

**A comparison of the bone-like apatite
formation potency between Hydroxyapatite
and β -Tricalcium phosphate
in Glass ionomer dental luting cement.**

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and β -Tricalcium phosphate
in Glass ionomer dental luting cement.

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감사의 글

논문이 완성되기까지 자상하게 지도해 주시고 관심을 아끼지 않고 격려해주신 최형준 지도 교수님께 감사드리며, 실험설계부터 논문이 끝나기까지 항상 조언을 해주신 이용근 교수님과 애정어린 관심과 격려를 아끼지 않으신 소아치과의 모든 교수님께 감사드리며, 실험실의 모든 것 하나하나를 가르쳐주신 치과재료학 교실 여러분께도 깊은 감사의 말씀드립니다.

모든 실험과정과 논문쓰기까지 사랑과 용기로 감싸주신 병원 가족 모두에게 감사드리며, 실험실을 항상 친구처럼 같이해준 박상욱 선생님, 채문희 선생님, 논문의 완성에 좋은 의견을 제시해준 이정진 선생님, 홍은경 선생님께도 감사드립니다.

3년을 넘게 항상 옆에서 나의 든든한 후원자가 되어주고, 격려해주며 슬픔과 기쁨을 함께해준 아내 김선희에게 감사의 뜻을 전합니다.

끝으로 지난 긴 시간동안 논문이 완성되기까지 항상 제 든든한 버팀목이 되어주셨던 아버님, 어머님, 장모님, 형, 형수님, 처제께 제 지난 날의 사랑을 담아 이 논문의 영광을 바칩니다.

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Abstract

A comparison of the bone-like apatite formation potency between Hydroxyapatite and β -Tricalcium phosphate in Glass ionomer dental luting cement

An increase in bonding strength due to the formation of bone-like apatite when of glass ionomer dental luting cement(GIC) and hydroxyapatite(HA) is bonded to human teeth has been reported in previous studies. However, the amount of bone-like apatite formed was rather low due to the low solubility of HA. The purpose of this study was to determine whether a mixture of highly soluble β -tricalcium phosphate(β -TCP) and HA added to glass ionomer cements increases bone-like apatite formation in the tooth-cement interface. Also, this study aims to determine whether the bioactivity of biphasic calcium phosphate(BCP: a mixture of HA and TCP), which is known to have both the physical properties of HA and the high solubility of β -TCP, is affected when used in dental luting cements.

Considering the fact that 15% HA-GIC(RelyXTM luting, 3M ESPE, USA) had the best physical properties in previous studies, the same composition of β -TCP and HA was used. In addition, an 85/15 mixture

BCP of HA and β -TCP, which has been reported to have the greatest osteoinductive potency(HUIPIN YOAN et al., 2002) was used, Total of four groups: the control group using pure glass ionomer, and the 3 test groups using 15% HA-GIC, 15% BCP-GIC, 15% TCP containing 85/15 mixture of HA and β -TCP, respectively, were established. The necessary requirement to be used as a dental luting cement is the film thickness, setting time, and compressive strength. Therefore, the experiments were conducted with an emphasis on these three factors, following ISO standard 9917-1:2003(E) regulations. Five samples from each group were bonded to natural teeth and stored in simulated body fluid(SBF) at 36.5°C. After four weeks, the bonding strength was measured, and the luting surface was analyzed using a scanning electronic microscope(SEM).

No significant differences in film thickness were observed between the 15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC groups, which were thinner than that of pure GIC. The setting time of the pure GIC group was shorter than that of the three test groups, but no significant inter-group differences were observed. Compressive strength was proportional to the amount of HA, with 15% HA-GIC having the greatest value, followed by 15% BCP-GIC, 15% TCP-GIC, and GIC, in

decreasing order. 15% TCP-GIC and 15% BCP-GIC had the highest bonding strengths, and a SEM analysis of cross sections revealed that bone-like apatite formation was proportional to the amount of β -TCP.

Improvements in physical properties were observed in the HA, BCP, TCP-GIC groups, and in particular, bonding strength - which was the most important aspect of interest in this study - showed remarkable improvements. The higher bonding strengths of 15% TCP-GIC group compared to 15% HA-GIC group may be explained by the high solubility of β -TCP, which facilitates bone-like apatite formation in the tooth-cement interface. However, further long-term studies involving experiments with various compositions of HA and β -TCP will pave the way for advancements in dental cements.

