

Mechanical Properties of Maxillofacial Silicone Elastomer and Approaches for their Improvement: A Literature Review

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Abstract

Silicone elastomer is the most used material for construction of maxillofacial prostheses. However, it has some deficiencies; most important is inadequate mechanical properties which directly affects its clinical behaviour. Considering the functional and psychosocial effects of the failure of maxillofacial prostheses on patients, many research studies have been devoted to developing a maxillofacial silicone elastomer with enhanced mechanical properties. Despite of the advances and improvements that have taken place, there is still an enormous demand for a material with improved physical, mechanical and biological properties. This article aims to review and discuss the mechanical properties of clinical significance of maxillofacial silicone elastomer and to present a comprehensive review of the scientific evidence available in the literature regarding the effect of different approaches employed to improve the mechanical properties of maxillofacial silicone elastomer.

Keywords: *Maxillofacial Silicone Elastomer; Mechanical Properties; Tear Strength; Tensile Strength; Shore a Hardness*

Introduction

Rehabilitation of patients with maxillofacial defects presents a difficult challenge for the reconstructive surgeon, the maxillofacial prosthodontist and the maxillofacial technician. These defects may be congenital or due to trauma or due to disease most commonly cancer. Curative treatment of aggressive cancer lesions requires a multidisciplinary treatment in which surgical resection can be part of. The main goal of surgical treatment is the macroscopic removal of tumor in order to save the patient's life. The primary disease may be cured but a secondary disease is acquired after the surgical resection, called 'Post resection defect'. These defects lead to patients retracting from their families and society and living a life of seclusion and depression [1].

The multidimensional impact of maxillofacial defects on a patient's life has led to an increased interest in improving the quality of life of these patients. Despite the recent advances in maxillofacial surgery, Surgical reconstruction of maxillofacial defects is technically demanding and generally entails several operations to achieve an aesthetically satisfying outcome. Furthermore, the final result may be aesthetically and functionally compromised. Hence, prosthetic rehabilitation is indicated in such cases. Maxillofacial prostheses aim to restore the normal anatomy and appearance, protect the tissues of a defect, and provide improved function. This is of great psychological benefits to the patients by improving their self-esteem and the quality of life which aid in patient re-socialization [2].

Silicone rubber is the most widely used material for construction of maxillofacial prostheses due to its desirable properties as biocompatibility, chemical inertness and durability. But, it has some deficiencies; most important is inadequate mechanical properties which directly affects its clinical behaviour. The clinical success of silicone elastomers for maxillofacial applications depend on their mechanical

properties, because in clinical service they are exposed to a wide range of forces; as continuous weak pressure from the skin and mucosa during resting, pressure from facial movement, and the applied pressure from the outside. Moreover, they are exposed to body fluids and a wide range of temperatures because they contact the atmosphere and the tissues [3].

Taking in to account the imperative role of the material properties in the success of the maxillofacial prosthetic rehabilitation, this paper aimed to review and discuss the mechanical properties of silicone maxillofacial prosthetic material of clinical relevance and the changes that take place in the material characteristics as well as the methods employed for enhancement of its mechanical properties to provide a foundation for future research on development of new methods for reinforcement of silicone maxillofacial elastomer to achieve an improved maxillofacial prosthetic material that can satisfy the patient's growing demands. To the best of the author's knowledge, such a review is lacking in the contemporary literature. The Medline, Embase and Cochrane databases were searched for experimental studies related to mechanical properties of maxillofacial silicone elastomer from 1970 to 2018.

Factors affecting the mechanical behaviour of silicone polymer

Silicone elastomers were introduced for use in maxillofacial applications by Barnhart in 1960. Silicones are chemically termed as poly-dimethyl siloxane (PDMS), they are a class of organo-silicone synthetic material based on molecular chains of alternate silicone and oxygen atoms. It is a combination of organic and inorganic components [4].

