In-vitro Comparison of Biomechanical Efficiency of Three Cannulated Screws for Arthrodesis of the Hindfoot

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ABSTRACT

Background: Sufficient inter-fragmentary compression is helpful to achieve successful bony fusion in hindfoot arthrodesis using internal fixation by screws. Beside bone quality, the design of a screw influences inter-fragmentary compression. Compresive force is achievable for any kind of screw system; however, the primary deformation of the bone is different for the different screw systems. The work necessary to achieve compressive force for primary stability was measured for different screw systems and compared to an AO screw with washer. Materials and Methods: The compressive force was determined as a function of screw advancement for 3 different cannulated screw types (7.3-mm AO screw with and without washer, the 6.5-mm Herbert screw and the 6.5-mm Ideal Compression Screw (I.CO.S) using different synthetic bone density (0.16, 0.24, 0.48 g/ccm). Compressive force was measured indirectly, via screw tension measurement with strain gauges. Results: We calculated the work to reach a limit of 60 N and the corresponding ratios to the value of the golden standard: I.CO.S (35.2%), Herbert (89.0%), AO screw without washer (116%). Conclusion: All screw systems yielded acceptable results but the I.COS did produce greater compresion. The essential differences were the primary deformation of the bone before reaching the sufficient compressive force for primary stability.

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Key Words: Cannulated Screw; Compression; Biomechanics; Synthetic Bone

INTRODUCTION

Isolated or combined arthrodesis is the standard procedure for treatment of arthritis, deformities and instabilities at the hindfoot. Internal fixation with bone screws is the most frequently used technique, first described by Wagner and Pock. Several authors modified this technique. Generally, adequate primary stability and inter-fragmentary compression is necessary to achieve successful bony fusion. Non union is the most prevalent complication, which was reported to occur up to 16% for isolated subtalar arthrodesis, 5% to 26% for isolated tibiotalar arthrodesis and 3% to 18% for triple arthrodesis.

Besides bone quality, the design of a screw strongly influences the work needed to achieve inter-fragmentary compression for healing. The necessary pressure can be obtained with several different screw systems. Application of a torque using a screwdriver generates a tensile stress in the screw, which is at equilibrium with the compressive force between the fusion parts. From a mechanical point of view, the main problem is the load transfer at the screw-bone interfaces. Overload of the load-transfer area may cause failure at the interface with a loss of compressive force

The purpose of this study was to measure the work necessary to reach a certain compressive force using different screw types, which are clinically used for hindfoot arthrodesis. We feel that maximum compression is not a practical parameter for comparison between distinct screw systems because it exceeds the necessary force for any screw clinically. Therefore, the goal of this study was to determine which screw type can create adequate compression with the minimum of work, which would thus lead to a minimum of bone deformation. The primary deformation of bone causes rheological and biomechanical reactions (Wolff’s law) after insertion, which should be kept as small as possible.

MATERIALS AND METHODS

Screws

Three different cannulated screws were tested: the AO/ASIF 7.3-mm with 16-mm thread length (Synthes, Davos, Switzerland), the Herbert 6.5-mm bone screw (Zimmer Inc., Warsaw, IN) and the Ideal Compression Screw (I.CO.S)
6.5-mm (Newdeal Inc., Vienne, France). All were made of titanium and had a length of 90 mm (Figure 1). The tests of the AO screw were conducted with and without a washer. The AO screw has a classic screw design with a head on the proximal end and a thread at the distal end. The two other screws both have a leading and a tailing thread of different pitches at both ends, creating compression.

Study Design

The principal experimental strategy was to evaluate the deformation of the bone. Equ. 1.1 shows the functional relationship between the compression force “$F_c$”, the primary deformation, presented by the characteristic measure “$\Delta l$” (Figure 2) and the given screw-systems, marked by the parameter “$p$.”

$$F_c = \varphi(\Delta l, p)$$

(1.1)

The method to measure the compression force “$F_c$” is chosen according to the equilibrium conditions: the tension in the screw is equal to the resultant compressive force in the contact surface of the bone. This means that the compressive force “$F_c$” is determined by measuring the tension in the screw using the strain-gauge method.\(^\text{21}\) This method is very sensitive and does not depend on the geometry (size and form) of the contact surface.

