VIBRATIONAL BEHAVIOUR ANALYSIS OF A FEMUR-THA STEM STRUCTURE USING FEA

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Abstract: *Vibrational analysis as a tool for monitoring and evaluating the early loosening of THA holds promise to improve the sensibility and reliability of diagnostic. This tool is based on the changes appearing in bone-implant stiffness.*
Key words: *FEM - finite element modelling, THA - total hip arthroplasty, THR - total hip replacement, vibration analysis, FRF - frequency response function, cementless, prosthesis stem*

1. INTRODUCTION

The finite element (FE) simulations are viable alternatives for prediction of primary fixation of THR and loosening evolution. Vibration analysis as a method based on the mechanical properties of the bone/implant system holds promise for improving diagnostic sensitivity of implants loosening. This technique has been successfully used to determine bone mechanical properties and monitoring the fracture healing [5]. The first documentation using this technique on patients compared vibration-testing results with radiographs in 23 patients with total hip replacements experiencing hip pain without X-ray evidence of loosening [2].

In the past decade, several studies were performed using resonance frequency for the quantitative assessment of fracture healing in long bone [3], to detect the change in bending rigidity early in the treatment of tibia fracture, but problems with soft tissue and interpretation of results suggest that the technique has a limited clinical usefulness.

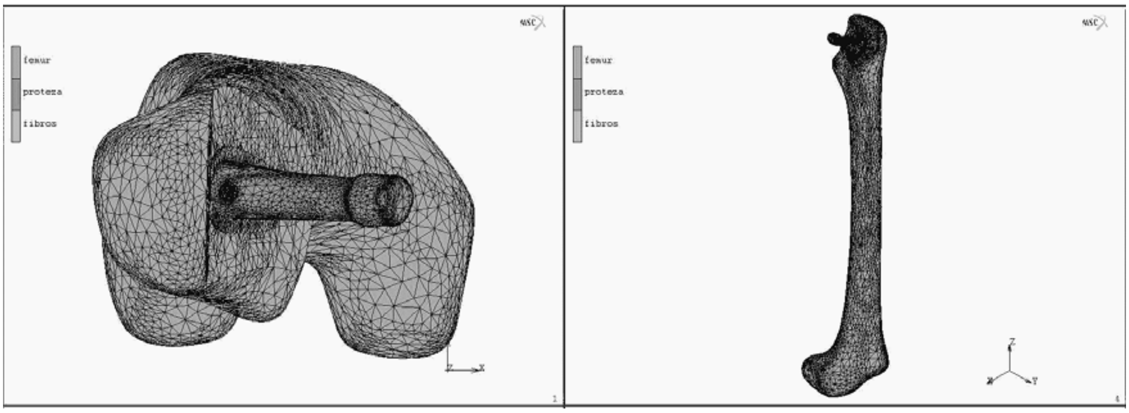
The successful outcome of total hip arthroplasty (THA) requires not only good initial fixation, but that this fixation is maintained in the long term. The initial fixation is obtained by interlock of the polymethymethacrylate (PMMA) into the surrounding cancellous bone; this bond may be partially or completely lost through the development of a fibrous tissue layer at the bone–cement interface, and subsequently lead to aseptic loosening. The formation of the tissue is associated with many causes, such as wear debris, bone necrosis, bone resorption due to stress shielding, implant motion. While long-term THA failure may be due to a combination of these causes, the transfer of loads from the stem to the bone is dependent on the interface conditions.

Finite element methods have been previously employed to study the mechanical consequences of a fibrous tissue layer on the load transfer characteristics of both cemented and cementless joint replacements. Such computer simulations have been recommended [4, 6], where it was previously found to be difficult to model experimentally the condition of a fibrous tissue layer. Brown et al. (1988) [1] in a three-dimensional finite element analysis of a cemented replacement with a compliant layer reported a general increase in stress levels for the proximo–lateral cancellous bone, with a decrease in the proximo–medial cancellous bone. There were also increases in the bending moment transmitted through the prosthesis and the cement mantle, especially around the mid-stem. Weinans et al. (1990) [7], using a two-dimensional FE model, investigated various soft tissue models (linear and non-linear). They observed that a non-linear model is important where relative displacements between implant and bone are to be evaluated.

