J Mater Sci: Mater Med (2009) 20:633–641
DOI 10.1007/s10856-008-3551-x

Static and fatigue mechanical characterizations of variable diameter fibers reinforced bone cement

Yan Zhou · Weimin Yue · Chaodi Li · James J. Mason

Received: 18 December 2007 / Accepted: 21 July 2008 / Published online: 21 October 2008

Abstract Fibers can be used to improve the mechanical properties of bone cement for the long-term stability of hip prostheses. However, debonding of the fibers from the matrix due to the poor fiber/matrix interface is a major failure mechanism for such fiber reinforced bone cements. In this study, a novel fiber (variable diameter fibers or VDFs) technology for reinforced bone cement was studied to overcome the interface problem of short-fiber composites. These fibers change their diameters along their length to improve the fiber/matrix interfacial bond by the mechanical interlock between the VDFs and the matrix. A novel composite made from novel ceramic VDFs incorporated in PMMA matrix was developed. Both static and fatigue tests were carried out on the composites. Conventional straight fiber (CSF) reinforced bone cement was also tested for comparison purposes. Results demonstrated that both the stiffness and the fatigue life of VDF reinforced bone cement are significantly improved ($P < 0.05$) compared with the unreinforced bone cement. VDF contents of 10% by volume increased the fatigue life over unreinforced bone cement by up to 100-fold. Also, the fatigue life and modulus of toughness of VDF reinforced cement were significantly greater than those of CSF reinforced cement ($P < 0.05$ and $P < 0.001$, respectively). Scanning electron microscopy (SEM) micrographs revealed that VDFs can bridge the matrix cracks effectively and pullout of VDFs results in much more extensive matrix damage than pullout of CSFs increasing the resistance to fatigue. Therefore, VDF reinforced cement was significantly tougher, having a greater energy dissipation capacity than CSF reinforced cement. VDFs added to bone cement could potentially avoid implant loosening due to the mantle fracture of bone cement and delay the need for revision surgery.

Y. Zhou · W. Yue · C. Li · J. J. Mason (✉)
Department of Aerospace and Mechanical Engineering, University of Notre Dame, 365 Fitzpatrick Hall, Notre Dame, IN 46556, USA
e-mail: jmason1@nd.edu; dukeprice@gmail.com

1 Introduction

Poly(methyl methacrylate) (PMMA) bone cement has been the widespread choice in orthopaedics for decades. It has been perceived as the weak link in maintaining the mechanical integrity of cemented total joint arthroplasties [1, 2]. Failure or fracture of the PMMA mantle can lead to loosening and ultimate failure of the prosthesis. Fatigue failure has been shown to be a predominant in vivo failure mode of bone cement, and is now recognized as a critical step leading to oseolysis and eventually aseptic prosthetic loosening [3, 4]. Improvement of the mechanical properties of bone cement, specifically increasing the resistance to fracture and fatigue, remains essential for increasing the longevity of cemented total joint arthroplasties.

The fracture resistance of bone cement can be enhanced by chemical modification or the introduction of a reinforcement phase [5]. The use of a fiber phase is attractive because short fibers can be incorporated into bone cements that are in current clinical use. Different reinforcements have been added to the polymerizing matrix in order to improve the fatigue properties and fracture toughness of the PMMA. These include fibers made of polyethylene [6], hydroxyapatite [7, 8], PMMA [9], Kevlar [10, 11], carbon [12, 13], titanium [14] and steel [15, 16]. These additions have resulted in improvements in mechanical properties over unreinforced PMMA and demonstrate the potential advantages of fiber reinforcement. However, only modest improvements in mechanical properties have been achieved in bone cement [17, 18]. Scanning electron microscopy (SEM) reveals that the poor interfacial properties between the fiber and PMMA matrix result in debonding between them [12, 19]. This is a major failure mechanism for fiber-reinforced bone cement [12, 17]. It suggests that further improvement in the mechanical properties of bone cement may be realized by enhancing the bonding between the fiber and matrix [18]. Chemical treatments and/or fiber coating might improve the interfacial bonding between the fibers and the polymer matrix [20]. However, the toxicity of these chemicals prevents their use in implants. Further, some treatments do not perform well in the aqueous environments, such as the human body [21].

Research shows that composite mechanical properties greatly depend on the fiber shape [22–26]. A stronger load transfer mechanism between the fiber and matrix can be obtained by modifying the morphology or shape of short fiber. One new fiber morphology has recently been proposed in the literature: variable diameter fibers (VDFs), which result from shape optimization maximizing the load transfer between the fiber and matrix [26]. These fibers change their diameters along the length, much like threaded bolts, in order to provide greater mechanical interlocking between the fiber and polymer matrix. The advantages of such fibers as they are described here are multiple. Due to mechanical interlocking, VDFs do not rely solely on shear at the fiber/matrix interface to transfer load from matrix to fiber. It is expected that VDFs can both bridge matrix cracks effectively and improve the mechanical properties of the composite. Consequently, VDFs facilitate an increase in the amount of strengthening attainable for any given fiber volume fraction, length, or degree of orientation, relative to standard straight fibers of fixed diameter. Furthermore, increased strengthening is provided without chemical bonding between fiber and matrix. Therefore, virtually any fiber can be used to reinforce any matrix without the need for coupling agents or surface treatments. Based on economics, processing feasibility and biocompatibility, ceramics, such as zirconia (ZrO_2) and alumina (Al_2O_3), comprise the most logical choices for the reinforcement phase due to a significantly higher stiffness, strength and hardness relative to polymers. Also they are bioinert and already accepted in the orthopaedics community [27]. This investigation explores this strategy in more depth to see if the mechanical properties can be improved enough to warrant use in orthopaedic applications.

