Mechanical Behavior of Bone Cement under Dynamic Loading

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Abstract

In orthopedic surgery and particularly in total hip arthroplasty, The fixation of the implant is generally made by the surgical cement, constituted essentially by polymer (PMMA), It is necessary to know the strengths applied to the prosthetic articulation during the current activities exercised by the patient in their life, to know the distribution of the constraints in the system (bone - cement - implant).

This study aims to analyze numerically using the finite element method, the effect of activities (dynamic loads) of the patient on the level and distribution of stresses generated in the components of total hip prosthesis. Five activities, the most frequently performed by the patient such as normal walking, the up and down stairs, sitting and up from chair, were selected for this study. For this purpose, a three-dimensional model of the total hip prosthesis has been developed. The results obtained from this model show that the total hip replacement components and especially the bone cement are more highly stressed during the process of climbing stairs. These excessively high loads can lead to damage of the cement and thus the loosening of the prosthesis.

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1 Introduction

Total hip replacement is an intervention of a biological articulation with prosthesis; her effectiveness depends on the quality of reconstruction, on architecture, on the mechanics of the hip, integrity and equilibrium, for those used as medical implants for orthopaedic and dental surgery (80,000 hip replacements and 30,000 knee prostheses are implanted annually worldwide). According to the Swedish Register of ATH over a period of 19 years, 7.1% of patients who underwent cementless ATH will need a second surgery for replacing the first; this number is 13% for patients who underwent cementless ATH. The main reason revisions ATH is aseptic loosening with 71% of cases, followed by infection with 7.5% (Herberts and Malchau, 2000 [1]). RCRA, 55% of cases require revision due to aseptic loosening. Two ways of implanting femoral exist, the choice being made by the surgeon according to the age of the patient and the quality of the bony support. If the beginnings of arthroplasty to the 19th century, it was not until the late 1950s when an English surgeon Sir John Charnley [2] used an acrylic resin cold, poly methyl methacrylate (PMMA) of surgical quality.

PMMA is the only material currently used for fixing prostheses in the bone during cementing arthroplasty; this attachment is via a mechanical coupling due to the overlay of the cement to the bone surface defects (anchor). The roughness is a determining parameter for fastening the bone-cement-implant system; it can lead to a more secure attachment of cement by [3], but it can also be the origin of the initiation of cracks due to notch effect in cement which can lead to loosening [4].

Cement is the weakest link of the chain transfer load implant-bone-cement; its damage is responsible for loosening interface at the cement-implant where micro-cracks over time fatigue develop greater extent and take more significant sizes. These fissures lead to both the destruction of the cement and to mobility of the implant within the bone. Such behaviour results in pelvic fracture in the patient [5] with unbearable pain.

Several authors have published values coxofemoral contact forces established from different phases of the analysis of a movement and calculated using simplified models muscle during various patients' daily movements, only Brand et al.

[6] compared calculated and measured by the data which were obtained, however, at the different moments. The strengths of contact hip measured in vivo with instrumentation implants were obtained at first by Rydell [7], Englais [8] and Kilvington [9]; these forces solicit the implant-bone-cement system, and determine the lifetime of this system. This affects the performance, reliability and stability of cementless total hip arthroplasty. Several research works were dedicated to the analysis of the damage of the cement which is largely responsible for the loosening of the total hip prosthesis. So Mr. Bouziane et al. [10] Zouambi et al. [11], Benouis et al. [12] analysed by the finite element method the effect of the microcavity on the behaviour of hip cement. These authors showed that these defects weaken the material due to notch effect and interaction. M. Benbarek et al. [13] have studied numerically by the MEF the mechanical behaviour of this binder containing a foreign body; they show that the latter is the stress concentration seat. In another numerical study, authors [11] reported the effect of the presence of micro-cavities on the crack propagation path in the primed cement; they conclude that this path is strongly influenced by the stress field around micro defect. Serier B. et al. [14] analysed the effect of walking forces on the fracture behaviour in cement. They show that the risk of initiation and propagation of the crack is more likely in the case load at an angle between 40 ° and 120. This risk is due to the low tensile strength of the cement. Other numerical work [12] showed that the existence of microcavity promotes the initiation and propagation of cracks of the cup of cement. The stress intensity factor is strongly influenced by the nature of the bone-cement interface, cement-implant. Propagation mode is dependent on the priming site of the cracks initiated microcavities.

