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Comparison of Opening Torque of Grooved and Non-grooved Memory Alloy Screws for Mandibular Fracture Fixation

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Abstract

Background and aims. The aim of this study was to assess the opening torque of grooved and non-grooved screws made of shape memory alloys (SMA) for fixation of mandibular fractures.

Materials and methods. In this in vitro study the opening torques of ten SMA screws with grooves and ten SMA screws without grooves were compared in 20 holes (10 holes for grooved screws and 10 holes for non-grooved screws) placed in the bovine mandible. Statistical significance was set at $P < 0.05$.

Results. The mean opening torque was 2.27 ± 0.43 nm for screws in the control group and 2.05 ± 0.45 nm for screws in the grooved group ($P < 0.05$, $P = 0.08$).

Conclusion. This study revealed that the torque needed for opening the grooved screws was higher in comparison with current fixation screws; however, the difference was not statistically significant.

Key words: **Bone, implant, mandibular fixation, shape memory alloys (SMA), torque.**

Introduction

One of the most important factors in the healing process of bone fractures is the stable and rigid fixation of fractured parts together, which is achieved through fixing screws and plates. Most of these devices are made of titanium (Ti), polyglycolic acid (PGA) and polylactic acid (PLA). If screws do not sufficiently engage the bone due to reasons such as low bone density or faulty technique it can cause movement of the segments, create a gap, cause further bone loss or graft loss and even create cartilage or fibrosis instead of normal bone tissue (nonunion).[1]

As mentioned before, primary instability of screws occurs in people with low bone density due to diseases such as osteoporosis or physiologic conditions like menopause.[2,3]

So far, some methods have been proposed to overcome this problem; Pitzen et al[4] studied the effect of thickening the screws on increasing their rate of fixation and found nothing significant.

Several studies have examined the effects of changes in the screw thread design,[5] use of higher force while placing the screws,[6,7] heightening the screws[8] and use of various types of cements.[9,10] However, these methods have not significantly increased screw fixation and have resulted in problems such as loss of time, increasing the cost, adding trauma to the surgical site and even requirement for further surgery.[9]

Shape memory alloys (SMA) are very interesting metals in terms of mechanical properties and physical behaviors, which are known for more than 30 years in the field of medical sciences.[11] Along with their general properties such as high biocompatibility, high corrosion resistance, excellent heat tolerance, low fatigue and high reversible strain, these exclusive alloys have two important and special properties: 1) Their super-elasticity that results in very high flexibility at relatively high temperatures so that they can be deformed up to about 10% without any permanent deformation; 2) Their shape memory property at lower temperatures. This means that their original form is induced at specific temperatures and pressure conditions during construction and it is kept in their memory. They are extremely deformed in the effect of force application but they return to their primary shape if heated.[12]

The mechanism that causes shape memory alloys to have such a wonderful behavior is related to their phasic changes. In general, a shape memory alloy has a martensitic phase at low temperatures and an austenitic phase at high temperatures. In the event of force application and deformation of this material at low temperatures, twinned martensitic loses this state and becomes oriented martensitic that is called the non-twinned martensitic phase. If in this state, the temperature of the material rises, the martensitic phase turns to the austenitic phase.[12-14]

Regarding the interesting features and properties of intelligent alloys and their wide applications in various sciences such as medicine, we sought to compare the opening torque of standard-design and fluted screws made of shape memory alloys. These alloys may be used for the following applications: 1) as dental implant;[15] 2) for fixations of various areas;[16-18] 3) for different types of osteotomies and distractions;[19,20] 4) to treat cleft palate and align maxillary bones;[21] 5) as gap-free dental implants;[22] 6) as rods and punches in the treatment of various spinal disorders;[23,24] 7) as a stent in cardiovascular surgeries;[11] 8) as a guide wire in delicate surgeries;[12] and 9) in orthodontic wires and endodontic files.[25]

The advantages of using SMA have been emphasized in all these studies. In most of them, it has been mentioned that it allows for shorter periods of repair, less problems in treated patients, more stable fixation and better healing, less scar, better tissue repair and generally more appropriate treatment selection. We performed a study to assess opening torque of grooved and non-grooved screws made of SMA for mandibular fracture fixation.

