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PMMA: An Essential Material in Medicine and Dentistry

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ABSTRACT: The first use of polymethyl methacrylate (PMMA) as a dental device was for the fabrication of complete denture bases. Its qualities of biocompatibility, reliability, relative ease of manipulation, and low toxicity were soon seized upon and incorporated by many different medical specialties. PMMA has been used for (a) bone cements; (b) contact and intraocular lens; (c) screw fixation in bone; (d) filler for bone cavities and skull defects; and (e) vertebrae stabilization in osteoporotic patients. The many uses of PMMA in the field of medicine will be the focus of this review, with particular attention paid to assessing its physical properties, advantages, disadvantages, and complications. Although numerous new alloplastic materials show promise, the versatility and reliability of PMMA cause it to remain a popular and frequently used material.

KEY WORDS: polymethyl methacrylate, PMMA, bone cement, intraocular lens, vertebral stabilization, cranioplasty

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I. INTRODUCTION

Vulcanite, also known as ebonite or hard rubber, was a standard denture base material for many years. Its chief disadvantage was its lack of esthetic qualities; it was completely opaque to light, and therefore it did not exhibit the translucence necessary to simulate gingival tissues.[1] This disadvantage prompted the development of a wide range of thermoplastic polymers as alternatives.[2] Of particular interest to the dental field was the development of acrylic chemistry. Acrylic acid and its derivatives were well known by the 1890s; however, it was not until 1901, with the availability of solid, transparent polymers of acrylic acid, that this field started to emerge. Derivatives of acrylic monomer, methyl, and ethyl acrylate were developed that produced perfectly clear solid polymers.[3]

In 1931, commercial production of the harder polymethyl methacrylate (PMMA) occurred, with introduction of Plexiglas® in sheet form.[3] By 1937, this material was also available in granules and molding powders.[4] Neurosurgeons began using PMMA during World War II for cranioplasties because of its strength and light weight.[5] By 1946, PMMA materials represented approximately 95% of the denture base market.[4] It was not until after World War II that self curing or so-called "room-temperature polymerization" was made available on a wide basis.[6] Medical research rapidly progressed until in the 1950s PMMA was used for the fixation of orthopedic prostheses of femoral bones.[5]

Continuous medical research has led to the expansion of PMMA use in several areas. Some of the most common uses of PMMA have been: (a) bone cements; (b) contact and intraocular lens; (c) screw fixation in bone; (d) filler for bone cavities and skull defects; and (e) vertebrae stabilization in osteoporosis patients.[6] PMMA's use throughout the dental profession exploded during the 1950s and 1960s. Not only was PMMA used in the fabrication of complete and removable partial dentures, but it was also used in the fabrication of acrylic temporary crowns and bridges, orthodontic appliances, external fixation of mandibular fractures, and maxillofacial prostheses such as maxillary obturators in cleft palate patients and intraocular prostheses.

II. PHYSICAL PROPERTIES

Most modern denture base resins are polymerized by heat activation, which produces free radicals that initiate and propagate the polymerization of the methyl methacrylate monomer.[7] Dentures are placed in a metal flask and invested with a gypsum product. The denture flask is submerged in a tank of water heated to 73.9 °C and cured for 12 hours. The induction system most commonly used in medicine is chemically activated (cold cure) at ambient room temperature. Such systems consist of at least two reactants that, when mixed together, undergo a chemical reaction that generates free radicals. Chemically induced systems consist of two or more parts that must be kept separate during storage.[7] An example of such a system is the use of a tertiary amine such as dimethyl-*p*-toluidine (the activator) and benzoyl peroxide (the initiator), which are mixed together to initiate the polymerization of so-called "self-cured" resins at room temperature. This process, in fact, is a special case of heat activation, because the presence of the amine reduces the thermal energy required to break the initiator into free radicals at ambient temperature.[7]

