Original Article

Assessment of failure of cemented polyethylene acetabular component due to bone remodeling: A finite element study

Rajesh Ghosh *

School of Engineering, Indian Institute of Technology Mandi, Mandi 175001, Himachal Pradesh, India

ARTICLE INFO

Article history:

Received 18 December 2015

Accepted 6 March 2016

Available online 29 March 2016

Keywords:

Hip joint

Pelvic bone

Acetabular component

Bone remodeling

Cement stress

Fatigue

ABSTRACT

The aim of the study is to determine failure of the cemented polyethylene acetabular component, which might occur due to excessive bone resorption, cement–bone interface debonding and fatigue failure of the cement mantle. Three-dimensional finite element models of intact and implanted pelvic bone were developed and bone remodeling algorithm was implemented for present analysis. Soderberg fatigue failure diagram was used for fatigue assessment of the cement mantle. Hoffman failure criterion was considered for prediction of cement–bone interface debonding. Results indicate fatigue failure of the cement mantle and implant–bone interface debonding might not occur due to bone remodeling.

© 2016 Prof. PK Surendran Memorial Education Foundation. Published by Elsevier, a division of Reed Elsevier India, Pvt. Ltd. All rights reserved.

1. Introduction

Fixation of the acetabular component has been one of the debatable issues for past few years. The type of fixation (cemented or cementless) mainly depends on different factors such as, bone quality of the patients, age of the patients and others. Long-term performance of cemented polyethylene acetabular components has been the major causes of concern during past few decades.¹ Cemented acetabular component may lose due to wear induced osteolysis, implant-induced adaptive bone remodeling, and failure of the cement mantle. In the case of cemented fixation, cement mantle has been reported weakest link among all other components of the implanted bone, since bone cement is weak in tension as compared to compression.² In order to deal with normal and active life, patients might face varieties of physiological loading activities, such as – normal walking, stair climbing, running, jogging etc. However, after implantation, normal or slow walking has been preferred by the clinicians. Owing to variations in forces (hip joint force and muscle forces) during entire normal walking, stresses in the cement mantle might vary from the maximum value to the minimum one, which might causes fatigue in the cement mantle. The material properties of the supported bone would also affect the value of stresses in the cement mantle

and stresses in other components.³ So, bone remodeling might have effect on fatigue failure of the cement mantle and eventual loosening of the acetabular component.

Finite element (FE) analysis has now been widely used for pre-clinical analyses and evaluation of the orthopedic implants, due to less computational and financial cost. Earlier studies of cemented acetabular component were mainly focused on immediate post-operative condition.^{3–9} Either they have focused on investigating the effect of stress/strain shielding around cemented acetabular component⁴ or the effect of cement thickness on stress in cement mantle^{3,6–9} or increasing the strength of cement–bone interface.^{5,6} The fatigue failure of the cement mantle of the cemented acetabular component was also studied by some authors.^{10–12} A clinical study on bone remodeling around cemented polyethylene acetabular component by Shetty et al.¹³ indicated less bone density reduction in and around periphery of the acetabulum. A computer simulated study of bone remodeling around cemented acetabular component indicates eventual bone density loss around the acetabulum.¹⁴ To the author's knowledge, fatigue failure of the cement mantle and cement–bone interface failure due to bone remodeling around cemented acetabular component has not been studied till now.

A FE study by Lamvohee et al.³ showed an increase in thickness of cement mantle would reduce the tensile stress in the cement mantle. However, their study considered only immediate post-operative condition. Irrespective of failure due to osteolysis, it is anticipated that bone remodeling around the acetabulum would

* Tel.: +91 01905 237930; fax: +91 01905 237924.

E-mail address: rajesh@iitmandi.ac.in

change the stresses in the cement mantle as well as the fatigue behavior and cement–bone interface debonding. Immediate post-operative condition might not accurately predict the loosening of the implant. Changes in stress in the cement due to bone remodeling needs to be monitored to evaluate risk of loosening. Considering these hypotheses and background, the objective of this present study is (a) to determine the bone remodeling around cemented acetabular components, (b) to investigate the fatigue failure of the cement mantle due to bone remodeling and, (c) to investigate the cement–bone interface failure due to bone remodeling.

