sciendo ASIAN BIOMEDICINE

DOI 10.1515/abm-2021-0003 — Asian Biomed (Res Rev News) 2021; 15(1):19–26

Technical report **Open access**

Injectable calcium phosphate and styrene–butadiene polymer-based root canal filling material

Hala B. Kaka[1,*], Raid F. Salman[2]

Abstract

Background: Three-dimensional obturation of the root canal system is mandatory for a successful root canal treatment. Using a filling material with optimal properties may enable the root canal to be sealed well and therefore obtain the desired obturation.

Objective: To develop a new injectable paste endodontic filling material using calcium phosphate powder and a styrene–butadiene emulsion polymer.

Methods: The powder phase comprised an equivalent molar ratio of tetracalcium phosphate, anhydrous dicalcium phosphate, bismuth oxide, and calcium chloride. The liquid phase comprised a styrene–butadiene rubber emulsion in distilled water. The powder and the liquid were mixed to achieve a paste consistency. The paste was subjected to various tests including flow, setting time, dimensional change, solubility, and radiopacity to indicate its suitability as a root canal filling material. All these tests were conducted according to the American National Standards Institute–American Dental Association for endodontic sealing materials. After passing these tests, the paste was submitted to an injectability test.

Results: The material showed acceptable flowability with 19.1 ± 1.3 min setting time and 0.61 ± 0.16% shrinkage after 30 days of storage. We found the highest solubility at 24 h (6.62 ± 0.58%), then the solubility decreased to 1.09 ± 0.08% within 3 days. The material was more radiopaque than a 3 mm step on an aluminum wedge. Furthermore, the material showed good injectability of 93.67 ± 1.80%.

Conclusions: The calcium phosphate powder in styrene–butadiene emulsion met basic requirements for a root canal filling material with promising properties.

Keywords: American Dental Association; calcium phosphates; root canal filling materials; solubility; styrene-butadiene rubber

The objective of root canal filling procedures is 3-dimensional obturation of the root canal system after complete cleaning and shaping, to eliminate bacteria and tissue debris from within them [1]. The procedure should be directed toward the filling of lateral canals as well as filling the main root canals because lateral canals may contribute to lateral root abscesses unless sealed off from the periodontal ligament [1, 2].

Microleakage in canals filled with gutta-percha and sealer leads to ingress and propagation of bacteria with

***Correspondence to: Hala B. Kaka,** Department of Conservative Dentistry, College of Dentistry, Hawler Medical University, next to Hawler Teaching Hospital, 60 m Street, Erbil 44001, Iraq, e-mail: hala.kaka@hmu.edu.krd

[1] Department of Conservative Dentistry, College of Dentistry, Hawler Medical University, Erbil 44001, Iraq

[2] Department of Conservative Dentistry, Endodontics, College of Dentistry, Hawler Medical University, Erbil 44001, Iraq

Open Access. © 2021 Kaka and Salman, published by Sciendo. BY-NC-ND This work is licensed under the Creative Commons Attribution NonCommercial-NoDerivatives 4.0 License.

ASIAN BIOMEDICINE

consequent infection [3]. Although Resilon was introduced with the concept of adhesive bonding to root dentine, it fails to provide a complete seal [4, 5]. Consistently, newly developed bioceramic sealers (EndoSequence BC Sealer and MTA Fillapex) do not fulfill the required chemical and physical properties as ideal root canal filling materials [6].

The inability of currently used root canal filling materials to adhere to the root canal dentine [3], their lack of ability to provide an optimum seal to the root canal system, and the heterogeneous composition of dentine and anatomical complexity of root canals [4] have motivated researchers to develop new filling materials with more promising properties to increase the success rate of endodontic treatment.

Because of their chemical similarity to bones and teeth, calcium phosphate materials have received a lot of attention. They are attractive biomedical materials due to their excellent biocompatibility and the nontoxicity of their chemical components [7, 8]. A mixture of tetracalcium phosphate (TTCP) and dicalcium phosphate anhydrous (DCPA) [9] with a liquid phase like water is a useful cement as a bone substitute material because the ultimate product is hydroxyapatite [10]. Water-soluble biocompatible polymers may be added to the aqueous phase used to dissolve calcium phosphate powder [11, 12]. The incorporation of water-soluble biocompatible polymers into the formulation of calcium phosphate cements may improve their properties such as injectability, cohesion, and setting time [11, 13].

The present study aimed to investigate whether calcium phosphate powder can be used with a styrene–butadiene emulsion polymer as an endodontic injectable filling material.

