LUBRICATION MODEL OF A KNEE PROSTHESIS, WITH NON NEWTONIAN FLUID AND POROUS ROUGH MATERIAL

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***Abstract*— Tibial component of knee prostheses, made of ultra high molecular weight polyethylene (UHMWPE), experiences a high degree of wear and may be expected to last twelve years on average. In this work, a steady state one-dimensional lubrication model of a knee prosthesis is solved through a numerical technique based on the Finite Element Method. The model takes into account a non Newtonian synovial fluid, its ultra filtration mechanism and the surface roughness of a porous elastic layer on the tibial component. The benefits of a porous compliant material placed at the top of the metallic tibial component are shown taking into account the stiffness and exudation capacity of the material and hyaluronic acid concentration of synovial fluid.**

***Keywords*— knee prosthesis, non-Newtonian fluid, elastohydrodynamic lubrication, wear, finite elements method.**

I. INTRODUCTION

Ultra high molecular weight polyethylene (UHMWPE) wear debris are currently the major cause of aseptic loosening of knee prostheses (Williams, 2003). These wear particles remain within the synovial fluid surrounding the implant and stimulating ostheolysis at the bone-implant interface. By this fact, the prosthesis is loosened and may be expected to last on average twelve years.

The interest in giving a solution to failures of knee prostheses is linked to the increasing demand of these implants by patients under 40. Young patients are generally active people, who might require several revisions (new replacements) during their lives. Therefore, the development of new materials and the improvement of the current polyethylene mechanical response are required in order to produce more durable prostheses.

Previous theoretical works showed that the optimization of the joint lubrication mechanism is a fundamental issue in improving the mechanical performance of the healthy knee. Both the low rigidity and porosity of the natural cartilage make the knee a well lubricated joint when it is working under high loading conditions. The self-lubrication mechanism promoted by the cartilage porosity has been shown to be one of the main reasons in producing very low friction (Corvalán *et al.*, 1999; Di Paolo *et al.*, 1998). By this mechanism, the surfaces in contact of the joint appear to be separated by a lubricant film for any loading condition, avoiding direct contact.

In the past few years, much interest has been directed toward the application of the elastohydrodynamic theory to artificial joints design. The minimum film thickness was predicted by Jin *et al.* (2003), based on the analysis of an artificial hip joint consisting of a metal femoral ball articulating against UHMWPE cups. Various comparisons of the predicted minimum film thickness were made between different bearing surfaces, elastic models, loading conditions and steady-state and transient conditions. They concluded that the minimum film thickness under cyclic conditions can be accurately estimated by a steady-state model using the column model for the computation of the elastic UHMWPE deformation.

The elastohydrodynamic lubrication theory applied to linear contacts was used by Di Paolo and Berli (2006) for a model of a metal-on-compliant layer knee prosthesis. The model considers the non-Newtonian (pseudoplastic) characteristic of the synovial fluid and the capacity to exude and absorb fluid by the compliant layer. It was shown that the pseudoplastic characteristic of the fluid benefits the lubrication mechanism by reducing the fluid viscosity when the film is narrowing. Consequently, the shear stresses at the compliant surface are reduced. In the same work, it was predicted that a material with the stiffness of the UHMWPE generates thinner fluid films than the surface roughness, thus promoting abrasive wear. At the same time, the generated pressure within the lubricant exceeds the design limit proposed by prostheses manufacturers, which could lead to an early failure by fatigue. Such predictions indicated that a material with similar stiffness to that of the cartilage would work more efficiently than the current polymeric replacements. Specifically, the film thickness for cartilage-like materials appears to be eight times higher and the maximum pressure three times lower than those for the current polyethylene. One of the main restrictions of Di Paolo and Berli (2006) is that their model does not consider the surface roughness, which may considerably influence polyethylene wear rates (Berli *et al.*, 2005).