Keywords: Glass ionomer cement, hydroxyapatite, BCP, β -TCP, Bone like apatite

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I. Introduction

Dental luting cement is a medium to bond the restorative material to the teeth. It has been studied for improvement of bonding strength with the human teeth. Glass ionomer cement was developed(Wilson& Kent, 1972) and is being widely used clinically nowadays. Glass ionomer cement bonds with tooth not only mechanically just like other cements, but also chemically(Erickson& Glasspoole, 1994). However, chemical bonding was almost insignificant in clinical application of glass ionomer cement because of the weak nature of chemical bonding and the majority of bonding depends on mechanical retention, just like

other dental luting cements. This weakness could lead to secondary dental caries or restoration exclusion due to bacterial invasion or microleakage(Pachuta& Meiers, 1995). In addition, restorations could be fallen out easily when insufficient amount of tooth material is available because bonding does not occur directly between the tooth and the cement. Therefore, not only the physical property of the dental luting cement itself, but its changes at the interface with the teeth are very important. To overcome these shortcomings, for instance, various fillers had been added to glass ionomer dental luting cement such as silver-cermets, stainless steel powder, carbon and alumino-silicate fibers, and hydroxyapatite(Kawano, 2001; Xu, 2000). In the previous study, bone-like material forming ability by mixing hydroxyapatite with several bone cements was reported in a protein-free acellular simulated body fluid with ion concentrations nearly equal to those of the human blood plasma. Accordingly, hydroxyapatite has been of considerable interest as a biomaterial for implants and bone augmentation since its chemical composition is close to that of bone and tooth(M. Okumura, 1997). According to a previous study, 15% hydroxyapatite, when added to the glass ionomer cement, increases the bonding strength with human teeth by forming bone-like apatite(Y.

Sangil, 2004). However, due to low solubility of hydroxyapatite, only limited amount of bone-like apatite formation was observed under SEM. In this study, we expected to increase the formation of bone-like apatite by using 15% β -tricalcium phosphate, the same apatite system as hydroxyapatite but with higher solubility and smaller size of particle. But, Pure β -tricalcium phosphate has poor physical property compared with hydroxyapatite. Therefore, 15% biphasic calcium phosphate, a mixture of pure hydroxyapatite and pure β -tricalcium phosphate was used in this study. Biphasic calcium phosphate was prepared as 85/15 mixture of hydroxyapatite and β -tricalcium phosphate, the ratio known to have the highest osteoinductive potential(Huipin Yuan et al, 2002).

The aim of this study was to enhance the bonding strength of glass ionomer cement at the interspaces with teeth, by utilizing bioactivities of hydroxyapatite and high solubility of β -tricalcium phosphate, which result in increased apposition of bone-like apatite at the interspaces with teeth. This will provide a good foundation for the wide application of the bioactivity of apatite in other fields of the dentistry.

II. Materials and Methods

1. Materials

Commercially available glass ionomer cement(GIC), hydroxyapatite(HA) and β -tricalcium phosphate(β -TCP) were purchased to prepare the composite of glass ionomer cement and HA or β -TCP. RelyX™ GIC (3M/ESPE, USA) was selected in this study. Fluka(Sigma-Aldrich Inc., USA) was selected in this study as β -TCP. It's molecular fomula is $\text{Ca}_3(\text{PO}_4)_2$ and molecular weight is 310.2. Calcium phosphate tribasic(Sigma-Aldrich Inc., USA) was selected as HA. It's molecular fomula is $\text{Ca}_5(\text{OH})(\text{PO}_4)_3$ and molecular weight is 502.3. To provide an environment similar to intraoral, test specimens were maintained at 36.5°C similar to body temperature in simulated body fluid(SBF) which has a composition as saliva.



Fig. 1. RelyX™ GIC (3M/ESPE, USA).



Fig. 2. β -tricalcium phosphate.



Fig. 3. Calcium phosphate tribasic(HA).

Table 1. The composition of simulates body fluid in this study

List	Material	1L (g)
1	NaCl	7.996
2	NaHCO ₃	0.35
3	KCl	0.224
4	K ₂ HPO ₄ .3H ₂ O	0.174
5	MgCl ₂ .6H ₂ O	0.305
6	1M-HCl	40 ml
7	CaCl ₂	0.278
8	Na ₂ SO ₄	0.071
9	NH ₂ C(CH ₂ OH) ₃	6.057

2. Methods

1) Preparation of sample

According to previous studies, where 15% HA-RelyXTM GIC was shown to have most improved physical property, we limited the concentration of apatite-GIC to 15%(Y. Sangil, 2004).

