THE INFLUENCE OF STEM SURFACE IN MICROMOBILITY AND CEMENT BONE STRESSES

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Abstract. Cemented hip prostheses have produced excellent clinical results and THR is at that moment one of the most successful surgical techniques, with good success rates. Cemented fixation represents 87% of the total number of replacements according to the Swedish orthopedic register. Therefore, it is important to reduce revisions and understand why revision happens. The purpose of this study was to access the micro mobility and stresses developed in a cemented hip replacement. An in vitro cemented Lubinus SPII stem replacement was performed in synthetic femurs and sectioned. Section 5 was analyzed after fatigue test and was observed to be the most critical in crack incidence. A CAD model of this section was built considering bone and cement boundary geometry. The finite element model was built and the influences of different interface conditions of the cement interfaces (bone and stem) were analyzed. The interface stiffness associated with stresses for interface failure was used to simulate different surface roughness and time after surgery. The surface roughness associated with the interface strength did not present significant influence relatively to cement interface stresses and micro mobility of the stem. The type if interface changes the stress and strain distribution of bone and the most severe factor is friction at the cement/bone interfaces. The cement/bone interface debonding increase the bone strains and suggests pain.

1 INTRODUCTION

Cemented hip prostheses have produced excellent clinical results and THR is at that moment one of the most successful surgical techniques, with good success rates [1]. The success rate of all implants has increased and for 10 years is of the order of 93.5% (1991-96) [2]. The Swedish orthopedic register describes an increase number of revisions and present 13.6% of all revisions. For this reason it is important to reduce revisions and understand why these happen. An important reason for implant revision is aseptic loosening and biomechanical factors. The mechanism of aseptic loosening is not completely understood and is mutifactorial according to some studies [3]. Mechanical factors are fundamental to trigger the aseptic loosening process of a cemented implant. Other factors are type of cement and surface finish that have been correlated with long term failure scenarios. The fatigue failure of the cement mantle has been identified as a possible loosening mechanism. In other studies, failure has been associated with the thickness of the cement mantle [4, 5]. Other indicator associated to the long term failure is the micro motion and migration of the stem. Some experimental in vitro studies have correlated these parameters with long term results. The fatigue process results from normally activities with repeated loads [6, 7]. In some studies, micro motions between interfaces have been associated with the failure process with the formation of debris in the interface, promoting the creation of fibrous tissue [8, 9]. Aseptic loosing is one of the most important failure scenarios, being the cement damage accumulation an indicator of potential mechanical failure.

The purpose of this study was to access the micro mobility and stresses developed in Lubinus SPII stem hip replacement. A finite element model was built based on sections obtained image capture (section 5). Four stems of a Lubinus SPII were implanted in synthetic femurs and loaded and tested in fatigue, using a moment from stair climbing loading profile during one million cycles. The mechanical properties of bone were the same of those used by other studies, and different friction coefficients and interface stiffness were simulated. This study gives insight to identify the influence of the interface stiffness on the cement stress and bone strains.

2 MATERIAL AND METHODS

2.1 Experimental model

The experimental procedure involved the replacement of synthetic femurs with Lubinus SPII stems. Figure 1 presents the stem geometry of the Lubinus SPII system. The Lubinus SPII hip is one of the most used in Sweden, accounting for almost 57% of the total number of prostheses applied in 2007 [2]. It appeared in the market recently and at the moment presents a success rate of 98% for 10 years in 25.620 cases and 87.2% for 16 years [1].

Figure 1: Lubinus SPII cemented stem.

Annaratone et al. [10] refer that the Lubinus SPII is "one of the best" in the market of cemented stems. In vitro cemented hip replacements were performed using synthetic femurs (large left femur, mod. 3403, Pacific Research Labs, Vashon Island, WA, USA). These femurs have proven to be suitable for static and dynamic experimental simulations [11, 12]. The stems were implanted into synthetic femurs by an experienced orthopedic surgeon with adequate surgical instrumentation and CMW3® cement bone was used.

The loading applied allowed the combination of bending moments, a torsion moment and axial loads. The load was applied with the femur positioned at 11º on frontal plane and 9º on the saggital plane, as specified by the ISO 7206 standard and used in other studies [6, 7].

The load applied was of sinusoidal type, which replicates loads induced by stair climbing and is considered the most sever [12]. The loading is approximately representative of 24 years of stair climbing, considering a hip reaction force of 2250 N for the intact femur. A maximum load and a minimum load resulted from the loading (table 1) configuration used after the adequate correction relatively to the intact femur head center was changed (intact and implanted femur) [13]. The femur was fixed at the distal condyles and one million load cycles at a frequency of 2Hz were applied through a pneumatic device.

