# UHMWPE Wear by Plastic Deformation in Hip Endoprosthesis

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*The UHMWPE has a plastic behaviour but it is specified only the surface wear, unless the evaluation of the great importance of this behaviour of plasticity for the calculation of the depth of penetration. This work tries to demonstrate the fact that there are components of the wear by plastic deformation. Also, by this type of wear an explanation of smooth surface of the UHMWPE in the contact zone, after millions of gaits cycles may be explained*

*Keywords: hip endoprosthesis, wear, penetration depth, roughness, plastic behavior*

The ultra high molecular weight polyethylene (UHMWPE) is extensively used as a bearing material in orthopaedic implants. Compared to ceramic or metallic versions the UHMWPE bearing is more biocompatible [1-3].

Nevertheless, it has become evident that the wear of UHMWPE may be the limiting factor that compromises the long-term performance of joint prosthesis [4]. As it results from the studies reported in the literature [5-8], the surface, which has a higher degree of the wear, (in fact, a bigger penetration depth) is very smooth. Atkinson was the first who observed that there is not a straight relation between the mass and the volume of the material removed by the wear. In other words, the amount of removed material corresponds to a volume inferior to the measured volume. It has been confirmed that worn polyethylene acetabular sockets exhibit three distinct regions: a relatively smooth high wear region in the superior half of the cup, a rougher low wear area in which the original machining marks are still visible and a low ridge separating the two.

Wang [9], also observed that in the zone of maximum penetration depth, the surface is very smooth, and he cannot explain this phenomenon, because he expects that in this area the surface must be rougher. The phenomena which explain smooth surface in high wear region are due to plastically deformation of the polyethylene asperity's peaks, that is, maintaining a constant value for the maximal rugosity characteristic of UHMWPE surface or a variation of that value within a very narrow range, whose average would indicate the value of the functional rugosity. Maintaining the value of the roughness between certain limits may be described in this way: in the process of the wear, it is noticed a growing of the maximum roughness value until the value for which the real pressure is bigger than the flow pressure. In this case, the tips of rugosities are plastically deformed until the value of the real pressure is equal to the elastic's limit of UHMWPE.

When there is an elastic deformation, in the same time it appears a removal of the material by adhesive wear (due to the smaller value of the roughness), which will increase the maximal roughness. The process is quite periodical repetition, leading finally to a supplementary depth of penetration considering the one, which was produced by the material's removal and of course, to maintain a smaller roughness in the region of maximum penetration (contact area).

## Experimental part

The main aim of this paper was to explain why the surface, which exhibits a higher degree of the wear, is very smooth. For the beginning, some consideration regarding the hip biomechanics needed to be done for proving the loading variation for one gait cycle; also the cinematic of this joint: flexion – extension, abduction – adduction and

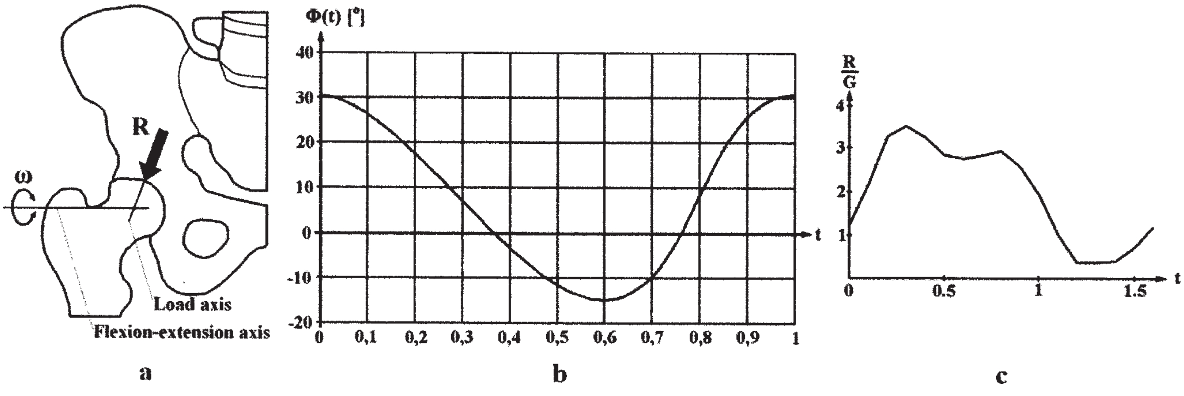


Fig. 1a . Reacting force and flexion-extension axis at one hip joint; b - Variation of the ratio R/G during one gait cycle; c - Variation of the flexion – extension, during one gait cycle

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internal – external rotation movements. Considering this data, a testing device of the tribological parameters for the biomaterials used for hip endoprosthesis was designed, with an accurate respect of the biomechanics of the hip joint (the physiological conditions).

