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Biomechanical criterion for selecting cancellous bone screws: arthrodesis in the hindfoot

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The aim of the paper was to compare primary biomechanical stability of different arthrodesis screws (7.3 mm AO screw with and without washer, 6.5 mm Herbert screw and 6.5 mm Ideal Compression Screw (I.CO.S)). The work necessary to achieve an adequate compressive force with them was compared to the measurement with the AO screw with washer, because this method is for the time being the most commonly used one and is called *the golden standard*. Compressive force was measured indirectly, via screw tension measurement, with strain gauges method. From the measurements we calculated the work to reach a limit of 60 N and the ratios corresponding to the value of the golden standard: I.CO.S (35.2%), Herbert (89.0%), AO-screw without washer (116%). The I.CO.S showed superior results. Only in the case of extremely poor bone quality, a clear advantage of I.CO.S could be expected in practice.

Key words: arthrodesis, biomechanical stability, screw, compression

1. Introduction

Isolated or combined arthrodesis is a standard procedure for the treatment of primary and secondary arthritis, deformities and instabilities of the hindfoot [6]. Internal fixation with bone screws is the most frequently used technique, first described by WAGNER and POCK [22]. Other authors modified this technique [11], [12], [13], [15], [17]. Generally, an adequate primary stability and inter-fragmentary compression are required to achieve successful bony fusion and secondary stability. However, no medical data on the needed pressure are available in the literature. Only HINTERMANN published some results [10] obtained for different screw types, mostly used clinically for hindfoot arthrodesis. It will be shown that Hintermann's results correspond quite closely to our own results obtained with sawbones models of low density (16 g/cm^3). We therefore based our comparison between the different systems on a compression level of 60 N given by Hintermann. Non-union may be caused by a decreasing compression due to rheological properties of the bone or due to physiological processes, resorption of bone substance [4], [23] in overloaded areas in the load-transfer zones between the arthrodesis screws and the bone after surgery treatment.

2. Material and method

Beside bone quality the design of the screws strongly influences the work, and finally the required inter-fragmentary compression and the primary state of stress in the load transfer zone are reached. Of course, the compressive force necessary for primary stability is reachable with any kind of screw systems; however, the primary deformation of the bone and with it the initial state of stress and strain after mounting is different for the different screw systems.

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Therefore the mechanism of load transfer, the ratios between the necessary compression needed for primary stability and the work to achieve it may lead from the mechanical point of view to a criterion for using a special type of screw system under given medical conditions. This means that the purpose of this study is to compare the compression force between the fusion parts, and the work necessary to reach the needed one may yield the desired criteria.

The method and the conceptual formulation of this study are therefore given by the following question: which screw types allow the biomechanical goal to be reached with a minimum of effort? This means that the comparison is based on the ratio of achieved effect to necessary work.

In order to measure the pressure between both bony partners of an arthrodesis, a direct use of any kind of pressure-sensitive foils seems to be unuseful because this method is very sensitive to artificial effects and it strongly depends on the shape and the size of the fracture surface. However, the measurement of the tensile stresses in the screw using adequate strain gauge bridges (EA 30-125RD-350/option L, Measurements Group Vishay, Lochham, Germany) is due to equilibrium equation an integral measure for this pressure which is very sensitive and does not depend on the geometry (size and form) of the contact surface. Therefore this method is used in this study.

2.1. Preliminary photoelastic considerations

In the first step, the stress distribution in the near vicinity of the screw threads was studied based on phototelasticity. A model designed consisted of three parts; the parts at the bolt head and that at the end-thread were photoelastic ones. These parts were composed of three layers in order to analyse the isochromatic fringes in the plane of the screw. Two layers made of the photoelastic inactive acrylic glass were beside the middle-layer made of Araldite B (Tiedemann.Betz). The third part in the middle of the model, where the wires for the strain gauges were led out to the instruments, was made from acrylic glass only. The torques applied and the effects achieved were recorded.

Figure 1 shows the screw-head and the threaded end of an AO screw and demonstrates that these load transfer areas are the most endangered parts of the bone–screw system. Overloading may lead to failure of the threads and to a decreased primary stability as well. This danger is of course reduced to some degree by washers. However, also in this case the load transfer is mostly concentrated on the outer line of the washer. Analysing the stress distribution at the threaded end of the screw it can be concluded that the load transfer is better distributed over more than one thread, and may be after a small but concentrated deformation in the vicinity of the first two–three threads.

