6

Biomechanical Considerations

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Percutaneous vertebroplasty (PV) has enjoyed rapid acceptance as a procedure with which to stabilize vertebral compression fractures (VCFs) and to prevent fractures in vertebral bodies weakened by osteolytic tumors. The procedure is being performed with increasing frequency, and scientific investigations into basic questions regarding the clinical efficacy and technical aspects of the procedure are becoming more common. This chapter reviews the current body of knowledge regarding PV fundamental research and attempts to place into clinical perspective the results from that research.

Mechanism of Pain Relief

The augmentation and stabilization of vertebrae using acrylic cement as an open procedure (vertebroplasty) has been practiced for many years (1–10). However, the percutaneous introduction of cement into a vertebra was first reported in 1987 (11). The procedure consisted of injecting polymethylmethacrylate (PMMA) cement through a large-bore needle into a painful vertebral hemangioma that had aggressively consumed a C2 vertebra. The vertebral hemangioma was injected primarily to prevent subsequent collapse of the involved vertebra, but the procedure also reportedly resulted in marked pain relief (11). The procedure was quickly adapted to stabilize osteoporotic VCFs (12). Since the introduction of PV, retrospective and prospective studies have reported pain relief in approximately 90% of patients treated for osteoporotic VCFs (13–19) and in approximately 70% of patients treated for various tumors (20–23). Although the exact mechanism of pain relief is unknown and may differ in patients with osteoporotic VCFs and those with tumors, possible mechanisms include thermal, chemical, and mechanical factors (24,25). Histologic studies of retrieved specimens report a zone of necrosis around the cement. This zone has been attributed to thermal damage, cytotoxicity from the methylmethacrylate (MMA) monomer, and ischemia (26,27). Because the specimens describe a single point in time, one can only speculate as to the cause

of the necrosis. Retrieved specimens from animal models did not indicate necrosis around the cement (28).

Thermal

It has been hypothesized that the heat of polymerization causes thermal necrosis of neural tissue and is therefore the mechanism responsible for pain relief (24). When PMMA polymerizes, heat is generated in the exothermic polymerization reaction (29). Concern about potential thermal tissue injury caused by the heat of polymerization has been the topic of orthopaedic investigations, with particular reference to arthroplasty (29–32). Thermal injury illustrates an Arrhenius relationship in which temperature magnitude and exposure time are both critical factors. Thermal necrosis of osteoblasts occurs when temperatures are higher than 50°C for more than 1 minute (33,34), but apoptosis occurs when osteoblasts are exposed to lower temperatures for longer periods of time (35). Some investigators have measured temperatures as high as 122°C during polymerization (36), but the volumes of cement required to generate such temperatures are substantially greater than those typically used in PV (35). Neural tissue may be more sensitive than osteoblasts to temperature (37).

A previous ex vivo study suggests that temperature is not a mechanism of pain relief (38). In that study, thermocouples were placed at three locations inside vertebral bodies (Figure 6.1) to assess the risk of thermal injury to interosseous nerves, periosteal nerves, and the spinal cord. The vertebral bodies received concurrent bipedicular injections totaling 10 mL of PMMA cement. Although temperatures exceeded 50°C for more than 1 minute at the anterior cortex and in the center of the vertebral body, the authors concluded that temperature was an

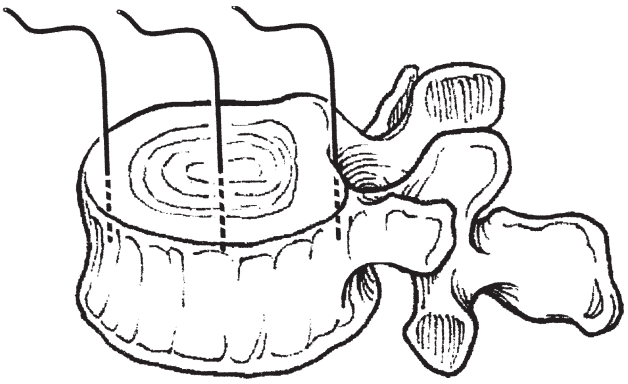

Figure 6.1. Schematic of a vertebral body instrumented with thermocouples to measure temperature elevation caused by polymerizing PMMA cement. Thermocouples were placed at the anterior cortex, at the centrum, and under the venus plexus of the spinal canal. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

