CHAPTER 21

RELATIVE STRENGTHS OF INTERNAL FIXATION IN OSTEOTOMIES AND ARTHRODESIS OF THE FIRST METATARSAL

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The purpose of this project is to quantitate and compare the relative response of internal fixation techniques to specific loads in osteotomies and arthrodesing techniques of the first metatarsal. Common techniques used for fixation in distal metatarsal osteotomies, base wedge osteotomies, and first metatarsal-cuneiform arthrodesis are evaluated. Mechanical testing was performed using a loading-rate controlled tensiometer and solid foam core foot models. Various modes of internal fixation were examined to determine the relative fixated stability of each configuration for several osteotomies. The relative stability imparted by each combination will be reported.

For the podiatric physician, evaluation and surgical treatment of first metatarsal deformities are common practice. Over the past several decades, the development of numerous osteotomies addressing these deformities has only been matched by the variety of fixation techniques utilized to stabilize them. The authors have taken a close look at these various techniques and evaluated their strengths and weaknesses. Which osteotomy is the most stable? Which fixation is superior? What factors does the surgeon introduce to the overall success of these procedures? These questions can best be answered after performing a quantitative study of the common techniques, and critically analyzing the results.

The purpose of this project is to evaluate the mechanical design of osteotomies and their fixation, therefore selection of an appropriate model is necessary. This model should minimize any additional variables. For this purpose, solid foam core foot models were utilized. There have been several other first metatarsal studies, yet these have all utilized cadaveric specimens. The tremendous variability between cadaveric specimens obscures the purely geometric factors that are to be observed within this study.

MATERIALS AND METHODS

This study was designed to compare the relative load-bearing ability of several osteotomies with different types of commonly used fixation techniques. In order to avoid confounding the geometric issues of osteotomy and fixation configurations with variations in biologic materials, a highly uniform substrate was needed. Therefore, a solid foam core polyurethane, first metatarsal model with attached medial cuneiform was used for the study. The following configurations of osteotomy and fixation were examined:

Distal Metaphyseal Osteotomies:

Austin	(0.062 K-wire) (4.0 mm screw)
Kalish-Austin	(0.062 Threaded K-wires) (2 - 2.7 mm screws)
Reverse Kalish-Austin	(2 - 2.7 mm screws)
Scarf	(2 - 2.7 mm screws)
Inverted Scarf	(2 - 2.7 mm screws)
Mau	(2 - 2.7 mm screws)

Base Wedge Osteotomies

Transverse	(24 Ga. Horizontal I.O.L.) (24 Ga. Perpendicular I.O.L.)
Oblique	(1 - 4.0 mm screw) (2 - 4.0 mm screw) (2 - 4.0 mm screw with plantar "tension band")

Metatarsal-Cuneiform Arthrodesis
- Crossed 0.062 K-wire
- Crossed 4.0 mm Screws
- 5-hole Medial plate
- 5-hole Medial plate with 4.0mm Interfragmental Screw

Mechanical testing was performed on a tensiometer (Instron, Canton, MA, model 4201, screw type). Load was applied at a constant rate in a purely sagittal plane direction at the level of the distal metatarsal. The load was applied by passing a flexible surgical wire around the model through an indented region representing the location of the first metatarsal cuneiform joint (Fig. 1). The cross head of the tensiometer travelled vertically at a rate of 500 mm/min. It has been shown previously that this rate is within the physiological loading range in a postoperative patient.[1] A voltmeter was used to zero out the load on the specimen prior to advancement of the crosshead. Data was sampled for 5 seconds at a rate of 200 Hz utilizing a personal computer with data acquisition software.

All specimen preparation, including the cutting of osteotomies and insertion of internal fixation devices was performed by the same surgeon (TC) in order to diminish the variability between specimens. All fixation devices were manufactured by Synthes, Inc. (Paoli, PA), and were inserted using standard AO technique. Once

