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Prediction of Lubrication Regimes in Two-Piece Metacarpophalangeal Prostheses

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Abstract

Various designs of two-piece finger prosthesis with conforming spherical surfaces have been proposed. These were compared by calculating the lubrication regime for the material combinations and operating conditions expected at the metacarpophalangeal joints of the fingers. Consideration was given to a range of loads from 2 to 50N, a range of entraining velocities from 0mm/s to 30mm/s, and a range of prosthesis radii from 3mm to 10mm. This theoretical lubrication analysis indicated that the optimum material combination of those available for two-piece metacarpophalangeal prostheses is in the order: ceramic-on-ceramic; metal-on-metal; pyrocarbon-on-pyrocarbon; and metal-on-polymer. However it should be recognised that other

factors may take precedence when choosing a material combination for a design of finger prosthesis.

Keywords: metacarpophalangeal; prosthesis; lubrication regime

Introduction

Currently, there is a debate regarding the potential long-term success of metal-on-metal 'resurfacing' hip prostheses [1-3]. These implants tend to employ relatively large articulating diameters of the order of 55mm and utilise cobalt chrome femoral heads rubbing against a matched cobalt chrome acetabular cup. This arrangement is in contrast to a 'conventional' Charnley hip prosthesis which would have a stainless steel femoral head of 22.225mm diameter which articulates inside an ultra high molecular weight polyethylene (UHMWPE) acetabular cup. While the short term (to 5 years) results of the metal-on-metal resurfacing hip prostheses have been excellent [4-5], it remains to be seen if they will match the long term (15 to 33 years) success of the Charnley and similar designs [6-8]. One of the theoretical advantages of the metal-on-metal resurfacing hip prostheses is that during part of the walking cycle they can operate with fluid film lubrication. This is due to their relatively large size, low roughness values, appropriate radial clearances and the materials employed [9]. During fluid film lubrication, the articulating surfaces are separated by a film of fluid, so that wear and friction should be minimized. This prediction is in contrast to that for the Charnley and similar hip prostheses which are forecast to operate in the boundary or mixed lubrication regime [10]. Under such conditions, surface interaction is expected with the potential of wear and increased friction.

A range of concepts for metacarpophalangeal (MCP) arthroplasty have been proposed over the years, with single-piece silicone prostheses, particularly the Swanson implant, dominating this market [11]. However, various two-piece designs which tend to aim for a more anatomical solution by having spherical bearing surfaces in a ball and socket arrangement have also been put forward. These have used a range of biomaterial couples including ceramic-on-ceramic, metal-on-metal, pyrocarbon-on-pyrocarbon and metal-on-polymer [11]. The aim of this paper was to compare the predicted lubrication regimes for these various biomaterial couples in the application of a two-piece MCP prosthesis.

In terms of specific implants, a ceramic-on-ceramic prosthesis is currently produced by Moje [12]. It employs zirconia as its material and comes in a range of sizes for the metacarpal head from 11mm to 15mm spherical diameter. To date no clinical results regarding this finger prosthesis appear to have been reported in the literature. A metal-on-metal MCP implant is offered by Biomet, named the Andigo® [13]. Both components are manufactured from cobalt chrome. No clinical results appear to have been detailed in the literature. A pyrocarbon-on-pyrocarbon implant is currently offered by Ascension [14]. Clinical results for this implant have been reported [15-16]. At eleven years follow up, for 71 prostheses implanted into 26 mostly rheumatoid patients, an average increase in range of movement of 13° was reported [16]. While 94% of implants showed evidence of osseointegration, 12% of implants were revised. It should also be noted that in the study the prosthesis was used in patients whose MCP joints had little deformity or subluxation [16].

