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Dynamic mechanical behavior of PMMA based bone cements in wet environment

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The mechanical properties of three wet commercial bone cements, namely Braxel (from Bioland®), Simplex-P (from Howmedica®) and CMW1-G (from DePuy®) are investigated by means of stress relaxation and dynamic mechanical analysis (DMA). The geometry of loading that was used is the three point bending method (ASTM D790); all the tests were performed in a water chamber by means of temperature sweeps between 17 and 57 °C and spanning four frequency decades. The results show that viscoelastic properties are strongly dependent on specimen conditioning (i.e. water uptake and heat treatment).

The results also show that all the cements that were analyzed show mechanical properties which are intermediate between the ones of the cancellous bone and of the metals of which prostheses are normally made. As a consequence, the cement is able to reduce the stress concentrations due to the interfacing of materials which have very different stiffnesses. Moreover, the results of the DMA, particularly the ones concerning the damping factor (tan δ), indicate that at body temperature the bone cements tested show an increased capacity of dissipation, the higher is the loading frequency, thus displaying shock absorbing properties.

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Introduction

Total joint arthroplasty restores the biomechanics of contiguous bones, affected by severe arthritis, by substituting the damaged bone extremities with biomaterials (e.g. total hip or knee replacement) [1–5]. The prosthetic joint device is held in place by means of cemented, cementless or hybrid fixation of its components to the bones [6]. Cemented prosthesis are especially indicated when the conditions of the hosting tissue don't meet general requirements of conformity and of a healthy structure [7–9]. Moreover, despite the promising performances of new cementless prosthesis, ten year follow-up studies have shown that the successful rates of cemented and early cementless prostheses are very similar, the major cause of failure being aseptic loosening related to wear debris originating from the polyethylene cup [10–14]. Partially cemented prosthetic components is another solution which combines the features of cemented and cementless fixation [15]. In any case, the cost-effectiveness demand, patient's needs, activities and age are other significant issues concerning the choice of the kind of prosthesis [16, 17].

Cements based on self-polymerizing poly(methylmethacrilate) (PMMA) are currently the only commercial synthetic biomaterials for anchoring the prosthetic components to contiguous bones using the cemented fixation technique. They consist of a solid powder phase made of PMMA or related copolymer and a liquid monomer phase (MMA) [18]. After mixing the solid and liquid phases, the polymerization takes place via a free radical reaction and the kinetics is regulated by the concentrations of the initiator (e.g. benzoyl peroxide) and the activator (e.g. N,N,-dimethyl-p-toluidine) which are present in the solid and liquid phases respectively [19–22]. Radio-opacifier (e.g. barium sulfate or zirconium dioxide) is added to the solid phase in order to differentiate bone from cement with radiographic analyses [23].

Even though bone cements are widely used, a number of drawbacks are well recognized: thermal necrosis due to the exothermal curing [21, 24], chemical necroses related to the release of unreacted monomer [25], concentrated stresses originating from the mismatch between the mechanical properties of the bone, the cement and the prosthesis, residual stresses related to shrinkage during polymerization [26–28], and periprosthetic inflammatory response due to wear debris of cement [29, 30]. The effect is the failure of weakly linked regions (i.e. zones at the interfaces between bone, cement and prostheses). Thus, in more than four decades several formulations and setting techniques of PMMA-based bone cements have been proposed in order to improve the quality of these materials [31].

Antibiotic (e.g. gentamicin) loaded cements are suggested in order to improve the resistance to septic loosening, and are particularly indicated for revisions of infected joints [32, 33]. However, the antibiotic effect is related to the release of the drug and often it is not sufficient to reduce the osseous infection since the cement matrix is impermeable to gentamicin and thus transport properties are mainly related to interconnecting cracks and voids in the cement bulk [34]. Mechanically the disvantages of the presence of antibiotics are a slight lowering of elastic modulus and strength in wet environment [18].

Since the physical properties of the cured PMMA are strongly dependent on the preparation technique, improved cement mixing methods are developed (e.g. vacuum mixing [20] or centrifugation [35]) in order to reduce the content of pores [23] which act as stress raisers [36, 37].

The use of poly(buthylmethacrilate) (PBMA) or poly(ethylmethacrilate) (PEMA) is suggested in order to lower the temperature during the exothermal reaction and to reduce the unreacted matter, thus reducing thermal and chemical necrosis of bone. Mechanically the disvantages are lowered elastic modulus and strength, both in static and dynamic conditions, while the advantages are higher strain to failure and improved damping factor if compared to PMMA bone cements [38, 39].

The introduction of a bioactive phase (e.g. hydroxyapatite) in the PMMA matrix is suggested in order to enhance the quality of the bone–cement interface [40], while the use of a self-reinforced PMMA composite is proposed in order to improve the mechanical bulk properties of the cement [41].

The cement layer has the main task of resisting and transferring the loads between the natural and synthetic coupled materials, while functioning as a mechanical buffer, reducing stress concentrations and absorbing mechanical shocks [42–44]. Since the adhesion between the bone and the cement and between the prostheses and the cement is mechanical, a number of tests (e.g. tensile, compression and bending) have been developed to characterize bone cements in a variety of environments and temperature ranges (e.g. air, water, Ringer's and intralipid solution), by using static and dynamic analyses [45].

Half a century of experience has brought to a multitude of mechanical data available in the literature, especially regarding static tests. During the polymerization, the PMMA beads powder swells and partially dissolves in the liquid phase of MMA [46], thus bone cements have a microstructure which is characterized by at least two-phases, with static mechanical properties that are lower than the industrial PMMA either in tension or bending [47] or compression [48]. Static testing, in the same environment and specimen conditioning, show that bending properties (e.g. Young modulus and strength) of bone cements are higher than tensile properties, while the strength in compression is higher than the other loading conditions [31]. PMMA beads size and distribution slightly affect static properties both in tension and compression, while affecting the temperature increase during polymerization [46]. The increase of the amount of barium sulfate in PMMA improves the static compression properties [48] and reduces the tensile properties [49], while zirconium dioxide leads to enhanced tensile properties of the cement if compared to barium sulfate [23]. At room temperature the bone cement in compression is harder and more brittle than at body temperature [50]. Water absorption reduces the bending modulus and the glass transition temperature [18], and the same conclusion is drawn when comparing tensile properties at room temperature [51] and those measured in a wet environment at body temperature [31].