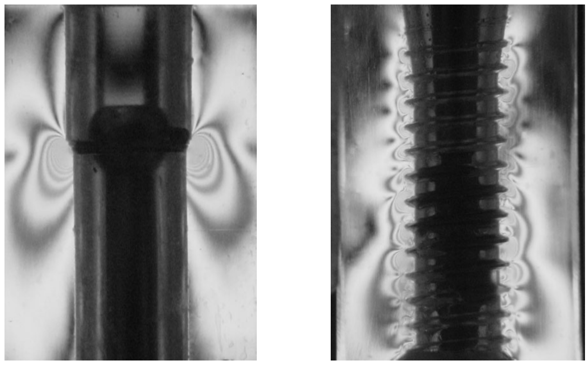


Fig. 1. Isochromatic fringes in the load transfer areas of an AO screw in the vicinity of the screw head and the threaded end

2.2. Measurement strategy

The principal experimental strategy is oriented to technical experience with the deformation behaviour of materials with low mechanical stiffness. Figure 2 shows a sketch of the reached compression force F_c as a function of a whole deformation of the system, Δl serves as its measure:

$$W = \int_{\Delta l} F_c . d(\Delta l) , \qquad (1a)$$

$$F_c = \varphi(\Delta l, p) , \qquad (1b)$$

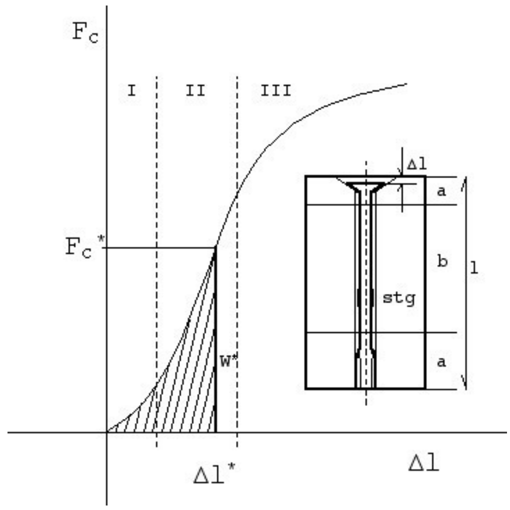


Fig. 2. Compression force as a function of a whole deformation of the system (Δl serves as its measure)

p is a parameter indicating the screw system used to reach primary stability. Due to a comparative nature of a present study this function was determined experimentally using synthetic bone models for each screw system to obtain the work W necessary to reach a certain compressive force F_c between the fusion partners. Because at present the AO screw with washer represents the *golden standard* in hindfoot arthrodesis, we compared all the results obtained with the golden standard. The comparison between the different screw systems is therefore based on the ratio

$$\lambda = W / W^*, \tag{2}$$

where W^* is the work needed for the same level of pressure when applying the golden standard.

The work W serves also as a measure of the whole primary deformation of the specimen. This state of deformation is increasing with the rise of work W which is not at all homogeneously distributed but dominantly concentrated over the load-transfer areas (see parts "a" in figure 2). The accompanying photoelastic studies showed that the deformation of the middle range of the specimen "b" in figure 2 is negligibly small and practically homogeneously distributed.

The force–displacement relationship (figure 2) is divided into three phases. Phase I is mostly caused by the deformation of the load-transfer areas due to the high contact stresses, especially by using AO screws. The practically linear phase II is dominated by the overall deformation. The non-linear phase III is due to failure in the load-transfer areas.

2.3. Synthetic bone models

In order to find out with which screw system a certain pressure level between the fusion parts of the bone is reachable with a minimum of energy input, it is sufficient to investigate this problem using synthetic Sawbones only. In accordance with literature [2], [3], [5], [21] synthetic bone models were chosen because of a high degree of reproducibility in contrast to cadaver bone tests [20]. Due to the lack of information and for the possibility of interpolation, three densities [2], [3] of solid rigid polyurethane foam (ASTM F-1839 Standard, Sawbone Europe AB, Malmö, Sweden) were used: 0.16 g/cm^3 (low density), 0.24 g/cm^3 (medium density) and 0.48 g/cm^3 (high density). Specimens of 40 × 20 × 20 mm dimensions were produced. The influence of cortical bone of a higher density was not studied because cortical bone of the calcaneus and talus is very thin [19] and therefore of less relevance.

