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Contents

Focus and Scope

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Seasonal greetings to all

It gives me immense pleasure to announce the first issue of ***International Journal of Dental Materials*** (IJDM), an open access peer reviewed journal. The aim of this journal is to provide a platform for the researchers from various disciplines to showcase the findings of research on dental materials. Though we find innumerable journals for materials science, there are very few journals dedicated to dental materials. The advancements in the multidisciplinary specialty of dental materials are rapidly progressing with the introductions of new materials and techniques in different areas of dentistry at a rapid pace. A good quality evidence based research governs the success of innovations in Dental Materials Science and IJDM will strive to support such practices by facilitating free and open access to the literature of researchers, academicians and clinicians in this field.

IJDM invites manuscripts from various dental specialties that focus on the aim and scope of the journal. Our goal is to encourage international collaboration among researchers to submit genuine and pioneering research articles for publication. In addition, we also promote, explore and propagate the concepts of young researchers by publishing their research.

IJDM was initiated to recognize scientists in the field of Dental Materials Science and to share their findings with the broader scientific community. Our editorial team is committed to get the journal indexed in various indexing databases and therefore achieve global recognition by selectively publishing high quality journal articles.

At the outset, I value your exceptional contributions for empowering knowledge in the field of Dental Material Science. I strongly believe that potential research contributions will be advantageous to those working in this field and I confidently trust that this will enhance the quality of our Journal. Furthermore, they also offer budding investigators with the ambition to explicate their concepts in emerging fields.

My special thanks to the authors for their contribution and also our editorial board members, reviewers for their timely response and cooperation to release the first issue of IJDM. I am confident that IJDM will reach to the best position in its field with your enormous support and research contributions.

Rama Krishna Alla
Editor-in-chief
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Comparative evaluation of fracture resistance of self-adapting PFS and elastic FRC post and core systems – An invitro study

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ABSTRACT

Background: Endodontically treated teeth (ETT) with extensive coronal destruction are more prone to fracture, so restoring these teeth with techniques that will not compromise the integrity of remaining tooth structure with the use of Post and core systems to retain full and final crown restorations seems mandatory. Anatomic posts have been introduced which have better adaptability to the canal anatomy and conserve more amount of tooth structure. **Aim:** This study was done to compare the fracture resistance of ETT restored with two anatomic post systems elastic FRC post (everStick) and self-adapting PFS (Spirapost). **Materials and Methods:** Twenty single rooted maxillary central incisors were selected for the study. All the samples were endodontically treated and randomly divided into 2 groups (n=10) according to the post system used (PFS post – Group I, FRC– Group II). In all the samples, post space preparation was done and the posts were luted using dual cure resin cement (Para core, Coltene, Mumbai, India). The remaining core was built using composite resin (Filtek, 3M, ESPE, USA). The samples were stored in saline for one week. All the samples were thermocycled for 500 cycles from 5 to 55 ^{0}c ±5 ^{0}c with a dwelling time of 30 seconds in each bath and a transfer time of five seconds. Fracture resistance of the samples was measured using universal testing machine. The obtained data was statistically analyzed by using independent t test. **Results:** There was no statistically significant difference between fracture resistance values of FRC and PFS groups. 30% and 70% of the samples of PFS and FRC showed favourable fractures respectively. **Conclusion**: The fracture resistance of PFS was comparable to that of FRC post.

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1. Introduction

Endodontic treatment is mostly preferred on teeth significantly affected by caries and/or fracture. ETT are considered more brittle, when compared to their non-treated counterparts, due to moisture loss [1] and loss of collagen cross-linking in the dentin after endodontic treatment [2] and loss of structural integrity associated with the access preparation [3]. Successful treatment of such badly broken down teeth with pulpal disease depends not only on good endodontic therapy, but also on good post endodontic restoration. Therefore, there is a need to rehabilitate these teeth with techniques that will not compromise the integrity of remaining tooth structure with the use of post and core systems to retain full and final crown restorations [4].

Different types of posts have been evolved since past, which include custom made and pre-fabricated posts. Pre-fabricated posts may be either metal or non-metal posts. Conversely metal posts showed several shortcomings such as corrosion, inflammatory reaction, high incidence of catastrophic root fractures, discoloration.

Fibre-reinforced composite (FRC) posts were developed to overcome the disadvantages of metallic posts. The preference and popularity of FRC posts can be attributed to its elastic modulus which is similar to that of dentin so that the occlusal stresses are distributed evenly resulting in favourable root fractures that are easy to repair. They can be easily removed from the root canals in the cases of retreatment [5]. The other advantages are the adhesion and micromechanical bonding characteristics of these fiber posts to the resin luting agent, dentin, and composite core which gives life-like appearance.

Pre-fabricated glass fibre posts are available for many years but these posts have some disadvantages like require shaping of the canal walls to fit the dowels, leading to dentin loss which increases the incidence of root cracks and fractures, poor adaptability in oval, curved and flared canals [6].

The post and core system with poly fibre strands PFS called Spira-post (Zenith Dental, DMG America, USA) which naturally flexes to conform to the shape of the canal thereby it can be angulated to offer the benefits of a custom fit for different root canal configurations. Due to its unique design, Spira-post is able to adapt to the irregularities of the canal, so it minimizes the removal of tooth material. Spira-post has many more advantages such as being lesser technique-sensitive, highly esthetic, even distribution of stresses, easy removal during retreatment [7].

An elastic Fibre reinforced composite (FRC, everStick, GC, India) which is a soft, flexible and un-polymerized glass fibre post which can be individually adapted to the shape of the root canal whether it may be curved or oval as well as in widen canals. It can be angulated to certain degree, according to the altered existing occlusion to offer benefits of a custom fit for all root canal configurations. It has some added advantages: more tooth structure preservation, high flexural strength, elasticity very similar to the natural elasticity of dentine, even distribution of occlusal stress on the root structure thus reducing the risk of fracture [8]. The purpose of the present in vitro study was to evaluate and compare the fracture resistance between FRC and PFS post systems.

2. Materials and methods

Twenty single rooted maxillary central incisors were collected from the Department of Oral and Maxillofacial Surgery, and stored in 0.1% thymol solution.

2.1. Root canal treatment

Teeth were decoronated accordingly to maintain uniform tooth length of 16mm from root apex. After making standard access cavity preparations, root canals were instrumented up to ISO size 30 with 4% taper M two rotary files (VDW, Germany). At every instrument change, the root canals were irrigated with 3 ml of 3% Sodium hypochlorite (Prime Dental products, Mumbai, India) followed by 1 ml of 17% Ethylene diamine tetra acetic acid (RC help, Prime Dental PVT LTD, India) and final irrigation was done by 10 ml of 0.9% saline. The root canals were obturated with corresponding gutta percha (Dentsply, USA) up to 5mm of root length by sectional obturation technique using AH plus sealer. The specimens were randomly divided in to two groups with 10 specimens (n=10) in each group and then the root canals were cemented with fibre posts. The root canals cemented with Poly Fibre Strands (PFS) called Spira-post (Zenith Dental, DMG America,

USA) and Fibre Reinforced Composite post (FRC, everStick, GC, India) were allocated to group I and II respectively.

2.2. Sample preparation for PFS

The post space was refined and enlarged upto gates Glidden (#4 Mani, Japan) according to manufacturer's instructions. The length of the remaining canal was measured using endodontic file and once the desired length was obtained, the Spirapost was measured and cut to the required length such that 4mm of the post was protruding out of the canal.

2.3. Sample preparation for FRC

The post space was refined and then everStick post foil bag was opened, and markings were made on the post according to the required length and the post was cut to a suitable length with sharp scissors. The length and suitability of the everStick post was checked by inserting it into the root canal till the post space and 4 mm of the post was protruding out of the canal. The fit of the post was finally checked and additional post material was shaped and attached tightly to the main post both coronally and inside the root canal by means of lateral condensation and the post was light cured before placing into the canal.

2.4. Cementation of post

The canal was rinsed to remove any debris and etched using 37%phosphoric acid and rinsed the canal with distilled water and dried using a paper point. The bonding agent was applied to the canal and the portion of the post being inserted into the canal. The canal space was filled with dual cure resin cement (Para core, Coltene, Mumbai, India) and post was inserted into the canal space using gentle pressure to allow any excess material to escape and the excess material was carefully removed and light cured for 40 seconds.

The core of dimensions 4x5 mm (length x width) was built using composite resin (Filtek, Z350 3M, ESPE, USA) according to manufacturer's instructions and final length of the prepared tooth was 6×5mm.

The samples were stored in saline for one week. All the samples were thermocycled for 500 cycles from 5 to 55 ^{0} c ±5 ^{0} c with a dwelling time of 30 seconds in each bath and a transfer time of five seconds.

2.5. Periodontal ligament simulation and fracture resistance test

To perform the compressive testing, each specimen was mounted in self cure acrylic block such that 1mm of root trunk is exposed and the socket was relined using poly vinyl siloxane impression material to simulate periodontal ligament. This specimen was placed into a fixation device (Jig) of universal testing machine (Instron 8801, USA) and a compressive load was applied at 45^{o} angulation on the palatal surface to the long axis of the tooth at a crosshead speed of 1mm/ minute until failure occurred. The amount of the load required to cause failure was automatically recorded in MPa using the machine itself. The modes of fractures were also observed and were classified into supragingival and subgingival fractures. Supragingival fractures are considered favourable in which fracture line occurs 1mm above the acrylic resin. Subgingival fractures are considered unfavourable in which the fracture line occurs 1mm below the acrylic resin.

The data was collected and tabulated for statistical analysis. Statistical analysis of data was performed using the software statistical package for the social sciences (SPSS, version 20.0) and analyzed by independent t-test.

3. Results

Means and standard deviations were calculated for both the groups. The highest mean fracture resistance was demonstrated by the PFS (G-I) which showed mean fracture strength of 898.30±355.25MPa, followed by FRC (G-II) which showed mean fracture strength of 667.08±363.151MPa (Table1). The difference in the mean fracture resistance among the groups was found statistically insignificant (P=0.167). Maximum number of the samples (70%) in the FRC group (G-II) showed favourable type of fractures, where as in PFS group (G-I) 30%samples showed favourable fractures (Table2).

4. Discussion

The present in vitro study was attempted to evaluate and compare the fracture resistance and also the mode of failure of endodontically treated teeth restored with of two novel post and core systems; PFS post and FRC post.

Groups	Mean	SD*	P Value
PFS (G-I)	898.30	355.25	0.167**
FRC(G-II)	667.08	363.15	

Table 1. Comparison of fracture resistance. Where *Standard Deviation, **not statistically significant.

