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# Static shear strength between polished stem and seven commercial acrylic bone cements

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**Abstract** The stem–cement interface is one of the most significant sites in cemented total hip replacement and has long been implicated in failure of the whole joint system. However, shear strength at this interface has rarely been compared across a range of commercially available bone cements. The present study seeks to address this issue by carrying out a comparative study. The results indicated that the static shear strength was more dependent on cement type than cement viscosity and volume. However, both cement type and viscosity were contributory factors on porosity and micropore size in the cement surface. There was no significant difference between Simplex P and Simplex P with Tobramycin. Although the bone cements were all hand mixed in this study, the static shear strength was significantly larger than the values recorded by other researchers, and the porosity and micropore size showed much lower values. Bone cement transfer films were detected on the stem surface, typically about 4–10 μm thick. They were considered to be an important factor contributing to high friction at the stem–cement interface after initial debonding.

## Introduction

Acrylic bone cement has been used in cemented total hip replacement (THR) for more than 40 years, the primary functions of which are as an intermediary material between the prosthesis and the bone to stabilise the femoral stem and to transfer physiological loading of the patient during normal activities [1]. Commercial bone cement is typically supplied as two components: a fine powder consisting of pre-polymerised polymethylmethacrylate (PMMA) or PMMA-based copolymers, benzoylperoxide (BPO) as an initiator for polymerisation reaction, a radiopaque agent commonly barium sulphate ($BaSO_4$) or zirconium dioxide ($ZrO_2$), and a vial of liquid composed of Methylmethacrylate (MMA) monomer, *N,N*-dimethyl-*p*-toluidine (DMPT) as an activator for polymerisation reaction and hydroquinone (HQ). Upon mixing the powder and liquid, a dough is formed which is then introduced manually or under mechanical pressure into the bone cavity [2]. Despite clinical application for many years, such problems as thermal necrosis due to exothermic reaction of polymerisation and chemical necrosis as a result of unreacted MMA monomer remain unsolved. "Modern cementing techniques" have been reported to significantly reduce porosity at the stem–cement–bone interfaces as well as in the bulk matrix [3]. Figure 1 displays a scanning electron micrograph (SEM) of Cemfix 3 bone cement surface, where micropores are formed after polymerisation. These micropores are considered to play a critical role in fatigue crack generation and propagation in the cement mantle, and in subsequent aseptic loosening and malfunctioning of cemented THR [4]. The stem–cement interface has consistently been cited as a weak link due to the absence of chemical bonding, and is often a fundamental factor in premature failure of THR [5, 6].

There are many brands of PMMA bone cement commercially available, all are similar in composition but have inherently different characteristics such as viscosity, porosity and mechanical properties during and following

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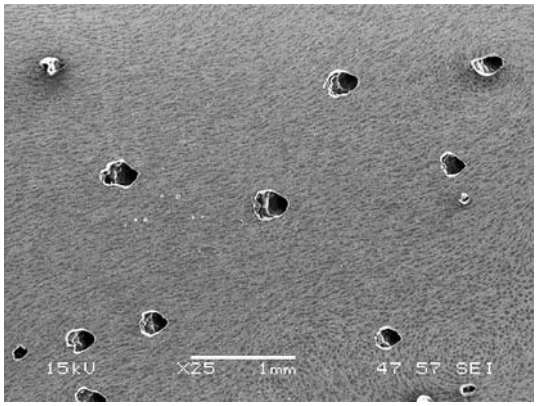