Fatigue properties of bone cements suggest strong correlations between *in vitro* testing results and clinical outcomes [51]; it is interesting to observe that although static properties of bone cements are lower than industrial PMMA, no difference is shown in the fatigue behavior in the same testing conditions [41].

Creep and stress-relaxation tests are suitable to assess the viscoelastic properties and the long term stability of materials [52]: PBMMA bone cements show larger subsidence rates if compared to PMMA when tested in unconstrained manner by means of stress-relaxation tests [53]. Viscoelastic tests are more sensitive to bone cement conditioning (e.g. aging and water absorption) and water based environments give the largest creep rates on PEMA based bone cements [38]. The factors which affect viscoelastic properties of acrylic bone cements, whether by means of stress-relaxation or creep, in tension or bending, always fall into the same categories: age of the cement, environment, temperature and stress-level [54].

Fracture mechanics testing is used to evaluate the fracture toughness of materials [55]: zirconium dioxide leads to enhanced toughness properties of the cement if compared to barium sulfate [23], while higher concentrations of N,N,-dimethyl-p-toluidine yield to lower values of the fracture toughness [56].

It is clear that the mechanical properties of PMMA-based bone cements are strongly affected by the composition, the mixing technique, testing methods and specimen conditioning, thus it is difficult to compare data affected by so many variables.

The mechanical behavior of PMMA is strongly time-dependent; in fact, fatigue and crack propagation of PMMA are sensitive to changes in strain rate associated to cyclic frequency [57–59], moreover the frequency sensitivity of classic fatigue is not consistent with the frequency sensitivity of fatigue crack propagation [57, 59, 60], this inconsistency is partially due to the different temperature profiles which develop into the specimens for the different testing techniques [57, 61, 62]. In this context it is appropriate to characterize the behavior of bone cements under cyclic loading condition. DMA is a very powerful and reliable methodology to determine various viscoelastic properties of polymers, such as the storage modulus, the loss modulus, the damping factor, the glass transition temperature and the presence of unreacted monomer [63]. DMA data can be obtained through temperature or frequency sweeps. Moreover this technique, investigating the behavior of the system at various frequencies, can be used to simulate the mechanical response of the material undergoing any periodical stress history through Fourier analysis coupled to the Boltzmann superposition

TABLE I Composition of the powder and liquid phases of the three bone cements

	Powder (g)		Liquid	
CMW 1 Gentamicin	Polymethylmethacrylate	33.89	Methylmethacrylate	18.22 g
	Gentamicin sulfate	1.69	N,N-dimethyl-p-toluidine	0.15 g
	Barium sulfate	3.60	Hydroquinone	25 ppm
	Benzoyl peroxide	0.82		
Simplex P	Polymethylmethacrylate	29.4	Methylmethacrylate	18.31 g
	Styrene		N,N-dimethyl-p-toluidine	0.48 g
	Polymethylmethacrylate	6.0	Hydroquinone	80 ppm
	Barium sulfate	4.0		
	Benzoyl peroxide	0.6		
Braxel	Polymethylmethacrylate	34.92	Methylmethacrylate	13.84 g
	Barium sulfate	4.0	Butylmethacrylate	2.16 g
	Benzoyl peroxide	1.08	N,N-dimethyl-p-toluidine	0.39 g
			Hydroquinone	20 ppm

principle [64]. This methodology may also provide useful information about the impact response of the material, which in fact is related to the results of the dynamic mechanical tests at high frequencies.

DMA of PMMA, in the amorphous or isotactic state, has been widely investigated in dry condition [65–70], however, even if a wet environment strongly affect viscoelastic properties of bone cements [38, 39, 54, 58], DMA in wet condition has never been undertaken. The effects of environmental factors have important consequences on both theoretical *in vitro* studies and clinical *in vivo* applications. The aim of this work is to characterize experimentally three different bone cement formulations through DMA in a water chamber by means of temperature sweeps between 17 °C and 57 °C and spanning four frequency decades. The testing is completed also by stress-relaxation tests.

Materials and methods

Braxel (from Bioland®), Simplex-P (from Howmedica®) and CMW1-G (from DePuy®) bone cements were tested. The first cement is characterized by the addition of a 13.2% by weight of butylmethacrylate to the liquid monomer, CMW1-G cement contains gentamicin sulfate antibiotic, while Simplex-P is a copolymer of MMA with styrene. Table I lists the composition of the powder and liquid phases of these bone cements. The specimens were prepared according to the manufacturer's instructions.

Each bone cement dough was poured into molds of rectangular cross section (125 × 50 × 2.5 mm) and a pressure of 0.02 MPa was maintained during PMMA curing. PMMA sheets were kept at room temperature for 24 h or one week to allow further polymerization and then machined into $25 \times 50 \times 2.5\,\text{mm}^3$ specimens with a low speed diamond wheel saw (Isomet, Buehler Ltd, IL, USA). In order to investigate the dependence of material properties on water uptake and monomer release, each cement sample was randomly divided in five groups each of five specimens and conditioned as described in Table II.

A precision balance (Mettler Toledo AB 54-S, Switzerland) was used to monitor, over three weeks, water content of each cement (specimens of 3 g with a thickness of 2.5 mm) kept in distilled water at 37 °C in a temperature controlled device (Thermoblok TA120, Lurano, Italy). The equation used is:

$$wc = \frac{w_t - w_0}{w_0} \tag{1}$$

where w_t and w_0 are the specimen weight at time t and dry, respectively. At each time point measurement, the specimens were removed from the water bath, bottled dry with filter papers, weighed and replaced in the water bath.

Mechanical tests were carried out using the three point bending method (ASTM D790) in a water chamber controlled by a Techne TE10D (Cambridge, UK) heater mixer constrained to a servomagnetic Enduratec ELF3200 (Minnetonka, USA) with a 220N loading cell. Stress and strain in the outer fibers (σ and ε respectively) were computed according to the following equations:

$$\sigma = \frac{3PL}{2bd^2} \tag{2}$$

$$\varepsilon = \frac{6dD}{L^2} \tag{3}$$

where P is the applied load, D is the middle span deflection, L is the span, b and d are the specimen width and depth, respectively.

Stress-relaxation tests were performed by applying strain steps between 0.2% and 1% with an initial ramp of $1.2\,\text{min}^{-1}$. The stress decay was monitored in each relaxation test over a period t_D reported in Table II. Stress vs. time, reduced stress function (i.e. the ratio between the stress at a given time t and the stress at the initial instant t_0) vs. time and elastic modulus vs. time plots were used to compare the behavior of the three bone cements.