2. Material and methods

Earlier study reported that, for elderly patients, cemented acetabular component performs better as compared to cementless acetabular component.¹⁵ For this reason, CT-scan datasets of 62 years female patient was chosen for the present analyses. From this CT-scan datasets, three-dimensional FE model of the right hemi-pelvis was generated. The detailed descriptions of the FE modeling procedure of the pelvic bone was presented in earlier studies.^{16–20} The generation procedure of the FE model of the pelvic bone was validated and verified experimentally.^{21,22} Both intact and implanted pelvic bone model were generated to evaluate causes of failure of the cemented polyethylene acetabular component. The solid model and finite element model (submodel) of the implanted pelvic bone along with all components is shown in Fig. 1. For fixation of the acetabular component, 3 mm constant

Table 1

Young's Modulus and Poisson's ratio of different materials.

Component	Young's Modulus (GPa)	Poisson's ratio	References
Cortical bone	17	0.3	23
Cancellous bone	Location dependent (from CT scan data)	0.2	23
Bone cement	2	0.33	6
UHMWPE	1.174	0.4	20
CoCrMo	210	0.3	16

thickness of the polymethylmethacrylate (PMMA) cement was considered.^{3,12} The acetabular component was positioned and orientated at an angle of 45° inclination angle and 15° anteversion angle.¹⁶ Ultra High Molecular Weight Polyethylene (UHMWPE) was considered for the material of acetabular component, having 48 mm outer diameter and 28 mm bearing diameter. A spherical femoral head was taken for application of the hip joint reaction force (Fig. 1).

2.1. Material properties

The pelvic bone was separated into two parts, namely spongy cancellous bone and thin cortical bone.²³ The material properties for each component are summarized and presented in Table 1. The material property of the cancellous was assumed isotropic, heterogeneous and elastic. The heterogeneous material property of the cancellous bone was assigned, using CT-scan data as similar to earlier studies.^{16,18} The cortical bone was assumed elastic, isotropic and homogeneous having Young's Modulus of 17 GPa, Poisson's ratio of 0.3 and density 1.73 g/cm³. The Young's Modulus of the UHMWPE acetabular component and cement mantle were considered as 1.174 GPa and 2 GPa, respectively and the Poisson's ratio for each of these materials was taken as 0.4 and 0.33.^{6,20} The spherical femoral head was made of CoCrMo alloy and for this material, Young's modulus was considered as 210 GPa and Poisson's ratio of 0.3.¹⁶

2.2. Loading conditions and bone remodeling simulations

Gait cycle or normal walking cycle was considered for loading cycle. This entire period of normal walking cycle was divided into eight static load cases.²³ For each static load cases, the action of twenty one muscle forces and hip-joint reaction force were considered as applied loading condition.¹⁸ Details of muscle forces and hip joint forces for eight static load cases were presented in earlier paper.¹⁸ At the pubis and the sacroiliac joint, fixed constraint was prescribed.^{9,18} A submodelling technique was implemented in this present investigation.¹⁶ Strain energy-based time-dependent adaptive bone remodeling algorithm was used in this study.¹⁶ A complete description of the bone remodeling algorithm was presented in earlier studies.^{16,19} However, instead of two load cases, eight static load cases of a normal walking cycle were considered in this study. The results of bone remodeling were presented in terms of months and years, where the time (month or year) was calculated based on bone adaptation rate reported by Weinans et al.²⁴ Hip joint reaction force was applied at the center of the spherical femoral head.

2.3. Interfacial conditions

In the case of cemented fixation, the acetabular component is fixed to the bone with cement. For this reason, fully-bonded implant–cement and cement–bone interfacial condition was assumed.¹² Frictionless contact was simulated at the interface of acetabular component and femoral head,¹² to represent well

Fig. 1. (a) Solid model of the implanted pelvic bone showing all components and boundary conditions, (b) Finite element model (sub model) of the implanted pelvic bone.

lubricating articulating surface. A 0.5 mm radial clearance was provided between femoral head and acetabular component. An asymmetric surface to surface contact element was used for the contact simulation. Convergence of solution for nonlinear contact analysis is dependent on contact parameters, namely contact stiffness and penetration tolerance. In this present analysis, contact stiffness 100 N/mm² and penetration tolerance 0.1 (factor) were found to be suitable for convergence of solutions. The finite element model of the intact and implanted pelvic bone in combine with bone remodeling algorithm, was solved using ANSYS v 14 (ANSYS Inc. PA, USA).