This study aims to analyze the effect of the daily activities of the prosthesis on the mechanical behaviour of the system (bone - cement - implant). This behaviour is analyzed in terms of the level and distribution of the stresses induced in the components of the artificial joint. This study was performed using the software (ABAQUS) for dynamic tests on a real model of the hip, and the different phases of movement established by Bergman after total hip arthroplasty [15].

2 Study of patient's movement
### 2.1 Dynamic movement of the hip

Bergman and al studied 9 activities of the current life at four patients from 51 to 76 years old, whose weight varied from 702 to 980 N, which had been subjected to a complete arthroplasty of the hip owing to a coxarthrosis (in 3 cases) or of a vascular necrosis of the successive femoral head in a fracture (in 1 case).

The forces exerted on the total hip replacement are related to the nature of the movement made by the patient. Table 1 are grouped on the forces applied to the femoral head for each activity:

Table 1 Amplitude of effort and cycle time for each activity.

| Activity	Nr of patients	Effort (% the body weight)				Cycle Time (S)
		-Fx1	-Fy1	-Fz1	F	
slowly walking	3	51	36	235	243	1.25
normal walking	4	54	32	225	233	1.1
fast walking	3	52	33	243	251	0.95
climb stairs	3	60	61	237	252	1.59
down stairs	3	60	39	253	261	1.44
sits on chair	3	43	08	150	156	3.72

The effort of the activity considered in this study corresponds to the maximum force of the curve "force versus time shown in Fig. 1. This stress is reached for a 0.9s time for a normal gait cycle.
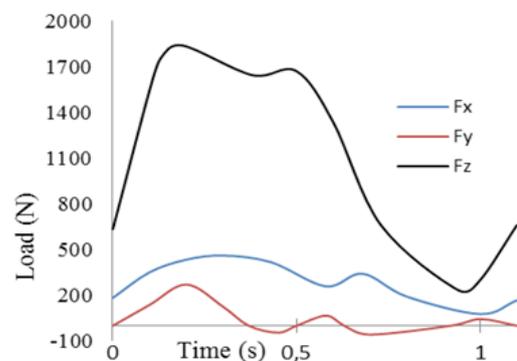
Fig. 1 Decomposition of a normal operating cycle
Corresponding to the efforts applied to the femoral head

2.2 Load applied to the femur

The strength of contact at the level of the hip joint, the amplitude F and of components-Fx1, -Fy1 and -Fz1, transmitted by the implant acetabular in the femoral head was determined in the system of axes of the left of femur R1 = (O, x1, y1, z1) in Fig. 2.
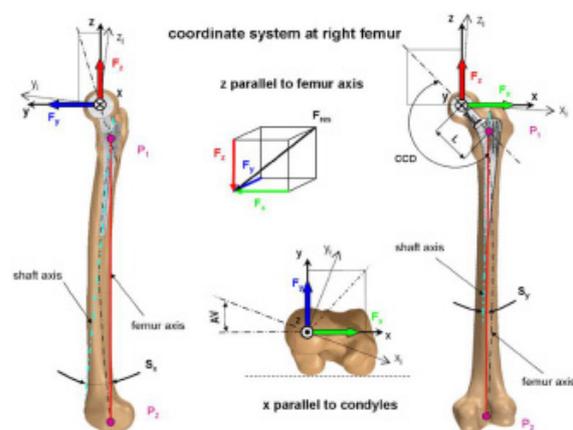
Fig. 2 The adopted coordinate system: Fx, Fy, Fz are the component of the resulted force applied on the femoral head, , (O, X i , Y i , Z i) is the implant system which is actually use by the computer to calculate Fx, Fy, Fz and the resulting force F $_{res}$.

3 Materiel and Method

3.1 Model and Design

In this work, we used the third generation model, of CHARNLEY MILLER KERBOUL (CMK3) who is represented in the Fig. 3. The choice fell on this model for its availability on the market.
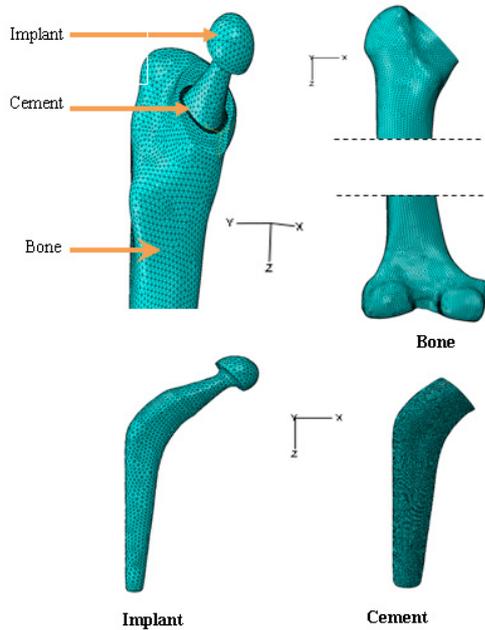