**Fig. 1** Micropores on Cemfix 3 bone cement surface, measured by SEM (JEOL JSM 6060 LV, Oxford Instruments)

polymerisation [7, 8]. Many studies have concentrated on mechanical properties of bone cement and the bond strength at the stem–cement–bone interfaces [9–11], in which some influencing factors, such as stem surface roughness, pre-heating the stem, pre-chilling the cement, cementing techniques and bone interdigitation are investigated. The most commonly used method to test bond strength is a push/pull out test, where a metallic rod is often employed to represent the femoral stem [12–14]. Finite element analysis (FEA) has also been used to simulate the physical conditions at the stem–cement interface, from which it is demonstrated that debonding at this interface is primarily dominated by shear failure [15]. To date however, few comprehensive studies have been performed to establish comparative data for the major clinical bone cements. Therefore, such a study would be particularly useful if it determined the shear strength at the stem–cement interface, taking into consideration its contribution to debonding of this interface, and by extension to the survivorship of cemented THR. There are quantities of femoral stems prevalent on the market for the surgeons to choose from, which vary in geometrical design, material and surface finish. Nowadays, there remains controversy as to the optimum stem surface finish, mainly focusing on whether matt stems can form permanent fixation at the stem–cement interface. It is generally accepted that matt stems can provide a greater bond strength due to enhanced bone cement integration, whereas polished stems form a much lower bond strength, hence with a higher probability of micromotion [16, 17]. It has been well documented that matt stems do give satisfactory results at 10–20 years [18]. However, clinical data has also shown significant superiority for some stem designs with a highly polished surface finish, e.g. the Exeter stem, whereas the matt surface finish has been proven to be less successful [19]. The great success of highly polished Exeter stem has been attributed to the fact that this specific stem is designed to subside within the cement mantle to promote stability and the polished surface finish aids this process. All these confusing clinical data and laboratory-based mechanical strength results require a deeper insight into the mechanical characteristic at stem–cement interface. This current study therefore aims to comprehensively investigate the interaction and the static shear strength between polished femoral stems and the major clinical bone cements through pull out tests, and to obtain a better understanding of the contributory factors such as cement type and viscosity. An additional factor—cement volume is also studied to ascertain whether or not more heat involved in a larger cement volume will significantly influence the static shear strength at the stem–cement interface.

## Materials and methods

In this present study, the static shear strength between polished femoral stem and seven commercially available PMMA bone cements was investigated. The details of the bone cements are shown in Table 1. For each cement type, two kinds of stainless steel rods were manufactured, enabling different cement volume sets to be tested. The stainless steel rods were highly polished to obtain a surface roughness of Sq about 10 nm, measured by an optical interferometry at ×50 magnification, this value being directly comparable to commercial polished femoral stems, Fig. 2a. A cylindrical holder made of mild steel was fabricated for the bone cement to be poured into, leaving a cement mantle thickness of 7 mm and 9 mm for the two diameters of rods (12 mm and 8 mm) respectively, Fig. 2b. The stainless steel rod was fixed using a milling machine chuck, which ensured accurate axial alignment of the rod within the cement mantle, Fig. 3. The bone cements were all hand mixed at room temperature, according to the manufactures' instructions. A metallic ring was connected to the cylindrical holder by screws in order to apply pressure on the cement during polymerisation process and to make sure that the interfacial failure will not occur between the bone cement and the cylindrical holder. The specimen was laid aside for 24 h to fully cure before being tested on a Hounsfield Test Machine H20K-W, Fig. 4. The load–displacement plot for the pull out test was then recorded. All the tests were performed at a constant speed of 2 mm/min by displacement control. Repeated tests were carried out five times for each cement type and each cement volume to provide statistical viability. The stainless steel rods were repolished after each test to ensure identical surface finish grade.

After each pull out test, the bone cement was cautiously extracted from the cylindrical holder and cut longitudinally

**Table 1** Relative viscosity and composition of the seven commercial PMMA bone cements

| Bone cements | Viscosity | Powder (w/w) | Liquid (w/w) | Suppliers |
|---|---|---|---|---|
| Cemfix 3 | Low | PMMA—87.6; BPO—2.4; $BaSO_4$—10 | MMA—84.4; DMPT—2.4; BMA—13.2; HQ—20 ppm | Teknimed S.A., France |
| Coriplast 3 | Low | PMMA—45; PMMA/MA—45; $BaSO_4$—10 | MMA—98; DMPT—2; HQ—45 ppm | Corin Medical Ltd., UK |
| Simplex P | Medium | PMMA—15; PMMA/ST—75; $BaSO_4$—10 | MMA—97.4; DMPT—2.6; HQ—60 ppm | Howmedica International Inc., Ireland |
| Simplex P-T | Medium | PMMA—15; PMMA/ST—75; $BaSO_4$—7.5; T—2.5 | MMA—97.4; DMPT—2.6; HQ—60 ppm | Howmedica International Inc., Ireland |
| CMW 3 | Medium | PMMA—88; BPO—2; $BaSO_4$—10 | MMA—97.5; DMPT—2.5; HQ—25 ppm | DePuy International Ltd., UK |
| CMW 1 | High | PMMA—88.85; BPO—2.05; $BaSO_4$—9.1 | MMA—99.18; DMPT—0.82; HQ—25 ppm | DePuy International Ltd., UK |
| Palacos R | High | PMMA/MA—84.25; BPO—0.75; $ZrO_2$—15; C—200 ppm | MMA—97.87; DMPT—2.13; HQ—64 ppm; C—267 ppm | Biomet Merck Ltd., UK |

*Note*: PMMA/MA—Polymethylmethacrylate/methylacrylate; BMA—Butylmethacrylate; PMMA/ST—Polymethylmethacrylate/styrene; T—Tobramycin; C—Chlorophyll