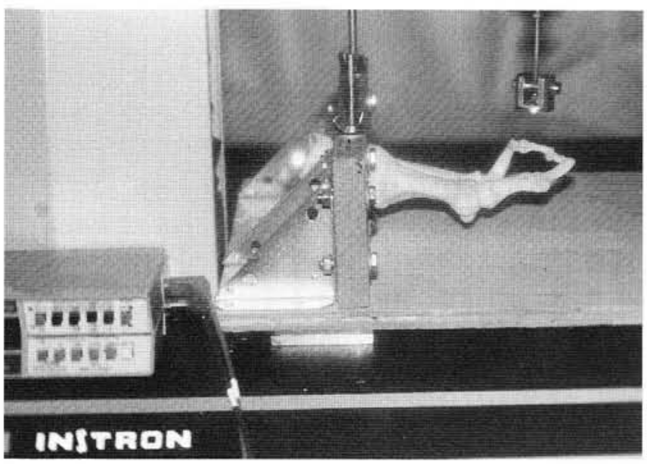

Figure 1. Instron Tensiometer testing device utilized for the application of constant load to the first metatarsal . A Voltmeter was utilized to zero-out the load prior to applying stresses in each specimen.

prepared, the specimens were rigidly attached to the testing jig. The jig, which was constructed of hardwood and steel, held each specimen in a completely immobile position, proximal to the metatarsal-cuneiform joint (Fig. 2).

Initial tests were performed on intact bone models (n=4) to gather baseline data, and to confirm the reproducibility of the testing system. Next, various combinations of arthrodeses and fixation were tested (n=2 for each configuration). All experiments were recorded on video tape for careful analysis of the mechanism of osteotomy/fixation failure. For each experiment performed, plots of load (newtons) versus displacement (millimeters), and load (newtons) versus time (seconds) were generated. Plots of the data were examined to determine the failure point. By convention, the failure point was determined to be at the transition point between elastic and plastic deformation, as indicated by a decreasing or zero slope on the load versus time plot (Fig. 3). Statistical analysis of all data was performed with Statview 512 software on a Macintosh computer.

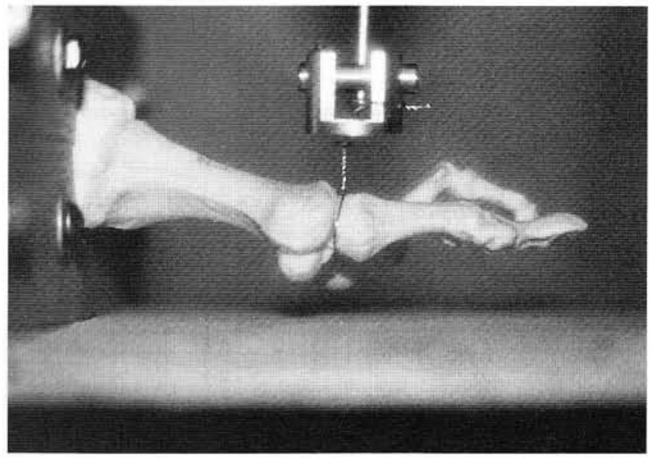

Figure 2. A wooden jig was constructed to securely hold the polyurethane models completely immobile at the proximal base. The jig was designed to approximate the metatarsal declination angle at 15-20 degrees.

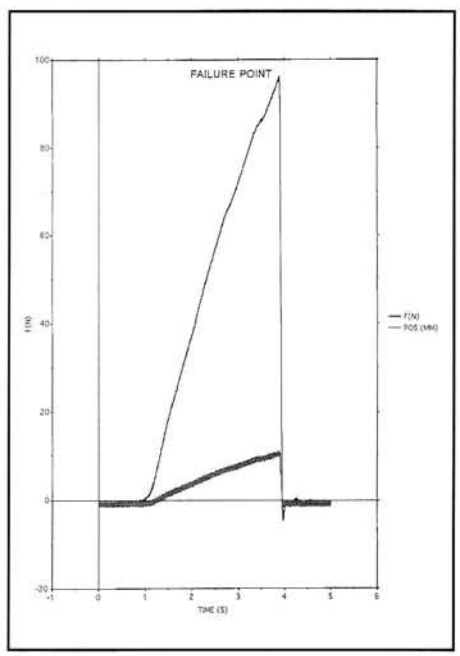

Figure 3. Example of the force vs. displacement plots for all specimens. Data was sampled for 5 seconds at a rate of 200 hertz. (1000 samples/5 seconds)

RESULTS

The average load required for failure of each osteotomy/fixation design is demonstrated in Figures 4A-4C with two-dimensional plots of Load (newtons) versus Time (seconds). Based on this series of experiments, it was determined that the most stable distal first metatarsal procedures were the Austin osteotomy with either 4.0 mm screw or 0.062" K-wire fixation, and the Reverse Kalish-Austin osteotomy with two 2.7 mm screws for fixation. The proximal base wedge osteotomy was found to be most stable when tension band fixation was used with the tension band traversing the plantar surface in the sagittal plane. Similarly, first metatarsal-medial cuneiform arthrodesis was found to be the most stable when the fixation device prevented plantar gapping, and methods which utilize tension bands placed below the neutral axis, combined with axial compression were most stable.

