Mandibular reconstruction revisited; on modeling and fixation techniques of the fibular free flap

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Chapter 5

Fixation methods in mandibular reconstruction using fibula grafts: A comparative study into the relative strength of three different types of osteosynthesis

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Introduction

After oncological resection of the mandible, reconstruction using a free vascularized bone graft has become the predominant treatment of choice. The free fibula graft is most commonly used (1). In order to restore the contour of the mandible it is necessary to make one or more osteotomies in the bone graft, depending on the size of the defect. The use of a reconstruction plate as a template alleviates the problem of positioning and fixation between the residual bone segments. These segments are fixed to the reconstruction plate with two screws (2). The trend in free bone graft surgery is however to avoid large plates in favor of an optimal minimum of fixation material of a much smaller dimension, like miniplates.

In recent years nickel titanium (NiTi) alloy staples suitable for use in mandibular fracture osteosynthesis have become available. The advantage these staples have over plates lies in the fact that they have only minimal effect on the periostial blood supply. Little dissection is necessary to place drill holes and the area of contact between fixation material and periostium is minimal. A particular characteristic of the NiTi alloy is the shape memory effect, induced by temperature changes. The staples can be modeled in a cooled state and on warming up, regain their original form, exerting compression.

To assess the usefulness of this mode of fixation, the Memory staple was compared with two other fixation techniques currently in use; stainless steel interosseous wires and 2.0 titanium miniplates. Their relative strength on compression and under torsional stress was measured using a fibula model.

Method

Mechanical testing osteosynthesis techniques in mandibular reconstruction is possible using in vitro models or cadaveric specimens. To avoid the inherent interspecimen variation of these studies, a model with an approximate geometry of a fibula was carved from a piece of beech dowel to imitate a mandibular reconstruction. This removes questions regarding the ostheosynthesis material model interface. Beech dowel was chosen because it is particularly homogenous and its elasticity modulus is comparable to that of bone (3). Two saw cuts at an angle of 50° were made and a wedge shaped piece of material removed. The cut surfaces of the two halves were placed against each other. Fixation occurred in the anterolateral (= future frontal) aspect and the dorsal (= future occlusal) aspects at an angle of just under 90° in relation to each other. All the fixation materials were sited in an identical location. Interosseous stainless steel wire and Memory staples were introduced through identically situated drill holes.

The following materials were employed in the fixation methods:

* Amp® Memory staples 12 mm wide with a leg length of 13 and 16 mm respectively (Amp, Villeurbanne, France). The 16 mm long staple was positioned on the anterolateral aspect and the 13 mm on the dorsal aspect. These lengths ensured that the points of the staples were buried just underneath the surface of the wood. Using a drill gauge, two holes of 2.2 mm for each staple were drilled at a distance of 12 mm from each other. The staple was positioned after having been modeled in the cooled state (-15°C).

* Interosseous wires of stainless steel 0.5 mm in diameter were used. They were tightened to a level where there was no discernible movement between the two parts.

* Synthes® (Mathys Medical Ltd., Bettlach, Switzerland) Craniofacial 4 hole titanium plates 0.9 mm thick and 25 mm long, fixed with four 2.0 self-tapping cortical screws of a minimum length of 8 mm (Fig. 1).

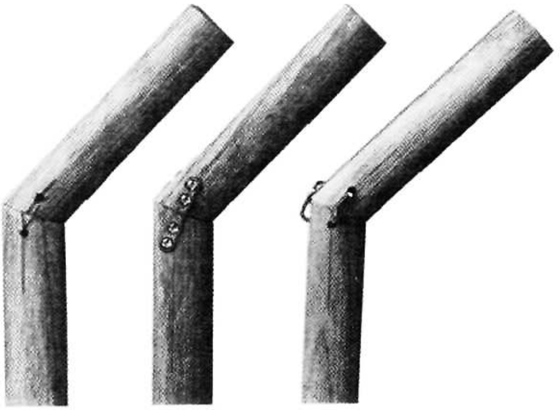

Fig. 1. Three forms of fixation (interosseous wire, Synthes® miniplates and Amp® Memory staples), applied to the fibula model. Note that all the fixation materials were sited at an identical location.

Measurements were carried out at a temperature of 35°C.

Two different types of measurement per fixation method were performed; each of which was repeated five times. During the compression tests forces were exerted on the experimental model using a compression/distraction bench micrometer (Tensometer type "W", Monsanto Instruments Akron, OH, USA) with a displacement speed of 3.18 mm/min. Compression on the model resulted in distraction at the site of the fixation material (Fig. 2).