Groups	Favourable	Unfavourable
PFS (G-I)	3	7
FRC (G-II)	7	3

Table 2. Different modes of failures

In this study, the endodontically treated teeth with different fibre post systems exhibited superior fracture resistance. Among the post systems studied, endodontically treated teeth with PFS post showed highest mean fracture resistance (898.30±355.25 MPa). However, no statistical significant difference (P=0.167), was found between the groups.

The higher fracture resistance of PFS group was due to its innovative composite structure of surgical stainless steel wires twisted around natural colour poly fibre strands which can compress or expand according to the diameter of the canal which reinforces the remaining root structure which further improved the fracture resistance of the tooth[7] poly-fibre strands create a homogenous unit by integrating with resin cement and core material in the canal resulting in a mono-block type of restoration. This creates a strong structure that absorbs and distributes external forces, and minimizing the risk of failure. These strands does not hold memory, so there is no concern for undue stresses within the canal.

The presence of a high molecular weight polymethyl methacrylate chains in the FRC post act as stress-breaker. A-glycidyl methacrylate matrix, decrease stress concentration at the interface of fibre matrix during deflection, and absorption of emerging stresses through the matrix[9]. Also, during manufacturing of FRC posts, the rehabilitating effect of unidirectional impregnated fibres can be created. These impregnated fibres are soaked with resin matrix in a pre-stressed tension that released after curing causing fibres compression which can absorb the tensile stresses under flexural forces [10].

Moreover, these fibres facilitate stress dissipation, supports the fillers of composite layers, and act as a crack stopper. The more increase of fibres in the matrix, the more increase of the post resistance to micro cracking. The bonding of FRC post with composite resin and also with the luting agent was improved by an interdiffusion bonding mechanism resulting in a 'Monobloc' type of restoration. This is attributed to the inherent character of the Interpenetrating polymer network (IPN) of the FRC post.

Comparing the modes of failure, more numbers of catastrophic/unfavourable failures were observed in group-I that is PFS group. Among the 10 samples in this group, 7 samples have reported subgingival fractures and the remaining 3 samples have reported supragingival fractures, which indicate 70% of the sample showed unfavourable fractures. Similarly, seven samples showed supragingival fractures and remaining three samples showed subgingival fractures among FRC post group. The results indicated that 70% fractures are favourable fractures.

The greater number of unfavourable fractures in Spira post group could be due to their structure of fibres and central metal core. The fibres which are too thin and flexible, couldn't withstand the forces for a longer time which resulted in uneven distribution and concentration of stresses along the metal core that have been transferred to the deeper and along the root structures resulting in catastrophic root fracture.

The greater number of favourable fractures in FRC post group was possibly due to similar elastic modulus of accepted range between dentin and everStick post resulted in equal dissipation of forces through overall length of the root without its fracture. Therefore, it can be attributed that FRC post act as stresses absorbers [11]. A layer of polymethylmethacrylate (PMMA) is present on the external surface of these posts. Adhesive resins which have solubility parameters close to that of PMMA can diffuse into the FRC post. Stick resin is one such adhesive resin that contains 2,2-Bis-[4-2-hydroxy-3-methacryloyloxy-propoxyphenyl-propane (bis-GMA), Tri-ethylene-glycol-dimethacrylate (TEGDMA), camphoroquinone and 2-dimethyl amino ethyl methacrylate. The bis-GMA and TEGDMA present in Stick resin have solubility parameters close to that of PMMA, thus enabling the penetration of the resin into the PMMA present on the outer surface of the FRC po-

sts which becomes interlocked into the IPN polymer matrix after polymerization[12,13].

5. Conclusion

Within the limitations of the study, it can be concluded that fracture resistance of PFS post is comparable to that of FRC post. However, FRC group showed predominantly more number of favourable fractures.

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Disinfection of Impression Materials: A Comprehensive Review of Disinfection

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ABSTRACT

Impression making is one of the most common procedures that are performed by dentists in day-to-day practice. These impressions can act as vehicles of transmission and carry various types of microorganisms, which further cause diseases like Hepatitis B, C, HIV, Tuberculosis etc. This contamination and cross contamination of microorganisms can be prevented by disinfecting the impressions immediately after removing from the mouth and label them as disinfected. Usually the impressions are placed under running water to remove saliva and blood, but this will eliminate the disease-causing microorganisms, so a standard protocol to disinfect the impressions and casts should be known to dentists and dental personnel. Various methods of impression disinfection like chemical disinfection, Microwave, Autoclave, Ultraviolet radiation have been described in literature having their own advantages, disadvantages and effects on impression material and casts. Recently antimicrobials and nanoparticles have been incorporated into the impression material itself to make it self- disinfecting. This will not only disinfect the impression material from inside but also disinfect the impressions from the time it is inserted in patient's mouth. A broad search on the literature available was performed to provide knowledge about mechanism of action, concentration of usage along with commercial preparations available of different disinfectants. This review article will enhance the knowledge and improve the behavior of dental health care workers about impression disinfection.

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1. Introduction

Dentistry is a branch of surgery that involves exposure of a person or materials to saliva/blood and other potentially infectious materials either directly or indirectly. On an average, 1ml of a healthy person's saliva contains about 750 million microorganisms [1]. Numerous studies have reported the colonization of distinct bacterial communities on different oral structures and tissues, and about 280 bacterial species from the oral cavity have been isolated [2]. Most commonly observed microorganisms in oral cavity of patients wearing prosthetic dental appliances, removable orthodontic appliances include *Staphylococcus, Streptococcus, Lactobacillus, Actinomyces*, and *Candida* species [3]. Health care professionals, especially dealing with oral diseases, are more vulnerable to cross infections during treating patients [4]. Furthermore, dental technicians are also susceptible to these infections as they handle various dental materials which are directly carried from the patient's oral cavity to the dental laboratory. Chidambaranathan AS *et al* (2017) reviewed and compared the various disinfection techniques available in the literature and they reported that 67% of the materials received in laboratories were contaminated with *Streptococci, Staphylococci, Candida* species*, methicillin-resistant S. aureus (MRSA), or P. aeruginosa* [5]. It was reported that dental personnel have a 5–10-fold chance of acquiring Hepatitis-B infection than the general population [6]. numerous studies also stated that Tuberculosis and Hepatitis-B microbes can survive up to 7 days or longer at room temperature [7].

Practicing high standards of infection control and safety measures are essential to control cross-contamination and occupational exposures to blood and saliva-borne diseases. The British Dental Association (BDA) stated that "Infection control is a core element of dental practice" [8] and certain recommendations are applicable to all levels and fields of dentistry including personnel involved in providing dental care directly or indirectly [9].

2. Dental Impression – A Possible source of cross-contamination

Impression making is an important practice to be carried out in dental treatment, specifically, involving in making a replica of the oral structures. During impression procedure, impression materials often encounter with saliva and blood, which may be infected with infectious diseases such as AIDS, herpes, hepatitis, or tuberculosis [10]. Dental impressions that are exposed to patient's saliva or blood, contaminate stone casts [11] and serve as a source of infection to dental personnel who handle or deal with the impressions or casts [12]. The personnel who works on such contaminated casts can cross-contaminate one patient casts to other and finally to the dentist and other patients. Therefore, Infection control is an essential and imperative issue in the dental practice to prevent the spreading of infection from one patient to another and also to provide protection to the dental health care providers. This article gives an insight on importance disinfection of dental impressions in preventing crosscontamination and also emphasizes the various disinfection modalities recommended for various impression materials.

BDA had recommended to decon-taminate and disinfect the dental impressions before they were sent to the dental laboratory [8] and it has evidently specified that the dentist is solely responsible for disinfection of the impression before it is being sent to the laboratory [9]. In 1998, FDI guidelines suggested that all impression materials, before transferring to laboratory, should be disinfected [13]. American Dental Association (ADA) [14] and Center for Disease Control (CDC) [15] also suggested disinfection of impressions or impression materials to prevent cross-infection and this can be accomplished by either immersion or spraying with disinfectants or other different methods.

Disinfectants should function as effective antimicrobial agents without showing adverse effects on the accuracy and the dimensional stability of the impression material. Various disinfectants such as sodium hypochlorite, glutaraldehyde, iodophors, and phenols are advocated for disinfecting impression materials [16]. The better practice to follow is the cleaning of the impression under running water with subsequent disinfection process, and also coordinate with the laboratory by labelling the device to specify disinfected status, as there is a possibility of repeated disinfection of an impression in the dental laboratory. This repeated disinfection may have detri-mental changes in dimensions of the impressions made [16].

Microorganisms can survive on, or even inside, the im-

pressions. However, the number of microorganisms decrease rapidly after impression making, and they are further eliminated by rinsing the impression under running water immediately after removing it from the patient's mouth [7]. Though this was the common practice and recommended procedure for disinfection of the impressions until 1991, numerous studies reported that such practice can only eliminate 40% of bacteria, viruses, and fungi, hence there is potential for infection [17]. Therefore, an appropriate disinfection treatment of impressions is mandatory to eliminate potential risks. Several disinfection methods such as chemical disinfection (immersion method/ spray method), autoclave, microwave, ultraviolet radiation are proposed to disinfect the dental impressions and models [10].

3. Chemical disinfection methods

Chemical disinfection preferably immersion, seems to be the most reliable and practical method, provided it does not adversely alters the dimensional accuracy of the impressions. Immersion disinfection is considered as an effective method as it ensures that all surfaces of the dental impression are exposed to the disinfectant solution. However, this method is not indicated for hydrophilic impression materials like hydrocolloids and polyethers as they can imbibe the disinfectant solution that results in dimensional inaccuracy of the impression [18]. In addition, immersion disinfectants should be discarded after every use (except for glutaraldehyde) [19], and it is a time consuming and expensive method.

The spray method is the other chemical disinfection modality which reduces the chance of dimensional changes or distortion, especially in hydrocolloid and polyether impression materials. However, it may not effectively disinfect the impression as this technique uses less amount of disinfectant solution and it may be inadequate to reach the areas of undercuts. Additionally, chemical disinfectants must be freshly prepared and also possesses poor shelf life [20].

3.1. Solutions used as chemical disinfectants for impression materials

3.1.1. Iodophors

Iodophor was discovered by H. A. Shelanski and M. V. Shelanski. This bactericidal, sporicidal, viricidal, and fungicidal compound is a complex of polyvinyl pyrolidone (PVP, povidone) and elemental iodine. They need more contact time with the impression material to achieve disinfection which may results in the dimensional inaccuracy of impressions. In addition to applications in dentistry, Iodophors are often used for the thermometers, disinfection of blood culture bottles, hydrotherapy tanks, and endoscopes [21].

3.1.2. Glutaraldehyde

Glutaraldehyde is a pungent colorless oil and can be used as disinfectant in liquid and gaseous forms. It is widely used to sterilize medical and dental instruments and also as preservative in industries. Glutaraldehyde possess bactericidal, viricidal, fungicidal, sporicidal, and parasiticidal activity. Their efficiency is increased in the presence of lower concentrations of organic material. They are recommended to be used with suitable protective equipment in a ventilated environment under the supervision of a trained person [21].