In an experimental work, Kawano *et al.* (2003) induced osteoarthritis in 40 rabbits and exposed a group of them to periodic intraarticular administration of high molecular weight Hyaluronic Acid (HA). Their results showed that the administration of HA reduces the friction coefficient of damaged cartilages to the values found in healthy ones. This shows the crucial role of HA on the knee lubrication process and thus the importance of including their effects in knee prostheses models.

In this work a lubrication model of a knee prosthesis with a metallic femoral component and a tibial component with a compliant porous layer is presented. The

hypothetical material adhered to the tibial component is capable of absorbing and exuding synovial fluid by means of a porous structure. The surface roughness is modeled as a harmonic wave and a power law is employed to account for the non Newtonian (pseudoplastic) behavior of the synovial fluid. Furthermore, the fluid viscosity dependence on the HA concentration is taken into account. The highly coupled equation system was simultaneously solved by means of a technique based on the Finite Element Method (FEM).

The material features and the fluid properties are the most important factors for a good operation of the artificial joint. For this reason, a set of numerical experiments were performed in order to evaluate the mechanical behavior of the hypothetic material. The aim of this work is to study the influence of the elastic modulus, the exudation factor and the concentration of HA on the joint operation conditions.

II. MODEL

Figure 1 depicts a scheme out of scale of the considered geometry to predict the performance of the proposed knee prosthesis. The femoral component (upper part) is metallic and the tibial component (lower part) is coated by a porous compliant material with harmonic surface roughness. In this work, a steady state relative motion between both components is assumed. Thus, this working condition represents the most exigent situation during the stance phase of the normal gait, where the applied load on the knee is three times greater than the body weight (Andriacchi and Hurwitz, 1997).

The equivalent model is represented by a cylinder in linear contact over a plane with roughness (see Fig. 1) (Dowson and Higginson, 1977). The governing equations come from the principles of mass and momentum balances applied to the simplified geometry in Fig. 1 under the following hypotheses:

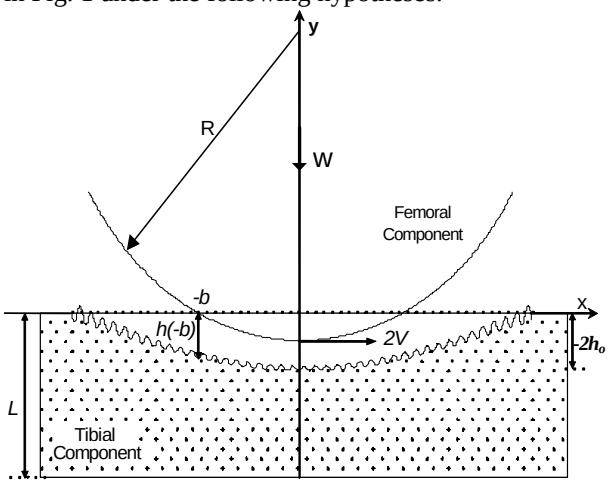

Figure 1. Equivalent model of a knee prosthesis having a rough porous tibial material. The scheme shows the joint's movement, when the tibial component of the equivalent model has been deformed by the fluid pressure.

1. Lubrication approximation.
2. Steady state, laminar, incompressible and one-dimensional flow.
3. Non Newtonian fluid modeled according to a constitutive power law equation (Di Paolo and Berli, 2006; Ribitsh, 1990; Wang and Zhang, 1987).
4. Variable viscosity depending on the HA concentration.
5. Non-deformable femoral component.
6. Tibial component covered with a thin porous layer restricted to plain strain.
7. Constant elastic material parameters.
8. Lubrication zone (contact area) wider than the thickness of the deformable porous layer.
9. Negligible flows into the material.
10. Surface roughness modeled as a harmonic wave (Dowson, 1990; Berli *et al.*, 2005).
11. Constant temperature.