The measurements were registered by a line amplifier (Philips PR 9340 (Philips Electronics N.V., Eindhoven, The Netherlands)) on an XY recorder (Philips PM 8134). All compression measurements were continued until the fixation material broke or extruded through the wood.

For torsion tests the model was so positioned in a jig that the axis of rotation ran through the center of the contact area between the two halves of the dowel (Fig. 3).

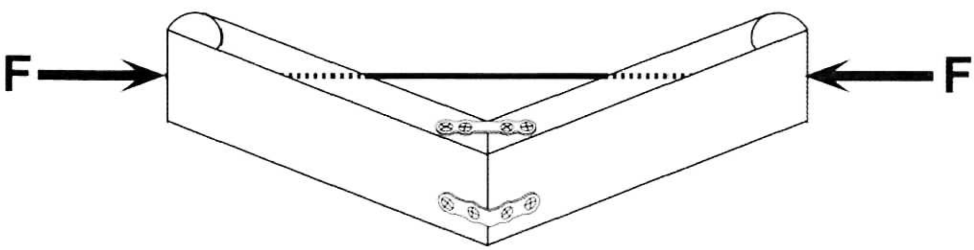

Fig. 2. Schematic representation of the compression test setup. A force is exerted (arrows) along the line connecting both ends of the model. This results in bending in the contact area and thus distraction on the fixation material.

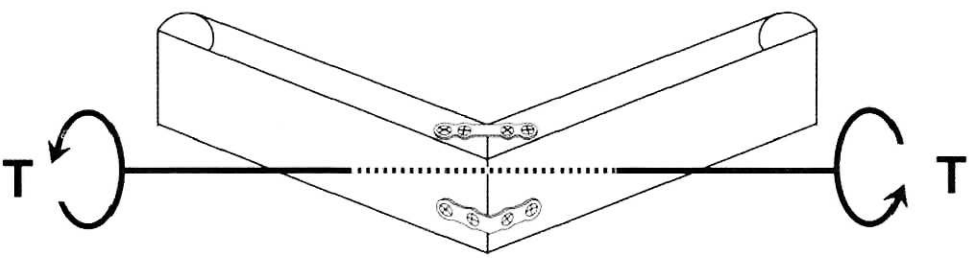

Fig. 3. Schematic representation of the torsion test setup. The model was so positioned in a jig that the axis of rotation ran through the center of the contact area between the two halves of the dowel. The two halves are rotated in opposing directions (arrows)

The test was carried out with continuous rotation of 0.3 degrees per second in a torsion bench developed by the Department of Medical Technical Development at the Academic Medical Center, Amsterdam, The Netherlands. Torsion measurement was carried out to a maximum of 85° (arbitrarily chosen).
The measurements were registered by a line amplifier (Philips PR 9340) on an XY recorder (Philips PM 8134).

For mutual comparison of the curves the magnitude of the forces (F_4 in Newton) at a displacement of 4 mm was used.
For the torque the value (T_{35} in Newton.meter) at 35° is taken.
As measures of rigidity (Rc) or angular stiffness (As) the displacement (mm) or the rotation (degrees) and their driving agent in N or N.m were computed from the linear part of the curves produced. The manner of the failure of the fixation material in relation to the forces exerted was recorded. The data were analyzed using the statistical program InStat (Graphpad Software, San Diego CA) on a Macintosh Power PC 8500/120 (Apple Computer Inc., Cupertino, CA, USA).

Results

Compression tests

In all cases the fixation in the frontal aspect was the first to fail. The stainless steel wires underwent a short period of elastic deformation followed by gradual untwisting. This occurred in the nearly horizontal part of the curve (Fig. 4).

The Memory staples deform first in the bridge area followed by bending of the legs and their sliding out of the drill holes. In two cases the staples fractured in the eye-shaped bridge area. The plates showed deformation of the first hole adjacent to the contact area, after which in three cases, the screws broke out of the wood and in the remaining cases the plate fractured. The force (F_4), that can be exerted on the fixation material at a displacement of 4 mm together with standard deviations are reproduced in Table 1.

The reliability of the measurements of the interosseous wire, staples and plates is shown by the Student's t test (P<0.0001). The Tukey-Kramer multi comparison test shows a difference between the F of all three osteo-synthesis methods. The biggest differences are between the interosseous wires and the plates (P<0.0001) and between the staples and the plates.

We could not demonstrate a difference between the rigidity (Rc) of the interosseous wires and the staples at a 5% level of significance. However, there is clear difference when comparing interosseous wires and staples to the plates (P<0.001).

