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Physical and chemical aspects of calcium phosphates used in spinal surgery

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Abstract In 1983, Brown and Chow [10] discovered the first calcium phosphate cement (CPC). This discovery opened a new era in the use of calcium phosphates (CaP) as bone substitute. The first commercial CPC have now been introduced, and porous CaP with totally new properties should follow soon. In this article, the CaP traditionally used in spinal surgery are compared with

these new CaP. Particular emphasis is placed on the use of CPC for bone augmentation.

Keywords Calcium phosphate · Cement · Bone · Substitute · Spine

Introduction

Calcium phosphates (CaP) have been studied as bone repair materials for the last 80 years [1]. CaP only began to be used in spinal surgery more recently, but CaP such as β -tricalcium phosphate (β -TCP), hydroxyapatite (HA), and their composites (biphasic calcium phosphate, BCP) are commonly used as granules or blocks in interbody fusion and scoliosis surgery [29, 41, 43]. Recently, new CaP denominated calcium phosphate cements (CPC) have been introduced [10]. These materials are made of one or several CaP powders and an aqueous solution. When the CaP powders and the aqueous solution are mixed together, a paste is obtained that sets within a few minutes. CPC are usually injectable and have therefore been proposed as a possible alternative to polymethylmetacrylate (PMMA), e.g., to reinforce osteoporotic vertebral bodies [4, 44] and thoracolumbar burst fractures [33] and to restore pedicle screw fixation [36]. CPC can also be used to make granules and blocks out of low-temperature CaP [6]. These CaP are very similar to the CaP present in bone. They possess a very large specific surface area, which makes them biologically much more (re)active than traditional CaP. Despite the increasing use of CPC in medicine, particularly in spinal surgery, there are still very

few articles in the medical literature describing CaP other than β -TCP, HA, and BCP [7]. The aim of this article is therefore to describe the differences between these new CaP and the more traditional CaP. Two different categories of CaP should be distinguished: (a) CaP obtained by precipitation from an aqueous solution at or around room temperature (*low-temperature CaP*) and (ii) CaP obtained by a thermal reaction, typically above 1000°C (*high-temperature CaP*). All CPC (e.g., α -BSM, Bone-Source, Norian SRS) belong to the first category, whereas most other CaP products (e.g., chronOS, Endobon, BCP, Pro Osteon, Triosite) belong to the second category. In the first part of this article, low-temperature CaP will be compared with high-temperature CaP; in the second part, CPC will be presented. Particular emphasis is placed on the use of CPC as bone augmentation material in spinal surgery.

Low-temperature calcium phosphates

Low-temperature CaP have a different chemical composition and structure than high-temperature CaP (Table 1), except for stoichiometric hydroxyapatite (Ca:P molar ratio of 1.67), which can precipitate at room temperature and which is stable up to 1300°C. As hydroxyapatite can precipitate over a large range of Ca:P molar ratios (1.50–

Table 1 Main calcium phosphate compounds

Name	Formula	Ca/P	Mineral	Symbol
Monocalcium phosphate monohydrate	$\text{Ca}(\text{H}_2\text{PO}_4)_2 \cdot \text{H}_2\text{O}$	0.50	–	MCPM
Dicalcium phosphate	CaHPO_4	1.00	Monetite	DCP
Dicalcium phosphate dihydrate	$\text{CaHPO}_4 \cdot 2\text{H}_2\text{O}$	1.00	Brushite	DCPD
Octocalcium phosphate	$\text{Ca}_8\text{H}_2(\text{PO}_4)_6 \cdot 5\text{H}_2\text{O}$	1.33	–	OCP
Precipitated hydroxyapatite	$\text{Ca}_{10-x}(\text{HPO}_4)_x(\text{PO}_4)_{6-x}(\text{OH})_{2-x}$	1.50–1.67	–	PHA
Tricalcium phosphate	$\text{Ca}_9(\text{HPO}_4)(\text{PO}_4)_5(\text{OH})$	1.50	–	TCP
Amorphous calcium phosphate	$\text{Ca}_3(\text{PO}_4)_2 \cdot n\text{H}_2\text{O}^a$	1.50	–	ACP
Monocalcium phosphate	$\text{Ca}(\text{H}_2\text{PO}_4)_2$	0.50	–	MCP
α -Tricalcium phosphate	$\alpha\text{-Ca}_3(\text{PO}_4)_2$	1.50	–	α -TCP
β -Tricalcium phosphate	$\beta\text{-Ca}_3(\text{PO}_4)_2$	1.50	–	β -TCP
Sintered hydroxyapatite	$\text{Ca}_5(\text{PO}_4)_3\text{OH}$	1.67	Hydroxyapatite	HA
Oxyapatite	$\text{Ca}_{10}(\text{PO}_4)_6\text{O}$	1.67	–	OXA
Tetracalcium phosphate	$\text{Ca}_4(\text{PO}_4)_2\text{O}$	2.00	Hilgenstockite	TetCP
Carbonated apatite	$\text{Ca}_{8.8}(\text{HPO}_4)_{0.7}(\text{PO}_4)_{4.5}(\text{CO}_3)_{0.7}(\text{OH})_{1.3}$		Dahlite	CAP

