



HAL
open science

Dynamic modeling of the hip joint after implantation of innovative total hip prosthesis

Brahim Mahiddini, Yacine Benabid, Khaled Benfriha, Améziiane Aoussat

► To cite this version:

Brahim Mahiddini, Yacine Benabid, Khaled Benfriha, Améziiane Aoussat. Dynamic modeling of the hip joint after implantation of innovative total hip prosthesis. 9th International Conference on Modeling, Optimization & SIMulation, Jun 2012, Bordeaux, France. hal-00728585

HAL Id: hal-00728585

<https://hal.science/hal-00728585>

Submitted on 30 Aug 2012

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

Dynamic modeling of the hip joint after implantation of innovative total hip prosthesis

B. MAHIDINI
LPF/ Ecole Militaire
Polytechnique BP 17
BEB Alger, Algérie
Brahim_mahiddini@
yahoo.fr

Y. BENABID
LSM/ Ecole Militaire
Polytechnique BP 17
BEB Alger, Algérie
ybenabiddz@yahoo.fr

K. BENFRIHA
LCPI/Arts & Métiers
ParisTech, 157 Bd de
l'hôpital 75013, Paris
Khaled.BENFRIHA@en
sam.eu

A. AOUSSAT
LCPI/Arts & Métiers
ParisTech, 157 Bd de
l'hôpital 75013, Paris
Ameziane.AOUSSAT@
ensam.eu

ABSTRACT

The dynamic study of the hip joint after implantation of Modular Total Hip Prosthesis (THP) has allowed us to verify the exactness of the quasi-static model used by comparing our results with literature, this study confirmed that the tailor modular THP is an ideal solution to replace the destroyed natural joint of the hip. The realization a tailored femoral stem prosthesis starting by femoral modular canal has been the subject of various studies; the main difference between this works is how each author reconstructs the channel modules. The approach to reconstruction of the modular channel of our patient is via a simple and feasible from a time perspective and programming, it relies mainly on the cloud points obtained after the segmentation of DICOM images. Also, the automatic positioning of the modular prosthesis in the patient femur gives a more realistic model. After validating the simulation results of the THP module with the literature, there has been proud that the acetabular cup UHMWPE (ultrahigh molecular weight polyethylene) is the component that undergoes the greatest deformation. Because the acetabulum is directly related to the femoral head in our model, it is therefore necessary to analyze the dynamics of THP without the standard socket UHMWPE because the same constraints on the contact interface acetabulum-femoral head in our model.

KEY WORDS: total hip prosthesis (THP), dynamic modeling, DICOM images, FEA.

1. INTRODUCTION

The hip joint is the articulation of the proximal lower limb. In terms of biomechanics, its role is to support the weight of the body. It transfers the efforts, mainly those of gravity, for lower limb. This transfer is widely studied in the literature (Daniel et al, 2006), (Kayabasi and Ekici, 2007). Indeed, the distribution of body weight is very different depending on whether one is standing on both feet and one bears unipodal. Understanding how this transfer takes place in different situations is essential to treat various diseases of the hip joint and the development of a prosthesis performance.

Over the last 40 years, total hip Arthroplasty (THA) has been a highly successful operation, with documented benefits of significant reduction in pain and improvement in quality of life (Martin and al, 2011).

The loose sitting of total hip prosthesis gives a pain to the patient (Kayabasi and Erzincanli, 2006).

The finite element method (FEM) is an advanced simulation technique that has been used in orthopedic biomechanics. It is an important tool used in the design and analysis of total joint replacements and other orthopedic devices. FEM modeling and analysis present a non destructive design approach for bone implant hip prosthesis (Kayabasi and Ekici, 2007).

Dynamic factors result in abnormal stress and contact between the femoral head and acetabular rim with

motion of the hip. These mechanical stresses result in reactive hip pain related to activities that typically require movement of the hip into the flexed position, resulting in abnormal engagement between the femoral head and acetabulum (Bedi and al, 2011).

There are three parts to the artificial hip joint (Figure 1), the metallic femur head, the UHMWPE layer and, sometimes, the metallic cup covering the UHMWPE layer for better fixation on the pelvis (Hai-bo Jiang, 2007). The characterisation of hip joint contact forces provides essential information for prosthetic joint design and testing (Stansfield and Nicol, 2002).

In terms of angle, the neck-shaft angle and femoral anteversion angle play an important role in the transmission of forces (Kaku and al, 2004). They showed that the tilt of the pelvis in the frontal plane is also essential for optimal transfer while the orientation of the acetabulum forwards in the sagittal plane has very little influence.

