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Flexural and fatigue properties of a bone cement based upon polyethylmethacrylate and hydroxyapatite

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Polymethylmethacrylate (PMMA) bone cement is commonly used in surgery to fix joint replacements into the bone. Although the operations are generally successful, loosening of the prosthesis does occur with fracture of the bone cement treated as the source of failure in some instances. Polyethylmethacrylate (PEMA) bone cement offers a promising alternative to PMMA due to its high ductility, low toxicity and low exotherm. In addition, hydroxyapatite (HA) particles can be added, while retaining the ductile properties of the material. In this study, the flexural and fatigue properties of this experimental cement, with and without HA reinforcement, have been examined. It was found that up to 40wt.% HA could be added with increases in both flexural strength and modulus. Specimens were subjected to tension-tension cyclic loading at a number of stress levels until catastrophic failure occurred. In comparison with a commercial PMMA cement, tested at relatively high stresses, the PEMA cement failed at lower cycles to failure. However, the data converged at the lower stresses employed which are closer to the physiological loading situation. With the addition of HA, although the cycles to failure were decreased, the deformation experienced by the PEMA-HA cement whilst being cycled was reduced.

1. Introduction

One of the most challenging aspects of total joint replacement is the fixation of the implant into the bone. The most common method employed to achieve this is to use polymethylmethacrylate (PMMA) bone cement. However, implant loosening does occur, with bone cement being implicated in some cases [1]. Thus, although PMMA based bone cement has widespread use, it does not possess the ideal mechanical characteristics required. As a brittle material, PMMA bone cement has relatively low fracture energy [2], one through which cracks can easily propagate causing fracture to occur at stresses lower than the ultimate tensile strength. Cracks initiate at stress concentrations due to pores and inclusions or at mechanical interdigitations on the bone cement interface. In addition, PMMA bone cement reaches temperatures up to 100 °C [3] during polymerization, which results in bone necrosis. Previous approaches to enhancing mechanical properties of bone cement have been directed towards improving existing PMMA bone cements, rather than developing new cements. The modifications attempted initially concentrated upon the addition of fibres to the cement [4–6]. However, the bone cements produced possessed much increased stiffnesses and displayed poor intrusion characteristics. In order to overcome these difficulties, research has been directed to the use of particulate composites, which offer the possibility of strengthening, without presenting problems of stress distribution or of compromising flow characteristics [7–9]. The addition of stiffer, brittle particles leads to further increases in the modulus of the PMMA and decreases in fracture energy. An alternative approach is to use a lower modulus cement as the matrix material. A more flexible cement would yield or flow before fracture and result in a more even stress distribution. The biomechanics at the bone cement interface will also be improved by a lower modulus cement as demonstrated by Litsky *et al.* who have developed a bone cement based upon polybutylmethacrylate with methylmethacrylate monomer [10].

The bone cement in this investigation was first reported by Weightman *et al.* [11], who developed a bone cement that possessed a low exotherm (55 °C), relatively low modulus (700 MPa) and high ductility, (50% strain to failure). The cement, based upon polyethylmethacrylate (PEMA) and n-butylmethacrylate (nBMA) monomer, has been shown to have other distinct advantages over conventional bone cements, both physiochemically and biologically. Residual monomer studies of the experimental cement showed lower extractability of n-butylmethacrylate compared to methylmethacrylate [12]. Toxicity studies [13] and a biological response investigation [14] also demonstrated the excellent biocompatibility of the PEMA-nBMA cement. However, although clinical trials have found the cement to be generally satisfactory, certain

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prostheses have experienced high creep. In order to overcome this problem and to offer the potential of increased bioactivity, hydroxyapatite (HA) particles were introduced. Behiri *et al.* [15] showed that the Young's modulus increased and strain to failure decreased upon the addition of HA. These properties were enhanced by the introduction of a silane coupling agent to the surface of the HA [16].

Since *in vivo* bone cement is subjected to cyclic loading of approximately one million cycles a year during normal walking at a frequency of one step per second, one of the most important properties of bone cement is that of fatigue. Therefore, in this study the fatigue characteristics of the experimental cement along with flexural properties were examined.

