Piezoelectricity in dental materials, a conceivable cause of postrestorative sensitivity

Göran Sjögren, Maud Bergman and Kenneth Johansson
Department of Dental Materials and Technology, University of Umeå, and Laboratory of Electronics, Faculty of Medicine, University of Umeå, Umeå, Sweden

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With the increased use of tooth-colored posterior inlays reports of postrestorative sensitivity have also increased. One of the symptoms the patients complain of is a sharp pain when the inlays are loaded through chewing and biting. Many explanations for the causes of dissimiliar types of postrestorative sensitivity have been offered, but one conceivable explanation that has not hitherto been studied is the direct piezoelectric effect in dental materials. Direct piezoelectric effect means that when certain anisotropic crystals are mechanically loaded, a charge is generated on the surface. The aim of the present study was to examine whether this physical phenomenon occurs in certain materials intended for dental use. Specimens of four different dental ceramics and one indirect composite resin for inlays were mechanically loaded with various forces, and the current was directly recorded. Currents of up to 0.9 μA with a pulse duraton of 0.4 msec were extracted, and it cannot be excluded that the piezoelectric phenomenon and related properties may cause postrestorative sensitivity. This has to be taken into consideration when posterior inlays of the types concerned are used. □ *Dental ceramics; piezoelectricity; postrestorative sensitivity; resin inlays*

Göran Sjögren, Department of Dental Materials and Technology, Faculty of Odontology, University of Umeå, S-901 87 Umeå, Sweden

During the past decade the use of ceramic and resin inlays has increased, and the demand for non-metallic restorative materials has initiated the development of ceramic and resin materials with improved physical and chemical properties, as well as new techniques, such as the CAD/CAM technique.

Following the increased use of tooth-colored restorative materials reports of sensitivity after the placement of posterior inlays have also increased. Thus it has been reported that up to a third of teeth provided with tooth-colored inlays have postrestorative sensitivity (1, 2). In these cases the patients often complain, immediately after cementation, of a sharp pain when the inlays are loaded. The symptoms may persist for weeks or months (3–5). Several factors have been discussed as the possible causes of postrestorative sensitivity. For example, factors such as desiccation of exposed dentin, polymerization contraction of the resin cement, deflection of composite under stress transmitting pressure to the odontoblastic layer, toxicity of the composite, and etching of dentin with phosphoric acid have been suggested (5, 6).

However, one conceivable cause of postrestorative sensitivity that has not been discussed until now is the piezoelectric phenomenon occurring in certain dental materials.

Piezoelectricity means that a charge is generated on the surface of certain anisotropic crystals when a mechanical stress is applied. The stress may be a compression, an extension, or a shearing stress—which is closely related to a compression—and there is a linear relationship between the applied force and the generation of electric charge. This physical phenomenon is called the *direct* piezoelectric effect. With a compression the charge appears on those surfaces that are loaded (the longitudinal mode) or on the surfaces perpendicular to the direction of the force applied (the transverse mode) as with a shearing stress. Longitudinal and transverse mode are terms that refer to different orientations of the crystal lattice with regard to the direction of applied force.

When the stress causes extension, the electric charge is created on the surfaces as with a compression, but the sign of the charge changes. In all cases the charge disappears when the load is withdrawn. In certain crystals, such as in quartz, piezoelectricity can be generated in three different modes—longitudinal, transverse, and shear.

The *converse* piezoelectric effect, also called reciprocal or inverse effect, means that the crystal alters its dimensions when a polarizing electric field is applied. The alteration is proportional to the magnitude of the polarizing field (7–9).

In March 1880 the direct piezoelectric effect was described for the first time by Jacques and Pierre Curie. They had found the effect in different materials such as zinc blende, sodium chlorate, boracite, tourmaline, quartz, cane sugar, and Rochelle salt (9), and the expression piezoelectricity was introduced 1881 by W. G. Henkel from the Greek *piezein*, which means 'to press'.