DMA, in a displacement control mode, was performed after each stress relaxation step, thus the mean strain level and the dwelling time coincide with the values of strain and t_D respectively, as reported in Table II. Sinusoidal waveforms of 0.04% strain amplitude (ε_0) between 0.01 and 100 Hz

$$\varepsilon = \varepsilon_0 \sin(\omega\, t) \tag{4}$$

were used for the frequency sweep characterization at 37 °C. Particularly sample 4 of each bone cement (Table II) underwent a frequency sweep combined with a temperature sweep at 5 °C increments between 17 °C and

TABLE II Samples conditioning and testing protocol for each bone cement. F denotes frequency sweep between 0.01 and 100 Hz and T_{17-57} denotes temperature sweep between 17 and 57 °C

Sample	Conditioning			Mechanical testing		
	Air		Water at 37 °C	Stress-relaxation in H_2O		
	Temp. (°C)	Time (h)	Time (h)	t_D (h)	Strain ‰	DMA
1	20	168	168	1	3,6,9	—
2	20	24	2	3, 5, 15, 24, 48	3	F, T_{37}
3	90	2	2			
4	20	168	168	1	6	F, T_{17-57}
5	20	168	168	3, 24, 48, 96, 144	3, 6, 9	F, T_{37}

57 °C. The time lag between the excitation signal (the strain) and the stress response

$$\sigma = \sigma_0 \sin(\omega t + \phi) \tag{5}$$

is reflected in the phase angle ϕ. This angle is independent of the specimen geometry and equals to the phase angle between the controled or input signal (the displacement) and the monitored or output signal (the load).

Fig. 1 shows the mechanical setup system where dynamic data were interpreted by means of the in phase or storage modulus (E'), the 90° out phase or loss modulus (E'') and the damping factor ($\tan\delta = E''/E' = \tan$).

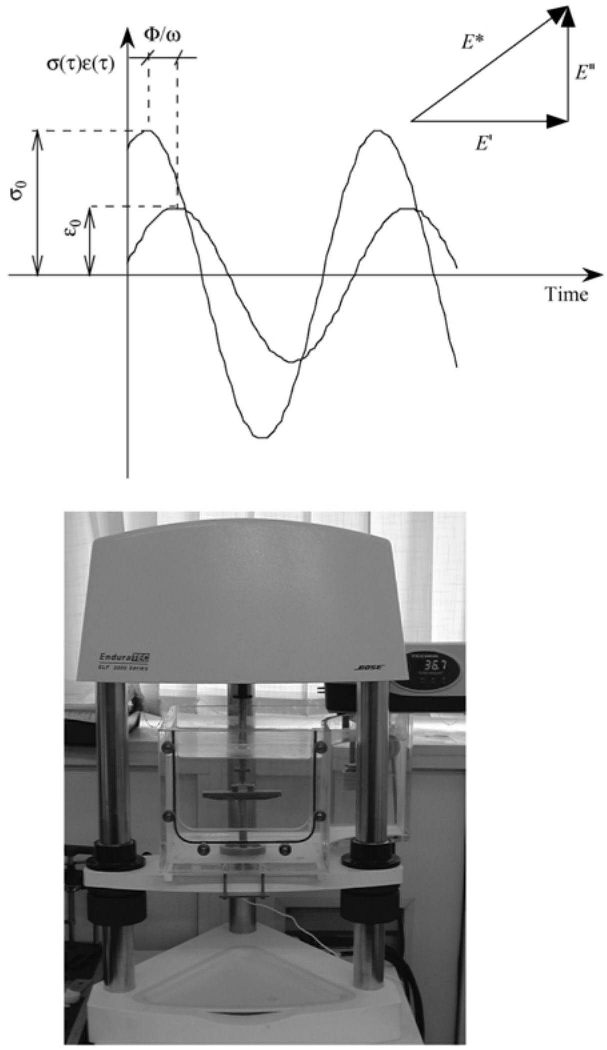

Figure 1 Mechanical setup system. The phase angle is independent of the specimen geometry and equals to the phase angle between the controled or input signal (the displacement) and the monitored or output signal (the load).

The statistical significance of the results was assessed using the one-way ANOVA variance test [71].

Results

Water uptake over three weeks for each bone cement is shown in Fig. 2. In the first stage the cements behave in the same manner, uptaking about 1% by weight of water in the first 20 h. The second stage is characterized by a lower water absorption rate and the differences between the cements are related to the different composition, particularly the lower absorption rate of the Simplex-P cement may be due to the hydrophobic nature of the styrene group. The weight gain for the Braxel cement reaches a steady-state value after about two weeks.

The Young moduli computed at 1.2 min^{-1} at 37 °C for the Braxel, CMW1-G and Simplex-P (samples 1 and 5) are presented in Table III. These values are related to the ramp of each stress-relaxation test up to 0.3% of strain. These moduli are consistent with those found by Khun and Edge who used the four point bending method in the same testing conditions [18, 72]. An elastic modulus of 2.6 GPa for Simplex-P was found by Gilbert *et al.* (as reported in the bone cements review by Lewis [30]), who also used the three point bending method. Bargar *et al.* [73], on the other hand, found a slightly higher value of the elastic modulus (2.9 Gpa) for the same cement, prepared according to manufacturer's instructions, aged at 37 °C in saline solution for 7 days and using the three point bending method.

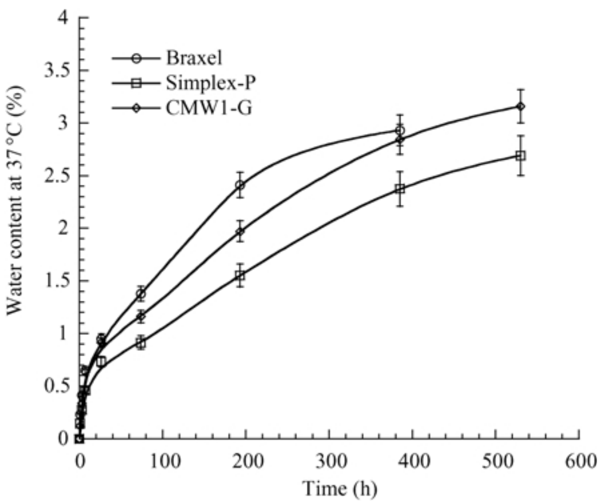

Figure 2 Water uptake over three weeks for each bone cement. The absorption rate of the Simplex-P cement may be due to the hydrophilic nature of the styrene group.