2. Materials and methods

2.1. Specimen preparation

The experimental bone cement consisted of polyethylmethacrylate powder, containing benzoyl peroxide initiator (Bonar Polymers, Newton Aycliffe, Co. Durham, UK) and n-butylmethacrylate monomer including N,N-dimethyl-p-toluidine (Aldrich Chemicals, Gillingham, Dorset, UK). The polymer to monomer proportions were a 2:1 weight ratio, generally used in batches of 40 g of powder and 20 g of liquid. The polymethylmethacrylate bone cement used for comparison was a commercially available low viscosity cement. All cements were prepared in a similar manner. The polymer powder was mixed with the monomer liquid for 60 s before being transferred to the syringe body of a cement gun. At dough time, measured in accordance with ASTM F451-86 [17] for each cement, the mixture was extruded into a PTFE mould. The filled moulds were subsequently placed under a pressure of 1.4MPa for approximately 20 min and the resulting specimens stored in dry conditions for one week at 37 °C prior to testing.

In order to investigate the influence of hydroxyapatite (HA) reinforcement on the PEMA bone cement, the HA was sieved together with the PEMA powder prior to being mixed with the monomer. Different weight fractions of powder were achieved by removing PEMA polymer from the 40 g of powder and replacing with equal weight of hydroxyapatite particles. This powder mixture was then mixed with a reduced amount of monomer in order to maintain a 2:1 polymer to monomer weight ratio. The maximum amount of HA that could be added in this way corresponded to a weight fraction of powder composition of 40%. The doughy mixture was thumbed into the appropriate mould and prepared as described previously.

2.2. Flexural testing

The specimens employed in the flexural tests were flat rectangular bars of dimensions, 80 mm length, 10 mm width and approximately 5 mm thickness, in accordance with ISO 178 [18]. The thicknesses of the specimens, which varied from batch to batch depending on the amount of excess filling of the mould, were measured and a mean calculated. They were tested on an Instron 6025™ testing machine in three point bending using a cross-head speed of 1.7 mm/min. The distance between the supports was calculated using the following equation:

$$L = (16 \pm 1)h \quad (1)$$

where L = distance between supports, h = mean thickness of set of specimens.

Tests were terminated after the maximum stress had been reached or following failure, as was the case with the PMMA specimens. The flexural modulus was calculated as described in ISO 178.

2.3. Fatigue testing

The fatigue test specimens were 75 mm in length, with a reduced cross-sectional area of 20 mm^2, corresponding to half-size ISO 527 multi-purpose test specimens [19]. The testing was conducted on a Bionix 858™ MTS electrohydraulic testing system. The machine was controlled using a 486PC and the number of cycles, testing time and extension of the specimen were recorded continuously until macroscopic failure. Specimens were tested in air at room temperature. Fatigue testing was performed under load control at a frequency of 2 Hz using a sinusoidal loading pattern. The specimens were cycled in tension–tension with the upper stress level varying from 30–70% of tensile strength. The lower stress level used in each loading cycle was 0.3MPa. Ten specimens were tested at each stress level, with the exception of the 40 wt%HA cement for which six specimens were tested. All fractured surfaces were preserved for scanning electron microscopy.

In order to compare the loading stresses to the tensile strength, the maximum tensile strength was measured for each bone cement composition. Specimens of the same dimensions used for the fatigue characterization were tested on the Instron 6025™ using a cross-head speed of 5 mm/min. The tensile strength was calculated from the maximum force obtained.

The results obtained from the fatigue tests were analysed using a Weibull model to establish fatigue life survival probability. The ordinate (W) used on the probability curves was determined from:

$$W = \log(1/(1 - P)) \quad (2)$$

where P = median rank.

2.4. Scanning electron microscopy

The fracture surfaces were examined using a JEOL 6300™ series Scanning Electron Microscope with an accelerating voltage of 10 kV. The specimens were gold coated prior to examination and images recorded on photographic film.

3. Results

3.1. Flexural tests

Fig. 1 demonstrates how the inclusion of up to 40 wt% HA in PEMA bone cement increases the

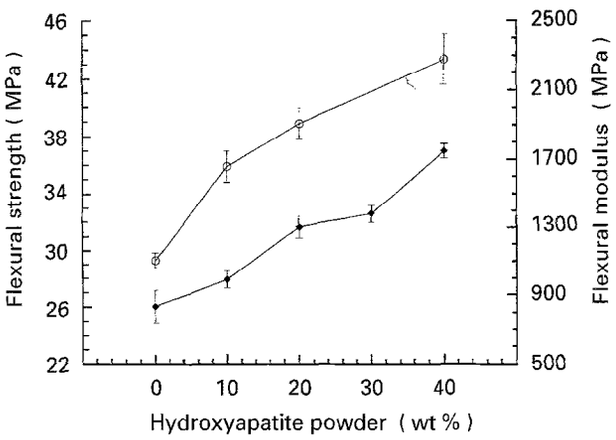

Figure 1 Influence of hydroxyapatite powder on flexural strength and modulus of PEMA bone cement. (– ○ –) flexural strength (– ◆ –) flexural modulus.