Due to the comparative nature of the study this function was measured experimentally using synthetic bone for each screw system in order to obtain the work “$W$” (Equ. 1.2) necessary to reach the compressive force “$F_c$” at the arthrodesis site. The AO screw with washer served as the control for this study. We compared the results of the other screws and the AO screw without washer to this screw.

Primary deformation increases with increasing work “$W$”. Overload may lead to increasing deformation of the load transfer areas and to decreased primary stability due to a failure of the threads. From the mechanical point of view, the screw system which generated a certain compression force with the lowest amount of work would be preferred.

$$W = \int_{\Delta l} F_c.d(\Delta l)$$

(1.2)

Previous stress optical investigations demonstrate that the load is concentrated in the transfer areas and the accompanying primary deformation is nonhomogeneously distributed.\(^\text{11}\) Figure 3A shows the very concentrated and nonhomogeneous fringe distribution for AO screw in the load transfer area in the vicinity of the screw head. Figure 3B shows that the load transfer at the screw threads is better distributed because of the great difference of stiffness of the materials (stress optical material/screw). Of course, this result cannot be directly quantitatively transferred to the combination screw/bone, but the qualitative information is useful.

Therefore, the measurable overall deformation “$\Delta l$” is roughly divided into two zones of deformation: the load-transfer areas at the end of the screws (a) and that in the
middle-range (b), which is nearly homogeneous. The non-linear force displacement relationship is divided into three phases (Figure 2). Phase I is mostly caused by the primary deformation of the load-transfer areas due to the high contact stresses, especially by using AO screws. The linear phase II is dominated by deformation of the specimen reacting as a unit due to homogenization during phase I. The non-linear phase III is up to failure in the load-transfer areas.

Synthetic Bone Models
Due to the wide range of bone density, three densities of solid rigid polyurethane foam (ASTM F-1839 Standard, Sawbone Europe AB, Malmö, Sweden) were used: 0.16 g/ccm (low density), 0.24 g/ccm (medium density) and 0.48 g/ccm (high density).\textsuperscript{2, 3} Specimens with a size of 40 × 20 × 20 mm were produced.

Strain Gauges
The original screwdrivers (biaxial: measurement of torsion only) and the screws (rosette: measurement of torsion and axial force) were instrumented with strain gauge-bridges (EA 30-125RD-350/option L, Measurements Group Vishay, Lochham, Germany). The measurements were done using a calibrated multimeter method (EDC 120 Walter & Bai company).\textsuperscript{21}

Experimental Procedure and Testing Apparatus
The gauge factors relating the electrical signal of the strain gauge bridges to the mechanical quantities to be measured were determined in calibration tests.\textsuperscript{22} Strain gauge bridges measured the achieved tensile load of the screw. The specimens were fixed to a holder, which was also supplied with a strain gauge bridge for controlling the measurements of the axial force applied with the screwdriver (Figure 4). Following the recommendations of the manufacturers all screws were inserted into the synthetic bone models using the original instruments. The screwdrivers were also instrumented to measure torque (Figure 4). The compressive

Fig. 3: Stress optical image of AO screw with washer (A), and of Herbert screw at the proximal thread of the screw (B). The fringe distributions of AO screw without washer is similar to (A), the distribution of I.C.O.S. is similar to (B).
force was recorded as a function of screw advancement $\Delta l$ (Figure 3) measured by the thread lead and the angle of rotation. The tests were continued with a constant rate of two revolutions per minute. The screw-driver was supported by the holder and the influence of the flexible shaft was negligible.