TABLE III Young's moduli computed at 1.2 min^{-1} and the maximum stress reached at each step level for the three bone cements. These values are related to the ramp of each stress relaxation test up to 0.2% strain. Numbers inside the brackets are the standard deviations

	Young's modulus (GPa)	Stress at 0.1 s (MPa)		
		0.3% Strain step	0.6% Strain step	0.9% Strain step
Braxel	2.1 (0.3)	5.3 (0.4)	8.2 (0.7)	11.0 (0.9)
CMW1-G	2.3 (0.2)	5.9 (0.5)	8.5 (0.6)	11.2 (0.9)
Simplex P	2.6 (0.3)	6.9 (0.5)	9.1 (0.7)	11.7 (0.8)

Fig. 3 shows the stress-relaxation over 1 h for the three bone cement (sample 1 of Table II) at 0.3%, 0.6% and 0.9% of strain. These diagrams show that there is a proportionality between the superimposed strain step and the resulting stress relaxation curve, thus suggesting linear viscoelastic behavior. Table III shows the initial stress values for each bone cement sample at strains of 0.3%, 0.6% and 0.9%. The same reduced stress function is observed for the CMW1-G and Simplex-P cements, while the Braxel cement differs from the other two just in the starting relaxation stage, showing an initial faster decrease, after which, the same decrease rate is observed. Yetkinler *et al*. [39], using compression tests on Simplex-P at 37 °C in aqueous environment, measured 19.79% relaxation at 1% of strain after one hour of stress-relaxation experiments. This value is consistent with the relaxation percentage between 1min and 1h of the Simplex-P cement measured in our experiments (18.1%) up to 0.9% of strain.

The DMA results at 37 °C, performed at various mean strain levels (up to 0.9%) and time intervals (up to 144 h), suggest that for each bone cement (sample 5 of Table II) the storage modulus, the loss modulus and the damping factor are independent of the mean strain level and of the dwelling time t_D. Fig. 4 shows the bar plot of the damping factor vs. frequency for the tested cements and the error bars represent the standard deviation. The damping factor for all the cements increase with frequency, however in the same frequency sweep range the Braxel damping factor increase is slightly higher. The differences between the damping factor mean values up to 10 Hz are not significant, however, in the 10–100 Hz decade the Braxel cement has a slightly higher damping factor. Table IV reports the p and F values (the upper and lower numbers respectively in each cell) of the ANOVA analyzes for the damping factor distributions for the three bone cements.

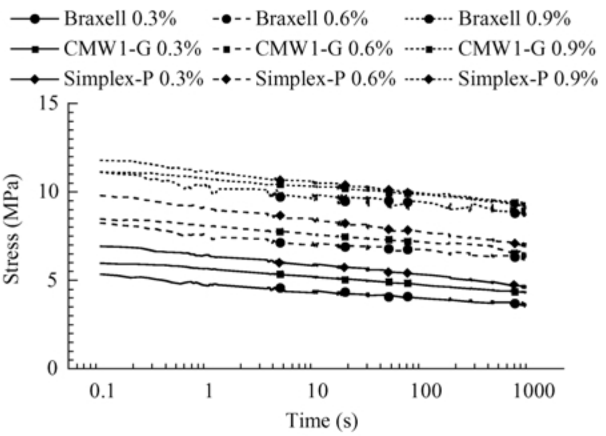

Figure 3 The stress relaxation curves of the three bone cements (sample 1 of Table II) at 0.3%, 0.6% and 0.9% initial strain. Up to this level of strain the materials exhibit linear scaling.

The dependence of the damping factor on the temperature and frequency for the Simplex-P cement, hydrated for 1 week (sample 4), is shown in Fig. 5. The secondary β-transition in the glassy region is clearly observed as a distinct relative maximum at temperature and frequency below 30 °C and 5 Hz, respectively. In spite of the merging with the α-transition peak at frequency higher than 5 Hz, this maximum clearly shifts at higher temperature as the frequency increases. Moreover, the cross points of the frequency parametrized curves occurring at temperatures higher than 40 °C suggest that the glass transition temperature will decrease the lower is the frequency. Thus in a wet environment both the α and β peak shift at higher temperature with the increase of frequency. The same trend is observed for the CMW-1 and Braxel cements, particularly the position of the β-peak for the Braxel cement at frequency up to 1 Hz is lower than 20 °C. It is interesting to note that at body temperature all the tested cements are characterized by an increase of tanδ with frequency, at least in the 0.01–100 Hz range (Figs. 4 and 5).

The effect of specimen conditioning (length of time since mixing, temperature, hydration and dwelling time) for the Symplex-P cement is shown in Figs. 6 and 7. Particularly the reduced stress function (Fig. 6) shows that sample 2 relax much faster than heated specimens (sample 3) and than specimens which underwent prolonged hydration (sample 5). This behavior is consistent with the damping factor measurements (Fig. 7) at low frequency, both heating and hydration have significant effect on the dissipation capability of the cement, particularly at 0.01 Hz the damping factor of sample 3 is higher than the other samples and this difference is statistically different ($p < 0.001$). However, while at 37 °C the stress-relaxation test is suitable to depict the effect of the conditioning on the mobility of the polymer main chain, at the same temperature the DMA test better describe the effect of the conditioning on the mobility of the polymer side chain, particularly at high frequency the damping factor (Fig. 7) of cured specimens (sample 3) increase with the dwelling time, hence with hydration, tending to the tan δ values of sample 2.