When pure methyl methacrylate monomer is polymerized, the density changes from 0.94 grams per cubic centimeter to 1.19 grams per cubic centimeter. This change in density results in a volumetric shrinkage of 21%. PMMA by itself is not used to a great extent. Rather, the liquid monomer, methyl methacrylate, is mixed with powdered PMMA polymer in a 1:2 to 1:3 ratio.[8] The monomer partially dissolves the polymer to form a plastic dough. By using prepolymerized polymer, the volumetric shrinkage is reduced to 5–7%.[7,9] It can be calculated on this basis that the acrylic should shrink linearly approximately 2%. The growth of the polymer chain ceases when the reactive center is destroyed by a termination reaction. The entire addition polymerization process can be pictured as a series of chain reactions. The process

occurs almost instantaneously. Any impurity in the monomer that can react with free radicals inhibits or retards the polymerization reaction. The length of the induction period is influenced by the presence of inhibitors as well as the degree of polymerization. A small amount of inhibitor, such as methyl ether or hydroquinone, is added to the monomer to inhibit spontaneous polymerization. The reaction is exothermic, and considerable amount of heat is released.[7]

III. MEDICAL USES

III.A. Orthopedics

PMMA has multiple uses in the field of orthopedics. It was first introduced as bone cement in the early 1960s by Charnley and Smith.[5,10] It has been used widely in joint replacement surgery to fill the space between prostheses and the surrounding bone. A recent publication documented the first PMMA cemented hip implant that had survived for longer than 50 years.[11] PMMA is not a true adhesive, but instead it mechanically interlocks with the surrounding cancellous bone.[12] Although there have been tremendous advances in joint technology for the treatment of hip and knee arthritis, the use of PMMA bone cement has changed little since Harris's description of the third-generation cement technique. The use of PMMA as a bone cement is still a critical element in joint replacement.[13] Research is continuing for ways to decrease the toxicity of bone cements and to improve the integration of implants to bone through the introduction of additives to PMMA.

With the increase in the elderly population within the United States, hundreds of thousands of elderly patients are hospitalized each year due to severe pain caused from osteoporotic vertebral body compression fractures (VCFs). Thrombosis and pneumonia are common complications from osteoporotic compression fractures. Vertebraplasty and kyphoplasty are percutaneous techniques that have been successful in treating painful osteoporotic compression fractures. In 2002, approximately 38,000 vertebroplasties and 16,000 kyphoplasties were performed in the United States.[14] Both procedures have been shown to decrease bed rest, narcotic analgesia use, and the need for bracing. PMMA bone cement is injected through an 11-gauge spinal biopsy needle under radiographic guidance into fractured and collapsed vertebra filling the voids. As the bone cement hardens, the vertebrae are sealed and stabilized, resulting in a reduction in pain in 70–90% of patients.[15]

Kyphoplasty is similar to vertebroplasty in that it internally stabilizes osteoporotic VCFs, which decreases pain and improves function. Kyphoplasty has the ability to reduce osteoporotic fractures and thereby improve spinal alignment by using an inflatable balloon (Kyphon, Inc., Sunnyvale, California is the sole supplier of equipment for kyphoplasty.)[15] Fracture reduction is achieved by using the inflatable balloon to create a void within the vertebral body that allows for the injection or placement of PMMA in a thick, doughy state under low pressure, thereby reducing the risk of emboli and extrusion outside the vertebral body. Kyphoplasty provides the benefits of vertebroplasty, but it also permits the surgeon to improve spinal alignment and reduce spinal deformity as well as decreasing the morbidity associated with kyphosis and loss of height.

Kyphoplasty is almost always performed under general anesthesia and requires overnight hospitalization for observation. In contrast, vertebroplasties are generally performed as an outpatient procedure with local anesthesia. Both procedures can be relatively safe and provide a similar decrease in pain, but kyphoplasty is more costly as a result of hospitalization, special equipment, and anesthesia.

