Properties of Bone Cement: Testing and Performance of Bone Cements

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Summary

Although all commercially available bone cements are based on polymethylmethacrylate and other acrylic co-polymers, they all differ in their precise chemical formulation and composition. This results in different physical properties like viscosity, heat release, and mechanics. These differences affect surgical handling and clinical outcome. Various testing methods of bone cement are discussed in this chapter. Clinically most relevant is fatigue testing and traditional cements perform best.

Introduction

General

All cements which found widespread use in orthopaedic surgery are based on polymethylmethacrylate (PMMA). This acrylic resin is used now for over 50 years for the fixation of orthopaedic implants. The first published artificial joint implantations occurred in 1949 at Copenhagen by M.S. Kiaer and in 1951 in the hospital for Joint Diseases at New York by E. Haboush [5]. The first commercially available bone cements were released to the market in the beginning of the seventies. Since then many types and makes of bone cement have been introduced. Only few stood the test of time.

Alternative Cements

Many attempts have been made to improve the physical properties of bone cements and many alternatives for acrylic were tested, like:

- glass-ionomeric cements,
- bioactive glass cements,
- resorbable cements.

The main advantages of ionomeric bone cements are the absence of heat generation during polymerization and its adhesive properties to bone. The main disadvantage is its low mechanical strength which makes it unsuitable for load bearing applications. Bioactive glass cements are a composite of bioactive calcium-phosphate (CaP) powder and a high molecular weight acrylic matrix. The mechanical strength is 2 to 3 times higher than of acrylic cement. It has less heat generation and less shrinkage during polymerization. Main disadvantage is its high rigidity and brittleness. It is weak in tensile fatigue loading. Resorbable cements like CaP and polypropylene-fumarate cements all suffer from brittleness and insufficient strength in load bearing applications [3].

Improvement of acrylic cements has been tried (▶ chapter 3.6) in many ways like:
- addition of CaP powders,
- addition of artificial fibres,
- modification of the curing mechanism,
- modification of the radiopacifier.

By addition of CaP powders as a filler material to cement, one has tried to enhance bony ongrowth to the cement surface and bony ingrowth into the cement mantle. Simultaneously, it would decrease the exotherm reaction. To obtain ingrowth high amounts up to 30 to 50 w/w% of CaP powder are needed to obtain a sufficient open structure. This open structure weakens the strength of the cement considerably. In the Far East such cements have been applied for the fixation of endoprotheses.

Addition of artificial fibres is meant to increase the mechanical strength. Fibres will increase the static fracture

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strength, the modulus of elasticity and the fatigue strength. Creep is diminished and fracture toughness is increased. Many fibres like Kevlar, carbon, glass and PET have been tested in a magnitude of 1 to 2 w/w%. The major drawback of artificial fibres is the long term biological effect of small wear particles. Many materials which are fully biocompatible as block material will give rise to tissue reactions if they are released on a microscopical scale. No artificial filler materials are at present applied clinically.

Test Standards

ISO standard 5833, which was first released in 1979 and latest revised in 2002, is a standard which describes a number of test methods and minimal requirements for acrylic bone cements [6]. All commercially available cements have to fulfill the requirements set forth in this standard. Unfortunately, the test methods and requirements are set on a low level and can be easily met. Therefore, this »standard« is not capable to discriminate whether a cement is suitable for clinical application or not. The ISO 5833 would for example find a setting time of just 3 minutes acceptable for a doughy cement. Even Boneloc cement, which had dramatic clinical results, fulfilled all requirements of the ISO 5833 standard.

In particular, a straight forward tensile test is missing in ISO 5833. Bone cement is remarkably weak in tension, but relatively strong in compression. It is also much more brittle in tension than in compression. In the 1992 version of this standard a bending test was added. Bending does include a tensile component, but the requirements of this bending test will be easily passed by all available cements. Most importantly there is not any type of fatigue testing in the ISO standard. This type of testing was recently described in ASTM standard F2118–2001 [1]. This standard accurately describes a method for a fully reversed tensile and compression cyclic loading test of acrylic bone cement. Unfortunately, the test does not state a minimum requirement.

Running a fatigue test is a very time-consuming procedure and therefore expensive test. The test results will highly depend on the mixing conditions (temperature, vacuum) and the resulting porosity of the test specimens. Only very few papers have been published which compare fatigue data of bone cements.

Effect of Chemical Composition

Polymers

The type of polymer powder is the most important factor which characterizes the performance of a particular type of bone cement. The most commonly used polymer powders are methylmethacrylate (MMA) homopolymer, methacrylate (MA) copolymer, butylmethacrylate (BMA) copolymer and styrene copolymer. They are applied in various commercially available bone cements (■ Table 3.1).

The addition of MA in the MMA-MA copolymer results in a change of physical properties as compared to MMA homopolymer. MA is a small molecule which makes the cement more hydrophilic and flexible (■ Fig. 3.24). The hydrophilic nature of MA speeds up the monomer absorption and powder dissolving. Hence the higher the MA concentration the higher the cement's initial viscosity will be. The addition of MA will mechanically result in a more flexible cement with a higher failure strain, relative low compression strength and a relative higher strength and failure strain in tension. Bone cement is a brittle material, which tensile strength is very susceptible for stress risers like air voids. MA cements will be less influenced by porosity due to this flexible behaviour.

Addition of a small percentage BMA, which has a higher molecular weight than MMA, gives the powder a more porous open structure. This may enhance the bond between the polymer matrix chains, which will entangle with the outer surface of the beads. A small percentage of BMA is claimed to result in better mechanical properties [7].

Styrene cements have a more hydrophobic behaviour. The time needed to obtain a homogenous mixture will take longer than for an MMA-MA cement. Addition of styrene copolymers is thought to be beneficial for the fatigue strength. No data exists whether this is true.

■ **Table 3.1.** Types of polymer used in various bone cement powders

Type of Polymer	Cement Brand
MMA homopolymer	CMW1, CMW3, Cemex RX, Cemex System, Zimmer regular+LVC
MMA-MA copolymer	Palacos R, Palamed, Osteopal, SmartSet HV, Versabond
MMA-BMA copolymer	Sulfix-6, Boneloc, Biolos
MMA-Styrene copolymer	Surgical Simplex RO, Osteobond, CMW Endurance

Hydrophilic	Methylacrylate	Compact
↕	Ethylacrylate	
	Methylmethacrylate	
	Butylmethacrylate	
Hydrophobic	Styrene	Voluminous

■ **Fig. 3.24.** Properties of various types of acrylic molecules

Monomers

The types of monomer molecule in the cement's liquid are of particular importance for the amount of heat generation. By reducing the number of molecules per gram of powder, the exotherm can be reduced. This can be accomplished by increasing the powder/liquid (P/L) ratio or by using high molecular weight monomers like BMA (M=140) or even isobornylmethacrylate (M=223) and n-decyl methacrylate (M=226). This method is applied by various manufacturers, who wish to market a reduced temperature bone cement, like e.g. Boneloc, Sulfix-6 and Cemex RX (◘ Table 3.2).