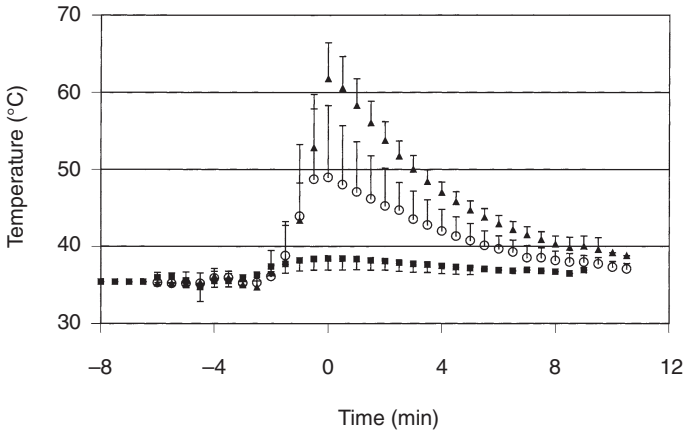

Figure 6.2. Typical temperature-versus-time response of a vertebral body injected with 10 mL of PMMA cement. Temperatures of 50°C for more than 1 minute cause necrosis of osteoblasts. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

unlikely mechanism of pain relief. The study was recently reconducted (39). To reflect the smaller volumes being injected in contemporary practice of PV, a 6-mL cement volume group was added to the previous study protocol (Figure 6.2). Another important change in the experimental design was that the cannulae were removed during cement polymerization to prevent inadvertent heat transfer through the cannulae into the bath. Even with smaller volumes injected (i.e., 6 mL), peak temperatures were higher and dwell times above 50°C were longer than those previously measured. For some specimens, peak temperatures were in excess of 110°C. Although the potential for thermal injury cannot be ruled out, the role of temperature remains unresolved. The ex vivo model did not account for active heat transfer secondary to blood profusion, which would be expected to remove much of the heat in vivo. In another study, temperatures measured in an in vivo goat model were below those needed to cause thermal injury (40). The low temperatures may be explained by the effect of blood profusion, but they also may be a consequence of the small volume of cement injected relative to that used in humans. The average volume of cement injected into the goat spines was 0.8 mL, an order of magnitude lower than the volume injected in the human cadaver studies. It is doubtful that the thermal energy and resulting temperature elevations can be scaled linearly based on the size of the vertebral bodies from the respective species. Until temperatures are measured in vivo in human patients, the risk of thermal injury during vertebroplasty will remain undetermined.

Temperature may, however, play a role in slowing tumor growth (31). A recent study indicated that apoptosis likely occurs in osteoblasts exposed to 48°C for 10 minutes or more (35). If similar results are found

for tumor cells, apoptosis and diminished tumor cell proliferation may result from exposure to polymerizing PMMA.

Chemical

Methylmethacrylate monomer is cytotoxic (41), but it is unknown if concentrations present in vivo immediately after PV are sufficiently high to be neurotoxic and therefore a mechanism of pain relief (24). In vitro concentrations exceeding 10 mg/mL have been shown to be toxic to leukocytes and endothelial cells (41), yet there are no reports that suggest in vivo concentrations reach such magnitudes. During knee arthroplasty, blood serum levels immediately after cementation and tourniquet release have been measured as high as 120 μg/mL, but such levels typically are much lower (<2 μg/mL) and drop precipitously minutes after cementation (42). During total hip replacement, blood serum concentrations between 0.02 and 59 μg/mL have been measured (43). The volumes of cement used for hip and knee arthroplasty are two to three times larger than those typically used with PV, and the monomer concentrations measured for those procedures are 10 to 100 times less than MMA concentrations reported to be cytotoxic to tissue cultures (41). Even though the cement used with PV typically is prepared with a greater monomer-to-polymer ratio than that of cement used for arthroplasty, it seems unlikely that MMA toxicity is responsible for pain relief experienced with PV.

Cytotoxicity also has been implicated in the antitumoral effect noted clinically (44). However, a recent cell culture study (45) suggested that MMA monomer is cytotoxic to breast cancer cells in concentrations similar to those for leukocytes and endothelial cells (41). Thus, it also seems unlikely that MMA monomer leachate from cement injected during PV has an antitumoral role. Nevertheless, until intravertebral MMA concentrations are measured in vivo, the hypothetical cytotoxic effect of MMA monomer will remain in question.

Mechanical

Mechanical stabilization of the affected vertebral body appears to be the most likely mechanism of pain relief. As with fixation of fractures in other parts of the human skeleton, internal fixation (in the current case, by PV) likely stabilizes the fracture and prevents micromotion at the fracture site, thereby limiting painful nerve stimulation (46,47). In tumors, the pain relief mechanism may be more complex. If the vertebral body contains regions of instability resulting from osteolytic activity by the tumor, PV may prevent micromotion and subsequent pain. If the cement injected during PV has some antitumoral effect (44), then the pain associated with rapid tumor growth may be diminished. The antitumoral effect may be thermal or chemical, as mentioned above, but it also may result from ischemia caused by the mechanical displacement of tumor tissue by the cement and resulting hydrostatic pressure. Thus, injecting PMMA cement into tumors of the spine may have the triumvirate effect of vertebral body stabilization, pain relief, and tumor growth impediment.