flexural strength from 29.3(± 0.54) MPa to 43.3(± 1.75) MPa and the flexural modulus from 835(± 95.5) MPa to 1746(± 43.7) MPa. The error bars and numbers shown in brackets are the standard deviations of six specimens. The values obtained for the PMMA bone cement were 67.7(± 5.7) MPa for the flexural strength and 2909(± 298) MPa for the flexural modulus. The PEMA-HA specimens did not fail but deformed significantly before the test was terminated. The PMMA specimens however, failed in a brittle manner, with limited deformation observed prior to fracture.

3.2. Fatigue tests

The fatigue results obtained were used to construct a conventional stress versus number of cycles to failure curve as shown in Fig. 2(a). The two cements represented are the PEMA, without HA addition, and PMMA bone cements. The stress indicated on the y-axis represents the upper stress level used for the loading cycle, which was varied to obtain the *S-N* curve. The upper stress values were normalized with respect to tensile strength in order to demonstrate the influence of tensile strength upon the fatigue characteristics of the materials and plotted against the cycles to failure as shown in Fig. 2(b). The ordinate in Fig. 2(b) represents the upper stress level as a percentage of the maximum tensile strength. Fig. 2(b) shows that, compared to tensile strength, the PEMA exhibits a superior fatigue resistance over PMMA cement. The tensile strengths were measured to be 25.0(± 0.8) MPa for the PEMA bone cement and 39.5(± 4.84) MPa for the PMMA bone cement. The maximum strengths represent the mean of six tensile tests and are followed by the standard deviations. It should be noted that the variation in strength values for both flexural and tensile tests was greater for the PMMA cement that for the PEMA cements.

The HA reinforced PEMA cements were tested in fatigue with a loading pattern of 0.3–15 MPa representing 60% of tensile strength. The tensile strength of each reinforced cement was approximately 25 MPa. Fig. 3 shows the Weibull probability plots obtained

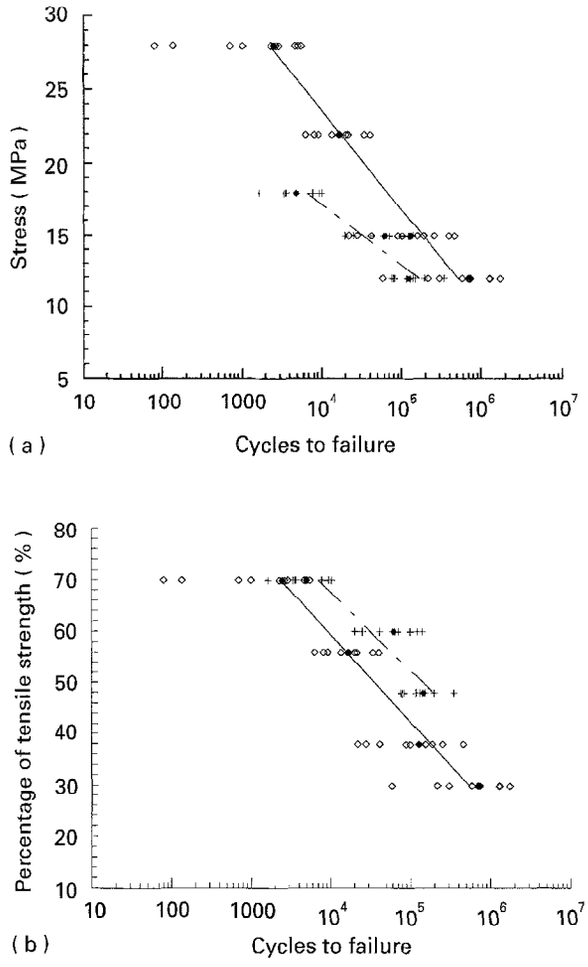

Figure 2 (a) Stress versus cycles to failure for PEMA and PMMA bone cements: (◇) PMMA, (+) PEMA, (– ● –) median (PEMA), (– ◆ –) median (PMMA). (b) Normalized stress versus cycles to failure for PEMA and PMMA bone cements: (◇) PMMA, (+) PEMA, (– ◆ –) median (PMMA), (– ● –) median (PEMA).