3.1.3. Sodium Hypochlorite

Sodium hypochlorite is a chemical with formula of NaOCl. It is composed of sodium cation and hypochlorite anion. It is water soluble. It is often used in industries for bleaching, surface purification, odor removal, and disinfection of water. Hypochlorite removes stains from clothes at room temperature. Hypochlorous acid and Sodium hydroxide is formed by adding water to the hypochlorite and it can be further dissociated into hydrochloric acid (HCl) and oxygen (O). The oxygen atom is a very strong oxidator [21]. The concentration of available chlorine and the pH of the solution governs the efficacy of NaOCl as disinfectant. Hypochlorous acid (HOCl) is a weak acid and it dissociates to the hypochlorite ion (-OCl) and proton (H^+) depending on the pH of solution. Generally, it is considered that HOCl is the dynamic species in the bactericidal activity, while the concentration of -OCl is a key factor that determine the cleaning efficacy. It is assumed that HOCl penetrates into the microbial cell across the cell wall and inhibit the enzyme activity essential for the growth, damage the cell membrane and DNA, and perhaps an injury to membrane transport capacity [22].

3.1.4. Benzalkonium Chloride (0.25%)

This is a quaternary ammonium (QA) chloride salt in which the nitrogen is substituted by a benzyl group, two methyl groups, and even alkyl chains. It has antibacterial, antiseptic, detergent, and surfactant action. However, Benzalkonium chloride (BC) is not effective against fungi, viruses, and bacterial spores. QA disinfectants possesses a strong positive charge which combines well with negatively charged surfaces. This makes the QA as good cleaning agent. QA compounds show low toxicity, but prolonged contact may irritate the tissues. Usually they are used for environmental sanitation like floors, furniture, and walls [21].

3.1.5. Isopropyl Alcohol

Isopropyl alcohol is a 2-propanol with the formula of C_3H_8O or C_3H_7OH. It is a colorless chemical compound with strong odor. It is commonly used as a topical antiseptic, and also to disinfect the surface of medical devices. It is suggested that the alcohol-based solutions should be stored in a cool, and well-ventilated area as they are highly flammable. Alcohol irritates the tissues and evaporates rapidly [21].

3.1.6. Ethyl Alcohol

Ethyl alcohol is more bactericidal than bacteriostatic, also tuberculocidal, fungicidal, and virucidal against enveloped viruses. Alcohols are not effective against bacterial spores and non-enveloped viruses. They denature the bacterial proteins, thereby inactivating the microorganisms. The ideal bactericidal concentration in water is 60% to 90%, and the bactericidal activity decreases on diluting the concentration below 50%. Ethanol has shown clear bacterial growth inhibition, especially when used in high concentrations against *S. mutans* and *S. aureus* [23].

3.1.7. Chlorhexidine

Chlorhexidine (CHX) is a positively charged molecule that binds with the negatively charged sites of the cell wall and destabilizes it. Hence, it interferes with osmosis of the cell wall. The CHX then attacks the cytoplasmic membrane and leaks the components that lead to cell death. In high concentrations, CHX causes the cytoplasm to congeal or solidify. The bacterial intake of CHX is very rapid (<20 seconds) [26]. No antifungal activity of CHX has been observed in the agar diffusion test in low concentrations, but 2% CHX showed antimicrobial activity against *S. aureus*, *E. coli*, and *B. subtilis*, but not *C. albicans* [24].

3.1.8. Ozone water

Ozone, is a gaseous inorganic molecule with the chemical formula of O_3. It is less stable than O_2 and easily breaks down to normal dioxygen in the lower atmosphere. Ozone is formed by the action of atmospheric electrical discharges and ultraviolet (UV) light from dioxygen. It exists in low concentrations (0.6 ppm) in the atmosphere. It is a potent oxidizing and antimicrobial agent. Ozone is an unstable compound that decomposes very quickly (half-life 40 minutes at 20°C). Ozone affects the cell membrane, vital proteins, unsaturated lipids, and the intracellular enzymes of microorganisms and may also cause DNA degradation [25].

3.2. Levels of disinfection with chemical disinfectants

Chemical disinfectants can be classified into three categories based on their efficiency against vegetative bacteria, tubercle bacilli, fungal spores, and viruses. The level of disinfection for various impression materials with different disinfecting agents is detailed in Table 1. *High-level disinfectants* are capable of inactivating bacterial spores and all other microbial forms, which is an essential criterion for high-level disinfectants. Commonly used high-level disinfectants include Ethylene oxide gas or Glutaraldehyde solutions. *Intermediate level disinfectants* accomplish destroying microbes like tubercle bacilli, however, they do not have any effect on spores. Formaldehyde, Chlorine compounds, Iodophors, Alcohols, and Phenols are widely used Intermediate level disinfectants. The chemical agents that show narrow antibacterial activity are considered as *Low-level disinfectants*. These are undesirable for disinfection of impressions [26]. Quaternary ammonium compounds, Simple Phenols, and Detergents are classified as Low-level disinfectants.

For effective disinfection, the contact time between impression and disinfectant must be at least equal to the time for tuberculocidal activity as recommended by the manufacturer of the germicide. It is essential to rinse the impression immediately after disinfection to remove residual disinfectant from the surface of impression. Kotsiomiti *et al.,* (2008) conducted a review on

Level of Disinfection	Disinfecting Agent	Impression Material
High Level Disinfection	Glutaraldehyde	Irreversible Hydrocolloids Zinc-oxide Eugenol Paste Polysulphide Polyether Silicones
Intermediate Level Disinfection	Sodium hypochlorite Complex phenolics Iodophors Chlorhexidine Alcohols	Irreversible Hydrocolloids Zinc-oxide Eugenol paste Polysulphide Polyether Polysilicones Impression Compound
Low Level Disinfection	Quaternary Ammonium compounds Simple phenols Detergents	Not recommended for disinfection of impressions

Table 1. Levels of disinfection for various impression materials with different disinfectants

accuracy and stability of impression materials subjected to chemical disinfection and suggested that the disinfection modalities should be restricted to methods that show least distortion on impressions and on the chemical nature of the impression material. Hydrocolloids should be disinfected for a limited time period. Immersion is more secure than spraying and self-disinfecting materials are efficacious, but better accompanied by immersion. Polyethers, on the other hand, can be effectively disinfected by spraying. Although this seems to be the preferred method for disinfection of these water-friendly materials. Modern polyethers seem to withstand immersion, even long-term. Little information could be traced considering the stability of hydrophilic silicones upon prolonged immersion disinfection. Until more sound evidence is available, long-term exposure of them in the disinfectants should be applied cautiously, as they may have an enhanced absorption potential. Hydrophobic elastomeric materials can be safely immersed in disinfectants and left for a long period [27].

4. Microwave disinfection methods

Microwave disinfection is an effective and versatile method, which is quick, easy, and inexpensive method. This method can be easily performed by dentists, assistants, and technicians. Thermal and non-thermal are the mode of actions used with microwave disinfection. Thermal effect is conversion of microwave energy into heat by prolonged kinetic motion of polar molecules, where as non-thermal effect is by direct interaction of electromagnetic field with the biologic molecule, creating effects that cannot be caused by thermal action alone [28]. Microwaves are responsible for antimicrobial action by disrupting the cell membrane integrity and cell metabolism of microbes [29].

5. Autoclave disinfection methods

Steam autoclave is a device used to sterilize equipment, surgical instruments in medicine and dentistry by subjecting them to high saturated steam pressure at 121°C or more for 15 to 20 minutes. An autoclave also works at 115°C/10 psi, 121°C/15 psi, and 34°C/30 psi. The standard settings can kill most bacteria, spores,

viruses, and fungi at 134°C, which can be achieved in 3 minutes. The color change indicates that the object inside the package or under the tape, has been autoclaved. Addition or condensation silicone materials could be sterilized producing less than 0.5%-dimensional change at 134°C, without showing relevant changes in tear strength [30].

Ethylene oxide gas autoclave is the other method used for disinfection of dental impressions. Holtan et al, (1991) showed that ethylene oxide gas sterilization allowed inclusion of gasses into the vinyl poly siloxane (VPS) impression, and these gasses would release later that results in producing bubbles in dies if poured instantly. This can be avoided by pouring the dies after 24 hours. They also suggested that steam autoclaving was a suitable method, specifically, if the impressions were made for the fabrication of removable prostheses [31]. Olin et al., (1994) reported that the use of ethylene oxide gas autoclaving of heavy- and light-body addition silicone impression material in custom trays showed significant structural changes (>0.5% change) due to the distortion of the trays or incapability of preventing expansion of the impression material [32].

6. UV light as disinfectant

The effectiveness of UV rays in disinfection depends upon the time, intensity, humidity, and access to the microorganism. Since dental prostheses provide a number of sites for housing microorganisms, UV light must be reflected from many directions. UV light exposure has shown to drastically reduce *C.albicans* colonies as compared to direct-current low discharge. It has been observed that a higher-watt UV light tube decreases the colony count in less time. The maximum killing efficiency with UV light exposure has been obtained with 24 watts (3750 $\mu w/cm^2$). The higher wattage required less time to decrease the colony count of *C. albicans* to zero [33]. Samra *et al*., (2018) recommended the Ultraviolet method as more suitable for disinfecting impressions without compromising their dimensional stability [34]. Various impression materials and their disinfection choices are mentioned in Table 2.

7. Disinfection of casts and models

Disinfection of casts and models is also an ideal practice to prevent the cross-contamination. However, the casts that were once properly disinfected can be re-contaminated during subsequent clinical and laboratory procedures. Many methods have been experimented for disinfecting the casts and models. These methods include spraying of disinfectant on the cast, immersion of cast in disinfectant solution, incorporation of disinfectants into dental casts[35] and also by using microwave oven [36].

Various studies reported that the immersion of cast in 0.525% NaOCl show no adverse effect on dimensional accuracy, surface detail reproduction and compressive strength of casts. Microwave irradiation is also an indicated method for cast disinfection [37]. However, autoclaving of the dental cast may results in reproduction of poor surface details, and immersion of the cast in chemical disinfectant dissolves gypsum, thereby decreasing the compressive strength. Microwave oven disinfection is therefore a potentially convenient solution [38]. In an in-vitro study, it was shown that the microwave irradiation significantly decreased the strength of dental casts after one hour of pouring, and did not show much effect after 24 hours. Therefore, it is worth waiting for 24 hours when using microwave irradiation for disinfection gypsum casts [39].

8. Recent trends in disinfecting impression materials

Recently, attempts have been made to incorporate antimicrobial agents into impression materials to avoid the conventional disinfection procedures as mentioned previously [7, 40-43]. Though these efforts were proved to be efficient in preventing the cross-contamination, an increased risk of dermal and mucosal irritation was observed in the patients when they were repeatedly exposed to certain compounds present in these impression materials [44].