DISCUSSION

Distal Osteotomies

The strongest osteotomy/fixation combination in the tests were the standard Austin osteotomy with either 4.0 mm screw or 0.062" K-wire fixation, and the Reverse Kalish-Austin with two 2.7 mm screws. Careful examination of these models reveal a common mechanical design. The superior stability of these specific designs is probably due to transfixation of the plantar arm of the osteotomy. Compressive forces in this plane resulting from fixation and/or loading of the osteotomy perpendicular to this plane produces increased friction between opposing surfaces. This results in minimal shear or sliding along this interface. As a result, weight-bearing forces at the metatarsal head are transferred along the intact proximal plantar aspect of the bone. Additional stability provided by fixation across this arm enhances the inherent stability of the osteotomy. In the K-wire model, initial stability is seen as dorsal forces are loaded across the plantar arm, yet the lack of compressive fixation eventually causes the slippage which leads to failure. With physiologic specimens, the K-wire pulls through the medullary canal of the apex and ultimately results in fracture through the dorsal cortex.

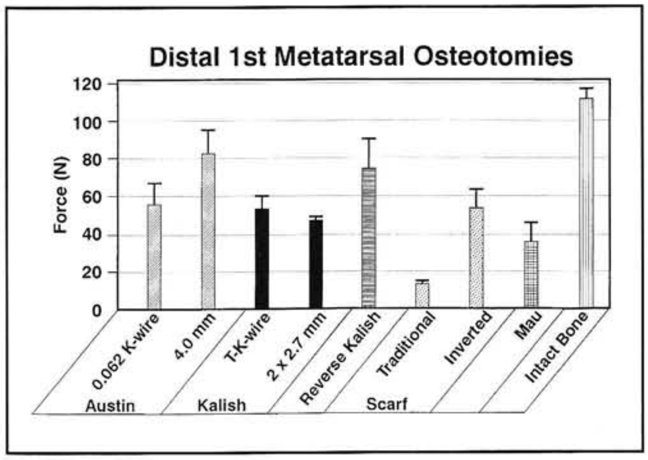

Figure 4A. Graph depicting the force applied at failure for various designs of distal metatarsal osteotomies

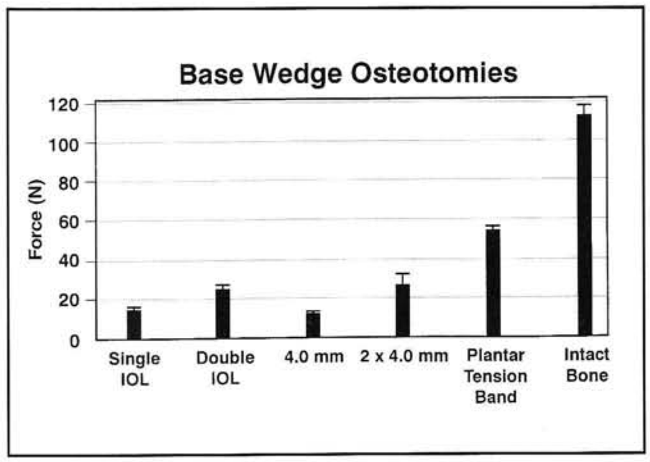

Figure 4B. Graph depicting the force applied at failure for base wedge osteotomies

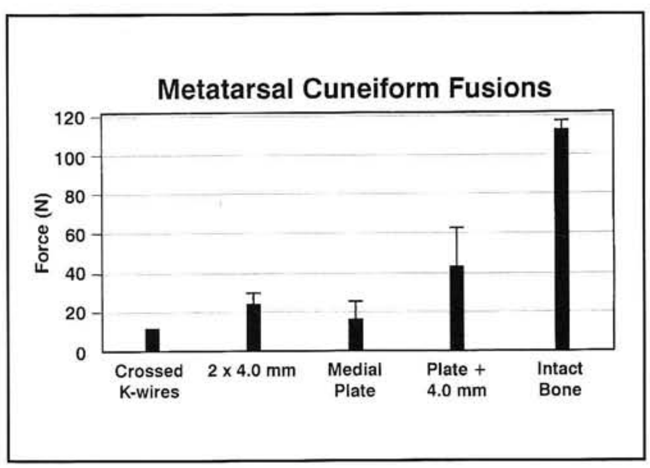

Figure 4C. Graph depicting the force applied at failure for first metatarsal-cuneiform arthrodesis.