Nevertheless, in the orthopaedic area there seems to be ongoing interest in the application of this rather simple testing technique. Since in vivo signals reflect information arising from a variety of sources such as articular cartilage, ligaments, tendons, muscle, joint capsules, bone, and prosthesis a computational virtual study is the only means to isolate and define responses capable of arising from a single factor; for instance, the failure of a femoral stem and cement interface.

2. THE FE ANALYSIS FOR CEMENTLESS FEMUR-ABG PROSTHESIS SYSTEM

The model was made starting from CT images obtained in laboratory and using two computational programs from Materialise Software package, Mimics 8.0 and Magics 9.0, for both parts the assembly, the femur and the ABG - prosthesis. These two parts were assembled; the prosthesis was fixed in the proper position inside the femur. After this, the cutting procedure has been applied to the femur in a surgical way and every part of the model has been separately remeshed. Then, the model was exported in <STL/ACSII> format in Patran where surface mesh was checked for avoiding errors. Modal analysis was made using MENTAT, a computational program of the MARC system.



This model was modified for simulation of increasing stages of loosening by application of a layer of fibrous tissue at the bone-prosthesis interface, simulating an instable interface between bone and prosthesis. The fibrous tissue thickness was between 1-3 tetrahedral elements.

The layer of fibrous tissue was modelled in MENTAT in three parts (proximal - Gruen zones 1 and 7, medial - Gruen zones 2 and 6, and distal - Gruen zones 3, 4 and 5). For each combination of those three parts of fibrous tissue the corresponding stages of loosening were simulated, starting from a good fixation (meaning that no fibrous tissue is present at the interface bone-prosthesis) and ending with the stage of severe loosening (meaning that the entire layer from the interface have properties of fibrous tissue, corresponding to a very instable interface). After the fibrous tissue was modelled, the rest of the assembly, meaning the femur and the prosthesis, was imported in MENTAT as <dat> files, one by one.

Material	Young modulus (MPa)	Density (kg/mm^3)	Poisson ratio
Femur (bone)	17000	1.70e-009	0.3
Fibrous tissue	2	0.42e-009	0.17
ABG (Co-Cr)	250000	2.50e-009	0.3

Table 1: Materials properties for the femur-prosthesis assembly

The material properties (table 1) needed for starting a vibrational modal analysis are Young modulus, Poisson ratio, and the mass density, and were taken from literature. For the fibrous tissue, the material properties are those of the callus tissue at four weeks after the fracture happened. This layer is presumed to induce the loosening process by stopping the osseointegration process. The cancellous bone forms this fibrous tissue layer in the healing phase, after the surgical intervention.

3. RESULTS

In FE work, the increase in resonance frequency shift with increasing complexity of the mode shape of the femur/prosthesis system was observed. Also, the most sensitive frequency band for observing defects in the femur/prosthesis connection is above 2500 Hz.

In the table 2 are presented the frequencies of the first seventh bending modes of a femur-prosthesis assembly, in four different stages of fixation: well fixed corresponding to the absence of fibrous tissue, early loosening corresponding to the proximal presence of fibrous tissue (Gruen zones 1 and 7), advanced loosening corresponding to the presence of fibrous tissue in proximal and also in medial position (Gruen zones 1, 2, 6 and 7) and severe loosening corresponding to the situation that at the interface, all the prosthesis stem is surrounded by fibrous tissue.

incr.	**A no fibrous**	**B proximal**	**C proximal-medial**	**D all fibrous**	**bending mode**	***A-B***	***A-C***	***A-D***
1	312.2	312.2	312.1	312.1	**1**	0	0.1	0.1
4	984.2	984.2	983.9	983.8	**2**	0	0.3	0.4
6	2010	2009	2009	2009	**3**	1	1	1
9	3401	3397	3396	3395	**4**	4	5	6
12	4845	4826	4823	4820	**5**	19	22	25
15	6252	6233	6221	6215	**6**	19	31	37
19	7715	7693	7690	7686	**7**	22	25	29

Table 2: The frequencies of the first seven bending modes of femur - ABG prosthesis assembly

In the figure are presented the differences between resonance frequencies of the first seven bending modes (only in frontal plane).