**Statistical Methods**

Although this study was primarily concerned with the load-transfer mechanism of the screw types under investigation, a two-way analysis of variance (ANOVA) was performed followed by multiple comparison test (Mathematica 5.1 additional package, Wolfram Research, Oxford, UK). A level of 95% was considered significant ($p < 0.05$).

**RESULTS**

We used 3 different densities and 4 types of screws (Figure 5). The size of the nonlinear range strongly increased with decreasing density of the Sawbone, especially for the AO screw without washer and in a more moderate form for the AO screw with washer and the Herbert screw in all three densities. The force versus screw diagram for the AO-screw without washer shows, especially for the lowest density, a nonlinear behavior (0 to 20 N) due to the compression in the near vicinity of the load transfer area (Figures 3A and 5A). This range is followed by a practical linear range up to about 80 N. Further screw advancement produces some rearrangements in the structure followed by stiffening again due to these rearrangements. A similar behavior in the range of 80 to 100 N was found for the AO-screw with washer. These processes cannot be found for higher densities, indicating that special care is necessary for very low densities. For each density the overall deformation to reach the force of 60 N was measured and the work necessary was calculated (Equ. 1,2), listed in Table 1.

![Figure 5: Relationship between displacements and the compressive force for three different Sawbone densities (A, 0.16 g/ccm; B, 0.24 g/ccm; C, 0.48 g/ccm) and four types of arthrodesis screws (• I.CO.S, ▲ Herbert Screw, ■ AO screw plus washer, and ♦ AO screw without washer). Values found at a level of compressive force of 60 N are given in Table 1.](image-url)
From the measurements we calculated the work necessary to reach 60 N for each screw type and compared it to the corresponding value of the AO screw with washer (100%): I.C.O.S (35.2%), Herbert screw (89.0%) and AO screw without washer (116%).

**DISCUSSION**

Today internal fixation with screws is the most frequently used technique for arthrodesis of the hindfoot. The maximum value of compressive force is mainly limited by the bone quality, but the design of the screw also affects compression. Overload at the screw-bone interface can damage the thread bone interface resulting in a loss of compressive force. This failure can be a significant problem during surgery. Therefore, selecting a screw system which minimizes the danger of a breakage at the bone-screw interface.

In general compressive force is applied qualitatively during surgery. However, no data about an “optimal” compressive force are available in literature. Only Hintermann et al. published data on the compressive force at the contact area of the fusion site. Therefore, we choose the reported threshold (60 N) to compare the different screw systems. In order to determine criteria for practical decisions between different screw types an “in–vitro study” was performed with 3 different screw systems, which are clinically used for hindfoot arthrodesis. The compression at the fusion site as a function of the work necessary to reach a sufficient compressive force was measured using strain gauges for three different densities of synthetic bone.

For better reproducibility, we decided to perform this study with polyurethane foam. This has been successfully used in several studies on the mechanical behaviour of screws. Cadaver bone requires a larger number of tests to achieve significance because of the high degree of cadaver bone. The influence of cortical bone with a higher density was not studied because cortical bone of the calcaneus and talus is very thin and therefore of less biomechanical relevance.

Our results show that the limit of compression given by Hintermann et al. can be reached with all three of these screw systems. However, the work necessary to reach this value of compression was quite different for the screws. The I.C.O.S showed superior results. Only with extremely poor bone quality would a clear advantage of I.C.O.S be expected in bone practice. The deformation of the specimens with I.C.O.S was much smaller than for the other screws due to better purchase of I.C.O.S. The different results of the AO screws (with and without washer) was caused by the weak resistance against countersinking of the screw head especially for low density bone. Therefore, by inserting the I.C.O.S, a lower risk of failure in the load-transfer areas and a minimum of post-insertion deformation was noted. An additional advantage of I.C.O.S is a mobile head with a special designed screw nut at the proximal thread which could produce an additional compressive force. With respect to the three tested bone densities it was not necessary to activate the additional possibility of loading offered by I.C.O.S to achieve 60 N of pressure.

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