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Evaluation of stress distribution in bone at varying lengths and diameters of micro-implant

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Abstract
Introduction: Temporary Anchorage Devices play an important role to overcome the limitations of conventional anchorage control in fixed orthodontic practice.
Aims and Objectives: To evaluate the stress distribution in bone at varying lengths and diameters of the micro-implant.
Material and Methods: 3-D Finite element models of maxilla, mandible, and micro-implants with varying lengths and diameters were generated. The micro-implants were inserted 90^{o} to the bone surface. A force of 200 g was applied from the micro-implant to the power arm. Stress distribution and its magnitude were then analyzed with a 3-dimensional finite element analysis program.
Results: Increased diameter of micro-implant, showed significant negative correlation with the stress generated. The maximum von Mises stress was found for implant of diameter 1.2 mm and least for implant of diameter 1.8 mm.
Conclusion: Increasing the diameter of the micro-implant reduces the stress concentration in bone, thereby increasing the likelihood of implant stabilization.

Keywords: Anchorage, finite element method, micro-implant, von mises stress

Introduction
In this era of progression, the truths of today become the myths of tomorrow; practitioners of a scientific discipline are generally resistant to accept a new paradigm. Nonetheless after a paradigm shift has occurred, a veritable explosion of new ideas and information occurs, leading to rapid advances in the field. Orthodontics as of today is on a threshold of virtual deluge of knowledge, flowing in at a rapid pace.
Since the 17th century, the principles of anchorage during orthodontic treatment have been clearly understood. For every (desired) action there is an equal and opposite reaction. Anchorage, then, is the resistance to the reactionary force that is either provided by other teeth, or by other hard and soft tissues of the face. An important aspect of treatment is maximizing the tooth movement that is desired, while minimizing undesirable side effects i.e. anchorage loss.
The micro-implant introduced by Ryuzo Kanomi [1] in 1997 is now widely used as an orthodontic anchorage device for anchorage control since they have various advantages like small size, convenient insertion and removal procedures with the requirement of less patient cooperation, possibility for immediate orthodontic loading [2, 3] and are also cost effective when compared with traditional anchorage reinforcements such as trans-palatal arches, nance-palatal arches or extraoral appliances.
In the field of dentistry a variety of sophisticated procedures and equipment are used, which are based on basic concepts of engineering. One such important procedure is the Finite Element Method (FEM) which was developed by Richard Courant in 1943 [4]. It is a means of discretizing a continuous structure into sub-domains called “Finite Elements” which forms the basis for finite element analysis (FEA). This method was introduced to dental biomedical research in 1973 [5], since then it has been extensively applied to analyse the stress and strain in the alveolar supporting structures and especially in the periodontal ligament. In Orthodontics, the application of finite element study has been introduced in 1980’s [6].

The need for an orthodontist to know about FEM is in the scope for research and a need for basic understanding about the physiologic reactions that occur within the dento-alveolar complex [7].
The force applied at a specific point acting along the various surfaces can also be elucidated.
Hence, in this paper finite element models were constructed to evaluate the stress distribution in bone at varying lengths and diameters of the micro-implant.

Materials and Method
Three-dimensional models of the maxilla and the mandible along with dentition, periodontal ligament, brackets, archwire, micro-implant and closed coil spring were generated by using SOLIDWORKS 14.0 software [company: Dassault Systems (DS)].
3-D models of all the components were customized individually and then assembled to create three-dimensional finite element models to perform en-masse retraction of 6 anterior teeth.