15% concentration of HA, β -TCP, BCP(85/15 HA and β -TCP) were replaced for glass powder, keeping the water/powder ratio of 2. In other words, the subtracted powder(GIC) was replaced with the same amount of HA, β -TCP, BCP. For equal distribution of the mixed HA, Sonicator (SH-2100, Saehan, Korea) was used for sonication. Air bubble was completely removed in a vacuum oven below 50°C.

Table 2. Sample identification of GIC in this study

Sample I.D	GIC	wt% of calcium phosphate
Control	RelyX TM	0
HA-GIC	RelyX TM	15% HA
BCP-GIC	RelyX TM	15% BCP
TCP-GIC	RelyX TM	15% TCP

2) Film thickness

After GIC, 15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC were inserted between two optically flat, square or circular glass plates with a contact surface area of 200 mm², a force of 150 N was loaded vertically for 10 min. Film thickness was the difference in thickness of the plates with and without the cement film.

3) Setting time

GIC, mixed 15% apatite-GICs were placed in the cabinet(37°C, 90% relative humidity) right after it was mixed. Ninety seconds later, an indenter was lowered vertically onto the surface of the cement and remained there for five seconds. This procedure was repeated at ten seconds intervals until the needle(indentor) failed to make a complete circular indentation in the cement. The setting time was recorded as the time elapsed between the time at the end of mixture to the time when the needle failed to make a complete circular indentation.

4) Compressive strength

Mixed 15% apatite-GICs and GIC were preserved in the cabinet(37°C, 30% relative humidity) for 24 hr after mixture. The specimens were loaded by a mechanical tester, which is capable of loading at the rate of 50 N/min. The applied load was recorded when the specimen fractured and the compressive strength(C) was calculated in MPa using the following formula.

$$C = 4p / \pi d^2$$

p = the maximum load applied (N)

d = the measured diameter of the specimen (mm)

5) Bonding strength

Five specimens were fabricated for each group. First, twenty human second molars were prepared to expose the dentin. Four groups of GIC and apatite-GICs, prepared as mentioned above, were bonded to the four groups of teeth with exposed dentin. The polyethylene tube (diameter - 4mm) was used to bond apatite-GICs to the dentin. They were immersed in simulated body fluid (SBF) at 36.5°C for four weeks. Afterwards, the bonding strength of four groups was measured using testing machine (Instron, UK) and compared to the control group.



Fig. 4. Samples for the measurement of bonding strength.

6) Surface observation

The surface and cross-section of the specimens were observed using SEM (S 2000, Hitachi, Japan).

7) Statistical evaluation

The statistical significant differences were analyzed by Mann-Whitney U test, one-way ANOVA, Duncan's multiple range test with statistical software (SAS version 8.1). One-way ANOVA determines if a statistically significant difference exists between each group, and if the P value was less than or equal to 0.05 it was considered as statistically different. Duncan's Multiple Range Test means that the same letters are not statistically different. Data were summarized as median and range.

III. Result

1. Film thickness

15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC had thinner film thickness compared with the one in pure GIC (Table 3, Fig. 5). However, there was no statistically significant difference among the three groups of apatite-GICs ($P>0.05$). All the samples containing were satisfied the requirement of film thickness of ISO standard 9917-1:2003(E) regulations (maximum film thickness : $25 \mu\text{m}$).

Table 3. Film thickness of GIC-apatites

	Film thickness(mm)			
	Control	TCP-GIC	HA-GIC	BCP-GIC
1	0.020	0.013	0.011	0.012
2	0.021	0.011	0.012	0.011
3	0.025	0.013	0.010	0.011
4	0.020	0.012	0.011	0.013
5	0.021	0.011	0.012	0.011
Average	0.020	0.010	0.010	0.010