Materials and Methods

Study Design

This comparative observational cross-sectional study was conducted in 2012. Before initiation, the study protocol was approved by the official review board.
In this in vitro study the opening torques for ten screws with grooves and ten screws without grooves made of SMA were measured and compared in twenty holes (10 holes for grooved screws and 10 holes for non–grooved screws) placed in the bovine mandible.

Methods

Initially, prefabricated bars of SMA (NiTi) measuring 1.4 inches in diameter and 10 feet in length (MEMRY, Bethel, USA) were obtained. In order to eliminate the confounding factors such as different materials of screws or possible differences in the size and design of the threads, a decision was made to construct standard screws with the diametric properties of SMA bars for the control group (through an integrating program of screw construction and thread cutting with the same depth and distance in a lathe machine).

The number of screw holes was randomly determined at 20 samples for each group. Since the properties of shape memory alloys depend on transformation from austenitic to martensitic phase, the samples were first prepared in the form of screw (10 mm in length, 5 mm in body diameter and 6 mm in screw head diameter) and then heated to 80°C to reach the austenite phase. After converting the context phase to the austenite phase in all the samples, a groove measuring 0.2 mm in thickness was created in sample screws (from the screw head to half of the threaded part) by wire cut; therefore, the fabricated screw achieved the capacity to open towards its central axis. After opening the 0.3–0.4-mm screws from the cutting site (the final groove was 0.5–0.6 mm), they were maintained at 100ºC for 210 minutes. Then, to achieve 100% content of martensite, by quickly cooling down to -80ºC (according to the manufacturer's recommendations and after microstructure study by electron microscopy), they were cooled using liquid nitrogen.

In the martensitic phase, the samples returned to their original form (closed) by exerting force and remained in the same form until they were used.

In the control group, screws (without grooves) were made without thermal cycles and only by creating a thread.

Two fresh bovine mandibles were prepared due to the similarity of their density and mineral content to human bone.[26,27] To prevent dehydration, they were covered in a wet towel at a temperature of 4ºC. Then, because of appropriate density, 10 holes with a diameter of 4 mm were embedded at 2-cm intervals from each other by a 4-mm drill in the angle and ramus of one mandible under cooling with normal saline using an irrigating syringe. The holes were prepared with 94% reproducibility by a skilled periodontist. Then, an Allen wrench of 6 mm and an analog torque comparator with an accuracy of 0.5 nm were used by the same periodontist in a clockwise direction; screws with standard design were placed under a constant force of 2 Nm. After 30 minutes, the screws were opened by the same wrench and the removal torques were measured with a torque meter.

Then, for the grooved-screw group, 10 holes were built exactly the same as those in the other mandible in the angle and ramus under the same conditions as for the standard screw. Then, before wrenching, the grooved-screw was heated up to 50ºC until the groove opened.[24] Then, the screw was cooled up to 25ºC and the groove was closed by force, ready to be wrenched in the embedded holes. During wrenching, a small metal stainless steel plate with a thickness of 0.2 mm was placed to avoid closing the gap in the groove during cutting. Then, it was wrenched into the hole with a constant force of 2 Nm by the same Allen wrench and the torque meter; the standard screws were heated through an external heat source (hot water)[24] to reach 50°C. Due to the high thermal conductivity of SMA, just heating the screw head suffices to transfer heat to the entire screw. After this period, the screw reached its austenitic phase and the groove opened; loosening of the sheet metal in the groove was indicative of groove opening.

After 30 minutes when the screw reached equilibrium with the environment, it was opened by torque wrench while the connector wrench of 7 mm was used instead of Allen wrench of 6 mm and the removal torque was recorded. During opening the screw, 3 metal plates (totally equivalent to the thickness of 0.6 mm) were placed in the opened groove to avoid closing the groove.

Like the standard screw, this screw was also prepared, wrenched, heated and opened in 20 holes and the removal torque was calculated separately for each hole.