The first six compounds precipitate at room temperature in aqueous systems. The last six compounds are obtained by thermal decomposition or thermal synthesis.

^a $n=3\text{--}4.5$; 15–20% H_2O .

1.67), a distinction is made between the low-temperature form, referred to here as precipitated hydroxyapatite (PHA), and the high-temperature form, referred to here as hydroxyapatite (HA). PHA is very similar to the mineral part of bone, whereas HA is the hydroxyapatite traditionally used in medicine. Low-temperature CaP are characterized by their large specific surface area (Fig. 1). For example, PHA can have a specific surface area as large as $100 \text{ m}^2/\text{g}$, whereas HA has a specific surface area of around $1 \text{ m}^2/\text{g}$. This large specific surface area makes low-temperature CaP biologically much more (re)active than high-temperature CaP. Low-temperature CaP have different solubilities than high-temperature CaP (Fig. 2).

Differences in the in vivo bioresorption rate can therefore be expected [14].

Among the low-temperature CaP, dicalcium phosphate dihydrate (DCPD; $\text{CaHPO}_4 \cdot 2\text{H}_2\text{O}$) and PHA are particularly interesting. DCPD is the most bioresorbable biocompatible CaP (MCPM is more soluble but is not biocompatible; Fig. 2), whereas PHA is the CaP most similar to the mineral part of bone. DCPD is the CaP that can be most easily synthesized. It has been detected in fracture callus [32], bone [38], and kidney stones [47] and is also sometimes considered to be a precursor of HA in bone [20]. DCPD is biocompatible and osteoconductive. At physiological pH, DCPD is the most soluble low-temper-

Fig. 1a,b Typical microstructure of **a** a high-temperature calcium phosphate (CaP) (chronOS, Mathys, Switzerland; specific surface area, $\approx 1 \text{ m}^2/\text{g}$) and **b** a low-temperature CaP (CDHA; specific surface area, $\approx 22 \text{ m}^2/\text{g}$). The width of the right photo is $8 \mu\text{m}$. (Courtesy of P. Buffat, Swiss Federal Institute of Technology, Lausanne, Switzerland)

Fig. 2 Solubility isotherms of several calcium phosphates (CaP) in water. The solubility is expressed as the total amount of calcium ions in solution. The solubility data were taken from Vereecke and Lemaître [46]

ature CaP (Fig. 2). It is therefore metastable and can be converted into other low-temperature CaP, such as dicalcium phosphate (DCP; CaHPO_4 ; at $\text{pH} < 6$), octocalcium phosphate (OCP; $\text{Ca}_8\text{H}_2(\text{PO}_4)_6 \cdot 5\text{H}_2\text{O}$; at $\text{pH} \approx 6-7$), or PHA (at $\text{pH} > 7$). In vivo DCPD is normally rapidly bioresorbed and replaced by new bone [39]. However, DCPD has been observed to transform into PHA, especially in a location poorly irrigated with blood [11]. The addition of small amounts of magnesium, which is an inhibitor of the precipitation of PHA, is believed to prevent the transformation of this magnesium-containing DCPD into PHA [8], hence making DCPD the most bioresorbable biocompatible CaP.