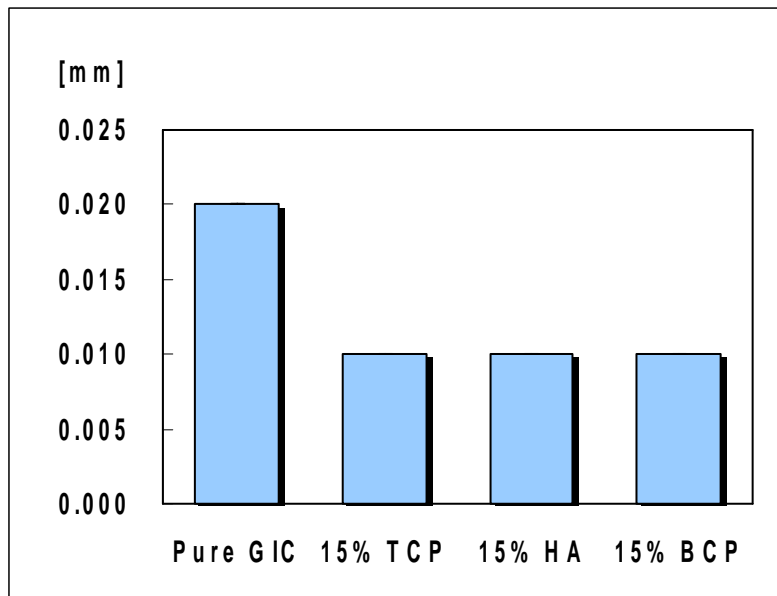


Fig. 5. Film thickness of GIC-apatites.

2. Setting time

15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC had longer setting time compared with the control group (Table 4, Fig. 6). However, there was no significant difference among the three groups ($P>0.05$), and their setting time was within the range of 3 to 8 minutes as indicated in the ISO standard 9917-1:2003(E) regulations.

Table 4. Setting time of GIC-apatites

	Setting time(sec)			
	Control	TCP-GIC	HA-GIC	BCP-GIC
1	225	303	301	300
2	211	292	305	306
3	235	307	299	303
Average	223.67	300.67	301.67	303

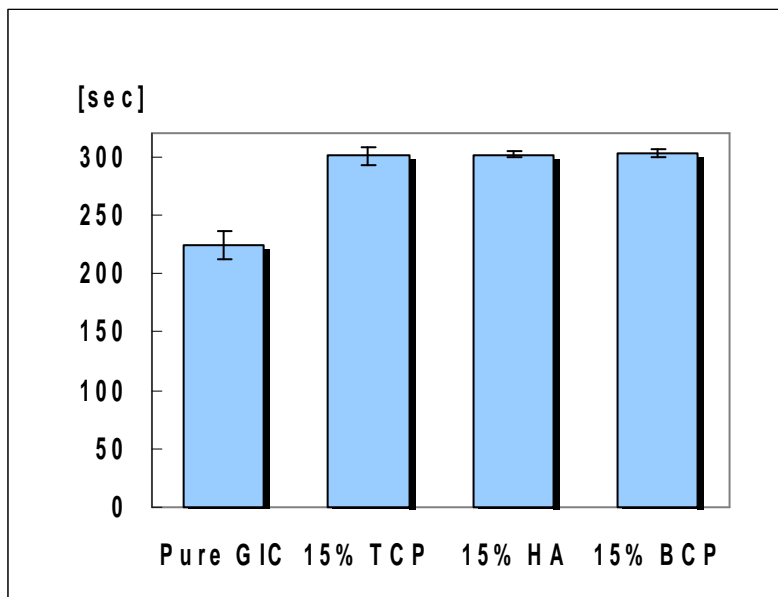


Fig. 6. Setting time of GIC-apatites.

3. Compressive strength

The more amount of HA component contained, the higher the compressive strength observed. In result, the compressive strength was greatest in 15% HA-GIC, and it decreased in order of 15% BCP-GIC, 15% TCP-GIC, and GIC (Table 5, Fig. 7). All the samples containing were satisfied the requirement of compressive strength of ISO standard 9917-1:2003(E) regulations(minimum compressive strength : 50Mpa) .

Table 5. Compressive strength of GIC-apatites

	Compressive strength (MPa)			
	Control	TCP-GIC	HA-GIC	BCP-GIC
1	90.01	128.01	170.01	165.01
2	87.02	127.75	168.05	163.04
3	88.03	130.61	165.35	161.75
4	90.10	131.11	172.11	163.75
5	93.11	129.30	173.65	159.08
Average	89.65	129.36	169.83	162.53

