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Study of factors affecting taper joint failures in modular hip implant using finite element modelling

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1. Introduction

The replacement of the natural hip with artificial components is a well established procedure in orthopaedic medicine to alleviate pain from the diseased joint. The total hip replacement (THR) consists of a femoral component or stem and an acetabular cup. These are made as either one piece or modular designs. Modular femoral components have become popular among surgeons because neck length and offset can be adjusted intra-operatively, providing increased versatility without any need for a large inventory. In addition, modular heads allow for mixed alloy systems such as the combination of a titanium alloy stem with a cobalt-chromium or ceramic head.

Apart from the neck/head connection, there are femoral stem designs with extra areas of modularity. Examples of these prostheses are the S-ROM (Joint Medical Products, Stanford, Connecticut), the Infinity (Dow Corning Wright, Memphis, Tennessee), the RHMS (Smith-Nephew Richards, Memphis, Tennessee), and PROFEMUR Hip System (Wright Cremascoli Ortho SA, France).

A major drawback for all modular orthopaedic devices is that each modular component interface becomes a potential site for corrosion, wear, fretting and fatigue of mating surfaces (Collier et al., 1992; Manley & Serekian 1994; Hallab & Jacobs, 2003; Goldberg & Gilbert, 2003; Hallab et al., 2004; Gilbert et al., 2009; Rodrigues et al., 2009). The products of these interface processes are believed to cause tissue reactions that lead to implant loosening and subsequent failure of the arthroplasty (Amstutz et al., 1992; Harris, 1994; Kraft et al., 2001; Goldberg et al., 2002).

Thus, to make implant designs successful in clinical applications, these concerns need to be adequately addressed during the design stage of the prosthesis in conjunction with the implant strength and integrity of the implant-host bone system.

Types of failures often encountered in modular hip implants are dissociation, corrosion, wear; fretting and fatigue of mating metal surfaces (Goldberg et al., 2002, Sporer et al., 2006; Rodrigues et al., 2009). There are varieties of design and material factors that may influence the failure of specific components. The most significant of these is fretting because is almost impossible to prevent and in many cases may lead to other form of failures.

Fretting is defined as a wear mechanism that occurs at low amplitude, oscillating, sliding movement between two mechanically joined parts under load. There are varying descriptions of the magnitude of the motion associated with fretting, but it is generally defined as ranging from few to 50 μm (Mutoh, 1995). Given the magnitude of loading, all modular junctions of total hip prostheses can be susceptible to fretting wear. Other failures associated with fretting are fretting corrosion and fretting fatigue.

Major concerns of fretting relate to the modular junctions with metal to metal contact surfaces. Even though fretting and associated problems of orthopaedic implants were recognized since late in 1960s and in 1970s (Cohen & Lunderbaum, 1968; Gruen & Amstutz, 1975), the concern is ever increasing because orthopaedic surgeons and implant companies are interested in implants with more areas of modularity and actually produce different designs such as those mentioned above. Because these implants pose more mechanical joints, the possibility of fretting damage increases markedly -- especially in titanium alloy materials. Notable examples of fretting damage are those described by Hallab & Jacobs (2003) and Bobyn et al. (1993).

Among the factors that promote fretting are the design characteristics of modular hip implants such as neck diameter, neck length and fabrication tolerances of the joined parts are particularly important. Looser manufacturing tolerances lead to smaller contact area, higher stress concentration, and higher interfacial motion. All are key factors in developing fretting wear. (Goldberg & Gilbert, 2003; Fessler & Fricker, 1989). Similar studies using FEA have been reported (Shareef & Levine, 1996; Kurtz et al., 2001).

In this study, a generic modular-neck femoral stem design was assessed for relative motion at the Morse taper junction. Non-linear finite element analysis (FEA) was used. The research was carried out with the objective to study the effect of different fits of the Morse cone and surface conditions on the extent of the relative micromotion at the mating taper interfaces.