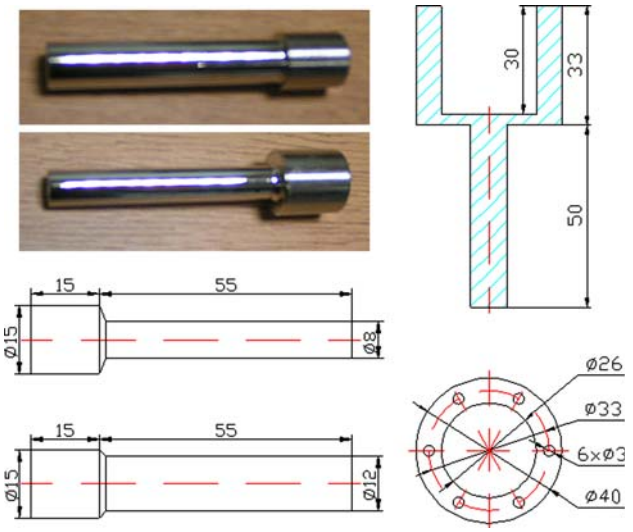


**Fig. 2** (**a**) Design of stainless steel rod (**b**) Design of cylindrical holder

into two equal parts. The inner surface of bone cement was cleaned with alcohol and stained using red dye to enable observation of porosity with a Leica optical

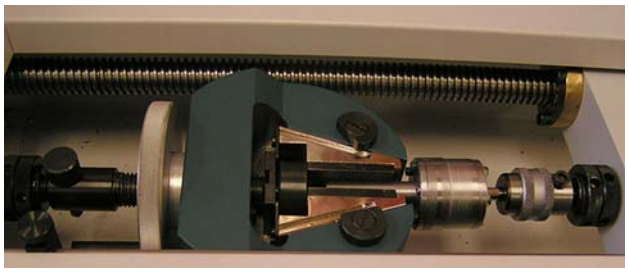

**Fig. 4** Pull out test set up on a Hounsfield Test Machine H20K-W

stereomicroscope MZ6. Totally 10 images were taken arbitrarily on the cement surface at ×10 magnification, with each imaged area being 4 $mm^2$. All the images were processed using Matlab and the micropores were recognised based on grey scale thresholds, Fig. 5. For each image, the porosity was determined by the ratio of the area of micropores to the area of whole cement surface. The number of micropores on the image was calculated, thus the mean area of one micropore could be obtained. The micropore size was then calculated as the diameter of the


**Fig. 3** Specimen set up using a milling machine chuck to ensure axial positioning

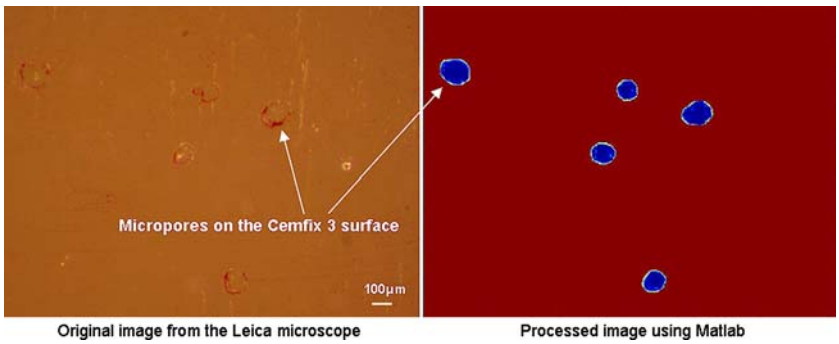


**Fig. 5** Micropores detection using Matlab

micropore, assuming it to be a perfect circle. Finally, the mean value of porosity and micropore size was calculated based on the 10 images for each test.

The static shear strength was determined using the initial debonding force divided by the real surface contact area:

$$\sigma = \frac{F}{\pi DL(1-\eta)}$$

where $F$ is the initial debonding force; $D$ is the rod diameter; $L$ is the internal length of rod within the cement mantle; $\eta$ is the porosity of the cement. The final static shear strength for each cement type and each cement volume was the mean value of the five tests carried out.

A three-way analysis of variance (ANOVA) was performed to investigate the influencing factors, i.e. cement volume, cement viscosity and cement type, on the static shear strength, porosity and micropore size for Cemfix 3, Coriplast 3, Simplex P, CMW 3, CMW 1 and Palacos R bone cements, Fig. 6. An unpaired student $t$-test was employed for Simplex P and Simplex P with Tobramycin (Simplex P-T) bone cements to establish the effect of the antibiotic additive on the results. The software used is SPSS 12.0 for windows.