Fig. 3 The conception of the model CMK 3 with Software ABAQUS (mesh boundary)

Three-dimensional numerical analysis by the finite element method was performed using the code calculations "software" ABAQUS. The patient's activities selected for this study are: normal walking up and down stairs, standing and sit on a chair. These patient's daily activities were simulated by dynamic variable load with time. To do this, a numerical three-dimensional model was developed (Fig. 2). This model allows representing geometrically a bone-cement-implant system and applying the mechanical laws as a deformable solid subjected to dynamic loads.

The finite element simulation requires a precise description:

- The implant (3D geometry, mechanical properties, characteristics of the cement-bone-implant interface) (Fig. 3);
- Bone structure (3D geometry, distribution of bone density, mechanical properties, behavior laws);
- The surgical cement (3D geometry, mechanical properties, the properties of bone-cement interface, cement-implant.
- System load conditions (joint contact forces, muscular strength). This description and the use of finite element method to determine a number of biomechanical variables such as the level and distribution of stresses in the bone-cement-implant system.
- The tetrahedral elements were used for the bone cement and the mesh of the implant as shown in Fig. 2. The reliability of the results requires a very fine mesh of this system.

3.2 Boundary conditions

The boundary conditions imposed on the structure studied are given in Table 2 and Fig. 4:

- A load applied to the femoral head as a function of time for each movement;
- A load applied in the proximal area of the outer part (abductor muscle force);
- A load applied in the proximal region of the anterior part (vastus lateral force);
- A flush imposed on the femur end (Fig. 3).

Table 2 External Forces to exercise PTH system

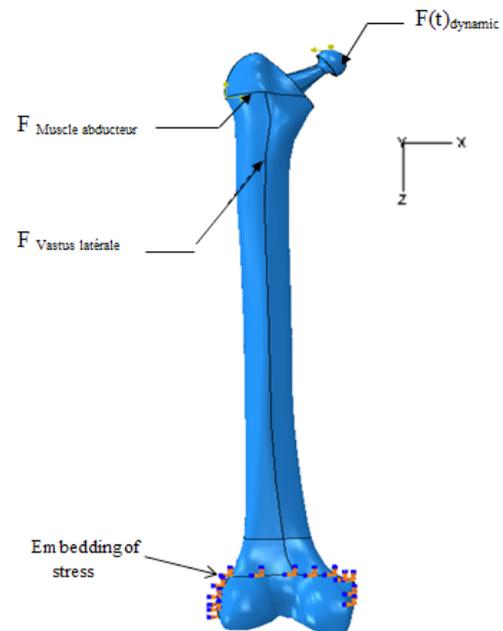| Effort (N)	X	Y	Z
F (t):Force applied to the head of PTH	Fx(t)	Fy(t)	Fz(t)
Abductor muscle	465.9	34.5	695.0
Muscle Vastus Side	-7.2	-148.6	-746.3

Fig. 4 Representation of the boundary conditions imposed on the analyzed structure

3.3 Materials and properties

The materials of the bone-cement-implant system, supposed to be homogeneous, used in this study are defined by their Young's moduli, their Poisson's ratios and densities (Table 3).

In the bone-cement-implant system, surgical cement (PMMA) represents the weakest link:

-Rupture strength:

- In traction 25 MPa
- In shearing 40 MPa
- In flexion 50 MPa
- In compression 80 MPa
- Elongation at break 5 %
- Resistance to fatigue in 108 cycles 14 MPa
- Crack resistance KIC 1 to 1.9MPa m1/2
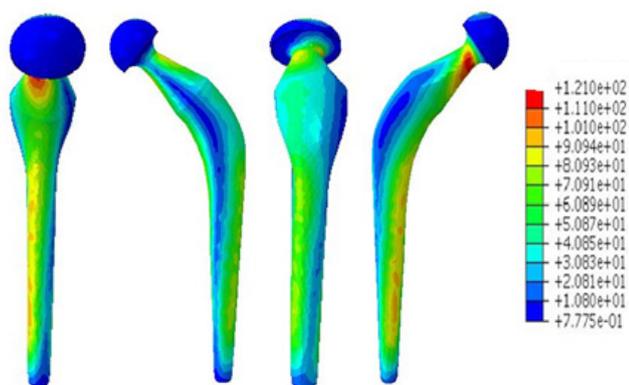
Table 3 Mechanical properties of cemented PTH system.