The goals of this study were to develop and fabricate a novel composite made from novel ceramic VDFs incorporated in PMMA matrix. Furthermore, both the static mechanical properties and fatigue behavior of this composite were evaluated and compared with the properties of the unreinforced bone cement and conventional straight fibers (CSFs) reinforced bone cement. Finally, the failure mechanisms in the VDFs reinforced bone cement were evaluated using SEM.

2 Materials and methods

2.1 Specimen preparation

A patented fiber processing technology [28, 29] was introduced to control the fiber morphology to attain the optimal fiber shapes. By modifying the fiber production technologies, the high quality zirconia VDFs, as well as CSFs, were successfully produced. Figure 1 shows the characterization of the VDFs. The wavelength of the diameter variation (λ) is about 530 µm. Diameters of enlarged nodes and narrow necks are about 175 and 115 µm, respectively. For CSFs, the average diameter is 120 µm. The continuous fibers were cut with a sharp roller blade into discontinuous filaments of the lengths of 6.35 mm (0.25 in.) resulting in a nominal aspect ratio of 50. The acrylic beads used in this study were poly(methylmethacrylate)-poly stryrene copolymer and contained 0.8% residual BPO. The solid component was provided without radiopacifier (supplied by Zimmer Inc., Warsaw, IN, USA) since zirconia fibers are themselves radiopaque. The liquid component consists mainly of methyl methacrylate (MMA) monomer, with 0.75% by volume *N*,*N*-dimethyl-para-touluidine (DMPT) and 75 ± 10 ppm hydroquinone.

Five groups of tensile test specimens were prepared by varying the fiber morphology (i.e. CSF and VDF) and content (i.e. 2 and 5 vol.%) (Table 1) in order to evaluate the effects of fiber shape and fiber volume percentage on the strength and stiffness of the composite bone cement. Specimens of untreated bone cement were included as controls. The amount of zirconia fibers (density 5.84 gm/cc) to be added to the mixtures was calculated by assuming a complete conversion of monomer to polymer, and assuming that the specific gravity of the monomer (0.936 gm/cc) would be changed to that of the polymer

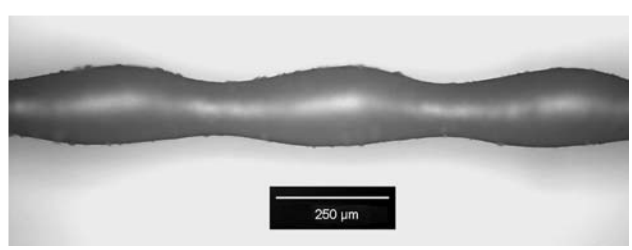

Fig. 1 Characterization of zirconia VDF

Table 1 Ratios of zirconia fiber (GMS), PMMA (GMS) and monomer (CC) used to create tensile specimens

Group	Fiber type	Fiber volume percentage (%)	Ingredient proportions		
			Zirconia (g)	PMMA (g)	Monomer (cc)
1	None	0	0	36	20
2	CSF	2	5.57	36	20
3		5	14.38	36	20
4	VDF	2	5.57	36	20
5		5	14.38	36	20

(1.17 gm/cc) on polymerization. The volume fractions of the reinforcements as well as the ratios of the zirconia, polymer beads, and monomer are tabulated in Table 1. The volume of the PMMA powders to the monomer was kept constant in all cases to facilitate mixing. In preparing tensile specimens, the appropriate weight of short fibers was mixed by gentle stirring of the fibers with the powder phase of the bone cement in an open mixing bowl. Care was taken to ensure that the fibers were thoroughly incorporated into the powder. The liquid MMA monomer was subsequently added to the powder phase of the bone cement. Further mixing was performed for 60 s at 1 Hz at room temperature (23°C) using a Zimmer Quick-Vac Vacuum mixing bowl, under a vacuum of 20–22 in. Hg. After mixing was complete, the cement mass was transferred by gravity flow into a cement cartridge and injected into a polysulphone mold using a commercial cement gun (Power Flo cement injection gun, Zimmer Inc., Warsaw, IN, USA). The tensile specimens used in this study were produced corresponding to ASTM D638-98, Type II [30]. Each group had ten specimens. After injection molding, specimens were cured in an oven at 37°C for 1 h. Specimens were taken out of the oven and removed from the mold before the excess material was trimmed. The samples were stored in a dry environment before testing.

The procedure for preparing fatigue specimens was identical to that for the tensile specimens. Five groups of fatigue specimens reinforced with zirconia fibers were prepared by varying the fiber morphology (i.e. CSF and VDF) and content (i.e. 2 and 10 vol.%) (Table 2). Also, a control group was prepared using untreated bone cement. The fatigue specimens were produced corresponding to ASTM F2118-01a [31]. After machining, all specimens were maintained in 37°C water before testing. Specimen surfaces were as uniform as possible to ensure that material characteristics, rather than preparation techniques, governed the fatigue failure process.

Due to the low viscosity, 5 and 10 vol.% of the fibers could be easily mixed into the cements, though this may not be possible for other formulations. There was an increase in the viscosity of the cements due to the addition of reinforcements. The increase in the viscosity was not significant enough to affect the delivery of the cements through a commercial cement gun (Power Flo cement injection gun, Zimmer Inc., Warsaw, IN, USA). Therefore, the formulations investigated are suitable for clinical use with current instrumentation.

2.2 Static test

Tensile test were performed on an ATS screw-driven universal testing machine (Series 910) within 3 and 14 days of specimen preparation. A crosshead speed of 2.54 mm/min (0.1 in./min), resulting in a strain rate of 7.43e-4 s^{-1} along the gage length, was used to load the samples to failure. A minimum of eight samples per group were tested. The stress at failure and modulus of elasticity were all measured for each test.