2.4. Assessment of fatigue failure of the cement mantle

Fatigue failure of the cement mantle was checked, using Soderberg criterion.²⁵ Owing to variation in hip joint force and muscle forces during normal walking, tensile stresses in the cement mantle changes from minimum value to maximum value. In the finite element analysis, tensile stress generated in every node within the cement mantle was calculated for each load cases of normal walking. The maximum and minimum value of tensile stress for each node was identified during normal walking (for every eight static load cases). After obtaining maximum ($\sigma_{\max}$) and minimum ($\sigma_{\min}$) value of tensile stress, the mean value of stress (σ_m) and stress amplitude (σ_a) was calculated, using following two equations.

$$\sigma_m = \frac{\sigma_{\max} + \sigma_{\min}}{2} \quad (1)$$

$$\sigma_a = \frac{\sigma_{\max} - \sigma_{\min}}{2} \quad (2)$$

Thereafter, Soderberg fatigue failure diagram was used²⁵ in order to check whether fatigue failure would occur in the cement mantle due to bone remodeling. In the Soderberg diagram, the

mean stress value was plotted in abscissa and stress amplitude was plotted in ordinate. A straight line as shown in Figs. 4–7, known as Soderberg line, was drawn, based on the value of endurance limit and value of tensile strength of the PMMA cement to evaluate fatigue failure of the cement mantle. In this present study, the value of endurance limit and tensile strength of the PMMA cement was considered as 9.2 MPa and 35.3 MPa, respectively.^{2,26} The safe zone of the fatigue failure is indicated in Figs. 4–7.

2.5. Assessment of cement–bone interface failure

The cement–bone interface failure (debonding) was checked, using multi-axial Hoffman failure criterion.²⁷ This criterion accounts for both normal stress (σ_n) and shear stress (τ_s) at the cement–bone interface. The value of σ_n and τ_s for each nodal points were calculated, using components of stress generated ($\sigma_x, \sigma_y, \sigma_z, \tau_{xy} = \tau_{yx}, \tau_{yz} = \tau_{zy}$ and $\tau_{xz} = \tau_{zx}$) at each node and the unit normal vector to the inclined plane parallel to the interface at the point. Thereafter, Hoffman number (FL) was determined for each node, using following equation.

$$FL = \frac{1}{S_t S_c} \sigma_n^2 + \left(\frac{1}{S_t} - \frac{1}{S_c} \right) \sigma_n + \frac{1}{S_s^2} \tau_s^2 \quad (3)$$

where S_t , S_s and S_c are the uni-axial interface tensile, shear and compressive strengths. These interface strengths were function of bone density (ρ) and determined by following equations.^{28,29}

$$S_t = 14.5\rho^{1.71}, \quad S_s = 21.6\rho^{1.65}, \quad S_c = 32.4\rho^{1.85} \quad (4)$$

For each and every nodal points, if $FL > 1$, then interface debonding will occur or if, $FL < 1$, then failure at this nodal point will not happen. The cement–bone interface debonding was

Fig. 2. Bone density distribution at different stages of bone remodeling.

assessed for immediate post-operative conditions and different stages of bone remodeling.

3. Results

Bone density distributions around the acetabulum and tensile stress distributions in the cement mantle at different stages of bone remodeling are presented in Figs. 2 and 3, respectively. Soderberg diagrams for fatigue assessment of the cement mantle at different stages of bone remodeling are shown in Figs. 4–7.

3.1. Bone remodeling around cemented polyethylene acetabular component

Implant induced bone remodeling around cemented polyethylene acetabular component indicate increase in bone density in the superior part of the acetabulum (Fig. 2). Increase in bone

density was also observed in the inferior part of the acetabulum (Fig. 2). Severe bone density reduction in and around the posterior part of the acetabulum (40–60%) indicates less load transfer after implantation across this region (Fig. 2). In the anterior-inferior part of the acetabulum, around 30–40% bone density reduction was predicted. Bone density reduction (5–10%) was less at the supero-anterior part of the acetabulum. Fig. 2 indicates less amount of bone density increase or decrease after 6 months of bone remodeling.