*Considerations regarding the biomechanics of the hip joint*

In normal gait, the muscular action and the weight of the patient contribute to generate a resulting variable force R which acts upon the femoral head Also, the relative movement of the femur head on the acetabular cup (with an angular speed ω) is due to flexion-extension (fig. 1.a).

The hip joint is a typical spherical joint with three-movement axis. So, it could perform three kind of movements: flexion-extension, abduction-adduction and internal rotation.

The angular variation of the flexion-extension (the principal movement) is presented in figure 1.b, during normal gait [4].

During gaiting, it is very difficult to estimate the action of the resultant force on the joint. The curves of the ratio between the reaction force R and the body weight G variation, during one-step, for a man, were represented in figure 1.c. A similar variation was determined by Franek [10].

*Kinematics scheme of testing device designed for experimental wear determination, for hip endoprosthesis biomaterials*

To evidence requirements of hip biomechanics, we made one testing device whose scheme is depicted in figure 2.

The facilities of this device are the following: simultaneous or alternative measurements of friction coefficient and wear; perfect synchronization of kinematics and dynamics simulation of hip joint; respecting angle between load axis and oscillation axis.

The device was built in such a manner, that the acetabular component (fig. 2, position 9) is only supported by the spheres which perform an oscillatory movement similar to that of a femoral head.

The double acetabular piece tended to move in the same time with the spheres due to the friction (fig. 2, positions 8 and 10), that was why it was fixed in one extremity with an elastic lamella which supported a force transducer, and the other extremity was immobile.

The measured friction moment is a projection of the real moment, the last one resulting from calculations. A mechanism cam - wedge - helical compression arc (fig. 2, position 12) achieved the specific loading for the hip joint; the cam (fig. 2, position 13) was synthesized corresponding with the loading diagram. The cam was geared in rotation by a chain driving (fig. 2, positions 4, 15, 16) and a shaft (fig. 2, position 14).

In addition, it was important to measure the loadings resulted after the synthesis of the cam also by a force transducer which was attached to the charging mechanism. The flexion –extension movement was described by a quadrilateral mechanism (fig. 5, position 5), which transformed the rotation of the chain wheel (figure 5, position 4) in an oscillation.

This movement had a frequency of 1 Hz (achieved by reducing the revolution of the electric engine, position 1, by the trapezoidal driving belts, position 2, and the recurrent reducing gear, position 3). The oscillation was transmitted to femoral head by a cogwheels chain, its ratio being 1:1 (fig. 5, position 5) and an universal jointed coupling (fig. 5, position 7). When the double acetabular piece is fixing, is possible to test the wear on the acetabular cup.

