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REVIEW

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Conventional and adhesive luting cements

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Abstract Luting materials for fixed prosthesis must fulfill special requirements in order to retain indirect restorations and fully maintain the integrity of abutments. The main requirements (inhibition of plaque accumulation, sealing of interface, possible antibacterial effects, acceptable solubility, wear, mechanical properties, adhesion, radiopacity, film thickness, type of curing, esthetics, storage, and cost) are reviewed to update clinical criteria on the selection of suitable materials. It can be concluded that there is no ideal luting material on the market. Alleged improvements in the physical data of newer materials do not necessarily result in better clinical performance. Only clinical trials can confirm the assumed benefits of materials.

Keywords Luting · Dental cements · Fixed prosthesis · Indirect restorations

Introduction

In recent years, many luting agents and dental cements have been introduced to clinicians with the claim of clinically better performance than existing materials due to improved characteristics. The luting of indirect restoratives to abutments is critical in achieving proper performance of indirect restorations. It is the final step in a chain of manipulations of diverse tissues, materials, and instruments. As for their main clinical usage, materials used for this purpose can be separated in two main groups: provisional and definitive.

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Provisional luting materials are of two main types: calcium hydroxide and zinc oxide cements (with eugenol or alternative substances). Many types of luting materials are considered definitive. Attending to adhesive potential, they can be divided into low (zinc phosphate, silicate cements), medium (polycarboxylate cement), or high (glass ionomer cements and filled or unfilled resins) luting materials. The curing of some resin materials is initiated either chemically, through the application of light, or both.

In general, dental luting’s two main functions are to establish or increase retention of the prosthetic appliance to abutments and to maintain its integrity. To succeed in both, an ideal material should fulfill specific biological, physiomechanical, and handling requirements [61]. Esthetic requirements are also very important, although not in a strictly functional way. The purpose of this article is to review how the presently available materials accomplish the more relevant functional requisites.

Biological requirements

The biological requirements of a luting material can be listed as biocompatibility, possesion of nontoxic and nonallergenic traits, inhibition of caries formation, and adequate sealing of the interface. Reviewing the biocompatibility and potential of toxic or allergenic reactions is beyond the scope of this paper, and for this the reader should refer to specific literature [20, 22, 63].

Inhibition of caries formation

Preventing caries formation is a complex issue. Only exclusion of bacteria from exposed dental tissue(s) and from the interface would impede the development of secondary caries. This is accomplished through avoiding plaque accumulation and physical exclusion of microorganisms from the exposed dental tissues and the interface by means of a perfect seal from the oral environment. A third way, considering that exclusion of microorganisms is far from possible, is to use a luting material with an antibacterial effect.

Inhibition of plaque accumulation

Plaque accumulation in dental cements and resin-type luting materials is mainly related to the possibility of achieving and maintaining a highly polished surface of the exposed, highly porous luting material.

Different luting materials have different polishing properties. In this regard, resinous materials are clearly superior to other product groups, since their organic phase can be easily polished. Its potential to be polished is then limited by the amount, size, and form of the inorganic filler phase. This is why a better-polished surface can more easily be produced in microfilled resin luting materials. However, such materials have an inherent drawback: a nonpolymerized surface layer. This layer is formed on the surface of the luting material that remains uncured due to oxygen inhibition and can be as thick as 270 μm [6]. There are two main ways to avoid this effect: placing a gel (glycerin, petroleum jelly, or similar material) on the external surface of the luting material space during curing, or permitting an excess of material that must later be removed. The second method has a clinical disadvantage in that overhangs of resin luting materials, once cured, are difficult to remove or even to find.

Little attention has been directed so far toward the problem of porosity in dental materials. Porosity influences a number of relevant material aspects, e.g., polishing. As pores can show, the surface will be imperfect (although polished) and make plaque accumulation easier. These pores have been shown to differ between different brands of hand-mixed luting materials and to be present in high numbers, from 4/mm^2 to 40/mm^2 of the material surface [43].

Sealing of the luting-tooth interface and hypersensitivity

It is generally accepted that the dentin-pulp complex is not totally sealed under physiological conditions [7] and that present adhesive systems do not hermetically seal the adhesive-tooth interface [13, 14, 15, 51, 59, 68], although several of these systems are considered clinically acceptable.