Statistical Analysis

All the observations were recorded by a blinded operator and the data were presented as means ± SD for continuous variables. Statistical differences between the groups were assessed by ANOVA. All the analyses were carried out using SPSS 20 (SPSS Inc., Chicago, IL, USA). Statistical significance was set at and $P < 0.05$.

Results

Twenty screws were made of SMA; ten screws had grooves and ten screws had no grooves (control group).

Pull-out strength (POS) was measured as torque of removal in each group. It should be noted that closing torque was 2 Nm in all the samples.

The mean opening torque was 2.275 ± 0.43 nm for screws in the control group and 2.055 ± 0.45 nm for screws in the grooved group ($P < 0.05$, $P = 0.08$).

The mean opening torques were compared between the screws in the control and grooved groups (Figure 1).

Discussion

Intelligent alloys are very interesting metals with unique physical properties and behaviors. Their two prominent properties are super-elasticity at relatively high temperatures and shape memory at lower temperatures; the former allows the elastic deformation up to about 10% and the latter is the ability to remember and return to their primary shape.

The main mechanism of this behavior is related to the phase changes in these materials. This study aimed to compare the opening torque of screws with standard design and grooved screws made of memory alloys. After putting the screw in bone and transferring heat to it, the groove of the screw opened (back to the original shape) and as a result, force was directed to the surrounding bone.[28] According to the principle of action and reaction, it seems that the force will increase the primary fixation of the screw in bone. This is a new discussion in the field of jaw and bone fixations; achieving this goal will undoubtedly lead to obtaining the best initial fixation for the healing process of injuries, surgeries and types of bone bonding particularly in patients with chronic bone problems.

One of the problems in this study was the heat (returning temperature) necessary to be transferred to the initial form of the screw with groove (the desired

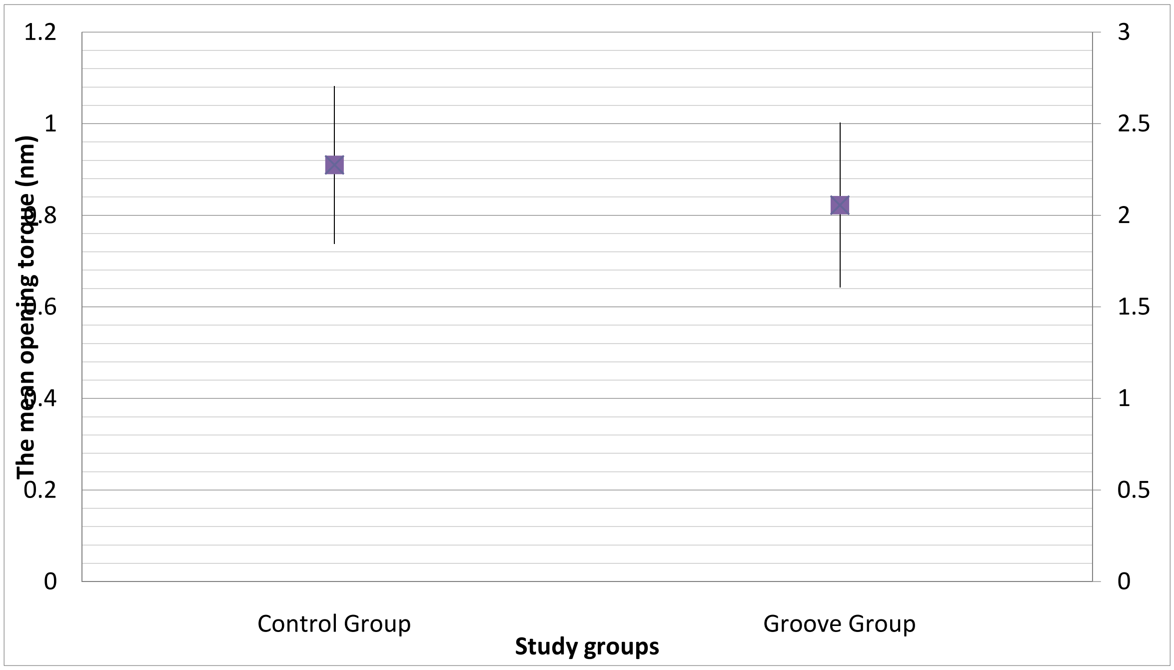

Figure 1. The mean opening torque in the control group and grooved group.