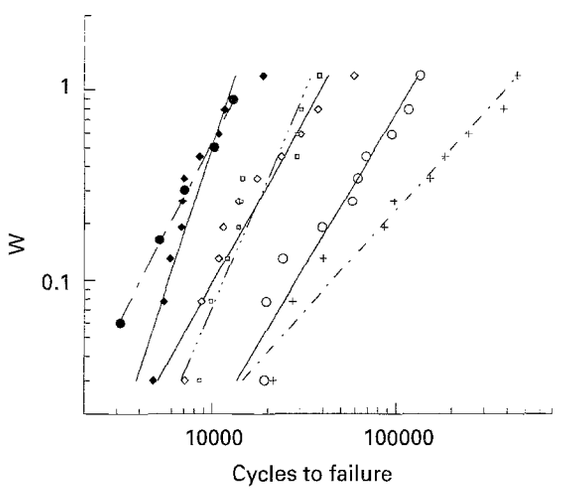

Figure 3 Weibull Probability Plots for 0-40 wt% hydroxyapatite reinforced PEMA bone cement cycled 0.3–15 MPa: (– ○ –) 0%PEMA(r = 0.987), (– ◇ –) 20%HA(r = 0.964), (– . + . –) PMMA(r = 0.989), (– ◆ –) 10% HA(r = 0.990),(– ● –) 40%HA(r = 0.991), (– □ –) 30%HA(r = 0.950).

for these tests, along with the data for the PEMA and PMMA cements. The maximum reinforcing effect was found at 20 and 30 wt% HA, producing median fatigue lives of 15640 and 14316 respectively. With 10 and 40 wt% HA the median fatigue lives obtained

were lower at 6906 and 7010 respectively. As the specimens are being cyclically stressed they gradually deform, not returning to their original gauge length on the return to a stress of 0.3 MPa. The deformation experienced by the PEMA cement specimens when cycled was observed to be up to 12 mm for some specimens, with a starting gauge length of 25 mm (i.e. 50% elongation). However, with the addition of HA this dynamic creep was reduced by a factor of 5–6.

3.3. Scanning electron microscopy

The fatigue fracture surfaces examined using scanning electron microscopy revealed that the PEMA cements possessed very low porosity whereas the PMMA cement often displayed pores of 1 mm or greater. At higher magnifications it was observed that the fracture surfaces of the PEMA cements were rough in texture (Fig. 4) compared to the PMMA cement (Fig. 5) which were much smoother.

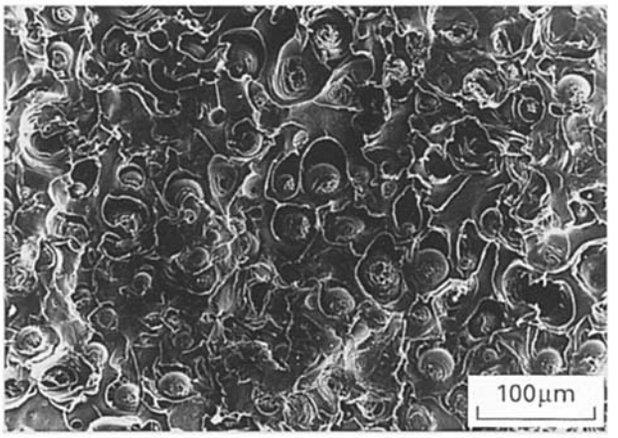

Figure 4 Fatigue fracture surface of PEMA bone cement.

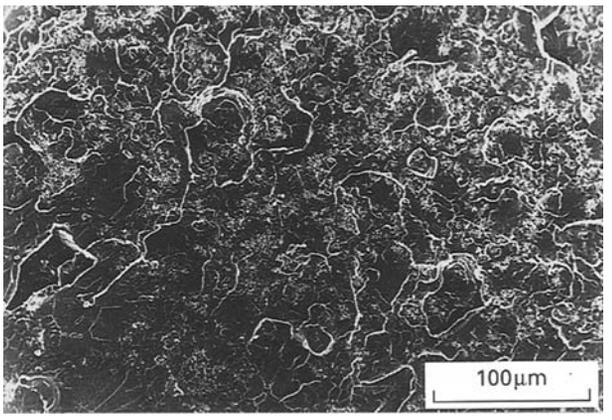

Figure 5 Fatigue fracture surface of PMMA bone cement.