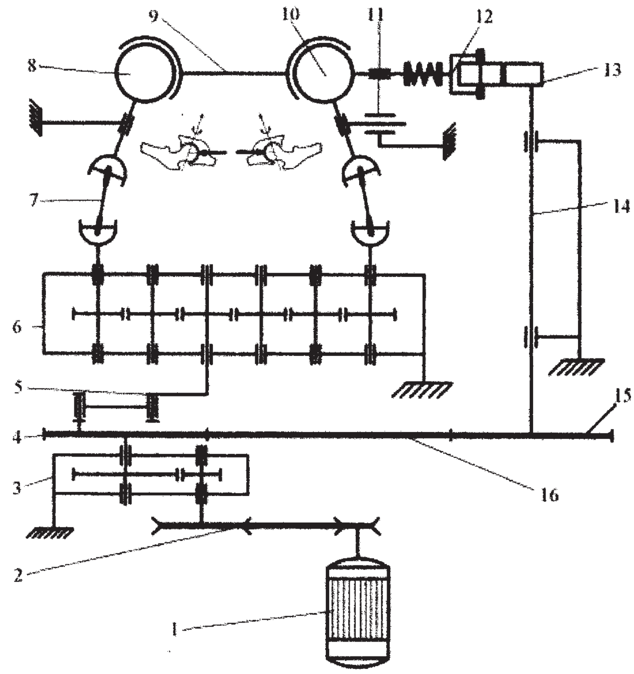


Fig. 2. Cinematic sketch of the testing device: 1 – electric engine; 2 – trapezoidal mechanism; 3 – reducing gear; 4 – leading chain wheel; 5- quadrilateral gear; 6 – recurrent reducing gear; 7 – universal joint; 8, 10 – femoral head; 9 – double acetabular piece; 11 – beam; 12 – wedge gear; helical arc, 13 – cam; 14 – shaft; 15 – led chain wheel; 16 – rings and rolls chain

For testing procedure, there are used: for acetabular coup: UHMWPE and for femoral head: stainless steel alloy based on CoCrMo. Testing period was 100 days, 5 h/day, which means 1.800.000 cycle.

The acetabular cup surface wear is illustrated in figure 3.

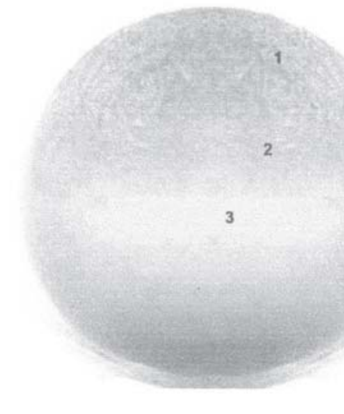


Fig. 3.A cetabular cup wear surface: 1 - a rougher low wear area; 2 - a low ridge separation; 3 - a relatively smooth zone

As in previous studies, the wear of acetabular cup shows three distinct regions: a relatively smooth high wear region in the inferior half of the cup, a rougher low wear in the superior half of the acetabular socket and a low ridge separating the two (fig. 2). The smooth area is deeper than rough area, with a volumetric wear and a depth penetration. To evaluate the depth penetration caused by roughness peak plastic deformation it is necessary to develop one calculation method.

*Calculation of the depth of penetration due to roughness peak deformation*

For determination of the surface of UHMWPE maximum roughness variation, it is calculated the real pressure - $p_r$ and after that, the limits for the pressure progress:

$$p_r = \frac{\overline{R}}{\eta \cdot A_n} \tag{1}$$

$\overline{R}$=1015N average load by a gait cycle;

$A_n = \pi \cdot a^2 = \pi \cdot 13{,}690965^2 = 588{,}868\,\text{mm}^2$ – nominal contact area; where $a$ is the hertzian contact radius (plan-sphere equivalent contact) defined by:

$a = a^* \left( \frac{3 \cdot \overline{R}}{E' \cdot \sum \xi} \right)^{\frac{2}{3}} \cdot \frac{\sum \xi}{2}$ where: $\sum \xi = \frac{1}{r_s} - \frac{1}{r_c}$; $r_s = 16$mm - femoral head radius ;

$r_c$ = 16.1mm - acetabular cup radius;

$E'$ - equivalent elasticity modulus: $\frac{1}{E'} = \frac{1}{2}\left( \frac{1-\nu_{cf}^2}{E_s} + \frac{1-\nu_{cc}^2}{E_c} \right)$

for femoral head (CoCr stainless steel): $E_s$ = 2.1 .$10^5$N/mm$^2$, elasticity modulus and $\nu_s$ = 0.3 Poisson coefficient; and for acetabular cup (UHMWPE): $E_c$ = 1400N/mm$^2$; $\nu_c$ = 0.3, $\Rightarrow E' = 3056{,}465$ N/mm$^2$ [10].