The application of hypotheses 1-4 to the conservation laws leads to a modified Reynolds' lubrication equation (Di Paolo and Berli, 2006). The integral form of this equation is:

$$p(x) = 12n \int_{-b}^{x} \mu_1(x) \left[\frac{h(-b)}{h(x)} \right]^{n-1} \frac{V\,h(x) - q_e(x)}{h(x)^3} dx \qquad (1)$$

where $p(x)$ is the pressure, $h(x)$ is the lubrication film thickness (gap between the femoral and tibial components), $q_e(x)$ is the flow rate and n is the exponent of the constitutive power law for the non – Newtonian characteristic of synovial fluid, whose *in vivo* value remains unknown. Therefore, n is used as a parameter in the computational code, whose values are ranged between 0 and 1. The n=0.60 value selected by Di Paolo and Berli (2006) in a previous paper was used for this work. Finally, $\mu_1(x)$ is the synovial fluid viscosity which depends on the HA concentration.

Since the size of the HA molecule is greater than that of the material pores (i.e., the HA does not enter into the compliant material), its concentration will be increased or decreased according to the amount of plasma absorbed or exuded by the deforming material, respectively. Thus, the HA concentration will change — depending on the flow rate— along the x-direction of the lubricating film.

Negami (1960) demonstrated that the relationship between the synovial fluid viscosity and the HA concentration is well described by a linear expression. Then, a macroscopic mass balance between the inlet section of the lubrication zone and a section located in a generic position allows to describe the variations of the viscosity as a function of the flow rates between both sections:

$$\mu_1(x) = \mu_0 \left\{ 1 + NVIS \left(\frac{q_e(-b)}{q_e(x)} - 1 \right) \right\} \qquad (2)$$

where $NVIS = KC_0/\mu_0$, K is a constant obtained from the bibliography (Corvalán *et al.*, 1999), and C_0, μ_0 and $q_e(-b)$ are the concentration of the HA, the synovial fluid viscosity (without HA) and the flow rate at the inlet of the lubrication zone.

The boundary conditions imposed to equation 1 are:

$$x = -b,\ p = 0 \qquad (3a)$$

$$x = \bar{x},\ p = 0 \qquad (3b)$$

$$x = \bar{x} \quad , \frac{dp}{dx} = 0 \tag{3c}$$

where $\bar{x}$ is a position at the lubrication channel exit whose localization is unknown and becomes a free boundary. Condition 3c avoids subambient pressure results (cavitations) which are characteristic of industrial bearings but not of human joints.

After applying boundary conditions (3) to Eq. (1), the following equations were obtained in order to determine the free boundary location and the entrance flow rate:

$$0 = \int_{-b}^{\bar{x}} \mu_1(x) \left[\frac{h(-b)}{h(x)} \right]^{n-1} \frac{V\,h(x) - q_e(x)}{h(x)^3} dx \tag{4}$$

$$V h\big|_{\bar{x}} = q_e\big|_{\bar{x}} \tag{5}$$

In this work, the surface of the tibial material is considered to be rough; this characteristic is represented by a harmonic wave. Since the pressure developed in the lubrication zone deforms the material surface (see Eq. 8 below), the film thickness is expressed as follows:

$$h(x) = -2h_0 + \frac{x^2}{2R} + A\,sen(\omega x) + d(x) \tag{6}$$

In Eq. (6), $2h_0$ is the intercrossing between the femoral and tibial components in an undeformed state, and $d(x)$ is the deformation of the compliant layer at each point throughout the lubrication zone. The roughness is represented by the term A sen(ωx), where the amplitude A for the harmonic wave is related to experimental measures of roughness as follows:

$$A = \sqrt{2}\, Ra \tag{7}$$

where Ra is the experimental value of the roughness.

The deformation (d) (see hypotheses 6-8) is proportional to the load applied at each point (local pressure) accordingly to a column model as expressed in the following relationship (Corvalán *et al.*, 1999; Di Paolo *et al.*, 1998; 2006):

$$d(x) = \frac{L}{E''}\; p(x) \tag{8}$$

where $E'' = (1-\nu)\, E \,/\, [(1-2\nu)\,(1+\nu)]$; E and ν are the elastic modulus and the Poisson's ratio, respectively, and L the compliant layer thickness.