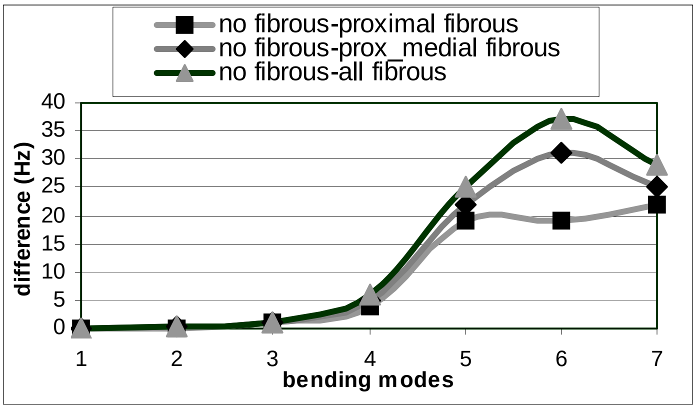


Figure 2: Shifts of the resonance frequencies of the Sawbone/ABG system between different stages of fixation, plotted for increasing complexity of the mode shapes

The resonance frequencies of the mode shapes of an intact Sawbone are affected by the insertion of prosthesis. The shifts in resonance frequency increase with increasing complexity of the mode shape. In the frequency range of the lower vibration modes (up to the second bending mode, below 1000 Hz) the change in resonance frequency is below 1 Hz. The first bending mode of the femur is hardly affected by the presence of the prosthesis. The effect of the prosthesis on this mode shape is mainly to increase the mass of the system, but the stiffness is hardly affected since the deformation of the proximal part of the femur is very small in the single bending mode. Only severe loosening will be detectable when the first bending mode of the femur is excited: in this case the connection between the prosthesis and the bone is too weak to allow the prosthesis to participate in the deformation of the femur and a rattling phenomenon will occur. The system then behaves in a strongly non-linear way and the interpretation of the shifts in FRF peaks is no longer valid.

4. CONCLUSIONS AND FUTURE WORK

Aseptic loosening is the most important cause of late failure in THR. The kinetics of loosening is different for cemented and cementless implants. In the case of cementless implants only a fraction of the stem surface comes in contact with fully mineralized bone. The rest of the implant surface contacts different types of tissue, all with very limited load-bearing capabilities. The initial soft tissue interface may appear at the operation time or later because of multiple causes. Depending on its extension and location, the presence of soft tissue interface may reduce the stability of the stem allowing large bone implant micro-movements to be induced by daily physiological loads. Larger micro-movements increase the extensions of the soft tissue interface, which further increases the micromotion, in a vicious loop that drives to the macroscopic loosening of the cementless stem. A more appropriate issue is the minimum extension of the soft tissue interface required to initiate the failure scenario described above. The future step is to build a model of a cementless femur-prosthesis assembly and to investigate another variable that can influence the loosening process, like the thickness of the fibrous tissue, location and extension of the non-bearing regions and of osseointegration regions. Also, a model for the femur and custom-made prosthesis assembly is required. A better understanding of the loosening process and of the fibrous tissue appearance is necessary for creating a valid model, similar with the reality.

Considering the fact that the femur doesn't have the same properties in each part, a future model should take into consideration the three parts of the femur (cortical, trabecular and marrow channel).

In real situations, the implant is submitted to many different loading conditions, which represent different daily activities. The formulation of the mechanical stimuli is of central interest for any remodeling models. Mechanically stimulated osteoblasts have completely different genetic expression than non-stimulated osteoblasts, with bone resorbing factors being upregulated. This indicates that the deformation of the cells may lead to bone resorbtion and to the formation of a fibrous tissue. Another important point is that the cells in contact with the prosthesis at the bone surface are not able to support all the deformation. A cell with a typical size of 5 μm is not able to support the shear strain induced by a micromotion of 150μm. The strain would be too high and would kill the cell. Furthermore, a model useful for a better understanding of the processes of failure of orthopedic implants is required. In the future, the parameters of the model should be determined more precisely and then applied to study more realistic orthopedic situations. A typical application of the model will be for designing new implants that minimize the formation of fibrous tissue.

5. REFERENCES

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