Modeling

From the point of view of classical linear viscoelasticity, the stress relaxation experiments and the DMA data are apparently not in agreement with each other. In fact, the experimental stress relaxation curves decay slowly without showing a well defined saturation point even

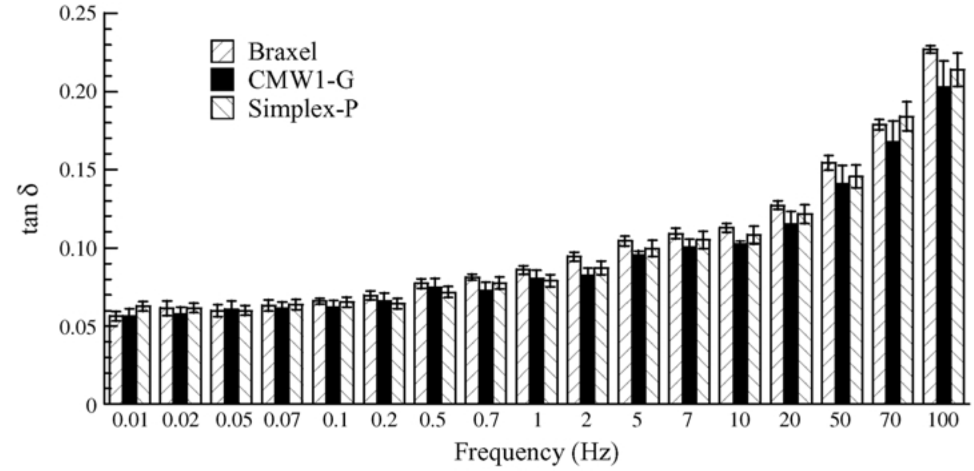

Figure 4 The damping factor vs. frequency for the three bone cements (sample 5 of Table II). Despite the different composition of the cements, the graphs are substantially equal. In the explored strain and frequency ranges, DMA suggests that the viscoelastic properties are independent of the mean strain level and of the dwelling time t_D, thus also independent of the stress level.

after 10 000 s, denoting a reasonably high relaxation time. On the other hand, the DMA results, particularly the ones concerning $\tan\delta$ show the presence of a maximum at frequencies higher than 100–1000 Hz, clearly indicating the presence of a rather low relaxation time, around 10^{-3}–10^{-4} s. Thus it is practically impossible to formulate a model based on only one of the two experiments and use it to predict the results of the other one, at least within the context of classical linear viscoelasticity; the two experiments will then be considered as complementary to each other and an exhaustive model for the material behavior will be obtained using information from both of them.

As the deformations involved during testing are rather small, we shall assume a linear viscoelastic behavior, with the stress depending on the history of the linear strain through an equation of the following kind [74]:

$$\sigma(t) = \int_0^\infty E(t-s)\frac{d\varepsilon(s)}{ds}ds \tag{6}$$

where σ and ε denote as usual the stress and the deformation respectively, while E is the viscoelastic modulus. The form of the viscoelastic modulus as a function of time needs to be prescribed with a constitutive equation. In our case we will assume that

$$E(t) = E_\infty + \Delta E(t) = E_\infty + C_1\exp(-t/\tau_1) + \int_0^\infty K(\tau)\exp(-t/\tau)d\tau \tag{7}$$

with

$$K(\tau) = \frac{C_2}{\tau} + \frac{C_3}{\tau^n} \tag{8}$$

The storage and loss moduli $E'(\omega)$ and $E''(\omega)$ are readily obtained from the expression for $E(t)$ using the known definitions [74]

$$E'(\omega) = E_\infty + \omega\int_0^\infty \Delta E(s)\ \sin(\omega\ s)\ ds \tag{9}$$

and

$$E''(\omega) = \omega\int_0^\infty \Delta E(s)\cos(\omega\ s)\ ds \tag{10}$$

where ω is related to the frequency f through

$$\omega = 2\pi f \tag{11}$$

while the damping factor $\tan\delta$ is the ratio between $E''(\omega)$ and $E'(\omega)$ as usual.

As we can see from Equation 7, this model has one main relaxation time (τ_1) superposed on a continuous spectrum of relaxation times (the last term in Equation 6). The exponential part of Equation 7 will enable a comfortable fitting of the material behavior with respect to the stress relaxation while the continuous spectrum

TABLE IV ANOVA anlysis of the damping factor distributions for the three bone cements. The upper and lower values in each cell represent the p and F values, respectively

tan δ	0.01 Hz		0.1 Hz		1 Hz		10 Hz		100 Hz	
	CMW1-G	Simplex-P	CMW1-G	Simplex-P	CMW1-G	Simplex-P	CMW1-G	Simplex-P	CMW1-G	Simplex-P
Braxel										
p	0.32	0.09	0.38	0.75	0.75	0.37	0.05	0.17	0.004	0.03
F	1.1	3.6	0.8	0.1	0.1	0.9	4.9	2.2	13.6	6.2
CMW1-G										
p		0.28		0.30		0.26		0.01		0.02
F		1.3		1.2		1.4		62		7.6

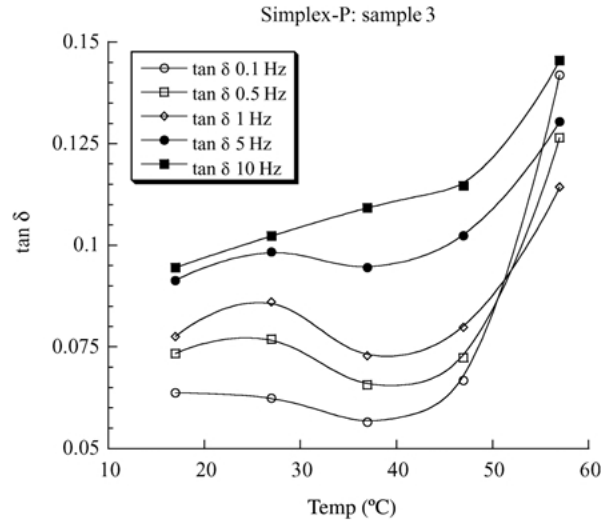

Figure 5 Dependence of the damping factor on the temperature and frequency for the Simplex-P cement, hydrated for 1 week (sample 4 of Table II). The secondary β-transition in the glassy region is clearly observed with a distinct relative maximum at temperature and frequency below 30 °C and 5 Hz, respectively. This maximum shifts at higher temperature as the frequency increases. The cross points of the curves suggest that also the a-peak will be first reached at lower frequency.

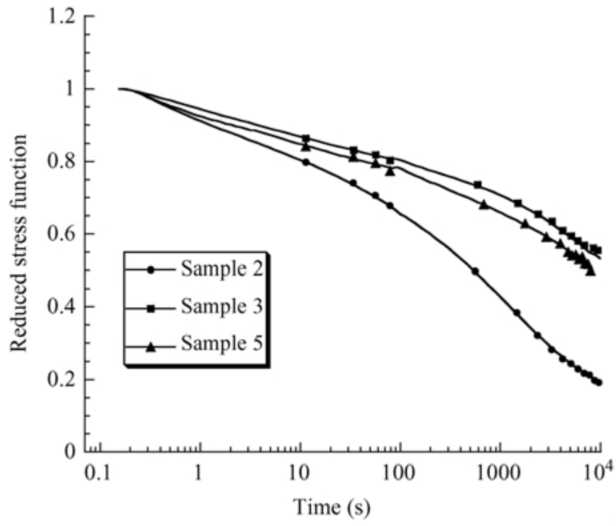

Figure 6 Effect of specimen conditioning for the Simplex-P cement on the stress relaxation at 37 °C in water environment. The reduced stress function shows that sample 2 relax much faster than heated specimens (sample 3) and than specimens which underwent prolonged hydration (sample 5). This effect is related to monomer content.

will be used to fit the data of the dynamic mechanical analysis. Models with continuous spectra of relaxation times have been successfully used to model the linear viscoelastic behavior of many biological materials [75].