The capability of the compliant layer to exude and absorbe fluid is expressed in the following flow rate equation:

$$q_e(x) = q_e(-b) + \;2\; V\,\theta\; d(x) \tag{9}$$

In Eq. (9) θ is the so-called exudation factor and represents the absorbed-exuded fluid volume per unit of deformed volume.

The fluid pressure supports the load avoiding the direct contact between solids. Therefore, the load w can be obtained from the following expression:

$$w = \int_{-b}^{\bar{x}} p(x)\; dx \tag{10}$$

The tangential force (friction) on each lubricated surface is obtained by integrating the shear stresses:

$$f_{0,h} = \int_{-b}^{\bar{x}} \tau_{0,h}(x)\; dx \tag{11}$$

where the suffix 0 represents the surface at y=0 (tibial component) and the suffix h represents the surface at y=h (femoral component).

Finally, the hydrodynamic friction coefficient for each surface is obtained as follows:

$$\phi_{0,h} = \frac{f_{0,h}}{w} \tag{12}$$

III. METHOD

Equations (1), (2), (6) and (9) constitute a highly coupled system of non-linear equations to be solved in a domain that is unknown. Thus, the model has no analytical solution and must be solved by numerical procedures. Key features of the computational technique are given below:

1. Finite element method —Galerkin formulation— using piecewise-linear trial functions (see Di Paolo *et al.* (1995) for a detailed explanation of the FEM-Galerkin method applied to the integral form of the Reynolds equation).
2. Simultaneous solution of the coupled system by the Newton method, involving the determination of the free boundary location at each iteration.
3. Numerical grid adaptation as function of the free boundary position
4. Parametric continuation algorithm with automatic step control.

A one-dimensional implementation of the Spines method (Kistler and Scriven, 1983, Di Paolo *et al.*, 2006) was used to determine the free boundary and to simultaneously update the nodal distribution. The only nodes whose positions are updated are those located between the center of the lubrication zone (x = 0) and the free boundary ($x = \bar{x}$), while the other nodes remain fixed. The program was implemented in FORTRAN and can be efficiently executed in personal computers. The primary variables to be determined are: the pressure at each node, the intercrossing between solids in their undeformed state ($2h_0$), the flow rate at the inlet of the lubrication zone ($q(-b)$) and the free boundary location. This means that if the domain is discretized with NN nodes there will be $NN+3$ unknowns. The other variables, (layer deformation $d(x)$, film thickness $h(x)$, and flow rate $q(x)$) are obtained in a post-processing stage. The power law coefficient n, the exudation factor θ, the load w and $NVIS$ were used as parameters in the parametric continuation process.

Since μ_0 and K are constant in this work, changes in $NVIS$ obey only to changes in the HA concentration (C_0).

Both the wavelength and the amplitude of the roughness remain constant. Their values were obtained from a previous work (Berli *et al.*, 2005). The amplitude value (see table I) was calculated from eq (7) considering a roughness value which belong to a roughness range that is common to the polyethylene and the natural cartilage (Kurtz *et al.*, 1999).

Table I: Physical and operative parameters of an implanted knee prosthesis.

Parameter	Symbol	Magnitude
Radius of the equivalent cylinder	R	0.70 [m]
Mean surfaces tangential velocity	V	$1.91.10^{-2}$ [m/s]
Synovial fluid viscosity (healthy)	μ_0	1.00 [Pa s]
Elastic modulus.	E	20 and 500 [MPa]
Viscosity exponent of the lubricant fluid	n	0.60
Exudation coefficient	θ	0.00 - 1.00
Poisson´s ratio	ν	0.40
Compliant layer thickness.	L	1.00×10^{-3} [m]
Load	W	$7.36 \cdot 10^{4}$ [N/m]
Sine wave amplitude	A	1.3 μm

Equations (1), (4), (5) and (10) are used to generate the *NN+3* weighted residues from which the unknowns are determined.