2. Materials and Method

A three dimensional (3-D) model of a generic hip stem system was created and analysed using a commercially available finite element (FE) software package: ANSYS software. The model was developed to simulate a modular hip implant system which consisted of a neck, part of the stem, and the interface between the two. Node to node contact elements were used to model the interface between two parts. These are non-linear elements; therefore, the finite element problem required a non-linear analysis. The nonlinearity of the model was based on the contact aspect only. The model was made to represent a hip stem of simple shape that could be manufactured in a good quality machine shop.

2.1 Finite element model generation

The outline of the stem was developed using keypoints and lines based on the dimensions of the PCA No. 5 hip prosthesis, but was simplified by leaving out the rounded corners and slight curvatures. As these simplified surfaces were relatively far away from the modular interface, the simplification would not affect the results. This also facilitated model changes during the analysis.

The model was meshed using all hexahedral (brick) volume structural element (SOLID45). A fine mesh was used around the tapered hole at the proximal end of the stem and a coarser mesh was used for regions distant from the hole and for the distal section of the stem.

Constraint equations were used to tie together the finer proximal and coarser distal mesh regions of the stem as indicated in Figure 1 that shows the final FE mesh used in analyses. The techniques that were employed to determine the extent of mesh refinement in the present work was to perform initial analysis with an assumed “reasonable” mesh. Then, the problem was re-analysed using finer mesh in critical regions, and the two solutions were compared. This process was repeated until the optimum mesh was obtained. A very fine mesh would have improved the results further but could take longer to run than it was feasible. The choice of the optimum mesh was based on both accuracy and solution run time. Further validation of results of this model was performed using experimental stress analysis as indicated in Figure 2.

Fig. 1. (a) Meshed stem and neck. Only one half of the stem was modeled using symmetry constraints along the sagittal plane that divides the stem into two halves. (b) Model alignment with respect to load. This arrangement was used for both FEA and experimental stress analysis.

2.2 Neck-stem contact definition

To simulate the neck-stem interface, the 3D node-to-node contact elements (CONTAC52) were used. These elements were used to represent two surfaces which could maintain or break physical contact and could slide relative to each other.

For successful contact problem, several contact element properties and options needed to be carefully selected. These properties include geometric input data, normal stiffness (KN),

sticking stiffness (KS) and coefficient of friction. Geometric input is controlled by the mesh of contacting bodies. User defined options include specifying the type of friction model (elastic or rigid Coulomb friction) and the contact time prediction control. In the determination of KN and KS, guidelines in the ANSYS reference manuals were followed so that the risk of numerical difficulties or slow convergence during the solution phase of the analysis could be minimized. Normal stiffness, KN, was determined based upon the stiffness of the surfaces in contact. KN had to be large enough that it could, reasonably, restrain the model from over-penetration, yet it had to be not so large that it could cause ill-conditioning of the stiffness matrix.

Fig. 2. Strain gauges installation. Two 3-element 45° staked rosettes and two single elements gauges were used. Numbers 1 to 8 represent gauge elements. Gauge elements 1, 4, 7, and 8 are aligned along the direction of principal stresses during the assemble loading.

Similarly, a suitable value for KS which would avoid numerical instability or excessive run times had to be determined. The default setting in ANSYS is $KS = KN$. Lower values reduce the run time. Value used were $KN = 100 \cdot E$ and $KS = 0.01 \cdot KN$; after being tried and found to reduce the run time without affecting the stress and micromotion results. Elastic Coulomb friction was used because practical fretting couples exhibit dual micromotion regimes, normally referred to as elastic regime and gross slip regime (Zhou and Vicent, 1995; Mohrbacher et al., 1995). The values for the coefficient friction (μ) used in various analyses models were based on experimental measurements reported by Fessler and Fricker (1989) and the data from Budinski (1991).