In the second phase of the investigation, only the two photoelastic parts were replaced by Sawbone models for the load transfer areas. The specimens were fixed to a holder, which was also supplied with a strain gauge bridge (figure 3).

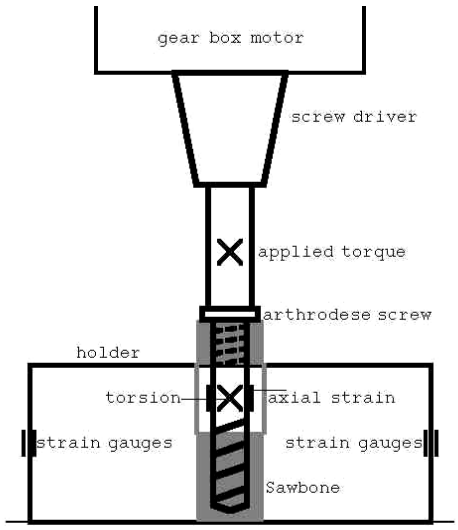


Fig. 3. Experimental arrangement for the Sawbone tests

Following the recommendations of the manufacturers all screws were inserted into the synthetic bone models by using the original instruments. The screwdrivers are equipped with strain gauges to measure the moment applied. Strain gauge bridges were applied to the screw shank to determine the achieved tensile load of the screw as a measure of the achieved compression between the two parts of the fractured bone. The compressive force was recorded as a function of screw advancement Δl (figure 2). The tests were continued at a constant rate of two revolutions per minute.

With this experimental procedure it is of course not possible to distinguish between the different contributions to the overall deformation, but equation (1) is within the accuracy reached, a reliable measure of the work put into the system. The moment applied, the forces reached in the screw, and the twisting of the screw were measured in order to quantify also the torsion of the fusion parts due to friction between the screws and the bone.

2.4. Screws

Three different cannulated screws (figure 4) 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., Warshaw; US) and the Ideal Compression Screw (I.CO.S) 6.5 mm

(Newdeal Inc., Vienne; France). All were made of titanium and have a length of 90 mm. The tests of the AO screw were conducted with and without a washer of titanium. The AO screw has a classic screw design with a head at the proximal end and a thread at the distal end. The two other screws are similarly characterized by a leading and a tailing thread with different pitches at both ends, creating compression. In contrast to the Herbert screw, the I.CO.S has an additional mobile head with a specially designed screw nut at the proximal thread. This head with an outer and internal thread could be inserted along the proximal end and allows us to produce an additional compressive force.

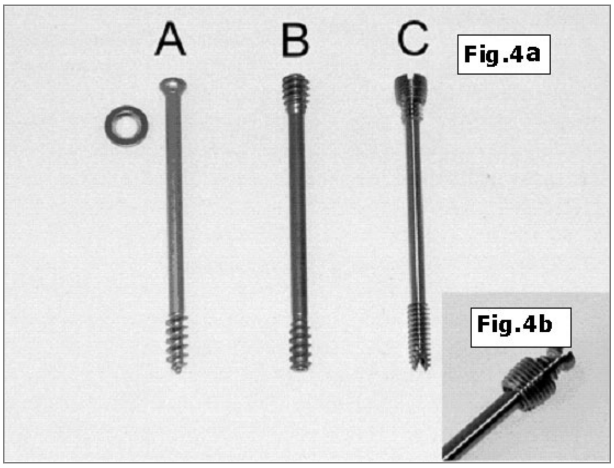


Fig. 4. Screw types used in this study:
AO +/– washer (A), Herbert (B), and I.CO.S (C)

Figure 4 shows the screw systems compared in this study. For the time being, the mostly used screw systems are as follows: the AO screw with and without washer "A", the Herbert-screw "B" and the I.CO.S "C" with a special mechanical device for an additional increasing the compression between the two parts of the fractured bone (figure 4b). For the time being, AO screw with washer is most used in arthrodesis and is known as "golden standard", therefore all results obtained for the different screw systems are compared to this standard.