Numerous experiments were performed on the alginate impression material as it is inexpensive and most commonly used to make different oral appliances. The disinfectant material added to the alginate must be effective with no adverse effects on the properties, accuracy and stability of the impression. Water solu-

Impression Material	Disinfecting Agent
Alginate	Iodophors and diluted sodium hypochlorite
Compound	Iodophors and diluted sodium hypochlorite
Polyether	Iodophors and diluted sodium hypochlorite, complex phenolics
Polysulphide	Iodophors and diluted sodium hypochlorite, complex phenolics
Silicone	Iodophors and diluted sodium hypochlorite, complex phenolics
Agar	Iodophors and diluted sodium hypochlorite
Zinc-oxide Eugenol Paste	Iodophors

Table 2. Impression materials and disinfection choices.

ble antimicrobial compounds such as quaternary ammonium compounds, bisguanidine compounds, quinoline compounds, dialkyl quaternary compounds, didecyldimethyl ammonium chloride, substituted phenols, chlorhexidine, and combination of these materials are typically employed [7]. Alginate microcapsules have been made by either blending the disinfectants physically or by coating onto the alginate powder. The disinfectant agent would be released on mixing these microcapsules with the liquid.

Similarly, disinfectants can also be added to the mixing liquid. Among these, CHX was the most widely used and efficient disinfectant without affecting properties and handling of alginates [45,46]. The addition of various concentrations of NaF to alginate powder produced a significant reduction of contamination with no significant effect on dimensional stability and details. In addition, the tear strength was significantly increased. However, it was also reported that the addition of NaF decreased the wettability of the impression. Among these methods, NaF solution is considered as a suitable disinfectant liquid for mixing with alginate impression material, as it did not significantly affect the properties of the material. It was shown that the properties of alginate were altered greatly at the higher concentrations of NaF. The optimum concentration of 2.0% of NaF was suggested to incorporate as it shown minimum effect on the properties of alginate impression material [47].

Numerous researchers have developed self-disinfecting impression materials by incorporating different antimicrobial nanoparticles in to impression materials. Numerous studies have reported that the addition of nanosilver is more effective against *S. aureus, Lactobacillus acidophilus*, *Actinomyces viscosus*, and *Pseudomonas aeruginosa* [41, 42, 48]. Particle size and concentration of the silver nanoparticles in alginate impression materials plays a significant role on antimicrobial activity. It was suggested that the silver nanoparticles with the average particle size of 80–100 nm impart superior antimicrobial property to the alginate hydrocolloid in a concentration-dependent manner than the finer nanoparticle size [41,42]. It was also reported that the addition of greater than 1.0wt% of silver nanoparticles affected the flow, gelation time and strength of alginate impression materials [41,42]. Antimicrobial efficacy of Zinc oxide and Copper oxide nanoparticles was also experimented by numerous researchers and these nanoparticles were also proved to be effective self-disinfecting agents for alginate impression materials with no significant negative effect on physical and mechanical properties [49].

9. Conclusion

Infection control is very important aspect in prevention of cross infection and safety of patients, Dentists and dental personnel. Impression disinfection can prevent spread of infection from dental clinic to dental laboratory technician, other patients and dental auxiliaries. It is the responsibility of the dentist to make appropriate choice of disinfection method for different impression materials.

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Surface Modifications of Dental Implants: An Overview

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ABSTRACT

Osseointegration is the key for long term success of endosseous dental implants. Implant surface properties like roughness, topography, energy and composition are the major surface features that influence the process of osseointegration. Several methods have been used to optimize implant surface roughness to increase surface area thereby improving the process of osseointegration. Blasting using alumina and titanium dioxide, acid treatment, anodization, and laser peeling are some of the subtractive methods used to optimize implant surface roughness. Additive methods, on the other hand, are used to coat HA onto the surface of endosseous implants and these include plasma sprayed HA, vacuum deposition technique, sol-gel and dip coating method, electrolytic process and nano-HA coating. Recently, biomimetic implant surfaces are being produced with calcium phosphate coatings under physiological conditions. These coatings may also act as vehicles for osteogenic agents like BMPs, GDFs and biologically active drugs like bisphosphonates, gentamicin, tetracycline, etc. Methods used for surface modifications of endosseous dental implants are vast and continuously evolving with the recently developed technologies. This article gives an overview of various surface modifications and current trends followed in the field oral implantology.

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1. Introduction

An implant is a biomaterial or a medical device, placed intentionally into human body either totally or partially buried beneath the epithelial surface [1]. Osseointegration of implants is a series of events leading to direct contact of living bone to the implant surface. This determines the ultimate success of endosseous implants at the tissue implant interface. Osseointegration process is affected by surface characteristics of implant such as roughness, topography, energy and composition [2,3,4]. Surface modification of implants is essential for seeking ideal osseointegration. Several methods are adopted with an aim of optimising the roughness and topography of endosseous implant surfaces in order to maximize the osseointegration. These endosseous implants have various surface characteristics ranging from machined/turned surfaces to more roughened surfaces by means of "blasting, acid etching, coating of ceramic particles or combination of various techniques" [5]. The goal of all the implant surface modifications is to achieve fine and rapid osseointegration [6]. Successful osseointegration is associated with osteoinduction, osteoconduction and osteogenesis [7,8]. Osteoconductive implant surfaces provide favorable environment for the bone to grow by acting as a scaffold. Osteoconductive surfaces require existing bone or differentiated mesenchymal cells for enhancing the new bone formation. Hydroxyapatite, calcium phosphate coatings on the implant surface are osteoconductive in nature. Osteoinductive surfaces enhance or induce bone regeneration from existing bone with the help of either bone morphogenic proteins (BMPs), growth factors and/or collagen- chitosan polymers. The aim of this review is to provide an overview of numerous surface modification techniques and provide an insight of the current trends followed in the field of oral implantology to improve osseointegration.

2. Methods of implant surface modification

Implant surface modifications are categorised into subtractive and additive methods. Various surface modification methods were enumerated in table 1.

2.1. Subtractive Methods

These methods involve in creating irregularities on the surfaces of endosseous implants.

2.1.1. Sand blasting

The objective of sand blasting is to improve surface roughness of the dental implants. This would increase the surface area of the implants that results in effective osseointegration. Sand blasting is done by subjecting the implants to various gritting agents like alumina (Al_2O_3) and titanium oxide (TiO_2) at high pressure. The efficacy of sandblasting depends on the number and size of the particles as well as the pressure and the speed of rotations to which implant is submitted [8]. It is simple and has no additional interface between implant and the surrounding bone. Blasting procedures leave residual particles on the surface of the implant, and this could modify the bone-healing process. Any sharp edges formed on the surface of implant can adversely affect the adhesion of bone onto the implant. It was observed that the sand blasted surfaces allow differentiation and proliferation of osteoblasts [9,10]. Numerous studies showed that adhesion of fibroblasts to blasted surface was difficult [11,12]. Wennerberg A *et al* (1995) reported that greater removal torque force was needed for Titanium dioxide blasted implants (35.4 N-cm) compared to titanium turned Implants (29.2 N-cm) placed in the tibia of rabbit [13]. This suggests that the blasted implant surfaces enhance proper bone formation around them compared to turned implants. Similarly, Piatelli A *et al* (1998) observed higher bone to implant contact (BIC) around the blasted implants compared to turned implants in femoral knee joints of rabbits [14].

Calcium phosphate in the form of hydroxyapatite (HA), beta tricalcium phosphate were also used as blasting media, and these are called as resorbable blast media (RBM). The base of titanium is submitted to blasting, followed by a passivity procedure to remove the calcium phosphate ($CaPO_4$) and finally, cleaning. The blast media is resorbed during these processes, and a surface of pure TiO_3 is produced that is free of contaminants [6].

2.1.2. Acid etching

This process was proposed to modify implant surface without leaving the residues found after sandblasting and for uniform treatment of surface of the implant. Etching of implant surface is performed using hydrochloric acid (HCl), sulphuric acid (H_2SO_4), hydrofluoric acid (HF), nitric acid (HNO_3) individually or in combinations. The acid mixture, etching time, temperature of bath affects the rate of etching process.

Subtractive methods	Additive methods	Current trends

• Sand blasting
• Acid etching
• SLA surfaces
• Anodization
• Laser peening
	• HA Coating techniques
	  • Plasma spraying
	  • Vacuum deposition
	  • Sol gel and dip coating method
	  • Electrolytic process
	  • Frit enamelling
	  • Hot isotonic pressing
	  • High velocity oxygen fuel spraying
	  • Ion associated deposition
	  • Sputter coating
		• Photofuctionalization
		• Biomimetic calcium phosphate coatings
		• Coating of osteogenic agents like BMPs, growth factors
		• Coating of bioactive drugs like bisphoshonates, gentamycin
		• Nano titania, nano HA coatings

Table 1. Various implant surface modification methods

It was reported that the implants etched with HCl and H_2SO_4 exhibited greater removal torque force (20.5 N-cm) than the turned implants (4.95 N-cm) after 2 months healing period in the femurs of rabbit [15]. Studies also reported that dual etching, specifically the combination of HF and HCl, is more beneficial in creating the rough surfaces on implants [16]. Chi SA *et al* (2003) implanted dual etched and turned implants into the tibia of rabbit and observed BIC after 12 months of placement. They reported greater BIC with etched implants (62.5%) compared to turned implants (39.5%) [16].

2.1.3. Sand blasted and acid etched surfaces (SLA)

This is a combination method in which the implant surfaces are treated with blasting followed by acid etching process. The standard SLA surface includes sandblasting with large grits of 0.25 to 0.5 mm subsequent etching with HCl/H_2SO_4.The resultant surface was constituted by uniformly scattered gaps and porosities, and it appeared to be slightly less rough than the surfaces subjected to plasma-spraying, which presented a deeply irregular texture that provided a less favorable environment for cell spreading [17].

Several studies compared the effect of SLA surfaces and acid etched surfaces on bone formation around the implants [18,19]. It was reported in the literature that SLA Implants exhibited greater removal torque force (186.8 N-cm) than acid etched implants (95.7 N-cm) three months after their placement in the miniature pigs [18]. In a human study, implants were placed into the palatal bone of the maxilla for orthodontic anchorage and BIC of 76.6% was observed on removal after 6 months [19].