Methods

Preparation of the material

The powder phase consisted of an equivalent molar ratio of TTCP (Applichem) and DCPA (Sigma-Aldrich), 5% wt bismuth oxide (ChemicalPoint), and 10% wt calcium chloride (Merck). The liquid phase comprised 50% styrene–butadiene rubber emulsion (Sika, Switzerland) in distilled water. To form the cement, one spoon of powder was mixed with 2 drops of liquid on a glass slab using a cement mixing spatula.

The material was subjected to a series of tests, all of which were performed according to the American National Standards Institute–American Dental Association (ANSI/ADA) specification No. 57 [14], except for the injectability test.

Flow

Using a graduated syringe, 0.5 ± 0.05 mL of the mixed experimental material was placed on a glass plate (40 mm × 40 mm, 5 mm thick; 20 ± 2 g). At 180 ± 5 s after the start of mixing, a load of 100 g and a top glass plate (also 40 mm × 40 mm, 5 mm thick, and 20 ± 2 g) were applied centrally on the top of the soft material. The load was removed 10 min after the start of mixing, and the minor and major diameters of the disc of the material formed were measured using digital calipers. If the disc was not uniformly circular or the major and minor diameters did not agree within 1 mm, the test was repeated. Ten readings were obtained from 10 different samples, and the mean of these 10 readings was taken as the flow of the material.

Setting time

The apparatus used comprised an indenter attached to a surveyor and a metal block (8 mm × 20 mm × 10 mm). The indenter weighed 100 ± 0.5 g and had a flat end diameter of 2 ± 0.1 mm. The indenter tip was cylindrical over a distance of at least 5 mm. The metal block and indenter needle were conditioned in an incubator at 37 ± 1 °C with a relative humidity of 95%.

A plaster of paris mold (internal diameter of 10 mm and thickness of 2 mm) was used. The material was mixed for 45 s at 23 ± 2 °C (room temperature) and 50 ± 5% humidity. The plaster of paris mold was fitted on a flat glass plate (microscope slide), then the hole of the plaster of paris mold was filled with the experimental material. After 180 ± 5 s from the start of the mixing process, the microscope slide with the mold containing the material was placed on the metal block and the whole apparatus was placed in an incubator capable of maintaining a temperature at 37 ± 1 °C and humidity of 95 ± 5%. As soon as possible after the specimen was placed in the incubator, an indenter needle was lowered vertically onto the surface of the material and allowed to remain there for 5 s. The indentation was repeated at 2 min intervals until the indenter impression ceased to be visible in the material. After each removal, the indenter tip was cleaned with a nonwoven swab. The net setting time was recorded as the time between the beginning of mixing and the time when the indenter did not make an impression in the material (**Figure 1**). The mean of 10 readings was taken as the setting time of the material.

Dimensional change

We mixed 2 g of the material with 0.02 mL of distilled water. A cylindrical split mold (12 mm height and 6 mm diameter)

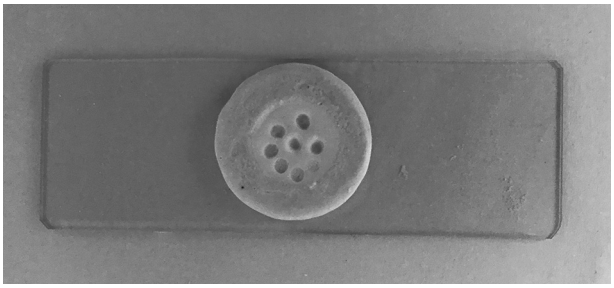

Figure 1. Fitted on a flat glass plate (microscope slide), a plaster of paris mold containing cement material for testing. An indenter needle was lowered vertically onto the surface of the material and allowed to remain there for 5 s. The indentation was repeated at 2 min intervals until the indenter impression ceased to be visible in the material. The net setting time was recorded as the time between the beginning of mixing and the time when the indenter did not make an impression in the material.

made of stainless steel was placed on a plastic sheet backed by a microscopic slide and was filled with the material to slight excess, another glass plate faced with a plastic sheet was pressed on the mold and the assembly held together firmly by the aid of a C-clamp. Five minutes after the start of mixing, the mold with the clamp was transferred to a cabinet that could be maintained at 37 ± 1 °C and relative humidity of not <95%. After the material had been set, the clamp was removed and the ends of the specimen were polished using 360-grit wet sandpaper. Subsequently, the specimen was removed from the mold and the distance between the flat ends was measured to an accuracy of 10 μm using a digital micrometer (Mitutoyo), and the result was recorded. The specimens were stored in distilled water for 30 days in the same cabinet. After this period, the specimen was removed from the distilled water, again the distance between the flat ends was measured to an accuracy of 10 μm using the digital micrometer.