When failure occurs with the 4.0 mm screw, fracture takes place in the diaphyseal (midshaft) region of the bone model, rather than at the osteotomy site. There is no observable failure in either the fixation or osteotomy (Figs. 5A, 5B). This results from the superior compression at the osteotomy site. With the Reverse Kalish design, again the plantar osteotomy is securely fixed, although this time with two 2.7 mm screws. Ultimate failure and fracture is again seen in the midshaft region of the metatarsal.

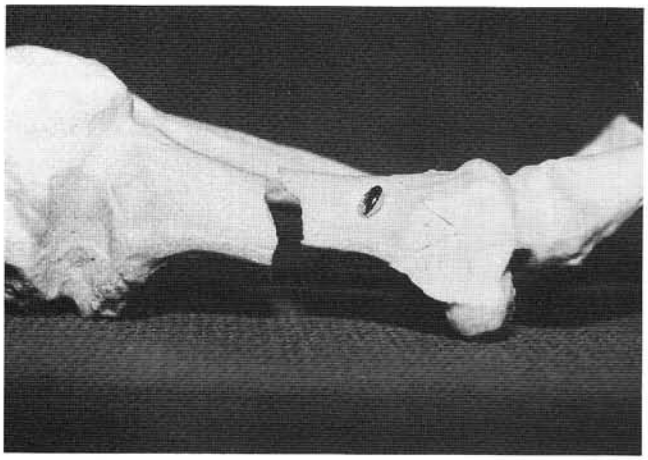

Figure 5A. Representative model of failure seen with a 4.0mm cancellous screw utilized for standard Austin fixation. Note no observable failure of the fixation or the osteotomy site; the model failed similar to an intact specimen.

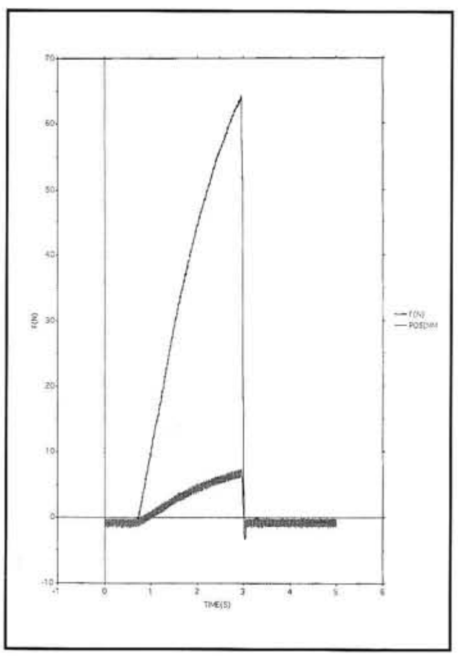

Figure 5B. The force plot for the standard Austin fixation.

Testing of the Kalish osteotomies clearly illustrates the role of stress risers in overall stability. When weight-bearing stresses are applied, ultimate failure of these osteotomies occurs by an interesting and reproducible process. Initial gapping is noticed across the unfixated plantar osteotomy. With increased weight-bearing loads, rotation of the capital fragment will occur around the distal apex of the osteotomy. Eventually, forces will be transferred to, and dissipated through, the first stress riser it encounters; a stress riser created by the dorsal fixation. In each test of this osteotomy and fixation configuration, failure is seen through a fracture at the most distal fixation site. This defect is created by either a 1.6 mm K-wire, or 2.7 mm diameter screw (Figs. 6A, 6B). The slightly higher stability (53N versus 47N) seen with 0.062" K-wires versus screws could be attributed to the smaller defect created in the dorsal wing of the osteotomy with the K-wire.

The ability of the Scarf osteotomies (traditional versus inverted) to bear loads was shown to vary tremendously from a mechanical perspective. Although the original design of the Scarf osteotomy provides a dorsal shelf to enhance weight-bearing stability, this can be misleading. In actuality, the traditional Scarf is only as strong as the proximal plantar osteotomy is shallow. One can imagine creating a plantar cut which rises dorsally into the base of the first metatarsal at a similar location to this proximal cut. The deeper the cut is carried into the metatarsal, the greater the stress riser and the thinner the intact dorsal section of the metatarsal. The proximal plantar cut in the metatarsal is a notch, and the deeper the notch, the easier it is to break (Figs. 7A, 7B). By simply inverting the Scarf osteotomy, the inherent mechanical stability dramatically increases. Inversion of the traditional design now produces a proximal dorsal cut with a distal plantar cut.