## Results

The static shear strength, porosity and micropore size (mean value and range) for each cement type and each cement volume are shown in Figs. 7, 8, 9, 10, 11, 12. It was evident by direct observation that there was no determinate relationship between the static shear strength and cement viscosity. It was however more dependent on cement type. In addition, the porosity showed much lower values for bone cements with low viscosity possibly because of greater facilitation for the air bubbles to escape from the bulk matrix, but there seemed to be no significant difference for bone cements with medium and high viscosity. The micropore size also appeared to be determined by cement type.

It was revealed from the three-way ANOVA that the static shear strength was not significantly influenced by cement volume and viscosity ($p > 0.01$), whereas there was significant difference among various cement types ($p < 0.01$). For porosity and micropore size, cement volume was again not a crucial factor ($p > 0.01$), while they were significantly influenced by cement viscosity and cement type ($p < 0.01$). The unpaired student $t$-test demonstrated that there was no significant difference in terms of static shear strength, porosity, and micropore size between Simplex P and Simplex P-T for both of the two cement volume sets ($p > 0.05$), indicating that an addition of tobramycin to the cement composition was not a contributory factor to the results.

The final mean static shear strength for each cement type is shown in Table 2, from which it was clear that

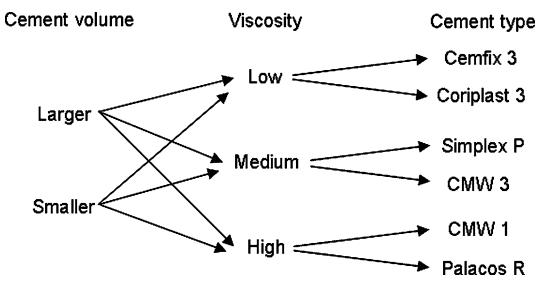


**Fig. 6** The model of three-way ANOVA

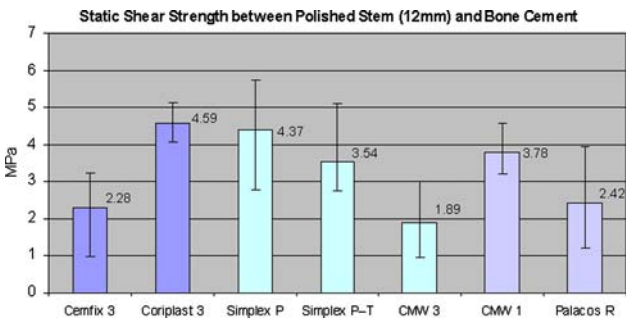


**Fig. 7** Histogram showing static shear strength for the larger cement volume set

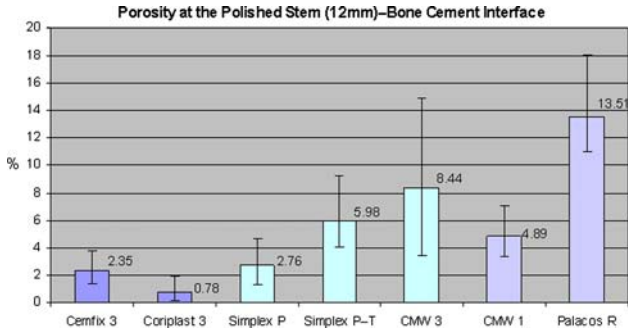


**Fig. 8** Histogram showing porosity for the larger cement volume set

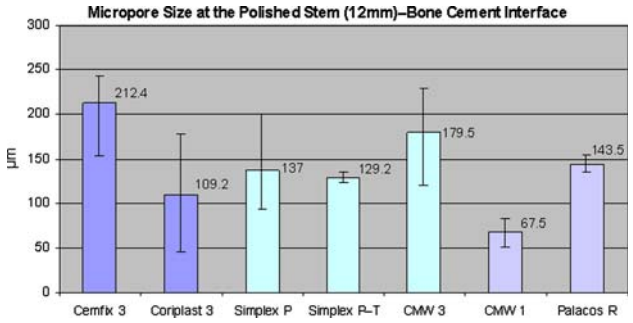


**Fig. 9** Histogram showing micropore size for the larger cement volume set

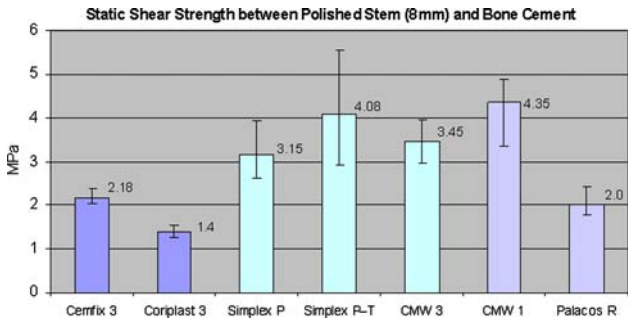