Different maxillofacial silicone elastomers vary in their mechanical properties due to differences in the compositional chemical formulations; including chain length of the polymers, type, amount and size of fillers, degree of cross-linking as well as molecular weight [5]. The polymer chain length within the silicone determines the viscosity of the material, as does the addition of fillers or other compounds such as colorants, antioxidants and vulcanizing agents in order to transfer the material from a liquid to a rubber during processing. The cross-linking between the long polymer chains result in a strong inter-linking between the components, therefore, cross-linking is essential to optimize tensile and tear strength and provide the silicone with resistance to heat and degradation during exposure to ultraviolet light (UV) [6]. However, the variation in cross-linker/elastomer ratio affects the mechanical properties of maxillofacial silicone elastomer, where it results in initial increase in mechanical properties which could be due to the formation of more cross-links and increasing cross-link density. After reaching a peak value, no more cross-linking sites are available, and the residual un-reacted cross-linker remains in the silicone, resulting in a decrease in mechanical properties with increasing cross-linker concentration. Thus, increasing the cross-linker/elastomer ratio causes a decline in the tensile strength of the elastomer. This is caused by the low molar mass silicone polymer which acts as a carrier for the actual cross-linking additive. Moreover, there is an increase in the hexane extractable content and a rise in the Si-H content of the elastomer. Clinically, this may result in colonization with micro-organisms as *Candida Albicans* [7].

Therefore, the mechanical behaviour of silicone polymers depends on three factors:

- a- The molecular weight distribution; blending both long and short chains of the same polymer creates a broader bimodal network which combines high tensile strength, tear resistance and elasticity. Tear strength can be optimized at low bimodal concentrations, but it declines at greater concentrations of low molecular weight polymer because at higher concentrations the cross-link network is tightened to reducing the flexibility and more brittle specimens with reduced tear strength are produced [6,8].
- b- Incorporation of surface treated silica fillers into the base polymer allows the polymer chains to uncoil and slide past neighbouring filler particles. This increases the cross-linking between neighbouring PDMS chains and provides the elastomer with a higher strength under deformation [8].
- c- Degree of cross-linking of elastomeric chains is essential in order to optimize tensile and tear strength. The higher the cross-linking, the harder, more brittle elastomeric material is produced, whereas low cross-link density provides lower tensile and tear strength [6].

Mechanical properties of clinical significance

Tear strength

It is the ability of the material to resist transverse forces and relates to the marginal strength and resistance to shearing forces during service. It is also defined as the maximum force needed to break the trouser shaped specimen divided by its thickness [9].

Clinical significance: Clinically it is the most important property to achieve successful aesthetic outcome, marginal integrity and accurate marginal adaptation, which are necessary to blend the facial prosthesis with the tissues of the patient, so very thin edges are needed to merge the prosthesis with the patient's tissues. However, thin edges result in very fragile margins which rupture easily during service. Low tear strength may result in tearing during removal of the prosthesis for maintenance and/or removal of adhesive from the margin for the purpose of cleaning. High tear strength indicates that the prosthesis can be produced in a fine finishing line with the skin to camouflage the edge of the prosthesis [10].

Tensile strength

It is an indication of the material's overall strength in general and its resistance to pulling forces [11].

Clinical significance: This characteristic contributes to the ability of the maxillofacial prostheses to overcome general wear, tear and weathering conditions [12].

Percentage elongation

It is the deformation that results from the application of a tensile force [9]. Ultimate elongation indicates the maximum flexibility of the material.

Clinical significance: It determines the capacity of the material to accommodate the movement of the adjacent soft tissue during function and the resistance of the material to rupture during service and maintenance. A material with a high percentage elongation is particularly advantageous when peeling a nasal or eye prostheses from the facial tissues [12].

Hardness

It is the resistance to indentation under specific conditions. The hardness of an elastomer is the result of two related properties; the inherent, and the processed hardness. This is based on the chemical structure of the material. The inherent hardness can be modified by vulcanization. The hardness of the processed elastomer is determined by the degree of adjusted inherent cross-linking [9]. According to Lewis and Castleberry [13], the ideal values of Shore A hardness should be between 25 and 35 units.

Clinical significance: It is related to the material softness and flexibility. Also, it reflects the tactile response of lifelike feel and it is used to determine how close they are to the consistency of human soft tissues. The ideal hardness is that which mimics the properties of the lost tissue [11,12].

Alterations in the material properties

Environmental factors such as exposure to solar UV radiation, temperature and moisture-the primary factors of weathering- as well as, other factors including air pollutants, cosmetics and strong solvents for cleaning and removal of the adhesive from the prostheses have negative effects on the longevity of prostheses [14].