Materials
1. The maxilla (D3 bone quality)
2. The mandible (D2 bone quality)
3. The periodontal ligament (thickness 0.25 mm evenly)
4. Brackets (slot 0.022” x 0.028”)
5. Stainless steel rectangular wire (0.019” x0.025”)
6. The power arm (height 5.5mm)
7. TAD’s (varying lengths (6 mm, 7 mm, 8 mm, 9 mm) & varying diameters (1.2 mm, 1.4 mm, 1.6 mm, 1.8 mm))
8. Nickel-titanium closed-coil spring (0.012” x 0.036”)

Methodology
Three-dimensional models of the above-mentioned components were generated as follows:
The jaws with the extracted first premolars (Fig1) were simulated with Type 3 (D3) bone quality present in the maxillary posterior region with 1 mm thin cortical bone surrounding a core of dense trabecular bone having favourable strength and Type 2 (D2) bone quality in the mandibular posterior region with 2 mm thick layer of compact bone surrounding the dense trabecular bone [8].

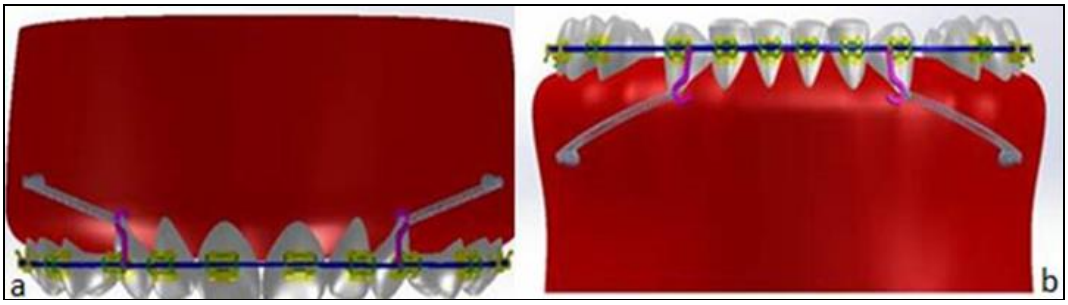

Fig 1: Showing 3D model of (a) Maxilla and (b) Mandible

The teeth were constructed with ideal crown and root dimensions according to Wheeler [9] and arranged according to Andrews six keys to normal occlusion [10]. The periodontal ligament was constructed having the thickness of 0.25 mm evenly to fit over the outer surface of the root, based on the studies of Kronfeld [11] and Coolidge [12], although its thickness may be different according to age, position, and individual variations. 0.022 x 0.028-in standard preadjusted edgewise brackets (3M Unitek, Monrovia, Calif) were made and attached to the centre of the clinical crown so that the facial axis point was at the centre of the bracket slot.

The power arms of height 5.5 mm were generated and attached distal to the lateral incisor bilaterally and perpendicularly to the archwire, to achieve bodily anterior tooth movement and to gain better control of the anterior teeth in sliding mechanics [13].
Abso Anchor TAD’s (Fig 2) were constructed having various diameters and lengths. The TAD’s were placed 8 mm in the maxilla and 11 mm in the mandible from the alveolar crest in the inter-radicular space between the first molar and the second premolar[14] and inserted at 90° to the bone surface [15].

