ASPECTS CONCERNING DYNAMIC MODELLING OF THE HIP JOINT

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ABSTRACT:
Although modern surgical techniques in orthopedics are now available, the treatment of the diseased hip joint does not end with surgical replacement, the postoperative rehabilitation being of the utmost importance. Present hip implants use an acetabular cup, which is constructed entirely of polyethylene or of metal with polyethylene lining. The debris from implant wear, particularly polyethylene is one of the contributory factors of the hip implants failure. Although the technology of hip implants has been intensively developed with high success rates, the mechanism of polyethylene wear and bone degradation are not fully understood. For this reason, our research efforts focus on the dynamic modeling of the hip joint, following regular human activity simulation in which can be determined the forces values from within the prothesed joint and then used in the designing of reliable components.

KEYWORDS: hip prosthesis, polyethylene, dynamic modeling, simulation

1. GENERAL CONSIDERATIONS
The realization of a hip joint prothesis requires a complex analysis from the biomechanics view-point, wich leads to the realization of a perfect functional prothesis of the human bone.

The prothesis metal-polyethylene is the most used-up, and it total hip joint replacement after the resection of the femoral and prepaition of the medullary femoral chanel and of the acetabulum.

The wear of the polyethylene and microsplints which results from the wear can cause osteolysis, the lose of osseous mass from around the prothesis, reduce the rezistance protheses and in the end it determines the remake of the surgical interventions. For this reasone, in the last past years, the reseach workers efforts have been headed for the optimization of the prime materials quality from which the prothesis are made of, and the friction conditions and the tensional state from the joint, but also the formative mechanisms of the microfissures.

If from surgical angle the artroplasty has became today routine procedure, from the angle of the postoperative behavior, the fulfillment of the hip joint protheses still raises many problems requiring a complex biomechanics analysis.

The final aim of the analyses is the manufacture of a durable component, biocompatible that can stand as a perfect functional replica of the human bone. The first problem and the most importance is related to the behavior in time of the couples of the materials used in the manufacturing of the protheses (life cycle of a hip joint protheses can vary between 10 and 15 years, after this period of time the remakeinig operation of the joint is inevitable).

In general, the hip joint prothesis is consisted of two parts (figure 1) the metallic part and the part made from plastic material, the fixation of the two parts is made with the help of a bone cement (methyl methacrylate). The metallic part is compounded from a ball attached to a stem, both of them are made from stainless steel, cobalt-chromium alloys or titan. The plastic material is called acetabular cup and can be made from PTFE, GUR 5113, GUR EP 4224.

In the last years for the total hip joint prothesis and for knee joint prothesis were used hight density polyethylene with hight wear resistant properties(UHMWPE). Hight molecular mass confers to the UHMWPE superior hardess and remarkable wear resistance comparative with the majority of biocompatibile plastic materials [2], [5], [8].

The wear of the polyethylene and of the microsplints resulted in the wear process may cause osteolysis, the lose of osseous mass...
around the prothesis, the decrease of the prothesis resistance and in the end the remake of the surgical intervention.

**Fig. 1. Replaced hip joint [4]**

For this reasone, we must pay special attention to the wear mechanism and the mechanism of the appearance of microsplints.

The wear of the acetabular cup from a prothesis with a femoral head of 32 mm [4], [2] is about 0.1 mm/year and it represents a 30% less than the wear registered in the case of utilizing PTFE by Charnley in the ‘50 [1]. Whereas the thickness of the acetabular cup is at least 6 mm and the wear rate is about 0.1 mm/year it results that the complete wear would occur in less than 60 years [8].

The experimental researches have been making evident the fact that though the complete wear of the acetabular cups made from UHMWPE is very reduced, the microsplints resulted following the wear processes can determine adversary reactions of tissues at the gradual absorption of the bone tissue – ostelysis [1].

Ostelysis causes the decrease of the protheses resistance, it causes the patient discomfort and in the end it causes the remake of the surgical intervention a fact which determine the researchers efforts to look in the direction of the optimization of the materials quality from which the prothesis are made of as well as the friction conditions from the joint so that in the end the percent of the microsplints resulted from the wear of UHMWPE to be very reduced [7].