The dimensionless real area:

$\eta = \frac{A_r}{A_n}$ is: $\eta = \frac{L^2 \cdot b \cdot \nu}{R(R_{\max})(d\nu - d - 1)} \left[ \left( \frac{D}{R_{\max}} \right)^{\frac{1}{d}} p_0 \right]^{\nu - 1}$ where:

$$D = (R_{\max}) \cdot \left( k_\nu \frac{1}{b} \right)^{\frac{1}{\nu}} \cdot \left( \frac{1}{\Delta_r} \right)^{\frac{d}{2}} \cdot \left( \frac{1-\nu_c^2}{E_c} \right)^d$$

$$k_\nu = \frac{\nu_1 \cdot \nu_2 \cdot \Gamma(\nu_1) \cdot \Gamma(\nu_2)}{(\nu_1 + \nu_2) \cdot \Gamma(\nu_1 + \nu_2)};\; d = \frac{2}{2 \cdot \nu + 1};\; \Delta_r = \frac{R_{\max}}{R \cdot b^{\frac{1}{\nu}}}$$

In this preview relations $b$ and $\nu$ are the curve lift parameters: $b=b_1+b_2$ and $\nu=\nu_1+\nu_2$.

For steel where $Ra=0.02$, $\nu_1$= 0.9, $b_1$ =1 and for UHMWPE $\nu_2$=3, $b_2$ =3.5 [6, 14].

The values of curve lift parameters are: $b$ = 4.5 and $\nu$ = 3.9.

$R$ – roughness average curvature radius: $R = \frac{R_1 \cdot R_2}{R_1 + R_2}$ where considering [7] $R_1$ = 45mm (steel) and $R_2$ = 35mm (polyethylene), consequently $R$ = 19.687μm.

$R_{\max}$ - maximum roughness defined:

$$R_{\max} = R_{\max S} + R_{\max P}$$

By considering [8], $R_{maxS}$ = 0.08μm (steel) and $R_{maxP}$ = 0.52μm (polyethylene), $R_{\max}$ = 0.6μm.

$L$ – specific length, $p_0$ – maximum hertzian pressure:

$$L = L^* \left( \frac{3 \cdot \overline{R}}{E' \cdot \sum \xi} \right)^{\frac{1}{3}}$$

It is considered that real pressure evolutes between two values equal to the elasticity limit $\sigma_c$=16.5MPa and the flow limit $\sigma_c$=22MPa of UHMWPE [8]. Also it is considered that the only parameter which varies in the real pressure expression is the $R_{max,p}$, as it is mentioned in [5-12], the roughness of the femoral head remains approximately constant over all the function period, growing only 10% in 15-20 years.

Determining $R_{max,P}$ for two limits of real pressure, $p_{r1} = c \,.\, \sigma_e$ = 49.5MPa respective $p_{r2} = c \,.\, \sigma_c$ = 66MPa, ($c$ = 3 [7]):

$A_u$ – contact area = wear area.

$$R_{\max P1}\big|_{pr1} = 5{,}12778 \cdot 10^{-4}\,\text{mm} \quad R_{\max P2}\big|_{pr2} = 5{,}631532 \cdot 10^{-4}\,\text{mm}$$

The evolution of the maximal roughness: when their value is maximum, $p_{r2}$=c . $\sigma_c$ = 66MPa, it is deformed until the real pressure is the same with the elastic limit and after that, the value of the maximum roughness is growing because of the remained material up the critical value of the real pressure. In this condition, the maximum roughness expression for UHMWPE surface is:

$$R_{\max P}(t) = \begin{cases} R_{\max P1} + H(t) & p_{r1} \le p_r \langle p_{r2} \\ R_{\max P2} + H(t) - \delta & p_r = p_{r2} \end{cases} \tag{2}$$

where:

$H(t)$ – removed material depth: $H(t) = \frac{V(t)}{A_u}$, $V(t)$, – removed material volume;

$A_u$ – calotte wear area:

$$A_u = 2 \cdot \pi \cdot r_u h \cdot \frac{h + 2 \cdot r_c}{2 \cdot (h + r_c - r_u)},$$

$h$ = 0.090184 mm – penetration depth, experimentally determined;

$r_c$ = 16.1mm;

$r_u$ = 16mm, $\delta = \varepsilon \,.\, R_{maxP2}$ - plastic roughness deformation, and $\varepsilon$ specific deformation.