By partially replacing methylmethacrylate (M=100) liquid molecules by a higher molecular weight monomer, the total number of monomer molecules is reduced. This results in less heat generation. A similar effect can be obtained by increasing the P/L ratio, which normally is 2.1 w/w. By increasing the P/L ratio to 3 w/w, the number of molecules and the amount of heat generation is reduced by 30%. From a chemical point of view a P/L ratio of 3 is acceptable. From a handling point of view it will become more difficult to obtain a homogeneous mixture.

Radiopacifier

To make cement visible on a radiograph it is needed to add a radiopaque element to the cement. Commercially used are $BaSO_4$ and ZrO_2. The advantage of ZrO_2 is that a relatively better contrast can be obtained than with a similar weight amount of $BaSO_4$. Another advantage of ZrO_2 is that it has no tendency to cluster like $BaSO_4$ (◘ Fig. 3.25). Clustering leads to inclusions in the cement mass which may decrease the mechanical properties. Attempts have been made to diminish the mechanical drawback from including a radiopacifier. Experiments showed that adding a radiopacifier of submicron size significantly increased the fatigue strength of bone cement [8]. Other attempts were made to replace the addition of a radiopaque powder by building iodine into the polymer molecules [2]. Such cement also showed a remarkable increase in fatigue strength. Due to risk of iodine allergy this development was never commercialized.

Handling

Handling is the most critical parameter for cement use in the theatre. In particular when applying modern cementing techniques like vacuum mixing and cement pressurization more working time is needed (◘ Table 3.3). The working time of a cement can be extended by applying a no-touch technique in which case the cement is handled immediately after the start of mixing inside a mechanical

◘ **Table 3.2.** Traditional and low temperature bone cements and their types of liquid and powder/liquid ratio

Cement Brand	Monomer	P/L Ratio w/w
Palacos R, Palamed, Osteopal, Surgical Simplex, CMW1+3, Zimmer regular + LVC	100% MMA	± 2.1
Boneloc	50% MMA + 20% isobornylMA + 30% n-decylMA	2.3
Sulfix-6, Duracem 3	85% MMA + 15% BMA	2.3
Biolos 1	86% MMA + 14% BMA	2.8
Cemex RX	100% MMA	3.0

◘ **Table 3.3.** First, second and third generation of cementing technique

Generation	Method	Year
First	Mixing with bowl and spatula	1965
	Cement kneading	1965
	Finger packing	1965
Second	Cement gun and syringe	1970
	Bone plug	1975
	Retrograde injection	1975
	Pressurization	1980
	Bone lavage	1980
	Low viscosity cement	1980
Third	Vacuum mixing	1985
	Prosthetic positioning by spacers	1987
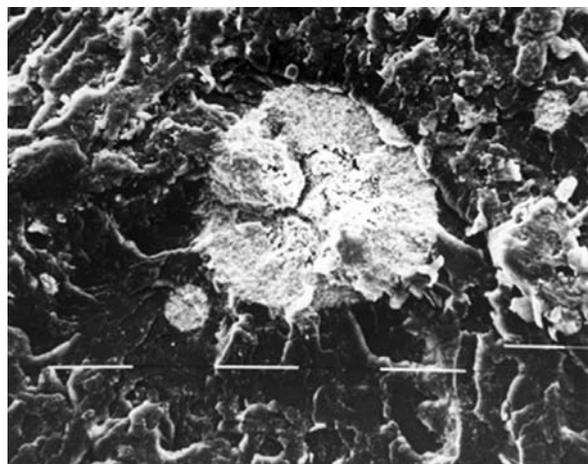
◘ **Fig. 3.25.** SEM of clustered $BaSO_4$ in fracture surface of tensile specimen (*white line* = 100 μm.)

Ø12.0

120.0

3 mm Bone cement

Dummy femur (POM)
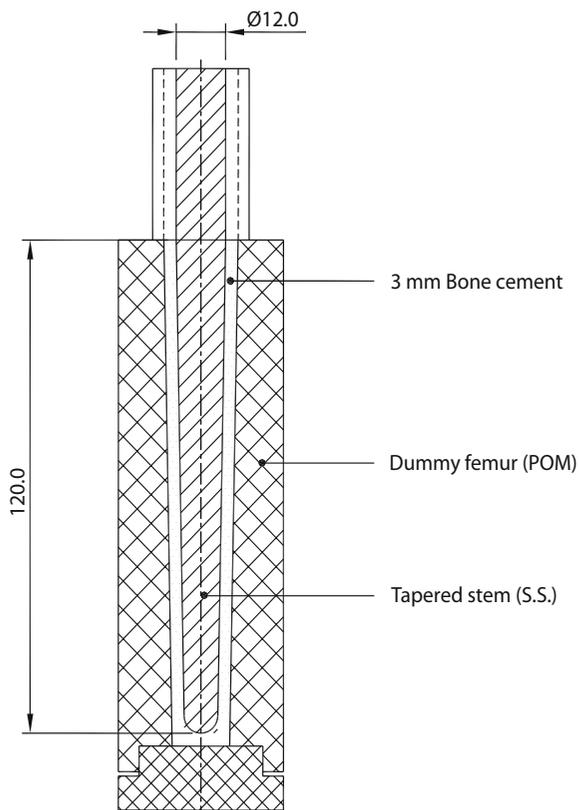
Tapered stem (S.S.)

Fig. 3.26. Test setup for a simulation hip implantation test

mixing system. If even then the working time is not sufficient, the polymerization process can be slowed down by pre-cooling the cement and mixing system.

Manual handling of cement can be divided in various stages: Mixing until homogeneous, doughing stage, kneading, working period and hardening. A time schedule for such a procedure can be derived from simulation hip implantation tests at specific ambient temperatures. ▣ Figure 3.26 shows a test set up which can be used for such an experiment.