The percentage of the material shrinkage was calculated as follows:

$$\text{Dimensional change following setting\%} = \left[\frac{\text{final length of the specimen} - \text{initial length of the specimen}}{\text{initial length of the specimen}}\right] \times 100$$

Solubility

We mixed 2 g of the material with 0.02 mL of distilled water. A split ring mold (1.5 ± 0.1 mm thickness and 20 mm internal diameter) was placed on a glass plate covered by a plastic sheet and was filled with the material to slight excess. A piece of dental floss was inserted into the material to allow the disc of the material to be suspended in the bottle without touching its walls. Another glass plate faced with a sheet of plastic was pressed on top of the experimental material and the mold was placed in the cabinet for 24 h to allow the material to set. After this, the mold was removed from the cabinet, and the specimen was carefully removed from the mold and the periphery of the specimen was finished to remove any irregularities or flash. The specimens were placed in a desiccator containing silica gel. After 24 h, the specimen was removed from the desiccator and its net weight was determined to the nearest 0.001 g using laboratory balance (model AL204; Mettler Toledo). A dry clean glass bottle was weighed to the nearest 0.001 g and was filled with 30 mL of distilled water and the sample was suspended in the bottle for periods of 24 h, 72 h, and 7 days alternatively.

After each time, the sample was removed carefully from the bottle and washed with 2 mL of distilled water; which drained back to the glass bottle, which was then dried in the desiccator for 24 h, then put into a new bottle (the same sample was used for different time intervals). The bottle containing water was placed in an oven at 110 °C for 24 h and then placed in a desiccator to allow complete cooling for another 24 h, then the bottle was weighed to the nearest 0.001 g using the laboratory balance.

Solubility was measured according to the following equation:

$$\text{Solubility\%} = \left[\frac{\text{final mass of the bottle} - \text{original mass of the bottle}}{\text{disc mass}}\right] \times 100$$

Radiopacity

In the powder constituents, 3 different concentrations of Bi_2O_3 were used: 3%, 5%, and 7%. A stainless steel ring mold of 10 ± 0.01 mm and a height of 1 ± 0.01 mm was placed on a microscopic slide and filled with the experimental material to slight excess, another microscopic slide was pressed over the ring mold to make a uniform 1 mm disc thickness between the 2 microscopic slides. The mold containing the material with 2 glass plates was placed in an incubator at 37 ± 1 °C and 95% humidity to allow the material to set. Subsequently, the disc of the material formed was removed from the mold and attached to paper with glue. Ten discs of the material were prepared for each concentration. The specimens were positioned on the center of an X-ray film along with an aluminum step wedge that was graduated from 1 mm