2.3 Angular mismatch

Three model cases were developed to simulate three possible scenarios which could be realised when assembling stem and neck. The three cases defined in Figure 3 are:

- i) No or zero angular mismatch
- ii) Positive angular mismatch
- iii) Negative angular mismatch

Tolerances of -2 to +2 minutes for the male taper were used, thus producing a maximum angular interference and clearance of 2 minutes. These tolerances are within the limits obtained by manufacturers. Design specifications for a particular manufacturer were $6^{\circ} \pm 2$ min. for female tapers and $5^{\circ} 58' \pm 1$ min. for male tapers. (Naesguthe, 1997).

Fig. 3. Definition of mismatch cases used in FE analysis. Angular mismatch between the neck and the stem was achieved by changing cone angle of the neck by two minutes.

2.4 Boundary conditions and loading

Boundary conditions for finite element analysis and load application were set as in the experiment to determine the endurance properties of femoral stem of hip prostheses according to the ISO standard (ISO 7206-4: 1989 (E)). The angle between the load line and anatomical axis of the femur was set to be 10° when viewed perpendicular to the plane that includes the stem and the neck (Figure 1). Boundary conditions were established such that the finite element nodes at the bottom of the stem were constrained in all degrees of freedom (d.o.f). Only one half of the stem was modelled using symmetry constraints along the plane that divides the stem into two halves, (see Figure 1). The half-stem model ignored out-of-plane loads but was sufficient and was able to satisfy the purpose of the analysis.

Model loading was applied in six steps:

Step 1: A load was applied at the top and along the axis of the neck, as shown diagrammatically in Figure 1(b). This was termed an “assembly load”; it was applied to simulate the force a surgeon would use when inserting the neck into the stem hole during the operation.

Step 2: The assembly load was removed.

Step 3: A load was applied at the same point as in Step 1, but at an angle of 10° from longitudinal axis of the stem, as shown in Figure 1(b). This was termed a “functional load”; it was applied to simulate the force that would be applied to the implant during walking.

Step 4: The functional load was removed.

Step 5: The functional load was re-applied.

Step 6: The functional load was removed.

Five ‘Load schemes’ code-named ‘First’, ‘Second’, ‘Third’, ‘Fourth’ and ‘Fifth’ were used in various analyses. The magnitudes of the assembly and functional loads used in these ‘Load schemes’ are shown in Table I. The assembly loads were used in FE models to represent three possible cases of moderate, high and no tapping loads, respectively. The function loads were used to represent possible physiological loads as reported in the literature (Bergmann, et al. 1993; Viceconti et al., 1996)

Designation of the loading scheme	Magnitude of the Assembly load (N)	Magnitude of the Functional load (N)
First	3114	5500 (7.5 x BW*)
Second	5500	3114 (4 x BW)
Third	5500	2000 (3 x BW)
Fourth	3114	2000
Fifth	0	2000

*BW = Body weight of average person of 75 kg.

Table 1. Assembly and Functional loads used in FE analysis

3. Results and Discussion

3.1 Validation Results

Experimental stress analysis was performed for the purpose of validating FE results. It was sufficient to measure stresses only instead of both stress and micromotion because, theoretically, in finite element analysis, stresses are calculated from the primary displacement values. For instance, the press-fit stresses resulting from the application of assembly load were determined from the amount of interference at the neck-stem interface which in turn is determined from how much the neck moved relative to the stem. Therefore, if the stresses were proved valid, so would be the displacements from which the stresses were calculated. In this case, no relative micromotion was measured experimentally. Graphs that compare experimental with numerical results of the FE model show a good agreement on the two sets of results. These are shown in Figure 4 and 5 for stress prediction during

assembly and functional loading, respectively. The differences between stress values obtained from strain gauges and stress values predicted by FE model were within 10% which is considered acceptable.

Fig. 4. Experimental and numerical stress values under asseby loading showing readings of gauges 1, 4, 7 and 8. The induced stresses at these gauge locations were mainly unidirectional (along x- direction).