During World War I the Allies began to use the converse piezoelectric effect in technical equipment. Using quartz plates, electrically stimulated, ultrasonic waves were generated for the detection of submarines and as depth-sounding devices. The direct effect was used for transforming acoustic waves into electric impulses, and later the methods have had many applications, such as diagnostic medical devices, in echo-sounding techniques, and in measuring instruments (7, 9).

Since the direct piezoelectric effect is described as occurring in quartz and in materials with anisotropic structure, the aim of this study was to examine whether this physical property occurs in some materials intended for dental use.

Materials and methods

Four different dental ceramics and one indirect resin for inlays were studied: a) Vitadur (Vita Zahnfabrik, H Rauter GmbH & Co KG, Bad Säckingen, Germany); b) Cerec Vitablocs Mark II-porcelain (Vita Zahnfabrik, H Rauter GmbH & Co KG); c) Dicor Glass-Ceramic, conventional castable (Dicor, De Trey Dentsply Int., Wiesbaden, Germany); d) Dicor-MGC Glass-Ceramic (Dicor, De Trey Dentsply Int.); and e) SR-Isosit (Vivadent-Ivoclar, Liechtenstein).

Specimens a, c, and e were made up in accordance with the manufacturer's instructions, whereas specimens b and d were in the form of prefabricated ceramic blocks intended for the CAD/CAM technique. Four samples of each material with a specimen size of $8 \times 8 \times 15$ mm were first tested. Two surfaces opposite each other were coated with a two-component electrically conductive epoxy adhesive (Electrolite, Panacol-Elosol, GmbH, Frankfurt, Germany). A copper plate with a thickness of 0.1 mm and measuring $7 \times 7 \times 14$ mm was fixed to these surfaces with the adhesive.

The specimens were placed in an aluminum can, which served as a Faraday cage, to prevent interferences from the environment. A movable steel rod with one end shaped into a 4-mm semicircular steel ball simulating a dental cusp was inserted through the lid of the aluminum can. Both the steel rod and the lid were connected with cables to the aluminum can, which was electrically grounded.

Shielded electrodes were fixed to the copper plates by soldering, and the opposite ends were connected to a charge amplifier with a low-input impedance (1 kΩ) and with outputs for both charge and current. This equipment was especially designed and constructed at the Laboratory of Electronics, Faculty of Medicine, University of Umeå, Umeå, Sweden. The signals from the amplifier were conducted to an oscilloscope (Hewlett-Packard, USA). All connections between the Faraday cage, the amplifier, and the oscilloscope were conducted through shielded cables.

The specimen was placed at the bottom of the aluminum can with the 'cusp' of the steel rod placed towards the specimen, and the specimen surfaces coated with the conductive epoxy adhesive and the copper plates with their cables free (Fig. 1).

The other end of the steel rod was loaded with various forces by dropping a steel ball,

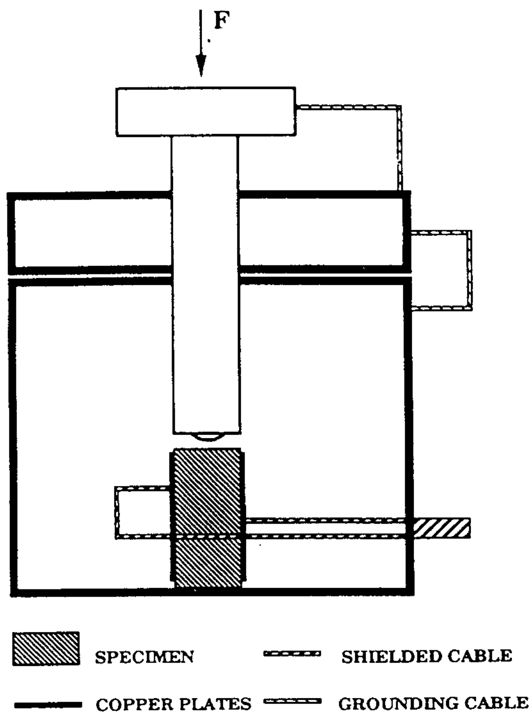

Fig. 1. The Faraday cage with a test specimen inside. F = force.

from different heights, through a tube of stainless steel placed on the top of the lid of the aluminum can. Holes were made in the sides of the tube to reduce influence from air resistance, and the tube was connected to the ground to prevent friction electricity, which could have disturbed the measurements.