	Materials	Young's modulus (Mpas)	Poisson Coefficient	Density (kg /m3)
Cortical bone	/	21.000	0.3	1990
Cancellous bone	/	132	0.3	600
PTH	Stainless steel (316L)	210.000	0.3	7900
Cement	PMMA	2000	0.3	1200

4 Resultant and analyse

An analysis of the level and distribution of the equivalent von Misses stress in the implant-bone-cement system based on the patient's activities was conducted. As it has been stated previously, the effect of five activities; normal walking (climbing and down Stair, sit and up from the chair) on the mechanical behaviour of the artificial joint (total hip replacement) in terms of variation of the stresses induced in each of the components of this joint. The objective is the detection of the activity leading to the reduction of the life of the prosthesis by increasing the likelihood of loosening or risk of losing by breaking the bone cement.

4.1 Stress analysis in the prosthesis

In this part of the work we analysed numerically by the finite element method the level and distribution of the equivalent stress of Von Misses induced in the implant during the five above activities and exerted by the patient. Fig. 5 shows the amplitude of the stress generated in this element of the structure during a normal gait cycle. This figure clearly shows that the stresses are highly concentrated on the top of the neck of the implant, part opposite that where efforts are applied to the structure. These constraints pose no risk of damaging to this highly rigid member.

The effect of other activities are summarized in Table 4 shows the intensity of the equivalent stress of Von Mises generated in the three parts of the implant (proximal, medial, distal), for each movement performed by the patient. Whatever the activity of the patient, these constraints are located on the proximal zone of the femoral neck of the implant. This behavior is due to the site (head of the implant). The forces applied (up to four time's patient weight in some activities). The form and size of the implant neck facilitate the stress concentrations in the region of this element of the structure. The level of stress decreases along the implant, the proximal part to the distal as shown in Table 4. This behavior is due to the conditions of limits on the structure.
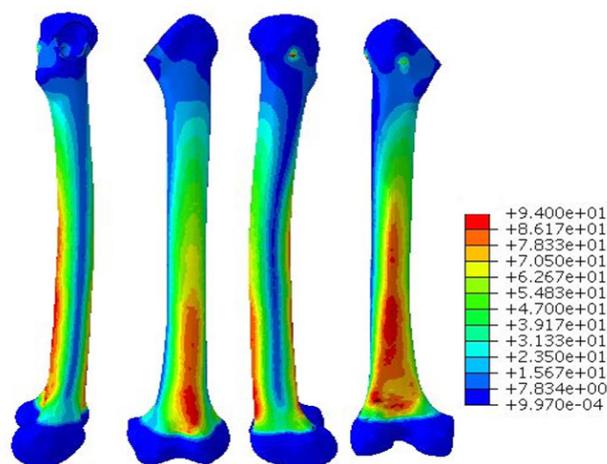
The movements climb stairs solicit more strongly the implant and especially the femoral neck.

Table 4 The effect of patient activities equivalent Von Mises stresses induced in the three parts of the implant

part of PTH / Activity	Proximal (MPa)	Medial (MPa)	Distal (MPa)
Down stairs	160	107	25
Climb stairs	140	105	34
Normal walking	120	100	7
Sits on chair	60	55	30
Up from the chair	60	55	8

4.2 Stress analysis in cortical bone

The damaged cortical bone, fixed to the implant through the bone cement received via this binder, efforts of this element of the prosthesis. Fig. 6 shows the magnitude and the distribution of the equivalent Von Mises stress induced in the bone during a normal gait cycle.

Fig. 5 Constraint Von Mises induced in the implant during a cycle of normal walking to the max point at t = 0.19s

Fig. 6 Von Mises Stress induced in the bone during a cycle of normal walking at max point t = 0.19s.

In the distal region where the stress reaches its maximum level. This behavior is due to the damage of the upper part in bone replaced with the implant. Like the prosthesis, bone is most loaded when the patient moves up or down stairs (Table 5).

Table 5 Effect of patient activities equivalent Von Mises stresses induced in the three parts of the cortical bone.