PHA is less soluble than DCPD, but accurate values are lacking due to the fact that PHA can have a Ca:P molar ratio of between 1.50 and 1.67 [9] and sometimes even outside this range [12] (Table 1). The solubility of PHA is believed to increase as the Ca:P molar ratio decreases. Another difficulty in determining the solubility of PHA stems from the fact that PHA can easily incorporate foreign ions such as carbonates, chloride, fluoride, and mag-

nesium ions. These ions may have a very strong influence on the solubility (e.g., fluoride). Finally, the size of the particle crystals is very small in PHA, typically close to 10–20 nm. In this range, the solubility depends on the particle size: the smaller the particle size, the greater the solubility. PHA with a Ca:P molar ratio of 1.50 is often called calcium-deficient hydroxyapatite (CDHA) or tricalcium phosphate (TCP). The latter term has led to it being confused with β -TCP, which has almost the same chemical composition, but a different crystallographic structure. The solubility of TCP is estimated to be close to that of β -TCP [13]. Therefore and contrary to HA, most PHA should be considered as bioresorbable. Apart from brushite CPC, all commercial CPC have PHA (Bone-Source, Cementek, α -BSM, Biopex, Mimix) or carbonated PHA (Norian SRS and CRS, Biocement D) as the main end product of the setting reaction. The main difference between the cements after the setting reaction lies in the Ca:P molar ratio, the presence of impurities (e.g., carbonate groups for Norian SRS and Biocement D), the cement porosity, and the crystal size of the PHA. All these parameters influence the bioresorbability of PHA.

With the discovery of CPC [10] and the use of CaP emulsions [6], low-temperature CaP can be produced in almost any shape and porosity. Granules and porous blocks can therefore be made out of CaP other than the traditionally used CaP, i.e., β -TCP, HA, and BCP. These new possibilities are particularly interesting for applications where a fast bioresorption rate or a large specific surface area is required, e.g., tissue engineering or drug delivery. An interesting assumption is that, having a large specific surface area, CaP can adsorb large amounts of endogenous proteins, such as growth factors. Low-temperature CaP might then be indirectly osteoinductive and could therefore be used in applications where fast bone ingrowth is required. To test this hypothesis, intervertebral body fusion cages were filled with an open macroporous CDHA block and implanted in sheep. The 6-month results are expected soon (T. Steffen, private communication). If this assumption proves to be correct, the use of CaP as bone substitute could spread very quickly. Presently,

Table 2 Main apatite calcium phosphate cements (CPC)

	Cement name	Company	Components	End product
ACP, amorphous calcium phosphate; CAP, carbonato-apatite; DCP, dicalcium phosphate; DCPD, dicalcium phosphate dihydrate; MCPM, monocalcium phosphate monohydrate; PHA, precipitated hydroxyapatite; TCP, tricalcium phosphate; TetCP, tetracalcium phosphate.	α -BSM	ETEX	ACP, DCPD [46]	PHA
	Biobone	Merck GmbH	—	—
	Embarc	Lorenz Surgical	—	—
	Norian SRS	Norian (Synthes-Stratec)	α -TCP, CaCO_3 , MCPM [12]	CAP
	Norian CRS			
	Bonesource	Leibinger	TetCP, DCP [10]	PHA
	Cementek	Teknimed	α -TCP, TetCP, $\text{Ca}(\text{OH})_2$, H_3PO_4	PHA
	Mimix	Lorenz Surgical	α -TCP, TetCP, citric acid	PHA
	Biocement D	Merck GmbH	α -TCP, DCP, CaCO_3 , PHA [26]	CAP*
	Biopex	Mitsubishi Materials	α -TCP, TetCP, DCPD [28]	PHA
	Fracture Grout	Norian (Synthes-Stratec)	TetCP, CaCO_3 , H_3PO_4 [21]	PHA
	chronOS Inject	Mathys Medical	β -TCP, MCPM	DCPD

the only commercial products on the market based on low-temperature CaP are CPC (Table 2). CPC have the disadvantage that they bioresorb layer by layer as they do not have any pores large enough to enable bone ingrowth.

High-temperature calcium phosphates

Until the discovery of CPC, it was not possible to synthesize CaP granules or blocks without thermal treatment (typically above 1000°C). All the traditional CaP used in medicine are therefore high-temperature CaP, mostly β -TCP, HA, and β -TCP/HA composites [3]. Due to the low toughness of CaP, the use of high-temperature CaP in spinal surgery has been mainly restricted to non-load-bearing applications. Nevertheless, high-temperature CaP have been used to make cervical cages for interbody fusion [27].