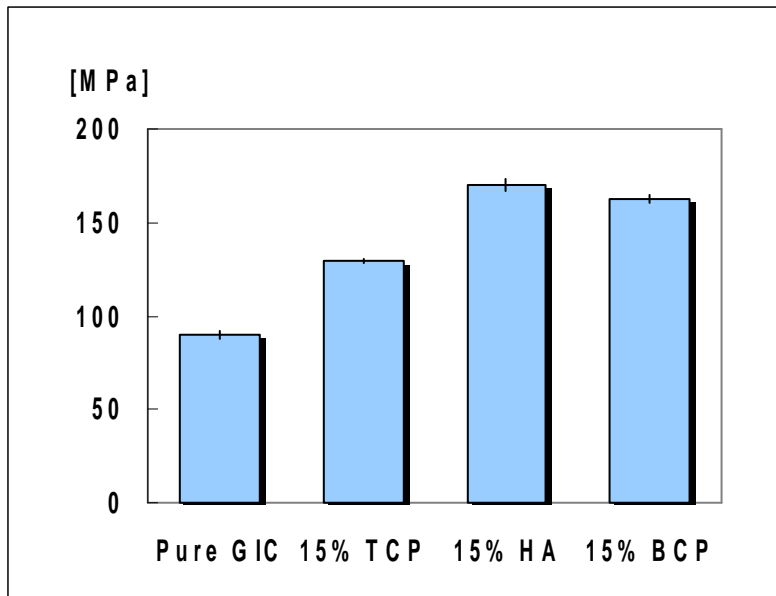


Fig. 7. Compressive strength of GIC-apatites.

4. Bonding strength

Bonding strength was greatest in 15% TCP-GIC, the group with highest amount of β -TCP involved (Table 6, Fig. 8). 15% HA-GIC and 15% BCP-GIC had subsequent bonding strength, and there were no statistically significant difference between them ($P>0.05$). The bonding strength of GIC group cited Lee's 2006 results which showed the lowest measurement(Lee, 2006).

Table 6. Bonding strength of of GIC-apatites

	Bonding strength (mN/mm²)			
	Control	TCP-GIC	HA-GIC	BCP-GIC
1	790.50	1555.1	1327.5	1507.3
2	584.90	1580.3	1502.5	1515.5
3	786.90	1532.1	1455.3	1499.7
4	803.00	1510.7	1391.4	1450.9
5	790.20	1521.6	1531.4	1478.9
Average	751.10	1539.96	1441.62	1490.46

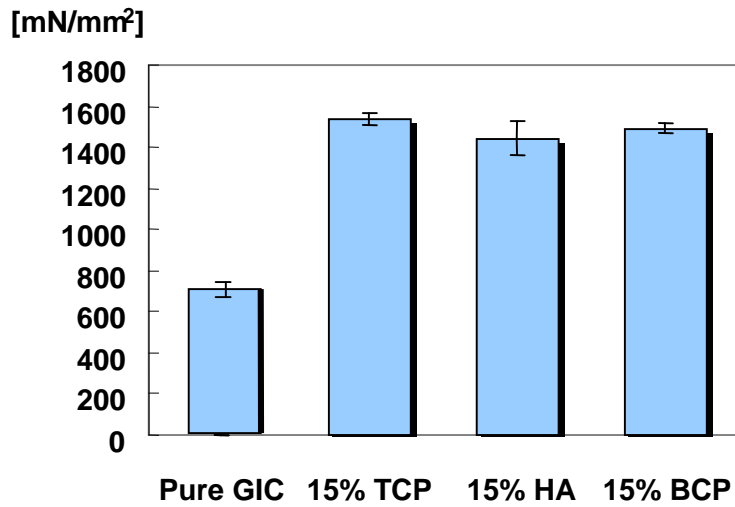
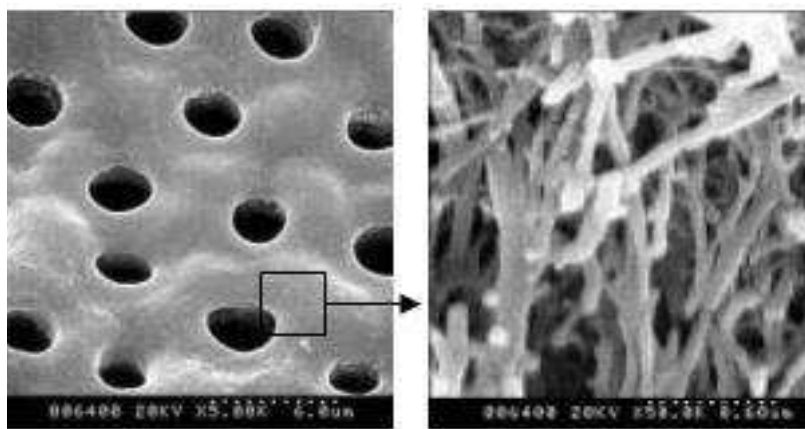


Fig. 8. Bonding strength of GIC-apatites.