Biomechanical Stabilization

Basic Biomechanics

The spine serves to transmit loads from the upper body through the pelvis into the lower extremities. The spine is conceptually divided into three columns: anterior, medial, and posterior. The medial and anterior columns serve to resist axial compressive loads (48) that increase in magnitude from the cervical region to the lumbar region. Because the center of gravity of the human body is located anterior to the spinal column, it creates a combined load resulting in axial compression and an anterior bending moment. For the spine to remain erect, tensile forces along the posterior column (i.e., paraspinous muscles and ligaments) need to act about the medial column, which serves as a fulcrum, while the anterior column acts to resist compression (Figure 6.3). During anterior flexion (e.g., bending over to tie a pair of shoes), the body's center of gravity moves anteriorly, increasing the bending moment on the spine and the compressive stresses on the anterior column. Bending over to pick up a load not only moves the center of

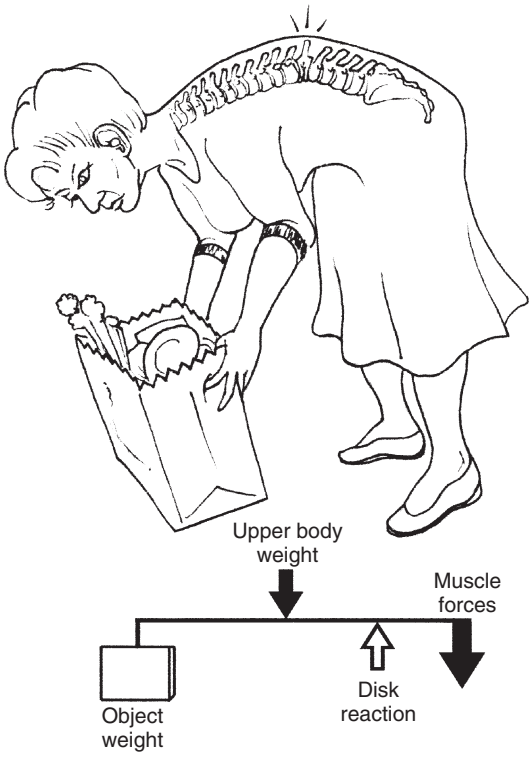

Figure 6.3. The body's center of gravity is anterior to the spine, creating an anterior bending moment and axial compression on the spine. Anterior flexion increases the anterior bending moment, thereby increasing the stresses on the spine and placing the spine at risk for fracture. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

gravity anteriorly, but it also increases the magnitude of the anteriorly located load, which, when combined with the increased moment arm, dramatically increases the compressive stresses on the anterior column. It is this excessive compressive stress that results in VCFs. By definition, VCFs exhibit disruption of the anterior column (48).

Compressive strength of vertebra is roughly related to the square of the vertebral bone mineral density (BMD) (49). When a patient's BMD is 2 standard deviations below the average for the sex-, height-, weight-, and race-matched young population, the patient is considered to be osteopenic. When BMD drops below 2.5 standard deviations, the patient is considered osteoporotic (50). In patients with osteoporosis, vertebral BMD might be half of what it was in their youth, which means the vertebral compressive strength may be as low as a one fourth of what it was in their previous young healthy condition.

Although many VCFs go undiagnosed (51,52), 700,000 VCFs are reported each year in the United States (53), 300,000 to 400,000 of which result in hospital admissions. Vertebral compression fractures that are diagnosed may be immediately radiographically apparent or may present with pain but little or no radiographically discernible deformity (54). The former fracture type is typically associated with an acute onset of pain during lifting, raising a window, and so forth, whereas the latter type suggests an initial weakening (perhaps as a result of microfractures) that reportedly progresses into radiographically diagnosable wedge fractures 6 to 16 weeks later (54).

Volume Fill

The goals of stabilization for VCFs are similar to those of stabilization for fractures in other sites in the body, namely, to prevent painful micromotion and provide a mechanically stable and biologically conducive environment for fracture healing to occur. The amount of strength and stability needed to provide the optimal mechanical environment for VCF healing is unknown and remains a point of controversy (55,56). Early in the PV experience, complete injection of the anterior column of the vertebrae was thought necessary (57), but recent clinical and experimental data have suggested that smaller volumes of cement may be sufficient (18,19,58). In one clinical study, 29 patients treated with PV received injected volumes ranging from 2.2 to 11.0 mL (mean, 7.1 mL) of cement; 90% of the patients experienced pain relief (13). Barr et al. (59) indicated that injection of 2 to 3 mL into the thoracic and 3 to 5 mL into the lumbar regions resulted in 97% moderate to complete pain relief. These results suggest that pain relief may be achieved with smaller volumes, but no correlation of level treated, volume injected, and clinical outcome was reported explicitly. In osteolytic metastases and myeloma, there is reportedly no correlation between the percentage of lesion filled and pain relief (60). A similar lack of relationship between cement dose and pain relief was suggested for osteoporotic compression fractures (61). A recent ex vivo study attempted to determine the relationship between cement volume injected and subsequent mechanical stabilization and found that only 2 mL of PMMA was

needed to restore strength in osteoporotic vertebral bodies (Figure 6.4), but that larger volumes (4 to 8 mL) were needed to restore stiffness (62). Because the correlation between volume of cement injected and restoration of mechanical properties was very weak, another study was undertaken to correlate the cement volume as a percentage of vertebral body volume with the restoration of mechanical properties (58). In this manner, the geometry of the vertebral body was removed from the analysis. Although the resulting correlation was similarly weak, it suggested that an injection of cement on the order of 30% of the vertebral body volume restored stiffness. A computational model of vertebroplasty reported that only 14% volume fill was needed to restore stiffness (63). Considering the variation in the experimental data, the experimental results and computational results are not necessarily inconsistent. Mechanical property restoration is a function of the volume of cement injected, the density of the host bone, and, to a lesser extent, the location of the cement.