The most soluble high-temperature CaP are tetracalcium phosphate (TetCP; $\text{Ca}_4(\text{PO}_4)_2\text{O}$) and α -tricalcium phosphate (α -TCP; $\alpha\text{-Ca}_3(\text{PO}_4)_2$). TetCP and α -TCP are about two orders of magnitude more soluble than HA at physiological pH (Fig. 2). TetCP and α -TCP therefore tend to transform into PHA when implanted in vivo. Both compounds are difficult to synthesize, so TetCP and α -TCP have mostly been used as the main constituents of CPC (Table 2). One known commercial α -TCP product is BioBase (BioVision, Germany).

β -TCP has the same composition as α -TCP, but a different crystallographic structure. As a result, β -TCP is one order of magnitude less soluble than α -TCP (Fig. 2). β -TCP can be obtained by thermal treatment above at least 650°C [5], e.g., by heating up CDHA above $\sim 700^\circ\text{C}$ [37]. It has been extensively used as bone substitute, either as granules or as blocks, and is bioresorbable, typically in the range of 1–2 years. Bioresorption occurs via osteoclastic activity [19]. There are a few products on the market, such as chronOS (Mathys, Switzerland), Vitoss (Orthovita, USA), and Biosorb (SBM, France). β -TCP has been successfully used in clinical practice, e.g., for interbody fusion in combination with spinal cages [23] and in scoliosis [29].

HA is the most stable CaP in an aqueous solution (Fig. 1) and is sometimes considered to be the most biocompatible CaP. It is also the CaP with the largest mechanical properties at a given porosity. HA is obtained by sintering stoichiometric PHA (Ca:P molar ratio, 1.67) above $\sim 700^\circ\text{C}$. At temperatures higher than $\sim 900^\circ\text{C}$, partial dehydration of HA may take place, eventually resulting in oxyapatite (OXA) [25]. This reaction only takes place in the absence of water vapor [14]. As OXA is difficult to detect, most commercial HA products may contain some OXA. The best-known commercial products are Pro Oseon (Interpore Cross, USA), Endobon (Merck, Germany), and Pyrost (Stryker Howmedica, USA). These products should be considered as non-bioresorbable, as they have a resorption time of decades rather than years.

Biphasic calcium phosphate (BCP) is a composite β -TCP-HA. It is obtained by calcining PHA (with a molar ratio Ca:P<1.67) above $\sim 700^\circ\text{C}$ [5]. BCP has intermediate properties compared with β -TCP and HA. Most commercial products, e.g., Triosite (Zimmer, USA) and BCP (Sofamor Danek, France), contain 60% HA and 40% β -TCP. BCP has been widely used in spinal surgery, e.g., for scoliosis [41, 43].

Calcium phosphate cements

CPC were discovered by Brown and Chow in the 1980s [10]. Since then, many compositions have been proposed [15, 16, 30]. However, despite the large number of formulations, CPC can only have three different end products: apatite (PHA), brushite (DCPD), and amorphous calcium phosphate (ACP). The only study performed with ACP [17] shows that the cement is rapidly converted into PHA. We can therefore classify all CPC formulations into two categories: (a) apatite CPC and (b) brushite CPC. Research effort has largely focused on apatite CPC, despite some interesting features of brushite CPC, mainly its large bioresorption rate. The following aspects of CPC will be considered here: chemistry, mixing, rheological properties, mechanical properties, porosity, heat release, volume change, and bioresorbability.

CPC are made of an aqueous solution and of one or several calcium phosphates. Upon mixing, the calcium phosphate(s) dissolve and precipitate into a less soluble CaP. Two types of setting reaction are possible. First, the setting reaction can occur according to an acid–base reaction, i.e., a relatively acidic CaP reacts with a relatively basic CaP to produce a relatively neutral CaP. A typical example is the cement developed by Brown and Chow [10], in which TetCP (basic) reacts with DCP (neutral) to form PHA (slightly basic):

Another example is the cement developed by Lemaître [31], in which β -TCP (slightly basic) reacts with MCPM (acidic) to form DCPD (neutral):

The second type of setting reaction occurs when the initial and final CaP have the same Ca:P molar ratio. Typical examples are ACP and α -TCP, which form PHA upon contact with an aqueous solution:

An important requirement for the clinician is the mixing procedure of the CPC. So far, most CPC are mixed by hand with a mortar and either a pestle or a spatula. To be injected, the cement paste must be transferred into a sy-

ringe. Three commercial products are sold with an appropriate mixing device. Norian SRS is sold as a reactant pack containing the powder mixture and the mixing liquid and requires the use of a rather complicated and not always reliable system, i.e., a very large mixing machine and an injection gun. The large pressure obtained with the gun makes it easier to inject the cement, e.g., into a porous and fine structure such as osteoporotic bone. α -BSM is also a two-component system. The powder is in a soft pouch, and the liquid is in a flask. After injecting the liquid into the powder pouch, the cement paste is mixed with the fingers. After 3 min, the paste can be injected. This method is easy and very elegant, but as the setting time of α -BSM is temperature dependent, results may not be always reproducible. Moreover, the pressure that can be applied on the cement is limited. In the product Biopex, the cement is mixed with a spatula in a syringe that can be opened from the front. After mixing, the front part is closed, a needle is inserted into this front part, and the cement paste can be injected by hand.