ASIAN BIOMEDICINE

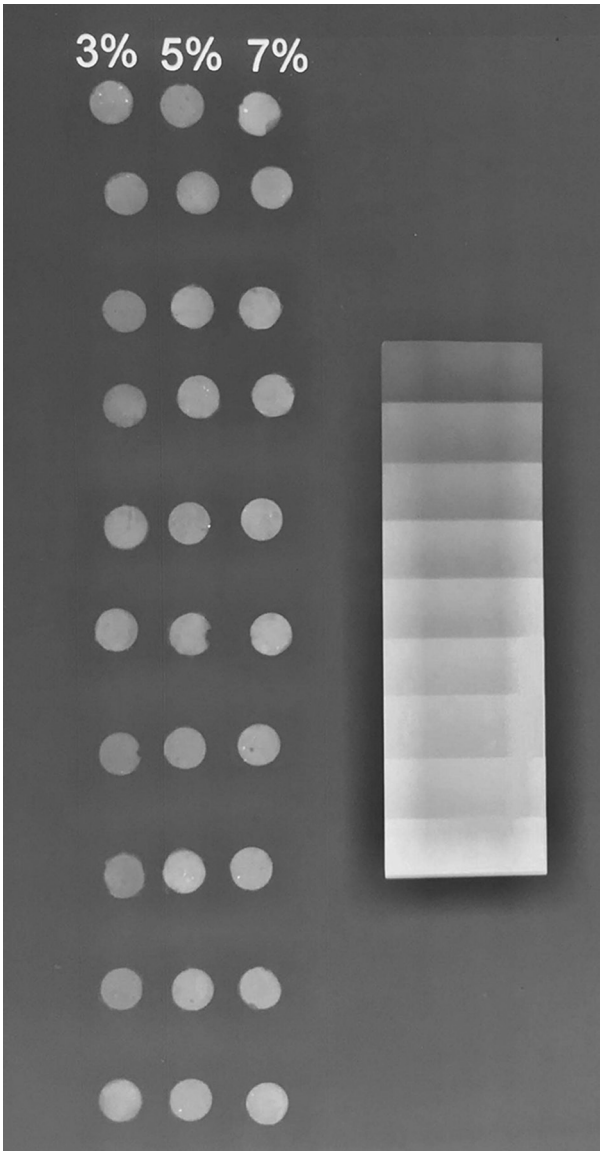

Figure 2. Digital radiograph of samples and an aluminum step wedge that was graduated from 1 mm to 9 mm thick. In the powder constituents, 3 different concentrations of Bi_2O_3 were used: 3%, 5%, and 7%. Discs of the material (10 mm diameter and 1.0 mm height) were prepared for each concentration. An X-ray machine operating at 65 kV, 10 mA was used to produce an image on a phosphorus plate film. The mean of 10 samples of each concentration of Bi_2O_3 was taken as the density of the material and was compared with the density of the aluminum step wedge using an optical densitometer (**Table 1**).

to 9 mm thick. An X-ray machine operating at 65 kV, 10 mA was used to produce an image on a digital phosphorus plate film (**Figure 2**). The focus film distance was 30 cm. A mean of 10 samples of each concentration of Bi_2O_3 was taken as the density of the material and was compared with the density of the aluminum step wedge using an optical densitometer (Heiland Electronic, Germany).

Injectability test

After the experimental material had passed the ANSI/ADA specification No. 57 [14] for flow, setting time, dimensional change, solubility, and radiopacity, the injectability of the experimental material was tested.

Injectability was quantified as the residual mass of the material retained into the syringe after loading with a constant force of 3 kg for a period of 5 s. The device used was composed of 3 parts: a tripod, a track (in which the syringe was placed), and a load mass.

Polymeric disposable syringes of 3 mL and a stainless steel cannula of 55 mm length and a diameter of 1 mm were used. An injectability test was performed after 2 min of mixing. The syringe was weighed when it was empty and the result was recorded, then it was filled with the experimental material using a metal spatula. To minimize air retention inside the syringe, it was tapped many times, then the syringe was weighed again, and subsequently it was filled with 2 mL of the material, and the result was recorded. The filled syringe was placed in the apparatus and the load was applied for 5 s. The syringe was weighed for the third time, and the result was recorded. All syringe mass measurements were made without the cannula.

The injectability was calculated according to the following equation:

$$\text{Injectability\%} = \left[\frac{(M1 - M0) - (M2 - M0)}{(M1 - M0)}\right] \times 100,$$

where:

M0 is the empty syringe mass,
M1 is the filled syringe mass, and
M2 is the syringe mass after injection.

Statistical analyses

Results are generally reported as mean ± standard deviation.

Results

Flow

The experimental material had a flow of 21.2 ± 0.08 mm.

Setting time

The setting time was 19.1 ± 1.3 min (**Figure 1**).

Table 1. The radiopacity of the aluminum step wedge

Aluminum thickness (mm)	Radiopacity†
1	1.12
2	1.00
3	0.95
4	0.87
5	0.81
6	0.72
7	0.66
8	0.60
9	0.37

†Radiopacity of the material sample has no units as values are related to the optical density of the radiograph and are assigned numerical values related to the amount of light that penetrates the film as measured using an optical densitometer, as density of the material sample increases the radiopacity values decrease in an inverse relationship as radiograph density decreases. The ANSI/ADA specification No. 57 [14] is that "an X-ray machine capable of producing radiation at (65 ± 5) kV and 10 Ma shall be used in conjunction with radiographic film of speed group D or E as specified in ISO 3665 to obtain a radiograph of the test specimen and the aluminum step wedge."

Dimension change

Extra samples were prepared because 5 of the samples were destroyed while stored in distilled water after 7 days. Similarly, some of the new samples were found damaged after 30 days of storage in distilled water. In total, there were 7 samples for the dimension test. The percentage of shrinkage of the experimental material was 0.61 ± 0.16%, which is within the scope of ANSI/ADA specification No. 57 [14].

Solubility

The solubility of the experimental material was different at different times; it was maximal after 24 h (6.62 ± 0.58%), but decreased after 72 h (1.09 ± 0.08%) and after 7 days (1.16 ± 0.18%).