4. Discussion

The test results demonstrate that the flexural properties of PEMA-nBMA bone cement can be improved with up to 40 wt% of hydroxyapatite. Even with such a high proportion of hydroxyapatite the material remains ductile and does not become brittle and fail catastrophically. In contrast to this result, for PMMA based cements, Castaldini and Cavallini [9] found that only up to 12.5 wt% hydroxyapatite particles could be added to Simplex-P bone cement before the mechanical properties deteriorated. Sogal and Hulbert [8] also reported that only up to 10 wt% hydroxyapatite particles could be added to Palacos R before tensile properties decreased. Perek and Pilliar [7] have obtained an increase of fracture toughness with up to 40 wt% hydroxyapatite in Zimmer Low Viscosity PMMA bone cement, but no other mechanical properties were measured.

The stress versus number of cycles to failure curve in Fig. 2(a) shows that at the relatively high stress levels used to date, the PMMA bone cement exhibits longer fatigue lives than the PEMA cement. This result is not unexpected, since the stresses employed represent higher percentages of tensile strength for the PEMA than for the PMMA bone cement, demonstrated by the normalized plot in Fig. 2(b). In Fig. 2(b) it can be seen that, for given percentages of the tensile strength, the PEMA cement has a better fatigue resistance. This is because PEMA bone cement is a much more ductile material than the PMMA bone cement. If the curves representing the median lives in Fig. 2(a) are extrapolated to lower stresses there will be a crossover point at approximately 6 MPa. This stress represents a higher value of bone cement mantle stress than has been calculated using a three-dimensional finite element model with a 3 kN load by Crowninshield *et al.* [20]. The highest tensile stresses found by the model were in the proximal–lateral region of the femoral canal, with a maximum of approximately 2.8 MPa. With a lower modulus cement, this stress will be expected to be reduced still further [21]. Therefore, unless an endurance limit is obtained with the PMMA cement, the results suggest the PEMA bone cement will possess a higher fatigue resistance at physiological loading.

The slopes of the Weibull plots are an indication of the scatter of the fatigue data. Fig. 3 shows that the PMMA bone cement exhibited a wide distribution of cycles to failure at the 0.3–15 MPa loading compared to the PEMA, with and without the addition of HA. The variability of data is also demonstrated by the large standard deviations obtained for the flexural and tensile strengths and flexural modulus. The coefficient of variation for the tensile strength of PEMA cement was found to be 3.2% as compared to 12.2% for PMMA cement. The scatter may be attributed to the variable and high porosity possessed by the PMMA and to the brittle nature of the cement. Davies *et al.* [22] demonstrated that upon reduction of porosity by centrifugation of Simplex-P bone cement, the slopes of the probability versus cycles to failure curves at the 0.08 and 0.05 strain levels were increased, as well as the fatigue lives. The PMMA cement usually failed from cracks initiating at the pores either on the surface or internally which is consistent with work by Topoleski *et al.* [23]. Porosity can be attributed to a number of causes, of which air introduced upon mixing is probably the most important. The much

lower porosity of the PEMA may be due to the ease of dissolving the PEMA powder into the nBMA liquid, resulting in a very low viscosity mixture initially, in comparison to the longer mixing required to obtain a thoroughly mixed PMMA bone cement. The differences in ductility of the PEMA and PMMA bone cements are shown in Figs 4 and 5. The fatigue fracture surface of the PEMA was rough in texture, with deformation of the polybutylmethacrylate matrix and PEMA bead pullout, compared to the flat fracture surface obtained for the commercial cement, with fracture through both the PMMA matrix and PMMA bead. Due to these large differences in ductility, the PMMA cement will be a lot more sensitive to the presence of pores compared to the PEMA cement.

The addition of hydroxyapatite particulate to the PEMA bone cement resulted in changes in fatigue resistance as shown in Fig. 3. The highest fatigue lives obtained with the HA addition resulted from the 20 and 30 wt% cements with the 10 and 40 wt% cements failing more easily. Whilst the specimens are being loaded they also undergo creep, this deformation was reduced depending on the amount of hydroxyapatite added. The mechanisms involved in this dynamic creep process will be the subject of future investigations with the PEMA-HA bone cement.