2.1.4. Anodization

In this process, implant surfaces are treated in strong acids like Phosphoric acid (H_3Po_4), H_2SO_4, HNO_3 and HF at high current density (200 A/m^2) or voltage potential (100v). Resultant surfaces produce oxide layer with thickness more than 1,000 nm. Microstructure and crystallinity of titanium dioxide layer were modified in this process [20,21]. These surfaces were compared with turned implant surfaces. Six weeks after placement in the tibia of rabbit, anodized implants showed greater removal torque force (12.9 N -cm) than turned surfaces (7.5 N-cm) [22]. Munhoz *et al* (2015) compared the effect of sand blasted - acid etched (SLA) and anodized implants in the tibiae of rabbits. It was found that SLA surfaces showed greater BIC compared to anodized surfaces[23].

2.1.5. Laser peening

This process uses a high intensity (5-15 GW/cm2) nanosecond pulses (10-30 ns) of laser beam to create small spherical uniform honeycomb pattern of small pores over the surface of implant [6]. Laser peening produces a surface with refined grain structures, compressive residual stresses, and increased hardness in metallic materials. Micro patterns of 20 μm wide and 7 μm deep were imprinted on cpTi biomedical implant material through laser surface treatment that improves the implant surface corrosion resistance, mechanical and fatigue strength. Laser treated implant surfaces showed greater removal torque in compari-

son with turned implant surfaces placed in the rabbit tibia [24,25].

2.2. Additive Methods

Additive methods include coating of implants with calcium-based compounds. During the process of osseointegration, calcium phosphate released from HA coated implant surfaces into peri-implant region, increases the saturation of tissue fluids and precipitates a biological apatite layer on the implant surface. This layer contains endogenous proteins and acts as a matrix for osteogenic cell growth and attachment [6]. Calcium phosphate coated implants showed better clinical success rate than uncoated titanium implants [26,27]. Different ceramic materials used to coat calcium phosphate onto implant surface are HA, tricalcium phosphate, fluorapatite, calcium pyrophosphate, brusite and bioglasses. Various methods used to coat HA onto the surface of endosseous implants include plasma sprayed HA, vacuum deposition technique, sol-gel and dip coating method, electrolytic process, hot isotonic pressing, high velocity oxygen fuel spraying, frit enamelling, ion associated deposition, sputter coating and nano-HA coating [28].

2.2.1. Plasma Sprayed HA

In this process, powdered crystalline HA is introduced and melted by a hot, high velocity plasma gas and propelled onto the implant surface [29]. HA particles undergo partial melting and produces 50µm thick coating on the surface of implant. The characteristic features of plasma sprayed HA are greater surface area of bone apposition to the implant, enhancement of biomechanics and initial load bearing capacity of the system and increase in the bone penetrations that enhances fixation in areas of limited initial bone contact. Plasma sprayed HA adherence to titanium is purely mechanical and numerous studies reported adhesive failure between coatings and implant surfaces [30].

2.2.2. Vacuum deposition technique

Radiofrequency magnetron sputtering, beam sputtering, pulsed laser deposition are the various techniques used for deposition of HA through vacuum deposition [28]. Radio frequency magnetron sputtering is done in a mix of argon and reactive gases to derive a desired HA stoichiometry [31]. In this process, implants are mounted in a radio frequency magnetron sputtering apparatus with base pressure of 10^{-6} mb [28]. It allows very thin, stable, homogeneous coating on implant surface. Shams Mohammadi *et al*(2004) investigated long term bone response of implants coated with HA using radio frequency magnetron sputtering technique and observed that coated implants showed better long-term bone response and improved bone to implant contact [31].

3. Current trends in implant surface modifications

3.1. Photofunctionalization

This recent technique involves treatment of implant surfaces using UV light to improve physical, mechanical properties and to enhance osseointegration [32]. Photofunctionalization improves the biological effects of titanium implants by converting the implant surface from hydrophobic to hydrophilic and electronegative to electro positive. It removes hydrocarbon layer from the surface that was formed during aging of implants. As a result the attachment, retention, proliferation and expression of fundamental phenotypes of osteoblasts are remarkably increased [32].

Seinichi Suzuki *et al* (2013) evaluated the degree and rate of implant stability development of photofunctionalized implants in humans. About 78.0% implant stability quotient (ISQ) was reported after 6 weeks of implant placement, that was considerably higher than the average ISQ of 66%, reported in the literature. This indicates that photofunctionalization accelerated and enhanced the osseointegration of endosseous dental implants [32].

3.2. Biomimetic Calcium phosphate coatings

Biomimetic method of coating calcium phosphate onto implant surface has gained popularity which was developed by Kokubo *et al* in 1990 [33]. This original technique was refined by several investigators[34-41]. Biomimetic technique involves immersion of pretreated implant into a supersaturated solution of calcium phosphate under physiological conditions of 37^0C temperature and pH of 7.4 [42 – 47]. The advantages of this technique are simplicity, economical, can be applied on heat- sensitive, non-conducive porous materials with complex geometry. This biomimetic calcium phosphate layer is an excellent media to act as a vehicle for osteogenic agents. These agents are precipitated into the latticework of calcium phosphate for their slow and sustained release. BMP, BMP-2, growth and

Implant system	Surface modification
Nobel Biocare, Switzerland	Phosphate enriched titanium oxide
Straumann, Switzerland	SLA surfaces
Osstem, South Korea	Resorbable blast media (RBM) using Calcium phosphate hydroxyappatite
Pitt-easy, Germany	Titanium plasma spray
Biohorizone, Alabama	Laser peening
Adin, Israel	SLA, Resorbable blast media (RBM) using Calcium phosphate

Table 2. Various dental implant systems and their surface modifications.

differentiating factors (GDF) are some of the osteogenic agents that can be incorporated into biomimetic calcium phosphate coatings. Thickness of these coatings varies between 10-50 μm. The osteoinductive nature of BMPs was first reported by Urist in mid 1960s [48]. BMPs act on undifferentiated mesenchymal cells and induce them to differentiate into osteoblasts and chondroblasts[49]. Conventional methods like adsorption [50], binding to biofunctional proteins [51], chemical treatments [52] deposit osteogenic agents superficially on the surface of coatings. These superficial coatings released rapidly upon exposure to biological environment. Thus, osteogenic efficiency of these agents is short lived. Unlike conventional technique, in this biomimetic method osteogenic molecules are incorporated into the lattice of calcium phosphate coatings and are released at a slower and study rate, increasing the bioavailability of the agents for longer period of time [53]. BMPs, GDFs, transforming growth factor (TGF- b) are some of the osteogenic agents incorporated into the lattice of biomimetic calcium phosphate coatings. Various materials used to serve as carrier for BMP -2, including collagen, demineralized bone matrix, synthetic and natural ceramic materials and poly glycolic acid [54-60].

Liu *et al*(2007) conducted a study on incorporating the osteogenic agents onto implant surfaces and concluded that BMP-2 can be incorporated into biomimetic coatings and it retained its biological activity for longer period after implant placement. Another study proved that dental implants coated with BMP-2 incorporated calcium phosphate showed rapid osseointegration in adult miniature pigs than the uncoated implants [61].

Pharmacological agents like bisphosphonates are coated on implant surface to improve bone density in highly cancellous bone[62]. Investigations were carried out to study the affect of recombinant human BMP (rh BMP-2) coating on implant surface in animal models. It was proved that rh BMP-2 promotes initial integration of dental implants [63, 64].

3.2.3. Nano-coatings

Recently, nano-dentistry has focused on the delivery of molecules that promote hard tissue remineralization. In this regard, nano particles have shown a strong influence on the host response at both cellular and tissue levels. This made their use more popular for modifying dental implant surfaces. Various methods have been developed to provide nano textured thin film biocompatible coatings on implant surfaces. They include sol gel method, pulsed laser deposition, electrophoretic deposition, ion beam assisted deposition and sputter coating [65].

Nano-titania and nano HA coatings have gained popularity and are studied extensively among the nano coated materials because of their biocompatibility, increased surface area to volume ratio and especially the composition of HA is similar to that of bone [66]. Various commercially available implant systems and their surface modification methods are given in table2.

4. Conclusion

The goal of modern implantology is rapid peri-implant bone healing which results in rapid osseointegration allowing early implant loading. Surface texture of implants plays a significant role in osseointegration. Numerous surface modification methods are continuously evolving with the development of newer techniques and technologies, which are aiming to make rapid and more predictable osseointegration. Recently, biologically active drugs (bisphosphonates, tetracyclins) and osteogenic agents (BMPs, PDGFs, IGFs) were also coated on implant surfaces for faster healing and to ensure

immediate loading of implants. Clinicians should have sound knowledge on surface modification methods of dental implants for careful and suitable selection of implant system to ensure long term success of implant therapy.

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Casting alloys: The saga of their existence and the recipe of their blend

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ABSTRACT

Though a variety of metals and combinations have been in use since decades in the field of dentistry, there are only a few which have sustained the challenges of evolving material science. The history of these alloys, their constituent metals and properties impart the rationale of their use both in the past and advancing future perspectives. Also bearing the environmental hazards in laboratory and clinical environments, safe levels of exposure to these alloys and aspects of selecting the best option among the different alternatives is important.

1. Introduction

In dentistry, metals represent one of the four major classes of materials used for the reconstruction of damaged or missing oral tissues while the others being ceramics, polymers and composites [1]. But the question is whether we really use pure metals in dentistry or not. With the exceptions of pure gold foil, commercially pure titanium and endodontic silver points, all the other metallic substances used in dentistry are alloys [1]. The reason for this occurrence is that pure metals fail to meet the desirable properties individually [2]. So, alloys provided the best means to alter the properties of individual metals into desirable characteristics. (Note: The word 'metal' is used as a generalized term for metals and alloys in the field of dentistry). In spite of the advancements in ceramics and increasing demand of esthetics over durability, the role of alloys in dentistry still remains prominent because of their extraordinary combination of strength, wear resistance, biologic compatibility and long-term survival [3].

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Metals form the building blocks of an alloy in innumerable combinations and proportions but those suitable for intraoral biocompatibility are limited and are enlisted in Table 1.

2. History of metals in dentistry

Dentistry as a specialty is believed to have begun about 3000 BC. Gold bands and wires were used by the Phoenicians after 2500 BC [4]. Modern dentistry began in 1728 when Fauchard published different treatment modalities describing many types of dental restorations, including a method for the construction of artificial dentures made from ivory and metals. Gold shell crowns were described by **Mouton** in 1746 but they were not patented until in 1873 by **Beers**. In 1885, **Logan** patented porcelain fused to platinum post replacing the unsatisfactory wooden post previously used to build up intra-radicular areas of teeth [5]. The other important events in the evolution of dentistry are given Table 2.

3. Ideal requirements of alloys in dentistry [1, 6-8]

The dental casting alloys should possess certain requirements in order to use them in the oral cavity for various purposes. The ideal requirements of various dental casting alloys are discussed in Table 3.