Radiopacity

As shown on the radiographic film (**Figure 2**), the mean densities for material samples with any of the 3 concentrations of Bi_2O_3 used were greater than that of the 3 mm step on the aluminum wedge (radiopacity 0.95), as such, they met the ANSI/ADA specification No. 57 [14]. All samples with 5% Bi_2O_3 had densities greater than that of 3 mm step (**Table 1**). However, some samples with 3% Bi_2O_3 apparently had slightly less than the optimum density. The radiopacities were 3% Bi_2O_3: 0.89 ± 0.58 (range 0.65–0.98; median 0.92); 5% Bi_2O_3: 0.84 ± 0.076 (range 0.66–0.95; median 0.88; 7% Bi_2O_3: 0.81 ± 0.18 (range 0.61–0.89; median 0.83).

Injectability

The material showed 93.67 ± 1.80% injectability.

Discussion

Calcium phosphate cements have been used as bone grafts with great success in the last decade [15]. Because they have a chemical composition and a crystalline structure similar to tooth and bone apatite materials [16], they have seen clinical success in many dental and orthopedic applications. However, there is limited clinical use of calcium phosphate cements in applications requiring injection because of their relatively poor injectability [17], this problem can be improved by the addition of various polymers [18–20].

To our knowledge, the present study is the first attempt to use a synthetic rubber styrene–butadiene polymer emulsion with calcium phosphate cements.

The experimental material was prepared and tested to identify its suitability as a root canal filling material. The tests were conducted according to ANSI/ADA Specification No. 57 for endodontic sealing materials [14].

Flow is considered an important characteristic of endodontic filling material because it helps the material to reach the irregularities of the canal walls. At the same time, the high flow may result in apical extrusion, leading to injury of the periapical tissues [21]. An endodontic sealing material is said to have good flowability when a disc is formed according to the procedure mentioned has a diameter of not <20 mm [14]. In this work, the experimental material had a mean diameter of 21.2 mm. The composition, particle size, shear rate, temperature, and time from mixing are the main factors related to the flow characteristics of the material [21].

An acceptable root canal filling material should set sufficiently slowly [2] to provide enough time for the operator to introduce it into the root canal before it sets, although an assurance in the setting reaction is one of the key advantages for the successful clinical results when using a calcium phosphate cement [22]. It is worth noting that the calcium phosphate cement needs moisture during the setting process. Therefore, for setting time measurement, a plaster of paris mold was used and stored at 37 °C and >95% relative humidity for 24 h before use. The experimental material introduced in this work had a mean setting time of 19.1 min. In general, polymers act to

increase setting time, which may be related to the higher viscosity of the polymer-containing paste, which hinders ion diffusion in the matrix [11]. This setting time is consistent with the findings by Fukase et al. [23], who found the setting time of calcium phosphate cement composed of TTCP and DCPA to be approximately 20 min. Komath and Varma [24] had found similar results when the cement powder contained TTCP and DCPD, mixed in an equimolar ratio, and the wetting medium used was distilled water with sodium phosphate (Na_2HPO_4) as the setting accelerator in an optimized concentration. A year later, they introduced a novel formulation of calcium phosphate cement that is fully injectable with a setting time of 20 min [25]. By contrast, Engstrand et al. [26] found that the cement prepared from polymer microparticles with an acidic premixed calcium phosphate cement had a final setting time between 30 min and 35 min measured using the Gilmore needle method. Hirayama et al. [27] found that calcium phosphate cement composed of equimolar amounts of TTCP and DCPA would harden in about 30 min when water was used as a cement liquid. The suggested optimal setting time for calcium phosphate cement is 10–15 min. However, Unosson [28] found that the precipitated hydroxyapatite cement sets slowly and could take up to a few weeks to fully harden, while the brushite cement sets rapidly and hardens within a minute.

The chemical composition, particle size and sintering temperatures of the powders, the liquid phase, and the ratio of liquid-to-powder play crucial roles in determining the setting time of cement materials [29].

In the present study, the experimental material showed good injectability, this can be attributed to the styrene–butadiene emulsion that is present in the liquid phase [11]. However, obtaining a material with good flowability (injectability) with a short setting time is difficult because injectability and rapid setting can be opposing properties as a rapid setting paste may start setting in the syringe, increasing the paste thickness and rigidity, and thus reducing its injectability. Improvements in injectability via powder or liquid modifications may inadvertently result in a slow setting. Therefore, any enhancement in paste injectability should not compromise setting times [30]. In the present study, the material showed good injectability, and this did not affect the setting time. The injectability of the calcium phosphate cement varies inversely with its viscosity, powder-to-liquid ratio, and the time after starting the mixing of liquid and powder. Injectability can be improved by using smaller particle sizes, with shorter and larger diameter cannula, and lower flow rates [31].