2.3 Fatigue test

Fatigue tests were performed on MTS hydraulic test frames. Testing was performed in an aqueous environment of using a specially designed environmental chamber maintained at 37 ± 1°C. Specimens were subjected to uniaxial constant-amplitude fully reversed tension-compression loading (±15 MPa in a sinusoidal cyclic manner), at a frequency of 10 Hz, until fracture. The stress level followed the industry standard, Harris protocol for fatigue of bone cements [32]. The number of stress cycles to failure, N_f, was recorded for each test.

Table 2 Ratios of zirconia fiber (GMS), PMMA (GMS) and monomer (CC) used to create fatigue specimens

Groups	Fiber type	Number of specimens tested	Fiber volume percentage (%)	Ingredient proportions		
				Zirconia (g)	PMMA (g)	Monomer (cc)
1	None	5	0	0	36	20
2	CSF	7	2	5.57	36	20
3		5	10	30.35	36	20
4	VDF	4	2	5.57	36	20
5		8	10	30.35	36	20

Tensile and fatigue test results were statistically analyzed using analysis of variance (ANOVA) techniques and Tukey-Kramer HSD method to determine where statistically significant differences exist. Data were analyzed using JMP (JMP IN 5.1, SAS Institute Inc., Cary, NC, USA) with a significance level of $P = 0.05$.

Representative fracture surfaces from the specimens were analyzed using SEM. Each specimen was cut ~0.5 cm below the fracture surface. Fracture surfaces were coated with gold by spectra deposition. Specimens were imaged using a field emission scanning electron microscope (Model 4500, Hitachi Field Emission microscope) operated at an accelerating voltage of 30 kV.

3 Results

3.1 Static test

X-ray radiographs of zirconia fiber reinforced tensile specimens show that the fibers were completely and uniformly incorporated into the bone cement during the mixing process. The results of tensile tests for zirconia fiber reinforced bone cement are shown in Figs. 2–4. There was a significantly greater ($P < 0.05$) elastic modulus for 5% CSF, VDF reinforced cements compared with the 2% CSF, VDF reinforced cement (Fig. 2). The elastic modulus of the 2% CSF, VDF reinforced cement was significantly greater ($P < 0.05$) than that of the unreinforced bone cement (control). The elastic modulus increased by 73 and 168% for cements reinforced with 2 and 5% CSF, and increased by 46 and 152% for cements reinforced with 2 and 5% VDF, respectively. However, the differences between the CSF and VDF reinforced bone cements at either 2 or 5% by volume were not significant ($P > 0.05$). The ultimate strengths of all groups were shown in Fig. 3. The ultimate strength was significantly greater ($P < 0.05$) for the VDF reinforced cement compared to the CSF reinforced cement at 2 vol.%. However, the difference between CSF and VDF reinforced cements at 5 vol.% was not statistically significant ($P > 0.05$). Overall, the ultimate strengths were not statistically different for the groups reinforced with 2% VDFs, 5% CSFs, 5% VDFs and the control group. The average ultimate strength of bone cement reinforced with 2% CSFs by volume was lower than the control cement, but the difference was not statistically significant different ($P > 0.05$). Figure 4 showed the modulus of toughness of the reinforced bone cement. The modulus of toughness was significantly greater ($P < 0.001$) for VDF reinforced cement compared to the CSF reinforced cement at both 2 and 5 vol.%.

Typical fracture surfaces for the tensile specimens of VDF and CSF reinforced bone cement are shown in Fig. 5. For both VDF and CSF reinforced cement, fiber fracture dominated the fracture characteristics instead of fiber pull-out, which indicated that the fiber strength was not sufficient to allow fiber pull-out. Although, fiber fracture dominated the fracture characteristic, fiber/matrix interface

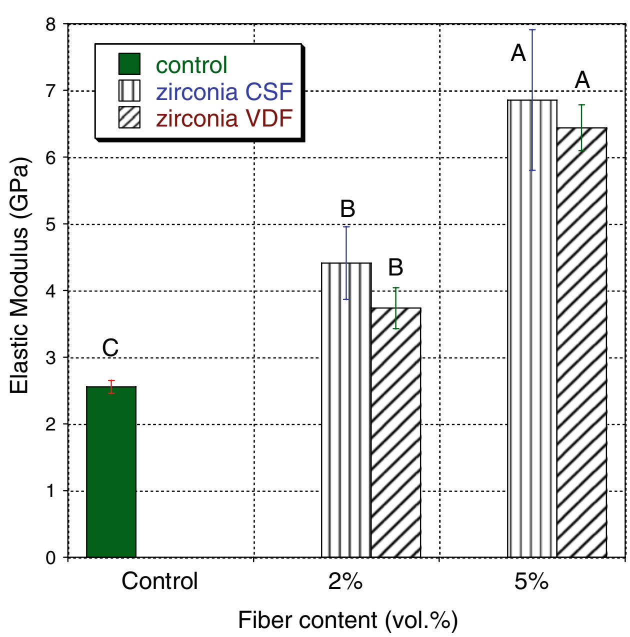

Fig. 2 Elastic modulus of unreinforced bone cement (control), bone cement reinforced with 2 and 5% zirconia CSFs and VDFs, respectively (levels not connected by same letter are significantly different, $P < 0.05$)

