Arch Orthop Trauma Surg (1994) 113:271–275

ORIGINAL ARTICLE

R. A. Jaskulka · E. Egkher · B. Wielke

Comparison of the mechanical performance of three types of unilateral, dynamizable external fixators

An experimental study

Received: 6 December 1993

Abstract Due to the increasing popularity of unilateral dynamizable external fixators for treating tibial shaft fractures, many new devices are being introduced onto the market. Especially in such half-frame fixators, the choice of any particular device depends above all on the stability of its construction. This study compares the biomechanical stability of three systems tested in axial compression, torsion, and both anterior-posterior and medial-lateral bending. In terms of the nondynamized phase, the AO/ASIF tubular fixator (as a one-plane, double-tube, unilateral frame) and the Martin Mono-Dynafix are, in general, less stable than the Orthofix fixator. After dynamization, the AO/ASIF system becomes particularly weak and offers low resistance especially to torque and any force that is perpendicular to the plane of assembly. The other two tested devices evinced much more stability; the Orthofix fixator seems superior to the Dynafix due to the different diameter of its screws.

Introduction

The objective of treatment with an external fixator is to combine rapid fracture healing – often under unfavorable soft tissue conditions – with optimal wearing comfort. For this reason the minimalist variant of a unilaterally inserted monofixator has lately prevailed. In addition to improved wearing comfort, this has the unique advantage that, after differing initial periods of rigid fixation, adjustment of axial fixation rigidity is possible at any time during the course of fracture healing.

R. A. Jaskulka (✉)
Medical Center East, Department for Trauma Surgery, Langobardenstrasse 122, A-1220 Vienna, Austria

E. Egkher
Clinic for Trauma Surgery, University of Vienna, Austria

B. Wielke
Institute for Solid State Physic, University of Vienna, Austria

Especially in the choice of a particular fixation device for routine use, certain problems may arise, as nowadays several versions are available. The variations go from the simple AO/ASIF tubular fixator to complicated instruments specially developed for dynamizable use.

Before making a decision, it is above all the stability of the constructions that must be taken into account. Although there is proof that absolute rigidity of the external fixator is less important for the healing of the bone fracture than was formerly supposed [7, 8], it should be remembered that a unilateral frame per se can offer less resistance to strain exerted on it than three-dimensional constructions or a closed frame. For even healing of a fracture and to prevent premature loosening of the screws, therefore the stability of these single-frame fixators therefore is still of outstanding importance.

The objective of this study was to compare the most common unilateral fixators as to stability behavior, and from that to develop recommendations for clinical use. Due to varying mechanical requirements in the different phases of treatment, the examinations were carried out both with locked and dynamized devices.

Methods

Three of the most frequently used dynamizable fixation devices were tested:

1. The AO/ASIF tubular external fixator in the form of a one-plane, double-tube, unilateral frame [11]
2. The Martin Mono-Dynafix (Gebrüder Martin, Tuttlingen, Germany)
3. The Orthofix Fixator (Orthofix, S.R.L., Verona, Italy).

Experimental design

The frames were mounted on round beechwood sticks with a diameter of 34 mm, representing the bones. The proximal end of one stick was clamped closely to the screw, rendering movements at this end negligible. The basic arrangement of the half-frames was kept constant in all models (Fig. 1): the distance between the Schanz screws $a = a'$ was 30 mm; the distance between the inner screws f was 180 mm; the distance between the inner screws and