In the metal-on-polymer category of MCP implants, Zimmer Germany offer a prosthesis named the Elogenics™ which has a titanium phalangeal component which articulates against an

UHMWPE capped metacarpal component [17]. In vitro testing of this implant has recently been reported [18]. Finsbury Orthopaedics offer a prosthesis having a cobalt chrome metacarpal component which articulates against an UHMWPE phalangeal component. The prosthesis is known as the total metacarpophalangeal replacement (TMPR™) and comes in four sizes, the metacarpal head size ranging from 9.5mm to 17mm diameter [19]. A clinical trial with 13 prostheses implanted and a follow-up of five years has been reported [20]. Of the eight patients in the clinical trial, seven had degenerative arthritis, while the eighth had rheumatoid arthritis but only with mild deformity. For this patient cohort an increase in range of motion from 27° to 60° was reported.

The aim of this paper was to calculate the theoretical lubrication regimes for the various two-piece designs of MCP prosthesis currently available. The objective was to see what lubrication theory can tell us about the potential of these designs and therefore predict which may be the most successful implant in vivo.

Method and Materials

Modelling the ball and socket implant as an equivalent ball-on-plane model and employing elastohydrodynamic theory [21-22] allowed the minimum effective film thickness (h_{min}) to be calculated from:

$$\frac{h_{\min}}{R_x} = 2.80\left(\frac{\eta u}{E^* R_x}\right)^{0.65}\left(\frac{w}{E^* R_x^2}\right)^{-0.21}$$

Where R_x is the equivalent radius (m), η is the viscosity of the lubricant (Pa s), u is the entraining velocity (m/s), E* is the equivalent elastic modulus (Pa), and w is the load (N). In turn, given that R_a is the surface roughness and assigning subscript 1 to the ball and subscript 2 to the socket of the MCP prosthesis under consideration, then the lambda ratios were calculated from:

$$\lambda = \frac{h_{\min}}{\left[(R_{a1})^2 + (R_{a2})^2\right]^{1/2}}$$

This allowed the lubrication regime to be identified, as $\lambda < 1$ indicates boundary lubrication, $\lambda > 3$ designates fluid film lubrication, and between these values mixed lubrication is indicated [23].

Before these calculations could be undertaken, the equivalent radius (R_x) was calculated from:

$$\frac{1}{R_x} = \frac{1}{R_1} - \frac{1}{R_2}$$

Where R refers to the radius of the component and subscript 1 refers to the ball and subscript 2 to the socket of the MCP prosthesis under consideration. The equivalent modulus of elasticity was determined from the equation:

$$\frac{1}{E^*} = 0.5\left(\frac{1-\nu_1^2}{E_1} + \frac{1-\nu_2^2}{E_2}\right)$$

Again, E refers to the Young's modulus of the component and subscript 1 refers to the ball and subscript 2 to the socket of the MCP prosthesis, similarly for the two Poisson's ratios.

A range of loads can be taken by the natural MCP joint. The greatest loading is associated with 'pinch' or 'grip' actions, typically when an object is being held. On such occasions there is no movement at the joint and so the lubrication analysis described in this paper would not be appropriate. When movement occurs at the MCP joint loads are generally lighter, for example a typical load across an MCP joint has been offered as 14N due to the balance of muscle forces alone [24]. Another paper described a model which predicted dynamic forces of between 5N and 24N for the MCP joints of an index finger [25]. Therefore 50N was estimated to be the greatest load at which movement of the MCP joint still took place, and for which this lubrication analysis would be appropriate. So, for the series of calculations which involved varying the load, the range of values was taken to be 2N to 50N, with an increment of 5N from 5N onwards chosen.

Finger joints clearly come in a range of sizes and can move at a range of speeds. If an 'average' speed of 1Hz is taken, for a 'typical' finger joint of 7.5mm radius (r) moving through an arc from 0° to 90° and back to 0°, then an average entraining velocity (u) of 11.8mm/s can be calculated using the equation:

$$u = r\omega/2$$

where ω is the angular velocity [9]. Again allowing for slightly higher frequencies and larger sizes of joints gave an estimated upper limit of 30mm/s for this series of calculations involving entraining velocity. The minimum speed was taken as zero and was incremented in 3mm/s steps.

Not all manufacturers disseminate the radii or diameters of their finger prostheses. One of the few that does is Finsbury Orthopaedics who offer MCP prosthesis radii from 4.75mm to 8.5mm for their TMPR™ implant [19]. Therefore, to cover all potential sizes, the series of calculations involving different radii was based on radii between 3mm and 10mm, in 0.5mm steps.