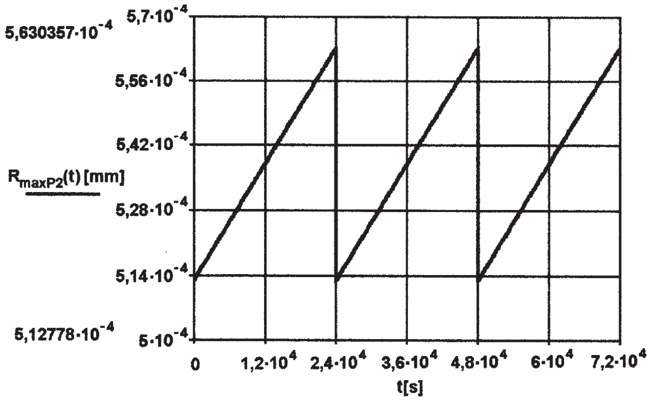


Fig. 4 Maximum roughness variation of UHMWPE

<u>Remarks.</u> The variations in polyethylene rugosity within the limits shown in figure 4, are purely theoretical, because in reality, periodicity is due to the size of the polyethylene particles and could vary between $10^{-5}$mm (in slow wear regime) to 0.5 mm (in fast wear at the end of time of functioning the prosthesis) [13].

Knowing the maximum polyethylene rugosity limits as well as the period during which a deformation takes place (24000s), it is possible to calculate the total deformation $h_d$ of the UHMWPE surface maximum rugosity for 1.8 x $10^6$s:

$$h_d = (R_{\max P2} - R_{\max P1}) \cdot \frac{1{,}8 \cdot 10^6}{2{,}4 \cdot 10^4} = 0{,}003775\,\text{mm}$$

This value is summed with penetration depth due to the removed material for the same time interval.

## Conclusions

As it was calculated and measured before, we can conclude that the mass wear is smaller than the volume wear, due to the plastically deformations of the roughness picks, which lead to a surface density increase.

As a result of laboratory testing, we conclude that the depth of penetration has two components; the first which corresponds to the volume of the removed material and which has the biggest percentage from the total depth of penetration – 96.31% and the second which is owed to the plastically deformations from the level of roughness picks, 3.69%.

As a medical suggestion for the patient with a prosthesis it is not only avoiding the excessive loading of the articulation by a big mass of the patient – it is contraindicated the obesity but also the reduction of the walking activity, as also avoiding standing very much for a long period. It is very important the professional reconversion, which costs, is smaller than a revision surgery (prosthesis reimplantation).

## References


1.PUSCA S., PAUN M.A., TOMA C., Mat. Plast, **44**, nr. 1, 2007, p. 39

2.FETECAU C, STAN FELICIA, POPA C., Mat. Plast., **42**, nr. 3, 2005, p.245

3.FETECAU C., STAN FELICIA, Mat. Plast., **44**, nr. 4, 2007, p. 263

4.BIGSBY R.J.A., HARDAKER C.S., FISHER J., Fifth World Biomaterials Congress, **1** University of Toronto Press, Toronto, May 29, 1996, p. 970

5.ABRAHAMS N., GOLDSMITH A. A., NICOL A. C., 12th Conference of the European Society of Biomechanics, Dublin, 2000, 226

6.TUDOR A., Contactul real al suprafeţelor de frecare, Editura Academiei , 1990

7.PAVELESCU D., Tribotehnica, Editura Tehnica -Bucureşti, 1983

8.ATKINSON J. R. WEAR No. 104, 1995, 217

9.WANG A., ESSNER A., POLINENI V. K., SUN D., STARK C., DUMBLETON J. H., Plenary and Invited Papers from The First World Tribology Congress, London, 8-12 sept., 1997

10.FRANEK F, PAUSCHITZ A., KELMAN P., Proc. of 10th International Colloquium, Technische Akademie Esslingen, 1996 p. 1495

11.TEOH S. H., CHAN W. H., THAMPURAN R., Journal of Biomechanics, issue 35, 2002, p 323

12.WILMANN G., Proc. of 10th Intemational Colloquim, Technische Akademie Esslingen,1996 p. 1505

13.CATES H. E., FARIS P. M., KEATING E. M., Journal of bone and Joint Surgery, 75B, 1999, p 249

14.BAJARIA H., BELLARE A., Medical Plastics and Biomaterials, March, 1998, p 132


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