5. Microstructure

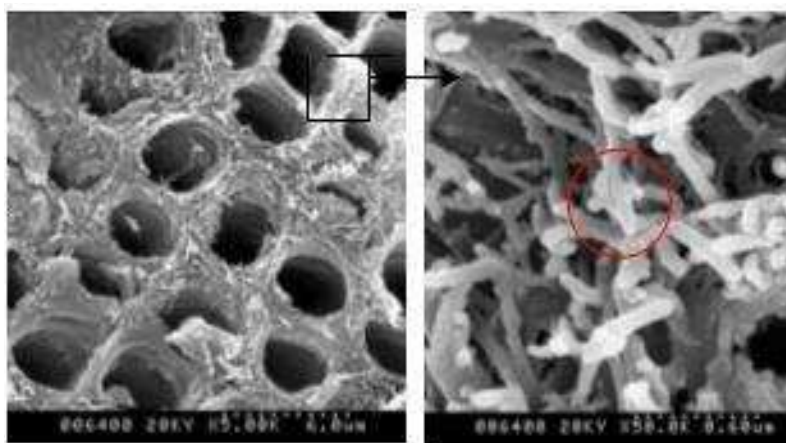
On the SEM image of the sectioned dentin surface, bone-like apatite crystal was deposited around the intertubular dentin in all three groups of 15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC (Fig. 10, 11, 12). Among them, 15% TCP-GIC group showed the highest apatite formation on the dentin surface, while there was no sign of bone-like apatite formation in the GIC group (Fig. 9).



(a)(×5000)

(b)(×20000)

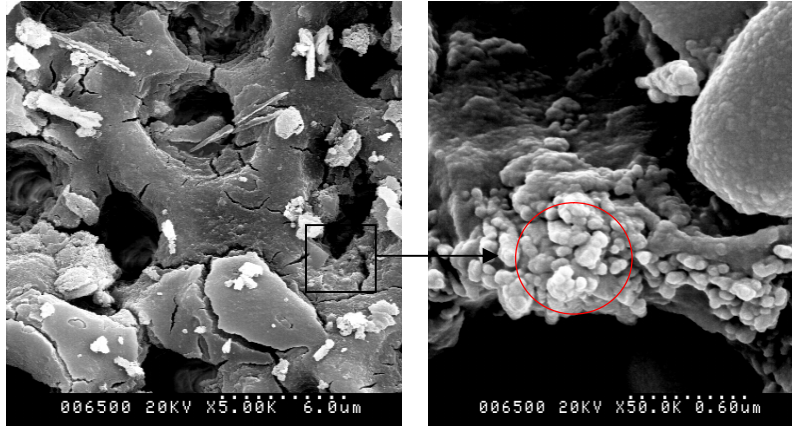
Fig. 9. Dentin surface of control group(pure GIC) under SEM.



(a)(×5000)

(b)(×20000)

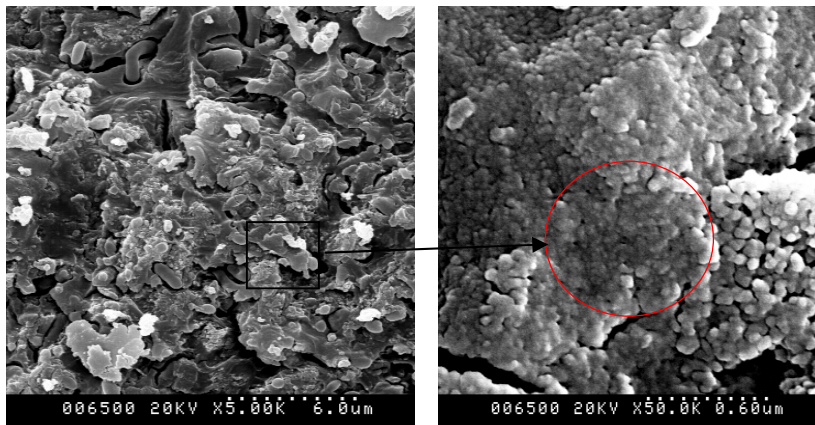
Fig. 10. Dentin surface of 15% HA-GIC group under SEM.



(a) (×5000)

(b) (×20000)

Fig. 11. Dentin surface of 15% BCP-GIC group under SEM.



(a) (×5000)

(b) (×20000)

Fig. 12. Dentin surface of 15% β TCP-GIC group under SEM.

IV. Discussion

GIC, due to its fluoride releasing and chemical bonding with teeth advantages, is a widely used dental luting cement. However, the weak strength and tooth bonding strength compared to other contemporary dental luting cement causes material fracture and secondary dental caries between the tooth interface. In order to solve such problems many studies have been done.