These factors result in deterioration of the prosthesis and degradation of the polymers in addition to changes to their optical and mechanical properties. The tensile strength and tear strength of the material decline, accompanied by an increase in the stiffness and hardness of the material. The prostheses lose elasticity and become increasingly rigid [15]. The peripheries of the prosthesis that need

to be finished as thin as possible are the most susceptible areas to degradation. Early rigidity is then followed by marginal tear due to daily application and removal of medical adhesive and exposure to the body secretions of the patient. Contact between the skin and the rigid border of the prosthesis can also result in micro-trauma, abrasion and ulceration of the underlying skin at the boundaries of the prosthesis and at the portions where the prosthesis moves against a fixed tissue bed or when the tissue moves against a fixed portion of the prosthesis. This movement usually creates an area of conflict at the elastomer tissue interface, which then leads to irritation of the skin and oral mucosa. This is especially important when a prosthesis contacts compromised tissues such as irradiated tissues [16].

Exposure to sunlight has an undesirable effect on the structural stability of silicone elastomer, where UV radiation results in molecular alteration and increases the material's inherent crystallinity. This may be explained by the formation of new highly flexible chain segments tending to rearrange themselves into higher-order formations, resulting in a chemicrystallization phenomenon, which may be responsible for the increased stiffness of the polymer [17,18]. Hence, solar radiation results in alteration of the mechanical properties and colour changes such as, gloss loss and colour fade. This phenomenon is the result of chemical reaction that occurs when UV radiation, in the presence of oxygen and moisture breaks down the silicone and pigment 'Photo oxidative degradation' within the prosthesis [14], and therefore alters the reflective properties of the material and changes its colour. The energy absorbed by the polymer on exposure to UV radiation results in the formation of free radicals within the polymer by the dissociation of the carbon-hydrogen bonds (C-H bonds) in the polymer chains, the steps are as follows [19]:

Initiation Step $PH \rightarrow P^* + H^*$

Once free radicals form, reaction with oxygen produces hydro-peroxides (POOH).

$P^* + O_2 \rightarrow POO^*$

Propagation Step $POO^* + PH \rightarrow POOH + P^*$

Carbonyl group compounds as Aldehydes and Ketones are formed due to the dissociation of the hydro-peroxides. These decomposition products can be employed as a chemical index for the degradation of the polymer. These free radicals continue to react via propagation reactions. These reactions terminate when of pairs of free radicals react leading to cross-linking.

Termination Step $P^* + P^* \rightarrow P-P$

The unstable free radicals go through chain reactions. This results in the formation of two smaller polymer chains, one of which is a free radical and is able to undergo further reactions and the other free radical may react with another one resulting in terminating the reaction [19]. Thus, the molecular weight distribution is altered due to chain scission and cross-linking. This may explain the different changes in different materials properties for instance; improvement in the tensile strength and decrease in hardness; when cross-linking is dominant the material loses its elasticity and become harder, conversely, the material becomes softer when chain scission is prevalent [20].

Weathering can alter the chemical composition of silicone elastomer. When three different PDMSs; TechSIL 25 (Technovent, England), Cosmesil M511 (Principality Medical, UK) and Elastomer 42 (Technovent, England) were examined and compared to chlorinated polyethylene (CPE) for the mechanical behaviour after outdoor weathering, TechSIL 25 and Elastomer 42 became harder and more brittle which was explained as a consequence of crosslinking, unlike M511 and CPE, which became softer and more ductile due to the domination of chain scission reactions [21]. Silicone elastomers may undergo cross-linking when exposed to high-energy radiation, the amount of cross-linking increase as the radiation dose and duration increase [22]. On exposure of a silica filled poly-dimethyl siloxane-poly-methylphenyl siloxane copolymer to accelerated artificial aging by Cobalt-60 gamma radiation, it was found that the polymer undergoes radiation induced crosslinking reactions which decrease the molecular weight between crosslinks, thus hardening the composite and reducing the motional dynamics of segmental motion [23]. An interesting finding was reported when room temperature vulcanized (RTV) and high

temperature vulcanized (HTV) silicone elastomers were subjected to tropical climate simulated weathering; RTV elastomers exhibited less change in hardness as compared to the HTV probably due to decrease in the initial polymerization of HTV elastomers and hence they undergo progressive hardening with time [22].

Contact of the facial prosthesis with human skin and/or mucosa for an extended period of time may cause it to absorb skin and/or oral secretions which may lead to degradation of the material and colour changes [14,24]. Where sebum fatty acids have a tendency to interact with silicone, breaking chain bonds and decomposing the elastomer by a phenomenon called as 'Reversion'. This is further accelerated with light radiation, giving rise to softer and weaker elastomer [22].