Part of Bone / Activity	Proximal (MPa)	Medial (MPa)	Distal (MPa)
Down stairs	41 MPa	114 MPa	125 MPa
Climb stairs	41 MPa	114 MPa	125 MPa
Normal walking	31MPa	86 MPa	94 MPa
Sits on chair	17 MPa	66 MPa	73 MPa
Up from the chair	17 MPa	53 MPa	48 MPa

The stress level obtained from the medial and distal regions is primarily due to the boundary conditions imposed on the bone-cement-implant system and in particular the frame of the lower part of the bone.

4.3 Stress Analysis in PMMA cement

Several studies showed that, under the same conditions and surgical preparation of bone cement techniques, the loosening is primarily due to the damage of cement by T.b. Moor et al. [17]. Cement plays different roles in: ensure the fixation of the bone to the implant, the load transfer of the bone to the implant and transporting gentamicin on the interface between bone and cement. To ensure the latter, the cement should contain some porosity density. The effect of this porosity on the behavior of cement has been the subject of several studies (David A. Hoey [18], Zouambi, 2013 [11]).

The latter weakens the cement by notch effect and inter-action. The durability of the total hip prosthesis depends of mechanical behavior of orthopedic cement. The analysis of these constraints is important for the performance of the artificial joint. This is therefore the objective of the latter part of this work. The results thus obtained are in the Fig. 7 Which shows the level and the distribution of the equivalent stress Induced in the orthopedic cement during a cycle of normal walking of the patient as a function of time. Analysis of these results clearly shows that the highest stresses are located on the proximal part of the cement. It is climb or down stairs that the patient more intensively applies the cement (Table 6).

In our boundary conditions, these higher stress level cement breakage threshold may initiate and propagating cracks in fatigue leading to the loosening of the cement rupture prosthesis. This behavior is in good agreement with other studies [E. BIALOBLOCKA [19], T.B. Maure [17], A.B. Lennon [20], D. T. Yang [21] Where loosening was observed. These authors explain Bone-implant deboning by breaking the cement Where fatigue cracks were observed. In another study,

Rudolf Marx et al. [22] have shown that one of the few problems of total prosthesis knee cemented (PTG) is the aseptic loosening of the tibia component due to the degradation of the interface between the bone cement and the tibia shaft metal component. The most popular activity is normal walking; it induces relatively low constraints and does not constitute any risk of damage to the cement. The most popular activity is normal walking; it induces relatively low constraints and does not constitute any risk of damage to the cement.

To avoid stress concentrations in the femur, patient activities must be limited by the orthopedic surgeon. Patient movements such as up and down stairs should be exercised with great caution. Other activities, such as the lifting of charges, running and jumping, must be systematically avoided.
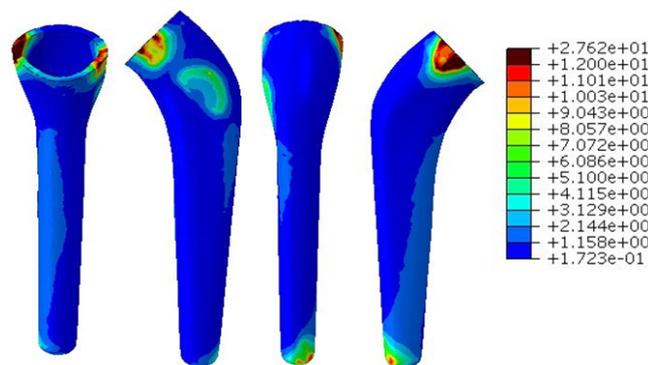
Fig. 7 Von-mise stress induced in the cement during a normal walking cycle at the max point (t = 0.19s).

Table 6 Effect of Patient Activities On Equivalent Von Mises stress induced in the three parts of the orthopaedic cement.

Part of Bone / Activity	Proximal (MPa)	Medial (MPa)	Distal (MPa)
Down stairs	36 MPa	3 MPa	12 MPa
Climb stairs	34 MPa	3 MPa	12 MPa
Normal walking	27 MPa	3 MPa	12 MPa
Sits on chair	13 MPa	2 MPa	13 MPa
Up from the chair	18 MPa	2 MPa	8 MPa

4.4 Stress states in orthopaedic cement

Surgical cement is the most fragile component of the implantology system; its damage is responsible for the loosening of total hip replacement. Therefore, an analysis of the states of stresses induced by a cycle of climbing stairs (Fig. 8) stressing the proximal part of the cement more strongly. This figure shows that this part is the seat of the most important stresses which are in compression and relative to the axes xx and zz. Compared to these solicitations, the stresses of tension Specific to these two axes are reduced by half. The existence of these two stress states clearly illustrate that the surgical cement is in flexion. These forces are essentially the cause of the propagation of cracks in mode of opening (mode I) observed by Serier et al. [23].