The total hip joint prothesis is a mechanical joint which must resist to the multiple biomechanical requirements for long period of time and which must fulfil the following desideratum [1]:

- biocompatibility;

- the possibility of personalization of the prothesis depending on the peculiarities (functional morphological particularities) of the considered case;
- proper resistance to support the persistent stresses caused by the body weight on all duration of its function, without fatigue fractures of its components;
- high resistance to wear;
- low friction between the articulated surfaces;
- solid and durable fixation at the bone level;
- simple and cheap technology for manufactureing the prothesis components.

The prothesis election is directed to an elaborate knowledge of the biomechanics implications and of the biocompatibility implications raised: the protheses kinematics, the forces whereat the prothesis is submissively to, the distribution of the tensions at the bone “interface” level, the friction forces, the wear, the stability and the mechanical properties of the materials that the prothesis is made of, the degree of rugosity of the articulated surfaces, the cementation etc. [1], [3], [6].

**Fig. 2. The load scheme from a replaced hip joint**

A very important role in the election of the type of the prothesis that might substitute better the complex biomechanics of the hip joint is the determination of the stresses that occur in the joint, within the realization of mathematical patterns to allow us to realize a static and an dynamic pattern of the joint.

In figure 2 is presented the scheme of the loads from a replaced joint. At the normal acetabular level, in the uniped support, the resultant $R$ will produce an elastic deformation of it. The elliptical configuration of the acetabulum tends to becomes
spherical ensuring a more intimate contact with the femoral head.
After implanting the acetabular cup, on account of its rigidity, the physiologically deformation can’t occur. This causes the appearance of a secondary stress strain which tends to expel the cup.
After Dietschi [1] researches, the internal pressure strains of the normal acetabululum touches values of 0.82 N/mm² and those values are more higher in the case of implanting the the acetabular cup of the prothesis.

The junction acetabulum-bone or bone-cement is submitted to two kinds of forces: shearing forces and strain forces (figure 3). In the condition of a good centred cup the forces are equilibrated. If the acetabulum isn’t centred, uncovered by the bone on its upper, the forces exercised on it will tend to mobilize and to upstand.
The femoral head is normally submitted to strain forces occurred by fact that the resultant R is perpendicular to the femoral head. Conversely the femoral neck, by reason of his architecture, is under the influence of a flexural torque, because the resultant R does not operate after its anatomic axis.

The median portion of the acetabulum support the strain stresses which, after Pauwels calculus, are reaching a maximum of 19,8 N/mm², and the lateral portion traction stresses which are reaching maximum values of 6,6 N/mm², for a resultant R of 22 kN.
The metallic piece of the prothesis inlaid in the femur is also subdued to a flexural torque which tends to accentuate the curvature. This flexural torque is increasing correlated with the more advanced position of the femoral piece into the varus. Valgization of the protheses reduces flexural torque but increases the resultant R (throughout the decrease of the arms lever of the muscular force).

The increase of the resultants size R however, is not favorable to the acetabulum. In practice it’s ought to find a compromise between the position in the valgus of the protheses and the size of the resultants R [1].

2. THE DETERMINATION OF THE STRESSES FROM THE HIP JOINT DURING ROUTINE ACTIVITIES

In order to determine the relations between the movements of the hip joint and the resultant force there are introduced the following coordinate systems [7]:
- XYZ the coordinate system in which is positioned the femoral head. The plan X-Z represents the frontal plan, X- the median lateral axis, Y- the posterior-anterior axis; X_C,Y_C,Z_C the coordinate system solidary to the cup and with the origin in the center of the cup;
- the spherical coordinate system solidary with the surface of the joint.
The angle $\beta$ represents the banking angle of the cup and it is definite as the angle between the axis X and the axis X_C. The angle $\eta$ is the antevertion angle of the cup and it is definite as the angle between the axis X and Z_C.