Degradation and discoloration of facial prostheses has been attributed to fungal colonization and the inhibition of fungal growth improved the stability and the longevity of the prostheses. Disinfection of silicone elastomers is another factor that is responsible for the adverse alterations in the colour stability and hardness in addition to changes in the surface properties as well as changes in the bulk of the silicone material as a result of extraction of some compounds from the matrix to the disinfecting solutions as demonstrated by several researches [24]. However, it was reported that the influence of chemical disinfection did not produce a significant difference in the hardness values nor the roughness of MDX 4-4210 and Silastic 732 RTV 224 (Dow-Corning, USA) when compared to control groups without disinfection [25]. In a study investigating different disinfection protocols; microwave exposure disinfection procedure was recommended for the PDMS material, while disinfection with sodium hypochlorite solution was not recommended [26]. Another study comparing the effect of Fittydent tablet, chlorhexidine gluconate 4%, and neutral soap disinfection concluded that; chemical disinfection led to significant alterations in colour and surface roughness whereas no significant effect on Shore A hardness [27].

The mechanical properties of silicone elastomer may be altered by the incorporation of colorants and pigments. Haug, *et al.* [28] demonstrated that liquid colorants as artist's oil paints as well as liquid facial cosmetics initially decrease hardness and tensile strength while increasing tear strength and percent elongation due to their effect on the polymerization of the silicone; where oil inhibits catalysis of silicone material, moreover; liquid colorant act as a second liquid phase without bonding to the silicone resin matrix. Nevertheless, ultimately the material becomes harder, most probably due to absorption or evaporation of the vehicle in which the colorant was dissolved. Thus, it is imperative to evade colorants which include components that may leach out or evaporate and deteriorate the mechanical properties. On the other hand, dry colorants, act as a solid filler without bonding to the silicone, thereby decrease the tensile strength and increase the hardness. Addition of rayon fiber flocking can as well increase the hardness of the silicone elastomer by 19% since it acts as fibrous solid filler.

Polymerization method can be a factor as well. It was reported that the mechanical properties of silicone elastomer A-2186 (Factor II Inc., USA) cured in a stone mould has lower mechanical properties than that cured in a stainless steel mould, this may be due to the decline in the degree of cure by the introduction of contaminants such as traces of amines, sulphites, organo tin-compound, nitrogen oxide and carbon monoxide, those impurities may be introduced through pigments and /or curing in stone moulds and inhibit the cross-linking reaction by competing with the unsaturated carbon- carbon double bond in the initial formation of the platinum/silicone vinyl complex (CH=CH₂) [29].

Approaches to improve the mechanical properties of maxillofacial silicone elastomer

Maxillofacial silicone elastomers perform well for a limited period; but, as time passes, deterioration of the mechanical properties and changes in appearance occurs. Considering the functional and psychosocial effects of the failure of maxillofacial prostheses on patients, it is necessary to improve the mechanical and physical properties of silicone elastomer used for maxillofacial prostheses so that it more closely emulates human skin and has longer service life [30].

A plethora of research have been conducted in attempt to enhance the mechanical properties of maxillofacial silicones, among the first trials was that of Firtell, *et al.* [31], they added foam RTV silicone to conventional RTV silicone to achieve a lighter-weight prosthesis, but

the tear strength decreased with the increase in the foam silicone. In 1977, MDX 4-4210 was altered with Dow-Corning 360 medical fluid (MF), although the silicone fluid reduced the hardness but it also weakened the tensile properties and hence, the unmodified silicone showed superior mechanical properties than the modified one [32].

In 1985, RTV silicone MDX 4-4210 was modified with MF100cs which significantly decreased the surface hardness, the tensile strength, ultimate elongation, and tear resistance of the material.

Therefore, if softer material is required, it can be accomplished with the addition of MF at the expense of reduction of other mechanical properties [33]. Later in 1987 an attempt was made to create a more pliable material emulating facial tissues by combining dimethyl siloxane-triacetoxy terminated silane (Type A adhesive) with PDMS with vinyl groups (MDX 4-4210). This addition decreased the hardness as well as the ultimate tensile strength, whereas the percentage elongation increased. Tear strength was comparatively constant for the different ratios with the exception of the 50:50, where tear strength decreased by 50% [34].