The magnitude and distribution of stress and displacement during walking are important to understand the working of the artificial joint dynamic modeling is the success key of innovative prosthesis.

The present paper reports a new result for dynamic modeling from innovative design of total hip prosthesis. The results help the orthopedic surgeon to identify the ideal points of joint implantation. This result includes the activities of walking.

2. MATERIALS AND METHODS

2.1 CAD model

The innovative model of total hip prosthesis is the result of long design job. In this search, the reverse engineering phase involves scanning bone samples in order to determine the geometric range of different models and form the basic surfaces to extract the new implant. Our objective is to determine three-dimensional hip motion with respect to readily identifiable bone landmarks, and to use the data to design the external joints themselves. First, geometrical properties of the hip were determined and used to define the axis system. Second, the motion of normal hips was measured with respect to the defined axes. Third, a computer graphics program was written to design external joints with certain parameters optimized (Walker and al, 1985). To create a geometrical model of bones (hip joint) a magnetic resonance imaging data (MRI) provided by central hospital of Algiers, was used.

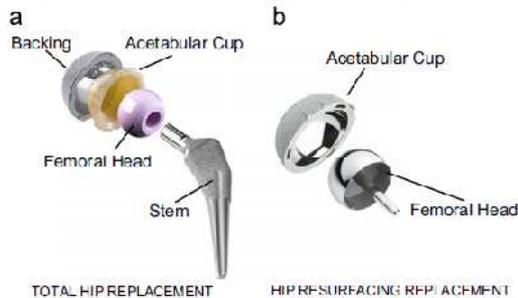


Figure 1: Main components of an artificial hip joint (Mattei and al, 2011)

2.1.1 Data acquisition

Figure 2 illustrates the main steps of this method. The input data are images in DICOM format with a scan interval of 0.3 mm, (obtaining 581 images at a 512 x 512 pixel of resolution 400 x 400 mm²).

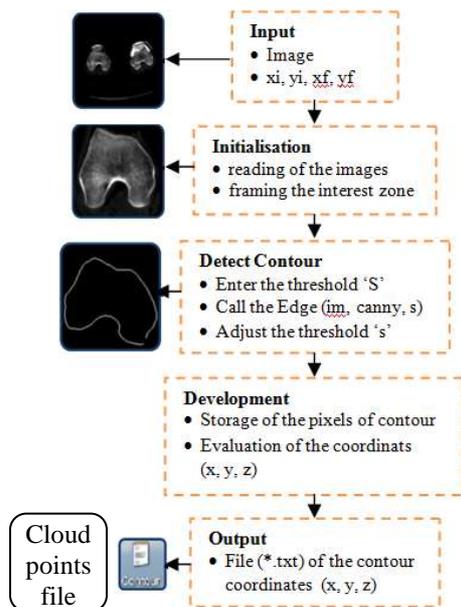


Figure 2: Flowchart of the algorithm proposed (Benabid and al, 2011)

2.1.2 Model design

The file of cloud points generated by the algorithm shown in figure 2 is used for the reconstruction of the 3D model of hip joint. We have also used ScanTo3D (SolidWorks) to filter the noise in cloud points and to generate curves of the sectional model geometries of both femur and pelvis. With the aid of spline functions, the loft function was used to generate the CAD geometry of hip joint.

