Regional Variation in Shear Strength of the Bone–Polymethylmethacrylate Interface

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Abstract: Many investigators consider the polymethylmethacrylate–bone interface the weakest link in the bone–cement–implant system. Push-out tests, frequently used for in vitro evaluation of the bone–cement interface, produced inconsistent results of shear strength. Therefore, a more reliable model for testing the shear failure of the interface was developed. Better understanding of intrinsic variations in bone quality and geometry of endosteal bone at the interface may yield important insight into the patterns of clinical failure of cemented total joint prostheses. Key words: PMMA, pressurization of PMMA, interface shear strength, variation, anatomic quadrants, bone–cement interface.

Total hip and knee arthroplasty are among the most commonly performed orthopaedic surgical procedures in the United States. The standard method of implant fixation is with polymethylmethacrylate (PMMA) or acrylic bone cement. The major long-term complication is implant loosening, and as the geriatric population increases, this problem will affect an increasing number of individuals. Although cementless fixation is gaining in popularity, it remains a developmental technique; therefore PMMA will probably have a place in total joint arthroplasty for many years to come.

Miller et al. (9) first proposed that there are two categories of implant failure. Stauffer (13) applied this concept in a clinical study reviewing the modes of failure in aseptic loosening of cemented total hip arthroplasties. Type I failure occurs at the prosthesis–PMMA interface due to circumferential hoop stresses in the cement secondary to subsidence of the prosthesis. Pores in the cement increase the likelihood of type I failure. Type II failure occurs at the bone–PMMA interface as a result of fracture of bone secondary to micromotion or necrosis of the trabeculae that interdigitate with the cement. As surgical techniques and strength improve, failure of the cement mantle (type I) may become secondary to failure at the bone–PMMA interface (type II). Pressurization (1, 4, 7, 11, 12), centrifugation (3), and vacuum mixing (15) have been shown to decrease cement porosity and in some cases improve the biomechanical properties. Modifications in technique directed at the interface include saline lavage and reaming to remove marrow products and loose or weak bone. Geometric modifications in the endosteal bone (6) have been suggested for improving interlock between PMMA and bone. Because trabecular bone at the interface is weaker than PMMA (2), it follows that the bone–cement interface could be the weak link in the bone–cement–implant system.
In vitro evaluation of the interface is essential in studying the effects of new techniques on interface strength. Traditionally, push-out tests of bone–PMMA composites (4, 5, 7, 8) have been used for this purpose, yet they may yield inconsistent results due to pushing out the cement at a location other than the true interface between bone and cement or fracture of the bone due to radial tension exerted by the PMMA plug.

We performed preliminary studies in our laboratory using conventional push-out tests on cross sections 10 mm thick. Failure occurred not at the bone–cement interface, but in a catastrophic manner in which the cortical ring shattered, leaving the interface intact. Because our effort was directed at the interface rather than bone, we were stimulated to design a more reliable model for testing interface strength. For this study, we used rectangular blocks from human femurs injected with PMMA. Biomechanical testing of the specimens in shear resulted in failure at the interface. We evaluated the variation in the strength of the interface by quadrants and along the longitudinal axis of the human femur.

Methods

Interfacial Shear Strength

Six pairs of fresh human cadaver femurs were used. Patient histories were reviewed to rule out significant musculoskeletal disease. Each femur was cleaned of excess soft tissue and cut to a length of 12.5 cm at the level of the lesser trochanter. The ends of the bone were embedded in an epoxy mold. The femurs were stored frozen in saline at a temperature of −15°C. Before injection with PMMA, the femurs were thawed at room temperature. The intramedullary canal was cleaned thoroughly with saline lavage using a Water Pik (Teledyne, Fort Collins, CO) and a bottle brush.

Simplex-P cement (Howmedica, Rutherford, NJ) was mixed at 2 Hz for 45 seconds at an ambient temperature of 24 ± 1°C. The cement was poured into a plastic canister, which was then coupled to an aluminum jig containing the embedded femur. A pressure transducer (Endevco, Anaheim, CA) was
placed in the proximal end of the injecting apparatus as well as in the intramedullary canal. PMMA was injected into each femur at a sustained pressure of 60 psi. Pressurization was maintained throughout polymerization (Fig. 1).

Following pressurization, each bone-PMMA composite was stored for at least 24 hours, then cut transversely into 10-mm slices using a diamond blade on a Bronvil milling saw. Blocks of the composite measuring $5 \times 10$ mm were cut from homotypic sites on the anterior, posterior, medial, and lateral aspect of each slice (Fig. 2).

Each composite was mounted in a specially designed testing clamp with the shear plane parallel to the bone-cement interface; they were loaded proximally to avoid a wedge effect (Fig. 3). The block was irrigated during testing to prevent desiccation of the bone. Each block was sheared to failure at a deformation of 0.5 mm/second. The load–deformation curve was recorded. Energy absorbed was calculated as the area under the load–deformation curve up to ultimate load. A total of 182 specimens were tested.

Apparent shear stress was defined as shear load divided by cross-sectional area (50.0 mm$^2$), and shear strain was defined as shear deflection divided by length of the specimen in the shear interface gap (1.5 mm). The shear strength represented the stress determined at the ultimate load. The apparent shear modulus was calculated as the slope of the linear portion of the stress versus strain curve.

