MECHANICAL BEHAVIOUR OF PRINTABLE SPLINT POROUS DESIGNS USING ADDITIVE MANUFACTURING: A FINITE ELEMENT ANALYSIS

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ABSTRACT

This study investigates the mechanical performance of splints with soft resin and acrylic biomaterials. The dental splint was created with a 3 mm height and divided into three layers. The top and bottom layers (external layers) differed by 1 and 2 mm. Body-centred cubic structures (BCC) were used to create a porous middle layer, which also varied by 1 and 2 mm. The top rigid layer was subjected to a compressive displacement in small steps to visualize the stress–strain plots for linear and non-linear regions. Results indicated that the larger the strut thickness (lower porosity) the lower stresses and reaction forces, regardless of the thickness of the middle layer. Additionally, variations in the thickness of the middle layer resulted in noticeable changes in both stresses and reaction forces. An increase in strut thickness led to higher stress levels, indicating an enhanced load-bearing capacity under the same applied deformation. In conclusion, incorporating porous layers within the splints can effectively distribute reaction forces and mechanical stresses. This design approach can offer several advantages, including enhancing patient comfort, reducing pressure on tissues such as the gingiva and peri-implant contact area.

Keywords: printable dental splints, finite element analysis, biomechanics, porous lattice structure.

1 INTRODUCTION

Restorations made of prosthetics are subject to a variety of biological and biomechanical effects. Bone loss from peri-implant infections of soft and hard tissue brought on by poor dental hygiene might be exacerbated in cases of low bone quality or increased sensitivity to infections [1]. In a similar vein, large biomechanical loads may result in excessive mechanical stress in the surrounding bone, which may lead to bone loss [2]. The distribution, direction, and magnitude of occlusal pressures acting on the implant, surrounding bone, and prosthetic reconstruction are critical factors for implant survival [3]. Bruxism patients receiving implant-supported prostheses cause parafunctional contacts and occlusal overload of the bone-implant interface [4]. However, clinical experience indicates that these individuals may develop mechanical and biological problems as a result of occlusal overload [5]. Due to their lack of a periodontal ligament and the absence of the proprioceptive feedback mechanisms of the masticatory muscles, dental implants are firmly attached to the bone. Therefore, marginal bone loss may be triggered by occlusal loading. The lifetime of an implant-supported prosthesis and the amount of bone loss surrounding an implant may also be correlated with the length and depth of implantation. This affects long-term osseointegration and is caused by variations in implant insertion depths, implant lengths, and contact areas of varied bone densities.

The art and science of employing detachable devices to produce a mechanical disadvantage for parafunctional forces and to attain neuromuscular harmony in the masticatory system is known as splint therapy [6]. Splints are advised for the therapeutic management bruxism patients in order to reduce the forces and lessen tooth wear [7]. The goal of intraoral occlusal splints is to achieve a balanced and consistent occlusal contact without irreversibly altering the dental occlusion or the mandibular rest position.

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Furthermore, a well-made splint encourages a harmonious interaction between the implant, bone, teeth, ligaments, joints, disk assemblies, masticatory muscles, and tendons. The justification for their employment is that durable, soft material exhibits a shock-absorbing [8] ability that may disperse severe loads linked to parafunctional behaviours. Since occlusal devices are usually worn at night, when parafunction episodes are most likely, a good fit is necessary to keep the device from moving while the patient is asleep. Furthermore, the anterior guiding should remove contact between the occlusal device and the posterior teeth during protrusion and eccentric movement, and there should be uniform occlusal contacts between the antagonist teeth and the device at maximum intercuspation. Given that patients typically store their occlusal devices in moist storage containers while not in use, a device must be precise and resistant to environmental changes in order to produce and maintain these harmonious contacts and disocclusion, regardless of the manufacturing process. Despite the fact that clinical practitioners generally agree that the occlusal splint is the most popular treatment, there is still debate regarding the ideal vertical thickness and the thorough criteria used to determine this thickness. According to published research, splints ranging in thickness from 1 to 15 mm have been used [9]. The efficacy of splints varying in thickness on symptoms related as well. As a result, choosing the right vertical thickness can have a significant impact on how well the treatment works.