The shrinkage of the experimental material was 0.61%, which is within the scope of ANSI/ADA specification No. 57 [14] because any dimensional change of the root filling material may affect the integrity of the bond between the filling and the root dentine [32]. While stored in distilled water for the 30 days, some samples were destroyed, the reason for this may be the initial high solubility of this calcium phosphate cement as shown in the solubility test, and hence the degradation of the material [33]. The solubility is an important property because it determines the dissolution, precipitation, hydrolysis, and phase transformation of calcium phosphate. Moreover, it plays an important role in biological processes that involve the formation and resorption of hard tissues [34]. The solubility of the experimental material was determined at 3 times [33]; at 24 h, it was highest (6.62%), but the solubility decreased to be within the ANSI/ADA specification No. 57 [14] at 3 days and 7 days. In general, all calcium phosphate compounds are more soluble in low pH [34]. Hydroxyapatite is considered as least soluble at neutral to basic pH, while DCPA is considered as the least soluble at acidic pH [28]. The DCPA content may be the cause of the early high solubility of the experimental material used in this study.

The final hydrated product of the calcium phosphate cement is of paramount importance because it determines the solubility of the CPC in vitro and in vivo. The hydrated product depends on the pH of the cement paste [29].

Root canal filling materials should have sufficient radiopacity to detect their presence, deficiency, extent, and apparent condensation. The radiopacity of 1 mm aluminum is approximately equivalent to a 1 mm thickness of dentine [35]. However, the use of a highly radiopaque material might give the impression of a compact root canal obturation despite the presence of imperfections including minor voids in the filling. Conversely, a less radiopaque material might be judged absent in areas where it is present in small amounts [36]. In the present study, all the samples were radiographed at the same time and digital film processing was used. This provides greater standardization of the densities, and it also reduced the time needed for conventional radiograph film processing. The digital imaging technique yields better contrast, visualization, sharp images, as well as being relatively quick [37]. It is well known that the radiopacity of pure aluminum (99.5%) is very close to that of human dentine [38]. Hence, an aluminum step wedge is used to compare the radiopacity of root canal filling material. Calcium phosphate has low radiopacity [29], therefore, a radiopaque agent needs to be incorporated into it, and should only impart the necessary radiopacity. The material should be inert, nontoxic, and be added in minimal amounts. To achieve the latter point, it is important to use elements that have high relative atomic mass [39, 40]. Bi_2O_3 is a common radiopaque agent and is incorporated into root filling material [41] because of its high radiopacity [42]. In the present study, 3 different concentrations of Bi_2O_3 were used in the powder: 3%, 5%, and 7%. The

material with 5% Bi_2O_3 produced a mean density that complied with ANSI/ADA specification No. 57 [14]; however, some samples with 3% Bi_2O_3 had less than the optimum density. This may be attributed to a "radiographic inhomogeneity" phenomenon, which is due to material microdensity variations, in which there is a small, but significant variation of radiopacity within and between successive specimens of the same geometry [38]. By contrast, samples with 7% Bi_2O_3 showed a density approximately equivalent to that of the 5 mm step on the aluminum wedge, although this may give a false impression of a perfect root canal obturation. Moreover, a study showed that the Bi_2O_3 used in mineral trioxide aggregates is not biocompatible [43] and it has a deleterious effect on the physical properties of mineral trioxide aggregates [44].

Conclusion

Calcium phosphate powder with styrene–butadiene emulsion met the basic requirements for a root canal filling material. Further research to study the biocompatibility and other important properties of this newly developed material is warranted.

Authors contributions. Both authors contributed to the design of the work; HBK contributed to the acquisition, analysis, and interpretation of data, and drafted the manuscript. Both authors revised the manuscript critically for important intellectual content; approved the final version to be published; and agree to be accountable for all aspects of the work.

Acknowledgments. We thank Dr. Hikmat Ali Mohamad; Assistant Professor in the Department of Chemistry, College of Education at the University of Salahaddin, for his advice and consultation about the chemical materials and their reactions. The authors did not receive any specific grant for this research from any funding agency in the public, commercial, or not-for-profit sectors.

Conflicts of interest statement. Both authors have completed and submitted an International Committee of Medical Journal Editors Form for Disclosure of Potential Conflicts of Interest. Neither author has any potential or actual conflict of interest to disclose in relation to the present study.

Data sharing statement. Statistical summaries of all data are presented to support the present work. The detailed raw data sets are available from the corresponding author for noncommercial purposes on reasonable request.