Table 1: Characteristics of the Lubinus SPII stem.

	Lubinus SPII
Size	LII (left)
Material	CrCo alloy
Head size	28 mm
Surface roughness (Ra)	0.98 (±0.02)
Maximum load (N)	1732 (±88))
Surface roughness (Ra)	681 (±94)

At the end of the fatigue testing procedure all specimens were transversely cutted in 11 sections. The first section was considered tangent to the collar. The cutting process was made using a high speed disc. The specimens were inspected with a non destructive technique using dye penetrating liquid [8].

2.2 Numerical model

The finite element model was built based on the in vitro femur replacements. The CAD model was generated as illustrated in figure 3. The image was scanned with a 10 µm per pixel and boundary conditions defined between cortical bone, cancellous bone, cement and stem. The load applied was a moment with a 0.79 N.m as used in other studies and referred to be the most critical [12].

The thickness of the cement layer was non uniform, which is biomechanically relevant [14, 15]. The numerical model was constructed by hexagonal elements of 8 nodes.

Table 2: Material properties of the numerical model.

	Young Modulus (MPa)	Poisson Coefficient
Cortical bone	17400	0.3
Cancel bone	280	0.28
Cement	3000	0.3
CrCo	210000	0.3

The finite element mesh was performed using Hyperworks V10, (Altair) software and solver MSc Marc used. The material properties were assumed to be linear elastic and properties supplied by Sawbones® (table 2) were considered. The FEM model contained 35695 elements and 46825 nodes (figure 3). Te surface roughness measured was 0.98 µm.

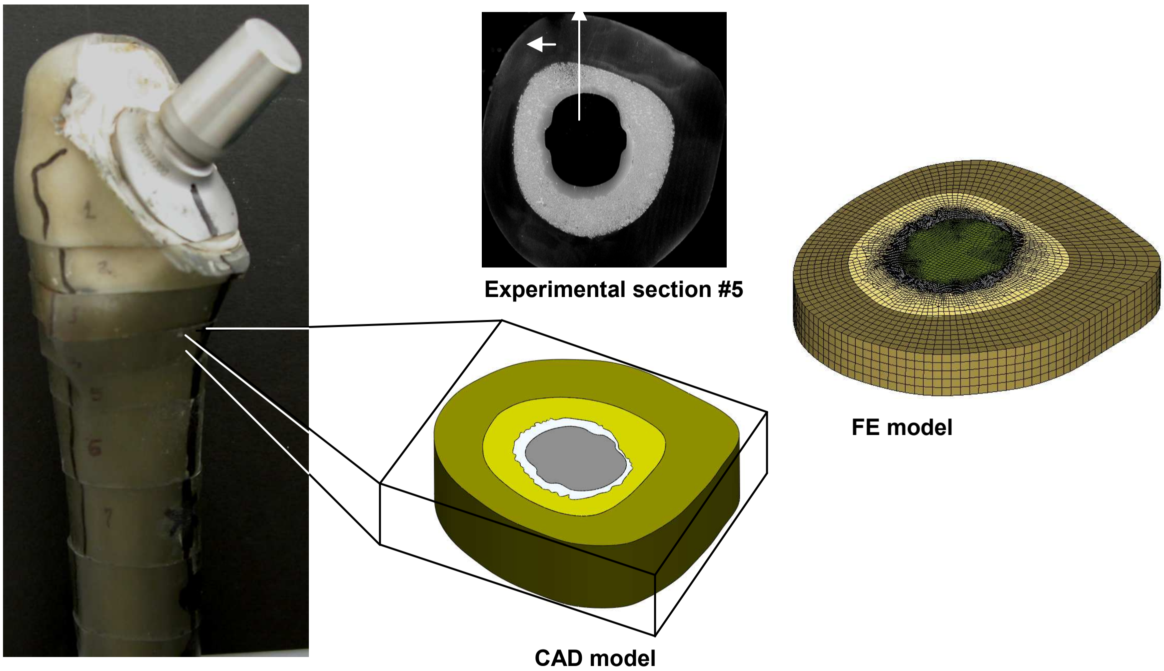

Figure 3: Procedure to build the CAD and FE model

The surface roughness is important to define the interface stiffness between stem-cement bone-cement interfaces. Zelle et al [16] discuss the importance of the normal stress in the interface which depends on the surface roughness. For the stem studied the authors presented a maximum normal stress of 0.50 MPa. Jui-Pin and Fu-Chai [17] defined the interface strength as 0.49 to 9.95 MPa for surface roughness from 0.89 to 2.76 respectively.