PMMA is also used to reconstruct and stabilize the spine after vertebral body (VB) resection of a metastatic tumor. VB defects are reconstructed using bone autograft or allograft, PMMA, and interbody spacers and/or cages.[16,17] Patients with vertebral metastases in many instances suffer from severe pain and difficulty in walking and at times are bedridden. Until the development of PMMA, replacement of bone due to malignant neoplastic fractures was difficult at best. Large irregular boney cavities can be success-

fully filled following the removal of cancerous bone. PMMA provides stability and strength to the afflicted bone almost immediately, permitting the patient to apply weight to the area and regain normal function with immediate relief of the pain associated with the movement of bony parts.[6,15,18]

Alloplastic materials for the reconstruction of cranial defects in the past have included stainless steel and titanium mesh,[18-20] hydroxyapatite,[21-23] alumina ceramic,[24] and silicone.[25] PMMA is commonly used because of its excellent tissue compatibility, the ease with which it is manipulated at surgery, and its strength, radiolucency, availability, low thermal and electrical conductance, and weight. PMMA is available for use as a moldable product alone, as a moldable product reinforced with metal mesh, and as a preformed molded implant for large defects.[26]

Small defects (5–15 cm^2) are corrected to protect the brain and to reestablish the normal contour of the skull by eliminating flat or depressed areas. The defects are small enough that the remaining skull in the defect area is sufficient to provide adequate protection.[27] The implant is made at the time of surgery and applied directly to the surgical site. The resin is prepared in a sterile stainless steel bowl or glass jar according to manufacturer's instruction. Following the initial mixing of PMMA, there are three stages of polymerization: doughing, handling, and setting. The dough stage is complete when PMMA loses its sheen after approximately 5 minutes. During the handling stage, the material is kneaded in a plastic sleeve and shaped. The dough-like material is applied to the defect, and the excess material is removed. The prosthesis must be held securely in position without movement until the resin is set. With a large and thick prosthesis, it may be necessary to cool the mass during the exothermic phase with sterile water or saline.[26]

Moderately sized defects (16–49 cm^2) are treated with PMMA reinforced with titanium metal mesh. A groove is created in the bone around the border of the calvarial opening. A pattern is made of the defect with sterile paper. The titanium mesh is made several millimeters larger than the pattern so that it can be domed to the proper shape. Several drill holes are made around the calvarial margin, and the mesh can be secured using strands of the mesh or nylon sutures. Before the titanium mesh is finally secured in position, a layer of saline-soaked Gelfoam™ is applied beneath the expected placement of the mesh. After the acrylic is thoroughly mixed, a layer of PMMA is rapidly applied across the mesh so that it penetrates the mesh and the calvarial groove. The mass must be cooled during the exothermic phase with sterile saline irrigation.[20]

Traditionally, large defects (>50 cm^2) were restored with the use of a preformed PMMA prosthesis made from a direct impression of the defect.[28] After the head is shaved, an impression of the defect is made, and a positive cast is poured in dental stone. In addition to the working cast, careful digital palpation and mapping of the area are necessary to locate the internal and external defect margins. The PMMA prosthesis is created indirectly out of wax using the positive cast of the patient's skull. The waxed prosthesis is invested in dental stone within a large metal flask and heat cured. The heat-curing process increases the strength of the prosthesis, reduces porosity, eliminates the problem of heat generation during polymerization, and reduces the residual monomer.[27] Holes are drilled in the preformed acrylic plate for in-growth of connective tissue. After the acrylic plate is sterilized with ethylene-dioxide, it is secured to the surgical area with Vicryl® sutures.[28] Most cranial implants need adjustment or augmentation at the time of surgery to compensate for lack of marginal fit and improper contours. Cranial implants made from external impression techniques often have flatter contours than the skull.[27]

Prefabricated cranial implants can now be fabricated with the combination of computed tomographic scans (CT) and the generation of a three-dimensional model of the cranial defect with CAD/CAM design reformation.[27] A written scan protocol is provided to the CT scan test site by the cranial cast manufacturer (Techmedia, Camarillo, California).[27] The protocol provides the specific settings required for the CT examination. The model is manufactured from standard image data produced by a GE CT 9800 scanner (General Electric, Milwaukee, Wisconsin).[27]