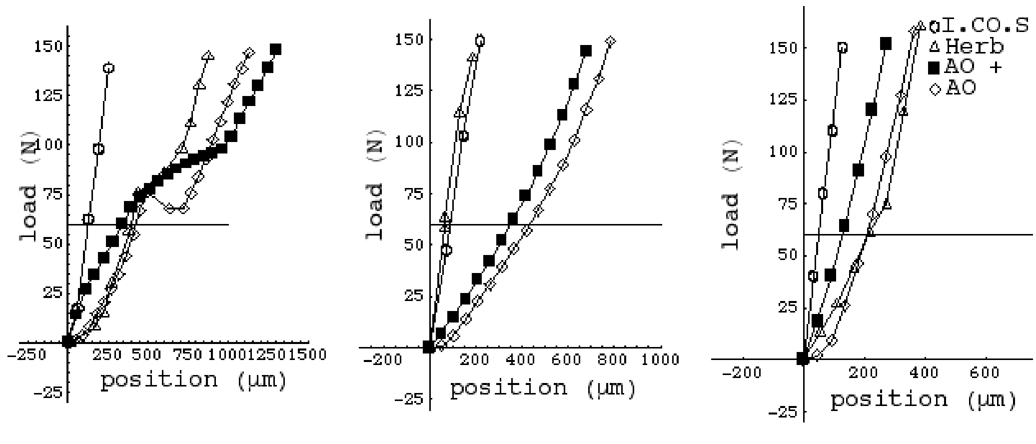
2.5. Statistical methods

In spite of the fact that this study is 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 ($p < 0.05$).

3. Results

Figure 5a–5c shows the results for the three densities (16, 24 and 48 g/cm^3) and for four screw systems (the golden standard is dashed).

The table presents the measure for the deformation Δl, the work W and the ratio λ according to equation (2), evaluated based on literature (HINTERMANN et al. [10]) for a compression force of 60 N. Figure 5a shows comparable results in the advancements of Herbert and AO screws (the latter with and without washer) obtained at the lowest density of Sawbone chosen for this investigation.



a b c

Fig. 5. Results of the Sawbone experiments as a function of Δl (according to figure 2)
for the densities used: 16 g/cm^3 (a), 24 g/cm^3 (b), and 48 g/cm^3 (c)

Table. Screw advancement (displacement) in Sawbone and the work occurring at 60 N compressive force. The results are obtained as a function of Sawbone density and screw type and expressed as mean (SEM)

Density (g/cm^3)	Screw	AO	AO+	Herbert	I.CO.S
16	Displacement (mm)	0.410 (0.040)	0.340 (0.038)	0.400 (0.039)	0.120 (0.021)
	Work (Nmm)	10.46 (1.85)	10.20 (1.62)	8.40 (1.19)	3.60 (0.78)
	Ratio (%)	97.5(33)	100	82.4(25)	35.3(13)
24	Displacement (mm)	0.430 (0.051)	0.350 (0.031)	0.090* (0.04)	0.095 (0.010)
	Work (Nmm)	10.97 (1,90)	9.98 (1.03)	2.70 (0.90)	2.85 (0.51)
	Ratio (%)	109.9(30)	100	27.1(12)	28.6(8.1)
48	Displacement (mm)	0.210 (0.021)	0.120 (0.020)	0.210 (0.018)	0.050 (0.005)
	Work (Nmm)	5.04 (1.01)	3.60 (0.85)	5.67 (0.81)	1.50 (0.26)
	Ratio (%)	140(61)	100	158(60)	41.7(17)

* Relative error in this case is serious compared to the values of the other screws.

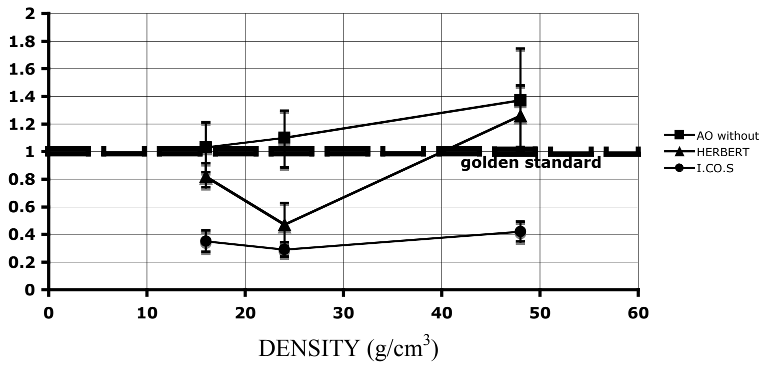


Fig. 6. Work of screw systems to achieve the compression of 60 N as compared to that of the AO screw with washer