Postvertebroplasty stiffness is the mechanical parameter likely to be linked most closely with pain relief (62). Restoring initial strength might be expected to prevent refracture of the treated vertebra, whereas restoring initial vertebral body stiffness likely prevents micromotion and the pain associated with it. However, fully restoring prefracture stiffness to vertebral bodies may not be necessary or even desirable. As with other fractures, providing some mechanical stability, even less than that of the prefracture state, may be sufficient to allow healing (64). If the repair is too stiff, stress shielding may occur and impede fracture healing. If the repair is not stiff enough, excessive motion at the fracture site may occur, resulting in nonunion. Furthermore, the remaining cancellous bone in the vertebral body is still osteoporotic and at risk of fracture. Thus, it is not surprising that there are some reports of refracture around the cement injected during a previous vertebroplasty (55,65). Some clinicians might be inclined to fill the vertebral body maximally in hopes of preventing secondary fractures, but this increases the risk of extravasation and subsequent pulmonary complications and theoretically may prevent the endplates from deflecting, thereby increasing disc pressure and placing adjacent levels at increased risk of fracture (66). However, disc pressure measurements ex vivo do not support this hypothesis (67).

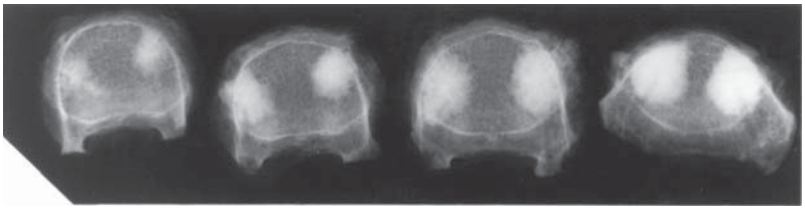

Figure 6.4. Radiograph of typical cement (Simplex P) distribution when 2, 4, 6, or 8 mL is injected into lumbar vertebrae. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

The volume and material properties of cement needed to achieve sufficient stabilization for healing and to prevent pain are yet unknown and can be determined definitively only by a prospective, controlled, randomized clinical study. Some of the conflicting opinions regarding the appropriate volume of cement needed for injection stem from the different goals of the procedure. Providing fracture stabilization to prevent pain and allow fracture healing may require a different cement volume than that needed to prevent fracture through prophylactic augmentation.

Unipedicular Injection

In another ex vivo study, Tohmeh et al. (47) found that vertebral body strength may be restored via a unipedicular injection of 6 mL of cement without risk of vertebral body collapse on the uninjected side (Figure 6.5). Both injection protocols in that study (6 mL unipedicular, 10 mL bipedicular) resulted in increased strength and restored stiffness to fractured vertebral bodies. These results (47), considered in conjunction with those of the previously mentioned volume-fill study (62), suggest that the injection of the appropriate cement volume is more important than the manner in which it is injected. The findings also were supported by a subsequent study in which injected volumes more closely reflect those in the contemporary practice of vertebroplasty. Despite results from a computational model to the contrary (63,68), a unipedicular injection of an appropriate volume of cement may allow adequate stabilization with the added benefit of reduced procedure time and risk associated with bilateral cannula placement. A similar ex

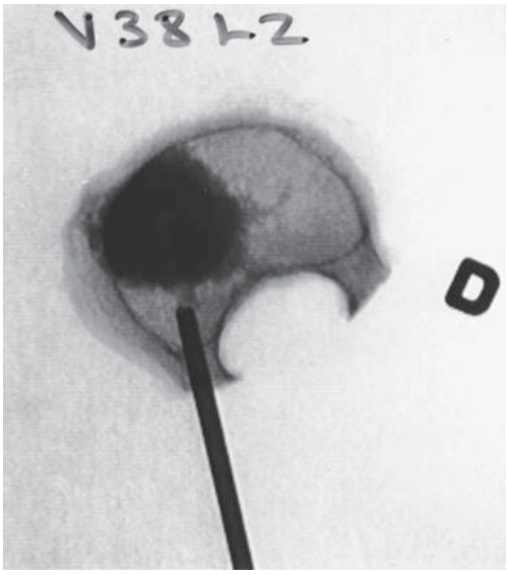

Figure 6.5. Typical distribution of cement after unipedicular injection of 6 mL of PMMA cement. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

vivo study (69) compared the compressive strength of vertebral bodies augmented prophylactically by a single posterolateral injection to those left unaugmented. Those investigators found that augmentation, even by modest (4.3 ± 1.6 mL) volumes of cement, increased vertebral body strength. Preliminary clinical outcome data on a limited number of patients in which the unipedicular procedure has been performed (59) support the ex vivo findings (47). However, it is unknown if unipedicular injections of volumes used in those ex vivo studies (47,54) would result in adequate mechanical stabilization clinically.