The number of slices needed to record the defect is determined by the technician. Increasing the number of CT scans improves the model accuracy. CT scans made 3 mm apart are adequate for bone edges perpendicular to the slice. However, scans made of bone edges parallel to the slice need to be closer to ensure accuracy of the model. A slice separation of 1 mm provides adequate accuracy for bone edges parallel to the CT scan beam. The completed scan can be reformatted and viewed as a solid body or in parts as defined by the technician using the image reformation functions available on the CT device.[26,27]

The data are archived onto a magnetic tape and sent directly to the cast manufacturer. The information is translated, and the resulting bone-edge contours are converted to machine-tool language and are used to drive a milling machine. Originally, the machine milled the casts from stacks of polycarbonate slices, which were indexed and glued together. Now the technique creates a life-sized cast from a solid plastic resin block.[29] The model manufacturer returns the cast for implant fabrication. The cast can have a solid or hollow core, depending on the size and location of the defect.[26,27] The implant is fabricated with the same technique used when the defect is recorded directly form the patient.

III.B. Ocular Devices

Contact lenses are classified as either hard or soft according to their modulus of elasticity.[30] Although more durable, hard lenses tend to be less well tolerated by the wearer and require longer adaptation periods. The first hard or rigid polymeric lenses were those manufactured by Kevin Tuohey in 1948 from PMMA.[30] PMMA lenses are lathe cut from rods or buttons of PMMA obtained by bulk free-radical polymerization of methyl methacrylate. Although PMMA possesses the favorable optical properties of light weight, surface wetability, and durability, the low oxygen permeability of this material limits the long-term wear of these lenses.[30] To reduce problems associated with corneal anoxia, PMMA lenses tend to be small in diameter and float on the precorneal tear film, thereby allowing oxygenation of the cornea via tear film exchange during blinking and movement of the lens.[30]

Cataract extraction and intraocular lens implantation is the most frequent form of ophthalmic surgical procedure, with over 1.6 million operations being performed every year in the United States alone.[30] The surgical method most frequently employed in the developed world for the treatment of this condition is extracapsular cataract extraction, which involves the subsequent insertion of an intraocular lens (IOL) to compensate for the loss of the natural crystalline lens. Of increasing popularity is the technique of phacoemulsification, in which the hardened nucleus of the crystalline lens is emulsified and removed through a 3.5 mm incision in the eye, as opposed to a 14 mm incision for an extracapsular cataract extraction. The advantages of this technique lie in the positive placement of the IOL into the capsular bag and less surgically induced astigmatism because of the small wound. This results in more rapid rehabilitation for the patient.[30]

PMMA has been the standard IOL material since the surgical approach was first developed by Harold Ridley in 1949. The IOLs are generally lathe cut from PMMA rods or buttons. The standard IOL consists of a central optic, which is supported by haptics, projections from the main body of the lens that provide support in the eye. The haptics are usually constructed from PMMA or the base material of the optic, although other polymers may be used. These posterior chamber lenses may thus be inserted wholly within the capsular bag or with the optic supported within the capsular bag remnants and the haptics lodged in the ciliary sulcus, the anatomical groove between the iris and the ciliary body.[30]

IOLs reside within the eye as foreign bodies and were largely regarded as inert until recently. Indeed, PMMA was selected as the material of choice because of its low weight and biocompatibility. Although this material has been used for over 40 years, major problems still occur as a consequence of its relatively low surface energy, which may result in both corneal endothelial damage on insertion and postoperative ad-

hesion of inflammatory cells to the IOL. Uveal contact with the IOL surface has also been shown to cause increased and prolonged postoperative intraocular inflammation, which may lead to iris adhesion to the IOL, uveitis, breakdown of the blood retinal barrier causing cystoid macular edema, and loss of vision.[31]