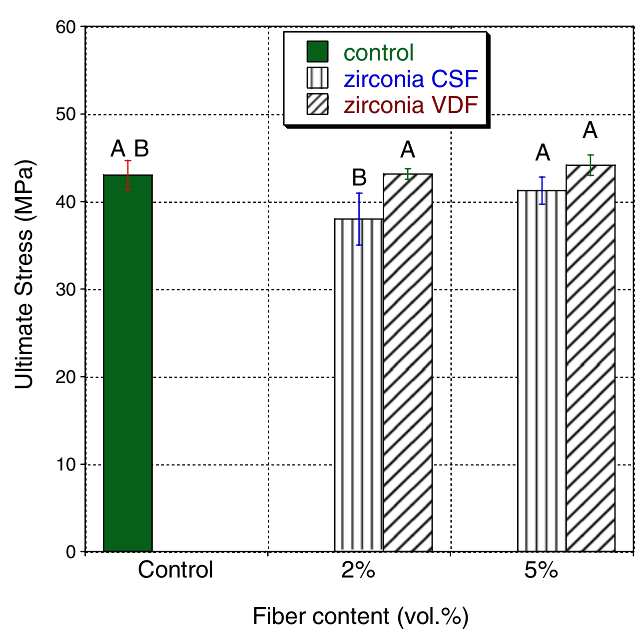

Fig. 3 Ultimate strength of unreinforced bone cement (control), bone cement reinforced with 2 and 5% zirconia CSFs and VDFs, respectively (levels not connected by same letter are significantly different, $P < 0.05$)

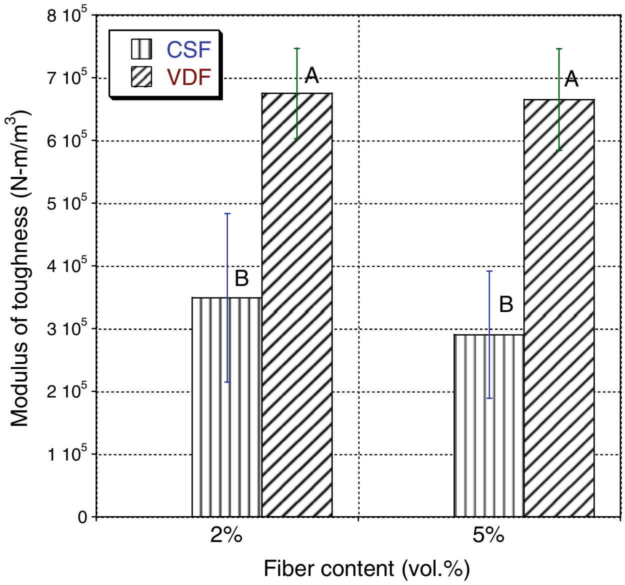

Fig. 4 Modulus of toughness of bone cement reinforced with 2 and 5% zirconia CSFs and VDFs, respectively (levels not connected by same letter are significantly different, $P < 0.05$)

debonding can be observed for both CSF and VDF composites under higher magnification, as shown in Fig. 6. Both the CSF and VDF were debonded from the matrix at the fiber/matrix interfaces. Compared with CSF, the fiber debonding length of VDF appeared shorter due to the mechanical interlock provided by the fiber morphology. Furthermore, Fig. 6a shows that portion of the matrix adhered to the fiber surface after fracture. This indicates that some bonding occurred between the fiber and the cement matrix. Such bonding may be related to the irregularities in the fiber surface, which cause micro mechanical interlock between the fiber and matrix.

3.2 Fatigue test

The fatigue results for the unreinforced bone cement (control), bone cement reinforced with 2 and 10% zirconia CSFs and VDFs are shown in Fig. 7. In the statistical analysis, the two run-out specimens after four million cycles of bone cement reinforced with 10% VDFs by volume were not included, and thus the analysis

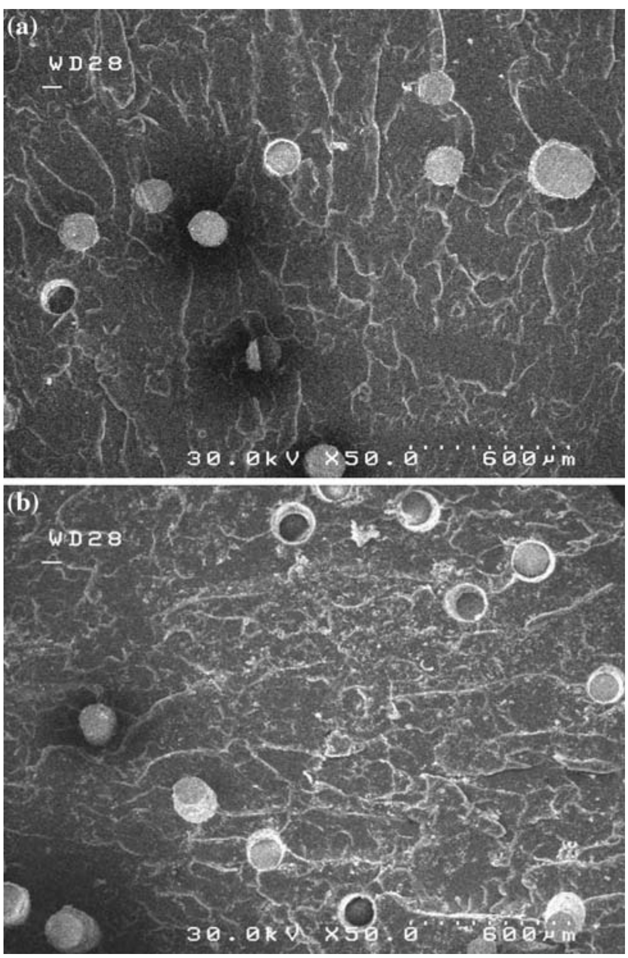

Fig. 5 SEM micrographs showing typical fracture surfaces for tensile specimens of (**a**) CSF reinforced cement and (**b**) VDF reinforced cement