Kyphosis Reduction

Restoration of height lost as a result of VCF and correction of the resulting kyphosis have the potential benefit of reducing postfracture sequelae such as loss of appetite, reduced pulmonary capacity, and diminished quality of life (70–74). Vertebral body height measured ex vivo suggests that minimal height (i.e., 1 to 2 mm) is restored after PV (75–77). To increase height restoration, a new device, the inflatable bone tamp, has been developed (75,78). The procedure used to place and inflate the bone tamp has been termed *kyphoplasty* (see Chapter 8 for a detailed description). Ex vivo tests indicate that the tamp treatment restores significantly more height than does standard PV treatment and achieves restoration of mechanical properties similar to that of PV (75,78). A recent report (79) suggests that similar height restoration may be achieved clinically, whether performing kyphoplasty (80) or not. The controversy over height restoration is presented in the chapter on kyphoplasty (see Chapter 8).

Injection Pressure

Another controversy regarding vertebroplasty concerns the pressure needed to deliver the cement into the vertebral body. Some investigators (81) report that creating a void allows cement to be injected under lower pressure than would be the case if cement were injected directly into the vertebral body. If a lower pressure were required to deliver the cement, the argument goes, then a more viscous cement could be used. Cement with greater viscosity is less likely to extravasate and result in clinical complications (82). Concern over injection pressure really stems from the tactile feedback clinicians receive during injection. Approximately 95% of the pressure required for cement injection is to overcome the friction in the cannula. This pressure can be substantial, especially when injecting cements that are or have become viscous (83). Only approximately 5% of the injection pressure is a function of the infiltration parameters of the vertebral body (82). The required pressure at the tip of the cannula is only that needed to displace the marrow, fat, and blood products in the vertebral body. A bench study reported that rapid injections of cement were required to produce a measurable increase in intravertebral body pressure (83). In that study, the cement was injected at a rate well in excess of what would be deemed clinically safe. Even then, the measured pressure was only 6 to 10 mm Hg above ambient pressure, but the pressure in the syringe exceeded 18,000 mm Hg.

Altered Kinematics/Adjacent Fractures

There is much concern about the potential increased risk of fractures occurring in the levels adjacent to vertebral bodies that have been treated with vertebroplasty. Retrospective clinical studies report conflicting results (84,85). Taking into consideration that risk of a subsequent vertebral body fracture increases 12.6 times after the initial fracture and that compression fractures are most prevalent in the thoracolumbar junction (86), it is difficult to differentiate which fractures would have occurred had vertebroplasty not been performed. None of the current clinical studies has sufficient power to make such a differentiation.

From a mechanical perspective, it is theoretically unlikely that stress concentration would occur at a level adjacent to one that had received vertebroplasty. Vertebroplasty typically restores or nearly restores the native strength and stiffness of the vertebral body. Thus, by definition, no stress concentration results. Even if large volumes of cement were injected, thus increasing the strength and stiffness of the vertebral body, most spinal motion occurs at the level of the disc. Unless the mechanics of the disc are altered (i.e., damaged, filled with cement) or the demands for motion increased (compensation for fused levels), no alteration in normal spine kinematics would be expected. Adjacent fractures occur most often when several levels are fused. In this instance, the normal kinematics of the spine is altered. In the normal spine, motion occurs in the flexible disc. After fusion, the levels adjacent to the fused levels are required to compensate for the lost motion. The resultant excessive motion places increased stress on those levels and puts them at risk for fracture. Interestingly, vertebroplasty is one of the procedures used in orthopaedic surgery to reduce the risk of fracture in the adjacent level. Should cement leak into the disc, however, the adjacent level is at increased risk of fracture (87).

A recent biomechanical study investigated the effect of vertebroplasty and kyphoplasty on adjacent disc pressures (67). Although disc pressure was reduced dramatically when an adjacent level was fractured, once the level was treated with either kyphoplasty or vertebroplasty, disc pressure increased, but not back to the prefracture normal level. These findings support the conclusion that vertebroplasty and kyphoplasty do not increase the risk of adjacent fractures, a finding that is in opposition to computational models (66). An ex vivo study of two-level functional spine units (FSUs) reported the augmented FSU was 19% weaker than the unaugmented FSU, although the difference was not significant (88). The investigators suggested that vertebroplasty may place adjacent levels at risk of fracture (88). It should be noted that that study may have introduced some experimental bias by always augmenting the caudal level of the FSU. The authors also injected a high volume (8.8 mL, on average) of cement relative to common vertebroplasty practice. Despite the attempts to identify biomechanically the risks of adjacent fractures, the true risk may be identified only through a carefully controlled, prospective, randomized clinical study.

