Effects of dynamic loading on the mechanical behavior of total hip prosthesis

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Abstract
Upon the arrival of the osteosynthesis, technological advancement has revolutionized the orthopedic surgery, producing increasingly efficient prostheses. Nevertheless, it turns out that over the years of lifetime these orthopedic implants presents complications in vivo. This requires in most cases a reoperation. To improve the total hip replacement lifetime it is imperative to investigate the effects of dynamic loading on the replaced joint. For this reason, we initially studied the mechanical behavior of their different components, according to the nature of hip’s movements, the exerted forces on the femoral head and particularly the muscular forces acting on hip-femur system. Then, we performed numerical simulations of a femur carrying a cemented hip prosthesis type (CMK3), the boundary conditions used in the simulations are identical to those applied in vivo during the daily activities.

Key words: Total hip prostheses, Surgical cement, Biomechanics, Muscle forces, Dynamic loading, Finite element analysis

1. Introduction

The durability of a total hip replacement depends on the mechanical behavior of each prosthesis component, especially the surgical cement. Until today the PMMA is the only material used to anchor the femoral implant in to the femoral bone during a cemented orthoplasty. In the cemented hip replacement, the cement is the most fragile material, which has two interfaces; one with the bone and the second with the prosthesis, this cement mantle is the weakest link in the load transfer system (implant – cement – bone).

However, despite the various disadvantages of PMMA, a recent study have shown that a lifetime of cemented hip orthoplasties in patients aged over 50 years is significantly high comparing to a non-cemented prostheses, at least 90% of those cemented prostheses reaches 15 years of lifetime in vivo (BALLARD T. W., et al, 1992).

This study comes within this context and with an objective to analyze the mechanical behavior of surgical cement under dynamic loading (real body loading). This behavior depends on interactions of the load transfer system (femoral implant - cement - cancellous bone - cortical bone) that are subject to very realistic loading conditions, based on measurement of contact force in vivo and the different EMG signatures of hip joint muscles, for an extended activities to include the major movements of the lower limb in the patient’s daily life activity.

2. The objective

The hip joint is the strongest joint in the human body and the most mobile of the lower limb, it finds its mobility due to the most powerful muscles of the human body, and it is held in place by a multitude of strong ligaments. This
anatomical specificity has secondary effects on the hip joint, like the irreversible injury, which requires sometimes a replacement of the damaged joint by an artificial articulation, called a total hip replacement or an orthoplasty, this artificial articulation must be strong enough to resist the most important forces in the human body.

The mechanical behavior of total hip prosthesis is, until now, under investigations. This study comes within this context as objective of a three-dimensional analysis of the mechanical behavior of total hip prostheses during the patient’s daily activities by the finite elements method. This behavior depends on the interaction of four components of the prosthesis exposed to a very realistic loading conditions and boundary conditions, based on measurements of the in vivo hip joint contact forces and different muscular efforts for an extended series of activities including the major movement of the lower limb in daily life of patients.

3. Materials and methods

To determine the mechanical behavior of total hip prostheses under dynamic loading, it is important to know the variation of muscular forces acting on the hip-femur system, and the resultant forces applied on the femoral head, during one cycle of each activity.

As it’s indicated in figure 5, the cement is supposed to interact with the femoral implant and the spongious bone, however in certain cases the surgeons confirms that they have reached the cortical bone zone when they were preparing the femur for the implantation, which is a new condition to be verified in the bio-competence, because of the cortical bone is much harder and stronger than the cancellous bone as we can realize in the material proprieties description below.

In this work, we will simulate the most unfavorable cases of the total hip orthoplasty, including the case of a direct contact : cement - cortical bone, and even walking with a 22 Kg load which will be simulated as an equally distributed weigh on both hands.

3.1 Muscular forces

The hip muscles are numerous, however, some have a low action on the hip joint articulation, or even negligible (BESNIER J. P., 1995). In this work, we have determined the efforts of the strongest muscles (such as the gluteus maximus, the gluteus medius, the small gluteus and the iliacus) (Fig. 1.a) (BERGMANN G, et al, 2000).