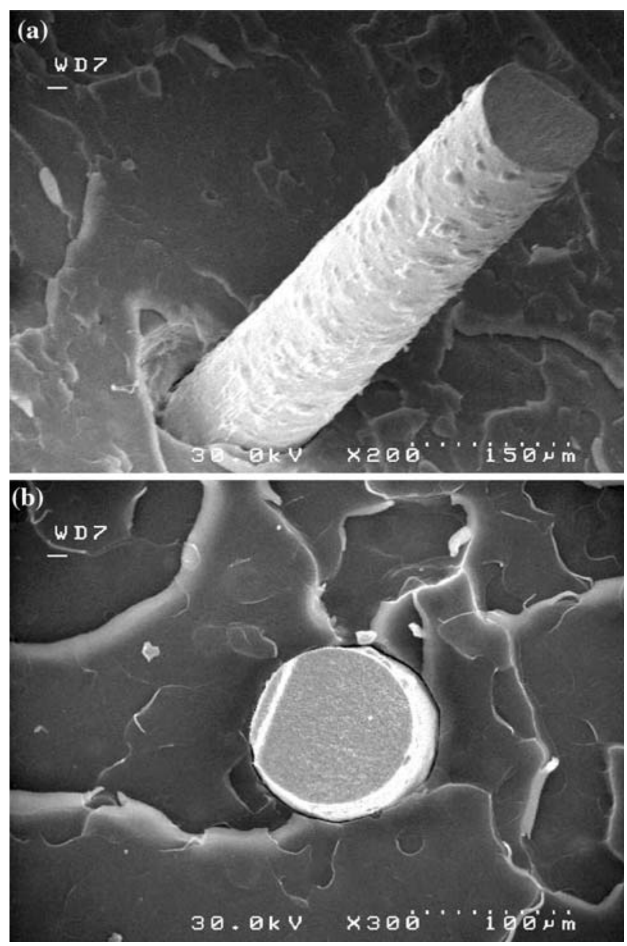

Fig. 6 SEM micrographs showing the fiber debonding lengths of (**a**) CSF and (**b**) VDF

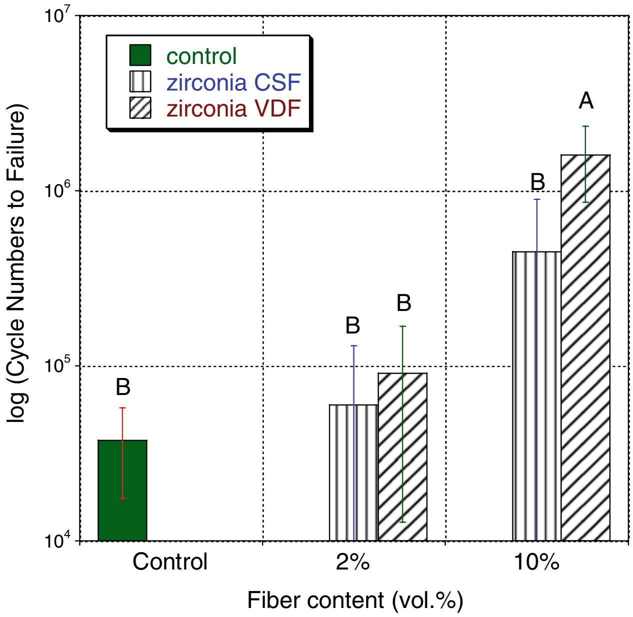

Fig. 7 Number of cycles to failure of unreinforced bone cement (control), bone cement reinforced with 2 and 10% zirconia CSFs and VDFs, respectively (levels not connected by same letter are significantly different, $P < 0.05$)

underestimates the actual fatigue data of the bone cement reinforced with 10% VDFs. The fatigue life of the reinforced groups was much higher than control bone cement. The number of cycles to failure increased by 59 and 141% for cements reinforced with 2% CSF and 2% VDF, respectively. Although the fatigue life of bone cement reinforced with 2% VDF appears higher than that of the cement reinforced with 2% CSF, there was no statistically significance ($P > 0.05$) due to the wide scatter in the fatigue results. The fatigue life was dramatically increased when the fiber volume percentage was increased to 10%. There was approximately an order of magnitude increase in fatigue life for the bone cement reinforced with CSFs over the control cement. Compared with the 10% CSF reinforced bone cement, the fatigue life of 10% VDF reinforced cement was about four times higher. The number of cycles to failure is significantly greater ($P < 0.05$) for the 10% VDF reinforced bone cement compared to the 10% CSF reinforced cement, which shows the advantage of VDFs.

Figure 8 showed typical fracture surface for the fatigue specimens of 10% VDF reinforced bone cement. The fibers are closely spaced to each other but they appear to have been uniformly dispersed into the PMMA matrix. The fracture surface was very rough. As the crack propagated, a few VDFs bridging the crack were pulled out, which resulted in extensive matrix damage (Fig. 8b), consuming large amounts of energy. Figure 9 showed the typical fracture surface of 10% CSF reinforced cement. The fracture surface