References

[1] Schilder H. Filling root canals in three dimensions. 1967. J Endod. 2006; 32:281–90.
[2] Chandra BS, Gopikrishna V, editors. Grossman's Endodontic Practice. 13th ed. New Delhi: Wolters Kluwer Health, Lippincott, Williams and Wilkins; 2014.
[3] Bansode PV, Pathak SD, Wavdhane MB, Kale D. Obturating materials present and past: a review. IOSR J Dent Med Sci. 2018; 17:27–33.
[4] Shrestha D, Wei X, Wu W-C, Ling J-Q. Resilon: a methacrylate resin-based obturation system. J Dent Sci. 2010; 5:47–52.
[5] Al-Maswary AA, Alhadainy HA, Al-Maweri SA. Coronal microleakage of the Resilon and gutta-percha obturation materials with Epiphany SE sealer: an in-vitro study. J Clin Diagn Res. 2016; 10:ZC39–42.
[6] Lee JK, Kwak SW, Ha J-H, Lee W, Kim H-C. Physicochemical properties of epoxy resin-based and bioceramic-based root canal sealers. Bioinorg Chem Appl. 2017; 2017:2582849. doi: 10.1155/2017/2582849
[7] Chen F, Liu C, Mao Y. Bismuth-doped injectable calcium phosphate cement with improved radiopacity and potent antimicrobial activity for root canal filling. Acta Biomater. 2010; 6:3199–207.
[8] Al-Sanabani JS, Madfa AA, Al-Sanabani FA. Application of calcium phosphate materials in dentistry. Int J Biomater. 2013; 2013:876132. doi: 10.1155/2013/876132
[9] Thepsuwan W, Monmaturapoj N. Preparation and characterization of calcium phosphate cement. Int J Chem Nucl Metall Mater Eng. 2014; 8:1236–39.
[10] Takagi S, Chow LC, Ishikawa K. Formation of hydroxyapatite in new calcium phosphate cements. Biomaterials. 1998; 19:1593–9.
[11] Perez RA, Kim H-W, Ginebra M-P. Polymeric additives to enhance the functional properties of calcium phosphate cements. J Tissue Eng. 2012; 3:2041731412439555. doi: 10.1177/2041731412439555
[12] An J, Wolke JG, Jansen JA, Leeuwenburgh SC. Influence of polymeric additives on the cohesion and mechanical properties of calcium phosphate cements. J Mater Sci Mater Med. 2016; 27:58. doi: 10.1007/s10856-016-5665-x
[13] Moseke C, Gbureck U. Tetracalcium phosphate: synthesis, properties and biomedical applications. Acta Biomater. 2010; 6:3815–23.
[14] Council on Scientific of the American Dental Association. American National Standard/American Dental Association Specification 57 Endodontic sealing materials. Chicago:ADA; 2000. Reaffirmed by ANSI: December 2006.
[15] Vasconcellos LA, dos Santos LA. Calcium phosphate cement scaffolds with PLGA fibers. Mater Sci Eng C Mater Biol Appl. 2013; 33:1032–40.
[16] Atmeh AR, Chong EZ, Richard G, Festy F, Watson TF. Dentin-cement interfacial interaction: calcium silicates and polyalkenoates. J Dent Res. 2012; 91:454–9.
[17] O'Neill R, McCarthy HO, Montufar EB, Ginebra M-P, Wilson DI, Lennon A, Dunne N. Critical review: injectability of calcium phosphate pastes and cements. Acta Biomater. 2017; 50:1–19.
[18] Jang J-H, Shin S, Kim H-J, Jeong J, Jin H-E, Desai MS, et al. Improvement of physical properties of calcium phosphate cement by elastin-like polypeptide supplementation. 2018; 8:5216. doi: 10.1038/s41598-018-23577-y
[19] Hablee S, Sopyan I, Mel M, Salleh HM, Rahman MM, editors. Antiwashout behavior of calcium phosphate cement incorporated

with Poly(ethylene glycol). IOP Conf Ser Mater Sci Eng. 2018; 290:012075. doi:10.1088/1757-899X/290/1/012075

[20] Zhang J, Liu W, Gauthier O, Sourice S, Pilet P, Rethore G, et al. A simple and effective approach to prepare injectable macroporous calcium phosphate cement for bone repair: syringe-foaming using a viscous hydrophilic polymeric solution. Acta Biomater. 2016; 31:326–38.

[21] Zhou H-M, Shen Y, Zheng W, Li L, Zheng Y-F, Haapasalo M. Physical properties of 5 root canal sealers. J Endod. 2013; 39:1281–6.