3.2 Micromotion results in three angular mismatch cases

Figure 6 shows the relative micromotion of the First load scheme at a medio-proximal point (B) of the neck-stem interface for three mismatch cases. The selected point was characterised with the highest micromotion during the load cycle as shown in Figure 7. It was also a point of highest stress during the functional load. In this regard, the selected point was critical in terms of potential surface failure due to fretting and other surface degradation mechanisms.

At any particular instant during the loading of the implant system, the relative micromotion at the stem-neck interface was a result of a free body displacement of the neck or an elastic deformation of the neck and the stem, or the combination of the two. In all cases, application of the assembly load caused a non-recoverable relative micromotion between the neck and stem. This micromotion is termed non-recoverable because upon the removal of the assembly load, the neck did not move back to its initial position.

Fig. 5. Experimental and numerical stress values under functional loading. Since the stress state was multi-axial, the experimental results plotted are those from the rosettes only.

The interfacial micromotion observed during the application of the functional load was mainly due to elastic deformation of the parts. This observation can be justified from the fact that in models that had the assembly load of 5500 N, the micromotion was fully reversed upon the removal of the load (Figure 8).

Fig. 6. Relative micromotion at the neck-stem interface for three mismatch cases at proximal medial point which experience highest relative motion.

Fig. 7. Variation of micromotion along the contact area. Highest micromotion is observed at point B indicated in Figure 1 (a).

Fig. 8. Relative interface micromtion during two functional load cycles for three mismatch cases – Second load scheme (load steps 3 to 6).

For the First load scheme, the application of the functional load resulted in the magnitudes of relative micromotion of 65 µm, 70 µm and 80 µm in zero, positive and negative mismatch cases, respectively. When the functional load was removed, the micromotion in zero, positive and negative mismatch cases reversed to 52 µm, 32 µm and 65 µm. The subsequent

cycles of functional loading and unloading resulted in repeated patterns of relative micromotion. This is clearly demonstrated in Figure 8 for micromotion variation curves for the Second loading scheme in two functional load cycles.

3.3 Results of other analysis models

Two models with zero angular mismatches were used to study the effect of the coefficient of friction and the magnitude of assembly load on variation of relative micromotion at the neck/stem mating surfaces. Figure 9 and 10, respectively, show the effects of coefficient of friction and assembly load on the neck-stem interface micromotion. The change of coefficient of friction had the greater effect on the reversible micromotion that occurred during the functional load cycle as shown in Figure 10. The increase in coefficient of friction from $\mu = 0.2$ to $\mu = 0.5$ had reduced the reversible micromotion from 50 μm to about 30 μm . Also, the magnitude of the assembly load affected the amount of micromotion caused by the subsequent functional load. Low assembly load resulted in highest micromotion as shown in Figure 10.

Fig. 9. Effect of coefficient of friction on the interface micromotion under assembly and functional load (load steps 1 to 4).

3.4 Model behaviour during the assembly load

The results of this study show that the magnitudes of the relative one time micromotion and reversible micromotion due to functional load depended on factors such as the magnitude of assembly force, coefficient of friction and the amount of angular mismatch between the male and female tapers.

The assembly load is used to achieve initial stability of the modular connection. A high assembly load is desirable in producing enough taper lock to prevent the neck from having further rigid body movement during the functional loading. This finding is consistent with the experimental study by Mroczkowski et al. (2006).

An extreme case of high interference is when the conical parts are pre-assembled as a shrink fit. This may result in considerable decrease in interfacial relative micromotion. Published experimental and retrieval data have indicated the absence of fretting damage in such press-fit situations (Brown et al., 1995; Mroczkowski et al., 2006).

Modular surfaces in real components are more complicated than they could practically be represented in FE models. Surface imperfections in real components will cause local yielding which will reduce the amount of mismatch when modular parts are assembled (Naesguth, 1998). Material nonlinearity due to localized plastic deformation of surfaces was not included in the FE model. Therefore, the overall numerical micromotion predictions were likely to be higher than the values that would be obtained in real components.

Fig. 10. Effect of assembly load on micromotion due to functional load. Third, Fourth and Fifth load schemes (load steps 3 to 6).