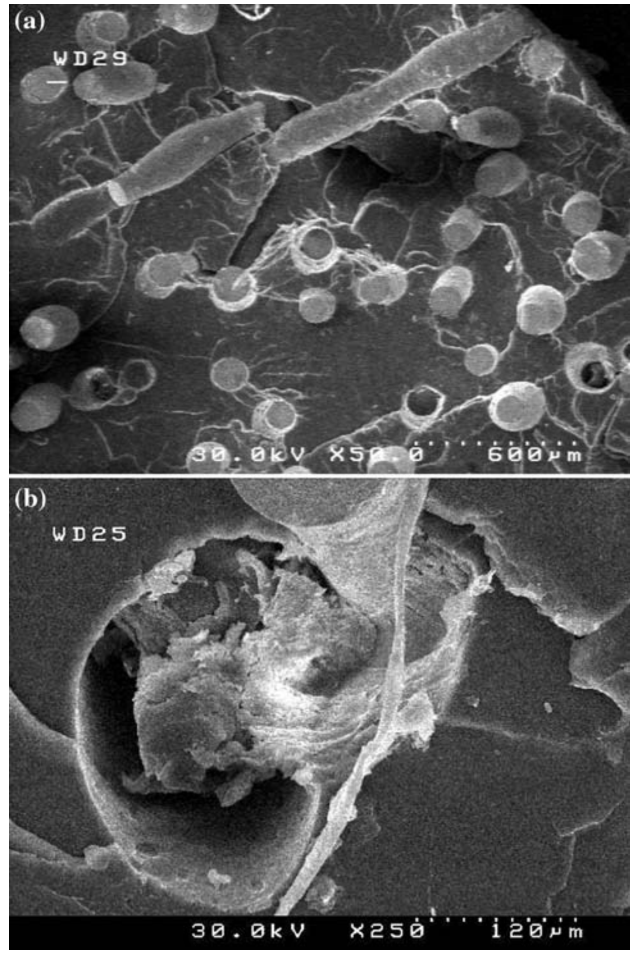

Fig. 8 SEM micrographs showing typical fracture surfaces for fatigue specimens of the 10% VDF reinforced bone cement

was smoother compared with the fracture surface of VDF reinforced bone cement. Therefore, the crack was able to propagate with less resistance for longer distances, leaving behind a flat fracture surface. Once the crack reached a critical size, it propagated through the entire cross-section of the specimen, causing fiber fracture or pullout, and leaving a relatively flat fracture surface with river marks, similar to the fracture surface of a brittle material. In contrast to VDF, the pullout of CSF did not result in significant matrix damage (Fig. 9b), which means less energy was consumed during the fracture process.

4 Discussion

4.1 Tensile test

The tensile tests demonstrated that incorporation of zirconia fibers significantly increased the elastic modulus of bone cement. The elastic modulus of fiber reinforced bone cement also increased with increased fiber content.

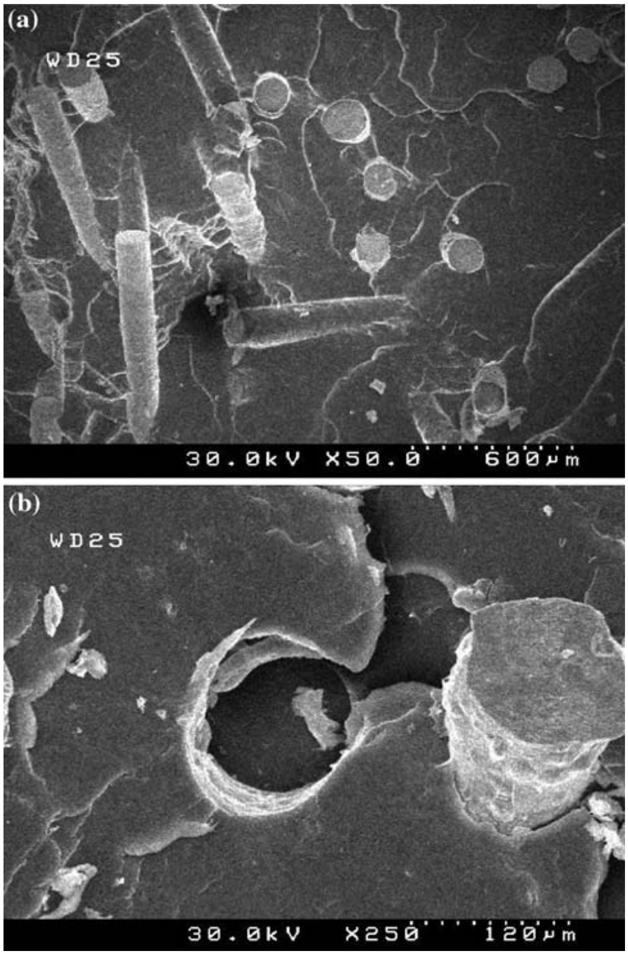

Fig. 9 SEM micrographs showing typical fracture surfaces for the fatigue specimens of the 10% CSF reinforced bone cement

However, the tensile strength of the fiber-reinforced cement was not significantly improved when compared with the control bone cement and the total failure strain was greatly decreased for the fiber-reinforced cement, especially for CSF reinforced cement. This is most likely due to the increased local stresses caused by fibers, which can be considered rigid inclusions in an elastic matrix [33]. The stress concentrations may have promoted crack initiation. Once the crack initiated, the stress intensity was so large that potential fiber reinforcing effects may have been overwhelmed [34]. This can also explain the reduction in strength of bone cement reinforced with 2% CSFs, in contrast to that of the control group. Compared with the CSF reinforced bone cement, the total strain to failure of the VDF reinforced cement was much higher (about two times) for both 2 and 5% groups. Corresponding to the higher total strain to failure, VDF reinforced cement has a significantly greater the modulus of toughness at either 2 or 5 vol.% (Fig. 4). More energy is needed to completely fracture VDF reinforced cement than to fracture CSF reinforced cement. Outside of the control group, VDF reinforced bone cement had the best tensile properties with the higher elastic modulus and failure strain.