Phacoemulsification, with its small incision, has encouraged the development of foldable IOLs for implantation. The need to insert these devices through a 3.5 mm incision has also encouraged companies to investigate other design modalities, including acrylic and silicone foldable IOLs. Between the two, acrylic lenses lead to a lower incidence of posterior capsule opacification and a higher rate of stability in the capsular bag.[32] In developing these materials, particular emphasis has been placed on the handling, foldability, and unfolding characteristics of the lenses, because they must be easy to insert, unfold slowly, and leave no crease mark. In this respect, it has been reported that the acrylic lens unfolds more slowly and in a more controlled fashion than the silicone lenses; the higher refractive index of the acrylic material gives rise to a thinner IOL.[30] Recent studies suggest that phosphorylcholine-based acrylate polymers may have application in the development of novel biocompatible foldable IOLs.[30]

IV. PROBLEMS WITH PMMA

There are advantages and disadvantages with any material used in medicine. There have been problems with infections occurring with prostheses cemented with PMMA, cement leakage, toxicity from leakage of methyl methacrylate monomer, tissue necrosis due to high thermal temperatures generated during polymerization, sensitization from handling PMMA, loosening of prosthetic implants, and inhalation of vapors by medical and dental staff. [6,33]

The development of a bacterial infection is a major problem that can lead to a total hip replacement (THR) prosthesis having to be removed, resulting in loss of function.[6] Staphylococcus and its variants are the most commonly occurring infection[34,35] (Table 1). In 1970, Buchholz and Engelbrecht[36] introduced PMMA bone cements loaded with gentamicin, followed in 1977 by Klemm introducing gentamicin PMMA beads to treat osteomyelitis.[36] Since antibiotic-loaded PMMA bone cements were first introduced in the 1970s, research in this area has never ceased. Research has shown that there is initially a rapid release of low doses of antibiotics from loaded PMMA bone cements or beads, followed by elution that progressively decreases over a period of time ranging from a few weeks to several months.[6,37-39]

TABLE 1. Most Prevalent Organisms Causing THR Infections

Organism	Percent
Staphylococcus aureus	50–60
*Staphylococcus epidermidis**	25–30
Bacteria, fungi, and mycobacteria	10–15

* *Staphylococcus epidermidis* is increasingly more prevalent as a pathogenic organism for infections about prosthetic joints.[35]

Today, antibiotics are delivered to the surgical site prophylactically in an attempt to decrease the occurrence of THR infections.[40,41] An evaluation of 10,950 THRs demonstrated that there was a lower incidence of THR revision when systemic antibiotic therapy was used in combination with antibiotic-containing cement.[42] If a THR infection does develop, a two-stage protocol is customarily used, with the implantation of a temporary antibiotic-loaded PMMA hip spacer (resembling the hip prostheses in shape) following the removal of the infected prosthesis and adequate surgical debridement.[40] Antibiotic impregnated PMMA bead chains on a surgical wire (resembling a pearl necklace) are used to treat local musculoskeletal infections.[41] The advantage of using beads compared to parenteral therapy is that they deliver a high concentration of antibiotics locally while avoiding high systemic concentrations. Bone cements containing penicillins, fluoroquinolones, and aminoglycosides are being investigated as carrier systems for the local delivery of antibiotics.[37,40,43,44]

During PMMA polymerization, there is a tremendous release of heat, which can be damaging to surrounding tissues. Temperatures exceeding 56 °C can cause protein denaturation and tissue damage.[5,45] Called *heat necrosis,* this phenomenon has been implicated as the cause of bone damage at the cement interface and subsequent component loosening. Research to investigate the phenomenon of exothermic heat generation in medicine has produced mixed results. In vitro testing produced temperature variations from room temperature to 100 °C, while in vivo temperatures generated at the cement/bone interface has been documented at 48 °C maximum.[5,46] These findings have left researchers questioning whether exothermic heat generated from polymerization was indeed the cause of bone interface necrosis.