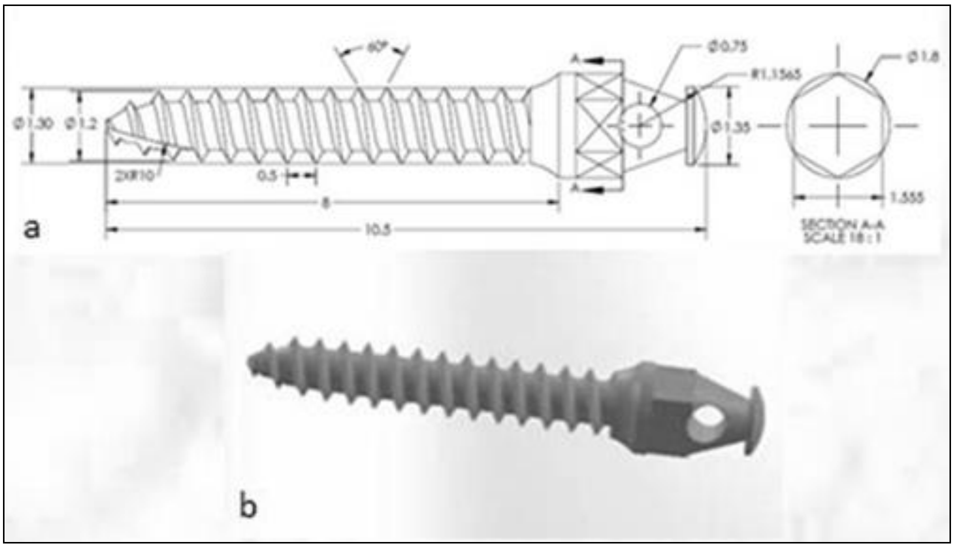

Fig 2: Showing Abso Anchor Temporary Anchorage device (SH1312-08 and SH1312-07; Dentos, Taegu, Korea)

Nickel-titanium closed-coil spring was designed and stretched between the TAD's and the hook (power arm) to deliver 200 g of force. Young's modulus of elasticity and Poisson's ratio for the teeth, periodontal ligament, bracket, and the wire were calculated according to the methods of Vollmer *et al* [16] and Reimann *et al*. [17] (Table 1). Young's modulus and Poisson's ratio for the trabecular bone varied depending on the quality of bone available at the site of implant placement.

Table 1: Material properties of various components used in the study

S. No.	Materials	Young's Modulus (MPa)	Poisson's Ratio
1	Tooth	20,000	0.30
2	Periodontal Ligament	0.05	0.30
3	Alveolar Bone 2,000 0.30	2000	0.30
4	Bracket/Arch Wire/ Power arm	200,000	0.30
5	Closed coil- spring	110,000	0.35
6	Temporary Anchorage Devices (Micro-implant)	110,000	0.35
7	Cortical Bone	13,700	0.30
8	Cancellous Bone (D3) 1	1600	0.30
9	Cancellous Bone (D5) 5	5500	0.30

The entire assembly was then exported for analysis with ANSYS Workbench (version 14.0; ANSYS, Canonsburg, Pa) through a bidirectional understandable translated system called initial graphics exchange specification (IGES).

Three-dimensional quadrangular and hexagonal element [(Solid98 (Tet10)] was used for meshing having 260984 nodes and 137171 elements in the maxilla and 187816 nodes and 103447 elements in the mandible. 16 models each for the maxilla and the mandible were developed for the study with varying the implant TAD's (6 mm, 7 mm, 8 mm, 9 mm) and varying the TAD's diameter (1.2 mm, 1.4 mm, 1.6 mm, 1.8 mm).

A simulated retraction force of 200 g was loaded mesiodistally to the centre of the TAD's, and stress distribution and its magnitude were analysed.

An assessment of the stress on the bone elements was performed by using von Mises equivalent stress (Fig 3).

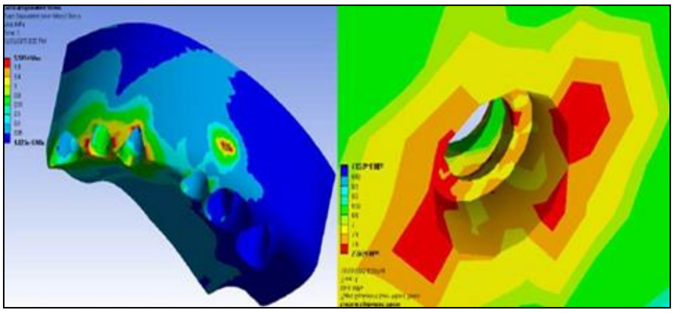

Fig 3: Showing cortical von-Mises stresses and cortical zoom around TAD

Results

A colour scale with 9 stress values was used to evaluate the stress distribution in the bone with different dimensions of the TAD's.

The scale for stress runs from lowest stress values i.e. blue, to the highest von Mises stress values i.e. red (Fig. 3).