4. Alloy constituents

Alloy is defined as a mixture of two or more metals or metalloids that are mutually soluble in the molten state; distinguished as binary, ternary, quaternary, etc. which again based on the miscibility is divided as solid solution, eutectic, peritectic and intermetallic types. Several alloys have been used in dentistry out of which eutectic and peritectic alloys are utilized maximum. But these metals cannot be mixed into one another in the conventional process, but require heat treatments in order to formulate and obtain the desired properties of the alloys [1].

4.1. Heat treatment of gold alloys [1, 8 –10]

Heat treatment of alloys is done in order to alter its mechanical properties. Gold alloys can be heat treated if it contains sufficient amount of copper. Only type III and type IV gold alloys can be heat-treated. There are two types of heat treatment such as softening heat treatment (solution heat treatment) and hardening heat treatment (age hardening).

Softening heat treatment increased ductility, but reduces tensile strength, proportional limit, and hardness [8]. It is indicated for appliances that are to be grounded, shaped, or otherwise cold worked in or outside the mouth. In this process, the casting is placed in an electric furnace for 10 minutes at a temperature of 700^0C and then it is quenched in water. During this period, all intermediate phases are presumably changed to a disordered solid solution, and the rapid quenching prevents ordering from occurring during cooling. Each alloy has its optimum temperature [8]. The manufacturer should specify the most favorable temperature and time.

Hardening heat treatment increases strength, proportional limit, and hardness, but decreases ductility. It is the copper present in gold alloys, which helps in the age hardening process [9]. It is indicated for metallic partial dentures, saddles, bridges and other similar structures. It is not employed for smaller structures such as inlays.

Hardening het treatment is done by “soaking” or ageing the casting at a specific temperature for a definite time, usually 15 to 30 minutes. It is then water quenched or cooled slowly. The aging temperature depends on the alloy composition but is generally between 200^0C and 450^0C. During this period, the intermediate phases are changed to an ordered solid solution. The proper time and temperature for age hardening an alloy are specified by the manufacturer.

Ideally, before age hardening an alloy, it should first be subjected to a softening heat treatment to relieve all strain hardening & to start the hardening treatment when the alloy is in a disordered solid solution [10]. This allows better control of the hardening process.

5. Classification

Dental casting alloys are classified in to various types and they are discussed in the subsequent sections from 5.1. to 5.3.

Metal	Metal type	Density (g/cm^3)	Melting point (^0C)	Boiling point (^0C)	CTE* ($x10^{-6}/^0C$)	Functions
Gold (Au)	Noble metal	19.3	1063	2970	14.2	Imparts ductility, improves tarnish and corrosion resistance
Silver (Ag)	Precious metal (not considered noble anymore)	10.4	961	2216	19.7	Strengthener , hardener, whitens the alloy, increases CTE
Palladium (Pd)	Noble metal	12.02	1552	3980	11.8	Hardener, tarnish resistance, whitens the alloy, increases fusion temp.
Platinum (Pt)	Noble metal	21.45	1769	4530	8.9	Strengthener, corrosion resistance, whitens the alloy, increases melting temp., grain refiner
Nickel (Ni)	Base metal	8.9	1453	4530	13.3	Hardener, Strengthener, whitens the alloy.
Cobalt (Co)	Base metal	8.85	1495	2900	13.8	Strengthener, Hardener
Chromium (Cr)	Base metal	7.19	1907	2671	6.2	Passivation effect
Copper (Cu)	Base metal	8.96	1083	2595	16.5	Hardener, Increases melting point
Molybdenum (Mo)	Base metal	10.28	2623	4639	4.8	Hardener, Strengthener, refractory material, corrosion resistance, grain refiner
Iron (Fe)	Base metal	7.87	1527	3000	12.3	Hardener
Manganese (Mn)	Base metal	1.74	650	1107	25.2	Oxide scavenger, increases CTE, whitens the alloy
Silicon (Si)	Metalloid	2.33	1410	2480	7.3	Hardener

Table 1: Various metals used for alloying. *Coefficient of thermal expansion.

Other noble metals – Iridium (Ir), Ruthenium (Ru), Rhodium (Rh), Osmium (Os) which act as grain refiners are added in small quantities.

Zinc (Zn) is a bluish white metal which acts as a scavenger for oxygen.

Gallium (Ga) used to decrease the co-efficient of thermal expansion is added in moderate amounts in silver free alloys and very minute amounts in silver containing alloys.

Beryllium (Be) though added to decrease fusion temperature and refines grain size poses a occupational hazard to the dental lab technicians.

Non-metals used for alloying: Carbon and boron are the two common non-metals used for alloying. While carbon increases strength, boron decreases the melting temperature, ductility and acts as a de-oxidizer.

Year	Event
1907	Introduction of Lost-Wax Technique
1933	Replacement of Co-Cr for Gold in Removable Partial Dentures
1950	Development of Resin Veneers for Gold Alloys
1959	Introduction of the Porcelain Fused-to-Metal Technique
1968	Palladium-Based Alloys as Alternatives to Gold Alloy
1971	Nickel-Based Alloys as Alternatives to Gold Alloys
1980s	Introduction of All-Ceramic Technologies
1999	Gold Alloys as Alternatives to Palladium-Based Alloys
1971	The Gold Standard: The United States abandoned the gold standard in 1971. Gold then became a commodity freely traded on the open markets. As a result, the price of gold increased steadily over the next nine years. In response to the increasing price of gold, new dental alloys were introduced through the following changes: • In some alloys, gold was replaced with palladium. • In other alloys, palladium eliminated gold entirely. • Base metal alloys with nickel as the major element eliminated the exclusive need for noble metals.
1976	Dental alloys were classified as passive implants by the Medical and Dental devices act.
1980	Introduction of All-ceramic technologies.
1996	The European Medical Devices Directive established the imposition of CE mark on any imports of dental devices.
1998	Gold alloys as alternatives to Palladium based alloys due to the Clean Air acts.

Table 2: Important historical events in the evolution of alloys in dentistry

5.1. ADA specification No. 5 classified dental gold casting alloys as [11-13]

- High Gold Alloys
  Type I – Soft – inlays, Class III and V restorations
  Type II – Medium – thick 3/4th crowns, pontics etc.
  Type III – Hard – full crowns, short span FPDs etc.
  Type IV – Extra hard – long span FPDs, partial denture frameworks etc.
- Low Gold Alloys
- White Gold Alloys

The hardness, proportionality limit and strength increase from Type I to Type IV whereas ductility and corrosion resistance decreases in the similar order.

5.2. On the basis of nobility the casting alloys have been divided into

- High Noble metal alloys – require at least 60 weight percentage (wt%) and a gold content of at least 40%.
- Noble metal alloys – require noble metal content at least 25 wt% (no stipulation for gold).
- Predominantly Base metal alloys – require noble metal content less than 25 wt%.
- Titanium and Titanium alloys – require at least 85 wt% of titanium content.

5.3. On the basis of their application

Two types of alloys based on their applications in dentistry. They include All-metal alloys and metal-ceramic alloys. Metal-Ceramic alloys have paved the gateway of esthetics along with the durability of that of the alloyed metals. The main function of metal-ceramic alloys is to reinforce porcelain, thus increasing its resistance to fracture.

Properties	Ideal Requirement
Biological	• Should be biocompatible. • Should not contain allergenic components. • Should not be irritant to the tissues. • Should not be toxic to the tissues.
Chemical	• Should be tarnish resistant. • Should be corrosion resistant. • Should be able to bond with porcelain. • Should have the ability to be etched.

Physical	• Should have enough strength.
	• Should be hard enough to facilitate
	  • Finishing and polishing and yet have sufficient
	  • Wear resistance
	• Should have optimum ductility in cases of RPD frameworks.
	• Should be malleable enough to withstand the compressive forces.

Thermal	• Should have a narrow melting range. Solidus temperature: Temperature at which an alloy becomes solid on cooling or at which the metal begins to melt on heating. Liquidus temperature: Temperature at which an alloy begins to freeze on cooling or at which the metal is completely molten on heating. • Should have good castability. • Should be easy for brazing and soldering. • Should have less solidification shrinkage. • Should have enough sag resistance. • Should have co-efficient of thermal expansion closer to porcelains.
Miscellaneous	• Should be inexpensive. • Should not discolor the overlying porcelain.

Table 3. Ideal requirements of alloys in dentistry

The alloys used for metal-ceramic purposes are grouped under two categories including noble metal alloys and base metal alloys. In case of noble metal alloys for porcelain bonding, addition of 1% base metals (iron, indium, tin etc.) increases porcelain-metal bond strength, which is due to formation of an oxide film on its surface. It also increases strength and proportional limit. The different metallic combinations available as per nobility are summarized in Table 4. [14, 15].

6.1. Noble metal alloys

The desired properties of noble metal alloys determine the selection of elements that constitute the alloys. These were the first alloys to be brought into usage in the field of dentistry. The reason for their selection may be due to their availability and their inertness towards the oral environment. Listed below are a few of the commonly used noble metal alloys-

6.2. Gold-Platinum-Palladium (Au-Pt-Pd) System

Alloy type	All-metal	Metal-ceramic	Removable Partial Denture alloys
High Noble metal	Au-Ag-Cu-Pd Au-Ag-Pd High noble metal ceramic alloys	Pure Au (99.7 wt%) Au-Pt-Pd Au-Pd-Ag (both 5-12wt% and >12wt% Ag) Au-Pd	Au-Ag-Cu-Pd
Noble metal	Ag-Pd-Au-Cu Ag-Pd Noble metal ceramic alloys	Pd-Au Pd-Au-Ag Pd-Ag Pd-Cu-Ga Pd-Ga-Ag	
Predominantly Base metal	CP Ti Ti-Al-V Ni-Cr-Mo-Be Ni-Cr-Mo Co-Cr-Mo Co-Cr-W Cu-Al	CP Ti Ti-Al-V Ni-Cr-Mo-Be Ni-Cr-Mo Co-Cr-Mo Co-Cr-W	CP Ti Ti-Al-V Ni-Cr-Mo-Be Ni-Cr-Mo Co-Cr-Mo Co-Cr-W

Table 4. Different alloy types available to suit the purpose classified according to nobility

These alloys were developed in an attempt to overcome the major limitations in the gold-platinum-palladium system. Poor sag resistance, low hardness & high cost, two variations on the basic combination of gold, palladium and silver were created and are identified as the either the high-silver.

The high silver alloy contains Gold (39% to 53%), Palladium (25% to 35%) and Silver (12% to 22%) including trace amounts of oxidizable elements are added for porcelain bonding. While The low silver alloys comprise increased Gold (52% to 77%) content and decreased Silver (5% to 12%) and Palladium (10% to 33%) content along with trace amounts of oxidizable elements for porcelain bonding.

These alloys improved the rigidity and sag resistance while maintaining the high nobility, co-efficient of thermal expansion and corrosion resistance while managing to keep the cost relatively cheaper than that of the Au-Pt-Pd alloys. These expensive alloys have high silver content that creates potential for porcelain discoloration.