The strongest shear stresses relate to the yoz plane of the os-cement-implant system. These constraints are responsible for the damage of Mode II cement observed by M. T. Achour et al. [24] and Serier et al. [23].

The surgical cement has a very low tensile strength versus compression. This is why we have shown in Table 7 the most important normal stresses of tension.
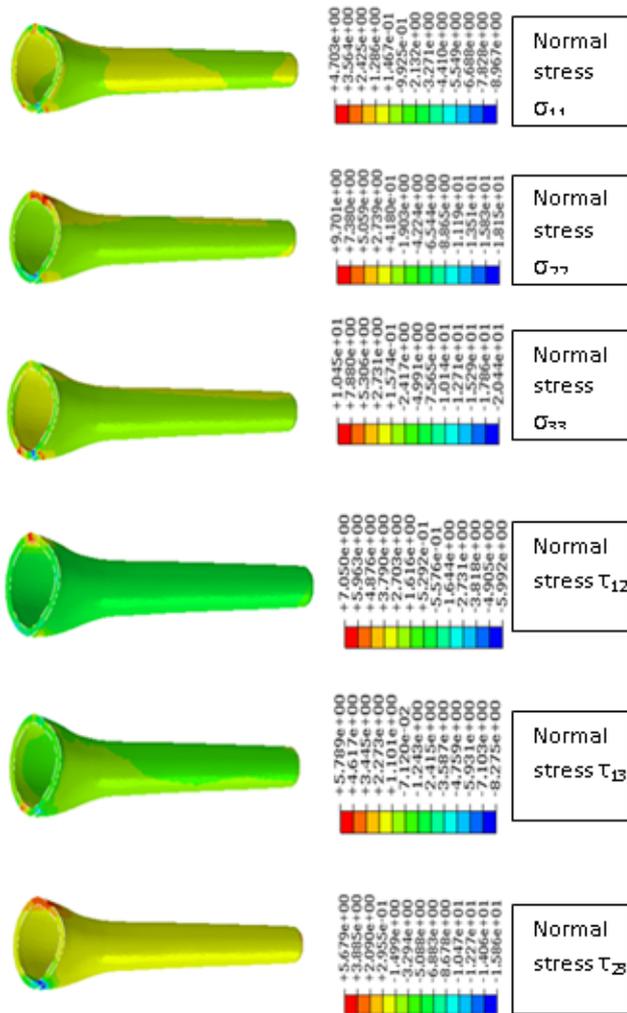

Fig. 8 Normal and tangential stresses induced in PMMA cement when climbing stairs at max point T = 0.3s

Table 7 Effect of patient activities on normaland tangential stresses induced in the cement.

Stresses (Mpas) Activity	σe	$\sigma 11$	$\sigma 22$	$\sigma 33$	$\tau 12$	$\tau 23$	$\tau 31$
Normal walking	26	6.5	8.6	9	9.8	6	5
Climb stairs	36	6.3	10	10.8	7.5	5.7	6.1
Down stairs	38	10.3	10.2	11.8	8	7.6	6
Sitting on chair	14	4.2	4.7	4.8	3	4	4
Up from the chair	18	4	5.8	5.7	3.8	4.3	4

5 Stresses following along the cement

In this part, an analysis of the stresses along the cement of left and right part from the proximal zone to the distal. These normal stresses and shear stresses induced by a climbing cycle of the stairs are shown in Fig. 9.
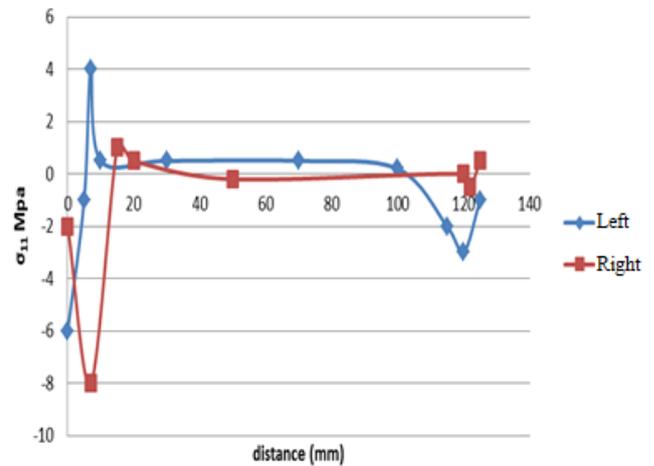