3.2. Tensile stress distribution in the cement mantle

Maximum tensile stress in the cement mantle was found for beginning of right single support phase of walking cycle, which is 13% of gait cycle and shown in Fig. 3. It has been observed that, minimum tensile stress in the cement mantle was found for halfway right swing phase (85% of cycle) of gait cycle. This caused

Fig. 3. Tensile stress distribution in the cement mantle.

Fig. 4. Soderberg diagram showing distributions of stress amplitude (alternating stress) and mean stress in the cement mantle after immediate post-operative condition. Marker indicates alternating and mean value of stress for different nodes.

Fig. 5. Soderberg diagram showing distributions of stress amplitude (alternating stress) and mean stress in the cement mantle after six-months of bone remodeling. Marker indicates alternating and mean value of stress for different nodes.

due to the magnitude of hip joint force was least among all eight load cases of the normal walking cycle. Considering immediate post-operative condition, maximum tensile stress was found at the superior part of the cement mantle (Fig. 3). Maximum value of tensile stress was found to be 4.28 MPa. Tensile stress in the inferior and the posterior part of the cement mantle was found to be low and ranging from 0.10 MPa to 1.50 MPa (Fig. 3). After bone remodeling, minor reduction in tensile stress was observed from central part of the cement mantle to the inferior part of the cement mantle (Fig. 3). Bone remodeling leads to marginal increase in tensile stress at the superior part of the acetabulum, where maximum tensile stress was found to be 4.43 MPa.

3.3. Fatigue assessment of the cement mantle

Fatigue assessment of the cement mantle for each time step of bone remodeling was monitored, however, results corresponds to immediate post-operative condition, after six months of bone remodeling, after one year of bone remodeling and after two year of bone remodeling are presented in Figs. 4–7, simultaneously. Maximum tensile (for 13% of gait cycle) and minimum tensile

(for 85% of cycle) stress in the cement mantle for each node was obtained, thereafter; mean stress and stress amplitude were calculated, using Eqs. (1) and (2). It was observed that, after immediate post-operative condition, mean stress value was varied slightly greater than zero to maximum of 2.63 MPa (considering all nodes in the cement mantle), whereas stress amplitude value was ranging from slightly higher than zero to maximum of 1.74 MPa (Fig. 4; Table 2). Maximum and minimum values of mean stress (MPa) and stress amplitude (MPa) in the cement mantle at different stages of bone remodeling are presented in Table 2. The data of mean stress and stress amplitude for each node within the cement mantle shows far below the Soderberg line (Fig. 4). There was no remarkable variation of mean stress and stress amplitude was observed at different stages of bone remodeling and all data were far below the Soderberg line (Figs. 4–7; Table 2).

3.4. Cement–bone interface failure assessment

Based on normal stress (σ_n), shear stress (τ_s) and interface strengths, Hoffman number (FL) was calculated for each nodal points after immediate post-operative conditions, after six months

Fig. 6. Soderberg diagram showing distributions of stress amplitude (alternating stress) and mean stress in the cement mantle after one-year of bone remodeling. Marker indicates alternating and mean value of stress for different nodes.

Fig. 7. Soderberg diagram showing distributions of stress amplitude (alternating stress) and mean stress in the cement mantle after two-years of bone remodeling. Marker indicates alternating and mean value of stress for different nodes.

Table 2

Maximum and minimum values of mean stress (MPa) and stress amplitude (MPa) in the cement mantle at different stages of bone remodeling.

	Immediate Post-operative		Six months after bone remodeling		One year after bone remodeling		Two year after bone remodeling	
	Maximum	Minimum	Maximum	Minimum	Maximum	Minimum	Maximum	Minimum
Mean Stress	2.63	~0	2.66	~0	2.67	~0	2.69	~0
Stress Amplitude	1.74	~0	1.76	~0	1.73	~0	1.73	~0

of bone remodeling, after one year of bone remodeling and after two years of bone remodeling to evaluate debonding of cement–bone interface. After immediate post-operative condition, the FL varied from 0.006 to 0.88 for nodal point adjacent to the interface, except twelve nodal points located at the central part of the acetabulum, where FL was found more than 1. The bone density at this region was varied from 0.10 to 0.28 g/cm³. FL number for all nodal points after six months, after one year and after two years of bone remodeling were found in similar range and these values were ranging from 0.035 to 0.97. However, at the medial portion and the inferior portion of the acetabulum, only twenty-three nodal points have found more than 1 FL number at different stages

of bone remodeling. Although the likelihood of interface debonding increase slightly due to bone remodeling, a few number of failure nodal points, 12–23 as compared to total 1204 interface nodes, suggested that interface debonding is less likely occurred due to bone remodeling.