temperature was 50ºC). This heat can be transferred to the surrounding bone while heating the screw. Since based on the principles of implantology and bone surgery, increasing bone temperature up to 47–50ºC can cause bone necrosis during preparation of holes or bone reshaping by a rotary device,[29] it seems that the process of heating can lead to the necrosis of the surrounding bone and loosening the screw in bone. Two points are important here: first, because of the high thermal conductivity of the alloy, wrenching the screw into the bone causes the body temperature (37ºC) to quickly transmit to the screw and if the returning temperature of the screw is about 37–40ºC, the screw groove will open before full entry and we will not achieve our goal. Second, according to several studies, oral temperature can reach to about 68–70°C when consuming hot foods and drinks.[30, 31]

Although it is observed that the tolerance of this temperature is not directly related to bone, it seems that natural sources such as hot drinks can be used instead of an external heat source to achieve the returning temperature and open the screw groove. On the other hand, even in the case of occurrence of heat-induced bone necrosis around the screw, it should be noted that the opening torque was applied 30 minutes after applying the heat (not enough time for bone necrosis to affect our result). Another confounding factor may be the groove design of the screw. Considering that the groove in the screw head is involved during opening by a wrench and the screw head is engaged by it, it may cause the groove to close and consequently loosen the screw. As a result, the recorded opening torque would be less than that of screws with conventional designs. In this study, we attempted to reduce this confounding factor by inserting metal plates between the grooves.

On the other hand, since the groove was placed in the coronal half of the screw, it has the most involvement in cortical bone. Since the cortical bone has less elasticity than the spongy bone, it is likely that this factor also causes the screw to loosen in the bone. Given these two points, it is suggested that the groove of screw be created in its apical half because of more contact with the spongy bone.

Given that the current study focused on the creation of grooves in screws fixing bone, there is no similar study to compare the results. However, there are many studies trying to seek factors causing further stability of screws in the bone with normal design.[4] Pitzen[4] has specifically mentioned that the bone density is a very important factor in the stability of the screw. Ricci et al[5] studied the effect of various designs of threads in terms of depth and distance of the rod on the entering torque and removal force of the screw in osteoporotic bones and specified that this factor could also slightly improve the POS in bones.

In another study, Daftari et al[6] investigated the correlations between screw hole preparation, insertion torque and pull-out strength. Their results showed that the greater the closing torque, the higher the removal force and the better the stability of the screw. Also, they reported that the more the cortical bone is involved, the higher the removal force. However, this finding was contradicted by Okuyama et al[7] in a similar study conducted a few years later. They reported that although in osteoporotic individuals the rate of fixation can be predicted, the study findings explained that the higher the bone density, the greater the removal force and also the stability. However, at the beginning of the discussion it was mentioned that the major problem of bones was their low density in their study.

Kohn et al[8] re-examined the effect of higher torque during wrenching and their results illustrated that the higher the insertion torque, the greater the fixation of the screw. But it cannot be a good predictor for the strength of the repaired bone. On the other hand, Kohn's study showed that the higher the length of the screw, the better the treatment will be. It should be noted that it is impossible to use longer screws in all the areas, especially in the face.

To the best of our knowledge, no study is available on grooved fixation screws made of intelligent alloys. While intelligent alloys are used in the medical and dental industries, almost all the studies have acknowledged the benefits of using these alloys over the conventional treatment methods. Liu et al[32] examined SMA fixation screws in their study.

Although in the current study, removal torque of screws made of SMA did not yield the expected results, their special and unique design distinguishes them from those used in other studies. Considering the limitations of this study, by eliminating the drawbacks, the desired and expected results might be achieved in future studies.