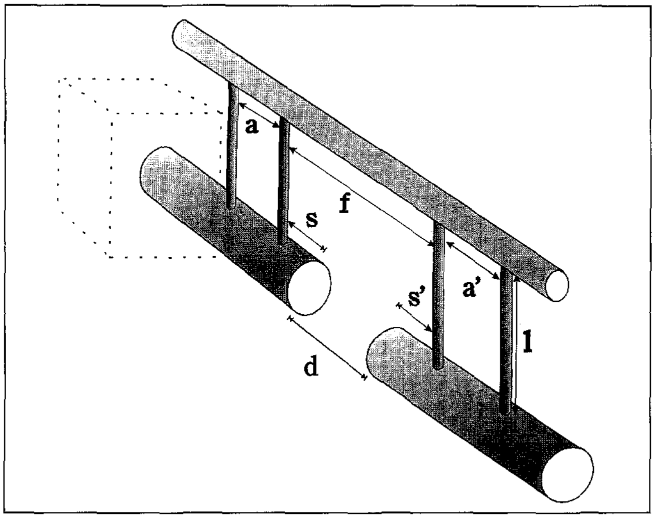

Fig. 1 Basic arrangement of the external fixators

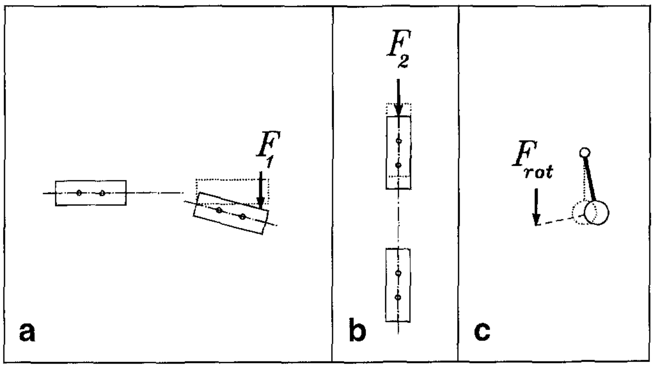

Fig. 2 a–c Physically possible stresses: **a** bending F_1, **b** compression F_2, **c** torsion F_{rot}

the fracture line $s = s'$ was 30 mm, and the free length of the screws *l* was 60 mm.

Since all kinds of forces that occur in practice are basically mixed forms of three physically different stresses, three cases of loading were taken into consideration (Fig. 2):

(a) Bending (F_1) – analogous to the impact of the weight of the extremity itself in a horizontal position
(b) Compression (F_2) – analogous to the weight in the horizontal direction of the extremity (standing).
(c) Torsion through the effect of the torque (F_{rot}) – analogous to the effect of twisting the extremity about its longitudinal axis.

Because of the asymmetrical construction of the half frames, the tilting of the fracture gap, the displacement x (measured at measuring point 1, MP1), and the torsion (measured at measuring point 2, MP2) were basically measured in two arrangements (Figs. 3, 4):

1. The loading impact parallel to the arrangement of the frames
2. The loading impact perpendicular to the frames.

Measurements

In all cases the impact of F_1 and F_{rot} were kept constant at a distance of 40 mm from the most distal screw. The length of the lever arm for F_{rot} was 100 mm.

In all variations mentioned, 2-kg weights were used, causing a load of 2 kg × 9.81 m/s² = 19.6 N.

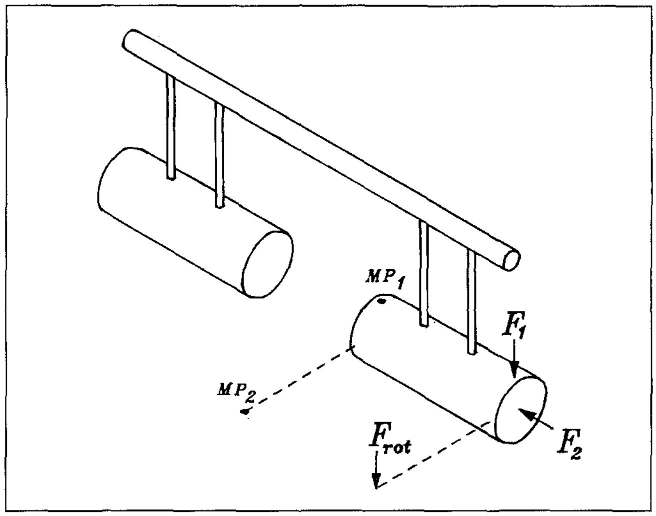

Fig. 3 Half frame with measuring points, showing loading impact parallel to the pins. For definition of measuring points, see text