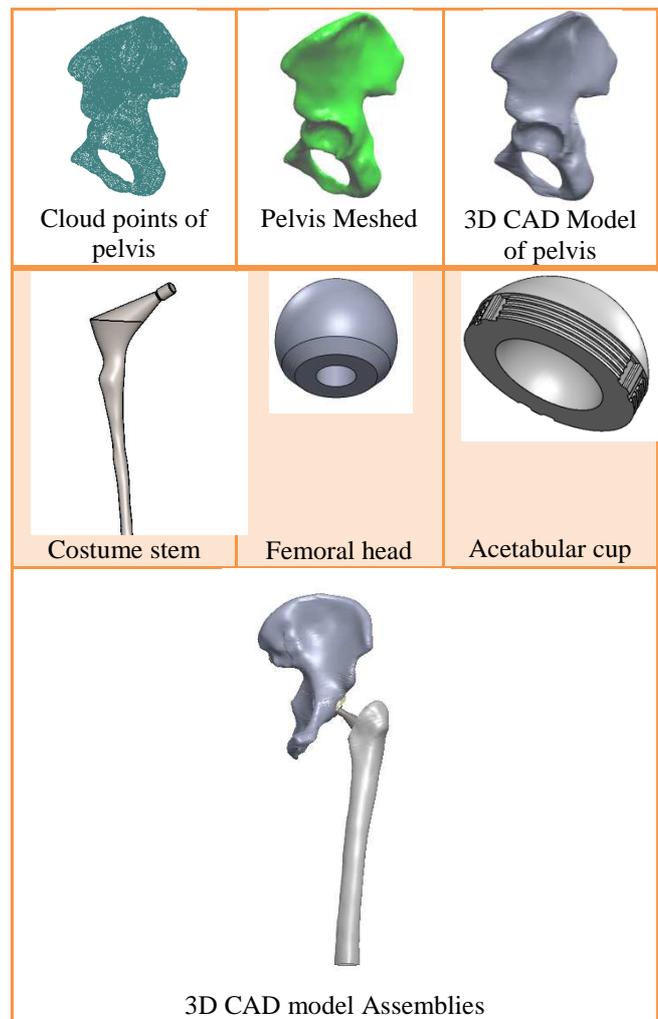


Figure 3: Model Design

2.2 Behavior during walking

Walking is a complex and difficult activity; usually humans walk without thinking about the execution of the movement. Found among mammal's bipedal walkers that Man is the only one to adopt the attitude erected as its natural position. During its evolution, bipedalism Vertical of Man makes it available for new activities, receiving and effectors (Briand and Bonnel, 1994). When human walking cycle, there are two phases: stance phase (figure 4 -1 to 8) and oscillating phase (figure 4 - 9 to 10).

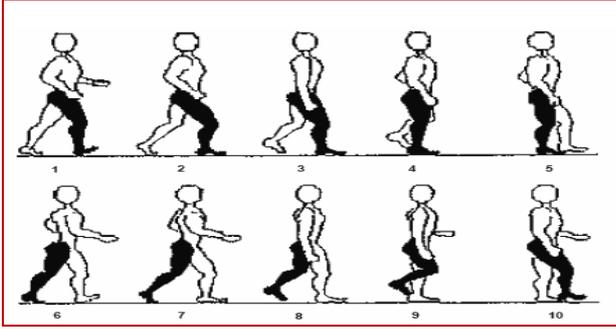


Figure 4 : The gait cycle:1/attacks heel;2/double former support of reception;3-4-5/bearing unilateral support; 6-7/double posterior support of dash;9-10/oscillating phase (D'angeli-Chevassut, Gaviria, 1994)

2. 3 Kinematic modeling

By definition, the kinematics is the study of motion without reference to the forces that generate this movement; it describes the temporal and spatial aspects of movement characterized by the relative position angles, velocities and accelerations of body segments and joints during locomotion. To determine the kinematics of the movement, the human body is treated as a kinematic chain consisting of rigid body links connected by perfect joints, each with one to three rotational freedom degrees (Pustoc'h, 2007).

Several concepts are used in the field of biomechanics; kinematic modeling we choose is using the basic sequence of rotations of the model because it is the most developed in the middle of the current biomechanics. Three-dimensional ranges of motion are calculated using the method of elementary sequences around moving axes. The rotation sequence chosen is the one recommended by the ISB (Wu and al, 2002) and many other authors (Doriot and Chèze, 2004) (Cappozzo and al, 2005).

2. 4 Kinetic modeling

The kinetic modeling aims to determine the mass and inertial parameters associated with each body segment. In this context there are several works that have been made to estimate the mass and position of center of gravity and moments of inertia of the segments. From anthropo-metric measurements (lengths and circumferences of body segments), the regression equations developed by (Seluyanov and Zatsiorsky, 1983), to assess the mass and inertial parameters of body segments such as the mass of the segments, the position of the gravity center G of each segment and the inertia matrix associated (Pustoc'h, 2007).

2. 5 Determination of dynamic torsor reduced at the hip joint

The principle of the dynamic method is used to isolate successively each body segment S_i of the articulated chain, the more distal than proximal. Knowing the dynamic torsor at the platform forces, the theorems of

Newton (or theorem of the dynamic resultant) and Euler (or dynamic moment theorem) are applied successively to the contact point of the foot on the platform, then ankle, and knee, and finally the hip (Figure 5). Are thus determined the forces and moments inter-segmental (articular) induced at each joint to balance the external mechanical actions (gravity, inertia forces, at support reactions) (Siegler and Liu, 1997). Indeed, the torser exerted by the dynamic distal segment S_{i-1} is known, one can deduce the torser exerted by the dynamic proximal segment S_{i+1} by applying the fundamental principle of dynamics: just write, every moment, equality between the torser external mechanical action torser system and dynamics of this system.