The stress in the TAD-bone interface as well as anteriorly in the bone adjacent to the power arm were generated both in the maxilla and the mandible.

Diameter of TAD's vs stress level

The maximum von Mises stress near the TAD's (Fig.4) was 6.15 MPa when the TAD's of 1.2mm diameter was used which gradually decreased to 4.47 to 3.51 to 3.07 MPa as the diameter increased from 1.4 to 1.6 to 1.8 mm respectively.

Thus, von Mises stresses generated in the cortical bone near the implant was the highest with 1.2 mm and lowest when 1.8 mm diameter.

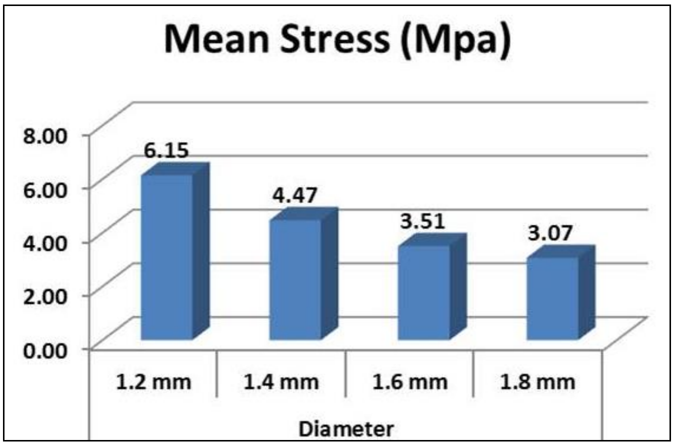

Fig 4: Graph showing correlation between stress and diameter

Length of TAD's vs stress level:

The maximum von Mises stress generated do not have any correlation with various lengths of TAD's (Fig. 5)

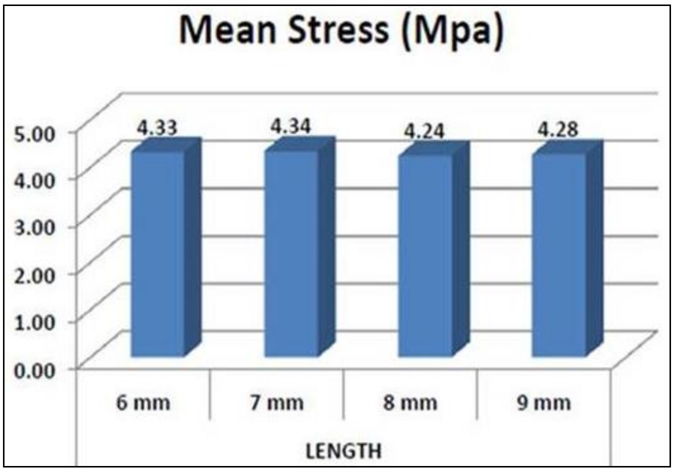

Fig 5: Graph showing correlation between stress and length

Mean stress in anterior region between maxilla and mandible

In the mandible, the maximum von Mises stress in the bone was 16.05 MPa generated anteriorly adjacent to the power arm compared to the maxilla in which the stress was 1.68 MPa which was the least anteriorly (Fig. 6).

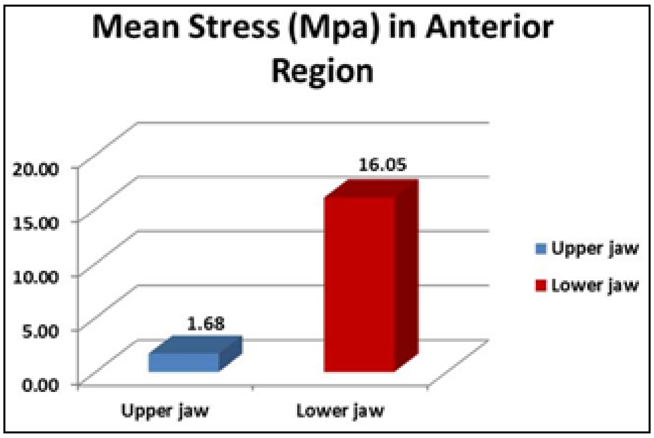

Fig 6: Showing comparison of mean stress between maxilla and mandible adjacent to the power arm