**Fig. 10** Histogram showing static shear strength for the smaller cement volume set

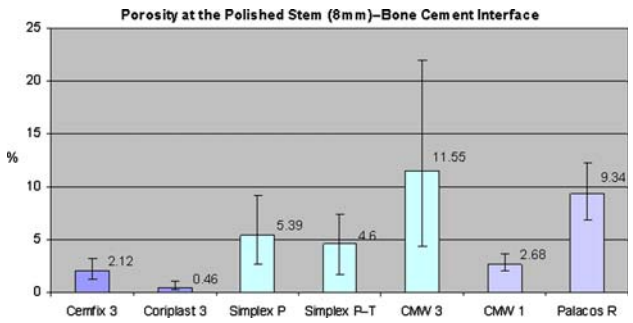


**Fig. 11** Histogram showing porosity for the smaller cement volume set

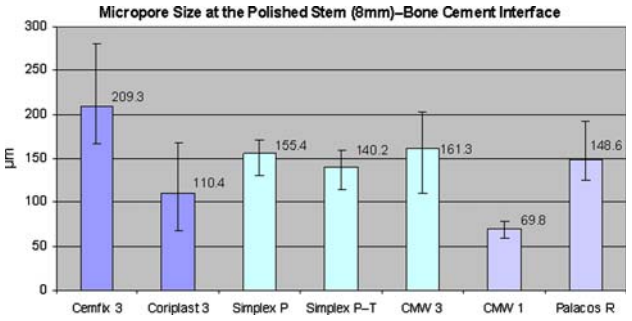


**Fig. 12** Histogram showing micropore size for the smaller cement volume set

**Table 2** Mean static shear strength for each cement type

| Bone cements | Viscosity | Mean static shear strength (MPa) |
|---|---|---|
| Cemfix 3 | Low | 2.23 |
| Coriplast 3 | Low | 2.99 |
| Simplex P | Medium | 3.76 |
| Simplex P-T | Medium | 3.84 |
| CMW 3 | Medium | 2.67 |
| CMW 1 | High | 4.06 |
| Palacos R | High | 2.18 |

CMW 1 gave the highest value, followed by Simplex P-T and Simplex P bone cements. A typical pull out test result obtained from the Hounsfield Test Machine H20K-W displaying the load–displacement plot is exhibited in Fig. 13. The plot shows an initial linear increase of load with incremental displacement until a peak value is reached. This point is defined as the initial debonding force. The force then drops to a lower value before cycling around 1.4 kN until the rod is fully pulled out from the cement mantle and the force returns to zero. The cyclical force was considered unusual; therefore an additional test was performed using CMW 3 bone cement on an Instron 1273 tensile test machine in order to discount machine error, Fig. 14a. The experimental conditions were the same as previous pull out tests. This test again showed a similar result, i.e. an initial high debonding force followed by a cyclical force around a significantly lower value. In this test the metallic rod was made from mild steel and the initial debonding force was recorded as 11 kN. Large areas of bone cement transfer films were detected on the rod surface, which it was considered to be involved in the cyclical force reading, Fig. 14b. It was thought that the cyclical force was a result of either frictional force between the debonded rod and the bone cement and or internal shear within the cement mantle. It was also considered important to investigate this phenomenon as it could have a bearing