The flask–pack–press method used to fabricate traditional dentures is similar to that used to fabricate occlusal devices. Although several materials can be employed in the traditional production of occlusal devices, the procedure is costly and time-consuming. Occlusal devices can now be manufactured both additively and thanks to recent developments in computer-aided design and computer-aided manufacturing (CAD–CAM) technologies. CAD–CAM techniques are highly efficient in terms of time, cost, and reproducibility. Occlusal devices made from prepolymerized [8] and standardized disks with superior mechanical qualities to those made traditionally can be produced by subtractive manufacturing. Nonetheless, additive manufacturing (AM) offers the advantage of producing several devices at once and producing less waste than subtractive manufacturing (SM). In many different biomedical domains, this method has become widely accepted. AM approaches are highly attractive for a range of applications because to their diversified methodologies and large range of materials with different physical properties. Furthermore, the design flexibility it offers makes it possible to produce intricate structures that would be challenging to build using conventional subtractive manufacturing techniques. As a result, digital occlusal splints are frequently made using 3D printing.

Finite element analysis (FEA) techniques have been used widely to evaluate the stress concentration pattern at the bone implant interface [10], [11]. In dentistry, there is growing interest in using occlusal load models based on FEA to examine various methods and materials [12], [13]

This study aims to design and investigate the behaviour of various splints design with porous architectures employing finite element method.

2 MATERIALS AND METHODS

2.1 Finite element modelling

Dental splints section’s 3D finite element models were created using SolidWorks 2023 (SolidWorks Corp., Dassault Systèmes, France). The splints’ overall thickness, including their solid and porous layers, was measured at 3 mm, as shown in Fig. 1. The solid upper and lower layers, as well as the porous inner splint layers, were designed to absorb forces and

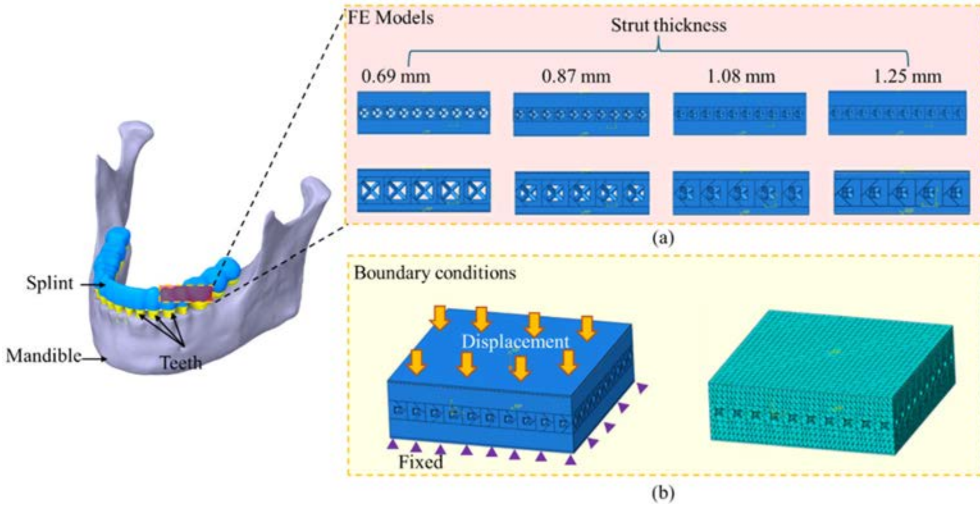

Figure 1: (a) Different splint models; and (b) mesh and boundary conditions.

reduce strains on the interfaces with the crown and between the implant and bone as well. The splints' top and bottom layers differed by 1 and 2 mm. Body-centred cubic structures (BCC) were used in the design of the middle layers, which also varied by 1 and 2 mm.

2.2 Materials assignment

Two types of biomaterials [14] were utilized to assign the splint model. Soft resin was used for the internal layer (middle layer), while acrylic was used to model the external layer (top and bottom layers). The material properties of the splints were listed in Table 1. Non-linear material properties were assigned to the analysis to accurately capture the mechanical behaviour of the splints.

Table 1: Material properties used for finite element investigation.