Applying the fundamental principle of the dynamic segment S_i (Equation 1 and 2) (Pustoc'h, 2007):

$$\sum \overrightarrow{F_{ext \rightarrow S_i}} = M_{S_i} \cdot \gamma(\overrightarrow{G_{S_i}}) \quad (1)$$

$$\sum \overrightarrow{M_{ext \rightarrow S_i}} = \frac{d(I_{S_i}(G_{S_i}) \cdot \overrightarrow{\Omega})}{dt} \quad (2)$$

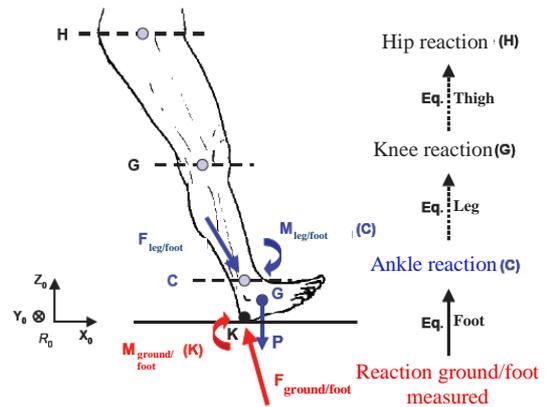


Figure 5: Modeling inverse dynamics of the lower limb during the stance phase of walking. The external mechanical action torser reduces the contact point K is known. The kinetic properties of the segments (foot, leg, and thigh) are estimated from anthropometric tables. The inter-segmental mechanical actions (H: hip, G: knee and C: ankle) are calculated by inverse dynamics (outside view-sagittal plane) (D'angeli and Gaviria ,1994).

In practice, the dynamic model is reduced to quasi-static at every moment of the movement. This dynamic approach is the method implemented to achieve the dynamic modeling of walking, (D'angeli and Gaviria ,1994).

2. 6 Determination of the articular contact reaction on the level of the hip

When the muscular efforts are determined, the response of articular contact $\overrightarrow{R_H}$ can be calculated by applying the theorem at every moment of the resultant dynamics applied to the system $S = \{\text{foot-leg-thigh}\}$ (Equations 3 and 4):

SEX	FEMALE
AGE	27 Years
SIZE	165 cm
WEIGHT	53 kg

Table1: Different characteristics from patient

$$\sum \overrightarrow{F_{ext \rightarrow s}} = M_{S_i} \cdot \overrightarrow{\gamma(G_s)} \quad (3)$$

$$\sum \overrightarrow{F_i} + \overrightarrow{P_{foot}} + \overrightarrow{P_{leg}} + \overrightarrow{P_{thigh}} + \overrightarrow{P_{ground \rightarrow foot}} + \overrightarrow{R_H} = M_{S_{foot}} \cdot \overrightarrow{\gamma(G_{S_{foot}})} +$$

$$M_{S_{leg}} \cdot \overrightarrow{\gamma(G_{S_{leg}})} + M_{S_{thigh}} \cdot \overrightarrow{\gamma(G_{S_{thigh}})} \quad (4)$$

The mass of the foot segment is negligible compared to the mass of other segments; the response of articular $\overrightarrow{R_H}$ contact at the hip is expressed as follows: $\overrightarrow{R_H} = M_{S_{leg}} \cdot \overrightarrow{\gamma(G_{S_{leg}})} + M_{S_{thigh}} \cdot \overrightarrow{\gamma(G_{S_{thigh}})} - \sum \overrightarrow{F_i} - \overrightarrow{P_{leg}} - \overrightarrow{P_{thigh}} - \overrightarrow{P_{ground \rightarrow foot}}$ (5)
By projecting this relation on each axis, the three components of the reaction of articular contact are determined at each time step during the stance phase. In general, the kinematic measurements give a complete description of the segmental movements in terms of positions, velocities and angular accelerations of the joints. Different characteristics on the subject are presented in Table 1, these characteristics are those used in the literature (Pustoc'h, 2007) and we will use later in our study to validate our results.

Our study concerns the analysis of the stance phase of walking (Figure 6). The average cycle of the walk we will adopt for our patients is 1.15 ± 0.05 seconds. It is consistent with the average described in the literature, the range of 1.11-1.39secondes.