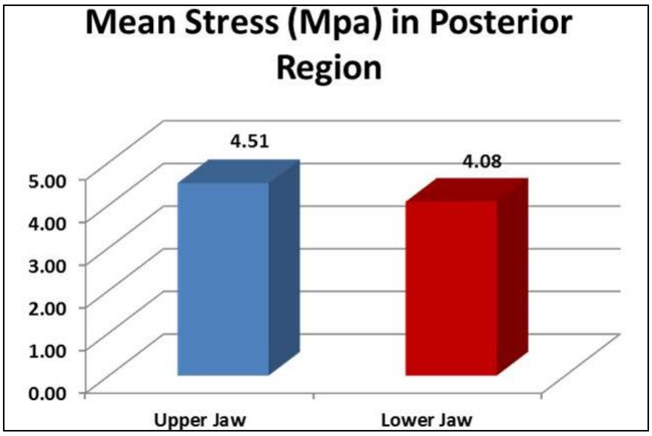

Fig 7: Showing comparison of mean stress between maxilla and mandible adjacent to the TAD

Mean stress in posterior region between maxilla and mandible

In the maxilla, the maximum von Mises stress in the bone was 4.51 MPa generated posteriorly adjacent to the TAD whereas in mandible the stress was 4.08 MPa (Fig. 7).

Length and Diameter of TAD's vs stress level:

The mean von Mises stress was found to be the least when TAD's of 1. 8 mm diameter and 8 mm length was used with stress generation of 3.004 MPa (Table 2).

Table 2: Comparison of mean stress at different lengths and diameters

Length (mm)	Diameter (mm)	Mean Stress
6mm	1.2	6.120
	1.4	4.582
	1.6	3.572
	1.8	3.047
7mm	1.2	6.212
	1.4	4.508
	1.6	3.490
	1.8	3.167
8mm	1.2	6.082
	1.4	4.380
	1.6	3.513
	1.8	3.004
9mm	1.2	6.169
	1.4	4.423
	1.6	3.453
	1.8	3.068

It was also found that at every length, TAD's with different diameters showed decreased range of stress (area of stress distribution-red area showing maximum stress) as the diameter increased from 1.2 to 1.8 mm (Fig 8).

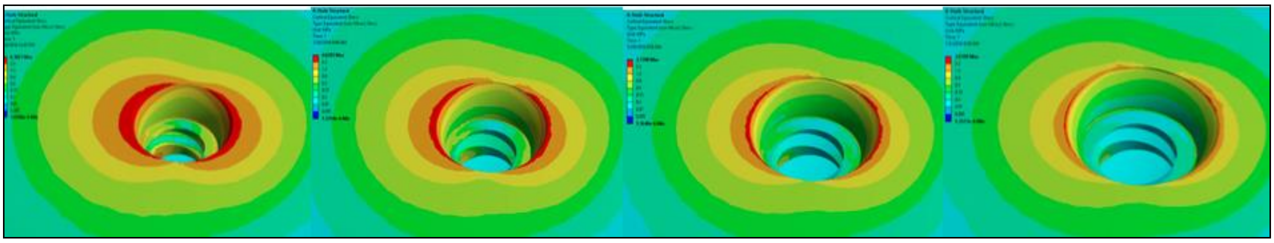

Fig 8: Showing correlation between stress and different diameters of TAD

Discussion

In this study, the stress pattern generated was evaluated for different TAD's lengths (6 mm, 7 mm, 8 mm, and 9 mm) and each of these TAD's with different diameters (1.2 mm, 1.4 mm, 1.6 mm & 1.8 mm) by using the 3D finite element method. The whole model was generated and analysed for obtaining the stresses produced unlike other studies [15, 18, 19]. who have considered a bone block representing the section of the inter-radicular bone or a portion of the jaw instead of analysing the whole model.