Fig. 1: a) Schematic representation of the hip’s strongest muscles:

Gluteus medius (1), gluteus maximus (2), TFL (3), gluteus minimus (4), semimembranosus (5).

b) Exerted forces by major muscles during a walk (determine by the instrumented femoral implant):

Gf: gluteus maximus, Pf: gluteus medius, Mf: gluteus minimus and the iliac muscle during normal walking.
The generated forces by each muscular contraction are determined by Electromyogram. The selected muscles exerts during all studied activities more than 90% of the total force exerted by all hip joint muscles (Fig. 1.b) (BOUZIANE M et al, 2009).

### 3.2 Boundary conditions

The instrumented THP implant with a telemetric data transmission allows the determination of contact forces generated at the hip, which is essential for the definition of total hip prostheses design (BERGMANN G, et al, 1998) (Fig.2.a). The processing of data that can be transmitted by a modern instrumented total hip prostheses, allows the determination of:

- the nature of exerted forces on the femoral implant (traction, compression, torsion or shear forces),
- the magnitudes of resulting forces and generated moments around the three axes (BERGMANN G. et al 1993),
- the tridimensional femoral implant deformation,
- the resulting temperature during practiced activities (BERGMANN G, et al, 2000).

![Fig. 2: a) Instrumented femoral implant, b) The variation of resulted forces $F_{res}$ in function of body weight during 1 cycle of the walking activities](image)

After having determined the muscular forces and the applied forces on the femoral implant of each activity, we projected all acquired forces (magnitudes and directions) on the adopted coordinate system (Fig. 3.a): the exerted force on the femoral head is simulated by the decomposition of resulting forces in the three main axes, the muscular forces are also decomposed according to the main axes but their application surface is on the bone and it is similar to their real application sites *in vivo*, while the knee is considered as a joint with two degrees of freedom, one displacement along the ‘Y’ axis and one rotation on the ‘X’ axis (Fig 3.b).

Then we proceed to simulate the mechanical behavior of the total hip prosthesis subjected to accurate body forces (dynamic loading). Six "6" essential activities exercised by the patient were selected for this study: normal walking, fast walking, slow walking, walking with an equally distributed load on both hands, ascending and descending stairs.
3.3 The analyzed structure

The numerical model of the analyzed total hip prosthesis is illustrated on figure 4. This model is composed of an implant (shaft and implant’s head) (Fig. 4.a), of a surgical cement (Fig. 4.b) and the femoral bone (cortical bone and spongious bone) (Fig. 4.c & d). The femoral implant and the femoral bone are attached together with the surgical cement as shown in Figure 5. The interactions between these components (implant – cement – bone) determines the lifetime of the cemented hip prostheses (BERRY D. J., et al, 1998).

This 3D bone model was realized directly from radiology images of the femur, in fact the CT-scan images allowed us to see the cross-section of the femur and according to the luminous intensity of tomographic images, two regions were
distinguished: cortical bone and spongious bone, then we imported these images by a 3D modeling software (solidworks) where we numerically built the two femoral component separately then with the assembly interface of ABAQUS software we arranged these two components in their precise positions and finely we cut the femoral neck and we created a housing in the femur’s upper extremity (proximal area) in which, the surgical cement and the femoral bone were inserted. (The patient weight is 980 N and his tall is 190 cm)

3.4 The adopted meshing and the material proprieties

For a better representation of the artificial hip joint components, the numerical models have been meshed separately using tetrahedral elements (Fig. 5). Because of the geometrical complexity of each component, we have reduced the element size by refining the mesh until having stable results (constant results even with smaller element’s size). The mechanical properties of bone and prosthesis’s components used in this simulation were defined according to (BOUZIANE M.M, & al, 2009) as follows:

|                | Young's modulus ‘E’ in MPa | Poisson’s ratio ‘ν’ |
|----------------|-----------------------------|--------------------|
| The cortical bone | 20000                       | 0.3                |
| The spongious bone | 132                         | 0.3                |
| The femoral implant | 210000                     | 0.3                |
| The surgical cement | 2000                      | 0.3                |

Fig. 5: Meshing illustration of the analyzed system (femoral implant, PMMA, spongious & cortical bone).