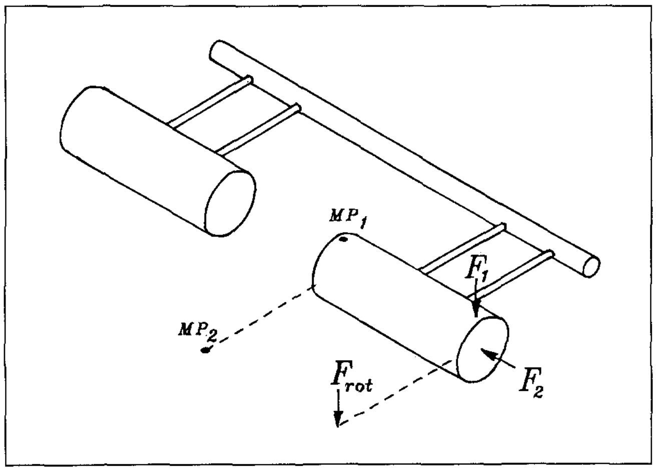

Fig. 4 Half frame showing loading impact vertical to the pins

The torque F_{rot} on the bone axis caused by a weight of 2 kg with a lever length of 100 mm is 2 kg × 9.81 m/s² × 0.1 m = 1.96 Nm. Taking the axis of the frame as the basis of rotation, the lever is 0.19 m long and the torque is 3.7 Nm on to this axis.

The vertical displacement x within the fracture gap was taken as a measure of the structure's flexibility/stiffness. It was therefore measured parallel to the load impact F_1 at the most distal bone, immediately in front of the inner Steinmann pin (measuring point MP1). The twisting was taken through lowering of a 100 mm long lever (measuring point MP2), measured from the axis (see Fig. 4).

Since F_{rot} causes displacement within the fracture gap mainly when the frames are arranged horizontally, it was measured only under a loading impact, perpendicular to the screws. Any other direction of force shows much less amount of displacement and bending.

All measurements were carried out five times and an average value was calculated. The measurement resolution was 0.01 mm; the reproducibility lay within the range of ± 0.1 mm.

The axial load was applied cyclically through an electronically controlled, static dynamic deforming device (Instron Co.) with a high resolution (< 0.001 mm) and the elastic hysteresis was measured. The slope of the stress-strain curve indicated the elasticity/rigidity. Loads of up to about 500 N (equivalent to a weight of 50 kg) were applied.

Results

Fixators locked

Loading type: F_1; *measuring point: MP1* (Tables 1, 2)

With the clamp vertical (loading impact parallel to the screws), the displacement in all three models was 0.3 mm (corresponding to a compliance of 0.015 mm/N). In a horizontal arrangement (load perpendicular to the screws), compression was: AO fixator 6.3 mm (0.32 mm/N), Dynafix 6.0 mm (0.31 mm/N), and Orthofix 2.7 mm (0.14 mm/N).

Loading type: F_{rot}; *measuring point: MP1*

The displacement of the most distal fragments with the torque impact and with a horizontal clamp arrangement was 7.6 mm (0.39 mm/N) with the AO fixator and 7.7 mm (0.39 mm/N) with the Dynafix. Significantly ($P < 0.001$) lower values were obtained with the Orthofix at 3.4 mm (0.17 mm/N).

Loading type: F_{rot}; *measuring point: MP2*

The pure torsion on measuring point MP2 with a vertical arrangement was as follows: AO fixator 1.72°/2 kg (0.88°/Nm), with Dynafix 1.95°/2 kg (0.99°/Nm) and with Orthofix 0.86°/2 kg (0.44°/Nm). In a vertical arrangement the torsion was: AO fixator 8.95°/2 kg (4.38°/Nm), Dynafix 8.02°/2 kg (4.09°/Nm) and with Orthofix 3.67°/2 kg (1.87°/Nm).

Loading type: F_2 (Table 3)

The shortening with axial loading impact was: AO fixator 0.014 mm/N, Dynafix 0.019 mm/N, and Orthofix 0.012 mm/N.