Other relevant values were taken from the literature, such as those given in table 1. In addition, a viscosity of the synovial fluid lubricant of 0.005Pa s was assumed [10]. For a 16mm diameter metal-on-metal hip prosthesis a radial clearance of 30μm has been reported [26]. In addition, tests of metal-on-metal hip prostheses with 22μm and 40μm radial clearances have been detailed [27]. Given that pyrocarbon and zirconia are finished by similar methods to that used for cobalt chrome in metal-on-metal hip prostheses, so it was assumed that similar radial clearances could be achieved. Therefore the radial clearance of each ceramic-on-ceramic, metal-on-metal and pyrocarbon-on-pyrocarbon joint was taken to be 30μm. However, the radial clearance for the metal-on-polymer prosthesis was taken to be 50μm, based on data from metal-on-polymer total hip prostheses having 28mm diameter femoral heads [28]. A summary of all the parameters used in the analysis is given in table 2.

Results

For the various material combinations, the equivalent elastic modulus and compound surface roughness values are given in table 3.

For a 'typical' application of a 7.5mm nominal radius prosthesis, under a 15N load and an entraining velocity of 11.8mm/s, it was found that the ceramic-on-ceramic and metal-on-metal combinations offered the potential of fluid film lubrication with λ ratios of 7.1 and 4.4 respectively, while the pyrocarbon-on-pyrocarbon prosthesis offered mixed lubrication with a λ ratio of 1.9, and the metal-on-polymer combination fell within the boundary lubrication regime with a λ ratio of 0.2.

By varying the entraining velocity from 0 to 30mm/s, the resultant changes in lambda ratio are given in figure 1. As can be seen, the greatest values of lambda ratio are seen with ceramic-on-ceramic (CoC), followed by metal-on-metal (MoM), pyrocarbon-on-pyrocarbon (PyoPy) and finally metal-on-polymer (MoP). Next, the prosthesis radius was varied between 3mm and 10mm. As shown by figure 2, under the chosen test conditions, for most of these sizes a ceramic-on-ceramic (CoC) implant would operate in the fluid film lubrication mode. A metal-on-metal (MoM) implant would be in the fluid film regime above approximately 6mm, and in the mixed lubrication mode below this value. A pyrocarbon-on-pyrocarbon (PyoPy) implant would mostly operate in the mixed lubrication regime and a metal-on-polymer (MoP) implant would always function in the boundary lubrication regime. Figure 3 offers the lubrication regime results when a variation of load over the range 2 to 50N was undertaken. Again, the order from enhanced lubrication to more difficult operating conditions is ceramic-on-ceramic (CoC), metal-on-metal (MoM), pyrocarbon-on-pyrocarbon (PyoPy) and metal-on-polymer (MoP). The results show that ceramic-on-ceramic and metal-on-metal combinations could operate with fluid film lubrication, pyrocarbon-on-pyrocarbon under mixed lubrication and metal-on-polymer in the boundary lubrication regime. As expected, the minimum film thickness increased with entraining velocity and with size of prosthesis, while it decreased as the load increased.

Discussion

The results show that, for the range of operating conditions considered, ceramic-on-ceramic provides the greatest opportunity for fluid film lubrication in the case of MCP implants. Ceramics provide a hard, scratch resistant surface which is likely to retain its low roughness values during operation. However catastrophic in vivo fractures of Moje first metatarsophalangeal prostheses have been seen [29]. Fracture of the proximal component of this toe prosthesis into approximately 50 pieces was reported. The failure of the device was put down to damage occurring during the process of manufacture. Therefore it should be recognised that quality control during production of these potentially brittle ceramic materials needs to be of the highest quality.

The metal-on-metal combination often achieved fluid film lubrication under the operating conditions considered, though occasionally mixed lubrication was predicted. It should be noted that were scratching of the articulating faces to occur, then this damage could increase the roughness and thus move the lubrication regime towards mixed and boundary. If the wear was substantial then this would lead to an increase in the clearance between metacarpal and proximal phalangeal components which in turn would have a negative impact on the lubrication regime through a decrease in the equivalent radius.