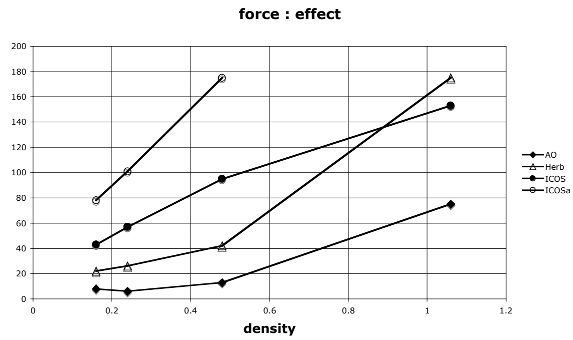


Fig. 7a

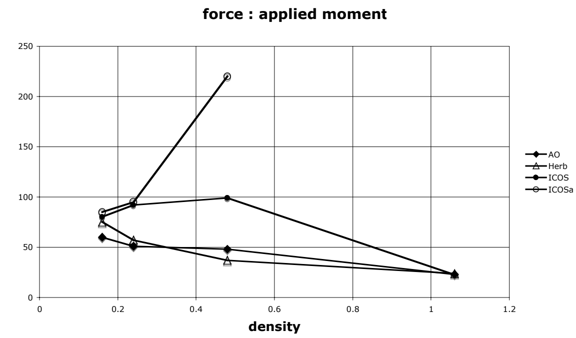


Fig. 7b

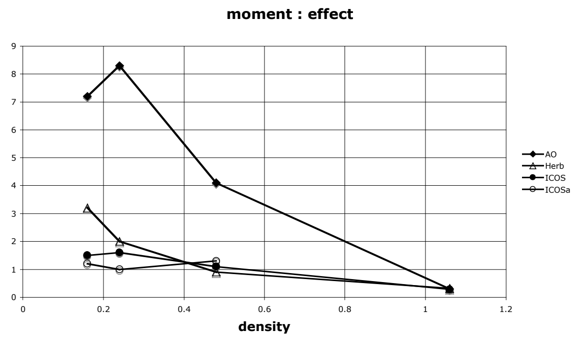


Fig. 7c

Fig. 7. Results of the Sawbone experiments as a function of the density (retrofitted with the results from the Plexi-Araldit-Plexi Models with a density of approximately 1.06 g/cm^3)

Figure 6 shows the ratio λ, according to equation (2), at the pressure level of 60 N.

Figure 7a shows the relationship between the compression force achieved and the effect applied as a function of the Sawbone density. Figure 7b shows the ratio of the achieved force to the applied screw moment. Figure 7c shows the ratio of the applied moment to the effect. It is shown that I.CO.S is the predominant system in all ratios in all density ranges.

4. Conclusion

Today internal fixation with screws is the most frequently used technique to get a certain bony union in the hindfoot [6], [9], [11], [13], [15], [16], [18], which is crucial to achieve primary stability in arthrodesis. The maximum value of compressive force is mainly limited by the bone quality [1], [7], [8]. But also the design of the screw system used is important for good compression. The overloading of the screw–bone interface may destroy the threads in the bone and results in a loss of the compressive force. This failure is the main problem during surgery. After operation the rearrangement of the bone structure known as *bone remodelling* [23] occurs. Rheological behaviour of bone as well as biomechanical processes initiated by operation are responsible for this process. Bone remodelling may lead to a loss of compression force before bone reaches a sufficient secondary stability. COWIN and HEGEDUS [4] as well as MARTIN [14] showed that the rate M of bone remodelling is governed by the following equation:

$$M = G: (\varepsilon - \varepsilon_0). \quad (3)$$

The tensor G (the coefficient of proportionality to $\varepsilon - \varepsilon_0$) has to be adopted via experimental results, with $\varepsilon - \varepsilon_0$ as the deviation of an actual primary strain distribution from the reference one. In the present paper, we compared the mounting strain distribution ε_0 of the golden standard with those of the other screw systems, with the result that the rate of bone remodelling reaches a minimum by using I.CO.S which is of special importance at poor bone quality.

The question arises with which screw system a sufficient primary stability is reached with a minimum of the work applied during insertion of the screw. The work W necessary to reach the chosen value of pressure (60 N) is due to the accompanying deformation of the load transfer zones quite different for the screw systems. Also in this context the I.CO.S is superior to the other screw systems. With respect to the three bone densities tested it was not necessary to make use of an additional possibility of loading offered by I.CO.S to increase the pressure applied. However, for extremely poor bone quality it may be necessary to use this additional device of a specially designed head, which could be inserted along the proximal end of I.CO.S.

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