Fixators dynamized

Loading type: F_1; *measuring point: MP1* (Tables 4, 5)

Compression with the AO fixator was 0.5 mm (0.025 mm/N), with the Dynafix 0.4 mm (0.020 mm/N), and with the Orthofix unchanged, and in the nondynamized state it was 0.3 mm (0.015 mm/N)

Loading type: F_{rot}; *measuring point: MP1*

Measuring of fracture compression under the torque impact F_{rot} was not practicable with the AO/ASIF fixator because of its complete instability. The value with the Dynafix was 8.0 mm (0.14 mm/N) and with the Orthofix 3.8 mm (0.19 mm/N).

Table 1 Loading impact parallel to the fixator frames when locked

	AO/ASIF	Dynafix	Orthofix
Measuring point MP1 Loading F_1	0.3 mm 0.015 mm/N	0.3 mm 0.015 mm/N	0.3 mm 0.015 mm/N
Measuring point MP2 Loading F_{rot}	3.0 mm 0.88°/Nm	3.7 mm 0.99°/Nm	1.8 mm 0.44°/Nm

Table 2 Loading impact perpendicular to the frames when locked

	AO/ASIF	Dynafix	Orthofix
MP1 Loading F_1	6.3 mm 0.32 mm/N	6.0 mm 0.31 mm/N	2.7 mm 0.14 mm/N
MP2 Loading F_1	8.8 mm 0.45 mm/N	9.5 mm 0.48 mm/N	3.2 mm 0.16 mm/N
MP1 Loading F_{rot}	7.6 mm 0.39 mm/N	7.7 mm 0.39 mm/N	3.4 mm 0.17 mm/N
MP2 Loading F_{rot}	15 mm 4.38°/Nm	14 mm 4.09°/Nm	6.4 mm 1.87°/Nm

Table 3 Loading impact axial with the frames locked

	AO/ASIF	Dynafix	Orthofix
Loading F_2	0.014 mm/N	0.019 mm/N	0.012 mm/N

Table 4 Loading impact parallel to the frames when dynamized

	AO/ASIF	Dynafix	Orthofix
MP1 Loading F_1	0.5 mm 0.025 mm/N	0.4 mm 0.020 mm/N	0.3 mm 0.015 mm/N
MP2 Loading F_{rot}	19.5 mm 5.70°/Nm	3.5 mm 1.0°/Nm	1.7 mm 1.50°/Nm

Table 5 Loading impact perpendicular to the frames when dynamized

	AO/ASIF	Dynafix	Orthofix
MP1 Loading F_1	13 mm 0.66 mm/N	6.0 mm 0.31 mm/N	2.8 mm 0.14 mm/N
MP2 Loading F_1	23.0 mm 1.17 mm/N	9.5 mm 0.48 mm/N	3.7 mm 0.19 mm/N
MP1 Loading F_{rot}	not measureable	8.0 mm 0.41 mm/N	3.8 mm 0.19 mm/N
MP2 Loading F_{rot}	54 mm/N ≈ 16°/Nm	15 mm/N 4.40°/Nm	7.2 mm/N 2.12°/Nm

Table 6 Loading impact axial with the frames dynamized

	AO/ASIF	Dynafix	Orthofix
Loading F_2	Not measureable	0.017 mm/N	0.013 mm/N

Loading type: F_{rot}; *measuring point: MP2*

The torque at the measuring point was 11°/2 kg (5.7°/Nm) with the AO arrangement, 2.0°/2 kg (1.0°/Nm) with the Dynafix, and 0.97°/2 kg (0.50°/Nm) with the Orthofix.

Loading type: F_2 (Table 6)

Testing of the axial loading impact in a dynamized state was carried out for the sake of completeness. All three fixators gave the expected results: the AO clamp slid together to the point of fracture compression; the Dynafix and Orthofix, after using up the dynamization gap, achieved similar values to those in the nondynamized state: Dynafix 0.017 mm/N, Orthofix 0.013 mm/N.