Nine femurs were prepared and injected with PMMA as described and cut transversely into 5.0-mm slices. The specimens at levels of 2.0, 2.5, 3.0, 3.5, 5.0, and 5.5 cm along the shaft of the femur were selected for evaluation. To enhance the resolution between bone and cement interface, we
stained the specimens with silver nitrate by placing them in a solution of 10% silver nitrate for 2 hours. The silver nitrate stains the bone dark grey to black by ion exchange, leaving the cement unchanged. This technique facilitated distinction between bone and cement at the interface. Each slice was divided into four quadrants: anterior, posterior, medial, and lateral. The trabecular orientation was classified as random, parallel, or perpendicular to the endosteal surface for each quadrant. Areas that were completely lacking in trabeculation were classified as undulating or flat (Fig. 4). The results for each specimen were compiled and an overall characterization of the gross appearance of the interface obtained.

Results

Interfacial Shear Strength

Ten of 192 specimens failed at the interface before they were mounted in the testing clamp. The mean quadrant interfacial shear strength is shown in Figure 5. Three-way analysis of variance, grouping data by quadrant, revealed a significant difference in quadrant shear strength between the medial and lateral quadrants ($P < .05$) and no significant difference in the shear strength of the four slices along the length of the femur or between matched pairs. Interfacial shear failure in a uniform mode was achieved by using specimens of bone–cement composite cut to precise dimensions. The highest shear strengths were observed medially and the lowest strengths laterally. Although the anterior quadrant was slightly lower than the posterior quadrant in shear strength, the difference was not significant.

Endosteal Architecture

There was a definite variation in endosteal architecture between medial and lateral interfaces. The medial interface was usually trabeculated with a trabecular orientation perpendicular to the endosteal surface. The lateral interface was less trabeculated. Numerous specimens showed a complete lack of trabeculation, with 42 of 54 (78%) lateral surfaces rated as undulating or flat (Figs. 4, 6).

Discussion

The sustained pressure used in this study (60 psi) is comparable to that achieved when a femoral prosthesis is inserted into a plugged intramedullary canal filled with cement (7). Pressurization of cement with a prosthesis, however, is (1) transitory, as demonstrated in our laboratory; (2) most effective distally; (3) only minimally effective proximally; and (4) inversely related to the optimal flow characteristics of cement. The sustained pressurization was initially applied when the cement was in a low-viscosity state and was maintained beyond the dough stage.

Regional variations in interfacial shear strength in
bone–PMMA composites of the human femur were observed in each case, with the medial interface highest in shear strength by an average of 30%. The posterior and anterior interfaces were similar in shear strength, with the values falling between those of the medial and lateral interfaces. All of the specimens that failed prior to testing were from the lateral quadrant.

Our observations on the macroscopic anatomy of the endosteal surface suggest that bone geometry is an important factor in bone–PMMA interface failure. Trabeculations at the interface provide for interdigitation and macrointerlock between bone and PMMA. A perpendicular orientation to the shear plane would maximize resistance to these stresses. This was the case for the medial interface, which was mechanically superior. The lateral interfaces tended to have fewer trabeculations, with the majority rated as undulating or flat. The specimens that failed prior to testing had smooth interfaces without any interdigitation. This medial to lateral variation in endosteal architecture offers a plausible explanation for the observed variations in regional interface shear strength.

Svensson et al. (14) considered the bond between acrylic and bone to be the weakest component in the bone–cement–implant system. Using finite element analysis, they demonstrated shear and tensile stresses across the bone–PMMA interface in the femur to be the most critical stresses in the system. Because of the irregularity of the endosteal surface and variability between samples, conventional push-out tests cannot isolate and test this interface in a reproducible manner. Our testing technique can accommodate the endosteal irregularity and effectively test this critical interface, and it allows evaluation of regional differences around the circumference of the interface.

Regional differences in mechanical properties have been demonstrated in other models. Halawa et al. (5) studied the variation of shear strength in bone across the coronal plane in the neck of the femur. They demonstrated higher stresses in the medial (8.0 MPa) than in the lateral (5.0 MPa) corticocancellous junction. They attributed this difference to the presence of primary compressive trabeculae medially and the primary tensile trabeculae laterally. These observations may be explained by Wolff's law, which
simply states that bone density is greatest where bone is under compression and less where it is under tension. A variation in endosteal shear strength also may be explained by the number and orientation of individual trabeculations at the interface. These trabeculations provide the macrointerlocking of acrylic to bone.

Shear strength did not increase from proximal to distal along the length of the femur. Using push-out tests, Oh et al. (10) demonstrated that the distal interface was stronger than the proximal interface. They attributed this to the fact that pressure head on the cement was lost proximally due to cement leakage. We attribute the absence of longitudinal variation in this study to the fact that our pressurization technique provides complete occlusion of the femoral canal. This converts it to a closed system with a relatively uniform cement penetration along the interface (4). Perhaps it is this uniformity in pressure and penetration that diminishes the variation in strength of similar anatomic sites along the length of the femur.

Summary

Based on this study, we draw the following conclusions. (1) Interfacial shear strength can be determined reliably using precision-cut rectangular specimens of the bone–cement interface. (2) The significant regional differences in interfacial strength demonstrated with this model would not be demonstrated using conventional push-out tests. (3) The medial quadrant interfaces were significantly greater in shear strength and the lateral quadrant interfaces lower in shear strength relative to each other and to the posterior and lateral quadrants. (4) Longitudinal variation in shear strength from proximal to distal was not observed using this system of pressurization. (5) The observed regional variations in shear strength in the femur may have important implications in the observed patterns of clinical failure.

Acknowledgments

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References