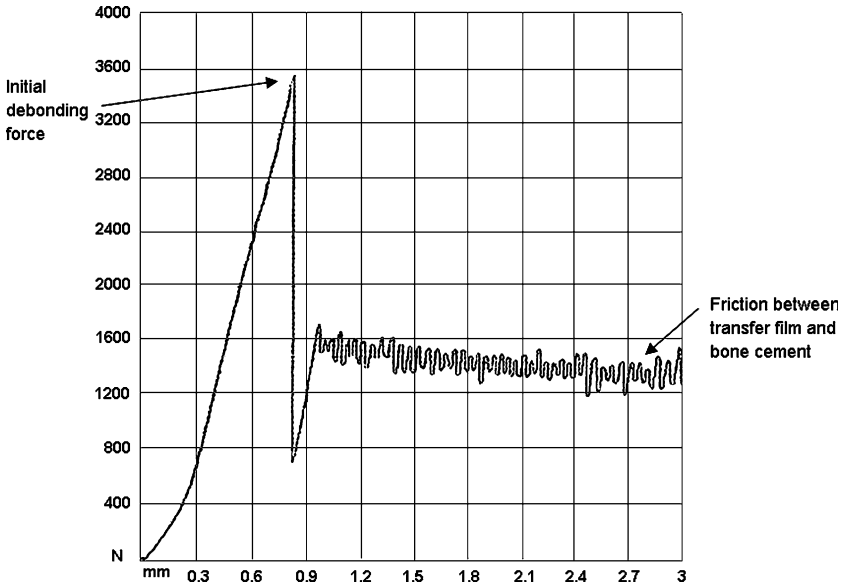


**Fig. 13** A typical load–displacement plot using CMW 3 bone cement

on stem movement in debonded femoral prostheses and in such stem designs where stem migration is common, e.g. the Exeter stem.

It is stated by Hutchings [20] that many polymers (including PMMA bone cement) sliding against hard counterfaces (e.g. metals, especially smooth surface) transfer detectable films onto the counterfaces. These films play an important part in the friction and wear of these polymers. Once the transfer films have formed, subsequent interaction occurs between the polymer and this layer of similar material, irrespective of the composition of the substrate. On further sliding the polymer may continue to wear by adding new material to the transfer films, since the interfacial bond to the counterface is often stronger than that within the bulk of the polymer itself. However, these transfer films whose thickness is of the order of several microns are usually removed subsequently as wear debris. This wear mechanism is further elucidated by other authors [21]. Based on this theory, it was assumed that in the present study the friction and cyclical force built up began with the "clean" rod trying to move against the cement surface. This interface was strong and the force increased until transfer films were formed and the cement material flowed over the cement transfer films, thus causing a drop in the measured force. The total contact area however was diminished gradually, resulting in a decrease of the cyclical force until the stem was fully pulled out. Although no large areas of bone cement transfer films were detected on stainless steel rod surface tested on the Hounsfield Test Machine (probably material-dependent), evidence of its presence was clear. Figure 15 shows the topography of the transfer film, measured by a Talysurf CCI interferometer at ×50 magnification. The height of the transfer films was calculated to be 4–10 μm, using Surfstand software V3.1.

## Discussion

It is generally accepted that long term durability of cemented THR requires meticulous care of three elements and two interfaces, which are femoral stem, stem–cement interface, bone cement, cement–bone interface and bone. With regard to the stem–cement interface, it has been drawing the attention of researchers for a long time due to its great contribution to aseptic loosening of femoral stems [22, 23]. Clinical studies have demonstrated that failure of cemented THR was initiated by debonding at this interface and fractures in the cement mantle [24]. Consequently great efforts have been made to improve mechanical properties of bone cement, and to enhance bond strength at the stem–cement interface. However, interfacial strength at this interface has not been compared for the major commercially available bone cements. In the present study, the static shear strength between polished femoral stem and seven bone cements were investigated through pull out tests. Porosity and micropore size were also calculated based on image processing. The results indicated that the static shear strength was more dependent on cement type than cement viscosity and volume. This complied well with the studies of other researchers [7], who drew the conclusion that effort should focus less on manipulating cement viscosity and more on making compositional changes. However, it was indicated in the present study that cement viscosity had an influence on porosity and micropore size. Previous laboratory tests have shown superiority of strength for low viscosity bone cements [25], but when they are applied for clinical use, those cements are more easily displaced from the irregularities in the bone by blood, thus providing a lower shear strength at the cement–bone interface. This is considered to be the reason