5. Conclusions

The flexural strength and modulus of the PEMA-nBMA bone cement were increased by the addition of up to 40 wt% HA particles. The material still exhibited ductile properties for all HA proportions tested.

At the stress levels used in this investigation, which are higher than those predicted in the cement mantle *in vivo*, the PMMA cement displayed higher fatigue resistance than the PEMA with and without HA reinforcement. However, when the stresses were normalized with respect to the tensile strengths measured for the PEMA and PMMA cements, the PEMA possessed superior fatigue characteristics.

The addition of HA to the PEMA cement reduced the number of cycles to failure obtained compared to PEMA cement specimens without HA, when tested at the same stress levels. The progressive deformation of the specimens during the cyclic loading was however, reduced by the addition of HA to the PEMA bone cement.

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References

1. T. A. GRUEN, G. M. MCNEICE and H. C. AMSTUTZ, *Clin. Orthop. Rel. Res.* **141** (1979) 17–27.
2. R. P. KUSY *J. Biomed. Mater. Res.* **12** (1978) 271–305.
3. H. C. AMSTUTZ and T. A. GRUEN, in "Current practice in orthopaedic surgery", edited by J. P. AHSTROM (The CV Mosby Company, St. Louis, 1973), pp. 158–182.
4. R. M. PILLIAR, R. BLACKWELL, R. MACNAB and R. CAMERON, *J. Biomed. Mater. Res.* **10** (1976) 893–906.
5. S. SAHA and S. PAL, *J. Biomechanics* **17** (1984) 467–478.
6. T. M. WRIGHT and P. S. TRENT, *J. Mater. Sci.* **14** (1979) 503–505.
7. J. PEREK and R. M. PILLIAR, *ibid.* **3** (1992) 333–344.
8. A. SOGAL and S. F. HULBERT, *Bioceramics* **5** (1992) 213–224.
9. A. CASTALDINI and A. CAVALLINI, "Biological and Biomedical Performance of Biomaterials", edited by P. CHRISTEL, A. MEUNIER and A. J. C. LEE. (Elsevier Science Publishers, Amsterdam 1986) pp. 525–530.
10. A. S. LITSKY, R. M. ROSE, C. T. RUBIN and E. L. THRASHER, *J. Orthop. Res.* **8** (1990) 623–626.
11. B. WEIGHTMAN, M. A. R. FREEMAN, P. A. REVELL, M. BRADEN, B. E. J. ALBREKTSSON and L. V. CARLSON, *J. Bone Joint Surg.* **69–B** (1987) 558–564.
12. K. W. M. DAVY and M. BRADEN, *Biomaterials* **12** (1991) 540–544.
13. P. REVELL, M. GEORGE, M. BRADEN, M. FREEMAN and B. WEIGHTMAN, *ibid.* **3** (1992) 84–87.
14. P. REVELL, M. BRADEN, B. WEIGHTMAN and M. A. R. FREEMAN, *Clin. Mater.* **10** (1992) 233–238.
15. J. BEHIRI, M. BRADEN, S. N. KHORISANI, D. WIWATTANDATE and W. BONFIELD, *Bioceramics* **4** (1991) 301–304.
16. S. KHORASANI, S. DEB, J. BEHIRI, M. BRADEN and W. BONFIELD, *Bioceramics* **5** (1992) 225–232.
17. ASTM F451-86 American Society for Testing and Materials, Philadelphia, PA (1986).
18. ISO 178: 1993(E), International Organization for Standardization, Switzerland (1993).
19. BS 2782: Part 9: Method 931A British Standards Institution, London, UK, (1988).
20. R. D. CROWNINSHIELD, R. A. BRAND, R. C. JOHNSTON and J. C. MILROY, *J. Bone Joint Surg.* **62-A** (1980) 68–78.
21. R. D. CROWNINSHIELD, R. A. BRAND, R. C. JOHNSTON and J. C. MILROY, *Clin. Orthop. Rel. Res.* **146** (1980) 71–77.
22. J. P. DAVIES, D. O. O'CONNER, D. W. BURKE, M. JASTY and W. D. HARRIS, *Clin. Orthop.* **229** (1988) 156–161.
23. L. D. T. TOPOLESKI, P. DUCHEYNE and J. M. CUCKLER, *J. Biomed. Mater. Res.* **24** (1990) 135–154.

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