3.5 Effects of coefficient of friction

Maximizing the value of μ is beneficial in reducing conditions that promote fretting and fretting fatigue. The coefficient of friction has significant importance in actual performance of modular junction. Since, the interface micromotion is influenced by the load applied to the connected components of the modular hip stem; high friction surface is desirable to minimize interfacial relative motion.

3.6 Effects of angular mismatch

Zero angular mismatch is not the best and yields more micromotion. In determining the optimum angular tolerances, the individual case has to be judged based on its loading arrangement. For the modular neck hip prosthesis used in the present work, a positive

mismatch is the optimum choice. The degree of mismatch should be limited to keep allowable stresses within the safe limits below the fatigue strength of the implant material.

Fatigue, fretting and corrosion: Model results have shown that application of a functional load results in relative interface micromotion which varies in magnitudes according to the nature of the modular connection, the surface properties, and the magnitude of both the assembly and functional load. The key parameter to fretting is slip amplitude which is defined as the peak-to-peak amplitude of the relative reversible micro-movement of the surfaces (Mohrbacher et al., 1995; Waterhouse, 1992).

In normal and fast walking, the load on the implant ranges from about 3 to 4 times body weight (Bergmann, et al., 1993). Even if the stress level on the component at these loads is lower than the failure values, the existence of relative micromotion at the modular interface may put the implant at a risk of fretting and fretting fatigue. Microscopic relative movement between mating surfaces of levels, as low as 0.125 μm or 3 μm , has been found sufficient to produce fretting debris (Mohrbacher et al., 1995; Waterhouse, 1992). The lowest slip range predicted in our models is higher than the above values. Fretting is therefore inevitable under predicted levels of reversible micromotion. Furthermore, in a surface micromotion characterised with sticking and sliding regimes, cyclic contact stresses can cause the formation of microcracks (Zhou & Vincent, 1997), which can lower the fatigue limit of the component by about 50% (Broszeit et al., 1985).

As shown in Figure 7 highest sliding micromotion occurs at medio-proximal location of the stem. This observation is consistent with findings reported by Viceconti et al. (1998). They observed repetitive parallel scars with an average length of 30-45 μm in specimens that were loaded at 300-3300 N. Our micromotion prediction at a functional load of 0-3114 N ranged between 13-41 μm .

Since the micromotion is inevitable, the only option available to minimize the fretting damage is to apply suitable fretting palliatives, as suggested by Beard (1988), that will reduce micromotion between mating surfaces. Even if the decrease of micromotion is apparently small, it can still have substantial effect in reducing fretting. Experimental data suggest that the specific wear rate (volume lost per unit load per unit sliding distance) varies as a function of the slip amplitude raised to the power of 2 to 4 (Beard, 1988).

4. Conclusion

A three dimensional, non-linear finite element model was used to analyse relative micromotion of the modular hip implant at the junction between the neck and the stem. Functional, design, surface and manufacturing features that can affect micromotion in the modular junction of hip implant were studied. From the results and discussions that followed, the conclusions are:

- A high assembly load reduces the magnitude of stress and micromotion fluctuations during ambulation, predicting lower fretting and fretting fatigue damage, hence, improved service life. Therefore, during operation, orthopaedic surgeons should aim at an assembly load of 6000 N or higher. The force to be used should be higher than the largest anticipated ambulatory load

- High friction at the modular interface with the coefficients of friction well above 0.5 up to complete binding between the surfaces is desirable in order to reduce the amount of relative micromotion at the modular mating surfaces.
- In the modular neck stem, a positive mismatch is the best. This means a cone angle of the neck 2 minutes above the female taper on the stem. Our model showed low relative interfacial micromotion in the stem-neck connection with positive angular interference. It is therefore assumed that fretting damage of the modular interfaces can be minimized by a proper control of manufacturing angular tolerances of the mating parts.