Materials and Tests

Cement Alterations

Since the publication of the first edition (89), several cements have received approval by the Federal Drug Administration (FDA) in the United States and by the Conformitè Europèene in Europe. Before this approval was given, many clinicians prepared their own mixtures of cement by altering the composition of PMMA cements that typically were approved for arthroplasty. Common alterations included (1) increasing the monomer-to-polymer ratio to increase working time and decrease viscosity (13,57,90), (2) adding radio-opacifiers to increase cement visualization under fluoroscopy (13,57,90), and (3) adding antibiotics (13). Altering an FDA-approved product is not considered off-label use; it creates a new device that needs to be FDA approved.

Monomer-to-Polymer Ratio

Increasing the monomer-to-polymer ratio decreases the compressive material properties of the cement (Figure 6.6) (91–93). Because cements altered for use with PV typically have monomer-to-polymer ratios of about 0.72 mL/g (compared with the manufacturer-recommended ratio of 0.5 mL/g), there likely is an increased amount of unreacted monomer available to enter the circulatory system (91–93). Even so, actual blood serum concentration during PV may be lower than that measured during total hip arthrodesis because the quantity of cement injected (<10 mL) is much smaller than that for hip arthrodesis (>40 mL) (41,42,94).

Radio-Opacification

Altering the concentration of radio-opacifiers significantly alters the material properties of the cement, as does the combined alteration of

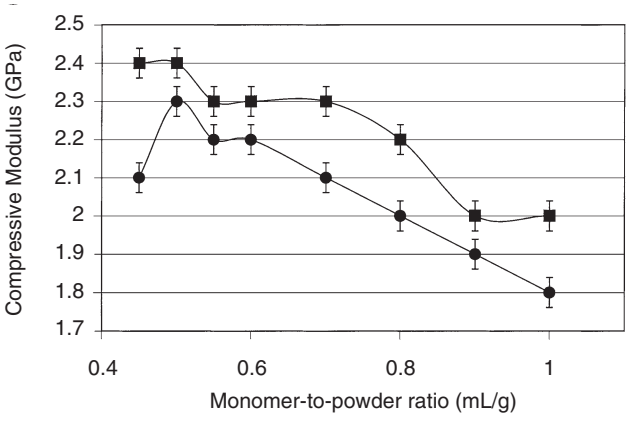

Figure 6.6. Cement compressive modulus as a function of the monomer-to-powder ratio for Simplex P. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

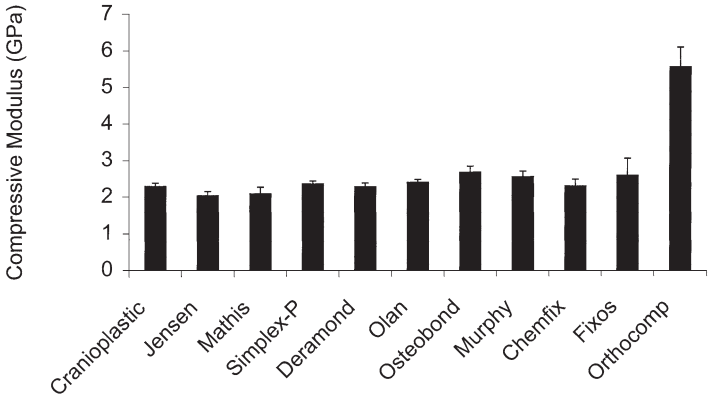

Figure 6.7. Relative compressive strengths for various cement recipes used in PV. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

monomer-to-polymer ratio and opacification (95). Although these modifications are statistically significant, they are of dubious clinical importance. In a recent study of tested cement recipes, the cement composition (Figure 6.7) that exhibited the minimum relative material properties (95) was the composition that has been used clinically during the past decade in the United States (13), but there have been no reports of complications associated with mechanical failure of that cement composition. Complications that have been reported are predominantly cement extravasation or the consequences of extravasation (13,96–98). The prevention of extravasation by means of adequate opacification and careful fluoroscopic visualization during cement injection is essential for the safe practice of PV (Figure 6.8). Thus, selecting a cement that can be injected easily and has proper opacification takes precedence over a cement that is unmodified and retains its original material properties.

Figure 6.8. Radiopacities of various mixtures of cement: A, Simplex P; B, Simplex P with 20% by weight $BaSO_4$; C, Mathis recipe; D, Cranioplastic with 10 percent by weight $BaSO_4$; E, Fixos; F, Chemfix3; G, Orthocomp; H, Murphy recipe; I, Olan recipe; J, Simplex P with 30% by weight $BaSO_4$; K, Deramond recipe; L, Cranioplastic with 20% $BaSO_4$; M, Jensen recipe; N, Cranioplastic with 30% by weight $BaSO_4$ (see Jasper et al. [92] for composition details). (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

Antibiotics

The efficacy of adding antibiotics to cement to reduce the risk of infection during PV is unknown. In contrast to arthrodesis procedures (99), the risk of infection from PV is extremely low (<1 percent). Therefore, elucidating the efficacy of prophylactic antibiotics would require a clinical trial with an extremely large population size for such a study to have sufficient statistical power. For immunocompromised patients, some clinicians routinely add antibiotics to the cement mixture (13).