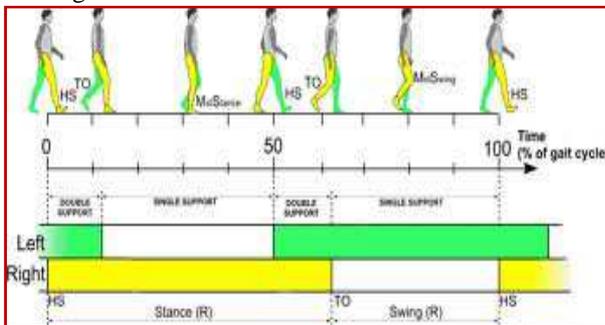


Figure 6: Definition of the stance phase and swing phase in the cycle of walking (Pustoc'h, 2007)

Concerning the evolution of the average transverse angle, it is almost constant throughout the duration of the

stance phase. This is not the angle (Figure 7) between 15 and 20 ° in our subject (Pustoc'h, 2007).

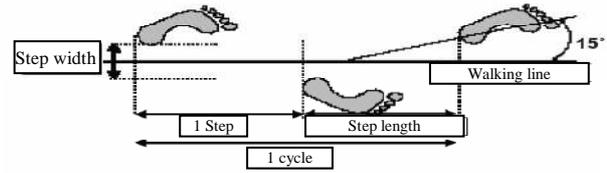


Figure 7: Spatial characteristics of walking: the step width, step length and step orientation generally equal to 15 ° (Pustoc'h, 2007)

2. 7 Determination of muscular effort

The muscular effort for the 14 muscles to the six movements of the hip are determined from the literature so that every effort be enabled or disabled during the entire gait cycle. And developing a dynamic model of human walking will be based primarily on the quasi-static approach. We will focus on efforts to reduce the level of the hip joint. For this we use the dynamic model described above and then use the parameters described in the literature (Pustoc'h, 2007).

2. 8 Determination of the reaction of articular contact

The reaction of articular contact is calculated from muscle strain by applying to every moment of the stance phase of the theorem resulting dynamics described in (§ 2.5). Figure 8 presents the results of the resultant joint exerted by the femur on the pelvis determined during the gait cycle (Wu and al, 2002).

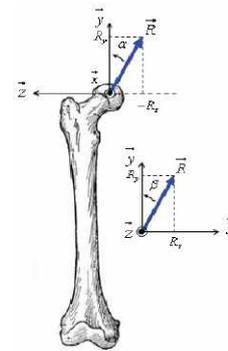


Figure 8: Orientation of the reaction of articular contact in the local anatomical landmark linked to the femur (Wu and al, 2002)

2. 9 FEM study of the hip joint after implantation of modular THP

In the simulation by the finite element method using a simplified dynamic model, using a quasi-static approach that allows us to calculate the effect of body weight reduces our patient at the center of the hip joint, then we make a simulation software ANSYS, the forces applied to the patient's body during a gait cycle are the center

point of application of the femoral head prosthesis of Figure 9 and the components shown in Figure 8.

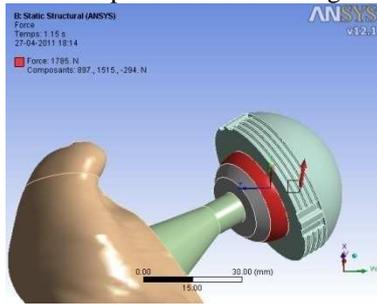


Figure 9: Positioning of the origin of our study

Cycle through the course of our patient is equal to 1.15 ± 0.05 seconds. The stance phase of walking is dividing in seven intervals. Curves (Figures 10, 11, 12) describe the three components R_x , R_y and R_z of force in Figure 8 for 6 simulated intervals. Components R_x , R_y and R_z are the components of the reaction of articular contact in the basin related to the anatomical landmark (Wu and al, 2002).

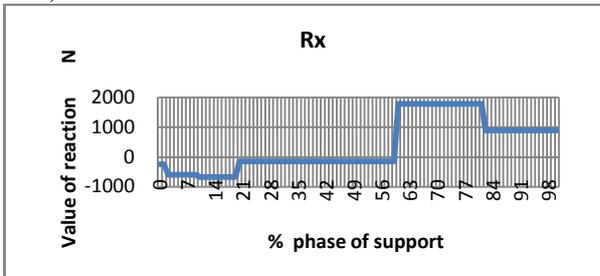


Figure 10: The component R_x force applied during the 6 simulated intervals (Pustoc'h, 2007)

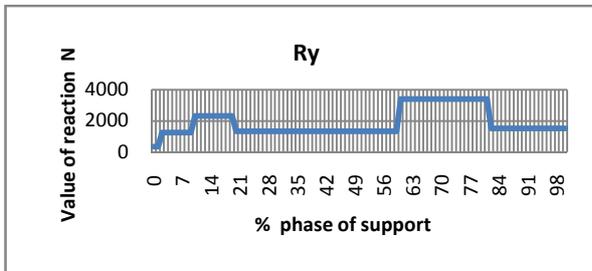


Figure 11: The component R_y force applied during the 6 simulated intervals (Pustoc'h, 2007)