The plots of the theoretical viscoelastic modulus E as a function of time, and of the storage modulus E' and the loss modulus E'' as a function of the frequency are compared in Figs. 8 and 9 with the correspondent experimental data for the three bone cements. As we can see, the theoretical curves are in a sufficiently close agreement with the experimental DMA data for all the bone cements. The experimental relaxation moduli, on the other hand, are lower than the theoretical prediction, with the exception of Braxel. The material parameters used in the curve fitting are reported in Table V for the three bone cements.

Discussion

In order to reduce stress concentrations between prosthesis and bone, an effective cement should have a modulus that is intermediate between the modulus of the prosthesis (usually made of metal) and the modulus of spongy bone. The results of the stress relaxation experiments show that all the cements we tested have this requisite. Since PMMA-based bone cements are applied between the bone tissues and metal implants, it is important to compare the mechanical properties of bone cements with those of bone (spongy and cortical) and metal (titanium and cobalt based alloys). Table VI shows that PMMA based bone cements have an elastic modulus and static strength which are in between those of spongy bone [76, 77] and titanium and cobalt based alloys [77–79], however the fracture toughness of bone cements is lower than cortical bone [80, 81] and titanium alloys [82, 83]. The hardness of PMMA [84] is also lower than both spongy and compact bone [85] and lower than titanium alloys. The same conclusion is drawn when comparing bone cements fatigue strength with those of cortical bone and titanium alloys. The average values of the damping factor and storage modulus of bone cements at 1 Hz are in between the values of metals [86] and trabecular bone. The data on cancellous bone have been obtained from unpublished tests performed on specimens taken from the tibial plateau in the coronal direction, tested in the same conditions as the ones used for bone cements in the present article. The damping factor and the storage modulus of bone cements are in fact of the same order of magnitude as those of compact bone [87].

Water transport in amorphous glassy PMMA occurs

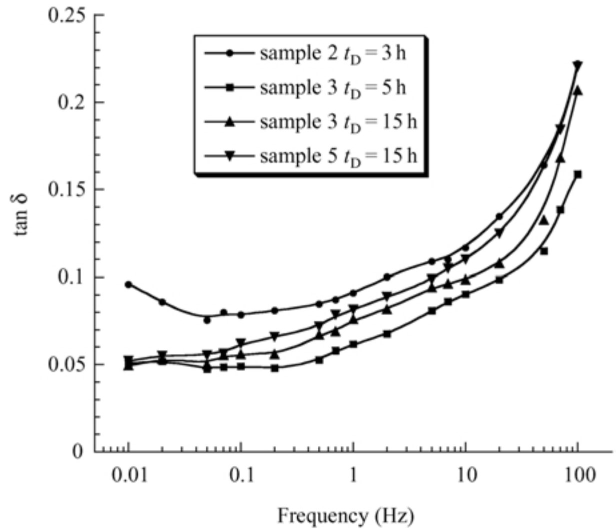

Figure 7 Effect of specimen conditioning for the Symplex-P cement on the dynamic mechanical behavior in water environment. At 37 °C DMA tests describe better than stress relaxation tests (Fig. 6) the effect of the conditioning on the mobility of the polymer side chain. At high frequency the dissipation of the material is similar for sample 2 and for specimens which underwent prolonged hydration, thus monomer and water have a same effect on the side chain mobility.

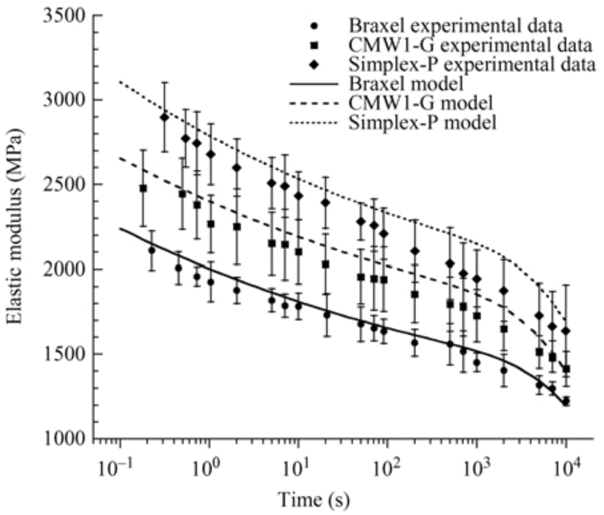

Figure 8 The elastic modulus vs. time for the three bone cements at 37 °C. The model that was used to fit the experimental data (Equation 7) is defined using information from both the stress relaxation and DMA tests.

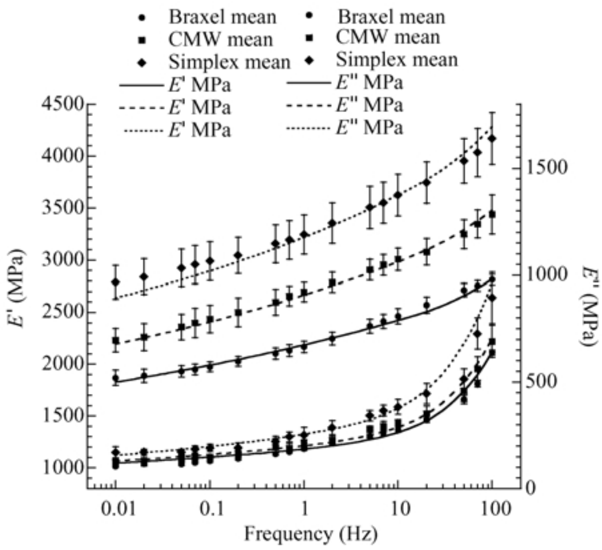

Figure 9 The storage and loss modulus vs. frequency for the three bone cements at 37 °C. While the storage modulus increase with frequency is similar for the three cements, the loss modulus increase differs: between 0.01 and 100 Hz the loss modulus increase for the Braxel, CMW1-G and Simplex-P cements are 500%, 450% and 425% respectively. The model that was used to fit the experimental data (Equations 9 and 10) is defined using information from both the stress relaxation and DMA tests.