This sealing is accomplished by the intimate apposition of the luting or adhesive material to tooth structures. The degree of contact and possible intermingling can happen at the submicroscopic level, as in actual adhesives, through the hybrid layer [47], or at the microscopic level, as in so-called frictional luting materials [1]. Both systems use phosphoric acid, a well-known dental eroding acid through which action dental tissues increase their roughness and wettability [60], although to varying extents. The goal is to avoid gaps between dental tissues and the luting material. Such gaps or empty spaces can appear during insertion of the luting material, its curing, or its function. The circumstances influencing this will be discussed later.

From a biological point of view, adequate sealing of the interface is necessary to impede or reduce movements of dentinal fluids that would exceed normal rates. Sudden changes in normal dentinal fluid rates, beyond a threshold, have for a long time been known to cause tooth hypersensitivity [8]. The rate of these movements is generally higher during preparation of the restoration, due to drilling, air drying, impression taking, and ill fit of temporary restorations. All these factors cause a transient hyperemia in the dentin-pulp complex [9], with a concomitant rise in pressure within it. This elevation stretches the distance between threshold and actual levels of movement, causing “normal,” reversible, dental hypersensitivity.

When the final restoration is luted and the temporary dentin-pulp inflammation resolves (usually within 1 week), no further dental sensitivity must occur. If it does, the cause (apart from an irreversible dentin-pulp complex inflammation or a reversible one due to premature occlusal contact) is a nonsealed interface caused by an opened interface or an internal gap. Both allow the interface to act as a pump, allowing fluids to move back and forth [9]. The symptoms may depend on the location of the gap. If it is external, cold and hypertonic solutions will probably cause discomfort or pain. If it is internal, usually only occlusal contact will cause pain, even in the absence of prematurities [48].

Antibacterial effect of the luting

Some metal ions released from glass ionomer or zinc phosphate may have a cytotoxic effect [72]. Furthermore, it has been proven that fluoride is released from certain dental materials, although at different rates and with different durations, depending on the material tested [25]. This ion is a well-known enzyme inhibitor [23] and has a caries-inhibiting effect [67]. However, it is still unclear to what amount fluoride is needed, how long the effect will last, and what effect this continued elution will have.

Nevertheless, a gap-free interface seems more important in preventing secondary caries than the release of fluoride or other substances such as 12-methacryloyloxydodecylpyridinium bromide (MDPB) [29, 30] from materials alone [58, 64] or the bioactive intrinsic potential of the adhesives [28].

Physiomechanical requirements

Ideal luting materials must remain stable in the luting space, resist mechanical loads, provide physical or chemical adhesion to abutments, and be radiopaque.

Stability

Luting materials are always exposed to the oral environment in the marginal gap between restoration and abutment, no matter which techniques or materials are used. This gap exposes the material to oral fluids and mechanical action of food and oral hygiene devices, and its width depends on clinical circumstances as well as on the dentist's and technician's skills.

Solubility and wear of luting materials are both responsible for a certain degree of removal of the luting material from the luting space, with the result of roughening the exposed surface and – eventually – loosening the restoration.

Solubility

There is a lack of correspondence between standard solubility tests and clinical results [73]. The reasons for this are differences in the periods studied and the types and pH of storage media. Clinical conditions vary, even within the same patient, making it virtually impossible to reproduce a natural environment.

Certainly, other conditions besides solubility influence clinical outcome. Not only the balance between the bonded and unbonded surfaces (the so-called C factor) [16, 18] rules the contraction and stress behavior of resin materials, but also the fact that some materials, when adhering to two parallel walls, are subjected to completely different internal and external stresses than when not adhered [3, 4, 17].

An in vivo study [26] with patients wearing luting specimens in the lingual flanges of inferior complete dentures showed that polycarboxylate and zinc phosphate cement dissolved more than a glass-ionomer cement. Under scanning electron microscopy, glass-ionomer and polycarboxylate cements showed pits and extensive cracks on their surfaces, while zinc phosphate showed a large number of pits. In general, it is accepted that resin luting cements are less soluble than other luting materials [33, 73].

Wear

According to Pallav [52], we can distinguish several types of wear that frequently appear simultaneously and/or sequentially and also interact: abrasive, adhesive, delaminating, chemical, erosive, and impact wear and surface fatigue. Wear will not be present to the same extent in different areas of the mouth or teeth, nor will it be caused by the same mechanisms, varying also with the type of restoration.

