiated (105 kGy) GUR-1020 medical grade UHMWPE (material purchased from Orthoplastics). They were all 9 mm in diameter and 12 mm in height. Vitamin

E was added to the virgin polymer prior to consolidation as stated in the test certificate of the material by the manufacturer.

The discs were all of diameter 28 mm and had a thickness of 10 mm. Prior to testing, the discs were polished to less than 0.05 μm Ra, as suggested in ASTM F732-17 for metallic surfaces [13]. The surface roughness of the pins and the discs were analysed using a Zygo 5000 non-contacting profilometer with a sensitivity of 0.001 μm , Sa [38, 39].

Every pin articulated against the same corresponding disc throughout the test. Approximately 16 ml of lubricant was added to each pin-disc assembly and the temperature inside the test bath was kept constant at approximately 20°C. The assemblies (figure 2b) followed an elliptical orbital motion under a constant load of 70.7 N, which corresponded to 1.11 MPa of nominal contact stress. The wear track path was set to 12 x 10 mm which was previously used in studies [23, 36] which tested the wear behaviours of XLPE and UHMWPE. This pin-on-disc device enabled a continuous change of the direction of sliding of the discs relative to the pins.

Wear was assessed in terms of weight loss using a Kern ABT 220-5DM balance, having a resolution of 10 μg . Wear rates, milligramme per million cycles (mg/Mc), and wear factors ($k = \frac{\text{mm}^3}{\text{Nm}}$) were calculated to make comparisons among groups. The wear factor is given by:

$$k = \frac{V}{LD} \quad (\text{Equation 1})$$

where V (mm^3) is the volume loss, L (N) is the load applied and D (m) is the sliding distance.

The area of contact, the force applied and the temperature inside the test bath were all kept constant throughout the experiment. Control pins and discs were used to monitor lubricant uptake for each group.

Results

The wear rates and wear factors calculated for the test pin groups are given in table 2.

Table 2 Summary of the calculated wear rates and wear factors for the test groups.

Test pin group	Wear Rate (mg/Mc) $\pm$ SD	Wear Factor (mm^3/Nm) $\times 10^{-6} \pm$ SD
Group 1	0.931 $\pm$ 0.790	0.405 $\pm$ 0.343
Group 2	0.074 $\pm$ 0.047	0.032 $\pm$ 0.020
Group 3	0.027 $\pm$ 0.021	0.011 $\pm$ 0.009
Group 4	0.034 $\pm$ 0.022	0.015 $\pm$ 0.009
Group 5	0.021 $\pm$ 0.009	0.010 $\pm$ 0.004

SD: Standard deviation

At the end of the experiment, the total weight gains of the control pins were measured as 0.03, 0.01, 0.00, 0.02, and 0.01 mg for each group respectively. There was no statistical difference between the

initial and final weight gains of the control pins (p-values for each control group were found as 0.464, 0.424, 0.425, 0.253 and 0.366, one-way ANOVA). Changes due to lubricant uptake were considered in the wear analysis.

Figure 4 Weight losses from the test pins articulated against CoCr discs.

The weight changes were recorded every 500,000 cycles for each group and the results are shown in figure 4. The weight losses were significantly different among pin groups (p-value:0.004, ANOVA), however, the weight losses recorded for pins from groups 3-5 did not show a significant difference throughout the test (p-value:0.846, ANOVA).

Figure 5 Visual analysis of sample pins from each test group before (a) and after (b) the experiment. For scale, recall that each pin is 9mm in diameter.

Visual analysis was conducted for the test pins, and sample images from each group are shown in figure 5. At the end of testing, apart from group 1 pins, machining marks were found to be visible for group 2-5 pins.

Figure 6 Surface roughness values of the pins before and after the experiment.

Compared to their initial values, surface roughness value of every pin group changed significantly, returning p-values of 0.000, 0.001, 0.000, 0.001, and 0.000 respectively (figure 6).

Figure 7 Changes in the surface roughness of the test discs

On average, the disc surface roughness value increased from 0.005 to 0.011 μm , see figure 7. This change occurred after 2.5 million cycles of testing and the change was found to be significant (p-value:0.000, ANOVA). However, after the first 500,000 cycles the change in surface roughness was insignificant (p-value:0.376, ANOVA). Put simply, the changes to disc surface roughness values occurred within the first 500,000 cycles.

After the wear test was completed, some mild scratches were observed on the surfaces of the test discs, on the wear track path followed by the pins. However, there was no detectable weight loss from the test discs after 2.5 million cycles of testing. When the change of weight of each group of test discs was compared (prior to the experiment and every time the experiment was stopped for measurements), the ANOVA test returned p-values of 0.997, 0.999, 0.998, 0.999 and 0.999 for group 1, group 2, group 3, group 4 and group 5 discs respectively. Therefore, the weight change of the test discs was found to be insignificant. Furthermore, no significant change in the weight of the control discs was observed throughout the experiment (p-value:1.000, ANOVA).

No polyethylene transfer to the metallic counterface was observed visually. The non-significant change in the weight of the test discs also supported this observation.

To aid comparison of the wear rates of UHMWPE, XLPE and VitEXLPE, the rheological behaviour of newborn calf serum (dilution ratio: 1:2, protein concentration: 22 g/L) which was used in the studies conducted by Harsha and Joyce [36] and Kandemir et al. [23] was analysed. The shear rate dependent rheological response of newborn calf serum was displayed in figure 8. Shear thinning behaviour was found for a shear rate interval of 0.1-100, then slightly increased reaching to 0.005 Pas at 1000 s^{-1} of shear rate.

Figure 8 Variation of diluted newborn calf serum (1:2) viscosity with shear rate.