To be injected *in vivo*, e.g., in an osteoporotic vertebra, the CPC paste must have two features: (a) injectability and (b) cohesion. Injectability means that the cement paste can be extruded through a long, small needle (e.g., 2 mm diameter and 10 cm length) without demixing. Demixing occurs when the mixing liquid is too fluid compared with the size of the cement powders, resulting in filter pressing: the liquid comes out without the CaP particles. The risk of demixing increases with the applied pressure. A cement paste with an appropriate cohesion sets in a fluid without disintegrating. This can be obtained by maintaining a high viscosity for the CPC paste. Some CPC pastes fulfill these two criteria, e.g., Norian SRS and α -BSM, but other CPC fulfill only one or neither of these requirements. For example, blood must be kept away from the implanting site of BoneSource and Cementek until setting. The loss of cohesion of the cement during mixing may lead to inflammatory reactions [34].

Several approaches can be adopted to improve the injectability and cohesion of a CPC. First, the powder-to-liquid ratio, P/L, can be decreased. The resulting paste is less viscous and can be more easily injected. A similar phenomenon is seen with PMMA cements when the monomer-to-powder ratio is increased. However, a decrease in the powder-to-liquid ratio of the CPC leads to a decrease in the mechanical properties of the CPC, and the cohesion of the CPC may be destroyed. This method is therefore only valid to a certain extent. For example, the manufacturers of Biopex suggest using a P/L ratio of 2.8 or 3.3 g/ml. A second approach is to decrease the particle size of the CPC components. For example, α -BSM, which has a small particle size, is well injectable. However, a small particle size requires a large amount of mixing liquid, which decreases the cohesion of the paste and also leads to greater porosity and lower mechanical properties. A third approach is to modify the interactions between the

particles by shifting the isoelectric point by adding so-called surfactants. This approach has recently been used by Friberg et al. [22], but has not been applied in commercial products. An indirect approach is to add CaP particles which act as spacers between other CaP particles. For example, DCP is added to the formulation of Bio cement D to improve the injectability of the paste. Finally, a fourth approach is to replace the mixing solution by a viscous polymeric solution [2]. Polysaccharides are particularly interesting due to their compatibility and good rheological properties. Most polysaccharide solutions are thixotropic, i.e., the viscosity of the solution decreases as the shear rate increases. A CPC made using a polymeric solution can therefore be easily injected and has a good cohesion in the absence of shear stresses. Only small amounts (a few weight%) of polysaccharides are normally needed to dramatically increase the viscosity of the mixing liquid. For example, Biopex uses a 5%w sodium chondroitin sulfate solution as mixing liquid. However, only very few polysaccharides are accepted for parenteral use. Moreover, the polymer solution can usually only be sterilized by ultrafiltration.

Another important aspect of cement injectability is the radiopacity. None of the CPC is as radiopaque as PMMA cement. Several research groups have tried to add a radiopacifier to CPC, but the biological outcome of the addition of small radiopaque particles to a bioresorbable CPC is not clear. Nevertheless, radiopacity differences can be expected between different formulations, i.e., the radiopacity should increase as the cement porosity decreases.

Considering their properties, CPC might potentially be used to reinforce osteoporotic vertebral bodies [4, 44] and thoracolumbar burst fractures [33] and to restore pedicle screw fixation [36]. Despite encouraging results, surgeons' expectations have not been met. This is not only due to the lower mechanical properties of CPC compared with PMMA cements, but also, and more importantly, to the greater difficulties encountered in filling the vertebral bodies. Several causes can be suggested: (a) in order to maintain their cohesion, CPC need to be more viscous than PMMA cement, hence rendering CPC less injectable than PMMA cement, and (b) CPC are hydrophilic, whereas PMMA cements are hydrophobic. CPC therefore tend to mix with body fluids and lose their cohesion, whereas PMMA cements tend to stay compact. To prevent these problems, several approaches have been proposed. In one approach, a cavity is created in the vertebral body, e.g., by means of an expandable balloon. The cavity is then filled with the CPC. In another approach, bone marrow is removed from the vertebra using a suction device [44], and the CPC is then injected.