In general, wear problems are of minor importance in classic, full-crown restorations [61] but take on relative significance in adhered, esthetic restorations. In these cases, susceptibility to wear increases when margins are located farther from the gingival area and approach the occlusal area [24, 40, 55], especially when the cavity margins are near occlusal stops or contact areas. It is generally accepted that wear is less pronounced in composite resin cements [21]. Proper fit of the restoration and the higher filler content [56] of composite resin luting cements increase wear resistance of the luting material [69], which decreases linearly with the increase in luting space [31, 35, 69].

Mechanical properties

Luting is the link between tooth and restoration, forming a complex interface. Consequently, as in most interfaces attached to dissimilar phases, it is subjected to complex challenges and must buffer the transition between and hold together two parts that differ greatly in – among other things – rigidity, wettability, color, direction(s) of movement, and chemistry.

In terms of retention, a luting material can be described as an all-or-nothing link: a partially broken cement film can (theoretically) still retain the restoration even while remaining attached at just one spot; however, microleakage and bacterial ingress due to microfracture may be present but clinically undetectable for long periods.

The elastic modulus (E) measures the ability of a material to resist elastic deformation under loading, representing the relative stiffness of the material within the elastic range. The E is a good measure of the ability of a luting material to transfer loads to the tooth, thus distributing stress. Although the ideal mechanical properties are unknown, it has been suggested [37] that a suitable luting material should have an intermediate (between tooth and restoration) E and it should have high resilience.

The E of resin-modified glass ionomers has been measured to be lower than that of dentin [37]. Resin composites, polycarboxylate, zinc phosphate, and mature glass-ionomer cements are in the lower, middle, and upper ranges of the values reported for dentin, respectively. These values were stable within 1 day for zinc phosphate and resin-modified glass-ionomer cement, within 1 month for resin composite, and continued to increase for 1 year for polycarboxylate or glass-ionomer cements. This increase will produce a very rigid layer of luting material, probably causing microfractures in the luting in the long run.

Resilience is defined by the amount of energy needed to deform a material permanently. This deformation would appear after repeated minimal displacement of the restorations in relation to the abutment, slowly flushing the interlock between the luting and the inner surface of the prosthesis and/or the outer surface of the tooth. Composite resin luting materials, resin-modified glass ionomers, and glass-ionomer cements have slightly greater resilience than zinc phosphate and polycarboxylate cements. The clinical significance of these differences is still unknown.

Fracture toughness (K_{IC}) of a material can be described as its ability to resist crack propagation [34] that may cause breakage. In general, composite resin luting materials have a significantly higher K_{IC} than resin-modified glass ionomers and are superior to conventional glass ionomers [34, 44]. As for the properties reviewed so far, composite resin luting materials represent the best luting material, probably followed by zinc phosphate cement [36].

Adhesion

As mentioned above, adhesion is accomplished by intimate apposition of the luting material to tooth structures. This intermingling may happen in two ways: at the submicroscopic level through the hybrid layer [47] or at the microscopic level in frictional luting materials [1]. Both systems benefit from the action of a phosphoric acid. As a separate clinical step, this acid allows the adhesive to penetrate demineralized dentin and, once cured, to form the hybrid layer. On enamel, this demineralization produces an enormous increase in surface roughness and wettability. On dentin and enamel, adhesives for resin composites produce the most predictable interlock via micromechanical retention.

As a part of the zinc phosphate cement, phosphoric acid erodes the abutment surface [65] to increase its wettability and roughness [60]. This material has not shown the same capability as actual composite resin adhesives to intermingle with dentin or enamel. That is why zinc phosphate is the “frictional” prototype of cement, as opposed to “adhesive” resin composites. Both interact with teeth in the same way, but at different levels and with diverse intensities.

Polyacid-based materials (glass-ionomer or polycarboxylate cements) also demineralize dental surface [65]. Their self-adhesiveness was demonstrated some time ago [66], and they have been shown to have a distinct interaction with dentin [74] via a transition zone, not exactly a hybrid between adhesive and collagen.