4. Discussion

The purpose of this study is to determine the mechanical causes of failure of cemented acetabular component which might occur due to bone remodeling, fatigue failure of the cement mantle and cement–bone interface debonding. In order to investigate these, a

finite element model of intact and implanted pelvic bone were developed, using CT-Scan data. Normal walking cycle was used to evaluate the fatigue failure of the cement mantle. Soderberg fatigue failure diagram and Hoffman failure criterion were used for fatigue analysis and implant–bone interface debonding. Previous studies on fatigue failure of the cement mantle were mainly focused on effect of cement mantle thickness, acetabular size, body mass index.^{3,10–12} To date, a complete analysis have not been performed yet to evaluate the mechanical causes of failure (fatigue failure of the cement mantle due to bone remodeling and cement–bone interface debonding due to bone remodeling) of the cemented polyethylene acetabular component.

Bone remodeling around cemented polyethylene acetabular component indicated increase in bone density around the rim of the acetabulum. This result was comparable to earlier clinical data published by Shetty et al.¹³ The study by Shetty et al.¹³ showed reduction in bone density around the medial part of the acetabulum up to 3 months of bone remodeling, thereafter, bone density reached its base line value after two years of bone remodeling. In this present study, bone density reduction was also observed at the medial part of the acetabulum, however, that never reached its base line value throughout the entire period of bone remodeling cycle (Fig. 2). A finite element based bone remodeling study by Bouguecha et al.¹⁴ indicated significant amount bone loss around the cemented polyethylene acetabular component and anticipated loosening of the cemented acetabular component, which is inconsistent to the current study, where increase in bone density was observed at the superior-posterior part of the acetabulum. This inconsistent of the result is due to FE modeling technique used by Bouguecha et al.¹⁴, where they did not considered the sandwich construction of the pelvic bone (no separate cortical and cancellous bone).

After immediate post-operative conditions, maximum tensile stress was found to be 4.28 MPa and observed at the superior part of the cement mantle for beginning of right single support phase of walking cycle, which is 13% of gait cycle (Fig. 3). Qualitatively this results was comparable to earlier published studies, however, quantitatively mismatch (in terms of tensile stress in the cement mantle) was found.^{3,5,6,9} This mismatch might be due to simplified loading conditions^{3,5,6,9} and assumption of homogeneous material properties for cancellous bone.^{3,5,6} These assumptions might not be entirely representative of the physiological conditions of the pelvic bone. In some studies, von Mises stress was used to evaluate failure and fatigue failure of the cement mantle,^{7,8} however, for brittle material, von Mises stress is not preferred stress for predicting failure.²⁵ In the year, 2007, Zant et al.¹² used 2-D simplified model of implanted pelvic bone to understand the fatigue failure of the cement mantle. This 2-D model might not be entirely representative of the physiological conditions of the pelvic bone in terms of loading, geometry and boundary conditions.

Owing to bone remodeling, a slight change in maximum tensile stress was observed at the superior part of the cement mantle, where 3.5% increase in tensile stress was observed in this region (Fig. 3). Bone remodeling led to reduction in tensile stress from central part of the cement mantle to inferior part of the cement mantle (Fig. 3). This might cause due to reduction in bone density around these regions. The maximum values of mean stress and stress amplitude 2.69 MPa and 1.76 MPa, respectively. These values were far lower than the endurance limit of stress and tensile strength of the cement mantle.^{2,26} Fatigue failure analysis (from Soderberg diagram) of the cement mantle at different stages of bone remodeling showed less chance of failure of the 3 mm constant cement mantle during normal walking.