Diameter of TAD's vs stress level

The results of this study indicated that the von Mises stress in cortical bone was affected primarily by the diameter of the TAD. A significant negative correlation was found between diameter and stress (Fig 9). As the diameter of the TAD was increased, the mean stress was significantly reduced with the least stress with 1.8mm diameter. It was also observed that as the diameter was increased the range of stress (i.e. the area of stress distribution) was reduced thereby having reduced chances of side effects to the neighbouring living tissue, the roots. The results of the present study were in accordance with the study of Duaibis *et al.* [18], but the stresses generated were more as compared to the present study. The variation may be due to the variation in the analysis of stresses generated, since they have analysed on a bone block as compared to the present study in which the whole of the maxillary and the mandibular model was analysed.

On the contrary, study conducted by Wu *et al.* [20], evaluated patients undergoing treatment with TAD's with diameters ranging from 1.2 to 2 mm and considered those TAD's as successful which were maintained for more than 6 months. By this evaluation they recommended the use of a TAD diameter equal to or less than 1.4 mm in the maxilla and TAD of diameter larger than 1.4 mm in the mandible.

The increase in the TAD diameter can efficaciously enhance the stability, implicating that the TAD can bear a mesiodistal orthodontic force optimally well. Hence, the advantage of using a larger diameter TAD, is its ability to distribute the forces applied over a greater area of bone with less production of bone stresses.

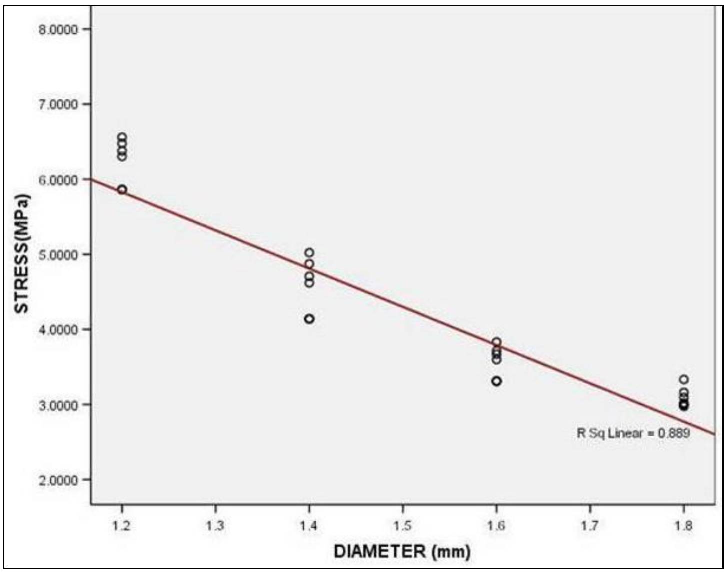

Fig 9: Showing correlation between stress and different diameter of TAD's

Length of TAD's vs stress level

Changing the length of the TAD did not have a considerable effect on the maximum von

Mises stress generated in bone at TAD site (Fig 10). The results are in accordance with the study of Duaibis *et al.* [18], but the stresses generated were more. The variation may be due to the variation in the analysis of stresses generated. On the contrary, a study by Deguchi *et al.* [21] with the use of computed tomography found the distance between the intercortical bone and the roots to be 2 mm on average, thickness of the cortical bone to be 2 mm and considered soft tissue thickness to be 2-3 mm and thus concluded the safest length of the TAD as 6 mm.