Whatever the mechanism is, the objective is to fix the cement to both walls of the luting space in order to attach the restoration to the tooth firmly. This situation of two large, practically parallel walls confining the luting space causes the major difficulty in adhering luting materials to both tooth and restoration. Many materials, particularly the resin-based ones, undergo shrinkage while curing. As the E increases and interlocking of the surrounding walls keeps the material attached, internal stress will develop perpendicularly [4, 10, 16] and parallel [32, 38, 39] to the interfaces, which may finally impair the material’s integrity and/or its attachment to the walls. This perpendicular stress can be mitigated at the moment – without disrupting the bond – by tooth or prosthesis deformation [4, 41] or internal porosity of the luting material. Internal porosity [43] would act as a stress reliever, as it adds an evenly distributed, internally nonadhered surface [2].

It has been shown [3, 4, 42] that, as tooth and all materials used in restorations have a certain elasticity, their strain deformation will reduce stress or at least its perpendicular component. Furthermore, this reduction will be more clinically relevant as the thickness of the luting space decreases: in thinner luting layers, minimal reductions of the distance between both limiting walls will bring about relatively high reductions in stress. Aside from other circumstances that also support this position, this should be enough reason to attempt the best possible fit of the restoration.

Little is known about the clinical relevance of shearing or tangential polymerization stress [38, 46], although it is likely to be considerable in luting layers adhering to two extensive surfaces, the abutment and the inner aspect of the restoration.

Radiopacity

Newer cements, especially adhesive ones, are normally color-matched to teeth for esthetic reasons. However, this may make excesses of cement in approximal surfaces, areas with anatomic surface variations, or gingival crevices [36, 45] difficult to locate and time-consuming to remove, especially when low-viscosity resin luting materials are used.

According to ISO 4049 specifications, the radiopacity of restorative materials should be higher than that of the same aluminum thickness. Presently there is no specification for radiopacity of luting materials, but it should be at least superior to that of dentin. Some resin luting materials include more radiopaque fillers, increasing radiopacity [49]. Proper radiopacity of a dental material allows differentiation between tooth and restoration to detect eventual gaps, secondary caries, overfillings, or underfillings. A material’s radiopacity is basically related to the atomic structures of its components. Components with higher molecular weight (i.e., metals) will retain more radiation (be more radiopaque) than plastic-derived or water-based materials (such as composite resins or glass-ionomer cements, respectively).

We recently measured the radiopacity of 250-μm layers of some typical luting materials [62], and all resin-composite and resin-modified glass-ionomer cements tested presented a remarkably lower radiopacity than zinc phosphate cement.

Handling requirements

Film thickness

As mentioned above, the luting space should be kept to a minimum to improve the fit of the restoration, expose a minimum of luting material to oral fluids, and minimize any polymerization contraction stress. There is no agreement on this minimum, but a 50–100-μm range seems convenient [36, 46]. Of course, this makes sense if the cement that must completely fill this hiatus is able to form a film of compatible thickness. It has been shown [71] that, for adhesively luted restorations, no significant differences occur with different designs of the retainer (whether flat or incorporating occlusal rests or grooves) with a mean luting space width of 75 µm. American Dental Association specification no. 8 restricts the film of zinc phosphate cement to a thickness between 25 µm and 40 µm [5]. This specification should be reviewed, because newer prosthetic materials have different needs and applications, and most luting materials incorporate different characteristics than frictional cements.

Various reports [50, 53, 57, 70] on the film thickness of a number of luting materials show that it can range from 152 µm to 10 µm, depending on the material tested. Regarding this, clinicians have to remember that dentin-bonding agents also have a measurable thickness. This can be acceptable (<50 µm) at the chamfer, the occlusal reduction, or the vertical walls, but it can be as high as 200 µm for some dentin-bonding agents at the dentin line angles between the chamfer and vertical walls [54].

Type of curing

Luting materials must undergo a chemical reaction to harden. This reaction can be initiated in three main ways: (1) mixing two or more different components of the material, which is improperly termed *chemical* curing (CC), (2) activating photosensitive molecules of the material in visible light curing (VLC), or (3) a combination of both methods called dual curing (DC). Relevant to clinicians is that the type of curing greatly influences three important aspects of the luting procedure: its control, its pace, and access.