The study has number of assumptions and limitations. Cancellous bone was assumed heterogeneous, isotropic and elastic; however, in real case cancellous bone might be anisotropic or viscoelastic in nature. For macroscopic analysis, heterogeneous,

isotropic and elastic material properties of the cancellous bone might be good assumptions as reported by earlier studies. Only normal walking cycle was considered for applied loading condition. Including more demanding physiological loading cycle, such as stair climbing, running, jogging, fast walking might be more effective for evaluation of fatigue failure of the cement mantle and interface debonding. Only one constant cement mantle thickness was considered in this present study. Variation of the cement mantle thickness might be useful for fatigue analysis. The bone remodeling data was presented in terms of months or years, where time (months or years) was calculated from bone adaptation rate reported by Weinans et al.²⁴ The adaptation rate of the bone might depends on, age of the patients, sex of the patients, bone quality etc. In order to obtain the real time history of the bone remodeling, adaptation rate of the same patients would be more useful for bone remodeling simulation. Only one acetabular orientation was considered in the present study. Improper positioning of the acetabular component might have different view of the failure of the cemented acetabular component.

Based on interface failure assessment at different stages of bone remodeling, cement–bone interface debonding would not occurred, where most of the nodal points, the FL number was below 1. A clinical study on failure analysis of the cemented polyethylene acetabular component showed that aseptic loosening of the acetabular component did not occur due to mechanical factors, and that loosening of the component was basically biological in nature.³⁰ Results of the present study also indicated mechanical causes of failure of the cemented polyethylene acetabular component are less during normal walking.

5. Conclusions

Based on finite element modeling and analyses of the cemented polyethylene acetabular component following conclusions may be drawn. Results of the present study shows increase in bone density at the superior part of the acetabulum, indicating more load transfer through this region. Bone density reduction at the posterior part and antero–inferior part of the acetabulum indicated less load transfer throughout these regions. The mean value of stress and stress amplitude in the cement mantle at different stages of bone remodeling were far below the Soderberg line, indicating less chance of fatigue failure of the cement mantle. Based on interface failure assessment, cement–bone interface debonding might not occur due to bone remodeling.

Conflict of interest

The author states that with regard to the submission of this research paper, there are no financial and personal relationships with other people and organizations.

Acknowledgements

Author would like to thank IIT Kharagpur and University of Southampton for providing the CT-scan data of the pelvic bone.

References

1. National Joint Registry. *National Joint Registry for England and Wales. 11th Annual Report*. Hemel Hempstead: NJR; 2014. September.
2. Lee C. The mechanical properties of PMMA bone cement. In: *The Well-Cemented Total Hip Arthroplasty*. Springer; 2005:60–66. ISBN:978-3-540-24197-3 (Print) 978-3-540-28924-1 (Online).
3. Lamvohee JMS, Ingle P, Cheah K, Dowell J, Mootanah R. Total hip replacement: tensile stress in bone cement is influenced by cement mantle thickness, acetabular size, bone quality, and body mass index. *J Comput Sci Syst Biol*. 2014;7:72–78.
4. Carter DR, Vasu R, Harris WH. Stress distributions in the acetabular region II: effects of cement thickness and metal backing of the total hip acetabular component. *J Biomech*. 1982;17:165–170.