Apatite is one of many calcium phosphate minerals and a main component of human bone and teeth. Enamel and human bones are consisted of 97% and 65% of apatite respectively. Apatite of bone and teeth are mainly composed of HA with extremely small amounts of calcium ions, metal ions and fluoride ions, thus being called a biological apatite. We in this study, tried to improve the physical properties of GIC by using HA, β -TCP, BCP(a mixture of HA and β -TCP) which are known to be highly bioactive(H. Yuan, 2002).

RelyX™ GIC (3M/ESPE, USA), used in this study, is resin reinforced glass ionomer cement, but it has not light curing reaction. It has two hardening reactions, acid-base reaction between fluoroaluminosilicate glass and polycarboxyl acid, and polymerization

reaction between methacrylate group of polymer and HEMA(2-hydroxyethylmethacrylate). Therefore, to estimate biocompatibility as dental luting cement according to ISO standard 9917-1:2003(E) regulations, film thickness, setting time, compressive strength, bonding strength were measured.

15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC(experimental groups) showed thinner film thickness values compared to pure GIC(control groups). Since the average particle size of GIC is 13~19 μm (<45 μm) and 5~10 μm in apatites, we could predict that small apatite particles move into the inter matrix empty spaces, thus explaining the reduction of film thickness.

Experimental groups had longer setting time compared to control groups. Such a result is probably due to the easy bonding between polyacrylic acid and calcium, aluminum, sodium ions of GIC powder and the strong bond between calcium and phosphate ions of apatite and hydroxyl groups resulting in delayed reaction with polyacrylic acids and apatites(A. Hideki, 2002).

Experimental groups showed higher compressive strength value compared to control groups. Such values were higher in experimental groups with apatite (15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC). Apatites of high calcium concentration seemed to accelerated ionic

bonding with polyacrylic acid(M.E. Lucas, 2003). Packed relatively small particle sized apatite in GIC particles seemed to stabilize its structure. Higher compressive strength of HA compared to β -TCP explains the highest strength value of 15% HA-GIC groups.

Experimental groups showed stronger bond strength compared to control groups. 15% TCP-GIC containing β -TCP showed the strongest value. The formation of bond-like apatite during the ionic change between the interface of calcium phosphate and SBF explains the increased bonding strength with tooth structures, such results could be proved in SEM examinations(L.L. Hench, 1996).

The strongest bonding strength of 15% TCP-GIC could be explained by the active ionic change in SBF due to the higher solubility of β -TCP compared to HA resulting in abundant bone-like apatite formation(T. Kokubo, 1991).

Improvements of mechanical properties of the material itself and tooth bonding strength by mixing apatite to GIC were discussed.

Thus, if studies on environmental factors to greatly improve bioactivity of apatite and ideal mixing ratio and methods could be done, improvements in longevity and problems regarding secondary dental caries could be solved.

V. Conclusion

In order to investigate changes in the material itself and between the interface with the human tooth, 15 wt % HA, BCP, β -TCP were mixed with GIC. Film thickness, setting time, compressive strength, bonding strength with dentin and the surface of dentin were evaluated. Test specimens were maintained at 36.5 °C in SBF for 4 weeks. The following conclusions were drawn from this investigation .

1. 15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC had thinner film thickness compared to pure GIC. However, there was no statistically significant difference among the three groups of apatite-GICs. The film thickness of the three groups were within the range of ISO standard 9917-1:2003(E) regulations and they did not affect the physical properties.
2. 15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC had longer setting time compared to the control group. However, there was no significant difference among the three groups. Setting time of the three materials were in the range of 3 to 8 minutes as indicated in the ISO standard 9917-1:2003(E) regulations.

3. Higher amounts of HA composition led to higher compressive strengths. The value was greatest in 15% HA-GIC and decreased in order of 15% BCP-GIC, 15% TCP-GIC and GIC.
4. Bonding strength of 15% TCP-GIC, the group with the highest amount of β -TCP was the greatest. 15% BCP-GIC and 15% HA-GIC showed subsequent values, no statistically significant difference between them were present. Bonding strength was lowest in GIC.
5. In the SEM image of the sectioned dentin surface, bone-like apatite crystal was deposited around the intertubular dentin in all three groups of 15% HA-GIC, 15% BCP-GIC, 15% TCP-GIC. Among them , 15% TCP-GIC group showed the highest apatite formation on the dentin surface.