Manual mixing in an open bowl generally takes 30 seconds for a hydrophilic MA cement and up to two minutes for a hydrophobic styrene cement. Doughing stage is the period during which the cement polymerizes until it can be picked up by hand. Then the cement is kneaded to a roll and filled in a syringe to be extruded into the bony cavity. During the working period the cement can be extruded, pressurized and a prosthesis must be inserted. The hardening period is the time needed to complete polymerization. ▣ Figure 3.27 shows a time schedule of non-cooled Palacos R for such a procedure, at various ambient temperatures. ▣ Figure 3.28 shows a time schedule for cooled Palacos R cement mixed in a Cemvac syringe vacuum mixing system. Mixing time is longer than for bowl mixing. Still the working period is much longer because there is no doughing time. The setting time is extended two minutes due to cooling of cement.
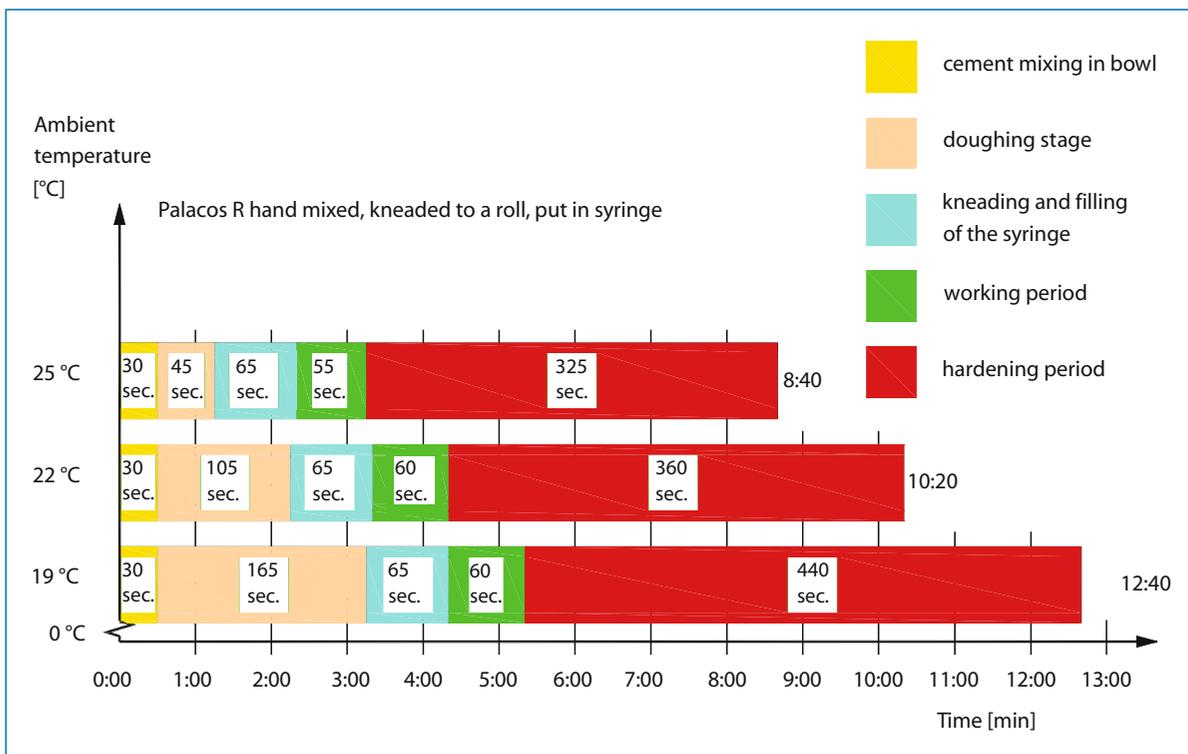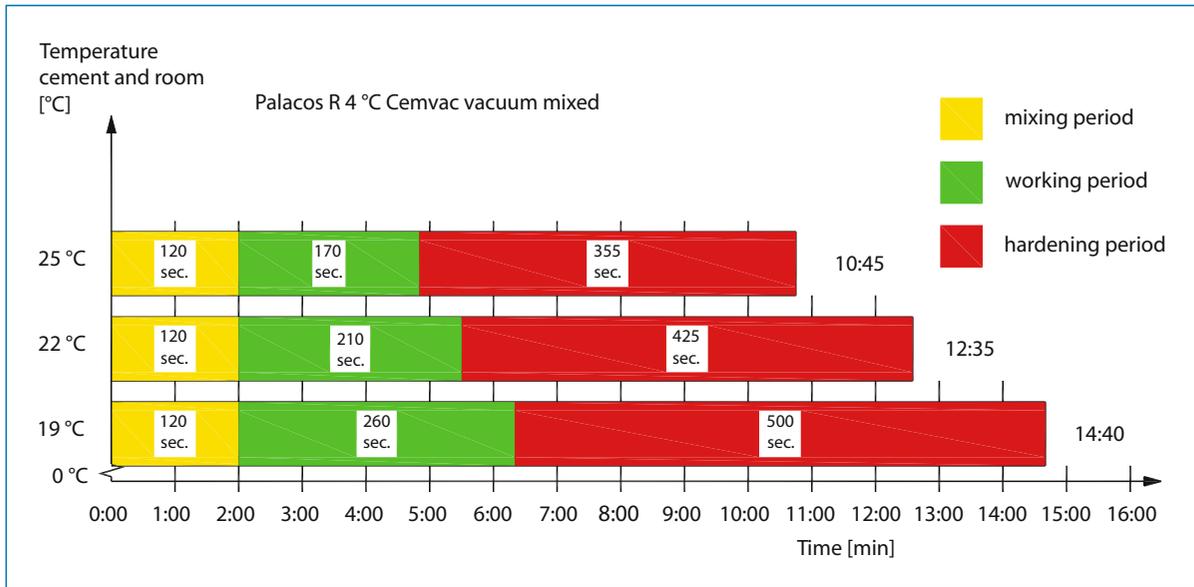

Fig. 3.27. Time schedule for manual kneading and syringe extrusion of non-cooled Palacos R bone cement at various ambient temperatures.

