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Influence of the Prosthetic Parameters on the Durability of the Material Couple Constituent: The Link of a Femoral Prostheses of Hip

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Abstract: The constant evolution of an artificial hip tries to put in presence of new articular surfaces perfectly spherical, susceptible to articulate in a prolonged way with reduced liberation of wear debris. The effect of the prosthetic parameters on the durability of the material couple used in the joint ball of the femoral side and cupule of the cotyledon side is studied. The distribution of the contact pressure has been calculated while introducing the three components of the load applied on the femoral head of the hip creating three components of the movement that have been respectively considered for the calculation of slippery distance. The analyses show that the volume of wear increases with the patient weight and the roughness of the articular surfaces. It increases up to a maximum with increasing of the femoral diameter and the modulus of elasticity then it decreases and stabilizes in an asymptotic manner.

Key words: Wear volume, total hip arthroplasty, mathematical modeling

INTRODUCTION

The prostheses of hip reduced to its simplest expression, it is the setting in contact of its articular faces (ball-cupule). The setting in contact doesn't function, intrinsically, that the price unavoidable of a wear. This wear entails with the passing of the years of use a wear of the articular surfaces and becomes cumulative with the passing of the time. The problem of wear is multi-factorial, susceptible to be influenced by prothetic parameters.

The production of the wear debris and the inflammatory answer of cloth are among the main problems limiting the clinical results of the total replacements of the hip (Harris, 1995; McGee *et al*., 1997). The wear of the mating situated between the implant and the cotyle can occur and depend on the couples of materials of the link. A field of research is developed to obtain a complete understanding of the factors influencing the behavior of use of the link, in order to reduce the periprosthetic load and develop more mating resisting wear. Some clinical studies are based on radiographic studies and on the analysis of use of the sought components.

It is well known from the literature that the primary wear rate is influenced by many factors, including the level of activity of the patient (Seedom and Wallbridge, 1985), the main femoral diameter (Hoeltzei *et al*., 1989; Hali *et al*., 1998), the condition of the surface (Wang *et al*., 1998), the cup orientation in the pelvis (Del Schutte *et al*., 1998), the conception of coupling (Bertei *et al*., 1985; Jin *et al*., 1994) and the quality of the polyethylene (Gomez-Barrena *et al*., 1998).

Because of the complexity of the problem, the statistical analysis cannot show any report between the clinical data and the specific parameters studied. The goal of our work is to gather a mathematical formulation of the problem in order to give an appropriated role to each of the variables identified for the study. Several approaches are used currently to study the factors influencing the rate of wear of the link kneecap situated between the implant and the cotyle in order to improve the couples of materials of the mating while increasing its durability.

A mathematical approach can imply the Finite Element Method (FEM), that can be used to simulate the real geometry, (Maxian *et al*., 1996, 1997). It was found that, (Bertei *et al*., 1985) a good agreement between the pressure of contact foreseen on the cups using the FEM and the analytic method.

In the present, the effect of contact in the link of the couple of thighbone pack-head materials and the influence of the physical and geometric parameters on the durability of the ball-cupule joint of the prosthesis are studied.

In this study the mechanical features of the total hip prostheses are reported in the following Table 1.

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Table 1: Mechanical characteristics of the materials of the total prosthesis of hip

Prostheses	Implant of titan	$E_h = 110GPa$	$v_h = 0.3$
femoral	Femur : AU4G	$E_j = 73GPa$	$v_f = 0.18$
	PMMA	$E_p = 3.5GPa$	$v_p = 0.3$
Cup	Cotyle	$E_c = 0.9GPa$	$v_c = 0.4$

MATERIAL AND MATHEMATICAL FORMULATION

The superior part of the implant possesses an elastic isotropic spherical femoral head that represents the part male of the link kneecap femur/pelvis. The female part of the link kneecap situated in the pelvis is a hemispheric cup manufactured to permit the installation of the cotyle. The couple of the materials are subjected to a normal load q normal applied on the surface. The wear volume of the surfaces (V) can be calculated by the following equation considering the sliding distance (x), (Archard, 1953):

$$V + Kqx \tag{1}$$

where K is the wear factor, which can be determined experimentally and depends on the materials of contact. Equation (1) is discredited in N times, as follows:

$$V(\theta,\phi) = k\sum_{t=1}^{n}\sigma_t(\theta,\phi)x_t(\theta,\phi) \tag{2}$$

Or σ (θ,ϕ) is the local pressure applied on the contact area and x_t (θ,ϕ) is the slide distance definite as the relative displacement of the existing point (Fig. 1).

$$\begin{aligned} x(\theta,\phi,1) &= r_x(\theta,\phi)\Delta\xi(_t) + r_y(\theta,\phi)\Delta\eta(_t) \\ &+ r_z(\theta,\phi)\Delta\varsigma(_t) \text{ or } r_x(\theta,\phi),\ r_y(\theta,\phi) \text{ et } r_z(\theta,\phi) \end{aligned} \tag{3}$$

are the distances between one point on the cup and the absolute axes X, Y, Z, respectively.