Fig. 9 Stress σ11 in climb stairs activity of the left and right side of PMMA cement

The variation of the normal stress Ϭ11 along the cement is illustrated in the Fig. 8 these stresses put the two parts left and right of the cement in tension and compression, they solicit more strongly the proximal area of this binder. Far from this zone (medial zone) the stresses drop considerably and then increase again in the distal part. The distribution of the normal stresses Ϭ22 is not homogeneous. Indeed, they are intensively concentrated on the upper left part. The normal stresses Ϭ33 induced along the axis (ZZ) of the structure and along the orthopedic cement are concentrated on its upper left area, they are almost doubly more intensive than the stresses induced in the opposite part of the cement. (Fig. 11). The normal stresses generated in the cement when climbing stairs by the wearer of the prosthesis in the medial and distal regions are lightly marked. Compared to the normal stresses developed in the two axes Ϭ11 and Ϭ22 of the structure, the third component more intensively applies the cement.

The tangential stresses τ12 Relative to the (XoY) plane of the structure is strongly concentrated on the cement edge as shown in Fig. 12 Indeed, the non-adherence of the cement to the implant and the bone leads to an increase in stresses in this site. Their highest level is reached in the proximal left part of the cement the Fig. 13 illustrates the variation in the shear stress in the plane (XoZ) of the cement while the patient climbing the stairs. These stresses concentrated on the upper left part of the cement, relax completely in the medial zone and then grow again in its distal part. The stresses relating to the right part of the cement have small amplitude.

The tangential stresses induced in the third plane of cement (YoZ), strongly concentrated on the left proximal zone of the
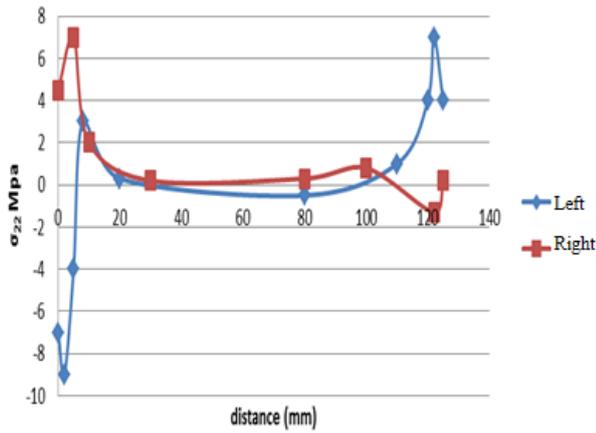
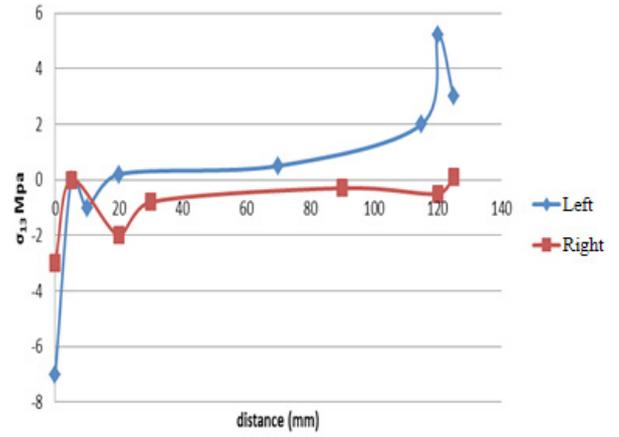**Fig. 10** Stress σ22 in climb stairs activity of the left and right side of PMMA cement

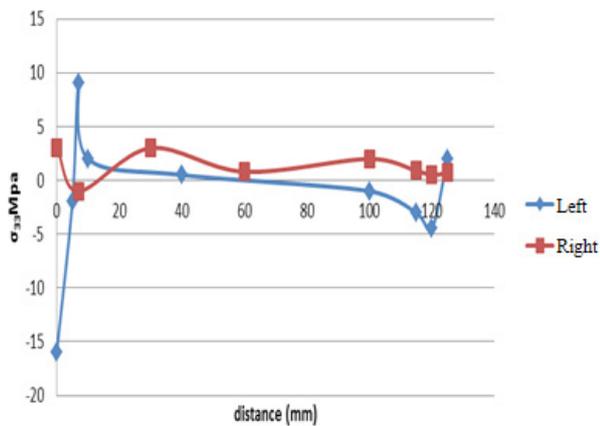**Fig. 13** Stress τ 13 in climb stairs activity of the left and right side of PMMA cement