Goldsmith et al found that their metal-on-metal hip prosthesis components tended to bed in by polishing each other [30]. This 'self-healing' ability of all-cobalt chrome articulations has been noted elsewhere [31]. As such, when metal-on-metal hip prostheses have been tested, researchers have described a bi-phasic wear pattern, consisting of a higher wear bedding-in phase and a lower steady state wear. As an explanation it has been argued that localised polishing takes place, which reduces roughness values and in turn improves the lubrication. Interestingly, for the smaller (16mm and 22mm diameter) metal-on-metal hip joints tested by Smith et al no bedding in wear was seen [26]. Here it was argued that although Ra values did improve, such an improvement was insufficient to create the positive conditions of full fluid film lubrication seen in the larger diameter prostheses. Extrapolating this opinion from metal-on-metal hip prostheses will probably mean that, for the relatively small size metal-on-metal finger prostheses, no such ‘self-healing’ ability will be likely.

The relatively small size of finger prosthesis may have a number of subordinate effects. From an engineering point of view it might be expected that the smaller components would be more difficult to polish. Therefore, it may be that the smallest sizes of prosthesis have the ‘worst’ surface finish. In addition, optimum polishing of the articulating spherical surface of the concave phalangeal components may be difficult to achieve, compared with the convex metacarpal head. In the case of larger spherical components such as artificial acetabular cups, roughness values of the acetabular cup an order of magnitude greater than those seen on the matching femoral component have been reported [32].

Regarding the pyrocarbon-on-pyrocarbon material combination, this gave results which showed the MCP implant to operate predominately in the mixed lubrication regime. The same concerns regarding increases in surface roughness could apply to pyrocarbon-on-pyrocarbon as to metal-on-metal. Furthermore, as scratch resistance is related to material hardness, the fact that pyrocarbon has a lower hardness than cobalt chrome raises concerns. Pyrocarbon is said to have a hardness of 240DPH (diamond point hardness) [33] compared to cobalt chrome with a hardness of 400DPH [34].

From the calculations in this paper it has been shown that the metal-on-polymer combination will operate in the boundary lubrication regime. Potentially this can have negative consequences due to higher wear and higher friction. However, tribology is a complicated science. For example, the particular lubricant may have an influence on friction and wear. Similarly, surface films may be formed which could promote or negate wear. Therefore appropriate in vitro testing is required. Of the metal-on-polymer implant designs mentioned above, the Zimmer Elogenics™ prosthesis was tested in a lubricant of dilute bovine serum, under a mean dynamic load of 12.5N, while the test prosthesis was oscillated though a 90° arc of motion at a speed of 1Hz. After 3000 of these dynamic loading cycles a static load of 100N was applied for 45 seconds before the whole combined load cycle began again. Therefore a test period when the entraining velocity was zero was included. Testing ran to 5,000,000 cycles of flexion-extension, at the end of which the wear of the two 5mm radius test prostheses was considered to be acceptably low for this metal-on-polymer combination [18]. Scratching of the polymer surfaces was seen, and this would likely indicate that mixed or boundary lubrication took place.

The positive results for ceramic-on-ceramic and metal-on-metal, and to a lesser extent pyrocarbon-on-pyrocarbon, need to be set against a recognition that when the finger is not moving, the entraining velocity is zero and so surface contact will occur, with a concomitant potential for wear. With wear will likely come an increase in the roughness of the articulating surfaces of the prosthesis, and an increase in the radial clearance. Both of these factors are likely to take the joint from fluid film towards a mixed lubrication regime in the case of ceramic-on-ceramic and metal-on-metal, and towards boundary lubrication, in the case of pyrocarbon-on-pyrocarbon.