There were almost no significant differences exist between tensile tests for the CSF and the VDF reinforced bone cements at either 2 or 5 vol.%, except the ultimate strength of the CSF and the VDF reinforced bone cement at 2 vol.%. The lack of a difference was likely because the fiber itself was not strong enough. The fracture surfaces showed that the failure mechanism in tensile tests for both CSF and VDF reinforced cements was fiber fracture. Although fiber/matrix debonding was observed for CSF reinforced bone cement, most CSFs fractured before being pulled out from the matrix, which shows that CSFs are weak (Fig. 5a). The failure mechanism was the same in VDF reinforced bone cement, although the fiber/matrix debonding distance was much shorter due to the mechanical interlock between the fiber and the matrix. Most VDFs also fractured before being pulled out of the matrix (Fig. 5b). VDFs are optimal for composites with a weak fiber/matrix interfacial bond, and where fiber pullout is the major failure mechanism. In this case, significant improvement is expected for VDF reinforced composites due to improved load transfer between the VDF and matrix, and the change in failure mechanism. This of course assumes the fiber is itself strong enough to resist failure. In the current study, the advantage of VDFs in static loading was masked by the weak fiber strength for both CSFs and VDFs. The fibers fractured before pull-out could contribute to strength. Implicit in this analysis is the assumption that the bonding between fiber and matrix is weak. The surface of the CSFs is very rough, which can cause micromechanical interlock between the CSF and the matrix. This mechanism was observed by bone cement adhering to the CSF surface after fracture (Fig. 6a). Bonding between the CSF and matrix also weakened the CSFs and VDFs reinforced materials by strengthening the bond between fiber and matrix.

4.2 Fatigue test

The fatigue tests demonstrated that incorporation of zirconia fibers increased the fatigue life of bone cement. Significant increases in the fatigue life of bone cement were realized by increasing the fiber content. For 2% fiber reinforced bone cement, although the fatigue life was much higher than that of the control cement, there was no significant increase due to the wide scatter in the results. There was also no significant difference between the VDF reinforced cement and CSF reinforced cement. SEM fractographs of fiber reinforced bone cement showed that the fracture occurred at different planes forming a pattern of river-like facets. Thus, fibers diverted crack propagation and absorbed more energy. However, due to the low fiber content, the crack was still able to propagate easily for a long distance before being inhibited by the fibers, leaving behind a flat fracture surface, which looked similar to the fracture surface of the control bone cement. Therefore, the fatigue life was not significantly improved when the fiber content was relatively low. As the fiber content was increased to 10 vol.%, there was approximately an order of magnitude or more increase in fatigue life for the CSF reinforced cement. Also, significant improvement in fatigue life was found for th eVDF reinforced cement compared with the CSF reinforced cement leading to a 100-fold increase in life from the control cement to the VDF cement. Compared with the control cement and 2% fiber reinforced cement, the fracture surface of 10% VDF reinforced cement was much rougher. The main crack propagated by coalescing with many smaller cracks. The smaller cracks were often not on the same plane as the main crack, resulting in a very rough fracture surface. Although SEM fractographs showed that the fiber/matrix interfacial bond was weak, most VDFs were kept anchored inside the matrix due to the more efficient mechanical interlock, which caused the diversion of the crack out of the original crack plane. In contrast to VDFs, more CSFs were pulled out due to the poor interfacial bond between the fiber and matrix and lack of sufficient mechanical interlock, which provided less resistance to crack propagation and poor bridging ability. Pullout of VDFs resulted in much more extensive matrix damage than that of the pullout of CSFs (Figs. 8b and 9b). This means that more energy was consumed for VDFs than CSFs during the pullout process. The results, along with the fracture morphology, clearly show that the VDF reinforced cement is significantly more resistant to fatigue, with a much greater energy dissipating capacity than CSF reinforced cement.

Fiber reinforcement affects the fatigue crack propagation phase of failure in bone cement, enhancing the fatigue crack propagation resistance. In general, the increased fatigue life seen in the VDF reinforced cement was the result of several deformation mechanisms operating in these specimens. These include increased load transfer at the fiber/matrix interface, and diversion of the crack out of the original crack plane by fiber splitting and fiber/matrix interface failure, both of which divert the crack from its fracture plane and increase the energy required for crack propagation. These modes of failure were seen in all of the VDF reinforced cement tested in this study.

In this study, the main failure mechanism was fiber fracture. This shows that the energy absorption capacity of fiber-reinforced cement could be increased by inducing more fiber pullout, which could be realized by improving fiber strength.

5 Conclusions

A VDF reinforced bone cement was developed. Both the static and fatigue mechanical properties were evaluated and compared with the properties of the unreinforced bone cement and a CSF reinforced bone cement. Results demonstrated that both the stiffness and the fatigue life of the VDF reinforced bone cement are significantly improved ($P < 0.05$) compared with the unreinforced bone cement. VDF contents of 10 vol.% increased the fatigue life over unreinforced bone cement by up to 100-fold. Also, the fatigue life and modulus of toughness of VDF reinforced cement was significantly greater than those of the CSF reinforced cement ($P < 0.05$ and $P < 0.001$, respectively). SEM micrographs revealed that VDFs can bridge the matrix cracks effectively and pullout of VDFs results in much more extensive matrix damage than the pullout of CSFs increasing the resistance to fatigue. Therefore, VDF reinforced cement was significantly tougher, having a greater energy dissipation capacity than CSF reinforced cement.

This study showed the feasibility of a novel fiber (VDF) technology for reinforced polymers. This fiber family significantly improved the fatigue life of bone cement at a very high level of reliability. Bone cements reinforced with VDFs could potentially inhibit implant loosening due to the mantle fracture of bone cement and delay the need for revision surgery.