5. References

- Amstutz, HC.; Campbell, P.; Kossovsky, N. & Clarke, IC. (1992). Mechanism and clinical significance of wear debris induced osteolysis. *Clinical Orthopaedics and Related Research*, 276:7-18.
- Beard, J. (1988). The avoidance of fretting, *Materials and Design*, 9 (4), 1988, 220-227.
- Bergmann, G.; Graichen, F. & Rohlman, A. (1993) Hip joint loading during walking and running, measured in two patients. *Journal of Biomechanics*, 26, 969-990.
- Bobyn, J., Dujovne, A. Krygier, J. & Young, D. (1993). Surface analysis of the taper junctions of retrieved and in vitro tested modular hip prostheses. In: *Biological Materials and Mechanical Considerations of Joint Replacement*, B.F. Morrey (ed.), 287-301, Raven Press Ltd., New York.
- Broszeit, E.; Kloos, K. & Shweighofer, B. (1985). The fretting fatigue behaviour of titanium alloy Ti 6Al 4V. In: *Titanium: Science and Technology*, 21-71, Deutsche Gesellschaft für Metallkunde e.V, FRG.
- Brown, S.; Flemming, C.; Kawalec, J.; Placko, H.; Vassaux, C.; Merritt, K., Payer, J. & Kraay, M. (1995). Fretting corrosion accelerated crevice corrosion of modular hip tapers, *Journal of Applied Biomaterials*, 6, 19-26.
- Budinski, KC. (1991). Tribological properties of titanium alloys, *Wear*, 151, 203-217.
- Cohen, J. & Lunderbaum, B. (1968). Fretting corrosion in orthopaedic implants, *Clinical Orthopaedics and Related Research*, 61, 167-175.
- Collier, J.P.; Michael, D.E.; Mayor, B.; Jensen, R.E.; Surprenant, V.A., Surprenant, H.P., McNamara, J.L.; & Belec, L. (1992). Mechanism of failure of modular prosthesis. *Clinical Orthopaedics and Related Research*, 285, 129-139.
- Fessler, H. & Fricker, D. (1989). A study of stresses in alumina universal heads of femoral prostheses, *Proceedings of the Institution for Mechanical Engineers. Part H: Journal of Engineering in Medicine*, 203, 15-34.
- Gilbert, J.; Mehta, M. & Pinder, B. (2009). Fretting crevice corrosion of stainless steel stem-CoCr femoral head connections: comparisons of materials, initial moisture, and offset length, *Journal Of Biomedical Materials Research. Part B, Applied Biomaterials*, Jan; Vol. 88 (1), 162-173.
- Goldberg, J., Gilbert, J., Jacobs, J., Bauer, T., Paprosky, W. & Leurgans, S. (2002). A multicenter retrieval study of the taper interfaces of modular hip prostheses. *Clinical Orthopaedics and Related Research*, 401, 149-161
- Goldberg, J. & Gilbert, J. (2003). In vitro corrosion testing of modular hip tapers. *Journal of Biomedical Materials Research: Part B, Appl Biomaterials*, 64B, 78-93.