The different models analyzed are defined in table 3. Different boundary conditions at each interface were simulated to consider the influence of time after surgery. . The interface with glue represents a bonded interface without separation. Model #2 and #3 presents possible stem-cement interface separation. Model #4 simulates contact conditions of both interfaces in the last step of femur replacement. The contact control was defined as Coulomb friction with a glue condition in the first step. . The two interface stiffness was simulated considering 0.5 MPa and 6.7 MPa until interface separation [18, 19].

Table 3: FEM boundary conditions.

	Interface stem/cement (μ/ MPa)	Interface bone/cement (μ/ MPa)
Model #1	glued	glued
Model #2	0.5/ 0.5	glued
Model #3	0.5/ 6.7	glued
Model #4	0.15/ 0	0.5/ 0

3 RESULTS

3.1 Experimental results

After the in vitro fatigue tests the femur replacements were sectioned and analyzed. Figure 4 presents the cracks distribution in the cement mantle. More cracks in the cement-bone interface of section #5 were observed. At medial aspect the stem cement interface presented more cracks in section #5. The most critical medial and posterior aspect was observed for section #5. For the lateral aspect, section #4 was the most critical in cement bone interface.

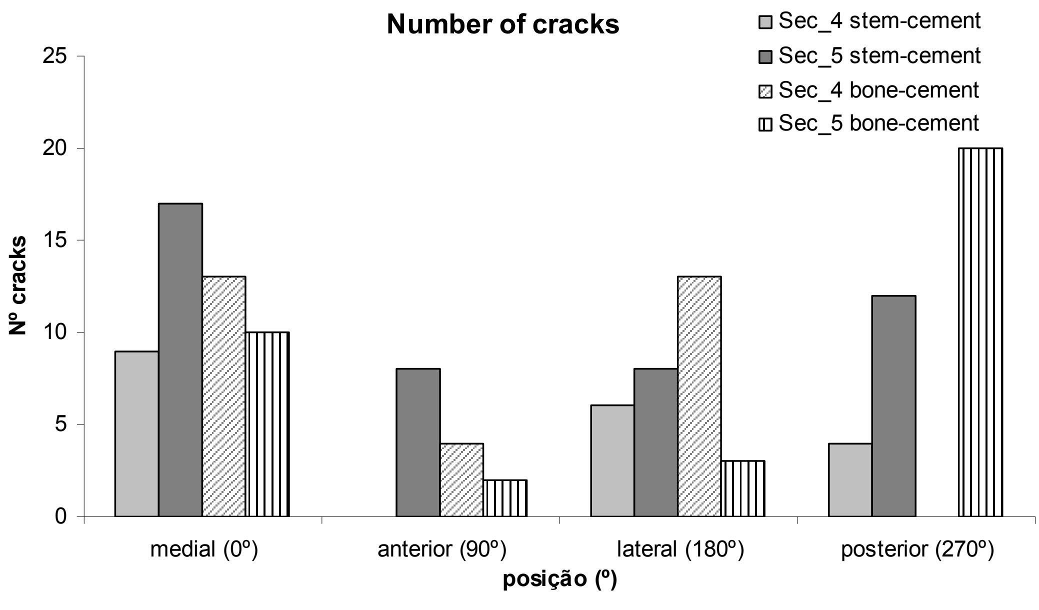

Figure 4: Number of cracks for section #4 and #5.

4.2 Numerical results

The displacement distributions in the three models are presented in figure 5 and 6. Model

#2 and #3 present similar results, showing small influence of the interface stiffness from 0.5 MPa (#2) to 6.7 MPa (#3). The results presents a similar behavior (relative displacement), only in the posterior aspect presents the same displacement.