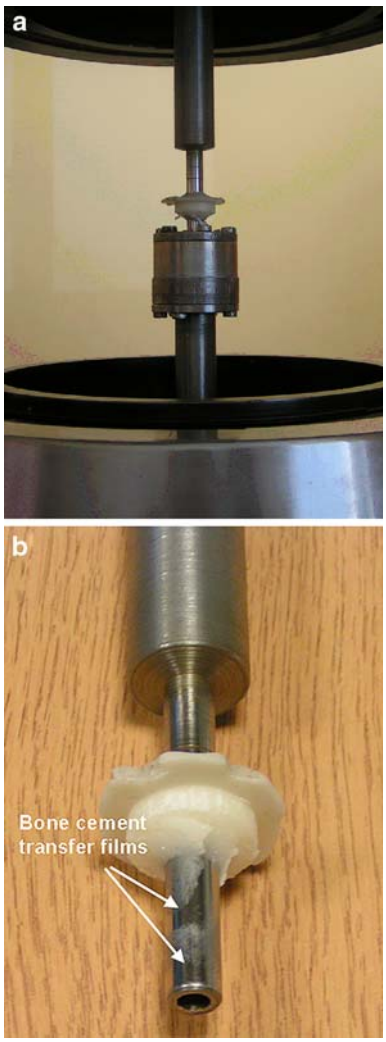


**Fig. 14** (**a**) The Instron pull out test (**b**) Result of the Instron test

why an increased revision rate has been found for those femoral components implanted with low viscosity bone cements [26]. Clinically, it is considered desirable to inject and pressurise low viscosity bone cements into the medullary canal to achieve optimum flow and mechanical interdigitation into cancellous bone [27]. The optimum thickness of bone cement has been recommended to be approximately 3–4 mm [28]. Any thickness lower than this value was considered to be prone to micromovement at the stem–cement–bone interfaces, while a thickness higher than 5 mm was believed to cause more thermal necrosis to the bone. This study has however suggested that heat is not highly detrimental to the static shear strength at the stem–cement interface. Clinical studies based on the data from the Norwegian Arthroplasty Register demonstrated that there were significantly increased rates of failure for Charnley prosthesis inserted with CMW 1 and CMW 3 bone cements [29], whereas the static shear strength for these two bone cements did not show the lowest value, indicating that there must be other factors influencing the failure mode of cemented THR. In addition, the femoral stem insertion rate, which was not consistent for all the pull out tests in this study, possibly had some influence on porosity and micropore distribution. However, a previous study has shown that there was no significant difference concerning porosity at the stem–cement interface for different insertion rates of femoral stems [30].

Surprisingly, the static shear strength was much larger in this study than the results of previous research which employed a similar test but in a push out mode. Wang et al. [12] reported that the static shear strength at the stem–cement interface was 0.53 MPa for Palacos R bone cement using "modern mixing techniques", whereas in the present study the strength was calculated to be 2.4 MPa and 2.0 MPa respectively for the two cement volume sets. It seemed that Wang et al. did not make area modification in their study, and this lack of "correction" could in part have accounted for their lower value of static shear strength. Additionally, both the porosity and micropore size of the bone cements, ranging from 70 μm to 210 μm, were much lower, although the cements were all mixed by hand. It was demonstrated by FEA that the interface conditions at the loading fixture played an important part in interface stress [31], which implied that the differences involved in test specimen preparation and experimental conditions potentially led to the significantly different results. Furthermore, Geiger et al. [13] concluded in their study that vacuum mixed bone cements did not appear to reduce porosity at the stem–cement interface or to improve mechanical properties for all bone cements. The static shear strength obtained in this study varied from 1.4 MPa to 4.6 MPa, which, in spite of its higher value, was still lower than the typical mean shear stress at the stem–cement interface, approximately 5 MPa [32]. Thus, debonding at this interface was considered to occur inevitably during in vivo service of the prosthesis. Femoral stems with a matt surface finish may initially form a higher bond strength at the stem–cement interface, thus prolonging the function of this interface. However, it should be noted that there will be more debris generation as well as more severe damage to bone cement for matt stems once debonding occurs [33]. It was further revealed from the present study that, for bone cements with similar viscosity, larger static shear strength was always obtained where lower porosity was generated at the stem–cement interface. This was consistent with the results of another study, in which Iesaka et al. [34] made a conclusion that increased porosity correlated with a reduction in shear strength after immersion in saline. Indeed, the effect of porosity should not be overlooked because the micropores not optically visible but present immediately below the cement surface could affect the interfacial shear strength. Porosity was also an important