Note the short working time of non-cooled high viscosity cement if used in a syringe application. The dummy femur was pre-heated to 30 ºC

Fig. 3.28. Cooled Palacos R mixed in a Cemvac vacuum mixing system. Note the increased setting time compared to Fig. 3.26 as a result of cement cooling to 4 °C. The dummy femur was pre-heated to 30 °C
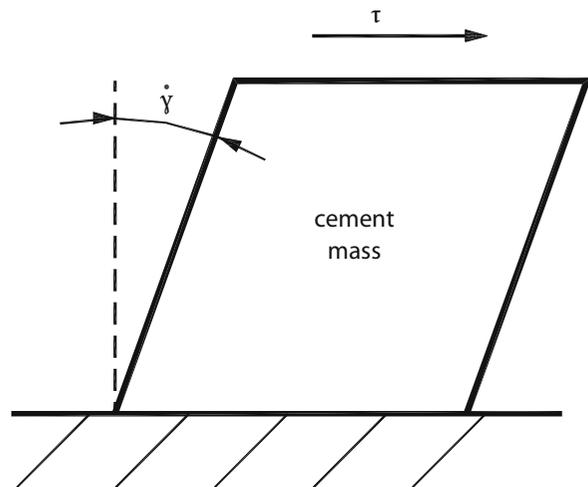
Viscosity

General

Viscosity is defined as the resistance of a fluid to shear deformation (■ Fig. 3.29):

$$\eta = \frac{\tau}{\dot{\gamma}}$$

The higher the viscosity of a bone cement is, the more difficult it will be for a surgeon to extrude the cement through a nozzle or to insert a prosthesis into the cement mass. Hence the development of the cement's viscosity during the polymerization process is an important parameter determining the handling of the cement.

When cement polymerizes it transforms from a fluid to a solid material. At the beginning of the polymerization process, cement is predominantly a fluid with viscous properties, at the end it is transformed to a solid phase with elastic properties. During the transformation, bone cement has both viscous and elastic properties and hence it is called a visco-elastic material with both viscous energy dissipating properties and elastic energy storage properties. This transformation can be demonstrated by measuring the dynamic visco-elastic properties with a rheogoniometer. In ■ Fig. 3.30 the dynamic viscosity development is shown for Sulfix-6 bone cement. In the beginning, when the mixed cement is more or less a suspension of polymer beads in the monomer fluid, the elastic properties are minor. During the polymerization process, the monomer will form elastic chains and they

Fig. 3.29. Viscosity η (Pa.s) is defined as shear stress τ (Pa) divided by the resulting shear rate γ̇ (rad/s)

will integrate with the dissolved outer surface of the polymer beads. Simultaneously, it will become more difficult to shear the cement. This results in an increase of the elastic and viscous properties. When the matrix of newly formed chains is more or less complete and the cement starts behaving as an elastic material, the dynamic viscosity drops again and can no longer be measured. Finally, a solid material with elastic properties remains. What rests from the fluid behaviour are the creep or so called cold flow properties. They are, however, measured on a much larger time scale.
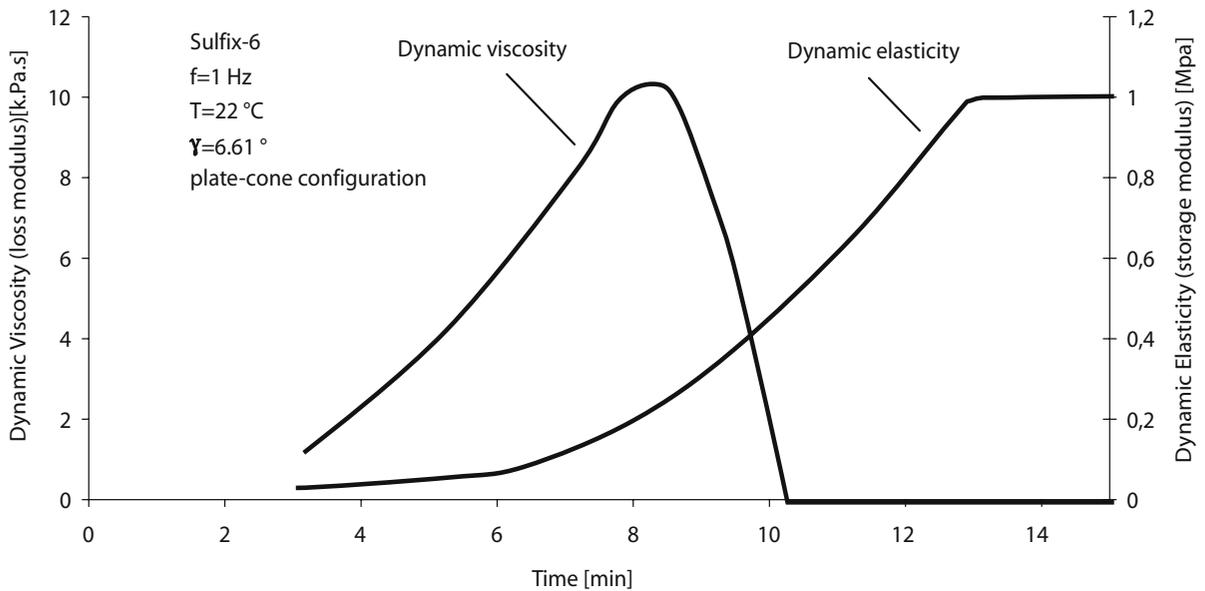
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Sulfix-6
f=1 Hz
T=22 °C
γ=6.61 °
plate-cone configuration

Fig. 3.30. Dynamic viscosity of Sulfix-6 bone cement. The graph shows the development of the viscous energy dissipating modulus and the elastic energy storage modulus as a function of time after mixing. The graph is measured with a Weissenberg rheogoniometer with cone-plate configuration at a frequency of 1 Hz

Apparent Viscosity

When cement viscosity is measured during simple shear at a constant shear rate, no distinction can be made between viscous and elastic properties. The measured viscosity then is called apparent viscosity. Cements are often named and characterized by the height of their apparent viscosity, to wit: high, medium or low viscosity. High viscosity cements are those which were originally developed for manual application, such as Palacos R and CMW-1. These cements can easily be rolled and kneaded and applied manually. They exhibit a high initial viscosity (Fig. 3.31). The high initial viscosity enables rapid manual handling. If high viscosity cements have to be used in a syringe system, cooling is recommended. Such cooling does not have to be at refrigerator temperature. Cooling to 15 or 18 °C already lowers the viscosity sufficient to enable syringe application.