4. Results and discussions

In this title we will expose and investigate the behavior of femoral bone and cemented hip prosthesis under accurate in vivo loading and boundary conditions and we will study the intensity and distribution of normal, tangential and Von Mises stresses induced in the three components of the total hip prosthesis (implant – cement – bone) where the ‘Z’ axis is the vertical one and it’s the 3rd axis, the second axis is ‘Y’ and it’s the frontal one and the ‘X’ axis is the lateral axis (normal stress $\sigma_{zz}$ and the tangential or shear stress $\tau_{xy}$ are represented respectively by $\sigma_{33}$ and $\tau_{12}$).
4.1 bone’s behavior

The bone’s response to in vivo dynamic loading is illustrated below; Figure 6 shows that these stresses are distributed inhomogeneously all over the bone. The most significant stresses are oriented according to the 'Z' axis direction. In fact, the bone is subjected to both tensile and compression stress on the 'Z' axis direction and this is due to the anatomical specifics of the femoral bone, which induce flexion stresses all over this bone. But the 'F_z' (forces applied according to 'Z' direction) acting on the femoral head, will generate a compression stress that will be added to the compression stress induced by the bone flexion and that will increase both : the resulted compression stress $\sigma_{33}$ and the compressed section (inner side of the bone) at the expense of tensile stress and tension section (outer side of the bone), Fig 6.d shows that the resulted compression stress is 42.73Mpa while the tensile stress magnitude is 39.65Mpa, in consequence the neutral plane is displaced from the inner side to the outer side, which is located in the central plane of the femoral bone in case of pure bending stresses. These compression and tension stresses are distributed on both sides along the bone and divide this element into two parts denoted herein as 'inner side' and 'outer side'. Consequently, the bone is subject to compound stresses:

- A compression stress due to the weight and muscular efforts;
- A bending stress due to the force application surface that is not on the anatomical axis of the femur, and this leads to the formation of a lever arm for the efforts and induces a flexion torque and a bending stresses.
- The normal stresses $\sigma_{11}$ and $\sigma_{22}$ have comparable intensity magnitudes and are much lower than the third stress level.
Fig. 6: Level and distribution of stresses in the femoral bone (in MPa).

$S_{33}$ and $S_{12}$ are respectively $\sigma_{33}$ and $\tau_{12}$

Compared to normal stresses, the bone’s most significant shear stresses are about four times lower than the highest normal stress amplitude. These shear stresses are related to 'YZ' and 'XZ' planes ($\tau_{23}$ and $\tau_{13}$), while the intensity of those induced in the plane 'XY' ($\tau_{12}$) is even lower.

For a better analysis of the stress distribution in the femoral bone we have shown in Figure 7.a & b the variation of normal, tangential and Von Mises stresses along the femoral bone in the interior side. Fig. 7.a shows that the normal stress $\sigma_{33}$ is generated along the bone, it reaches the highest amplitudes in the region included between 75mm and 300mm away from the upper extremity. The peak of compression stress $\sigma_{33}$ is located in about 117mm away from the upper extremity. The variation of the other normal stresses $\sigma_{11}$ and $\sigma_{22}$ along the length of the bone is relatively very weak (Fig.7.a).

The strongest tangential stresses on the 'XZ' plane ($\tau_{12}$) are located near the proximal and the distal epiphysis (close to the extremities of the femoral bone) (Fig. 7.b), they have almost the same amplitude but opposite direction. The most important stresses induced in the plane 'YZ' of the bone are located on the proximal-median area. Away from this area the shear stresses $\tau_{23}$ and $\tau_{13}$ have a low intensity.
The variations of deferent stresses along the outer side of femoral bone are illustrated in Figure 7, where it appears that tension stresses $\sigma_{33}$ have almost the same amplitude variation as the compression stresses in the inner side but with opposite signs (Fig. 7.a & c), the tension stresses $\sigma_{33}$ reach their highest values at 186mm from the upper extremity of the femur. Just like the inner side, the amplitude of normal stresses $\sigma_{11}$ and $\sigma_{22}$ on the outer side is very low (Fig. 7.c).