It should be noted that changes in lubricant viscosity would influence the calculated minimum film thickness and thus the lambda ratio. In this paper a viscosity of 0.005 Pa s was taken [10]. However, a higher value of 0.01 Pa s has been reported [35] as has a lower value of 0.003 Pa s [36]. In addition it is recognised that the viscosity of synovial fluid depends upon the shear rate at which it is measured and whether it has originated from ‘normal’, osteoarthritic or rheumatoid joints [37-38]. To indicate the influence of viscosity on lambda ratios, figure 4 shows the results for a ceramic-on-ceramic implant of 7.5mm radius at 15N load and at a range of entraining velocities of 0 to 30mm/s, at viscosities of 0.003, 0.005 and 0.01 Pa s. As can be seen increasing viscosity leads to greater lambda ratios and therefore improved lubrication. Similarly, were the radial clearance of the implant to increase, then the lambda ratios would be reduced and the lubrication regime could be negatively influenced. It would be useful to correlate the theoretical results reported in this paper with the surface topography of explants, but such data is currently unavailable in the literature.

Returning briefly to the resurfacing metal-on-metal hip prostheses mentioned in the Introduction, these can operate in the fluid film lubrication mode [9]. This outcome is due to their larger radii and greater entraining velocity compared with the MCP implants considered here, but is offset by the much higher loads seen at the hip compared with the MCP joints. It is felt that these differences, inherent in diverse joints around the body, should be appreciated by those concerned with lubrication analysis of various prostheses.

Conclusion

Lubrication theory indicates that, for the material combinations considered for application to two-piece MCP prostheses, ceramic-on-ceramic and metal-on-metal designs could operate with fluid film lubrication, pyrocarbon-on-pyrocarbon is expected to function in the mixed lubrication mode, while metal-on-polymer designs are likely to operate in the boundary lubrication regime.

However, it should be recognised that other factors may come into play when choosing a particular implant and material combination for the MCP joints.

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Figure 1: Variation of lambda ratio with entraining velocity