Finally, the load applied on the joint was three times the body weight of 736 N.

IV. RESULTS AND DISCUSSION

Abrasion and fatigue are the main mechanisms of polyethylene wear. The former is a consequence of the contact between polymer and metal during a full movement phase of the gait. The latter is related to inner fracture of the material caused by high normal cyclic loads and high shear stresses transferred to the material from the fluid.

A. Elastic Modulus

Figure 2 shows the film thickness for two different elastic moduli (i.e. 20 and 500 MPa) considering a layer with and without roughness. It can be seen that the film thickness, in all the situations, is smaller than the

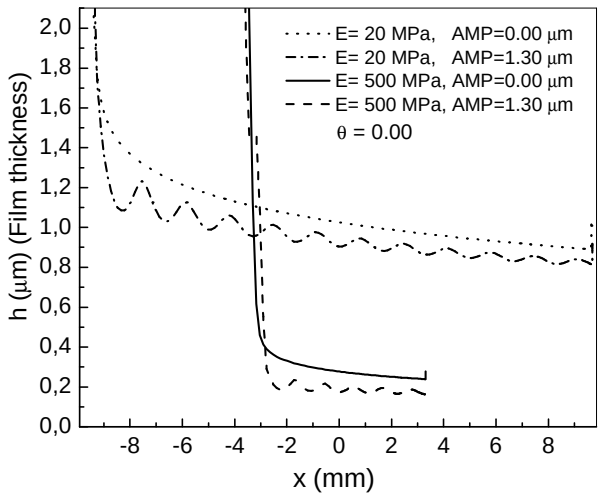

Figure 2: Film thickness for a compliant layer of two different elastic moduli: with and without roughness.

undeformed roughness (roughness before deformation). Under these circumstances, the existence of the lubricating film between the prosthesis components could be possible because the fluid pressure deforms the surfaces reducing their original roughness amplitude.

The film thickness obtained for the stiffer rough material (500 MPa) is approximately 28% smaller at the center of the contact than that of the same material without roughness. Figure 3 shows the pressure field in the lubrication zone for the same conditions shown in Fig. 2. The pressure exhibits a maximum value (16 MPa) that is 60% higher than the fatigue design limit for UHMWPE. This value is around 10 MPa, half the elastic limit of the polyethylene (i.e. 21 MPa, see Pappas *et al*, 1987; Lee and Pienkowsk, 1997). The situation becomes more severe in presence of surface roughness, for which the pressure takes a maximum value (18 MPa) up to 80 % above the fatigue design limit.

These results suggest that, under lubricating conditions, fatigue could be the main cause of wear since the normal stresses (pressure) on the material exceed the design limits in all cases, while the lubrication fluid film remains thick enough to avoid the direct contact between the prosthesis components. However, in real situations, a free-of-contact condition for the 500 MPa material can not be guaranteed, because in some cases the predicted film thickness is very thin (H < 0.16 μm, 82% thinner than the non deformed roughness). Otherwise, the lubricating film for the 20 MPa material is thicker than 0.8 μm (i.e. 13% thinner than the non-deformed roughness). This is a more reliable film thickness than that found for the rigid one.

B. Exudation factor

Polyethylene has neither the ability to exude nor to absorb fluid as the articular cartilage does. Previous works showed that this capacity permits the cartilage to self lubricate, being this one of the main causes of the low friction coefficients that living joints exhibit (Corvalán *et al.*, 1999; Di Paolo *et al.*, 1998).