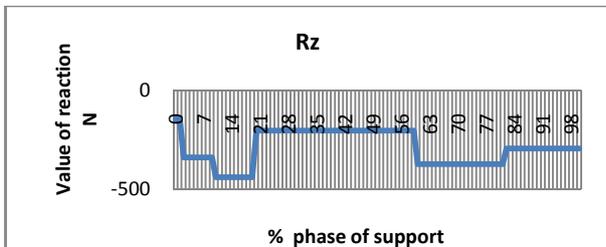


Figure 12: The component R_z force applied during the 6 simulated intervals (Pustoc'h, 2007)

3. RESULTS AND DISCUSSION

The prosthesis is implanted in an automatic way in the bone of the patient with an adequate design of the acetabulum UHMWPE according to clinical parameters, using a design factor of the femoral head and the creation of modular femoral stem from the channel modular patient. The subject of dynamic study is to test the resistance of the THP module to work due to the weight of the patient during the gait cycle. Thus, the results of the dynamics of THP module implanted in the patient's body are illustrated in Figure 15, it appears that the most solicited during the gait cycle is the part acetabular UHMWPE (figures 17 and 18), but despite this it has a maximum total strain of about $20\mu\text{m}$ and an elastic deformation equivalent to $93 \times \frac{10^{-5}\text{mm}}{\text{mm}}$ (figures 13 and 14), which is totally acceptable the scale of this study was to verify only the resistance of THP during a single gait cycle.

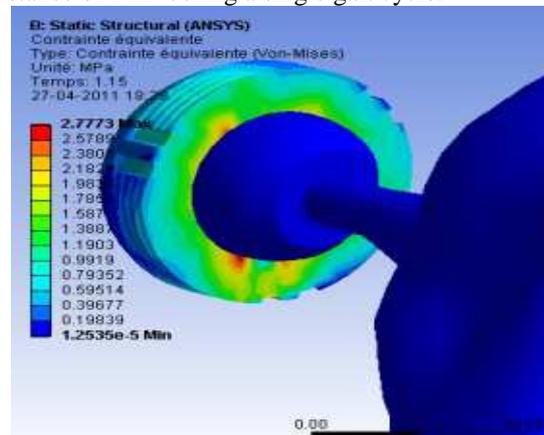


Figure 13: Distribution of equivalent stress in the chain PTH-bone-cement

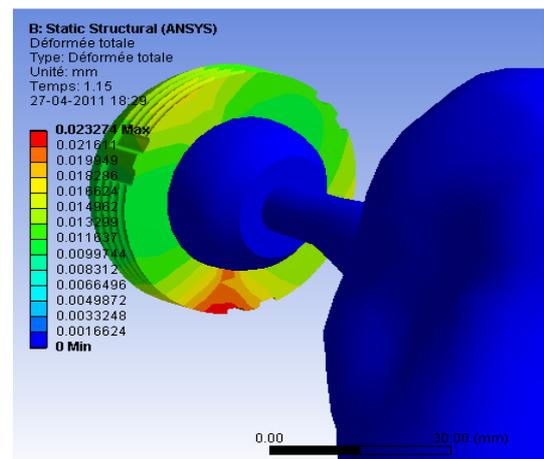


Figure 14: Distribution of the total deformation in the chain PTH-bone-cement

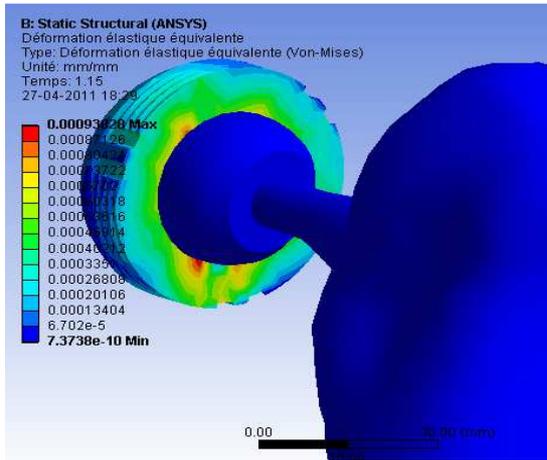


Figure 15: Distribution of the equivalent elastic strain in the chain PTH-bone-cement

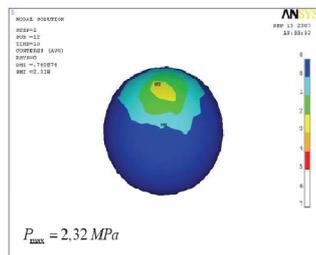


Figure 16: Pressure distribution to 10% of the support phase of the gait cycle (Pustoc'h, 2007).