- Gruen, T & Amstutz, H. (1975). A failed vitallium/stainless steel total hip replacement: A case report with histological and metallurgical examination, *Journal of Biomedical Materials Research*, 9, 465-475.
- Hallab, N. & Jacobs, J. (2003). Orthopedic implant fretting corrosion, *Corrosion Review*, 21(2-3), 183-213.
- Hallab, N.; Messina, C.; Skipor, A. & Jacobs, J. (2004). Differences in the fretting corrosion of metal-metal and ceramic-metal modular junctions of total hip replacements, *Journal Of Orthopaedic Research*, Mar; Vol. 22 (2), 250-259.
- Harris, WH. (1994). Osteolysis and particulate disease in hip replacement. *Acta Orthop Scand*. 65:113-123.
- Kraft, C., Burian, B., Diedrich, O. & Wimmer, M. (2001). Implications of orthopedic fretting corrosion particles on skeletal muscle microcirculation, *Journal of Materials Science: Materials in Medicine*, 12, 1057-1062.
- Kurtz, S. Srivastay, S., Dwyer, K., Ochoa, J. & Brown, S. (2001). Analysis of the stem-sleeve interface in a modular titanium alloy femoral component for total hip replacement, *Key Engineering Materials*, 198-199, 41-68.
- Manley, MT. & Serekian, P. (1994). Wear debris; An environmental issue in total joint replacement. *Clinical Orthopaedics and Related Research*. 298:137-146.
- Mohrbacher, H., Celis, J., & Roos, J. (1995). Laboratory testing of displacement and load induced fretting, *Tribology International*, 28 (5), 269-278.
- Mroczkowski, M.; Hertzler, J.; Humphrey, M.; Johnson, T. & Blanchard, C. (2006). Effect of impact assembly on the fretting corrosion of modular hip tapers, *Journal Of Orthopaedic Research*, Feb; Vol. 24 (2), 271-279.
- Mutoh, Y. (1995). Mechanisms of fretting fatigue, *JSME International Journal*, 38 (4), 405-415.
- Naesguth, E.A. (1998). Wear and corrosion analysis of modular hip implants: The ceramic head-metallic neck interface, *Ph.D thesis*, Queen's University, Canada
- Rodrigues, C.; Urban, M.; Jacobs, J. & Gilbert, L. (2009). In vivo severe corrosion and hydrogen embrittlement of retrieved modular body titanium alloy hip-implants, *Journal Of Biomedical Materials Research. Part B, Applied Biomaterials*, Jan; Vol. 88 (1), 206-219.
- Shareef, N. & Levine, D. (1996). Effect of manufacturing tolerances on the micromotion at the Morse taper interface in modular hip implants using the finite element technique, *Biomaterials*, 17(6), 623-630.
- Sporer, S.; DellaValle, C.; Jacobs, J. & Wimmer, M. (2006). A case of disassociation of a modular femoral neck trunion after total hip arthroplasty, *Journal Of Arthroplasty*, Sep; Vol. 21 (6), 918-921.
- Viceconti, M.; Ruggeri, O.; Toni, A. & Giunti, A. (1996). Design related fretting wear in modular neck hip prosthesis, *Journal of Biomedical Materials Research*, 30, 181-186.
- Viceconti, M.; Massimilia, B.; Stefano, S. & Toni, A. (1998). Fretting wear in a modular neck hip prosthesis, *Journal of Biomedical Materials Research*, 35, 207-216.
- Waterhouse, R.B. (1992). Fretting fatigue, *International Materials Reviews*, 37, 77-97.
- Zhou, Z. & Vincent, L. (1995). Mixed fretting regime, *Wear*, 181-183, 531-536.
- Zhou, Z. & Vincent, L. (1997). Cracking induced by fretting of aluminium alloys, *Journal of Tribology*, 119, 36-42.

Modeling Simulation and Optimization - Focus on Applications

Edited by Shkelzen Cakaj

ISBN 978-953-307-055-1

Hard cover, 312 pages

Publisher InTech

Published online 01, March, 2010

Published in print edition March, 2010

The book presents a collection of chapters dealing with a wide selection of topics concerning different applications of modeling. It includes modeling, simulation and optimization applications in the areas of medical care systems, genetics, business, ethics and linguistics, applying very sophisticated methods. Algorithms, 3-D modeling, virtual reality, multi objective optimization, finite element methods, multi agent model simulation, system dynamics simulation, hierarchical Petri Net model and two level formalism modeling are tools and methods employed in these papers.

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Kassim Abdullah (2010). Study of Factors Affecting Taper Joint Failures in Modular Hip Implant Using Finite Element Modelling, Modeling Simulation and Optimization - Focus on Applications, Shkelzen Cakaj (Ed.), ISBN: 978-953-307-055-1, InTech, Available from: <http://www.intechopen.com/books/modeling-simulation-and-optimization-focus-on-applications/study-of-factors-affecting-taper-joint-failures-in-modular-hip-implant-using-finite-element-modellin>

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