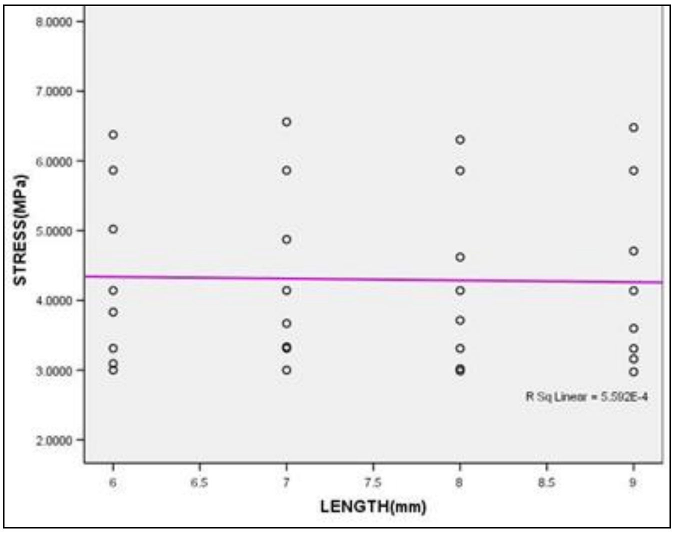

Fig 10: Showing correlation between stress and different length of TAD's

Mean stress in anterior and posterior region between maxilla and mandible:

The stresses in different bone qualities can be influenced greatly by the material properties of each layer. The material properties of cancellous bone varied according to its density in the posterior regions of the maxilla and the mandible with least stresses transmitted to the cancellous bone.

Comparison of the mean stress in the anterior region near the power arm was observed to be higher in mandible as compared to the maxilla (Fig.6). Whereas on comparison of the mean stress in the posterior region near TAD, a higher value was observed in the maxilla as compared to mandible though statistically it showed no significant difference (Fig. 7) unlike the study of Jasmine *et al.* [15].

Also a study by Kronfeld [11], showed a close relation between the density of the bone and the failure rate of dental implants (TAD's) which was found to be 3% for bone types 1, 2, and 3, but 35% for bone type 4. Also, more TAD failures were reported in areas with dense cortical bone and minimal trabecular bone when compared to areas with dense trabecular bone and dense cortical bone or dense trabecular bone with thin cortical bone. Similarly, the success of TAD's can also be influenced by the quality of bone [22].

Length and Diameter of micro-implant vs stress level

A significant difference in the mean stress level was found at different diameters of the TAD with no significant difference was seen for different lengths or comparison with lengths and diameters together. When 6 mm TAD length and different diameters (1.2, 1.4, 1.6 and 1.8 mm) were observed, it was found that 1.8 mm of TAD diameter showed the least stress generated (3.047 MPa). Similarly it was observed that TAD diameter of 1.8 mm also showed the least stress generated with 7, 8 and 9 mm length with the stress value of 3.167 MPa, 3.004 MPa and 3.068 MPa respectively as compared to 1.2, 1.4 and 1.6 mm of diameters (Table 2). Hence, irrespective of the length, the least stress was observed with the TAD of diameter 1.8 mm and considering different lengths at 1.8 mm diameter, the least stress was observed with 8 mm length of TAD (3.004 MPa). Thus, the recommended dimension of the TAD to be used is of 8mm length and 1.8mm diameter. Similar results were shown by Crismani *et al.* [23]. who concluded that screws under 8mm length and 1.2 mm diameter should be avoided.

Conclusion

It was concluded from the present study that with the increase in diameter of the TAD from 1.2mm to 1.8mm there will be significant reduction in stresses and reduced range of stresses i.e. area of stress distribution around the TAD, whereas the increase in length of TAD showed no relationship to the stress generated around the TAD. Least stress was found with 1.8 mm diameter and 8 mm length of the TAD. When 200 grams of force was applied from the TAD to the power arm, in maxilla the stresses distributed were higher around the TAD whereas in the mandible the stresses distributed were higher adjacent to the power arm. Therefore, with en-masse retraction of the anterior teeth, there are all possibilities to loose anchorage while applying forces from maxillary molars. Hence, it is recommended to use TAD in high anchorage cases while retraction of the anterior teeth.

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