During the setting reaction of the cement, CaP crystals grow and become partly interlocked, hence making the cement mechanically rigid. As CaP are brittle, the compressive strength of a CPC is always much greater than its tensile strength. Most CPC have a tensile strength of

1–10 MPa, whereas the compression strength varies between 10 and 100 MPa. For comparison, the tensile strength of PMMA is 30 MPa [40]. The mechanical properties of a CPC depend on its composition. The main determining factor is the ratio between the amount of cement powder (P) and mixing liquid (L). If this P/L ratio is large, the porosity of the CPC is low. As the mechanical properties decrease exponentially as porosity increases, low porosity always corresponds to large mechanical properties. As a rule of thumb, the tensile strength increases twofold with a 10 vol% decrease in porosity. The mechanical properties reported by CPC manufacturers are never documented, so it is difficult to compare CPC. Several authors have attempted to compare CPC cements. For example, Driessens [18] established compression strengths of 33 ± 5 , 8 ± 2 , 83 ± 4 , and 4 ± 1 MPa and setting times of 8.5 ± 0.5 , 17 ± 1 , 6.5 ± 0.5 , and 19 ± 1 min for Norian SRS, Cemetek, Biocement D, and α -BSM, respectively. Biocement D has by far the greatest compressive strength, but this value does not mean that Biocement D is the least breakable implant. In vivo, shear and tensile forces play a very important role. The tensile strength of the CPC therefore also needs to be considered, e.g., using the Mohr circle approach [42]. Finally, it should be kept in mind that the initial mechanical properties of the apatite CPC may vary with implantation time. Animals studies indicate that the mechanical properties of apatite CPC tend to increase continually [35], in contrast to those of brushite CPC, which initially decrease and then increase when bone grows [24].

CPC typically have a porosity of 30–60% vol depending on the cement formulation. The porosity is open, i.e., the pores are perfectly interconnected. The pore size is close to 1 μm . Thus the pores are too small to allow fast bone ingrowth and the CPC bioresorbs layer by layer. This feature is the main drawback of CPC when compared to open macroporous CaP blocks. The porosity of the CPC is due to the excess water used in the CPC composition. Attempts have been made to reduce the amount of water in order to decrease the porosity and hence increase the mechanical properties, but a decrease in porosity also leads to a decrease in the CPC bioresorption rate. Moreover, the

amount of water determines the rheological properties of the cement paste: a decrease in the CPC water content leads to a large increase in the viscosity of the cement paste, eventually leading to a nonflowable mixture.

Most CPC release some heat during setting. However, CPC are not as exothermic as PMMA cements or, if they are (e.g., BoneSource, a mixture of TetCP and DCP), the release rate is low enough to allow good dissipation of the heat.

In contrast to PMMA cements, CPC set at an almost constant volume. The final cement porosity can therefore be accurately predicted from the initial composition. The absence of shrinkage enables good contact between bone and CPC during and after setting.

The bioresorption rate of brushite CPC (e.g., chronOS Inject) is higher than that of apatite CPC (e.g., α -BSM), which in turn is higher than that of HA. However, the bioresorption rate of apatite CPC is still very slow, particularly due to the fact that CPC are not macroporous. For example, Young et al. [48] reported a 30% decrease in the amount of Norian SRS after 24 months in a rabbit femur. Some differences in bioresorption rate can be expected depending on the type of apatite CPC. Generally speaking, a longer bioresorption time is expected when the crystal size increases or when the porosity decreases. Norian SRS and α -BSM are therefore expected to bioresorb faster than BoneSource, Biopex, and Cemetek.

Conclusion

Traditional CaP (HA, BCP, β -TCP) have commonly been used as granules or blocks to accelerate or improve spinal fusion. The discovery of CPC has opened up new horizons in the use of CaP in spinal surgery. CPC can be injected into osteoporotic bone to reinforce it or can be used to make granules and blocks out of low-temperature CaP. Several CPC are now on the market, while low-temperature CaP granules and blocks are still being tested. However, CPC already need to be improved; in particular, their bioresorption needs to be accelerated and their injectability and mechanical properties improved.

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