5. Mootanah R, Ingle P, Cheah K, Dowell JK, Shelton JC. Total hip replacement: results of a postal survey of current practice on the cement fixation of the acetabular cup in the UK. *Hip Int.* 2004;14:155–162.
6. Mootanah R, Dowell JK, Cheah K, Ingle P, Shelton JC. Configuration of anchorage holes affects fixation of the acetabular component in cemented total hip replacement – a finite element study. *Comput Methods Biomech Biomed Eng.* 2007;10:439–445.
7. Kumar YS, Pant B, Darunkumar Singh K. Thickness effects on maximum von-Mises stress of a cement mantle in total hip replacement – a finite element study. *J Appl Biomater Biomech.* 2009;7:111–115.
8. Coultrup OJ, Hunt C, Wroblewski BM, Taylor M. Computational assessment of the effect of polyethylene wear rate, mantle thickness, and porosity on the mechanical failure of the acetabular cement mantle. *J Orthop Res.* 2010;28:565–570.
9. Zhang QH, Wang JY, Lupton C, et al. A subject-specific pelvic bone model and its application to cemented acetabular replacements. *J Biomech.* 2010;43:2722–2727.
10. Tong J, Zant NP, Wang JY, Heaton P, Hussell JG. Fatigue in cemented acetabular replacements. *Int J Fatigue.* 2008;8:1366–1375.
11. Adegbile PH, Zant NP, Tong J. In vitro fatigue behaviour of a cemented acetabular reconstruction. *J Biomech.* 2006;39:2882–2886.
12. Zant NP, Wong CK, Tong J. Fatigue failure in the cement mantle of a simplified acetabular replacement model. *Int J Fatigue.* 2007;29:1245–1252.
13. Shetty NR, Hamer AJ, Kerry RM, Stockley I, Eastell R, Wilkinson JM. Bone remodelling around a cemented polyethylene cup: a longitudinal densitometry study. *J Bone Jt Surg Br.* 2006;88:455–459.
14. Boughecha A, Elgaly I, Stukenborg-Colsman C, et al. Numerical investigations of the strain – adaptive bone remodelling in the prosthetic pelvis. *IFMBE Proc.* 2010;29:562–565.
15. Hailer NP, Garellick G, Karrholm J. Uncemented and cemented primary total hip arthroplasty in the Swedish hip arthroplasty register. *Acta Orthop.* 2010;81:34–41.
16. Ghosh R, Mukherjee K, Gupta S. Bone remodelling around uncemented metallic and ceramic acetabular components. *Proc Inst Mech Eng H: J Eng Med.* 2013;22:490–502.
17. Ghosh R, Gupta S. Bone remodelling around cementless composite acetabular components: the effects of implant geometry and implant–bone interfacial conditions. *J Mech Behav Biomed Mater.* 2014;32:257–269.
18. Ghosh R, Pal B, Ghosh D, Gupta S. Finite element analysis of a hemi-pelvis: the effect of inclusion of cartilage layer on acetabular stresses and strain. *Comput Methods Biomech Biomed Eng.* 2015;18(7):697–710.
19. Ghosh R, Mukherjee K, Gupta S. Bone remodelling around uncemented acetabular prostheses. *Bone Jt J Br.* 2013;95:184.
20. Ghosh R, Pratihari DK, Gupta S. Towards the optimal design of an uncemented acetabular component using genetic algorithms. *Eng Optim.* 2014. <http://dx.doi.org/10.1080/0305215X.2014.971779>.
21. Ghosh R, Gupta S, Dickinson A, Browne M. Experimental validation of finite element models of intact and implanted composite hemi-pelvises using digital image correlation. *Trans ASME J Biomech Eng.* 2012;134(8). 081003(1–9).
22. Ghosh R, Gupta S, Dickinson A, Browne M. Experimental validation of numerically predicted strain and micromotion in intact and implanted composite hemi-pelvises. *Proc Inst Mech Eng H: J Eng Med.* 2013;22:162–174.
23. Dalstra M, Huiskes R. Load transfer across the pelvis bone. *J Biomech.* 1995;28:715–724.
24. Weinans H, Huiskes R, van Reitbergen B, Sumner DR, Turner TM, Galante JO. Adaptive bone remodelling around bonded noncemented total hip arthroplasty: a comparison between animal experiments and computer simulation. *J Orthop Res.* 1993;11:500–513.
25. Bhandari VB. *Design of Machine Elements.* Tata McGraw Hill Education; 2010.
26. Bialoblocki-Juszczak E, Baleani M, Cristofolini L, Viceconti M. Fracture properties of an acrylic bone cement. *Acta Bioeng Biomech.* 2008;10:21–26.
27. Hoffman O. The brittle strength of orthotropic material. *J Compos Mater.* 1967;1:200–206.
28. Kaplan SJ, Hayes WC, Stone JL. Tensile strength of bovine trabecular bone. *J Biomech.* 1985;18:723–727.
29. Stone JL, Beaupre GS, Hayes WC. Multiaxial strength characteristics of trabecular bone. *J Biomech.* 1983;16:743–752.
30. Schmalzried TP, Kwong LM, Jasty M, et al. The mechanism of loosening of cemented acetabular components in total hip arthroplasty. Analysis of specimens retrieved at autopsy. *Clin Orthop Relat Res.* 1992;274:60–78.