Most low viscosity cements were developed much later for ease of handling in syringes with long thin nozzles for retrograde cement injection. Well known low viscosity cements are: CMW3, Sulfix-6, Osteopal, Palacos E-flow, Zimmer LVC, Cerafix and others. In Fig. 3.31 it is shown that the starting viscosity of these cements is very low. This makes it impossible to handle them manually in an early stage. As all cements polymerize roughly in the same time span, it means that the working time for low viscosity cements is less than for medium or high viscosity cements. They are difficult to contain if they are extruded too early and they polymerize more rapidly at
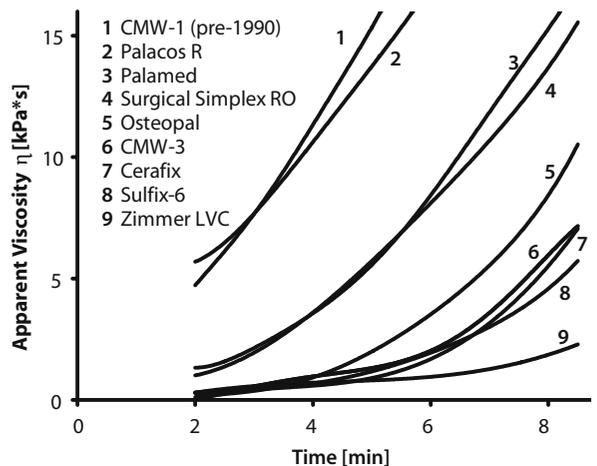

1 CMW-1 (pre-1990)
2 Palacos R
3 Palamed
4 Surgical Simplex RO
5 Osteopal
6 CMW-3
7 Cerafix
8 Sulfix-6
9 Zimmer LVC

Fig. 3.31. Apparent viscosity of various types of bone cement as a function of time after mixing. Note the difference in starting viscosity of high (*1, 2*), medium (*3, 4*) and low (*5, 6, 7, 8, 9*) viscosity cements

the end of the working stage. Figure 3.31 shows a more exponential viscosity increase for low viscosity cements and a more linear viscosity increase for high viscosity cements. In general the behaviour of low viscosity cement is therefore more critical to ambient temperature and time schedule.

A few cements exist with an intermediate viscosity development. Examples of these medium viscosity cements are: Surgical Simplex RO and Palamed. Their viscosity enables both manual application and syringe appli-

cation in an early stage. Even if one needs more working time for vacuum mixing and pressurization, these cements can be applied without the need for cooling.

Factors Affecting Viscosity

The speed of the polymerization process is temperature dependent. Therefore ambient temperature affects the viscosity development and setting time. ▣ Figure 3.32 shows the effect of ambient temperature on the apparent viscosity as a function of time after mixing. Roughly 1 °C ambient temperature increase results in ½ minute reduction of working and setting time.

Cooling cement prior to surgery will change its handling properties. The viscosity development slows down and the setting time increases. ▣ Figure 3.33 shows the effect of cooling Palacos R to 4 °C. Its viscosity curve now has moved to the medium viscosity area. If not only the cement but also the environment is cooled to 4° C, than this high viscosity cement behaves like a low viscosity cement.

Viscosity is determined by the speed at which the powder is dissolved in the monomer. Variables which affect this process are for example the amount of outer surface of the beads, the amount of hydrophylic molecules and the powder/liquid ratio. A hydrophylic molecule like MA will rapidly absorb the monomer and increase the cement's viscosity. The main difference in the Palacos cement family is the amount of MA as part of the total

cement mass. By decreasing the amount of MA from 10% to 5% w/w, the behaviour is modified from high viscosity cement to a low viscosity cement (▣ Fig. 3.34).
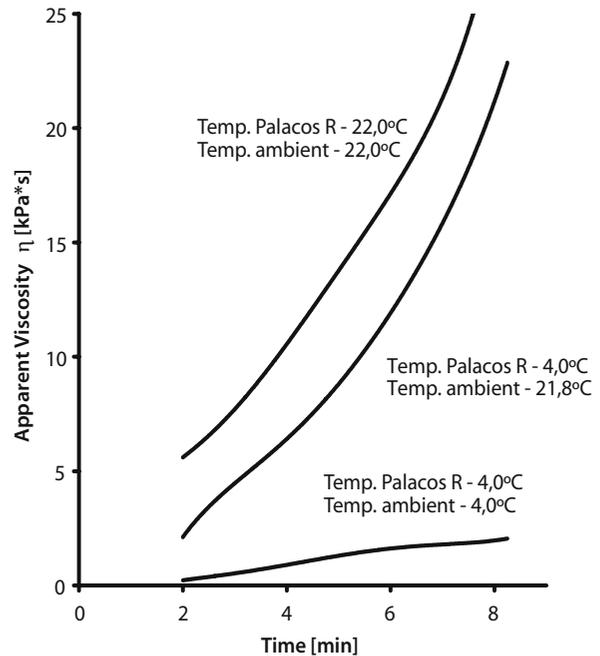
▣ **Fig. 3.33.** Apparent viscosity of Palacos R cement as a function of time after mixing at various cement and ambient temperatures.
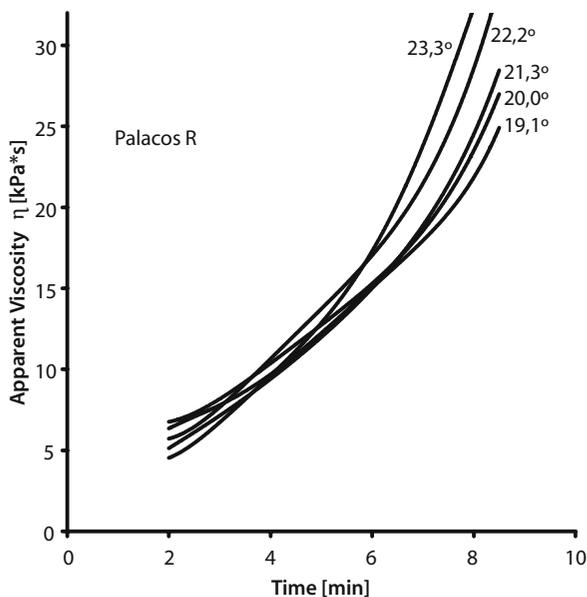
▣ **Fig. 3.32.** Apparent viscosity of Palacos R bone cement as a function of time after mixing at various ambient temperatures. Test shear rate = 0.358 s⁻¹, cone-plate configuration
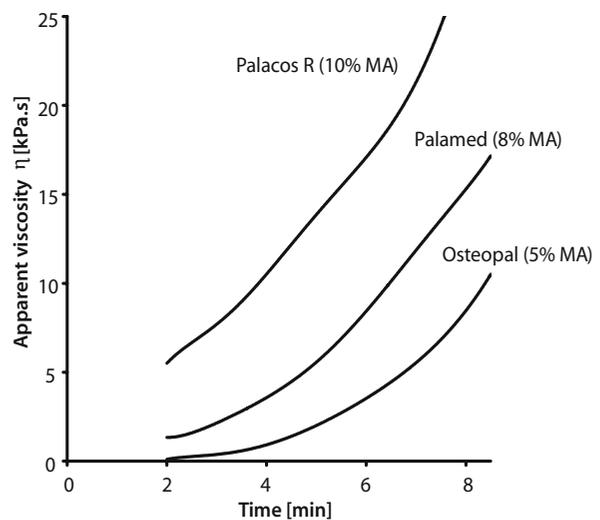
▣ **Fig. 3.34**. Apparent viscosity development for 3 bone cements which differ in their MA copolymer concentration