Material	Young's modulus (MPa)	Poisson's ratio	Yield strength (MPa)
Acrylic	2,400	0.35	110
Soft resin	1,100	0.30	45

2.3 Interactions

The top and bottom plates were modelled as rigid bodies and were connected to the splint surfaces using surface-to-surface tie constraints.

2.4 Boundary conditions

The top rigid layer was subjected to a compressive displacement of 0.5 mm and the bottom rigid layer was fixed in all directions. The compression was applied in small steps to visualize the stress–strain plots for linear and non-linear regions.

2.5 Mesh and element type

A mesh size of 0.2 mm was selected based on mesh sensitivity analysis to ensure results accuracy. Ten-node quadratic tetrahedral elements (C3D10) were used for meshing. The element type of 10-node quadratic tetrahedron (C3D10) was selected. The total number of elements ranged from 342301 to 109164 for different models.

3 RESULTS

The results of finite element analyses are shown in Fig. 2 and listed in Table 2. It was revealed that the larger strut thickness (lower porosity) showed lower stresses and reaction forces compared to the smaller strut thickness for the same thickness of the middle layer. Also, the change in stresses and reaction forces was seen when the thickness of the middle layer was changed. A higher thickness of the middle layer showed lesser stresses and reaction forces compared to a smaller thickness of the middle layer. In the case of a 1 mm thick middle layer with a strut thickness of 0.69 mm, the reaction force of 839 N was seen, however, this reaction force was increased to 1449, 2679 and 4154 N for the strut thicknesses of 0.87, 1.08 and 1.25 mm, respectively. Hence, a larger thickness of the strut can carry more load compared to a smaller thickness of the strut. Similarly, the von-Mises stresses in the case of a 1 mm thick middle layer with a strut thickness of 0.69, 0.87, 1.08 and 1.25 mm were seen as 116.9, 119.7, 121.2 and 144.2 MPa, respectively. The stresses were increased by increasing the strut thickness, which is evidence of carrying more load with the same amount of applied deformation. Similarly, in the case of a 2 mm thick middle layer, the stresses and reaction forces were 99, 74.7, 108.2, 147.9 MPa and 810, 1383.3, 2457 and 3840 N, respectively, for the strut thickness of 0.69, 0.87, 1.08 and 1.25 mm.

Table 2: Reaction forces and stresses in different splint models.

Layer thickness (mm)	Strut thickness (mm)	Stresses (MPa)	Reaction force (N)
1	0.69	116.9	839
	0.87	119.7	1449
	1.08	121.2	2679
	1.25	144.2	4154
2	0.69	99.0	810
	0.87	47.7	1383
	1.08	108.2	2457
	1.25	147.9	3840

The results showed that the higher the thickness of the porous layer can deform more for a specific load. Fig. 3 shows the von Mises stress distribution. It was observed that a 2 mm porous layer with a strut thickness of 1.25 mm resulted in reduced deformation. This suggests a higher energy absorption capacity, making it particularly suitable for splint applications under masticatory forces. In this configuration, stresses are efficiently transferred from the solid regions to the porous structure, where the struts play a key role in absorbing mechanical loads. This behaviour aligns with the well-known load-distributing characteristics of the BCC architecture. Designing splints with porous structures internally can enhance stress distribution and absorb occlusal forces more efficiently, reducing the risk of localized