[22] Ishikawa K. Calcium phosphate cement. In: Ben-Nissan B, editor. Advances in calcium phosphate biomaterials. Heidelberg: Springer; 2014, p. 199–227. (Wang M, series editor. Springer Series Biomater Sci Eng., vol. 2). doi.org/10.1007/978-3-642-53980-0_7

[23] Fukase Y, Eanes ED, Takagi S, Chow LC, Brown WE. Setting reactions and compressive strengths of calcium phosphate cements. J Dent Res. 1990; 69:1852–6.

[24] Komath M, Varma HK. Development of a fully injectable calcium phosphate cement for orthopedic and dental applications. Bull Mater Sci. 2003; 26:415–22.

[25] Komath M, Varma H. Fully injectable calcium phosphate cement—a promise to dentistry. Indian J Dent Res. 2004; 15:89–95.

[26] Engstrand J, Persson C, Engqvist H. Influence of polymer addition on the mechanical properties of a premixed calcium phosphate cement. Biomatter. 2013; 3:e27249. doi: 10.4161/biom.27249

[27] Hirayama S, Takagi S, Markovic M, Chow LC. Properties of calcium phosphate cements with different tetracalcium phosphate and dicalcium phosphate anhydrous molar ratios. J Res Natl Inst Stand Technol. 2008; 113:311–20.

[28] Unosson J. Physical properties of acidic calcium phosphate cements. [Thesis]. Lund: Uppsala Univ.; 2014. Digital Comprehensive Summaries of Uppsala Dissertations from the Faculty of Science and Technology 1195. 73 pp. Uppsala: Acta Universitatis Upsaliensis

[29] Sugawara A, Asaoka K, Ding S-J. Calcium phosphate-based cements: clinical needs and recent progress. J Mater Chem B. 2013; 1:1081–9.

[30] Burguera EF, Xu HHK, Weir MD. Injectable and rapid-setting calcium phosphate bone cement with dicalcium phosphate dihydrate. J Biomed Mater Res B Appl Biomater. 2006; 77:126–34.

[31] Dorozhkin SV. Self-setting calcium orthophosphate formulations. J Funct Biomater. 2013; 4:209–311.

[32] Kazemi RB, Safavi KE, Spingberg LSW. Dimensional changes of endodontic sealers. Oral Surg Oral Med Oral Pathol. 1993; 76:766–71.

[33] Schäfer E, Zandbiglari T. Solubility of root-canal sealers in water and artificial saliva. Int Endodont J. 2003; 36:660–9.

[34] Chow LC. Solubility of Calcium Phosphates. In: Chow LC, Eanes ED, editors. Octacalcium phosphate. Basel: Karger; 2001, p. 94–111. (Lussi A, Buzalaf M, series editors. Monogr Oral Sci., Vol. 18).

[35] Thu M, Kim J-W, Park S-H, Cho K-M. Physical properties of a new resin-based root canal sealer in comparison with AH Plus Jet. J Dent Rehabil Appl Sci. 2017; 33:80–87.

[36] Bodanezi A, Pereira AL, Munhoz EA, Bernardineli N, Moraes IGd, d CMB. Digital radiopacity measurement of different resin- and zinc oxide-based root canal sealers. Rev Odonto Ciênc. 2010; 25:74–7.

[37] Muhamedagic B, Muhamedagic L. Digital radiography versus conventional radiography in dentistry. Acta Inform Med. 2009; 17:85–9.

[38] Watts D, McCabe J. Aluminium radiopacity standards for dentistry: an international survey. J Dent. 1999; 27:73–8.

[39] Bowen R, Cleek G. X-ray-opaque reinforcing fillers for composite materials. J Dent Res. 1969; 48:79–82.

[40] Camilleri J. Evaluation of the physical properties of an endodontic Portland cement incorporating alternative radiopacifiers used as root-end filling material. Int Endod J. 2010; 43:231–40.

[41] Guerreiro-Tanomaru JM, Duarte MAH, Gonçalves M, Tanomaru-Filho M. Radiopacity evaluation of root canal sealers containing calcium hydroxide and MTA. Braz Oral Res. 2009; 23:119–23.

[42] Min KS, Chang HS, Bae JM, Park SH, Hong CU, Kim EC. The induction of heme oxygenase-1 modulates bismuth oxide-induced cytotoxicity in human dental pulp cells. J Endod. 2007; 33:1342–6.

[43] Camilleri J, Montesin FE, Papaioannou S, McDonald F, Pitt Ford TR. Biocompatibility of two commercial forms of mineral trioxide aggregate. Int Endod J. 2004; 37:699–704.

[44] Camilleri J. The physical properties of accelerated Portland cement for endodontic use. Int Endodo J. 2008; 41:151–7.