through a dual mechanism (that is, part of water is inserted into the polymer network and part is accommodated into pre-existing microvoids) and the equilibrium sorption can be described by the dual sorption model [88]. The water absorption experiments have evidenced differences among the tested cements. Particularly the lower rate absorption of the Simplex-P cement compared to the Braxel and CMW1-G cements is ascribable to the hydrophobic nature of the styrene group. It is important to note that the radical polymerization of PMMA does not proceed to competition [89, 90] and the non-converted residual monomer in the polymerized bone cement is about 2–6% [18]. Thus, beside water absorption, conditioned specimens (samples 1, 2, 4, 5) are characterized also by the release of unreacted monomer. It has been pointed out that at 40 °C the monomer loss rate in water is higher than in air [38].

The mechanical experiments that were performed have nonetheless evidenced certain differences among the various bone cements, both in stress relaxation and in dynamic mechanical testing. In the whole frequency sweep range the mean values of the storage modulus (Fig. 9) for the CMW1-G cement are always in between those of the Simplex-P and Braxel cements. For every bone cement the whole increase in E' is of about 55%. Particularly the storage modulus of the Simplex-P and the CMW1-G cements at 0.01 Hz are 2785 MPa (SD 162 MPa) and 2231 MPa (SD 114 MPa) respectively. Ma *et al*. [91], by using PMMA with a 12.4% ion content, compression molded at 200 °C, testing at 1 Hz and at 30 °C in dry condition, found a value of 3 GPa for the storage modulus. Gilbert *et al*. [41], by using the three point bending method on the Simplex-P cement at room temperature, testing in the high cycles fatigue at 5 Hz, measured a higher value of the storage modulus (4.1 GPa). This difference may be due to water effects; indeed even static tests show that water reduces the elastic modulus of about 10% [18], of course the damping factor in dry condition will be lower than in wet conditions. Fig. 9 also shows that the loss modulus

TABLE V Numerical values for the parameters used to model the bone cement behavior

	E_∞	C_1	C_2	C_3	τ_1	n
Braxel	100	1200	90	19	$6 \cdot 10^5$	1.5
CMW1-G	1000	600	100	3	$2 \cdot 10^5$	2.0
Simplex-P	1000	1300	130	15	$9 \cdot 10^4$	1.1

TABLE VI Comparison between mechanical properties of bone cement, bone and metals, used in prosthetic implants

	Elastic modulus (GPa)	Static strength (MPa)	Hardness (GPa)	Fracture toughness ($MPa \cdot m^{1/2}$)	Fatigue strength (MPa)	Storage modulus at 1 Hz (GPa)	tan δ at 1 Hz
Bone cement	3	> 50	0.3	1.3	10–20	2.2–3.2	0.06–0.07
Cortical bone	10–20	50–300	0.4–0.7	3.8	50–100	10.2	0.04
Spongy bone	0.1–1	2–20	0.3–0.6	—	—	0.2	0.09–0.10
Ti6Al4V	100	900	400	12–21	300–700	10^2	10^{-4}
CoCr Alloys	200	1000	100	50–200	600–800	$> 10^2$	10^{-4}

for the CMW1-G cement are in between those of the Simplex-P and Braxel cements. One-way ANOVA analysis suggests that these mean values are significantly different, reaching the highest statistical significance at 100 Hz when comparing the loss modulus of Braxel and CMW1-G ($p = 0.02$, $F = 8$). While the storage modulus increase with frequency is similar for the three cements, the loss modulus increase differs: between 0.01 and 100 Hz the loss modulus increase for the Braxel, CMW1-G and Simplex-P cements are 500%, 450% and 425%, respectively (Fig. 9). Robertson *et al.* [75], by using dynamic investigation in tension on industrial atactic PMMA, processed by means of compression molding at 170 °C, found, in dry conditions at 1 Hz, a loss modulus of about 200–300 MPa at 37 °C, which is consistent with the range of values of the loss modulus found in this work for the bone cements at 1 Hz (186.8–250.7 MPa). In particular, Simplex-P has displayed the highest values of the mechanical properties and Braxel has shown the lowest values, with the exception of a slight improvement in the damping factor, while CMW1 has shown properties, which are intermediate between Simplex-P and Braxel. This behavior is in general agreement with the different composition of these cements, since the rigid polystyrene elements of Simplex P, inserted into the main PMMA chain, result in a higher stiffness with respect to the pure PMMA main chain of CMW1, while the longer and softer butyl-methacrylate components of Braxel imply lower mechanical properties and a comparatively larger dissipation. The use of PBMMA as bone cement originated from the need of a material that better distributes the peak loads due to interfacing between different materials [39]. As we can see from the relaxation curves, this behavior is also coupled to a steeper reduction of modulus with time and may render this material dangerously prone to excessive creep and hence aseptic loosening.

It is quite remarkable that, even though the moduli of the three cements are quite different, the damping factor at body temperature is very similar for all the three cements, possibly indicating that the most important factor influencing $\tan\delta$ is the main constituent of the polymer chain, which is indeed PMMA for all the three cements. In any case the plot of the damping factor as a function of the frequency indicates a remarkable capacity of energy dissipation at 37 °C, which increases with the loading frequency, at least up to 100 Hz. As a consequence, the cements will comparatively dissipate more energy the higher is the strain rate and hence functioning as shock absorbers.

This is an extremely attractive feature of these materials because it allows the prosthetic implant to damp occasional impulsive overloads (e.g. loads applied in less than 0.01 s) and can be very beneficial in the case of patients who have sustained extensive bone loss, such as elderly patients [10–16].

It is important to note that, in the explored strain and frequency ranges, DMA suggests that the viscoelastic properties are independent of the mean strain level and of the dwelling time t_D (Fig. 3). The former independence is sensible since we are exploring the linear viscoelastic region of the cements (Fig. 2), while the independence of the dwelling time directly implies the independence of the mean stress level.