The shear stresses generated during those activities, in the outer side of the bone are shown in figure 7. Unlike the shear stresses induced in the compressed side of the bone (inner side), the most important shear stresses induced in this side are related to ‘YZ’ plane and they are practically localized on the proximal and median areas.

![Graphs showing stress variations](image)

**Fig. 7:** Stresses variation along the interior side (a and b) and the exterior side (c and d) of the femoral bone.

### 4.2 Implant’s behavior

The simulation results exposed in Figure 8 shows that the most significant normal stresses are $\sigma_{11}$ and $\sigma_{22}$ and they are located on the neck of the implant. The stress cartography shows that the outer side of this neck is in tension, while the compression is located in the inner side and that $\sigma_{11}$ stress about five times more intense than $\sigma_{22}$, which is due to the direction of applied forces on the implant makes. On this zone of the implant $\sigma_{11}$ tensile stress is almost equal to the compression stress $\sigma_{33}$ and it is distributed practically in the same way on either side of the implant neck. The $\sigma_{33}$ stresses are concentrated on the lateral sides of the implant (Fig.8.d), according to their nature (a compression in the inner side and tension on the outer side) this stress distribution reveals that this element is under bending stresses, in
fact this flexion loading is transmitted to the cement first then to the femoral bone. The tension and compression stresses are almost equally distributed on either side but the compressive stresses are slightly more intense than tensile stresses.

Fig. 8: Level and distribution of induced stresses in the femoral implant (in MPa).

On the figure 9.a it appears that the strongest normal stress $\sigma_{11}$ of the inner side of the implant is located at exactly 17mm from the upper extremity. The $\sigma_{33}$ normal stresses are highly concentrated not only on the neck but also on both sides. The banding stress induces compressive forces in the inner side of the implant.

Just like the normal stresses, shear stress $\tau_{13}$ is intensively induced in the implant, its maximal value is located on the neck at 17mm away from the superior extremity, the other shear stresses are relatively very low (Fig. 9.b).

The most intense stresses of the outer side are located on the neck at 8.5mm from the head extremity which make the neck area under high tensile stresses where the maximal amplitude is located at neck’s mid-distance. The geometrical form of the neck causes the stress concentration especially $\sigma_{11}$ and $\tau_{13}$ stresses (Fig. 9.c). This same figure indicates also that the distal inner and outer sides of the implant are exposed to an important $\sigma_{33}$ stress. The level of other normal stresses in the distal side is almost insignificant.
In Figure 9.d is presented the variation of tangential stresses along the outer side of the implant. The neck is exposed to high amplitude $\tau_{13}$ stress, located at a distance of 8.5mm away from the head extremity, and they are much stronger than $\tau_{23}$ and $\tau_{12}$. Far from the neck area, the intensity of all shear stresses drops considerably.

![Fig. 9: Stresses variation along the interior side (a b) and the exterior side (c d) of the femoral implant.](image)

### 4.3 cement’s behavior

Intended to ensure three essential functions: adhesion, antibiotic transport and load transfer; the surgical cement is the weakest link of the load transfer chain (implant – cement – bone), but also his mechanical behavior determines the reliability, the performance and the durability of cemented total hip prostheses.

The maximum Von Mises stresses are located on the proximal frontal side and the distal lateral zone (fig.10.a), this is mainly due to the absence of the cancellous bone in these areas, which makes a direct contact between cement and cortical bone; while the rest of the surgical cement is under low stress intensity.

The cartography of normal stresses $\sigma_{11}$, $\sigma_{22}$ and $\sigma_{33}$ is illustrated on figures 10.b, c and d respectively. The normal stresses $\sigma_{11}$ and $\sigma_{22}$ are intensively localized on the proximal and the distal zone and they are generated by a tension forces on the proximal zone and a compression forces on the distal zone; away from these areas, the intensity of normal stresses $\sigma_{11}$ and $\sigma_{22}$ is negligible.