The forces were measured with a miniature load cell (TH-UM, Hydronics Inc., Westerville, Ohio, USA) and a strain gauge indicator (BIN KI-D, Nobel Elektronik, Nobel Industries, Sweden) by screwing the aluminum can onto the miniature load cell. The load was applied at the top of the specimen a little off center, causing a shearing stress, and the current and pulse duration were calculated from the oscilloscope tracing.

To evaluate whether there were any differences due to the area of the specimens, four Cerec ceramic specimens measuring 4 × 12 × 15 mm were also tested as above.

To study the influence of cementation four Cerec ceramic specimens were then coated with a dual resin cement (Colténe Duo Cement, Coltène AG, Altstätten, Switzerland), and four were coated with a GPA cement (GC Fuji 1, G-C Dental Industrial Corp., Tokyo, Japan). Thereby two surfaces, perpendicular to the surfaces with the copper plates, on each specimen were coated with a thin layer of the cement after they had been etched with 4.9% HF acid for 1 min (Vita Cerec-Etch, Vita-Zahnfabrik) and silane-treated (Silicoup, Kulzer GmbH, Wehrheim, Germany). In addition, two specimen surfaces, perpendicular to the surfaces with the copper plates, on four Cerec ceramic specimens were coated with a dentin adhesive (Gluma, Bayer AG, Leverkusen, Germany). All the specimens were then tested in the same manner as above.

Results

Ranges for generated currents in the test series related to the corresponding ranges for applied forces and pulse durations are shown in Table 1.

Discussion

The cause of postrestorative sensitivity in teeth provided with posterior inlays seems to be multifactorial. Many of the explanations that have hitherto been described (for example, Refs. 5, 6) could, individually, explain the patients' symptoms.

In addition to all these explanations, piezoelectricity and related phenomena must be considered. The results of this study show that a current of up to 0.9 μA with a pulse duration of 0.4 msec was extracted from the surfaces of one of the dental ceramics studied when loaded with a dynamic force of 450 N, and since there is a linear relationship between the applied force and the generated electric charge (8), it is possible that higher values for the current could be obtained depending on the degree of alteration of the

Table 1. Ranges for current, force, and pulse duration for different materials

Material	Force, N	Current, μA	Pulse duration, msec
Vitadur	265–420	0.01–0.06	0.2–0.4
Cerec Mark II (size, 4 × 12 × 15 mm)	195–450	0.01–0.9	0.2–0.4
Cerec Mark II (size, 8 × 8 × 15 mm)	195–625	0.02–0.7	0.2–0.4
Cerec Mark II (size, 8 × 8 × 15 mm) with composite cement	175–425	0.02–0.45	0.2–0.4
Cerec Mark II (size, 8 × 8 × 15 mm) with GPA-cement	No measurable current		
Cerec Mark II (size, 8 × 8 × 15 mm) with Gluma	No measurable current		
Dicor conventional	170–509	0.01–0.65	0.1–0.4
Dicor MGC	120–480	0.01–0.6	0.2–0.4
SR-Isosit	175–505	0.01–0.5	0.2–0.4

dimensions of the ceramic specimens when loaded. In a clinical situation the fit of an inlay, the underlying dentin, and the size of the inlay, especially the thickness, may influence the values, because the deformation of an inlay also depends on these factors and not only on the force applied.