Figure 4 shows higher values of friction coefficient for the 20 MPa material than for the 500 MPa one. It

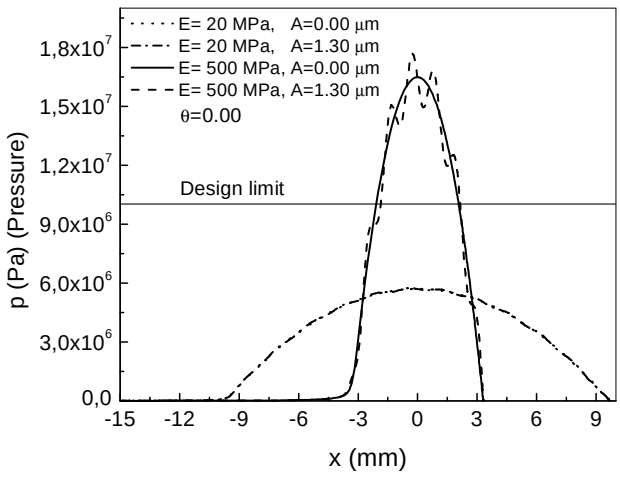

Figure 3: Fluid pressure in the contact zone for the fluid films shown in Fig. 2.

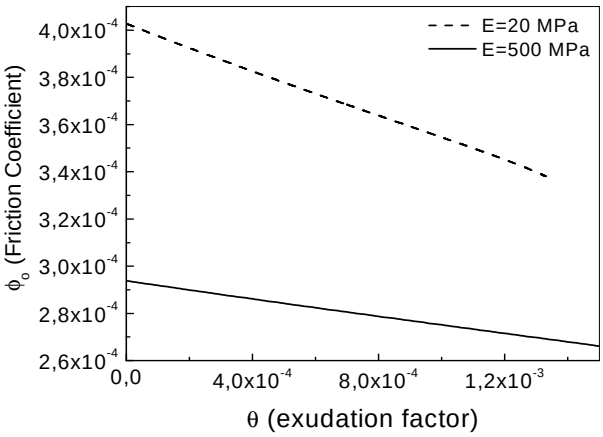

Figure 4: Friction coefficients versus exudation factor for *E*=20 MPa and *E*=500 MPa. Note that the slope of the curve for *E*=20 MPa is higher than for *E*=500 MPa.

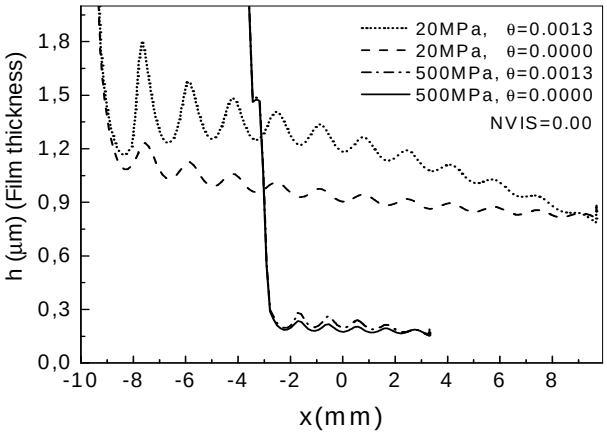

Figure 5: Influence of exudation factor on the film thickness.

should be noted that although the film thickness is 5 times thicker for the compliant material than for the 500 MPa one (see Fig. 5), the flow rate is 400 times greater (see Fig. 6) and the region where high shear stresses are distributed is 3 times wider for the first one. However, a reduction of the friction coefficient could be attained for the compliant material if it has non-zero exudation factor value, as Fig. 4 shows.

Although the material with elastic modulus of 500 MPa show minor sensitivity to variations of θ it is clear from Fig. 4 that it also could take advantage from a porous structure.

Figure 5 shows that the film thickness becomes thicker for both elastic moduli (i.e. 20 MPa and 500 MPa) as the exudation coefficient increases. On the other hand Fig 6 shows that a material with exudation capacity generates a greater flow rate in the lubrication zone than that observed for the material without exudation capacity. Thus, the film must be increased to allow such an increment of flow rate as Fig. 5 predicts.