$\Delta\xi(_t)$, $\Delta\eta(_t)$ and $\Delta\zeta(_t)$,are the angular variations of the components of the distance r(t), respectively $r_x(\theta,\phi)$ $r_y(\theta,\phi)$ and $r_z(\theta,\phi)$ with the absolute axes for a temporal step. When these angular functions describe the rotation of the hip during a cycle of gait (Sutherland *et al.*, 1980), the Eq. (3) brings closer the complete movement produces by the hip during the gait.

The transformation of coordinates of point m is the system (X_c, Y_c and Z_c) is given by the relations listed

$$\begin{cases} X_c = X\sin\eta + Y\cos\eta \\ Y_c = Z\sin\beta - (X\cos\eta - Y\sin\eta)\cos\beta \\ Z_c = Z\sin\beta + (X\cos\eta - Y\sin\eta)\cos\beta \end{cases} \tag{4}$$

Where, X, Y and Z absolute system and X_c, Y_c and Z_c mobile System linked to the cup (X_c plane Y_c of basis of the cup).

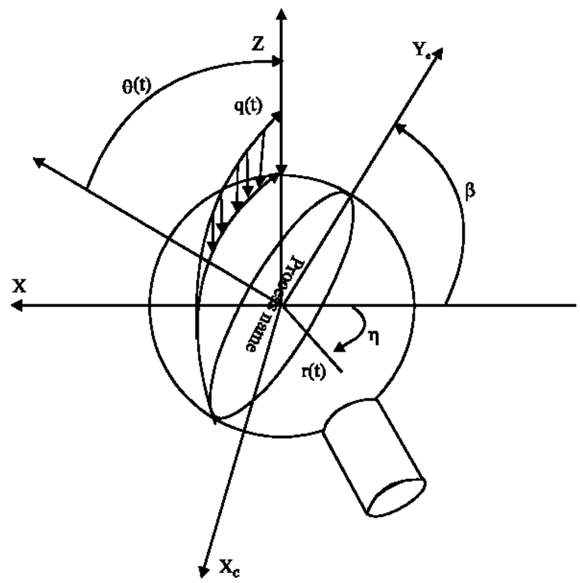

Fig. 1: Schema of the coordinate system of the couple materials (cup and femur)

β and η are the angles defining the orientation of the hemispheric cup: β, angle of slant situated between the X axes and, Y_c, η angle anteversion of the cup, situated between the X axes and Z_c.

The spherical coordinates of the point m are gotten finally below by the transformations:

$$\theta = \cos^{-1}\left(\frac{2Z_c}{D_c}\right),\ \phi = \tan^{-1}\left(\frac{X_c}{Y_c}\right) \text{et}\ \rho_m = \frac{D_c}{2} \tag{5}$$

In a total prosthesis of hip the femoral head (smaller) is tightened in the cup (bigger) under a variable resulting pressure (Bergmann *et al.*, 1993). The point of contact is a surface having a circular circumference whose the pressure is normal plane passing by the origin of the reference (Fig. 1). The pressure of contact on the circular cup surface has been calculated while considering the Hertzian theory for the elastic contact of two bodies possessing a non compliant geometric shape. The formula adopted in the case of a sphere in contact inside a sphere (Whitehouse, 1994). The local pressure of contact has been calculated by:

$$\sigma_t(\theta,\phi) = \frac{3q(_t)}{2\pi r^2(_t)}\left[1 - \frac{d_t^2(\theta,\phi)}{r^2(_t)}\right] \tag{6}$$

Or r(t) is the circular ray of surface of the contact, is:

$$r(t)=\left[\frac{3\pi}{8}q(t)\left(\frac{1}{\pi E^*}\right)\left(\frac{1}{D_h}-\frac{1}{D_c}\right)^{-1}\right]^{1/3} \tag{7}$$

$$\text{With } \frac{1}{E^*}=\left[\left(\frac{1-V_h^2}{E_h}-\frac{1-V_c^2}{E_c}\right)\right]$$

The distance of the generic point m to the surface of contact with the axe of load vector is given by:

$$d_t(\theta,\phi)=\frac{D_c}{2}\sin\left\{\cos^{-1}\begin{bmatrix}\sin\theta\sin\theta_p(t)\\ \cos(\phi-\phi_p(t)\\ +\cos\theta\cos\theta_p(t)\end{bmatrix}\right\} \tag{8}$$

While determining the variation using the Eq. (7), we obtain:

$$\Delta D=\left(\frac{d_c^2\left(r^3/\lambda\right)^{-1}}{1+D_c\left(r^3/\lambda\right)^{-1}}\right) \tag{9}$$

With

$$\lambda=\left[\frac{3\pi}{8}q(t)\left(\frac{1}{\pi E^*}\right)\right] \tag{10}$$

It is noticed that the variation of the diameter is a function of the second order, so it obtains an extremum. In this calculation we used mathematical formulation by introducing geometrical parameters for a patient of standard reference (Raimondi *et al*., 2001): D_c = 28.2 mm, D_h = 28 mm , η = 15°, β = 50°, Ra = 0.05 μm.