Several studies have been conducted to investigate the effect of additives such as glass fibers, silica fibers, ceramic whiskers, cellulose fibers, or short aramid but fibrillar reinforcements didn't enhance the mechanical properties nor did high modulus polyethylene fibers. Yet, the incorporation of large amounts of organic or inorganic fillers can be detrimental in that they deteriorate the mechanical and physical properties of the composite [16,35]. Andreopoulos, *et al.* [36] reinforced a silicone elastomer (C-50) with particulate silica powder. The tensile strength and elongation at fracture improved with increasing the amount of silica up to 35%, while the Young modulus - describes measure of elasticity of a material- showed small reliance on the amount of silica, and the tear resistance continued to increase with increasing the filler amount.

Numerous researches were performed to develop processes to include opacifiers into polymeric materials, producing a new class of materials that combine strength of the opacifiers and flexibility of the organic polymer. Opacifiers are rigid and have a superior shear modulus in comparison to pure silicone elastomer. The improved physical and optical properties as well as resistance to stress-induced cracking and aging achieved due to the incorporation of opacifiers in to polymeric material may be due to the high surface energy and chemical reactivity of the opacifier particles, giving them the capacity to interact with the silicone elastomer matrix to form a three dimensional (3D) network within the silicone matrix. Incorporation of opacifiers as barium sulphate and titanium dioxide opacifier can also affect the colour stability and hardness of the maxillofacial silicone [37,38]. It was reported that the addition of opacifiers to silicone elastomer did not prevent weathering degradation and the use of different opacifiers had different effects on silicone elastomer mechanical properties. Furthermore, although UV mineral-based light-protecting used as opacifier (with active ingredients mainly titanium dioxide and zinc oxide and with Sun Protective Factor 30) for pigmented silicone MDX4-4210, it was reported to cause substantial degradation of the mechanical properties. On the other hand, titanium white dry opacifier was reported to cause less degradation of the silicone elastomer mechanical properties [39].

Nano-particles exhibit intrinsic surface reactivity, high surface areas and can strongly chemi-sorb many substances. The size, surface structure and inter-particle interaction of nano-materials determine their unique properties and the improved performances. Nano-oxide particles are rigid and have a higher shear modulus than the pure silicone elastomer [40]. The addition of nano-particles into a polymeric matrix improve its properties, this may be due to the higher surface energy and chemical reactivity of the particles, allowing them to interact with the silicone elastomer matrix and form a 3D network within the silicone structure. The addition of nano-sized particles results in the improvement of the material characteristics and the control of biological, mechanical, electrical, magnetic, and optical characteristics as well [41]. The results from the study of Han, *et al.* [30] indicated that incorporation of CeO₂, TiO₂ and ZnO nano-particles at concentrations of 2 to 2.5% by weight into A-2186 elastomer improves hardness, tear strength, tensile strength, and elongation. However, when the concentration was 3%, the tear strength, tensile strength, and elongation decreased, and the hardness increased. Conversely, the research conducted by Wang, *et al.* [42] concluded that the addition of TiO₂ nano-particles up to 4% improved the tensile strength and Shore A hardness of MDX4-4210 elastomer. But, the elongation at break and tear strength for the 6% composite decreased.

Emphasis was given to nano-silica reinforcing fillers in several studies. Incorporating hydrophobic surface-treated nano-silica filler into MDX4-4210, was found to be improve the dynamic visco-elastic properties; Storage moduli, loss moduli and damping factor as reported by Santawisuk., *et al* [43]. In another study, the mechanical properties of MDX4-4210 were improved by adding hydrophobic surface-treated fumed silica filler up to 10% [44]. The study performed by Zayed., *et al* [45] concluded that the incorporation of surface treated silicon dioxide nano-particles at concentration of 3% enhanced the tear strength, tensile strength and the elongation of A-2186. This may be explained by the improved dispersion of the nano-particles within the silicone matrix due the surface treatment of the nano-particles. When the effect of addition of hydrophobic silaned silica, fumed hydrophilic silica and TiO₂ was compared. It was found that hydrophobic silaned silica and fumed hydrophilic silica increased the tensile strength of A-2000 and A-2006 silicone elastomers in comparison to the control and TiO₂ groups. For A-2000, the fumed silica and the hydrophobic silica groups revealed a remarkably higher percent elongation than the control group, while the TiO₂ group demonstrated the lowest value of percent elongation for A-2006. The highest hardness value for A-2000 was in the TiO₂ group but the lowest hardness value for A-2006 was in the TiO₂ group. Regarding the tear strength; there was no significant difference among the different groups for A-2000. But, for A-2006 the fumed silica and TiO₂ groups had significantly higher tear strength than the control group [46].