![Fig. 3. Coordinate systems associated to the hip joint](image)

At a given moment of time t from within the walk cycle, the resultant force vector $F(t)$ applied on the P placed on surface of the the joint (figure 3). The resultant of the forces that acts on the femoral head is

$$F(t) = \sqrt{F_x^2(t) + F_y^2(t) + F_z^2(t)},$$  \hspace{1cm} (1)

where $F_x(t)$, $F_y(t)$ and $F_z(t)$ represents the components on the X, Y respective Z axis.
The coordinate of the point \( P(X_p, Y_p, Z_p) \) can be calculated with the relations:

\[
X_p = R_c \cos \left[ \tan^{-1} \left( \frac{-F_x(t)}{\sqrt{F_x^2(t) + F_y^2(t)}} \right) \right],
\]

\[
- \sin \left[ \tan^{-1} \left( \frac{-F_x(t)}{-F_y(t)} \right) \right],
\]

\[
X_p = R_c \cos \left[ \tan^{-1} \left( \frac{-F_x(t)}{\sqrt{F_x^2(t) + F_y^2(t)}} \right) \right],
\]

\[
- \cos \left[ \tan^{-1} \left( \frac{-F_x(t)}{-F_y(t)} \right) \right],
\]

\[
X_p = R_c \sin \left[ \tan^{-1} \left( \frac{-F_z(t)}{\sqrt{F_x^2(t) + F_y^2(t)}} \right) \right],
\]

where \( R_c \) represents the cups radius.

The coordinate of the point \( P \) given by the relations (2) are transposed in the reference system \( X_C Y_C Z_C \) of the cup and obtaining the relation

\[
\begin{bmatrix}
X_C \\
Y_C \\
Z_C \\
\end{bmatrix} = \begin{bmatrix}
\sin \eta & \cos \eta & 0 \\
- \cos \eta \cos \beta & \sin \eta \cos \beta & \sin \beta \\
\cos \eta \sin \beta & - \sin \eta \sin \beta & \cos \beta
\end{bmatrix} \begin{bmatrix}
X \\
Y \\
Z
\end{bmatrix}
\]  

(3)

where \( \eta \) is the anteverton angle of the cup, \( \beta \) -banking angle of the cup.

The spherical coordinates of the point \( P \) are

\[
\begin{cases}
p = R_c; \\
\theta = \cos^{-1} \frac{Z_C}{R_C}; \\
\varphi = \tan^{-1} \frac{X_C}{Y_C}.
\end{cases}
\]  

(4)

The local contact stress is

\[
\sigma_1(\theta, \varphi) = \frac{3F(t)}{2\pi r^2(t)} \left[ 1 - \frac{d^2(\theta, \varphi)}{r^2(t)} \right]^{\frac{1}{2}},
\]  

(5)

where \( r \) is the contact surface radius,

\[r(t) = \left[ \frac{3\pi}{8} \frac{F(t)}{\pi E_H} \left( \frac{1 - \nu_H^2}{E_H} - \frac{1 - \nu_C^2}{E_C} \right) \left( \frac{1}{D_H} - \frac{1}{D_C} \right) \right]^\frac{1}{3}\]

(6)

and \( d_t \) is the distance between a point situated on the contact surface and the load vectors axis

\[d_t = R_c \sin \left[ \cos^{-1} \beta \right],\]

(7)

where

\[
\beta = \sin \theta \sin \theta_c(t) \cos(\varphi - \phi_c(t)) + \cos \theta \cos \theta_c(t).
\]

(8)

In (6) equation were taken into account the following notations: \( D_H \) is the diameter of the femoral head, \( E_H, \nu_H \) -Young modulus of elasticity resective Poisson coefficient of the femur; \( E_C, \nu_C \) -Young modulus of elasticity resective Poisson coefficient of the cup.

3. THE DYNAMIC MODELING OF THE HIP JOINT

In a problem of dynamics, when practical all the sizes that steps in the expressions (1), (2), (5) and (6) are aleatory functions of time, the variation of the reaction force from the hip is impossible to be exactly determined by the analytic solving of some equations. Even in the variant in which the differential equations of the motion can be written, is unlikely to solve them with an elevated degree of precision. Therefore, the only valid variant is to consider of numerical methods and to create calculus programs, which can make it possible to approach the dynamics problems of the human body or of his parts. The experimental research presumes the verification of the results obtained after the simulation of the action of the inferior limbs muscles on the hip joints made with stiff programs like Life Mod 2005 and MSC ADAMS. The software Lifemode 2005 it contains a specialized database on the human skeleton and soft tissues which surrounds it, and having exact date about the forces and the stresses generated by the soft tissues for given activities such as gait, running, laying on a chair, climbing the stair etc.
This database was created to carry through the simulations of given human activities and the comprehension of the way the forces that occurs in the structure of the osseous system as well as in the muscular tissues acts, in relatively short period of time and without needing complex input data. The realization of the simulation consists in some stages:
- generation of a partial human body;
- importing and attaching the geometry of the prosthesis to the body;
- ground reaction forces calculus;
- dynamic simulations for gain on horizontal and tilted ground.