The constant value in Fig. 6 represents the flow rate for the non-porous material, which is constant because no liquid can be exuded or absorbed from the polymeric matrix. Otherwise, the exudation capacity causes a noticeable increment of fluid at the lubrication zone, while the required flow from the surroundings (at the inlet of

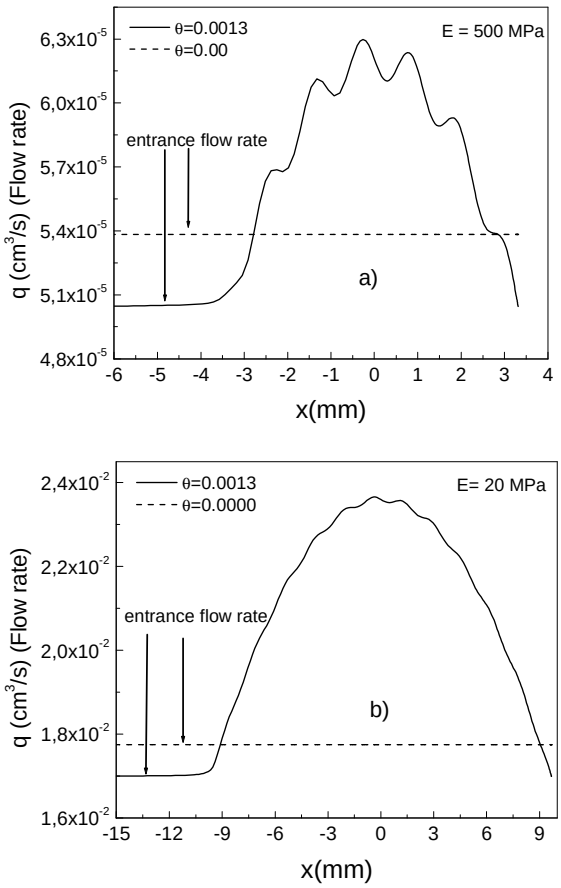

Figure 6: Flow rate at the lubrication zone for both a) 500 MPa and b) compliant material. The curves corresponding to porous materials (materials with exudation capacity) show lower flow rates at the inlet of the lubrication zone than the non-porous ones.

the lubrication zone) is reduced with respect to the non-exudation case. This phenomenon can be thought as an auto-lubrication mechanism (noted before by Corvalán *et al.*, 1999; and Di Paolo *et al.*, 1998), by which low friction coefficients can be obtained with a minimum requirement of surrounding fluid

C. HA Concentration

Application of HA intra-articular injections to patients with arthritic pathology is a usual medical practice to reduce knee pain.

The pseudoplastic characteristic of the synovial fluid is responsible for the viscosity reductions that counteract the increment of velocity gradients when fluid film stretches under high loads. At the same time, a material with the ability to exude and absorb fluid could increment the flow rate at the lubrication zone, diluting the HA and promoting an extra-viscosity-reduction (see Eq. (2)). This effect can be observed for both materials (see figs. 7 a) and b)). Although not shown for brevity, this mechanism does neither affect the fluid film thickness, nor the velocity gradients nor the pressure field. Thus, the friction coefficient is diminished only by viscosity reduction, as shown in Fig. 8. The range of explored values for the *NVIS* parameter was narrower for the