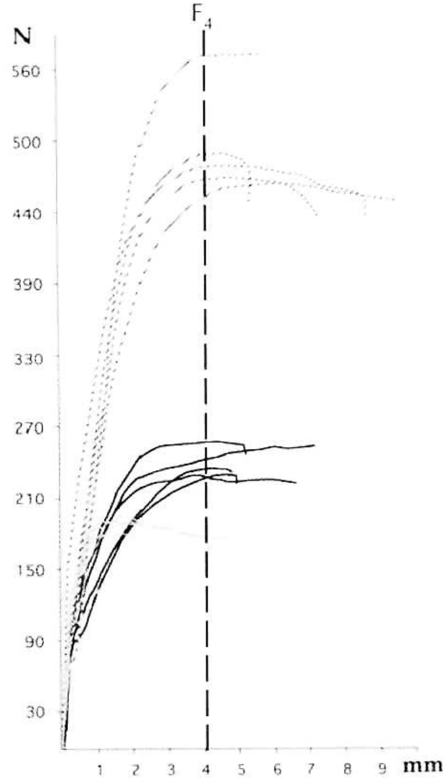

Fig. 4. Compression test force-displacement curves. The rigidity was deduced from the slope of the linear portion of the load deflection curve.
Legend: —— miniplates, —— Memory staples, —— interosseous wires. Displacement in mm, exerted force in Newton (N).

Table 1. Results of Compression testing. Mean of the value of force (F_4) in N and Rigidity (Rc) in N/meter. Standard deviation in parentheses.

Fixation Material	F_4(sd)	Rc(sd)
Interosseous wires	**176.4**(13.9)	**8.3**(2.4)
Memory staples	**238.4**(10.0)	**6.9**(0.8)
Miniplates	**490.2**(49.4)	**21.4**(3.3)

Torsion Tests

The torsion tests show a virtually identical pattern of fixation material deformation. Staple fracture and screw extrusion did not occur. The test results are shown in Table 2.

Initially the staples showed torsion between the legs followed by deformation of the eye shaped bridge area. Here the curve runs parallel to the curve formed by the titanium plates (Fig. 5).

The interosseous wires untwisted gradually. The interosseous wire did not break or cut through the surface of the wood. A high degree of reproducibility was seen in the torsion tests on all three modes of fixation. This is true for both values of torsional strength at 35° of rotation (T_{35}) and degree of rigidity (Rt) ($P<0.001$).

The torsional strength and angular stiffness of the interosseous wires differ clearly on the Tukey-Kramer multi comparisons test in relation to both staples and plates ($P<0.0001$). The value of torsional strength of plates at 35° of rotation is at approximately twice as high as the torsional strength of staples. The difference in the rigidity of staples and plates is marginal. It is noteworthy that when exertion of forces on the staple fixation was discontinued, the staples returned to almost exactly their original position. It would appear that up to 85° of rotation, the extent of plastic deformation incurred by Memory staples is limited, unlike that of the titanium plates.

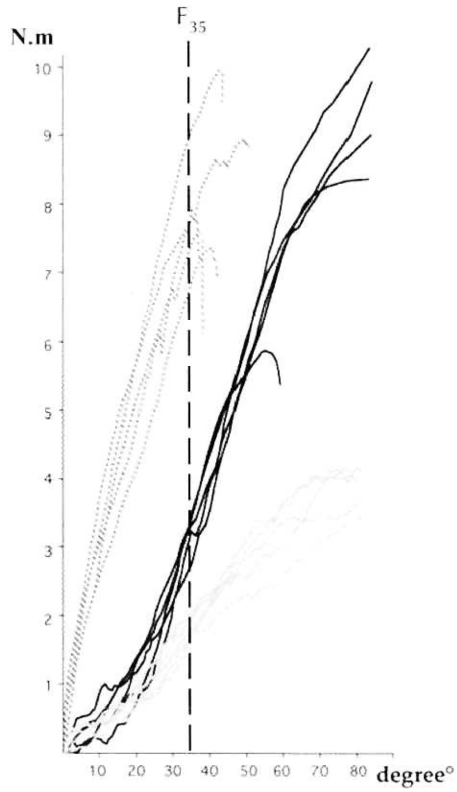

Fig. 5. Torque-rotation curves. The angular stiffness was deduced from the slope of the linear portion of the load deflection curve.
Legend: —— miniplates, ——Memory staples, —— interosseous wires. Displacement in mm, exerted force in Newton (N).