Research into bioactive bone cements (BBC) has been conducted as a result of the concern about high temperatures during polymerization. BBCs consisting of PMMA with hydroxyapatite (HA) or chitosan has shown that exothermic temperatures of the BBCs were considerably lower those that of pure PMMA.[47,48] Castaldini[48] reported that PMMA with HA was shown to reduce potentially harmful heat generated during PMMA polymerization.

In dentistry, during the fabrication of acrylic temporary crowns directly in the mouth, exothermic heat can be harmful to the dental pulp. Timing is of the utmost importance in removing the acrylic temporary crowns from the mouth just as the polymerization process enters the period of exothermic heat generation. Newer dental materials for making temporary crowns are available that consist of acrylic and bisphenol-*a*-glycidyldimethacrylate (Bis-GMA). Bis-GMA materials generate less exothermic heat and set up faster, and it is easier to correct fabrication errors with these materials, but they are more brittle than acrylic temporaries made of PMMA, especially when used for long-span bridges.

Yamamuro et al.[10] developed a BBC consisting of CaO-SiO_2-P_2O_5-MgO-CaF_2 (AW glass-ceramic) powder and Bis-GMA. Their research showed that BBCs exhibited overall better physical properties than PMMA. BBCs were superior to PMMA in generating less heat when measured in their center and outer surface and exhibited greater compressive, bending, and tensile strength and fracture toughness. Histologically, BBCs actually bonded to bone at 3 and 6 months postsurgery "through a Ca-P rich layer," whereas the PMMA cement revealed a fibrous tissue layer that intervened between the PMMA and bone. Crystallography confirmed that the Ca-P layer was an apatite layer measuring 30 μm in thickness, further proof that PMMA bone cement does not bond to bone, but is only mechanically interlocked with the cancellous bone.[10]

Although much research has been conducted on the effect of heat necrosis, not all researchers have been in agreement that exothermic temperatures generated during polymerization of PMMA cause are its cause. Research by Swenson et al.[49] in 1981 concluded that the heat generated during cementing of an implant was not the primary cause of junctional bone necrosis. Reckling and Dillion[46] concluded from their research that high temperatures didn't occur at the cement/bone interface but did in the interior of the polymerizing cement. Thick bone cement—i.e. greater than 10 mm—does have the ability to obtain the high temperature capable of the protein coagulation seen in bone necrosis. PMMA beads are not as controversial as PMMA bone cements because they are fabricated outside the human body. It is now believed that the early bone damage seen at the cement/bone interface is due to the local chemical effect of PMMA monomer as well as the trauma of the mechanical preparation of host bone.[5,49]

Surgeons have become sensitized and have developed contact dermatitis from repeated use of PMMA, even while wearing surgical gloves. Dental laboratory technicians are more prone to developing contact dermatitis because of hand mixing PMMA in its dough stage during the fabrication of acrylic dentures. The monomer component of PMMA can have toxic local and systemic effects. These effects are due to the release of monomer during the first 5–10 minutes of PMMA polymerization. Unfortunately, this period can not be bypassed, because PMMA bone cement is best applied during this timeframe.

Systemically, varied amounts of liquid monomer is released within the first 15 minutes of the polymerization process and is absorbed by the blood. A direct relationship between vasodilatation and subsequent hypotension during cementing has been made.[50] How much free monomer must be absorbed in the blood before hypotension can develop has not been determined. Other systemic effects include respiratory inhibition and pulmonary, hepatic, and renal cellular damage. These findings have resulted in animal studies after large doses have been administered but have not been confirmed in patients after bone cement has been used as in total hip replacement.[51] Systemic monomer is eliminated during the respiratory exchange process.[52]

After curing has been completed, there is as much as 10% of the residual unreacted methylmethacrylate monomer left in the mass.[6] Upon completion of the polymerization process, only minimal amounts of monomer is released during the life of the prosthesis. Any residual "unpolymerized monomer" is trapped within the cured cement and does not wash out over time. Chronic systemic toxicity has not been an issue.[6,50]