RESULTS AND DISCUSSION

The Fig. 2-4 show the variation of the wear volume according to the different prosthetic parameters (the variation of the diameters of the couple materials, the roughness and the modulus of elasticity of the femoral head) for three weights of the patient, respectively p = 50, 70 and 90 kg.

The volume of wear increase with the increase of the patient's weight and it is proportional to the applied load. Figure 2 shows a non linear increase of the wear volume with the femoral main diameter.

The articulation passes by an initial phase where the surfaces are polite *in vivo*. During this phase the liberation of metallic particles increased passing by a maximal value during this phase as shown in Fig. 2. After a few thousands cycle of gait will be an apparition of a steady-state, in which the wear remain constant. For the three cases of the patient's weight, this increase reach a maximum then it decreases with the increase of the difference of the diameters of the couple x material and finally it stabilize asymptotic to a value of the wear volume equal to the weight of the patient. It is clearly shown that the wear of a surface leads to a reduced wear volume when the surface is reduced.

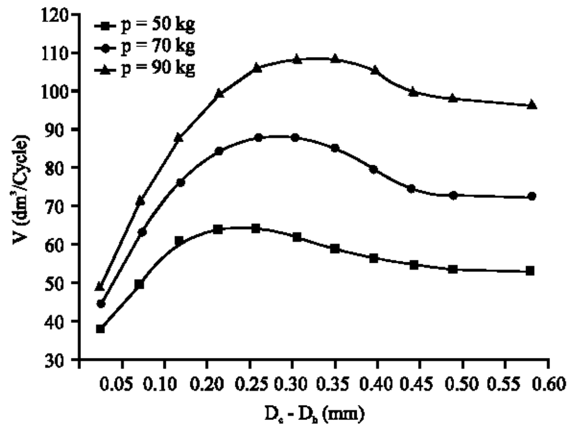

Fig. 2: Variation of the wear volume according to the variation of the couple diameter femoral head-cup

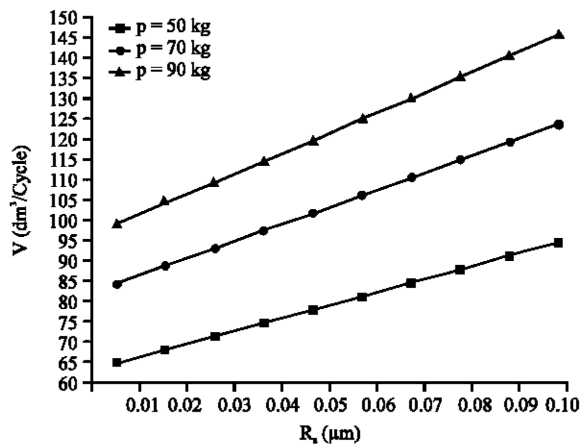

Fig. 3: Variation of the wear volume according to the variation the roughness contact surface

Figure 3 shows the variation of the wear volume according to the middle roughness (R_a). The tracings show a linear increase with the femoral main roughness. Using the calculation of regression we can deduct a mathematical model of the wear factor K. The wear volume is linearly proportional to the roughness of the femoral head.

The wear volume increase no linearly until a maximum and decrease with the increase of isotropic elastic modulus of the cup (Fig. 4). The curve takes a slightly ascending pace while reaching an extreme value of the wear volume. When the weight of the patients is weak, the variations of the isotropic elastic modulus have slight effect on the wear volume.

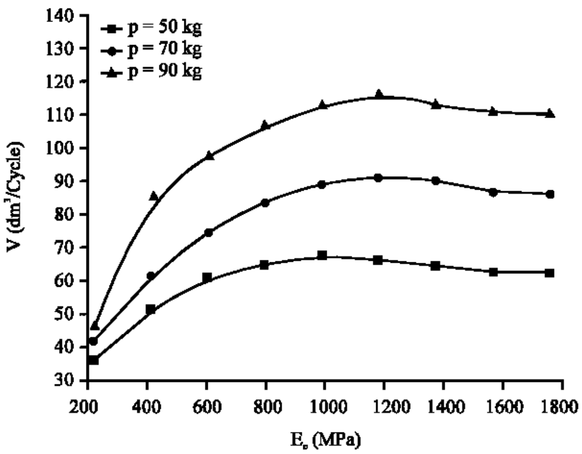

Fig. 4: Variation of the wear volume according to the Young modulus of the femoral head

The results show a very meaningful role of these parameters on the of wear volume for heavier patients.

CONCLUSIONS

The simulations of the models permit the calculation of the role of the different parameters affecting on the global wear volume and on the distribution of the pressure on the surface of the articular cup. The load applied on the head of thighbone produces a distribution of contact pressure that is not axisymetric, it explain the difficulties to the use of configuration of sphere. The analysis of the results permits the deduction of the following conclusions:

- The reduction of the surface of contact reduce the clearing of the remnants of wears
- The volume of wear increase proportionally and linearly with the roughness (R_a)
- Minimizing the surfaces of contact of the couple materials comes back to decrease the wear volume and therefore to the increase of the durability of the prostheses
- The patient weight influences directly the prosthetic parameters.

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