Table 2. Results of Torsion testing. Mean of the value of torque (T_{35}) in N.m and angular stiffness (As) in N.m/degree. Standard deviation in parentheses.

Fixation Material	T_{35}(sd)	As(sd)
Interosseous wires	**1.7**(0.2)	**0.1**(0.1)
Memory staples	**3,2**(0.3)	**0.2**(0.0)
Miniplates	**7,8**(0,9)	**0.2**(0.5)

Discussion

Currently, the fibula is the most commonly used free flap bone graft in mandibular reconstruction. It is long enough for an extensive jaw reconstruction and can be divided into relatively small segments of up to 2 cm without compromising vascularization (4).

The number of osteotomies and the length of the bone segments have a very definite effect on the degree of difficulty of the fixation method.

The larger the defect of the jaw, the more osteotomies are necessary and the more complex the procedure becomes. Small changes in the length or angle of the bone segments have a direct effect on the position and projection of the chin.

Fixation techniques can be divided into rigid and non-rigid forms of fixation.

Rigid fixation methods use screws, THORP 4.0, plates, vanous reconstruction plates and screws as well as miniplates.

A reconstruction plate can be used as a final fixation and as a template fixed to the remaining mandibular abutments, within which the bone graft is given a contour and to which the graft can be fixed by one or more screws. The use of a reconstruction plate greatly simplifies the procedure. The disadvantage of this procedure however, is the relatively large size of plate and screws. A high profile plate can distort the contour of the jaw. This is particularly true of central reconstructions. The fibula segments are sited within the plate, which has already been shaped to the original contour of the jaw. This always results in an ossal reconstruction that is smaller than the original jaw. 'Stress shielding' by the rigid reconstruction plate can sometimes delay consolidation (5). The fixation material may hamper the placement of osseointegrating implants for a dental prosthesis. Secondary positioning of implants often necessitates the removal of plate and screws (6, 7, 8). Miniplates may also have this disadvantage.

Since its introduction by Hidalgo in 1989, the use of the miniplate for mandibular reconstruction has proved to be a reliable method (9). These small plates are easily modeled and enable an accurate reconstruction of the contour of the jaw using small segments of bone. Two four-hole plates per osteotomy position are usually necessary also for fixation of the neomandibula to the remaining mandible. The use of small bone segments always carries the risk of devascularization and necrosis due to compression of the segmental periostial vessels (10, 11).

Non-rigid fixation utilizes interosseous wires, sometimes augmented by Kirschner wire fixation. When more than one osteotomy is necessary, modeling and fixation by wires can be a laborious and difficult procedure. The success of the wiring technique depends more on optimal contact between the bone segments than do the other procedures. However, studies were not able to demonstrate any difference in the rate of bone healing between the rigid and non rigid fixation methods in mandibular reconstruction (12, 13), in contrast with interosseous wire repair of fractures of the edentulous mandible. The reason for this difference is probably that a vascularized bone graft has a much richer blood supply than the fractured atrophic edentulous mandible (14).

In recent years an alternative fixation method has become available. This is the nickel titanium Memory staple (BRI®, BIO Research Innovations, La Seyne, France and Amp®, Amp, France). When an object made of this nickel titanium alloy is cooled below a critical temperature the metal matrix goes into the so-called martenistic phase. In this condition the metal matrix can be manipulated by bending without being damaged. When the object is warmed up, the metal matrix regains its former configuration and the original shape is restored. For medical use the nickel titanium proportion is such that the critical temperature is below 0°C, and the original crystal formation returns at body temperature.

BRI® and Amp® Memory staples are supplied with the points bent towards each other and an S-shaped or oval bridge. After cooling to below the critical temperature the legs are straightened until they are parallel and the S shape or oval straightened out. If proper leg length is chosen in order to penetrate both cortices, bicortical pressure will be created on reheating. Comparison of nickel titanium alloys with other titanium or stainless steel alloys in medical usage shows them to have a low elasticity modulus, high tensile strength and a high fatigue strength (15). This results in continuous dynamic compression at the site of the osteotomy. The biocompatability of these nickel containing alloys is comparable with that of titanium (16).

To date in the field of maxillo facial surgery, Memory staples have been used only in the treatment of fractures (17, 18).

In view of the good reports of Memory staples in these and other bone fixation techniques, we considered their use as a potential method of fixation of fibular osteotomies in mandibular reconstruction.