Acknowledgements The authors express their gratitude to Zimmer Inc. and Advanced Cerametrics Inc. for donations of some of the materials used in this study. The help of Mr. Don Yakimicki is greatly appreciated. This research is supported by the Indianan 21st Century Research and Technology Fund, the State of Indiana and NSF Grant # DMI-0419671.

References

1. M. Jasty, W.J. Maloney, C.R. Bragdon, D.O. Oconnor, T. Haire, W.H. Harris, J. Bone Joint Surg. Br. **73**, 551 (1991)
2. M. Spector, Orthop. Clin. North Am. **23**, 211 (1992)
3. L.D.T. Topoleski, P. Ducheyne, J.M. Cuckler, J. Biomed. Mater. Res. **24**, 135 (1990). doi:10.1002/jbm.820240202
4. S.P. James, M. Jasty, J. Davies, H. Piehler, W.H. Harris, J. Biomed. Mater. Res. **26**, 651 (1992). doi:10.1002/jbm.820260507
5. S. Deb, J. Biomater. Appl. **14**, 16 (1999)
6. B. Pourdeyhimi, H.D. Wagner, J. Biomed. Mater. Res. **23**, 63 (1989). doi:10.1002/jbm.820230106
7. E.J. Harper, J.C. Behiri, W. Bonfield, J. Mater. Sci. Mater. Med. **6**, 799 (1995). doi:10.1007/BF00134320
8. S. Shinzato, M. Kobayashi, W.F. Mousa, M. Kamimura, M. Neo, Y. Kitamura et al., J. Biomed. Mater. Res. **51**, 258 (2000). doi:10.1002/(SICI)1097-4636(200008)51:2<258::AID-JBM15>3.0.CO;2-S
9. J.L. Gilbert, S.S. Net, E.P. Lauthenschlager, Biomaterials **16**, 1043 (1995). doi:10.1016/0142-9612(95)98900-Y
10. B. Pourdeyhimi, H.D. Wagner, P. Schwartz, J. Mater. Sci. **21**, 4468 (1986). doi:10.1007/BF01106573

11. T.M. Wright, P.S. Trent, J. Mater. Sci. **14**, 503 (1979). doi:10.1007/BF00589852
12. R.M. Pilliar, R. Blackwell, I. Macnab, H.U. Cameron, J. Biomed. Mater. Res. **10**, 893 (1976). doi:10.1002/jbm.820100608
13. S. Saha, S. Pal, J. Biomed. Mater. Res. **17**, 1041 (1983). doi:10.1002/jbm.820170613
14. L.D.T. Topoleski, P. Ducheyne, J.M. Cuckler, J. Biomed. Mater. Res. **26**, 1599 (1992). doi:10.1002/jbm.820261206
15. S. Saha, M.J. Kraay, J. Biomed. Mater. Res. **13**, 443 (1979). doi:10.1002/jbm.820130309
16. S.P. Kotha, C. Li, S.R. Schmid, J.J. Mason, J. Biomed. Mater. Res. A **70A**, 514 (2004). doi:10.1002/jbm.a.30107
17. R.P. Robinson, T.M. Wright, A.H. Burstein, J. Biomed. Mater. Res. **15**, 203 (1981). doi:10.1002/jbm.820150208
18. S. Saha, S. Pal, J. Biomed. Mater. Res. **18**, 435 (1984). doi:10.1002/jbm.820180411
19. W. Krause, R.S. Mathis, J. Biomed. Mater. Res. A **22**, 37 (1988)
20. J.M. Yang, P.Y. Huang, M.C. Yang, S.K. Lo, J. Biomed. Mater. Res. **38**, 361 (1997). doi:10.1002/(SICI)1097-4636(199724)38:4<361::AID-JBM9>3.0.CO;2-M
21. K.L. Ohashi, R.H. Dauskardt, J. Biomed. Mater. Res. **51**, 172 (2000). doi:10.1002/(SICI)1097-4636(200008)51:2<172::AID-JBM5>3.0.CO;2-Y
22. R.C. Wetherhold, F.K. Lee, Compos. Sci. Technol. **61**, 517 (2001). doi:10.1016/S0266-3538(00)00217-7
23. N. Phan-Thien, Fiber Sci. Technol. **14**, 241 (1981). doi:10.1016/0015-0568(81)90016-6
24. Y.T.T. Zhu, I.J. Beyerlein, Mater. Sci. Eng. A Struct. **326**, 208 (2002)
25. Y. Zhou, C.D. Li, J.E. Renaud, J.J. Mason, Eng. Optim. **37**, 121 (2005). doi:10.1080/03052150412331298399
26. Y. Zhou, C.D. Li, J.J. Mason, Mater. Sci. Eng. A Struct. **393**, 374 (2005)
27. J.B. Park, *Biomaterials* (Plenum Press, New York, NY, 1979)
28. R.B. Cass, R.R. Loh, T.C. Allen, U.S. Patent 5,827,797, 1998
29. R.B. Cass, Am. Ceram. Soc. Bull. **70**, 424 (1991)
30. ASTM Standard D638-98, *Annual Book of ASTM Standards* (ASTM, PA, USA, 1998), p. 45
31. ASTM Standard F2118-01a, *Annual Book of ASTM Standards* (ASTM, PA, USA, 2001), p. 1673
32. J.P. Davies, D.O. O'connor, J.A. Greer, W.H. Harris, J. Biomed. Mater. Res. **21**, 719 (1987). doi:10.1002/jbm.820210604
33. J.D. Eshelby, Proc. R. Soc. Lond. A Mater. **241**, 376 (1957)
34. L.D.T. Topoleski, P. Ducheyne, J.M. Cuckler, J. Biomed. Mater. Res. **29**, 299 (1995). doi:10.1002/jbm.820290304