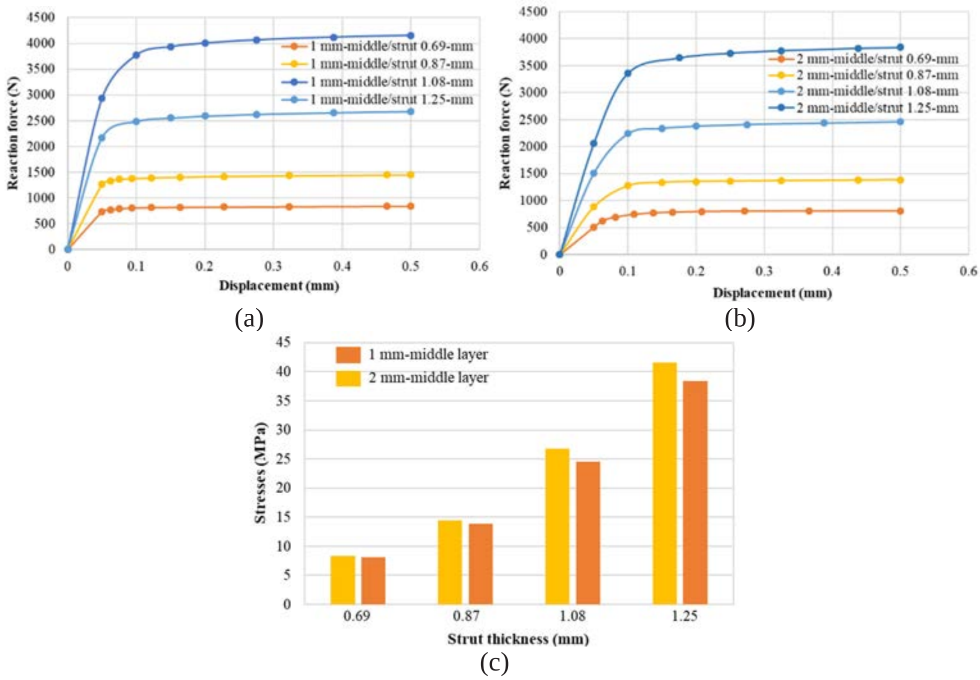

Figure 2: Results of finite element analyses. (a) Compressive behavior and reaction force for 1 mm thick middle layer splint; (b) Compressive behavior and reaction force for 2 mm thick middle layer splint; and (c) von-Mises stresses in different splint models.

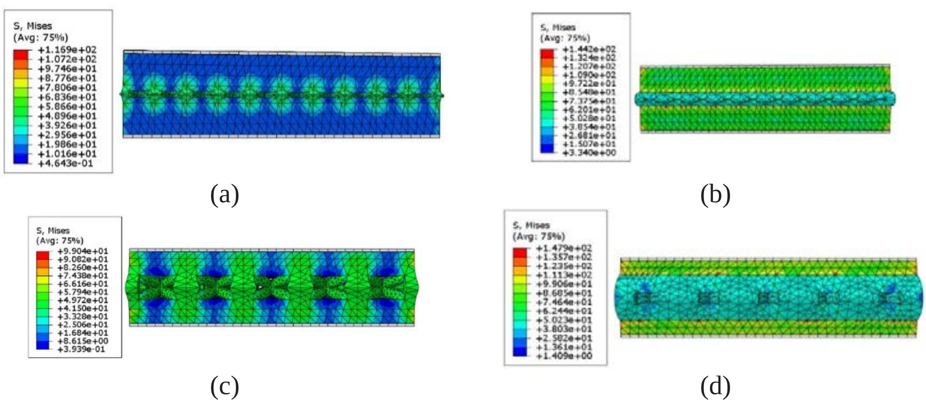

Figure 3: Ghraphical presentation of von Mises stresses and deformation. (a) 1 mm middle layer with 0.69 mm strut thickness; (b) 1 mm middle layer with 1.25 mm strut thickness; (c) 2 mm middle layer with 0.69 mm strut thickness; and (d) 2 mm middle layer with 1.25 mm strut thickness.

pressure points. As a result, patients may experience improved comfort, better protection of underlying tissues, and reduced mechanical stress on dental restorations or implants, making this design especially suitable for long-term use during functional activities such as chewing.

4 CONCLUSION

The results indicated that the porosity of the splints plays a vital role in the stress–strain behaviour when they were subjected to the loads. The higher the porosity, the lower the stresses and the reaction forces. Similarly, increasing the thickness of the porous layer from 1 mm to 2 mm results in a decrease in the stresses and reaction forces for the specific amount of deformation. These results indicate that incorporating porous layers within the splints can effectively distribute reaction forces and mechanical stresses. Thus, introducing the porous layers in the splint may enhance patient comfort with flexible deformation and absorb the stresses during the masticatory forces. Moreover, in the case of dental implants, excess stress transfer to the peri-implant bone can be reduced to prevent bone damage. This design approach can offer several advantages, including enhancing patient comfort, reducing pressure on tissues such as the gingiva and peri-implant contact area.

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