It is also unknown what effect adding antibiotics to PMMA cement prepared for PV has on the cement's material properties. The addition of antibiotics to PMMA cement used in arthroplasty reportedly does not affect the cement's fatigue properties (100) and may increase its compressive strength (99).

Mechanical Tests

Cement Tests

Most mechanical tests for determining the material properties of acrylic bone cements are performed based on the American Society for Testing and Materials (ASTM) standard F451 (101) or similar test standards. To measure compressive material properties of acrylic cement, the cement components typically are weighed, mixed, and then poured into a mold consisting of cylindrical holes, each 6 mm in diameter and 12 mm high. The mold is then placed between two stainless steel plates, compressed, and subsequently placed in a saline (0.09%) bath maintained at 37°C for a given period of time. The cement specimens are sanded flush with the mold, pressed out of the mold, and inspected for defects. Specimens containing defects greater than 10 percent of their cross-section are culled from the group of test specimens. The specimens then are individually placed between loading platens on a materials testing machine and compressed to failure. Stress and strain data, obtained by dividing the load and deformation data by a specimen's cross-sectional area and initial length, respectively, are plotted for each specimen (Figure 6.9). Ultimate compressive stress is defined as peak (maximum) stress. Compressive modulus is defined as the slope of the linear (Hookean) portion of the stress-versus-strain curve. Compressive yield strength is determined using the 2% offset method, in which a line is drawn parallel to the Hookean portion of the stress-versus-strain curve but offset along the strain axis a distance equal to 2% of the specimen's initial height.

Compression is the loading mode most often used to test cements for PV. Although the cement undoubtedly experiences shear and tensile stresses in vivo, the dominant stress likely is compressive. It is unknown if cement fatigue is of clinical concern for the practice of PV. There are no clinical reports describing mechanical failure (fatigue or otherwise) of the cement. Furthermore, it is unknown if the stress magnitudes, in vivo, are sufficient to cause fatigue. It is unlikely that the stress magnitudes typically experienced by bone cement used with hip arthroplasty are similar to those experienced in the spine. For example,

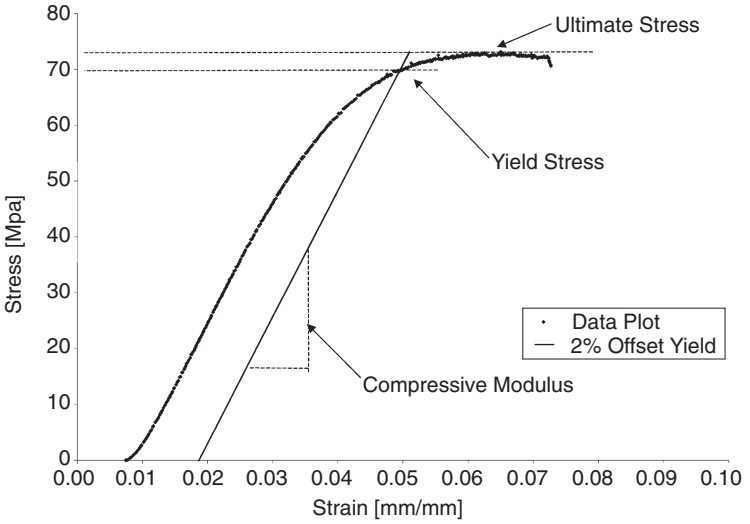

Figure 6.9. Typical compressive material behavior of cement specimens. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

the strength of one PMMA cement manufactured for use in vertebroplasty is 65 megapascals (MPa) (93), and an average cross-sectional area of a lumbar vertebral body endplate is 1,200 mm (2,102). An axial load on the spine of approximately 78 kilonewtons (kN) would be needed to generate enough stress to cause cement failure. For a 70-kg man, an axial load of 78 kN equates to 114 times body weight, which is well beyond the failure strength of a lumbar vertebra, even of normal density (102).

It is also unlikely that the cement used for PV would be exposed to enough cycles to cause fatigue. Most PV is performed on patients advanced in age (>70 years) whose remaining life span may not be long or active enough to elicit a fatigue response. Because of the relatively recent introduction of the practice of PV, no patients have follow-up of more than 20 years after treatment.

Vertebral Body Tests

As with tests conducted on isolated cement specimens, mechanical tests conducted on vertebral bodies to determine their prefracture (initial) and postrepair structural parameters have been almost exclusively compressive (46,47,62,75). Typically, impressions of the vertebral body endplates are made using a common epoxy to distribute contact stresses across the endplates during compression tests. The potted specimens are placed between loading platens on a materials testing

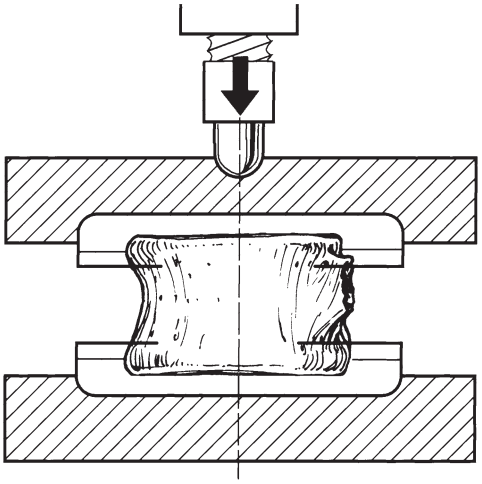

Figure 6.10. Compression test of an osteoporotic vertebral body. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

machine and compressed (Figure 6.10). In this manner, the initial stiffness and failure loads of the vertebral body are determined. The vertebral bodies then are repaired with the particular method under investigation and recompressed. Strength and stiffness values of the repaired specimens then are compared with the initial values to determine the biomechanical effect of the repair (Figure 6.11).