The simulation results for the contact pressures on the surface of the cartilage of the femoral head are shown in Figure 16. These results are from the literature (Pustoc'h, 2007), in isolation after the implantation of THP, there remains only the cartilage of the acetabulum UHMWPE. Comparing these results with the literature (Pustoc'h, 2007), which represents the distribution of contact pressure in the cartilage during the gait cycle for a normal hip. The simulation results we obtained for equivalent Von Mises stresses are shown in figures 17 and 18. The follow-up evaluation of the contact pressure of the femoral head-acetabulum, can detect the most dangerous areas during a walk cycle in 1.15 seconds. The results of this study will be used later to scale the standard prosthesis if it does not withstand forces applied during the gait cycle.

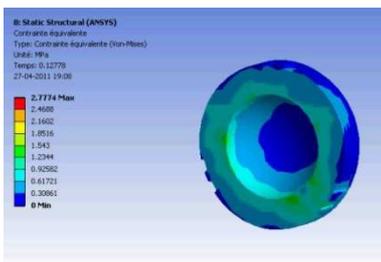


Figure 17: Distribution of contact pressure at the surface of the femoral head in 10% of the stance phase of gait cycle

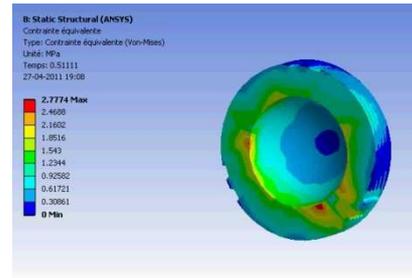


Figure 18: Distribution of contact pressure at the surface of the femoral head in 50% of the stance phase of gait cycle

After implantation of modular THP, it is clear that the pressure distribution in the contact area acetabulum-femoral head is the same as that of the natural hip joint, this is due to:

- The design tailored of the acetabular UHMWPE for the patient;
- The design of the femoral head from the channel modular, respecting the optimum thickness of 4 mm, to ensure proper distribution of pressure throughout the chain: Pelvis-THP-cement-bone this ensures that the patient's bone undergoes a fracture before THP or cement (Wroblewski and al);
- Automatic positioning of THP ensures that THP is the correct location (position and orientation), which makes it replaces the natural femur in order to maintain the optimal state of the articulation of the hip. The main role of the hip joint is to support the weight of the body during the various phases of the movement. Thus, it is useful to find a method that allows the proper replacement of this joint while preserving its natural characteristics before implantation of THP. The dynamic study allowed us to identify areas most in demand during the gait cycle, such as the acetabulum, which represents the center of this connection. The study of the hip joint after implantation of THP may be reduced to the study simply the prosthesis designed, ignoring that in the contact area of the femoral head-acetabulum effort are the same in both parties, for the acetabulum is directly related to the femoral head. With the aim of simplify the study for the standard prosthesis, one will make abstraction that only the study of the chains, head femoral – femoral stem, is necessary to determine the distribution of pressure during a walk cycle after implantation of standard THP.

4. CONCLUSION

The dynamic study of the hip joint after implantation of THP module allowed us to verify the accuracy of the quasi-static model used by comparing our results with those of literature; this study confirmed that THP modular measure is an ideal solution to replace the natural joint of the hip. The realization of a prosthetic femoral stem position by starting on the modular femoral canal has been the subject of various studies (PUSTOC'H, 2007), the main difference between this

work is how each author reconstructs the modules channel.

The approach to reconstruction of the modular channel of our patient is via a simple and feasible from a time perspective and programming, it relies mainly on the cloud points obtained after the segmentation of medical images.

Also, the automatic positioning of the modular prosthesis in the femur of the patient gives a more realistic model. After validating the simulation results of the module THP with the literature, we could verify that the acetabulum UHMWPE is the component that undergoes the greatest deformation. Because the acetabulum is directly related to the femoral head in our model, it is therefore necessary to analyze the dynamics of THP without the standard acetabulum UHMWPE because the same constraints on the contact interface acetabulum-femoral head in our model.