**Fig. 4. The position of total hip joint prothesis in different stages of the subjects motion**

In figure 4 are showed a couple of the subjects position with a total hip joint prosthesis implanted.

**Fig. 5. Right foot and the ground contact forces for given weights of the subject**

The modeling and the simulation of male subjects with weights of 60, 80 and 100 kg in the cases such as: the horizontal gait, the preparation in sight of sitting on a chair, sitting, the lift in the vertical posture and again the horizontal gait, permitted us to obtain the variation diagrams of contact forces between inferior limbs and the ground and the variation diagrams of the contact forces between the acetabular cup and the metallic stem represented in figures 5, 6 and 7.

**Fig. 6. Left foot and ground contact forces for given weight of the subject**

**Fig. 7. Stem/cup contact forces for given weights of the subject**

4. CONCLUSIONS

The charts analyse permitted determination of some conclusions presented in tables 1, 2 and 3 and notice that:
- the contact forces size variation between the right foot and the ground major than the ones of the left foot;
- the contact forces variations between the inferior limbs and the ground and the contact forces variations between the cup and the stem are not proportionately with the changing of the subjects weight;
- the contact forces between the cup and stem are major for the moment of the preparation in sight of sitting on a chair than the rest of the analysed movements.
Table 1 Right foot and the ground contact forces for given weights of the subject (see figure 5)

<table>
<thead>
<tr>
<th>The weight increase of the subject [%]</th>
<th>(60-80) kg 33.33%</th>
<th>(80-100) kg 25%</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>27.482</td>
<td>25.905</td>
</tr>
<tr>
<td>2</td>
<td>23.540</td>
<td>23.671</td>
</tr>
<tr>
<td>3</td>
<td>14.046</td>
<td>17.923</td>
</tr>
<tr>
<td>4</td>
<td>13.501</td>
<td>20.539</td>
</tr>
<tr>
<td>5</td>
<td>29.126</td>
<td>27.43</td>
</tr>
<tr>
<td>6</td>
<td>18.668</td>
<td>24.20</td>
</tr>
</tbody>
</table>

Table 2 Left foot and ground contact forces for given weight of the subject (see figure 6)

<table>
<thead>
<tr>
<th>The weight increase of the subject [%]</th>
<th>(60-80) kg 33.33%</th>
<th>(80-100) kg 25%</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>24.902</td>
<td>25.621</td>
</tr>
<tr>
<td>2</td>
<td>21.174</td>
<td>21.401</td>
</tr>
<tr>
<td>3</td>
<td>21.279</td>
<td>24.679</td>
</tr>
<tr>
<td>4</td>
<td>13.844</td>
<td>22.224</td>
</tr>
<tr>
<td>5</td>
<td>22.457</td>
<td>22.165</td>
</tr>
<tr>
<td>6</td>
<td>10.972</td>
<td>16.324</td>
</tr>
</tbody>
</table>

Table 3 Stem / cup contact forces for given weights of the subject (see figure 7)

<table>
<thead>
<tr>
<th>The weight increase of the subject [%]</th>
<th>(60-80) kg 33.33%</th>
<th>(80-100) kg 25%</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>40.401</td>
<td>25.536</td>
</tr>
<tr>
<td>2</td>
<td>17.228</td>
<td>14.188</td>
</tr>
<tr>
<td>3</td>
<td>37.756</td>
<td>22.224</td>
</tr>
<tr>
<td>4</td>
<td>19.41</td>
<td>19.078</td>
</tr>
<tr>
<td>5</td>
<td>27.993</td>
<td>24.129</td>
</tr>
<tr>
<td>6</td>
<td>40.401</td>
<td>25.536</td>
</tr>
</tbody>
</table>

REFERENCES


AUTHORS

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