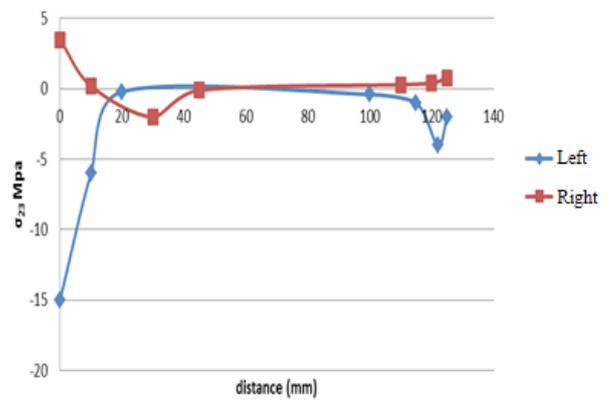**Fig. 11** Stress σ33 in climb stairs activity of the left and right side of PMMA cement

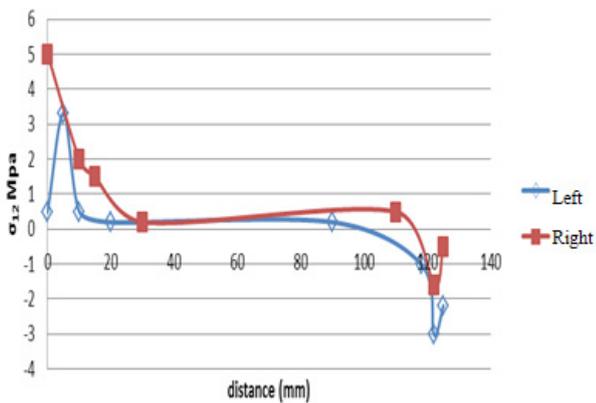**Fig. 14** Stress τ 23 in stair climbing activity of the left and right side of PMMA cement.

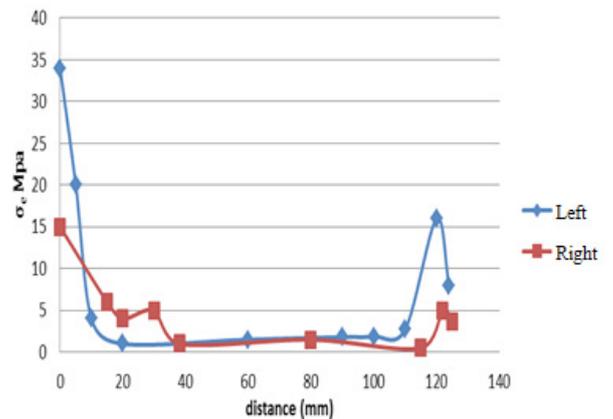**Fig. 12** Stress τ 12 in climb stairs activity of the left and right side of PMMA cement

Fig. 15 Von-misses stress (σe) in climb stairs activity of the left and right side of PMMA cement.

cement, are much greater than the two other shear stresses (Fig. 14), Their intensity is canceled on the medial part then increased slightly in the lower region of orthopedic cement.

On the right distal part the stresses are of a level about three higher than that resulting from the same part but on the left.

The stress concentration located in the proximal left area in the cement seat. The local equivalent stress is higher than that of the tensile strength of this binder (Fig. 15). It is in this area or the risk of damage by breaking the cement is most likely.

6 Conclusion

The results obtained in this work can be concluded that:

- Those activities conducted by the wearer of the total hip prosthesis induce equivalent stresses whose intensity is distributed homogeneously;
- The most important constraints are located on the top of the cement. This constraint whose level varies along the cement depends on the nature of the movement exerted by the patient;
- The most important stresses are located on the upper part of the femoral stem and the cement. These stresses, the level of which falls along these constituents from this zone to the distal one, depend on the nature of the movement exerted by the patient
- The highest equivalent stresses are localized in the proximal part of the cement, compared to the other two zones (medial and distal), the constraints in the cement resulting from the five activities exerted by the patient are not very intense.
- The equivalent stress of Von Mises induced localization in the orthopedic cement during climb and down stairs exceeds the threshold of breaking in tension of this binder. It is in the proximal part where the local stress is greater than the tensile stress of the orthopedic cement;
- The existence of normal stresses tension and compression show that, during the patient's activities, the surgical cement is subjected to bending forces.

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