Discussion

Polymer wear and its negative outcomes are recognised as major reasons for MoP prosthesis failures. The changes in the rheology of the synovial fluid, before or after joint replacement, have been found to influence the wear generated from implant surfaces [1]. These changes usually occur in the viscosity and the protein content of the synovial fluid [40].

Theoretically, lubricants with lower viscosities produce thinner lubricant films that result in more interaction between surface asperities thus, more wear. The proteins in serum-based lubricants, however, tend to form a boundary layer [40]. This boundary layer may influence the wear that is generated due to surface contact, therefore, in order to test the load bearing capacity of VitEXLPE in a range of lubricant viscosities (considering the worst and the best-case scenarios observed in diseased and healthy/normal joints), we sought to avoid the influence of proteins on wear. To our best knowledge, this study is the first to investigate the influence of lubricant viscosity on the wear of VitEXLPE articulated against CoCr.

Wear rates decreased as the viscosity of the test lubricants increased. This result was expected because according to the theoretical lubrication regime model proposed by Hamrock and Dowson [41], viscous lubricants generate thicker lubricant films that separate the articulating surfaces to reduce surface interactions. Only the CMC fluid used to test group 4 (0.112 Pas) did not follow this trend, the wear rate for this group was slightly higher than group 3 (0.089 Pas). However, the difference in the wear rates of group 3 and group 4 pins was not found to be significant (p-value: 0.846).

Visual changes on the surfaces of the pins are shown in figure 5. Machining marks only disappeared completely in the lowest viscosity case tested (group 1, tested with 0.002 Pas CMC fluid). The viscosity of the CMC fluid that group 1 pins were tested resembled the viscosities recorded for rheumatoid (0.005 Pas) synovial fluids [1]. For the other groups, the viscosity values of the CMC fluids were higher (0.056 to 0.155 Pas), and the machining marks were still present after 2.5 million cycles of testing. The remaining machining marks served to visually indicate the high wear resistance of VitEXLPE. A similar

observation for VitEXLPE acetabular liners was also made by Rowell and Muratoglu [27] when a serum-based lubricant was the test lubricant.

Using data from previous studies, [23] and [36] that used newborn calf serum (dilution ratio: 1:2) as the test lubricant, the wear rates of UHMWPE, XLPE and VitEXLPE were compared (figure 9). Group 1 VitEXLPE pins were selected to be used for comparison. The viscosity of the group 1 CMC fluid and its rheological response to changing shear rate were similar to what was observed for diluted newborn calf serum.

Figure 9 Weight losses from VitEXLPE, XLPE and UHMWPE pins articulated against CoCr discs.

As shown in Figure 9, the wear rate of VitEXLPE is significantly lower than that of UHMWPE ($p:0.041$). The weight losses from XLPE and VitEXLPE pins did not show a significant difference ($p:0.597$). However, it should be noted that the test lubricants used were different. For XLPE pins tested with diluted newborn calf serum, the lubricant viscosity was assumed to be constant (a constant viscosity of 0.01 Pas was recorded by Scholes and Unsworth [42] in an analysis conducted with artificial hip implants).

Theoretically, the wear rates calculated for group 1 VitEXLPE pins (0.002 Pas) of this study should be lower when tested in a serum-based lubricant which is assumed to have a higher viscosity (0.01 Pas). This is because viscous lubricants tend to produce thicker lubricant films between articulating surfaces. This in return can reduce asperity contacts and reduce wear [4]. In addition, proteins that are present in the serum-based lubricants have been shown to produce a boundary layer by adsorbing on the bearing surfaces which can also protect the articulating surfaces from wear [10]. Kandemir et al. [23] showed that the XLPE pins, which were tested against CoCr discs with diluted calf serum, produced a wear rate of 1.05 mg/Mc, whereas the group 1 VitEXLPE pins produced 0.93 mg/Mc when articulated against the same material with a CMC fluid. The difference was not found to be significant, however, this outcome contradicted the aforementioned theory. Even though the viscosity of the group 1 test

lubricant was lower and contained no serum proteins, the wear rates of VitEXLPE were found to be slightly lower than XLPE.

Our results clearly show that the viscosity of the lubricant matters. This is contrary to the common belief that with the flat-on-flat contact no viscosity effect can exist as there is no wedge or clearance effect. It appears that it is rather a question of the lubrication of the asperity contact than of the macroscopic shape of the specimens. For future work, it would be interesting to combine increased viscosity, by CMC fluid, with the presence of proteins. This could be a next step forward in the development of a synthetic replacement for serum in orthopaedic tribology.

This study has some limitations. First, this study tested a flat-on-flat configuration to observe the effect of viscosity on the wear behaviour of VitEXLPE against CoCr. This is different to the spherical surfaces in a total hip replacement where the differences in radii do influence theoretical lubrication regimes. However, the use of the high-capacity SuperCTPOD allowed us to relatively quickly investigate the effects of viscosity on the wear of VitEXLPE. Secondly, CMC fluids are not serum-based lubricants, which is a prerequisite suggested by international standards [13, 14] for in vitro wear testing. However, using CMC fluids allowed the effect of lubricant viscosity on wear to be observed, which was not possible with serum-based lubricants.

Conclusion

VitEXLPE pins were tested against CoCr discs with CMC fluids having different viscosities which covers viscosity of healthy and diseased synovial fluids. There was a trend for wear to reduce as the viscosity of the lubricant increased. Even for the group tested with the lowest viscosity (mimicking the severest case observed for rheumatoid synovial fluids), the wear rate calculated for VitEXLPE was significantly lower than the wear rate of UHMWPE and was not significantly different than the wear rate calculated for XLPE.

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Conflict of Interests

The authors have no conflict of interest to declare.

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