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Thermal Properties

General

During curing of acrylic bone cement, the liquid component consisting of mainly MMA monomer molecules polymerizes to a solid mass of MMA polymer chains. This process is accomplished by the C=C double bond of each monomer which opens up to form a –C-C-C-C- polymer strain. Opening of the C=C double bond is accompanied by the release of heat. For one gram of pure MMA monomer 588 Joule of heat is generated. In comparison, 420 Joule is sufficient to increase the temperature of 1 gram of water from 0 °C to 100 °C. This heat release results in a temperature rise of the curing cement mass and its environment. The speed of this process is temperature-dependent and the heat release results in a self accelerating polymerization rate and subsequent temperature increase. This is called the Tromsdorff effect and can be demonstrated by measuring the temperature of a curing cement mass. In this chapter the thermal behaviour of various types of cement will be discussed.

Heat Generation

The amount of heat generation is determined by the number of monomers per gram of bone cement. However, not all monomers have the same molecular weight and not all cements contain the same amount of monomer liquid. Therefore, not all cements experience a similar temperature increase. Various bone cements were developed in particular to decrease the cement's temperature increase. Various methods have been used by manufacturers to reduce the amount of heat generation (Table 3.4).

Methods 1 to 3 of Fig. 3.28 all result in less heat generation. Method 4 is distinctively different because the same amount of heat per gram cement is generated, but released at a slower rate. One can measure the temperature rise of a cement mass with a thermocouple. If the cement mass is large enough and isolated, the temperature in the centre will rise to an adiabatic steady value which equals the amount of heat generation divided by the specific heat value of the cement. Assuming that all cements have the same specific heat, the centre peak temperature will resemble accurately the amount of heat generation. Figure 3.35 shows the test setup for such an adiabatic temperature measurement. The measured adiabatic temperature rise for a number of bone cements is listed in Table 3.5.

Traditional bone cements based on MMA monomer and a P/L ratio w/w of 2:1 show a temperature increase of ±104 ºC. It means that starting at an ambient temperature of 22 ºC, a peak temperature of 126 ºC will be measured. The low temperature cements generate less heat and a temperature rise between 71 ºC and 94 ºC is measured. Reduction

Table 3.4. Methods to decrease the amount of heat generation

1. High Powder/Liquid (P/L) ratio (Cemex RX, Sulfix-6)
2. Use of high molecular weight monomers (Boneloc, Sulfix-6)
3. Addition of water to the liquid (Implast)
4. Decrease the polymerization rate (Palacos)

Table 3.5. Adiabatic temperature rise representing the amount of heat generation of low temperature and of traditional »normal temperature« types of bone cement

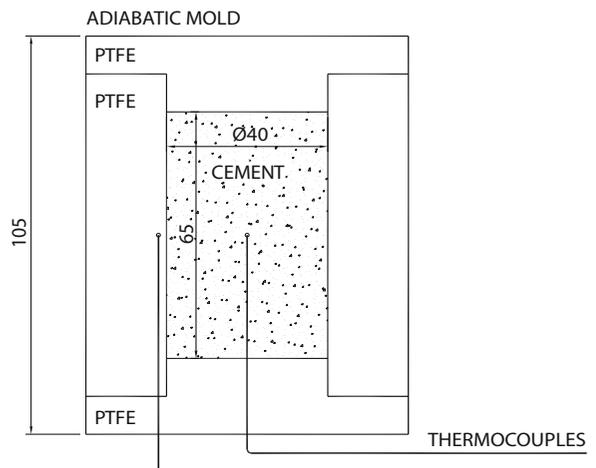| Type of Cement	Temperature Rise [°C]
Boneloc	71
Implast	76
Biolos 1	86
Cemex RX	90
Sulfix-6	90
Biolos 3	94
Zimmer LVC	104
Palacos R	104
Osteopal	104
CMW 3	105
Surgical Simplex Ro	105

Fig. 3.35. Test setup for adiabatic temperature measurement

of heat generation will reduce heat-induced bone-tissue injury. In this way less bone would have to be remodelled after cemented arthroplasty and the surgical reconstruction might have a better survival. This philosophy behind the development of low temperature cements has never been proven clinically. What has been proven is the clinical failure of some low temperature cements. The use of high molecular weight monomers in Boneloc has led to a very low modulus of elasticity and a high creep rate. Implast which had 20% w/w of its monomer replaced by water, was after

polymerization a highly porous cement which results in a low fatigue strength [4]. Reducing the amount of liquid may lead to difficult wetting of the powder. In tensile fracture surfaces of Cemex RX undissolved powder beads can be found (◘ Fig. 3.36). Reduction of heat generation is always at the expense of other cement properties. The best clinical results are still obtained with normal temperature cements.
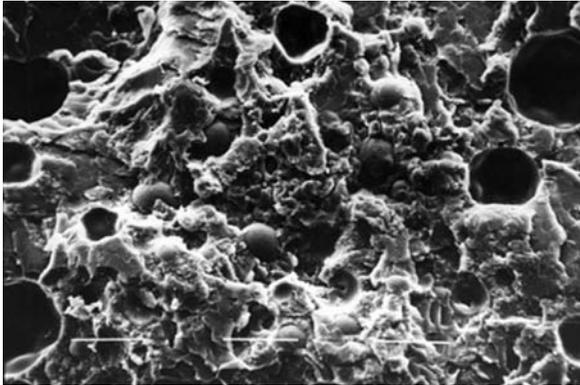