With regard to the pulse durations obtained in the present work it should be noted that they depend on the test arrangements—that is, the pulse duration varies depending on how the specimens are loaded.

The values for the charge generated also depend on the shape of the inlays and on the direction in which the crystals are loaded. The piezoelectric effect in the longitudinal mode is related only to the magnitude of the force applied, but in the transverse mode the effect also depends on the dimensions of the crystals. The charge generated increases if the crystals are long and narrow (10).

The currents generated on loading (Table 1) will be highly dependent on how the specimens are compressed under loading. During a series of experiments the first blow that hits the specimen will cause very fine microcracks in the outermost surface layer. The next blow that hits the same spot will therefore be somewhat attenuated. This means that the effect of the first blow cannot be fully reproduced since there is a minor change of the surface. In spite of the fact that a measuring series involved 6–27 separate blows, with the exception of the Vitadur series in which total fracture occurred already after 3–5 blows, it was therefore not meaningful to calculate mean values and standard deviations for the currents generated. In addition, the experimental set-up was only made to ascertain whether piezoelectricity occurs in the materials tested. Thus, the current values (Table 1) for the different types of specimens cannot be directly compared.

However, the most important factor for the value of the current generated is the rate of deformation when the inlays are compressed. By definition the charge is $Q = \int i\,dt$, where Q = charge (C), i = current (A), and t = time (sec).

From this follows: $i = dQ/dt$. But $Q = k \cdot F$, where k = a material constant (C/N) and F = the applied force in N.

Thus, $i = dQ/dt = k \cdot dF/dt$; that is, the current is highly dependent on the rate of change of the applied force. In other words, it is not *the value of the force* that is most important for the magnitude of the generated current. This reflects the fact that in the clinical situation the piezoelectric effect is dependent on the manner in which the inlays are compressed—that is, whether they are hit by a slow or rapid load. This might be one of the reasons why the symptoms do not appear every time the patient bites on the restoration.

With regard to the role of luting layers, the inlays will in many cases be quite close to dentin with no cement layer between the dental substance and the restorative material. The surface of dentin is not smooth, and there will be many irregularities preventing the cement from covering the whole surface, so areas without cement will appear.

Furthermore, the high capacitance makes it unnecessary for the restorations to be in direct contact with dentin or for the luting layer to be conductive; that is, charges can influence the pulp nerve even if there are small gaps between the inlays and dentin or a thin layer of cement.

The forces with which the ceramic specimens were loaded in the present study were measured by placing the Faraday cage, with the specimens inside, on a miniature loading cell. The mass of the Faraday cage and the elasticity of the specimens have of course influenced the force values registered, but the small differences from the exact forces applied to the specimens should in these cases be negligible, since the values measured give a good apprehension of the relationship between the applied forces and the generated current.

Mumford (11) reported that the pain threshold values of intact upper anterior teeth were in the range 0.7–4.5 μA when a pulse duration of 10 msec is used. For lower anterior teeth the range was 0.3–3.5 μA. Björn (12) reported values of 2.9 μA for the mandibular incisives and 15.5 μA for posterior teeth, with ranges of 0.7–10.0 μA and 3.7–38.0 μA, respectively (pulse duration, 10 msec). Virtanen (13) in a recent study reported a mean threshold value of 10.3 μA (pulse duration, 10 msec) for intact upper incisors.

In these cases (11–13) the pain threshold of intact teeth was tested with a current-measuring instrument with the stimulating electrode placed on the incisal edge or on a cusp. The stimulators were monopolar and were designed to overcome the high impedance of the external circuit. Nevertheless, the values for the pain threshold in the studies concerned cannot be directly compared with the current needed to influence the pulp nerve in a situation in which inlays, with their internal surfaces functioning as electrodes, are close to the dentin with its microcracks and tubules and when, furthermore, the rise time of the current is short. There seems to be a difference between the time needed to stimulate the pulp nerve through the tooth tissue of intact teeth and that needed to stimulate nerves that are directly ‘connected’ to the stimulating electrode. The time needed for the latter should be markedly shorter, since in this case there is a smaller amount of intervening tooth substance, with high resistance and capacitance, to delay and filter out the quickly rising current pulse. With regard to the rise time of the current pulse the lowest thresholds are obtained with the shortest rise time (12, 13).