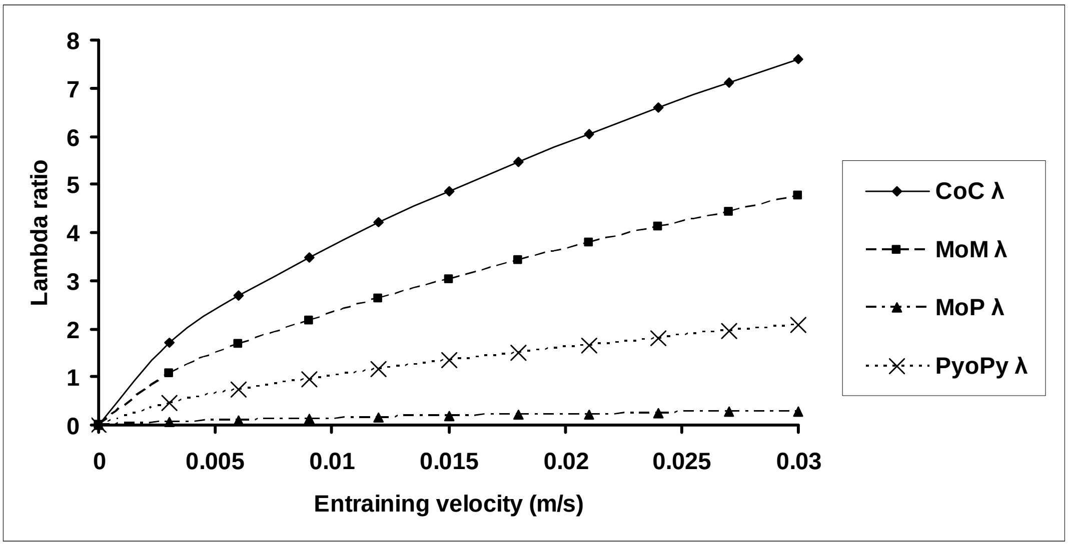

Figure 2: Variation of lambda ratio with prosthesis radius

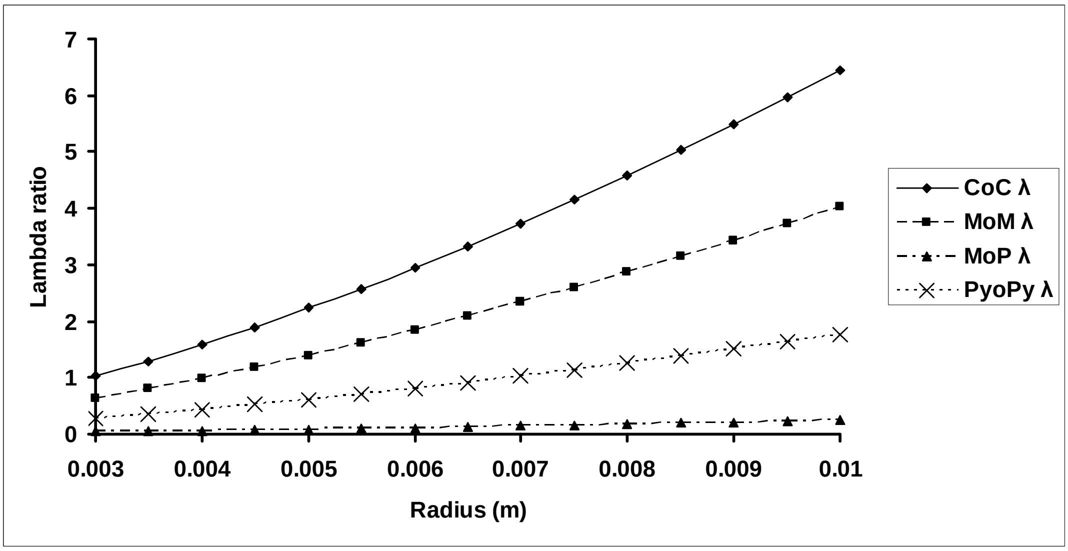

Figure 3: Variation of lambda ratio with prosthesis load

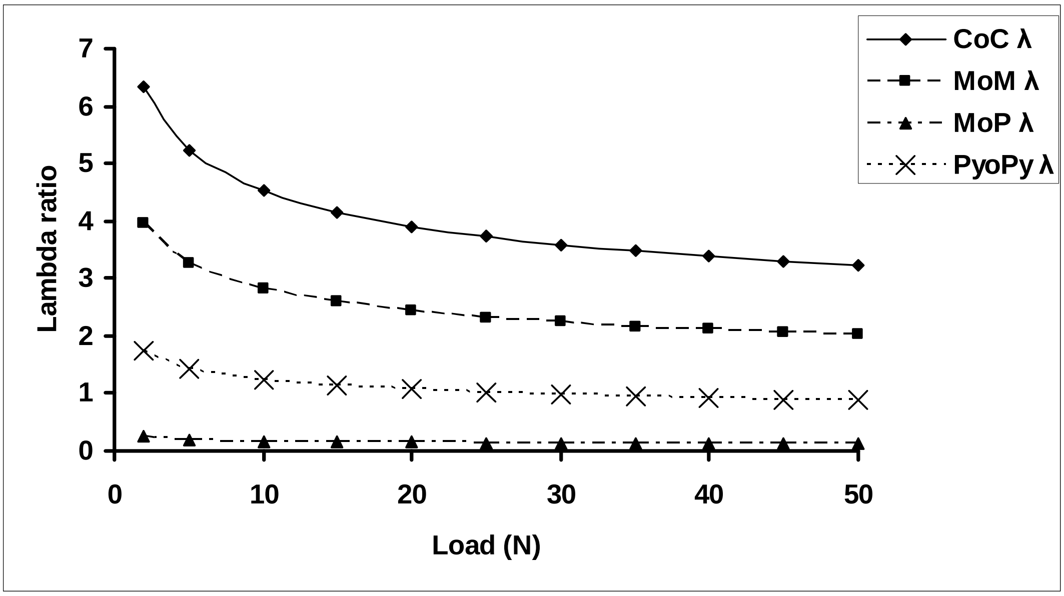

Figure 4: Variation of lambda ratio with entraining velocity for different viscosity values