◘ **Fig. 3.36.** SEM of fracture surface of tensile specimen made of Cemex RX cement. P/L ratio of 3 results in inhomogeneous wetting of powder and undissolved powder beads in fracture surface
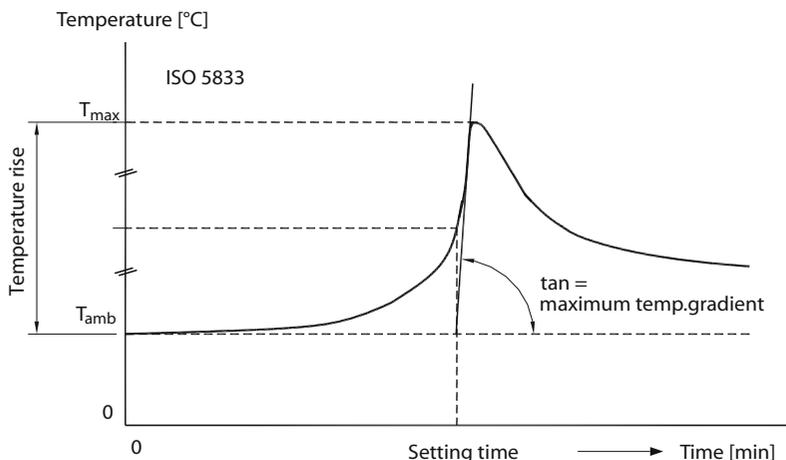
ISO 5833 Temperature Test

In the ISO 5833 standard a temperature measurement test is described for a cement mass of ø 60×6 mm thickness. In the centre of the cement mass the temperature development during polymerization is measured with a thermocouple. ◘ Figure 3.37 shows the test setup and a resulting cement mould. The outcome of this test is very susceptible for accurate placement of the thermocouple. ◘ Figure 3.38 shows a graph of an ISO 5833 temperature test.

The temperature rise will be less than for an adiabatic test because during polymerization heat will flow to the colder environment. ◘ Table 3.6 shows the temperature rise and setting time for a number of bone cements. The setting time is defined as the time as half the temperature rise is attained. This is during the self accelerating polymerization stage just prior to reaching maximum temperature. ISO 5833 temperature rise is lowest for Boneloc with 36 °C and highest for Surgical Simplex RO with 69 °C.

A similar temperature rise in an adiabatic test does not mean that a similar temperature rise is measured in an ISO 5833 test. Palacos R and Surgical Simplex Ro with roughly the same heat generation have a 13 °C difference
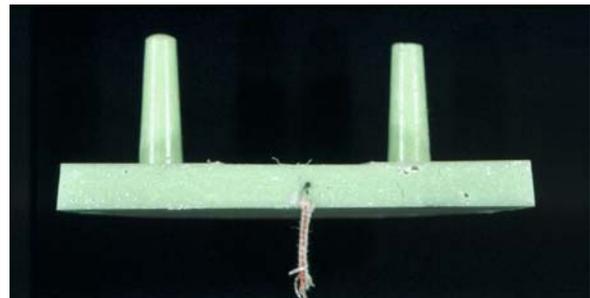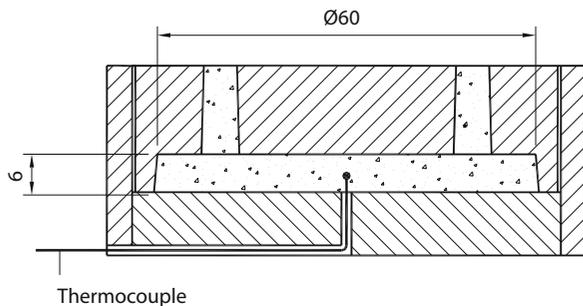
◘ **Fig. 3.37.** Test setup of temperature test according ISO 5833. Important is very accurate placement of thermocouple in the middle of the 6 mm thickness

◘ **Fig. 3.38.** Temperature graph measured in an ISO 5833 temperature test. Setting time is defined as time when temperature is middle between ambient and maximum temperature. Max. gradient is attained just before reaching maximum temperature

in the ISO 5833 temperature rise. A closer look at the ISO-temperature graph reveals that there is a high difference in polymerization rate during the self accelerating polymerization step. Palacos R has a maximum temperature gradient of 60 ºC/min and Surgical Simplex Ro of 179 ºC/min (■ Table 3.7). This means that in case of Palacos the generated heat has more time to flow to the environment and thereby reducing peak temperature.

■ **Table 3.6.** Maximum temperature rise and setting time for various types of cement measured in an ISO 5833 temperature test

Type of Cement	Temperature Rise [°C]	Setting Time [min:sec]
Boneloc	36	11:00
Cemex RX	44	13:20
Sulfix-6	48	10:50
Zimmer LVC	52	11:50
Palacos R Genta	56	10:40
SmartSet Genta	56	9:50
Osteopal	58	12:10
CMW Endurance	63	12:10
CMW 3	65	10:50
CMW 1 Genta	67	9:10
Surgical Simplex Ro	69	11:50

Mechanical Properties

Introduction

In vivo bone cement is subjected to a complex and varying load pattern. It is impossible to simulate experimentally such loading conditions in order to obtain information about the mechanical properties of specific types of bone cements. However regardless the complexity of the loading situation, only 3 modes of mechanical stress do exist: tensile, compression and shear stress. Rather than performing a complex loading simulation test, one can perform a much simpler test to establish the properties for each stress mode separately. Originally, the ISO 5833 standard only described a compression test at 24 hours after mixing. In the 2002 version a bending test at 24 hours after mixing was added.