Since $\tan\delta$ is independent of specimen geometry, the comparison between our DMA results in wet environment and literature data in dry condition [65–70] is sensible if related to the damping factor measurements. Our investigation clearly shows that in a wet environment the onset of the α peak reflected in $\tan\delta$ (Fig. 5) is shifted to lower temperature than in dry condition. Thus water plays a very important role on the viscoelastic behavior of bone cements, lowering the glass transition temperature and acting as a plasticizer. Important mechanical properties (i.e. toughness or impact strength) of glassy polymers are related to the local structure and to the time-scale of molecular motions of these polymeric materials at their working temperature. PMMA is a special polymer since secondary loss peaks are markedly observed by dielectric [65, 92–94], enthalpic [94–96], stress-relaxation [97] and DMA techniques (Fig. 5) [65–70]. Nuclear magnetic resonance and quasi elastic neutron scattering suggest that the origin of the secondary relaxation or β-relaxation is controled by the partial rotation of the carboxymethyl group [98–100]. This motion is hindered by intramolecular and intermolecular constraints, regarded as potential barriers [101]. If the barrier height is great compared with the molecular energy, the motion may be treated as harmonic oscillations in the potential wells and jumping between three equidistant angular sites (120° jumping) of energy minima [99–102]. In glassy amorphous PMMA, due to the differences between intermolecular interactions, there exists a distribution of potential barriers, thus each CH_3 group feels a different local environment and therefore rotate with different frequency [98, 102, 103]. Thus, the parametric curves of Fig. 5 are broad and their half-width decrease and height increase with increasing frequency.

The effect of specimen conditioning (length of time since mixing, temperature, hydration and dwelling time) for the Simplex-P cement is clearly shown in Figs. 6 and 7. It is remarkable that at 37 °C the stress-relaxation test is suitable to depict the effect of the conditioning on the mobility of the polymer main chain, at the same temperature the DMA test better describe the effect of the conditioning on the mobility of the polymer side chain. The much faster relaxation of sample 2 (Fig. 6) or equivalently the higher damping factor values of sample 2 at 0.01 Hz (Fig. 7) clearly indicate that the residual unreacted monomer plays an important role on the main chain mobility. As the ureacted monomer is released through a post curing process (sample 3) or through a prolonged conditioning process (sample 5), the values of $\tan\delta$ at low frequency drastically drops (Fig. 7), or equivalently, the cement relaxation is lower (Fig. 6). It is interesting to note that water uptake of the cements (sample 3 at high values of t_D, and sample 5) will not affect $\tan\delta$ at 0.01 Hz as the residual monomer does. A different effect is instead observed for the side chain mobility, which at 37 °C is only shown through DMA (Fig. 7). Indeed at high frequency the unreacted monomer increase the damping capability of the cement (sample 2). However, as water is taken up by the cement, the same $\tan\delta$ values are measured at 100 Hz (sample 3 at high values of t_D, and sample 5). This similar effect of the unreacted monomer and water at high frequencies may be related to the small role played by the solvent thermodynamic quality on the time-scale and activation energy for local conformational transitions. Thus under a mechanical point of view the residual unreacted monomer, *in vivo* applications of bone cements, mainly affect the relaxation or creep [38, 54] in a transient manner, while water absorption mainly increase the damping capability at high frequency, thus affecting the side chain mobility.

The model that was used to fit the experimental data has two relaxation times, one of which is evident from the constitutive expression (Equation 7), while the other one is implicit in the expression for $K(\tau)$ in Equation 8. Thus the complete plot of the theoretical damping factor vs. frequency would show two maxima. The frequency range, in which the bone cements were tested at 37 °C, is not wide enough to confirm experimentally the presence of two such maxima. Dielectric loss measurements suggest that at room temperature the peak related to the β transition appears at about 10^5 Hz [92]. Nevertheless, we can infer the presence of two relaxation times from the different behavior of the relaxation and of the DMA curves. Thus it is practically impossible to formulate a model based on only one of the two experiments and use it to predict the results of the other one, at least within the context of classical linear viscoelasticity, hence the stress relaxation and DMA experiments have been considered as complementary to each other and the model for the material behavior is obtained using information from both the experimental tests.

At 37 °C the bone cement is in a glassy state, characterized by adequate mechanical properties and brittleness. However, the damping factor increase with frequency denotes the presence of a secondary relaxation mechanism. This is a common feature of branched polymers, such as PMMA, one of the relaxation mode being linked to the mobility of the side branches, while the other one to the mobility of the main chain. This is an advantageous characteristic of the material, because the brittleness, that is typical of glassy polymers, is relieved by the secondary relaxation mechanism, which is active at the body temperature and in the frequency range in which the cements were tested, and this makes this material tougher than an ordinary polymeric glass.

Conclusion

The three cements have shown a similar qualitative behavior, characterized by the coupling of good mechanical and damping properties. From a quantitative point of view a few differences among the various cements have been found: Simplex P appears to be the cement with the highest moduli, while Braxel has shown the lowest properties, particularly regarding the results of the stress relaxation tests and the storage modulus.

As far as the damping properties are concerned, all the tested materials have shown an interesting behavior: the value of the damping factor, proportional to the part of external mechanical work that is dissipated, is an increasing function of the loading frequency, thus the material will better function as a mechanical buffer the higher is the loading frequency, and hence working well against the potentially bad consequences of an impulsive overload.

Interpreting the results of this analysis in terms of the different composition of the three cements, it can be concluded that the copolymerization of methyl-methacrylate with butyl-methacrylate has the desired effect of slightly increasing the damping factor of the material. However, this happens at the price of lowering the creep resistance of the material, as it can be deduced from the stress relaxation tests that have been performed. The copolymerization of methyl-methacrylate with styrene, on the other hand, increases the mechanical properties of the material, without compromising the damping factor, which remains anyway in the range of 0.1–0.25.

The presence of the antibiotic in the bone cement does not appear to excessively decrease the mechanical properties, although its efficacy inside the scarcely permeable bone cement yet remains to be proved.

At 37 °C stress-relaxation tests underline the behavior of the main polymer chain, while DMA tests better describe the mobility of the side chain.

At 37 °C, in the explored strain and frequency ranges, DMA suggests that the viscoelastic properties are independent of the stress level. Moreover, DMA in the wet condition suggests that the α transition shifts to lower temperature if compared to the dry condition.

The presence of unreacted monomer increases the relaxation rate of the cements, thus mainly affecting the main chain mobility, while water absorption mainly affect the side chain mobility, increasing the damping properties of the cement at 37 °C.

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