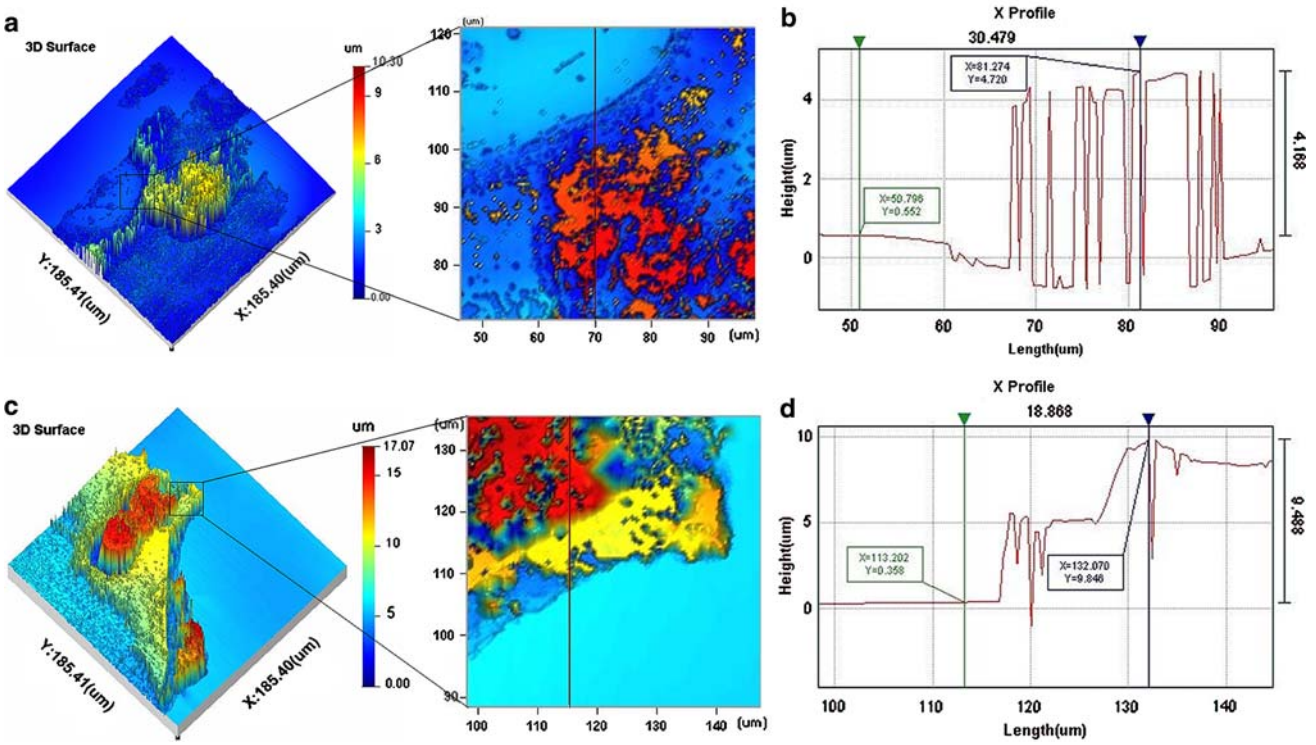


**Fig. 15** (**a**) Transfer films detected for CMW 3 cement in the Instron pull out test (**b**) 2D profile showing height of transfer films for CMW 3 cement in the Instron pull out test (**c**) Transfer films detected for CMW 3 cement in a Hounsfield pull out test (**d**) 2D profile showing height of transfer films for CMW 3 cement in a Hounsfield pull out test

factor that was considered detrimental to mechanical properties of bone cement as well as bond strength at the stem–cement–bone interfaces. The micropores were deemed to induce initiation and propagation of fatigue cracks in the cement mantle. Thus any porosity reduction both in the bulk cement and at the interfaces have been regarded as clinically beneficial [10].

Antibiotics such as gentamicin and tobramycin have been added to PMMA bone cement in order to prevent or treat infection [35] and clinically better results have been reported [26]. Despite their clinical benefit, it is generally accepted that mechanical properties of bone cement will be modified with additions of antibiotics, exemplified by alteration in density, bending strength, and an increase in viscosity [36]. In the present study however, it was indicated by the unpaired student *t*-test that an addition of tobramycin to Simplex P bone cement did not lead to significant differences to the static shear strength and porosity at the stem–cement interface nor did it influence micropore size greatly.

Bone cement transfer films were detected in the present pull out tests. These transfer films were considered to contribute significantly to the cyclical force following the initial debonding at stem–cement interface, where interaction occurred between bone cement and the transfer films. It was speculated that there remained a large bond strength between bone cement and the transfer films, resulting in the cyclical force after debonding as evident on the load–displacement plots. This cement transfer films have been previously observed on some retrieved femoral stems, the formation of which was deemed to play a role in friction and wear of the stem–cement interface [37].

## Conclusions

The following conclusions could be drawn from the present pull out tests:

1. The static shear strength between polished femoral stem and bone cement appears more dependent on cement type than cement viscosity and volume.
2. The highest mean static shear strength of the pull out tests carried out in the present study is obtained for CMW 1, followed by Simplex P with Tobramycin and simplex P bone cements.
3. The porosity and micropore size of bone cement are significantly influenced by cement type and viscosity, while cement volume is not a contributory factor.
4. There is no significant difference between Simplex P and Simplex P with tobramycin bone cements in terms of the static shear strength, porosity and micropore size.

5. The static shear strength between polished femoral stem and bone cement measured in this study is much larger in comparison with that of previous research, while the porosity and micropore size shows smaller values, although all cements were mixed by hand.
6. Bone cement transfer films are present on the stem counterface after the pull out tests, and they are considered to contribute significantly to the interaction at the stem–cement interface.

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