A common approach for improvement of the tear strength of facial prosthesis is to incorporate an acrylic resin substructure; a modification of this approach was described to incorporate a magnet in the acrylic substructure sealed by a polyurethane liner. The retention of the acrylic resin substructure to facial silicone prosthesis was improved by Velcro [47]. Udagama., *et al* [48] managed to line the tissue-surface of silicone prostheses with a thin prefabricated vacuum adapted thermoplastic urethane film to render it adheophilic, wettable, smooth to enhance the effectiveness of water-based adhesives and to be resistant to fungal growth as well as to increase the tear-resistance at the margins and provide better marginal integrity. A modified technique for lining facial prostheses using polyurethane sheets was described by Lemon., *et al* [49]. The major drawback of this techniques is de-bonding of the liner from the silicone which may eventually lead to failure of the prosthesis. Gunay., *et al* [16] embedded tulle in silicone elastomer (A-2186) to enhance the tear resistance of the edges of maxillofacial prostheses. Tulle was used because of its inherent flexibility and because it doesn't adversely affect the elastic behaviour of the silicone elastomer, consequently, preventing tear at the edges. The results of this study concluded that tulle improved the mechanical properties of silicone elastomer particularly in terms of tear resistance.

A novel approach to reinforce polymers was tested; it is by the use of Poly-hedral-silsesquioxanes (POSSs) as a reinforcing agent. POSSs are a nano-scale organic-inorganic hybrid. The cage-like molecular structure of an individual unit of POSS contains a 1.5 nm silica cage with eight pendant organic groups. Results demonstrated that tear and tensile properties were strengthened by the addition of POSS [50]. A modified silicone elastomer was introduced in 2013 by Liu., *et al* [51] the silicone elastomer is filled with hollow microspheres to obtain optimal maxillofacial prosthetic material with low density, low heat conductivity and better flexibility. Hollow microspheres are spherical particles with diameter from tens of nano-meters to several hundreds of microns. They are usually used as light weight fillers, high quality insulating materials and damping materials. The formulation with a volume fraction of 5% hollow microspheres demonstrated the most favourable properties to be used as maxillofacial prosthetic material, but its tear strength was significantly lower than that of pure silicone elastomer. Another research claimed that it is possible to optimize the physical and mechanical properties of MDX 4-4210 silicone elastomer by incorporation of a mixture of Expancel and SiO₂ hollow microspheres [52].

Several silicone rubber formulations have been developed and investigated for use in maxillofacial applications. A material based on methacryloxypropyl terminated polydimethylsiloxane (MPDS-MF) cured by free radical thermal polymerization and cross-linking was evaluated and compared to A-2186. Results indicated that its hardness is similar to that of A-2186. Nevertheless, its tensile strength, tear strength, and ultimate elongation, and adhesive bonding strength are higher than those of A-2186. The increased adhesion to non-silicone based adhesive may be related to the presence of methacrylate groups in MPDS-MF [53]. Bellamy., *et al* [6] developed silicone rubber formulation which is the base of a three layered maxillofacial prosthetic systems which includes an inner silicone gel and a thin external

polymeric coating. The material was prepared by combining the high and low molecular weight vinyl end blocked polymers. The silica filler and hydride functional silicone polymer were then added to the base of the polymer; finally, the platinum catalyst was incorporated. The new material was developed to have higher tear strength, low hardness and adequate viscosity when compared to the commercially available materials. Another formulation introduced by Aziz, *et al.* [54] is a hydroxy-end blocked PDMS formulation with improved tear strength by using base polymer mix of 80%: 20% high to low molecular weight polymers and 20% silica filler. However, this formulation had a higher hardness than commercially available materials.

Conclusion

Although substantial improvements have taken place in the mechanical properties of silicone elastomer used in maxillofacial prosthetic reconstruction there still remains an enormous demand for novel material with ideal mechanical, physical and biological properties that mimic the properties of native tissues and resist microbial biofilm formation. Consequently, more research should be dedicated to this area.

Conflicts of Interest

None.

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