The directions of the current are also important for the effect. The current is most effective when it flows parallel to the nerve fibers and weaker when it is perpendicular to the nerves (14).

Thus the pain threshold is very individual and can vary within wide ranges, and it therefore seems quite possible that the charge generated on the surfaces of ceramic and resin inlays might explain the sharp pain the patients sometimes complain of when the inlays are loaded by biting and chewing.

But what explanation is there for the fact that in most cases the symptoms disappear within a few months? One explanation is that the polymerization contraction of the resin cement admits the penetration of saliva along the lingual, buccal, approximal, or cervical margins of the inlay, thereby causing short-circuiting. Initially, short-circuiting through the saliva on the approximal surfaces of the inlay is prevented when the approximal cervical marginal conditions are absolutely sealed by the cement. When the marginal conditions and the cement admit saliva to penetrate, short-circuiting may be possible. In vitro testing showed that when short-circuiting occurs on the surfaces of the inlays and when specimens of resin cement are covered with a thin film of water or a physiologic saline solution, the conductivity increases and no current can be measured. That might be the reason no current was detected when specimens coated with a GPA cement or with the dentin adhesive (Gluma) were tested. The conductivity in the GPA cement and of the dentin adhesive may have caused short-circuiting of the surfaces.

The definition of piezoelectricity is that a charge is generated when certain anisotropic crystals are compressed. One interesting experience gained in this study was that it was not possible to measure any current in some of the conventionally manufactured specimens of the feldspatic ceramic Vitadur. One possible explanation may be that the traditional manufacturing technique for these specimens causes differences from time to time, so that the degree of anisotropic structure varies between specimens made of the same ceramic. Fired ceramic products may inherit a textural anisotropy that is derived, in part, from the preferred orientation of clay particles (15). The values of the current (Table 1) were also lower for Vitadur than for the machinable ceramics, which are fabricated using higher temperatures and pressures, which probably increase the formation of an anisotropic structure.

With regard to the Dicor glass ceramic the anisotropic structure is created by the growth of the mica crystals when heated above a temperature of 950°C (15). When the indirect composite resin specimen SR-Isosit was manufactured, a mold of an impression material was used, and initially a light pressure was applied to the resin when placing it in the mold, which might have influenced the orientation of the molecules and caused an anisotropic structure.

Another physical phenomenon that might influence the values of the current or the charge is friction electricity generated by minimal movements of the inlays when they are compressed by biting and chewing. Preliminary trials showed that friction electricity is generated very easily. During the construction of holders for the specimens and when looking for materials to use for embedding inlays or extracted teeth to be placed in the Faraday cage, we encountered many problems in avoiding friction electricity created from the materials used. Even when the surfaces of ceramic specimens were rubbed, with minimal movements, against plates of brass, friction electricity was developed, which disturbed the measurements. Therefore the specimens used in the present study were placed so that the surfaces to which the load was applied were tightly connected to the steel rod and to the bottom of the aluminum can, which were both electrically grounded so that the friction electricity could be conducted away.

In conclusion, piezoelectricity is a physical phenomenon that occurs in certain dental restorative materials, and the possibility cannot be excluded that postrestorative sensitivity may occur when these materials are used. There are many questions that must be studied further—for example, how does the presence of dental adhesives or the shapes and dimensions of the restorations—that is, crowns and different types of inlays and so forth—influence the generation of piezoelectricity and the consequences of this for clinical situations. Another interesting task is to evaluate how the specimen loading will influence the generated current—that is, the magnitude, the pulse duration, and the rise time.

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