6.3. Palladium-Silver (Pd-Ag) System

This was the first gold free system to be introduced in the United States (1974) that still contained a noble metal (palladium) [18]. It was offered as an economical alternative to the more expensive gold-platinum-silver and gold-palladium-silver (gold based) metals.

The low silver containing group consist of Palladium (55% to 60%), Silver (25% to 30%) and Indium and Tin in trace amounts while the high silver group is made of Palladium (50% to 55%), Silver (35% to 40%) with Tin, little or no Indium and trace elements of other oxidizable base elements.

These less dense alloys exhibit good tarnish and corrosion resistance in spite of the moderate nobility and also show good porcelain bonding and excellent sag resistance. These are suitable for long span fixed partial dentures and are cost effective. The alloys are prone to discolor the porcelain and have low hardness. They adsorb gases and hence should not be cast in carbon crucible.

6. 4. Gold-Palladium (Au-Pd) System

This particular system was developed in an attempt to overcome the major limitations in the gold - platinum -

silver system and Pd-Ag alloys namely; Porcelain discoloration and too high coefficient of thermal expansion & contraction.

These have decreased gold but increased palladium content. They contain indium for bonding, gallium and (Ga) to decrease the fusion temperature. These alloys are stronger, stiffer, and harder than the Au-Pt-Pd alloys and have higher elongation (more ductile) and casting temperatures (easier to solder).

These alloys are composed of Gold (44% to 55%), Palladium (35% to 45%), Gallium (5%), Indium & Tin (8% to 12%) and traces of grain refining elements. The alloys exhibit excellent castability, good bond strength and corrosion resistance, improved hardness and strength but have low density. These are not thermally compatible with high expansion dental porcelains and are expensive.

6. 5. High Palladium System

Several types of high palladium were originally introduced (Tuccillo, 1987) [19]. More popular composition, group containing cobalt and the other containing copper. Copper appears to be more popular. These white alloys have high strength and hardness, good stiffness and elongation, and low density. However, they have low sag resistance and form dark oxides.

The cobalt containing alloy comprises of Palladium (78% to 88%), Cobalt (4% to 10%) with gold (2%) added in some alloys and trace amounts of oxidizable elements (such as gallium and indium) are added for porcelain bonding. On the other hand, the copper containing alloy contains Palladium (70% to 80%), Copper (9% to 15%) and trace amounts of the oxidizable elements gallium, indium and tin are added for porcelain bonding with Gold (1% to 2%) and Platinum (1%) added sometimes.

The alloys of this system show good castability, sag resistance, corrosion resistance, high hardness and low density. On the other side, the alloys adsorb gases forming thick oxides and hence pose a problem of discoloring the porcelain.

6.6. Recycling Noble Metal Casting Alloy [20]

The alloy scrap should be recycled because of the high value of the precious metals. It can be collected and sent back to the manufacturer of it can be recast. These alloys are stable so it can be recast two or three times without much change in its composition. However, the more volatile base metals like zinc, indium, tin and iron may be lost. To compensate for this equal amount of new alloy should be added to the scrap during recasting. They should be carefully cleaned before reuse. Alloys of different types and manufacturers should not be mixed as it may change its composition and properties.

7. Base metal alloys

Base metal alloys have been the other group of alloys extensively instituted in dentistry. As these alloys are very reactive in nature, they form good chemical bond with porcelain and also have the advantage of good mechanical properties when used in specified proportions [21]. These are classified as nickel based, cobalt based, titanium and titanium based alloys. Alloys in both systems contain chromium as the second largest constituent.

7. 1. Nickel-chromium (Ni-Cr) System

These metal-ceramic alloys offer such economy that they are also used for complete crown and all metal fixed partial dentures [22]. The major constituents are nickel and chromium, with a wide array of minor alloying elements. The composition of two types of Ni-Cr alloys are shown in Figures 1 and 2. The system contains two major groups [22], those that contain Beryllium, and contain no Beryllium/Beryllium free. Among the two, Ni-Cr-Beryllium alloy are generally regarded as possessing superior properties and have been more popular [23].

Nickel-Chromium-Beryllium alloy is composed of 62 - 82% Nickel, 11-20% Chromium, 2% of Beryllium and traces of minor alloying elements like, aluminium, carbon, iron, molybdenum etc. while the Beryllium free alloy has similar proportions of Nickel and Chromium without Beryllium and traces of Boron, Molybdenum, Niobium, Tantalum etc. This alloy system have low density and high sag resistance. The major drawback of the alloys is the oxide formation on the surface that causes bond failure between the alloy and the porcelain. The Beryllium containing alloy is more easily castable than the Beryllium free one.

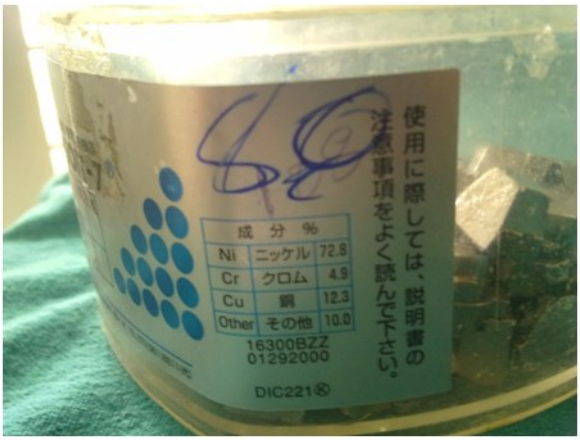

Figure 1. Ni-Cr pellets

Figure 2. Be-free Ni-Cr alloy

7. 2. Cobalt-Chromium alloys

Cobalt chromium alloys have been available since the 1920's. They possess high strength. Their excellent corrosion resistance especially at high temperatures makes them useful for a number of applications. These alloys are also known as 'satellite' because they maintained their shiny, star-like appearance under different conditions. They have bright lustrous, hard, strong and non-tarnishing qualities [24]. The applications find a wide range from partial denture frameworks to surgical implants. These alloys show similar properties as that of the Nickel-Chromium alloys.

The alloy contains Cobalt (55 to 65%), Chromium (23 to 30%), Nickel (0 to 20%), Molybdenum (0 to 7%), Iron (0 to 5%), Carbon (up to 0.4%) and Tungsten, Manganese, Silicon and Platinum in traces. According to A.D.A specification No. 14 a minimum of 85% by weight of chromium, cobalt, and nickel is required and classified based on fusion temperatures as type I – High fusing and type II – Low fusing.

7. 3. Titanium and Titanium alloys

Titanium is called "material of choice" in dentistry [25]. This is attributed to the oxide formation property which forms basis for corrosion resistance and biocompatibility of this material. The term '*titanium'* is used for all types of pure and alloyed titanium. Commercially pure titanium is used for dental implants, surface coatings, crowns, partial dentures, complete dentures and orthodontic wires [26]. The following are the properties of titanium.

- Resistance to electrochemical degradation
- Begins biological response
- Relatively light weight
- Low density (4.5 g/cm^3)
- Low modulus (100 GPa)
- High strength (yield strength = 170-480 MPa; ultimate strength = 240-550 MPa)
- Passivity
- Low coefficient of thermal expansion (8.5 x 10–6/°C)
- Melting & boiling point of 1668°C & 3260°C

7.3.1. Commercially Pure Titanium (CpTi)

It is available in four grades (according to American Society for Testing and Materials ASTM) which vary according to the oxygen (0.18-0.40 wt.%), iron (0.20-0.50 wt%) and other impurities. It has got an alpha phase structure at room temperature and converts to beta phase structure at 883°C which is stronger but brittle [27].

7.3.2. Titanium alloys

Alloying elements are added to stabilize alpha or the beta phase by changing beta transformation temperature e.g. in Ti-6Al-4V, Aluminum is an alpha stabilizer whereas vanadium as well as copper and palladium are beta stabilizer. Alpha titanium is weld able but difficult to work with at room temperature. Beta titanium is malleable at room temperature and is used in orthodontics, but is difficult to weld [28]. Pure titanium is used to cast crowns, partial denture, and complete denture.

7.3.3. Cast titanium

Cast titanium has been used for more than 50 years, and it has been recently that precision casting can be

obtained from it. The two most important factors in casting titanium-based materials are its high melting point (1668°C) and chemical reactivity [27]. Because of the high melting point, special melting procedures, cooling cycles, mold materials, and casting equipments are required to prevent metal contamination, because it readily reacts with hydrogen, oxygen and nitrogen at temperatures greater than 600°C. So casting is done in a vacuum or inert gas atmosphere. The investment materials such as phosphate bonded silica and phosphate investment material with added trace metal are used. It has been shown that magnesium-based investment cause internal porosity in casting. Because of its low density, it is difficult to cast in centrifugal casting machine. So advanced casting machine combining centrifugal, vacuum, pressure and gravity casting with electric arc melting technology have been developed.

7.3.4. Difficulties in casting Titanium

- High melting point
- High reactivity
- Low casting efficiency
- Inadequate expansion of investment
- Casting porosity
- Difficulty in finishing
- Difficulty in welding
- Require expensive equipment

A comparative assessment of the properties of different alloys mentioned in the article is illustrated in Table 4.

Properties	Noble metal alloys	Base metal alloys		Titanium
		Co-Cr	Ni-Cr-Be	
Biocompatibility	Excellent	Excellent	Fair	Excellent
Density	14 g/cm^3	7.5 g/cm^3	8.7 g/cm^3	4.5 g/cm^3
Modulus of elasticity	90 GPa	145-220 GPa	207 GPa	103 GPa
Sag Resistance	Poor to excellent	Excellent	Excellent	Good
Technique Sensitivity	Minimal	Moderately high	Moderately high	Extremely high
Bond to porcelain	Excellent	Fair	Good to excellent	Fair
Cost	High	Low	Low	Low

Table 5. Comparative properties of High noble alloys and Base metals for Metal-Ceramic prosthesis

8. Biological hazards

The potential hazards caused by casting alloys is due to beryllium and nickel content in the alloy. Other elements which elicit a irritant response are Palladium and Chromium. The dental laboratory technicians and patients are the two major groups at risk due to these hazards. Berylliosis and Contact dermatitis are caused by exposure to beryllium and nickel respectively [29]. Moffa *et al* (1973) reported that high levels of beryllium were accumulating during finishing and polishing when a local exhaust system was not used [30]. When an exhaust system was used, the concentration of beryllium in the breathing zone was reduced to levels considered safe by the authors. In 1982, Moffa *et al* reported that no correlation was found between the incidence of nickel sensitivity and the presence of intraoral nickel alloy prostheses [31].