Discussion

Our results show clear differences in the individual fixators' stability behavior, which are demonstrable in both the locked and, especially, the dynamized phase.

In the first phase of "treatment" (not dynamized), the differences in measured values between the AO/ASIF external fixator and Dynafix were minimal. The Orthofix clearly showed more stability in this case, particularly in regard to a load perpendicular to the screws, and to torsion as well. In both cases the measured difference was more than 50%, while a load with an impact parallel to the frame or one that worked axially, was neutralized to an almost equal degree by all three devices.

In the second phase (dynamized) the differences in stability became far more obvious. The AO/ASIF fixator became almost useless, because of its low resistance to torque or to a force appearing normally to the mounting plane. The instability was so extreme that some measurements could not be carried out at all due to the extent of displacement. This fixator offers resistance comparable to that of other models only to loads acting parallel to the assembly plane. As to the Orthofix, in this phase of treatment we confirmed the results of Aro et al. [1]: axial dynamization in our tests had little influence on the rigidity of the Orthofix fixator under rotation and under flexion. The Dynafix hardly changed its stability behavior compared to the nondynamized phase. Orthofix proved superior to Dynafix in all measurements in this phase as well, but the differences wer only minor, and of doubtful clinical relevance.

Free transmission of load between the fragments, which was decisive for the second phase of treatment, was present in all three models. In the case of the AO/ASIF fixator we found the ideal case of dynamization with complete sintering together of the simulated fracture gap and a free flow of force between the fragments under axial loading. The other two instruments also showed free axial mobility up to the limit due to construction, after which the resistance corresponded to that in a nondynamized state.

The causes of the instabilities to be found in all external fixators were different in the individual models. In each model a weak point due to construction could be identified. In the case of the AO/ASIF fixator it was clearly the connection of the adjustable clamp with the tubes that was responsible for the great loss in stability after dynamization. When the jaws are loosened there is no control over the acting torque. This explains the massive loss of stability during this phase.

In the case of the Orthofix, the sliding mechanism between the proximal and the distal bar proved to be the weak point. After dynamization, an instability – albeit slight – occurred in the form of a rotation, something also described by Ralston et al. [12]. The values measured by us with this model fell off accordingly, after loosening of the fastening screw, especially under the effect of the torque. The Dynafix has superior guidance at this connecting point, which was reflected in almost unchanging measured values in both "treatment" phases. The fact that the measured values of Orthofix were nonetheless better than those of Dynafix is explained by the diameter of the Schanz screws, which are larger by 1 mm, making them stiffer by a factor of 73%. This can be deducted from the results of Egkher [5] and Chao and Hein [2]. Egkher stated that the physical rigidity of a screw is proportional of the fourth power of its diameter, a fact that clearly proves the importance of this parameter.

Concerning the clinical use of these fixators, the measured differences in stability should entail clear effects on the fracture healing, as well as on the incidence of screw loosening and screw tract infection. This is demonstrated in the results of numerous clinical and experimental studies [3, 6, 9, 15], which have shown that a lack of stability is of special importance for the quality of the healing of the bone fracture and for the occurring of pin-bone interface problems. There are, however, various opinions about the extent of stability necessary for optimal fracture healing; micromovements in the fracture gap are regarded by some authors as advantageous for ideal fracture healing [7, 8]. On the other hand, Lewallen et al. [10] and Chao et al. [3] achieved optimal healing of bone fractures only under absolutely rigid fixation, especially in the early phase of treatment. However, according to our results, all three systems tested seem suitable for use in the nondynamized form. Without loosening of the gliding mechanism, no clinically relevant differences in stability are to be expected between the models.

After dynamization, a free transfer of load should be possible axially, whereas movements on other planes are undesirable in this phase [14, 15]. This requirement cannot be met by the AO/ASIF fixator, as can be seen from our results. Therefore, with fractures that allow early dynamization, this fixator appears of dubious value in clinical use. The other two models tested seem almost of equal quality and are definitely more suitable for these fractures.