![Image of normal stress distribution](image)
This cement–cortical bone direct contact in the two extremities of the surgical cement was created in order to know the mechanical responses of the weakest link of the loads transfer chain if the surgeon reaches the cortical bone zone when preparing the femoral implant housing.

In addition to the proximal frontal region, the distribution of $\sigma_{33}$ stress generated in the cement is totally different from the two other normal stresses (Fig. 10). Indeed, $\sigma_{33}$ is the result of the bending force which is transmitted from the femoral implant, and just like the other components it induces in the outer side of the cement tension stresses and in its inner side compressive stresses, and unlike $\sigma_{11}$ and $\sigma_{22}$, the normal stress $\sigma_{33}$ affects softly the distal zone of the cement.

In figures 10.e, f and g are shown the levels and distribution of shear stresses induced in the three planes 'XY', 'XZ' and 'YZ' of the surgical cement, denoted respectively $\tau_{12}$, $\tau_{13}$ and $\tau_{23}$, they are distributed almost in the same way on cement, they are concentrated on the distal zone.

Beyond the distal zone, the shear stresses induced in the surgical cement are relatively low, however, the direct interaction between cortical bone and cement has affected the stresses level in the contact regions but the highest amplitude of shear stresses is much lower than the ultimate shear stresses.

In the outer part of the surgical cement we noticed that the distal zone is under the highest $\sigma_{33}$ compressive stresses comparatively to $\sigma_{11}$ and $\sigma_{22}$ (Fig. 11.c).

The amplitude of shear stresses relating to the planes 'XY', 'XZ' and 'YZ' in the inner side are much lower than those resulting on the outer side (Fig. 11.b & d), while the maximal amplitude is recorded on the extremities of the inner side and only the distal zone of the outer side (Fig.11.b & d). It appears that the cement is exposed to a low intensity share stresses compared to the other components.
5. Comparative study

We have previously shown that the nature of the activities exerted by the patient determines the level of stresses which are induced all over the total hip prosthesis components. The comparative analysis of the results obtained during the walk activities (slow, normal and fast walking simulation) shows that the Von Mises stresses induced in bone and surgical cement are related to the speed of this activity. Indeed, more this activity is exerted at a slower rate: more the stresses induced in total hip prosthesis components are higher (Fig.12.a).

In comparison between these six activities (slow walking, normal walking, rising and descending stairs, fast walking, walking carrying a load on both hands), the activity that generates the most important tresses in the bone and surgical cement is the last one, indeed walking with an equally distributed load on both hands increase significantly the amplitude of the applied forces on the femoral head (Fig.12.b).

This is explained by the fact that the load is a foreign body that will affect directly the prosthesis through the charge transfer chain (hands to the arms to the spine to the pelvis then to the femur); so in addition to the weight of the load, the hip-joint muscles will exert supplementary force to create an equilibrium moment.

During the practice of the studied activities, each component was differently loaded: the strongest stresses induced in the surgical cement were located in its extremities (proximal and distal zone), whereas the major stresses generated in the bone were located along its median zone, while the femoral implant most important stresses appeared in the proximal zone, precisely in the implant’s neck.
6. Conclusion

The level of exerted forces on the femoral head depend on the nature of the performed activity, while the generated stresses depends much more on the forces directions than on their levels.

The femoral implant is the most loaded component of the total hip prosthesis and the maximal stress is recorded in the implant’s neck.

The femoral bone is subjected to bending stress all along its length, and it’s according to the vertical direction ‘Z’ that the most intense $\sigma_{33}$ stresses appears, specifically in the median zone. The compressed side is slightly bigger than the one in tension, and inner side is more loaded than that outer side.

The surgical cement is the component the less loaded and during all the simulated activities he verified the bio-competence conditions and requirements even in the presence of non-standard conditions (22 Kg load lifting and direct contact with the cortical bone); the highest stress amplitudes appeared on the proximal and the distal zones, so those regions are the most vulnerable ones, and they are the most exposed zones to the cracks initiations.

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