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국문 요약

글래스아이오노머 치과용 시멘트에 대한

하이드록시아파타이트와 제 3 인산칼슘의 골유사 아파타이트

형성능력의 비교

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글래스아이오노머 시멘트와 하이드록시아파타이트를 혼합하여 인간 치아와 접착시켰을 때 골유사아파타이트(bone-like apatite)의 형성으로 인하여 결합력이 증가되었다는 보고가 있었다. 하지만, 하이드록시아파타이트의 낮은 용해도로 인해 골유사아파타이트가 비교적 적게 형성되었다. 이번 연구의 목적은 용해도가 높은 제3 인산칼슘과 하이드록시아파타이트를 글래스아이오노머 시멘트에 혼합하여 치아에 접착하였을 때, 치아와 시멘트 사이의 계면에서 골유사아파타이트의 형성이 증가하는지를 알아보는 것이다. 그리고, 하이드록시아파타이트의 강한 물리적

특성과 제3 인산칼슘의 높은 용해성을 모두 가지는 이상인산칼슘 (biphasic calcium phosphate)의 생체활성능(bioactivity)이 시멘트에 혼합하였을 때도 발현되는지 알아보기 위함이다.

이전 연구에서 글래스아이오노머 시멘트 제품인 RelyX™ (3M/ESPE, USA)에 15 wt%의 하이드록시아파타이트를 혼합하였을 때 가장 높은 물리적 성질을 보인 점을 감안하여, 같은 비율의 제3 인산칼슘과 하이드록시아파타이트를 사용하였다. 그리고, Huipin 등의 연구에 따라 가장 골유도능이 높다고 알려진 하이드록시아파타이트와 제3 인산칼슘이 85 : 15의 비율로 혼합된 이상인산칼슘을 사용하였다. 따라서, 순수한 글래스아이오노머 시멘트는 대조군으로, 15 wt%의 하이드록시아파타이트, 이상인산칼슘, 제3 인산칼슘을 각각 포함하는 글래스아이오노머 시멘트를 실험군으로 설정하였다. 치과용 시멘트의 기본 성질인 피막도, 경화시간, 압축강도는 ISO 9927 기준에 맞게 측정되었다. 혼합된 글래스아이오노머 시멘트를 치아에 접착시킨 후, SBF에서 36.5 °C 수조에 4주간 보관하고 결합강도를 측정하였다. 그리고, 절단면의 표면 형태를 알아보기 위해 전자현미경 사진으로 관찰하였다.

1. 글래스아이오노머 시멘트의 피막도는 실험군이 대조군에 비해 얇게 측정되었다. 하지만, 실험군 간에 통계학적 유의차가 없었다

($P>0.05$).

2. 글래스아이오노머 시멘트의 경화시간은 실험군이 대조군에 비해 길게 측정되었다. 이는 ISO 9917의 기준에 만족하였고, 실험군 간에 통계학적 유의차가 없었다($P>0.05$).
3. 하이드록시아파타이트가 많이 포함되어 있을수록 글래스아이오노머 시멘트의 압축강도가 크게 측정되었다. 즉, 15 wt% 하이드록시아파타이트, 이상인산칼슘, 제3 인산칼슘을 포함한 글래스아이오노머 시멘트, 순수한 글래스아이오노머 시멘트 순으로 압축강도가 크게 측정되었다($P<0.05$).
4. 제3 인산칼슘의 양이 많이 포함되어 있을수록 글래스아이오노머 시멘트의 결합강도가 크게 측정되었다. 즉, 15 wt% 제3 인산칼슘, 이상인산칼슘, 하이드록시아파타이트를 포함한 글래스아이오노머 시멘트, 순수한 글래스아이오노머 시멘트 순으로 결합강도가 크게 측정되었다($P<0.05$).
5. 절단면의 전자현미경 관찰 결과, 위와 같은 순으로 골유사아파타이트 형성이 많이 관찰되었다.

이상의 실험 결과로부터 아파타이트를 첨가한 것은 기존의 글래스아이오노머 시멘트의 물성을 향상시켰을 뿐 아니라, 결합강도 또한 증가하였다. 특히, 용해도가 높은 제3 인산칼슘을 포함한 그룹이 하이드록

시아파타이트를 포함한 그룹보다 치아와의 계면에서 골유시아파타이트의 형성이 많이 이루어졌으며, 이로 인해 결합강도가 더 높았을 것으로 생각된다. 이러한 아파타이트의 생체활성능을 최대한 활용한다면 글래스아이오노머 시멘트 뿐 아니라, 여러 가지 다른 치과용 재료의 발전을 이룰 수 있을 것이다.

핵심되는 말: 글래스아이오노머 시멘트, 하이드록시아파타이트, 이상인산칼슘,

제 3 인산칼슘, 골유사아파타이트