REFERENCES

- Oguz Kayabasi.,and Fehmi Erzincanli, 2006. Finite element modelling and analysis of a new cemented hip prosthesis, *Advances in Engineering Software* 37, p. 477–483.
- Oguz Kayabasi., and Bulent Ekici, 2007. The effects of static, dynamic and fatigue behavior on three-dimensional shape optimization of hip prosthesis by finite element method”, *Materials and Design* 28, p.2269–2277.
- Asheesh Bedi., Mark Dolan., Michael Leunig., and Bryan T. Kelly, 2011. Static and Dynamic Mechanical Causes of Hip Pain, *Arthroscopy: The Journal of Arthroscopic and Related Surgery*, Vol. 27, No 2, p. 235-251.
- Robin Martin, Patrick E. Clayson, Serge Troussel, Brian P. Fraser, and Pierre-Louis Docquier, 2011. Anterolateral Minimally Invasive Total Hip Arthroplasty. *The Journal of Arthroplasty*.
- Hai-bo Jiang, 2007. Static and Dynamic Mechanics Analysis on Artificial Hip Joints with Different Interface Designs by the Finite Element Method. *Journal of Bionic Engineering* 4, p. 123-131.
- L. Mattei., F.DiPuccio., B.Piccigallo, and E.Ciulli, 2011. Lubrication and wear modelling of artificial hip joints: A review. *Tribology International* 44, p. 532–549
- B.W. Stansfield , and A.C. Nicol, 2002. Hip joint contact forces in normal subjects and subjects with total hip prostheses: walking and stair and ramp negotiation. *Clinical Biomechanics* 17, p. 130–139.
- Kaku N., Tsumura H., Taira H., Sawatari T, and Torisu T, 2004. Biomechanical study of load transfer of the pubic ramus due to pelvic inclination after hip joint surgery using a three-dimensional finite element model, *Journal of Orthopaedic Science*, Vol. 9, p. 264-269.
- Briand D, and Bonnel F, 1994. Adaptations morphologiques à la marche bipède dans l'échelle animale. *La Marche humaine et sa pathologie*, Editions Paris Masson, p. 6-9.
- D'angeli-Chevassut M, and Gaviria M, 1994. La marche humaine : description et repères sémantiques. *La marche humaine et sa pathologie*, Editions Masson Paris, p. 9-20.
- Pustoc'h Audrey., 2007. *Élaboration d'un modèle mécanique de l'articulation de la hanche sous sollicitations dynamiques*. These de doctorat Université Claude Bernard - Lyon 1, France.
- Wu GS., Allard P, Kirtley C., Leardini A., Rosenbaum D, and al, 2002. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion--part I: ankle, hip, and spine. *International Society of Biomechanics, Journal of Biomechanics*, Vol. 35, p. 543-548.
- Doriot N and Chèze L, 2004. A three-dimensional kinematic and dynamic study of the lower limb during the stance phase of gait using an homogeneous matrix approach . *IEEE Transactions on Biomedical Engineering*, Vol. 51, p. 21-27.
- Cappozzo A., Della Croce U., Leardini A, and Chiari L, 2005. Human movement analysis using stereophotogrammetry: theoretical background. *Gait & Posture*, Vol. 21, p.186-96.
- Zatsiorsky VM, and Seluyanov VN, 1983. The mass and inertia characteristics of the main segments of the human body. *Biomechanics VIII-B (Edited by Matsui H and Kobayashi K, Human Kinetics Publishers, Champaign, IL)*, p. 1152-1159.
- Walker P. S., H. Kurosawa., M.D Rovick, and R. A. Zimmerman, 1985. External Knee Joint Design Based on Normal Motion. *Journal of rehabilitation research and development*, vol. 22, No. 1, BPR 10-41, p. 9-22.
- Y. Benabid, A. Aoussat, T. Chettibi “Design of high flexion total knee prosthesis considering activities of North Africa peoples” *Computer Methods in Biomechanics and Biomedical Engineering* Vol. 14, No. S1, August 2011, 21–23
- B. M.wroblewski., P. D. Siney, and P. A. Fleming, The Charnley Hip Replacement – 43 Years of Clinical Success. The John Charnley Research Institute, Wrightington Hospital, Nr. Wigan, U. K.
- Siegler S, and Liu W, 1997. The three-dimensional passive support characteristics of ankle braces. *Journal of Orthopaedics and Sports Physical Therapy*, Vol. 26, p. 299-309.
- Daniel P. Nicoletta, Ben H. Thacker, Hamid Katoozian, Dwight T. Davy, The effect of three-dimensional shape optimization on the probabilistic response of a cemented femoral hip prosthesis, *Journal of Biomechanics* 39 (2006) 1265–1278.