Conclusion

New ideas have always been associated with success and failure. The passage of time and further studies with fewer errors are necessary to achieve the desired results. Within its limitations, this study revealed that the torque needed for opening the designed grooved screws made of shape memory alloys was greater in comparison with the currently available fixation screws; however, the difference was not statistically significant.

References

1. Hupp J, Ellis E, Tueker M. *Contemporary Oral and Maxillofacial Surgery*, 5th ed. Chicago: Mosby; 2008. p. 52-5.
2. Hitchon PW, Brenton MD, Coppes JK, From AM, Torner JC. Factors affecting the pullout strength of self-drilling and self-tapping anterior cervical screws. *Spine* 2003;28:9-13. doi: 10.1097/00007632-200301010-00004
3. Schultheiss M, Claes L, Wilke HJ, Kinzl L, Hartwig E. Enhanced primary stability through additional cementable cannulated rescue screw for anterior thoracolumbar plate application. *J Neurosurg* 2003;98(1 Suppl):50-5. doi: 10.3171/spi.2003.98.1.0050
4. Pitzen T, Franta F, Barbier D, Steudel WI. Insertion torque and pullout force of rescue screws for anterior cervical plate fixation in a fatigued initial pilot hole. *J Neurosurg Spine* 2004;1:198-201. doi: 10.3171/spi.2004.1.2.0198
5. Ricci WM, Tornetta P, Petteys T, Gerlach D, Cartner J, Walker Z, et al. A comparison of screw insertion torque and pullout strength. *J Orthop Trauma* 2010;24:374-8. doi: 10.1097/bot.0b013e3181c4a655
6. Daftari TK, Horton WC, Hutton WC. Correlations between screw hole preparation, torque of insertion, and pullout strength for spinal screws. *J Spinal Disord* 1994;7:139-45. doi: 10.1097/00002517-199407020-00007
7. Okuyama K, Abe E, Suzuki T, Tamura Y, Chiba M, Sato K. Can insertional torque predict screw loosening and related failures? An in vivo study of pedicle screw fixation augmenting posterior lumbar interbody fusion. *Spine (Phila Pa 1976)* 2000;25:858-64. doi: 10.1097/00007632-200004010-00015
8. Kohn D, Rose C. Primary stability of interference screw fixation. Influence of screw diameter and insertion torque. *Am J Sports Med* 1994;22:334-8. doi: 10.1177/036354659402200307
9. Jung MY, Shin DA, Hahn IB, Kim TG, Huh R, Chung SS, et al. Serious complication of cement augmentation for damaged pilot hole. *Yonsei Med J* 2010;51:466-8. doi: 10.3349/ymj.2010.51.3.466
10. Schultheiss M, Claes L, Wilke HJ, Kinzl L, Hartwig E. Enhanced primary stability through additional cementable-cannulated rescue screw for anterior thoracolumbar plate application. *J Neurosurg* 2003;98(1 Suppl): 50-5. doi: 10.3171/spi.2003.98.1.0050
11. Mantovani D. Shape memory materials for biomedical application. *Advanced Engineering Materials* 2002;4:91-104. doi: 10.1002/1527-2648(200203)4:3%3C91::aid-adem91%3E3.0.co;2-b
12. Azadi Boroujeni B. Constitutive modeling and finite element analysis of the dynamic behavior of shape memory alloys. A thesis submitted in partial fulfillment of the requirements for the degree of Doctor of Philosophy. The Faculty of Graduates Studies (Mechanical Engineering). The University of British Colombia. 2008.
13. Zhang C, Madbouly SA, Kessler MR.Biobased polyurethanes prepared from different vegetable oils. *ACS Appl Mater Interfaces* 2015;7:1226-33.
14. Yu YJ, Infanger S, Grunlan MA, Maitland DJ. Silicone Membranes to Inhibit Water Uptake into Thermoset Polyurethane Shape-Memory Polymer Conductive Composites. *J Appl Polym Sci* 2015;5:132.
15. Tallury SS, Spontak RJ, Pasquinelli MA. Dissipative particle dynamics of triblock copolymer melts: a midblock conformational study at moderate segregation. *J Chem Phys* 2014:141:244911.