Vreeburg *et al* (1984) concluded that the oral exposure of nickel and chromium to guinea pigs via a fixed appliance or the dietary intake of these elements as metallic powder or salts did not induce an allergic reaction to these metals [32]. Even more significant is that subsequent attempts to elicit an allergic response in previously exposed animals generally failed, whereas unexposed animals exhibited a higher incidence of hypersensitive responses.

8.1. OSHA (Occupational Safety and Health Administration) recommendations

Exposure to beryllium dust in air should be limited to a particulate beryllium concentration of 2 µg/m3 of air (both respirable and nonrespirable particles) determined from an 8 h time-weighted average [33]. The

existing OSHA standard specifies an 8-h time-weighted average (TWA) concentration limit of 1000 μg/m^3 or 1 mg/m^3 of nickel and nickel compounds.

8.2. NIOSH (National Institute for Occupational Safety and Health) recommendations

A limit of 0.5 μg/m^3 based on a 130-min sample in case of beryllium exposure and a standard to limit employee exposure to inorganic nickel in the laboratory or office to 15 μg/m^3 (air), determined as a TWA concentration for up to a 10-h work shift (40-h work week) is recommended [34].

9. Guidelines for selection of an alloy

Practitioner should select the suitable alloy with consultation from the dental laboratory with primary concern towards the systemic health of the patient i.e. regarding any allergies. Next in order follow the desired physical, chemical and thermal properties which will influence the fit and durability of the restoration along with the esthetic and cost factors that finally decipher the appropriate alloy option [35].

10. Conclusion

Each alloy has specific physical and mechanical properties that affect its manipulation and application. Tooth preparation and restoration design will determine the required physical and mechanical properties of the alloy, so all factors should be kept in mind during selection of an alloy. Other technologies are currently available to avoid the challenges and cost associated with metal casting process.

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EAER: An Alternative and Effective Remineralization non-invasive method in Restoration of carious tooth

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ABSTRACT

This article deals with the newer technologies that are used in the remineralization procedure of tooth thus avoiding usage of drills on the tooth structure and preserving the enamel as a whole. A number of finer and newer technologies are being developed so that procedures can be more friendly and acceptable to both the dentist as well as the patient, these newer strategies are also concerned in preserving the tooth structure as a whole without the introduction off any foreign material, thus increasing the life as well as the strength of the tooth structure. This paper gives an ephemeral information on various newer strategies of remineralization with more emphasis on the most recently introduced remineralization technique such as electrically accelerated and enhanced remineralization technique (EAER).

1. Introduction

Dental caries is the one of the most common dental issues among the various populations irrespective of age, gender and geographic regions. The treatment available is simple cost effective yet many people neglect it due to dental phobia and lack of dental treatment awareness. The deep hidden fear for the drills and injections has led to ignore the treatment by the patients. Various dentists have revealed that in order to control the anxiety of the patients, they sometimes have to advice sedation [1]. However, this treatment does not only remove infected part but also part of the healthy tooth structure in order to create room for restorative material and also to provide proper retention with it. Hence, it is necessary to explore an alternative treatment modality which should eliminate the phobia among the patients, less technique sensitive. In addition, that should also prevent the recurrent caries and strengthen the natural tooth. This article provide an overview of the various newer techniques developed on remineralisation procedures and gives more

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emphasis on mechanism, advantages and disadvantages of Electrically accelerated and enhanced remineralisation (EAER) method, so that this newer technique could be made more familiar to practitioners.

2. Pathophysiology of cavity formation

Dental cavities are not mere a stationary process of tooth loss but rather it's a dynamic process; there is constant conflict between the demineralization process and remineralization process. Various acids are produced by the microbial flora during the metabolism of the carbohydrates that results in the drop in pH. This leading to the dissolution of minerals such as calcium and phosphates in their ionic forms from the tooth surface as well as sub-surface leading to demineralization of the tooth and ultimately leading to cavity formation [2]. The various factors that are responsible for demineralisation that ultimately leads to caries formation are cariogenic bacteria such as streptococcus mutants, lactobacillus etc., fermentable carbohydrates such as sugar and the saliva. The remineralisation factors are antibacterial agents, saliva and remineralizing ions [3]. The remineralizing action takes place when minerals like fluoride are introduced in the surface of enamel, where it absorbs and holds minerals and fluoride which are present in the plaque and help in the remineralisation of the surface that is lost due to cariogenic activity [2]. The thorough detailed pathophysiology has allowed us to incorporate a more regenerative procedure rather than just curative. Various regenerative procedures are given in the subsequent sections.

2.1. Dentin phosphoprotein 8DSS peptides

The phosphorprotein is the non-collagenous extracellular element of the dentin [4]. The human dentin phosphoprotein peptides (DPP) have proven to have aspartate-serine—serine nucleotides which help in the production of hydroxy apatite (HA) crystals. The 8DSS peptide have two mineral binding sites that attaches to both free calcium, phosphate ions and even to HA crystals. When these peptides are applied to the enamel surface prevents the dissolution of these ions and further helps in attracting calcium and phosphate ions. Thus, helps in remineralization[5]. This procedure is still in laboratory phase and no commercial products are yet available in market.

2.2. P11-4 peptides

The fibilliar matrix has the ability to attract the calcium channels towards itself. It also acts as a new productive site for HA crystals, these peptides self-assemble in hierarchical 3-dimensional way at the site of lesion in response to high acidic pH of the surrounding. When low viscosity isotropic P11-4 is applied to the enamel surface it transforms into an elastomeric gel, thus leading to restoration of the fibrillar matrix and ultimately enabling the lost tooth structure to remineralize [6].

2.3. Leucine-rich amelogeninpeptides

This technique proves to be more successful in *invitro* studies. The *invitro* studies have proved that the tooth that is treated with amino acids containing leucine rich amelogenin peptide not only reduced the depth of the lesion but also provided the required elements for the growth of enamel peptides, hence helping in remineralization [7]. This amelogenin peptide only contains the C-terminal and N- terminal of the parent protein. These domains are seen to be responsible for the binding and linear growth of the mineral content.

2.4. Poly(amino amine) dendrimers

These are also called as artificial proteins as they mimic the matrix that is required for mineralization. The organic crystal that is formed is known to be formed through poly amino amine dendrimers template has the exact content shape and size as that of intact enamel, in fact the rods even tend to align parallel to the existing HA crystals [8,9].

2.5. Nanohydroxyapatite

The synthetic hydroxapatite have shown to be best in terms of biocompatibility [10], due to its nano size it can be compactly packed, increases the surface area to bind with other particles and also fills the tiny holes. The exact mechanism behind remineralization with the help of nanohydroxyapatitie (nHA) is still unknown, many researchers explain that nHA promotes remineralization by acting as a layer of enamel around the defective area. Other researchers suggest that nHA crystals acts as reservoir of calcium and phosphate ions, thus these ions are present in supersaturated state thus enabling remineralization and suppressing demineralization [11].

2.6. Electrically accelerated and enhanced remineralization

Among the remineralization procedures discussed from section 2.1 to 2.5, electrically accelerated and enhanced remineralization (EAER) has received more attention among the researchers and it was introduced by the researchers at King's college, London. EAER is based on the theory of iontophoresis, which is a widely used and accepted method for transdermal and ocular drug delivery [12, 13]. In iontophoresis, the charged particles get attracted towards the opposite electrode until and unless the electric field is lowered. EAER is a process that aims at enhancing the uptake of the various remineralization factors towards the damaged tooth, which results in shifting of the equilibrium from demineralization state to remineralization state. Therefore, the tooth is protected from decay and is restored well without any drill and fill [1,12].

The mechanism behind the EAER is quite simple and unique. Figure 1 describes the various steps that are involved in restoring caries lesions to the equivalent of healthy enamel using EAER. The tooth surface is cleaned and conditioned, initially. Then, the actual remineralization process takes place by increasing the concentration of all the natural minerals at the site which is to be repaired, with the help of electric field from a custom-made dental device. This helps in enhancing the natural repair of the damaged/infected tooth structure [14].

Pitts NB and Wright JP (2018) experimented with gold nanoparticles instead of remineralizing agents as they are easy to detect and locate the exact location in the lesion. They observed the presence of gold nanoparticles almost into the entire lesion, that is considered as activated sites with lesion and also in areas of organic substances which prevented surface activation.[14]

2.6.1. Advantages of EAER [1]

- Elimination of phobia needles, sounds, smell etc.,
- No loss of any health tooth structure.
- No introduction of foreign materials into the body thus eliminating many allergic and rejection reactions.
- No requirement of refilling of restorations which is generally done to conventional materials after a period of time due to polymerisation shrinkage.
- User friendly for the dentist.

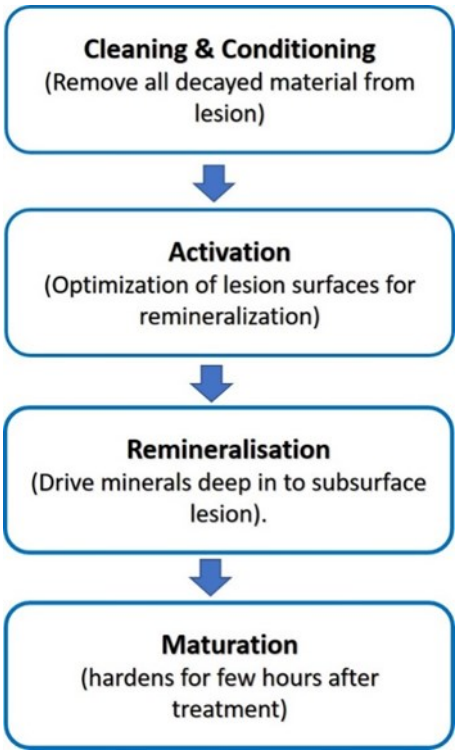

Figure 1. Steps involved in using EAER (Electrically Accelerated and Enhanced Remineralization) to restore caries lesions

- Teeth treated with EAER are stronger than those filled with fillings and even help in preventing further decay.
- Highly esthetic and has positive effect on patients.
- Repair the entire lesion.
- No damage to existing healthy tissues.
- Also helps in whitening the teeth, thus a better option than bleaching.

2.6.2. Limitations of EAER

- No clear evidence if this procedure could be used even in cases of pulpal involvement.
- Since biomimetic materials are to be used then can be more expensive than the conventional dental restorative materials.
- Less evidence of successful clinical trials.

3. Conclusion

The concept of remineralization is one of the most studied and researched topic since the early 20thcentury, but yet even till day we still practice invasive form of dentistry, with developing technology it's time for a shift from invasive form to a non-invasive form of dentistry, EAER is one the most ideal way to practice non invasive methods as it only saves the tooth but also st-

rengths it. The small "healing hand piece" placed on the damaged tooth eliminates the phobia associated with drill among the patients, even the electric current used is minimal, it's even less than that are used for electric pulp testing, thus making it not only just effective but one of the safest techniques. With these newer innovations the future dentistry looks promising and without drills.

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Contents