Taking the titanium miniplate method of fixation as one extreme and the cerclage method of fixation as the other one, within whose parameters good bone healing is to be expected, the results given by the Memory staples fall somewhere in between. Their ability to withstand torsional stresses is particularly important in central reconstructions and is comparable to that of the titanium miniplate. Memory staples have the additional advantage of exerting dynamic compression at the site of osteosynthesis. Should external forces cause the fixation to become distorted, the use of the staple would cause it to return to its original position. In the same situation, interosseous wire fixation would untwist and loose its stability completely.

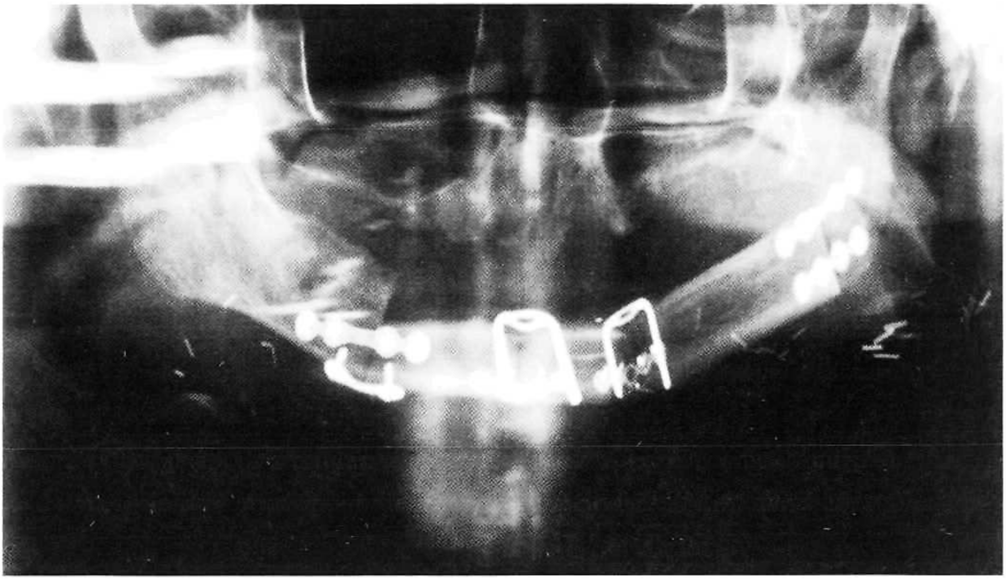

Fig. 6. Early postoperative panorex view of a patient following segmental mandibular reconstruction with a fibula free flap. The two fibular osteotomies were fixed with Memory staples and Synthes® 1.5 titanium miniplates. At that time no custom made staples were available for use in the frontal plane.

Miniplates exhibit plastic deformation and thus do not regain their original shape. This causes loss of bone contact resulting in reduced stability and slower consolidation.

Staples have a number of advantages over the titanium miniplates. Time is saved because of the reduced number of necessary surgical steps (the drilling of multiple holes, bending of plates and placing of screws all become redundant). The periosteum is only minimally compromised as the staples are small and have only a limited area of bone contact. Devascularisation of the smaller bone segments is unlikely. The bulk of the fixation material is less thus reducing the need for its removal in case of dental implants.

A disadvantage of the staple is its height in the frontal plane. In fixations of an angle less than 20°, the staple clearly juts out. The degree to which it juts out is however, limited. At an angle of 50° this corresponds with approximately 3 mm which is comparable to the height of a reconstruction plate and screws. This problem can be overcome by making staples with a preformed angle in the bridge area.

In setting up the trials, a model made from material independent of the variations found in cadaver material was deliberately chosen. Cadaver material is not homogenous and differs widely in quality. The geometry of cadaver material also varies, making identical and comparable positioning of the osteosynthetic material impossible.

It is feasible to demonstrate the differences in the strength of the various types of osteosynthetic material very effectively on a beech dowel model. Its force displacement curves show that it is strong enough to resist the forces exercised upon it. It was only during miniplate compression tests that the screws were occasionally pulled out of the wood. This was accompanied by the extreme plastic deformation of the screw holes. This is never observed in vivo. The moment of failure (miniplate fracture, loosening of screws) is much further on in the process than the point at which plastic deformation begins (Fig. 4).

No attempt to obtain absolute values was made, as in practice they are almost impossible to apply. The forces that act on a reconstructed mandible are highly variable and difficult to quantify. Our experimental model appeared to deliver values high enough to be relevant with a high degree of reproducibility.

Our results and recent clinical experience (Fig. 6) have shown that the Memory staple has a place as a suitable method of fixation in mandibular reconstruction using a free flap vascularized fibula bone graft.

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