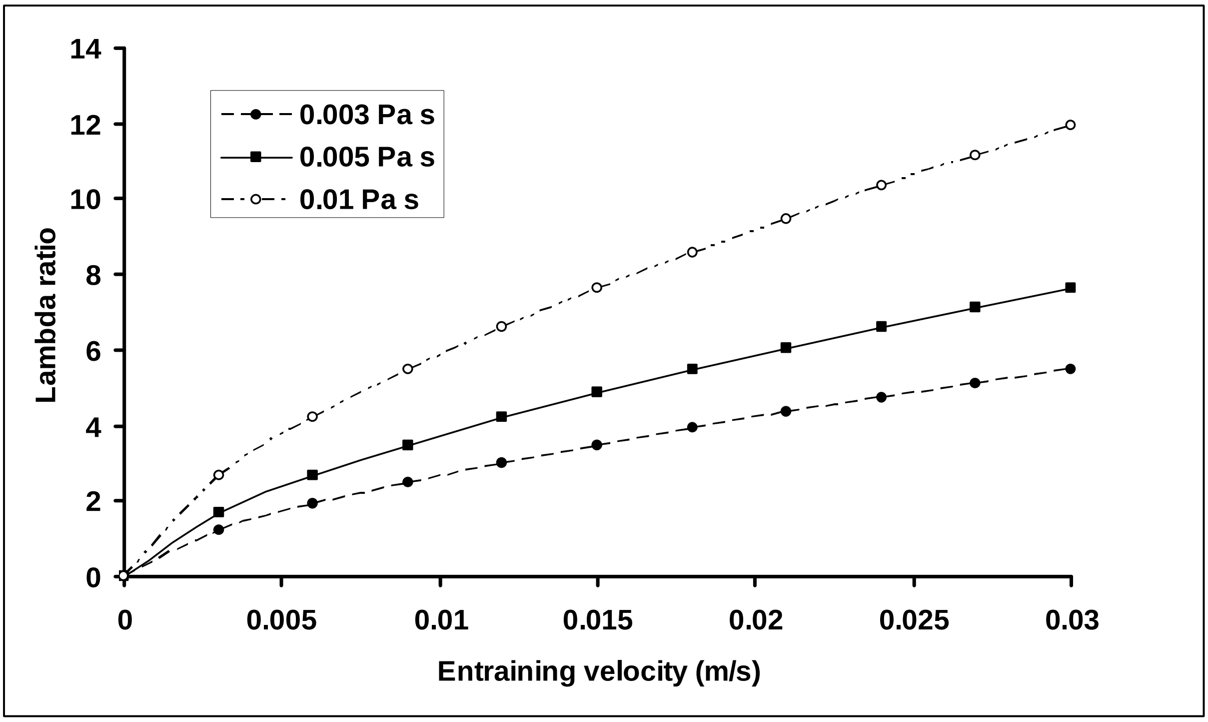

Table 1: Material properties

Material	Young's Modulus (GPa)	Ref.	Poisson's Ratio	Ref.	Roughness (nm) ball and socket	Ref.
Cobalt Chrome	210	[9]	0.3	[9]	3 10	[26]
UHMWPE	1	[22]	0.4	[10]	1290 (socket)	[32]
Pyrocarbon	29.4	[14]	0.3	[39]	40 40	[33]
Zirconia	198	[35]	0.29	[35]	3 6	[35] [40]

Table 2: Summary of values used in the analysis

Parameter	Constant value	Range
Load	15N	2-50N
Entraining velocity	11.8mm/s	0-30mm/s
Radius of metacarpal head	7.5mm	3-10mm
Viscosity	0.005Pa s	---
Radial clearance	0.03mm	---
Radial clearance (metal-on-polymer)	0.05mm	---

Table 3: Equivalent elastic modulus and compound surface roughness values

Material combination	Equivalent Elastic Modulus (GPa)	Compound surface roughness $\sqrt{Ra_1^2 + Ra_2^2}$ (nm)
Ceramic-on-ceramic (Zirconia and Zirconia)	216	6.7
Metal-on-metal (Cobalt Chrome and Cobalt Chrome)	236	10.4
Pyrocarbon-on-pyrocarbon (Pyrocarbon and Pyrocarbon)	32.2	56.5
Metal-on-polymer (Cobalt Chrome and UHMWPE)	1.9	1290