The local chemical effect of PMMA and the trauma of preparing the surgical site have now taken precedent as the cause of bone damage at the cement/bone interface.[51,53] Because of its low solubility, only a small amount of methyl methacrylate monomer is systemically absorbed. The vast majority of monomer resides at the interface and disseminates into the local tissues. Research has proposed that PMMA is extremely cytotoxic as a result of the local effect of unpolymerized monomer. The combination of the mechanical trauma of site preparation, the placement of the implant, and the local chemical effect of PMMA result in a zone of necrosis at the cement/bone interface.[6,46,54] The documented outcome is bone resorption followed by bone remodeling. A foreign body response similarly seen in other tissues is created by methylmethacrylate. Animal studies have shown a dense fibrous tissue formation encapsulating the cemented implants. Fibroblasts, multinucleated giant cells, and macrophages have all been reported as the different types of cells present. The response of these cells has been implicated as the possible cause of junctional tissue damage, lyses, and component loosening.[55,56]

The principal risk of vertebroplasty procedures is leakage of PMMA into paravertebral soft tissues, the spinal canal or neural foramina to the adjacent discs or the veins.[57] PMMA cement leakage occurs 34–64% of the time and is considered not clinically significant.[15,24,26,27] If leakage is detected during fluoroscopy, injection of PMMA should be stopped because mechanical consequences with adjacent vertebrae can develop. Reported complications include increased pain, radiculopathies, pulmonary embolism from PMMA, and infection. These complications relate to cement leakage leading to radiculopathy and spinal cord injury and potential complications of subsequent surgical decompression.

In contrast, kyphoplasty complications are reported by the surgeons to the manufacturer of the devices. Anaphylactic reactions to PMMA has been documented in the US Food and Drug Administration's Manufacturer and User Facility Device Experience (MAUDE) database, which details medical complications that occur with medical devices used in specific procedures. Nussbaum et al.[14] identified complications in 52 patients who underwent either kyphoplasty or vertebroplasty between 1997 and March 2003. Four deaths were reported to MAUDE as reactions to PMMA: one for kyphoplasty, three for transpedicular vertebroplasty, and one for lateral vertebroplasty.

Inadequate ventilation can be detrimental to neurosurgeons, orthopedic surgeons, and dental personnel who are occupationally exposed on a daily basis to methylmethacrylate (MMA), the liquid component of PMMA. It has been reported that exposure of medical staff to MMA vapors as high as 100 ppm has occurred.[58] Experimental and clinical studies have proven that MMA can cause irritation to skin, eyes, and mucous membranes, allergic dermatitis, stomatitis, asthma, paresthesias in the fingers, neuropathy, disturbances of the central nervous system, liver toxicity, and changes in the blood parameters.[59-64] Inhalation studies have been conducted as a result of widespread exposure to MMA.

The chronic exposure of MMA has been examined by various routes of exposure in rodents, and MMA was found not to be carcinogenic.[62] Most medical and dental personnel aren't aware that MMA vapors are a toxic organic vapor that will be inhaled even if face masks are worn. Wesley and Brinsko[65] strongly recommend that a filtering system be used to lower the levels of MMA that can be inhaled.

V. CONCLUSION

PMMA is used extensively within the medical and dental profession for a wide variety of procedures. Without the development of PMMA, modern medicine and dentistry would have difficulty in providing the quality of dental care it does today. PMMA is a relatively inexpensive material, it can be used indirectly or directly during clinical or surgical procedures, it is easy to use, and its properties have gradually been improved with additives. The use of PMMA has been a key factor in the advent of joint replacement as a surgical option, decreasing pain and increasing mobility for patients suffering from osteoporosis or vertebral metastases, or the placement of IOLs to correct vision deficiencies due to cataracts. There have been documented side effects that have resulted from the use of PMMA, but its use has far outweighed its negative characteristics.

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