Although the spine is loaded predominantly in compression, the effects of bending and torsional loading should not be ignored. Wilson

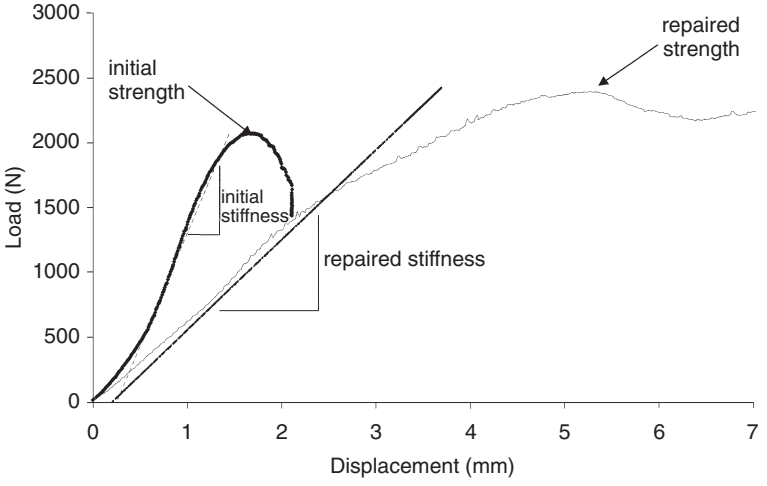

Figure 6.11. Mechanical behavior of an osteoporotic vertebral body during initial compression test and after repair. (From J.M. Mathis, H. Deramond, and S.M. Belkoff [eds], Percutaneous Vertebroplasty. New York: Springer, 2002, with permission.)

et al. (78) used a multisegment cadaver model to investigate potential altered kinematics as a result of kyphoplasty or PV. Although such models have the benefit of evaluating spine kinematics in a more clinically relevant manner than using isolated specimens, it is difficult in the multisegment model to create the simulated fractures needed to evaluate subsequent repairs. Thus, treatments in the study by Wilson et al. (78) were performed on intact (nonfractured) vertebral bodies, and it is unknown what effect the treatments might have on vertebral bodies mechanically weakened by VCFs.

Alternative Cements

Two factors have motivated the development of new types of cements and injection devices: the increasing frequency of the practice of PV and deficiencies in existing PMMA cements for use with PV (46,103–105). These cements are bioactive (106–109) or bioresorbable (103,105,110–113), are naturally radio-opaque (93,105), and have lower exothermic reactions (38,103,105) than PMMA cements.

Until recently (77,105), the use of calcium-phosphate cements in PV has been impeded substantially by their difficulty of injection (103). These more biocompatible cements may eliminate concerns about thermal necrosis and cytotoxicity and appear to result in mechanical stabilization of fractured vertebral bodies similar to that of PMMA (105). Yet, if thermal or chemical mechanisms are found to play an antitumoral role, then the non-PMMA cements may not be as effective for use in patients with tumors. Bioresorbable cements may be most appealing for use in prophylactic augmentation because injected vertebral bodies would be mechanically augmented immediately, whereas the cement would provide an osteoconductive material for subsequent bone repair and remodeling. The subsequent risk of fracture after the cement is remodeled or resorbed is unknown. Bioresorbable cements also may have application with PV for treating burst fractures in young healthy patients (104). Despite the allure of using such cements, some caution is warranted because the calcium may initiate coagulation and clot formation, thus placing the patient at risk for cardiac arrest (114). Many questions regarding the clinical use of these cements remain and need to be resolved through careful investigation.

Summary and Conclusions

The practice of PV has experienced explosive growth in recent years and, with it, many questions regarding the efficacy of the procedure and its optimal practice. Percutaneous vertebroplasty functions primarily to stabilize fractures, thus preventing pain and providing a stable environment for healing. The amount of cement needed to affect stabilization is unknown, but it is probably 4 to 6 mL rather than the volume needed to fill the vertebral body completely (>10 mL), as previously thought necessary. Altering the cement composition by adding antibiotics, opacifying agents, and more monomer alters the material properties of the cement, but with the availability of cements approved

by the Conformitè Europèene or the FDA, such alterations are of more academic than clinical interest. The primary concerns relative to cement selection are whether or not the cement can be injected easily and visualized properly under fluoroscopy.

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