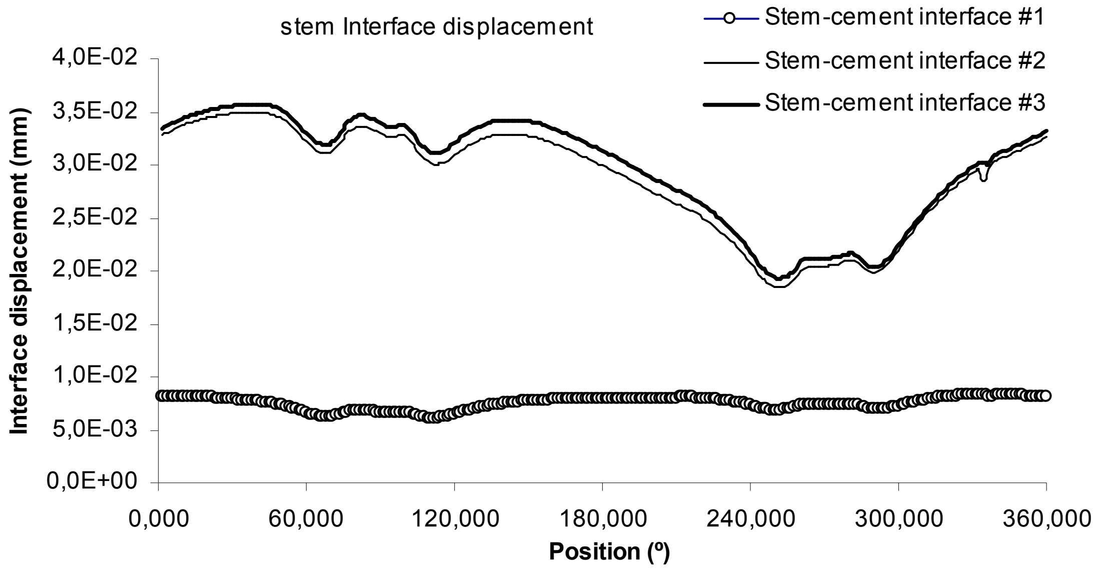

Figure 5: Interface dispacements for different interface conditions.

The bonded interfaces (#1) presented the most uniform distribution. The interface stem-cement with contact presented the most concentrated distribution in the anterior e posterior aspects. These models show the variation of displacements with durability of time of replacement.

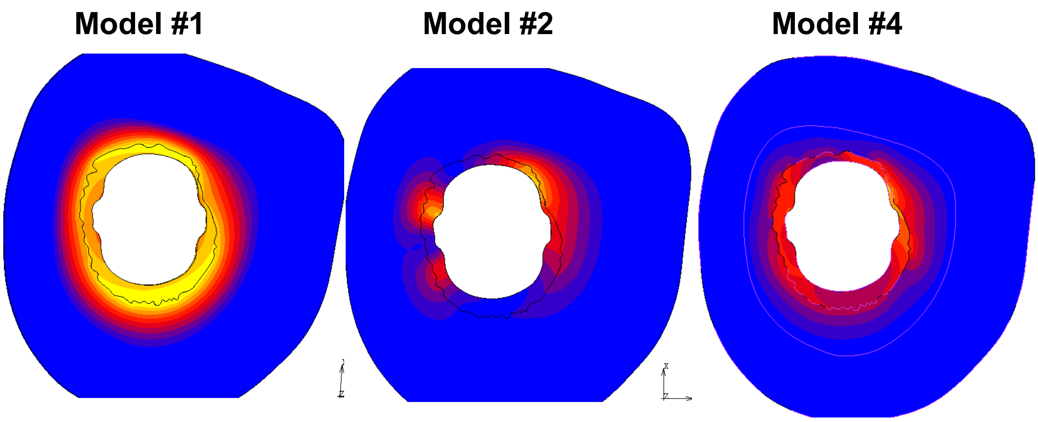

Figure 6: Interface dispacements for different interface conditions.

The micro mobility in the stem-cement interface for models #2 and #3 are very similar (figure 7). This could be explained by the high moment applied and contact conditions do not influence the micro mobility. For the two interfaces simulating debonding (#4) we observed that the micro mobility in stem-cement interface decrease.