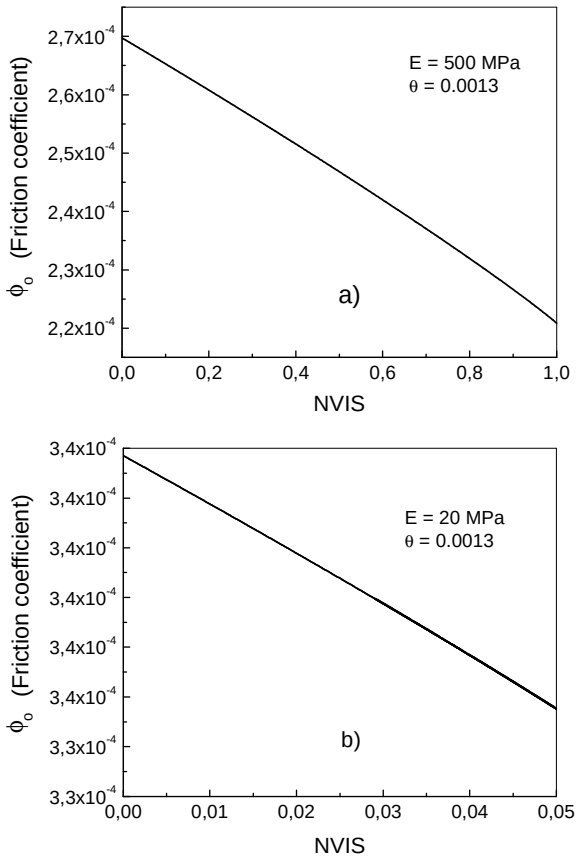

Figure 7: HA concentration effects on the friction coefficient at the lubrication zone for: a) material of 500 MPa and b) material of 20 MPa. NVIS= 0.00 corresponds to one phase fluid (C_0 = 0).

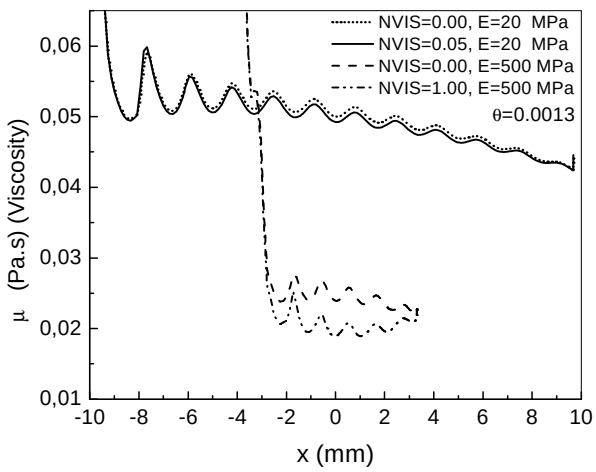

Figure 8: Fluid viscosity changes due to dilution process of HA at the lubrication zone.

compliant material than for the rigid one due to program convergence problems.

V. CONCLUSIONS

A lubrication mechanism model of knee prostheses was numerically solved, considering a rough compliant porous material on the tibial component; the synovial fluid was considered as non Newtonian fluid modeled by a power law constitutive equation, taking into account its viscosity dependency on the HA concentration.

The objective of this work was to study the influence of the elastic modulus and the exudation capacity of a compliant layer placed on the tibial component, together with the HA concentration in the synovial fluid on the principal physical variables affecting the prosthesis wear.

The results show that a compliant material with the capacity of exuding and absorbing fluid has some mechanic advantages over non-porous UHMWPE because of the following reasons:

- A larger deformation capacity promotes thick fluid films even with surface roughness, i.e., elastohydrodynamic lubrication.
- The pressure field in the lubrication zone is distributed over a wider region. As a result, the pressure values remain under the stress design limit. The maximum pressure for a rough UHMWPE exceeds 80 % the design limit.
- The material capability of absorbing and exuding fluid allows a reduction of the friction coefficient for both low and high rigidity materials.
- The exudation-absorption mechanism generates a self-lubrication process that maintains a larger amount of fluid at the lubrication zone. This characteristic would benefit both the compliant material and the UHMWPE by reducing the shear stresses on the material.
- The exudation process observed in the porous material generates a variable HA concentration at the lubrication zone that diminish the lubricant viscosity, and therefore, the friction coefficient on the material adhered to the tibial component.

Currently, the most promising compliant materials are some hydrogels (Yasuda *et al.*, 2005) which have similar properties to those of the natural cartilage, including capability of absorbing and exuding fluid.

The results presented in this work could guide the new materials experimentation for knee prostheses with the mechanical characteristics analyzed here.

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