Problems at the pin-bone interface are among the most frequent complications during treatment with the external fixator and are also related to the stability of the construction. Increased pin loosening is due to increased bone resorption around the screws, caused by decreased stability [6, 13, 15]. There are indications, however, that the dynamization maneuver reduces the risk of pin-bone interface failures in the second phase of treatment, due to decreased load on the screw-bone line [1]. This goes for the ideal case, when the axial dynamization does not alter fixation rigidity under either torsion or bending, while the compressive load is able to be transmitted freely through the bone ends at the fracture site. Taking into consideration our results, we can therefore deduce that with the AO/ASIF fixator, loosening of screws should occur more frequently than with other models.

To sum up, we found that the AO/ASIF fixator should only be recommended for clinical use as a monofixator in cases where dynamization is not desirable. It offers sufficient stability only in its nondynamized form. With this fixator, cyclical axial loading in the form of dynamization, with all its advantages for the healing of the fracture [7], can only be achieved at the cost of massive loss of stability in the other planes too. The two other models tested are excellent instruments and are almost equal in quality in both phases of treatment, the Orthofix fixator being slightly superior. This difference is not caused by the construction of the frames but by the different diameters of the pins used by the two systems, for the screw diameter is the main determinant of stability.

References

1. Aro HT, Kelly PJ, Lewallen DG, Chao EY (1990) The effects of physiologic dynamic compression on bone healing under external fixation. Clin Orthop Rel Res 256 : 260–273
2. Chao EY, Hein TJ (1988) Mechanical performance of the standard Orthofix external fixator. Orthopedics 11 : 1057–1069
3. Chao EYS, Aro HT, Lewallen DG, Kell PJ (1989) The effect of rigidity on fracture healing in external fixation. Clin Orthop Rel Res 241 : 24–35
4. De Bastiani G, Aldegheri R, Renzi Brivio L (1984) The treatment of fractures with a dynamic axial fixator. J Bone Joint Surg [Br] 66-B : 538–545
5. Egkher E (1985) Theoretische und experimentelle Studien zur Technologie der externen Knochenbruchstabilisierung. Wien Klin Wochenschr 97 [Suppl 158]
6. Evans M, Spencer M, Wang Q, White SH, Cunningham JL (1990) Design and testing of external fixator bone screws. J Biomed Eng 12 : 457–462
7. Goodship AE, Kenwright J (1990) The influence of induced micromovement upon the healing of experimental tibial fractures. J Bone Joint Surg [Br] 67-B : 650
8. Heitemeyer U, Claes L, Hierholzer G, Korber (1990) Significance of postoperative stability for bony reparation of comminuted fractures. An experimental study. Arch Orthop Trauma Surg 109 : 144–149
9. Huiskes R, Chao EY (1986) Guidelines for external fixation frame rigidity and stresses. J Orthop Res 4 : 68–75
10. Lewallen DG, Chao EY, Kasman RA, Kelly PJ (1984) Comparison of the effects of compression plates and external fixators on early bone healing. J Bone Joint Surg [Am] 66-A : 1084–1091
11. Müller ME, Allgöwer M, Schneider R, Willenegger H (1991) Manual of internal fixation. Techniques recommended by the AO-ASIF group, 3rd edn. Springer, Berlin Heidelberg New York
12. Ralston JL, Brown TD, Nepola JV, Williams DR, Marsh JL (1990) Mechanical analysis of the factors affecting dynamization of the Orthofix dynamic axial fixator. J Orthop Trauma 4 : 449–457
13. Schatzker J, Horne JG, Sumner-Smith G (1977) The reaction of cortical bone to compression by screw threads. Clin Orthop Rel Res 111 : 248–252
14. Williams EA, Rand JA, An KN, Chao EY, Kelly PJ (1987) The early healing of tibial osteotomies stabilized by one-plane or two-plane external fixation. J Bone Joint Surg [Am] 69-A : 355–365
15. Wu JJ, Shyr HS, Chao EY, Kelly PJ (1984) Comparison of osteotomy healing under external fixation devices with different stiffness characteristics. J Bone Joint Surg [Am] 66-A : 1258–1264