Chemical curing materials typically consist of two pastes or a powder and liquid that are manually mixed, although some materials are encapsulated to make mechanical mixing possible. Manual mixing has been performed by dental teams for almost a century and tends to produce consistent cement mixes within individuals but with considerable scattering in mixing ratios and resultant material characteristics [19] when mixtures from various persons are compared. Its advantages are that it is well-known and cheap, that the whole mixture is subjected to the process simultaneously, and that the induced chemical reaction is relatively gradual and slow. Gradualness and slowness are the most convenient features, since they guarantee the slowest pace of possible contraction stresses. Disadvantages of this system, apart from the above mentioned scattering of results, are that the chemical process cannot be controlled by the clinician once the mixing has started, and that it entails a great number of bubbles [43] in the mixture that will hinder optical and mechanical behavior. These bubbles, on the other hand, may also act as stress relievers [2].

Visible light curing is the most extensively used system of polymerizing resin-based luting and restorative materials [11]. The key problem is ensuring that *enough* light energy reaches *all* parts of the luting material, to guarantee that the photoinitiator starts the subsequent polymerization reaction. This system provides the most controllable environment for clinicians, since setting will be initiated only where the light acts on the luting material. However, it has some important drawbacks regarding rate of events and access. Even using the minimum indispensable amount of light energy, the development of polymerization contraction stresses is much faster than with CC materials, thus threatening the ability of the interfaces, the luting itself, the restorations, and the abutment to adapt. This will probably worsen when massive amounts of light energy are applied to the luting, as with modern high-energy lamps [12].

Problems of access appear with metallic restorations, in profound areas where the light is filtered by the outer parts of the restoration or the luting, and generally in shadowed areas such as endodontic posts or core buildups.

Other considerations

Esthetics

Presently, an esthetic appearance of luting materials is virtually a must in almost all nonmetallic restorations, particularly when margins are visible. In such regions, color-matched resin-based luting materials are clearly superior to any other type, mainly due to their translucency and excellent color match to dentin and enamel. It is even possible to introduce slight color changes or adaptations of thin, nonmetallic restorations. Ionomer-based luting materials may have also a good color match, but their translucency is somewhat inferior.

Shelf life

Any material should have a convenient period during which physical and mechanical properties are maintained within a clinically acceptable range (shelf life). Changes in some properties of expirable materials over time may be unnoticed by clinicians [27]. With suitable storage, zinc phosphate's viscosity remains stable for long periods, but the viscosity of glass-ionomer cements increases significantly after 24 months. Diametral tensile strength is also affected, but reductions occur after an apparently long lapse (40 months) [27].

Cost

The price of luting materials is not a critical issue if considered to be dose-related, but it can influence the clinician's choice when acquiring a new material with limited indications and dubious usage.

Taking the Spanish price of zinc phosphate in the year 2000 as a reference (0.12 Euro/g), prices of other luting materials were 3.3 (zinc carboxylate cement), eight (glass-ionomer cement), 35 (resin-modified glass-ionomer cement), and up to 175 times (resin composite cements) more expensive. Of course, the densities are not exactly comparable, but this is probably outweighed, considering the probable waste in dosage.

Final remarks on criteria for cement selection

There is no ideal luting material available on the market today. The interested dentist has several options for luting indirect restorations and should use specific selection criteria. Alleged improvements in the physical properties of newer materials do not necessarily result in better clinical performance. Only clinical trials will identify the purported benefits of materials. Moreover, some properties (for instance, adhesion) have not yet been proven necessary for all circumstances:the requirement of a luting merely to adhere may allow one to sidestep more effective cements if retention is not an issue.

Luting is only one link in the chain of restoring a tooth. Others are at least as relevant: restoration material, patient diet, occlusion, prophylactic habits, design and preparation of the abutment teeth, impressions, and laboratory work. Certainly it is possible to impair the results of a restoration with an improper luting material, but it is no longer acceptable to improve unsatisfactory restorations by using high-fashion luting products.

Dental materials are quickly changing. New materials are speedily introduced to the market. Unfortunately, by the time adequate, time-consuming clinical studies have been published, these products will probably be off the market or changed ("improved") by the manufacturers and thus no longer available in the tested formulation. Therefore, the market for dental material is probably one of the most manufacturer-driven. In such an environment, it makes sense for the practicing dentist to stay with well-known, reliable materials instead of testing new products, unless their supposed properties are proven to be real, worthwhile, and relevant.

New materials are commonly more sensitive to technique. As they are introduced to fill gaps in performance of already existing ones, their specificity tends to be high and the technology on which they are based more sophisticated. This makes their use more demanding in terms of clinical skill, may restrict their usefulness, and could raise their price.

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