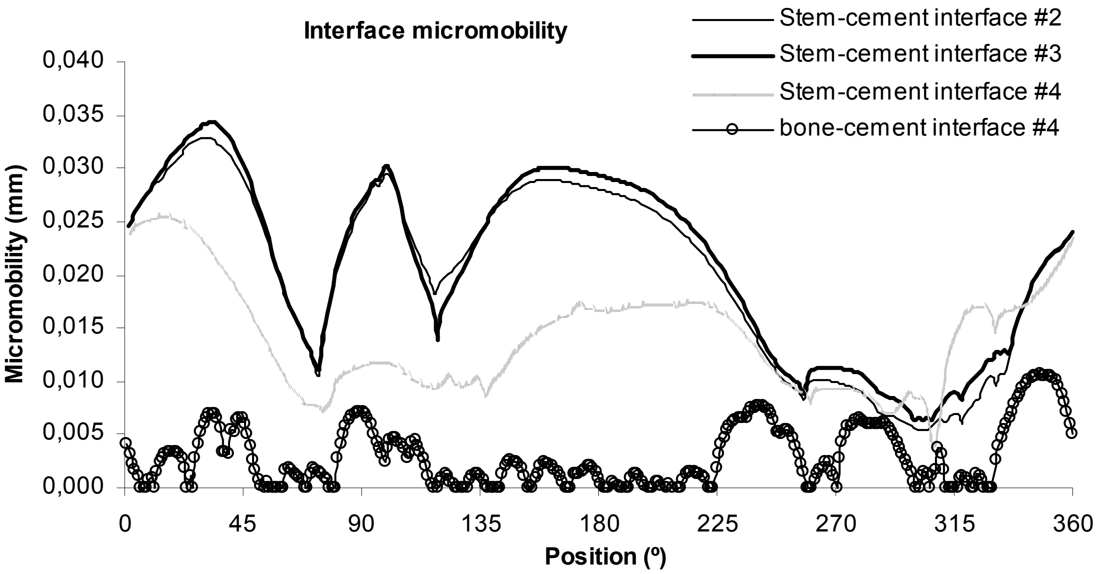

Figure 7: Maximum and minimum principal bone strains at the cement-bone interface.

The cement stresses were critical in the medial/posterior and lateral/anterior sides of the femur. The most critical conditions were observed for models #2 and #3 in some contact points. The bonded interface in model #1 presented the lowest principal stress. The maximum principal stress in the cement/stem interface was 39.6 MPa (#3), and in the interface cement/bone was 22.7 MPa (#3).

The maximum and minimum principal strains in the bone interface were most critical in model #4, with a maximum strain of 20000 µstrain. The maximum principal strain for model #1 with bonded interfaces was 4700 µstrain. Figure 8 presents the strain distribution for the three models. The anterior and posterior aspects presented the highest principal strain values. Not taking into account the posterior aspect, model #2 and #3 presented similar strain distributions. These values are in agreement of experimental crack results in posterior aspect at cement-bone interface.

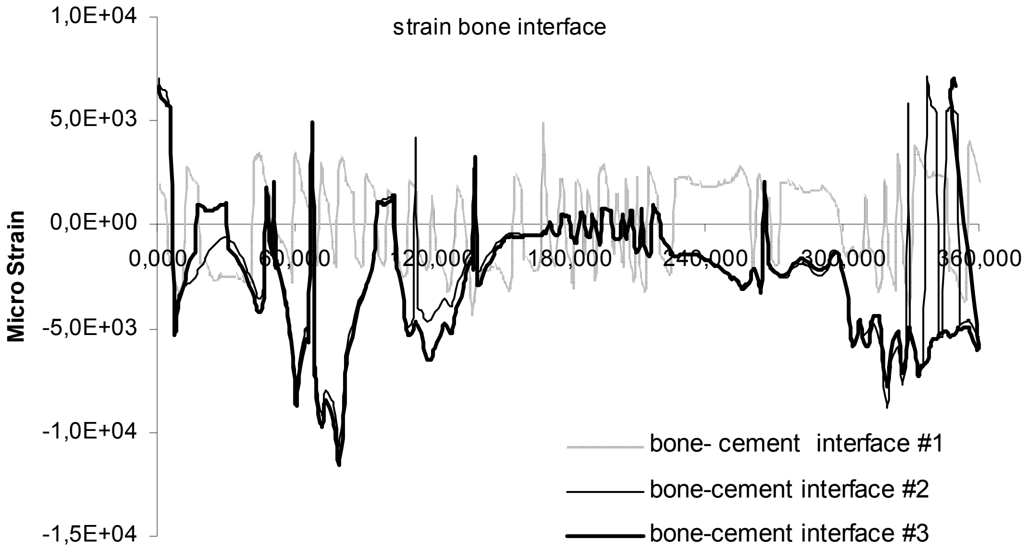

Figure 8: Maximum and minimum principal bone strains at the cement-bone interface.

CONCLUSIONS

- The loading configuration is an important issue to obtain representative results. The moment applied represents loads due to stair climbing and is the most severe;
- The interface stiffness does not present significant different results in terms of mobility, but changes the maximum stress in the cement interface up to 30%;
- The analysis of the sections of an implanted femur can be correlated with experimental results;
- The medial and lateral aspects present the most critical bone stresses;
- The micro mobility in the stem cement interface is critical in the first pos implanted period, because some debonding in the cement-bone interface occur the micro mobility decrease in stem interface. But debonding in bone interface increase the bone strain distribution and promote the pain.

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