One may question the validity of a test at 24 hours after mixing. There is a considerable amount of residual monomer after curing of bone cement and afterpolymerization will take several weeks. The effect of afterpolymerization in a compression and in a tensile test can be seen in ■ Tables 3.8 and 3.9. Palacos R was submitted to a compression test at 2 hours, 2 days and 28 days after mixing. The stress at failure, strain at failure and the modulus of elasticity showed an increase of resp. 48%, 15% and 30% over a 4 week period. In tension there is no effect from

■ **Table 3.7.** Different maximum ISO temperature rise despite similar amount of heat generation due to difference in self accelerating polymerization rate

Type of Cement	ISO 5833 Test		
	Adiabatic Temp. Rise	Maximum Temp. Gradient	Maximum ISO Temp. Rise
Palacos R Genta	104 ºC	60 °C/min	56 °C
Surgical Simplex Ro	105 ºC	179 °C/min	69 °C

■ **Table 3.8.** ISO 5833 compression test of Palacos R at various intervals after mixing

Compression Test Palacos R	2 Hours	2 Days	28 Days	Increase
Stress at failure [MPa]	73	86	108	48%
Strain at failure [%]	6.8	7.1	7.8	15%
Modulus of elasticity [MPa]	1920	2170	2500	30%

■ **Table 3.9.** Tensile testing of Palacos R at various intervals after mixing. Specimen shape according to DIN 53455 and crosshead speed of 25 mm/min

Tensile Test Palacos R	2 Hours	2 Days	28 Days	Increase
Stress at failure [MPa]	52	53	52	0%
Strain at failure [%]	2.9	2.7	2.6	−9%
Modulus of elasticity [MPa]	2720	3050	3020	11%

afterpolymerization on the strength and a limited effect on strain and elasticity. Beyond 4 weeks no major change in mechanical properties has to be expected *in vitro* or *in vivo*. Therefore, mechanical testing should better be performed at 4 weeks after mixing than after 24 hours.

The tensile strength of cement is 2–3 times lower than in compression. This is a result of air voids and other inclusions which act as stress risers in this brittle material. This effect is much higher in tensile and outweigh the effect of the afterpolymerization. In this chapter we describe tensile, compressive and shear properties in quasi-static tests.

In vivo cement is subjected to cyclic loading, but because of the time consuming aspect and high costs involved in fatigue testing, only few papers have been published on fatigue data.

Compressive Properties

An overview of several bone cements and their compression properties are seen in ◘ Table 3.10. Cements based on more brittle PMMA polymers seem to have a higher compressive strength and modulus of elasticity than cements like Palacos and Boneloc which are based on more flexible molecules. Cements with a low monomer concentration (P/L ratio of 3) like CEMEX RX exhibit less strength increase due to afterpolymerization than cements with a regular monomer concentration (P/L ratio of 2) like CMW 3 (◘ Table 3.11). In all cases, the compression strength is much higher than physiological compressive stress levels which are in the order of 5 MPa.

Tensile Properties

There is no specific standard for a quasi-static tensile test of bone cement. General standards for tensile testing of plastics, however, do exist, like DIN53455 and ASTM D638. ◘ Table 3.12 shows the results of tensile testing of specimens made according to DIN53455, stored in air at ambient temperature, and tested at 28 days after mixing with a crosshead speed of 25 mm/min. All specimens with voids of ø 1.5 mm or more at the fracture site are discarded. The failure stress varies between 29 and 52 MPa. There is a clear effect from the chemical formulation on the mechanical properties (◘ Fig. 3.39). The cements based on copolymers with ductile MA-molecules (SmartSet, Palacos, Palamed) show the highest tensile strength, while the more brittle cements based on copolymers with styrene molecules or PMMA homopolymers show the lowest tensile strength. Note that Boneloc has a normal tensile strength but the lowest modulus of elasticity, which resembles its high creep rate. The more ductile MA-copolymers have a strain to failure of ± 2.1% which doubles

Elasticity	Molecule	T_{glass} [°C]
Brittle	Styrene	120
	Methylmethacrylate	105
	Ethylmethacrylate	65
	Butylmethacrylate	20
Ductile	Methylacrylate	6

◘ **Fig. 3.39.** Mechanical properties of various types of acrylic molecules

◘ **Table 3.10.** Compressive properties of bone cements derived from a compression test according to ISO 5833

Type of Cement	Compression Tests at 24 Hours after Hand Mixing, According to ISO5833		
	Failure Stress [MPa]	Failure Strain [%]	Modulus of Elasticity [MPa]
CEMEX RX	101.8	7.1	2608
CMW 3	101.7	7.1	2518
CMW 1 Genta	96.5	7.0	2147
SULFIX-6	96.3	6.9	2461
CMW 3 Genta	95.9	6.9	2177
Palacos R Genta	80.7	6.2	1993
Boneloc	80.0	6.5	2177

◘ **Table 3.11.** Increase of compressive failure stress due to afterpolymerization of bone cements with various P/L ratios. Higher liquid portion gives more strength increase

Type of Cement	Compressive Failure Stress at 24 Hours [MPa]	Compressive Failure Stress at 28 Days [MPa]	Increase of Failure Stress [%]	P/L Ratio w/w
CMW 3	101.7	122.3	20.3	2.1
SULFIX-6	96.3	110.4	14.6	2.3
CEMEX RX	101.8	113.0	11.2	3.0

3

Table 3.12. Tensile properties of bone cements derived from a tensile test at 28 days after mixing. Specimens made according to ISO 53455, specimens stored in air at room temperature, crosshead speed 25 mm/min

Type of Cement	Tensile Tests at 28 Days after Hand Mixing		
	Failure Stress [MPa]	Failure Strain [%]	Modulus of Elasticity [MPa]
SmartSet HV	51.7	2.2	3068
Palacos R	49.8	2.1	3176
Palamed Genta	48.2	2.0	3283
Palacos R Genta	46.4	1.8	3300
SULFIX-6	40.7	1.6	2949
CMW 1 Genta	35.5	1.3	3031
Boneloc	35.4	0.9	2360
Surgical Simplex Ro	33.9	1.2	3017
CEMEX RX	30.7	1.1	3098
CMW 3 Genta	28.9	1.0	3050

the failure strain of ± 1.1% of the weaker and more brittle Styrene-copolymers and MMA homopolymers.

Shear Properties

Unfortunately, shear strength cannot be measured accurately experimentally. To design an experiment which causes an even shear stress distribution on the fracture site is very difficult, if not impossible. So called push out tests which are often used for shear testing will not generate an even shear stress. The best experimental setup to determine the shear strength is possible an AIA model [9]. However, even after extensive testing with this model we did not succeed to develop a well functioning shear test for bone cement.

Take Home Messages

- The ISO5833 standard is not suitable for cement quality assessment.
- The ASTM F2118 fatigue-test standard lacks a minimum strength requirement.
- Low viscosity cements have no surgical or patient benefits.
- Low temperature cements have no surgical or patient benefits.
- Medium viscosity cements can be handled in a syringe without cooling.
- Flexible MMA-MA copolymer cements perform best in tensile.
- Stiff PMMA and Styrene cements perform best in compression.
- The most important pre-clinical cement test is a fatigue test.
- The most important clinical cement property is handling.
- Traditional cements perform best.

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