16. Ashrafi MJ, Arghavani J, Naghdabadi R, Sohrabpour S. A 3-D constitutive model for pressure-dependent phase transformation of porous shape memory alloys. *J Mech Behav Biomed Mater* 2015;42:292-310.
17. Itro A, Garau V, Tartar GP, Colella G. Experience with a rigid fixation device in maxillofacial surgery using shape memory clips. *Minerva Stomatol* 1997;46:381-9.
18. Tomitsuka K. Study of mechanical properties of shape memory alloy plate for internal fixation of jaws. *Kokubyo Gakkai Zasshi* 1991;58:59-73. doi: 10.5357/koubyou.58.59
19. Idelsohn S, Pena J, Lacroix D, Planell JA, Gil FJ, Arcas A. Continuous mandibular distraction osteogenesis using superelastic shape memory alloy (SMA). *J Mater Sci Mater Med* 2004;15:541-6. doi: 10.1023/b:jmsm.0000021135.72288.8f
20. Zhou HZ, Hu M, Hu KJ, Yao J, Liu YP. Transport distraction osteogenesis using nitinol spring: an exploration in canine mandible. *J Craniofac Surg* 2006;17:943-9. doi: 10.1097/01.scs.0000236437.74850.26
21. Ishii N, Oyama T. Early maxillary alignment of unilateral cleft lip and palate using beta-titanium wire. *J Craniofac Surg* 2008;19:861-5. doi: 10.1097/scs.0b013e31816b1b06
22. Pautke C, Kolk A, Brokate M, Wehrstedt JC, Kneissl F, Miethke T, et al. Development of novel implant abutments using the shape memory alloy nitinol: preliminary results. *Int J Oral Maxillofac Implants* 2009;24:477-83.
23. Wang Y, Zheng G, Zhang X, Xiao S, Wang Z. Temporary use of shape memory spinal rod in the treatment of scoliosis. *Eur Spine J* 2011;20:118-22. doi: 10.1007/s00586-010-1514-7
24. Yeung KW, Lu WW, Luk KD, Cheung KM. Mechanical testing of a intelligent spinal implant locking mechanism based on nickel-titanium alloys. *Spine* 2006;31:2296-303. doi: 10.1097/01.brs.0000238967.82799.3d
25. Burstone CJ. *Current Concepts and Techniques*, Vol. 2. St. Louis: Mosby; 1985. p. 89-96.
26. Bauer J, Efe T, Herdrich S, Gotzen L, El-Zayat BF, Schmitt J, et al. Torsional stability of interference screws derived from bovine bone--a biomechanical study. *BMC Musculoskelet Disord* 2010;11:82. doi: 10.1186/1471-2474-11-82
27. Engelke W, Müller A, Decco OA, Rau MJ, Cura AC, Ruscio ML, et al. Displacement of dental implants in trabecular bone under a static lateral load in fresh bovine bone. *Clin Implant Dent Relat Res* 2013;15:160-5. doi: 10.1111/j.1708-8208.2011.00338.x
28. Zhou HZ, Hu M, Yao J, Ma L. Rapid lengthening of rabbit mandibular ramus by using nitinol spring: a preliminary study. *J Craniofac Surg* 2004;15:725-9. doi: 10.1097/00001665-200409000-00005
29. Newman MG, Yakei H, Klokkevold P, Carranza F. *Clinical Periodontology*, 11th edn. Philadelphia: Elsevier; 2011. p. 627.
30. Youngson CC, Barclay CW. A pilot study of intraoral temperature changes. *Clin Oral Investig* 2000;4:183-9. doi: 10.1007/s007840000060
31. Barclay CW, Spence D, Laird WR. Intra-oral temperatures during function. *J Oral Rehabil* 2005;32:886-94. doi: 10.1111/j.1365-2842.2005.01509.x
32. Liu XW, Wang PF, Fu QG, Zhang CC, Xu SG, Su JC, et al. Treatment of Seinsheimer type V subtrochanteric femoral fractures with dynamic hip screw and shape memory alloy bow-teeth screw. *Zhongguo Gu Shang* 2010;23:288-90.