The strength of the Inverted Scarf lies in understanding the tension side of the bone. In theory, as dorsally directed loads are applied, these forces are transmitted across the distal osteotomy site, and then ideally transferred directly to the proximal metatarsal segment, being absorbed primarily along the intact plantar cortex. An intact plantar cortex is the cornerstone of this sound mechanical design. In actuality, the inherent stability now mimics the design of the Kalish-Austin discussed previously, and ultimate failure in these

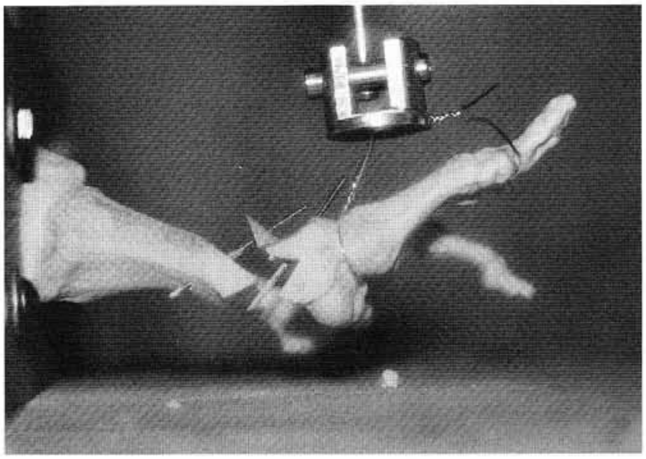

Figure 6A. Example of model failure for a Kalish-Austin with 2 0.062" threaded K-wires. There is observable rotation of the capital fragment around the distal apex with ultimate fracture occurring at the first stress riser encountered; the distal fixation device.

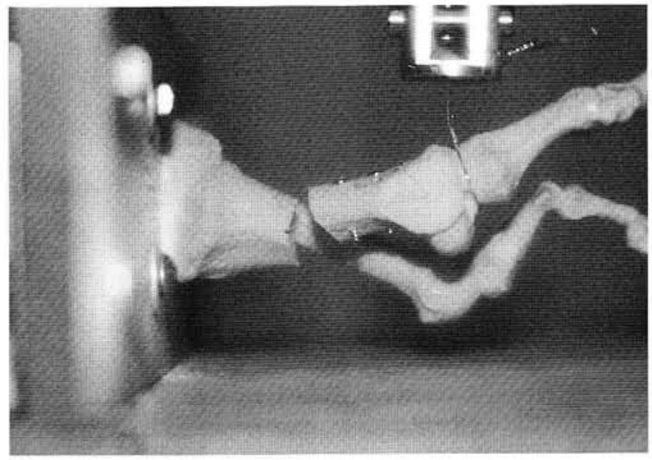

Figure 7A. Reproducible failure of a standard Scarf osteotomy with model failure progressing at the plantar arm through the proximal apex and ultimately progressing through the dorsal cortex. Again forces are centered around the stress risers of the osteotomy.

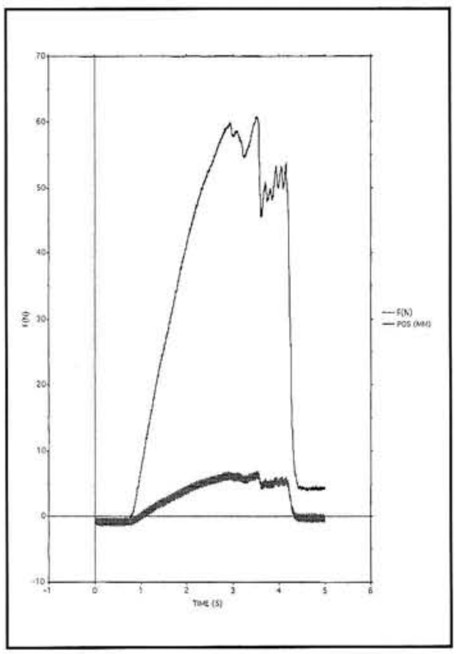

Figure 6B. The corresponding force plot for a Kalish-Austin with 2 threaded K-wires.

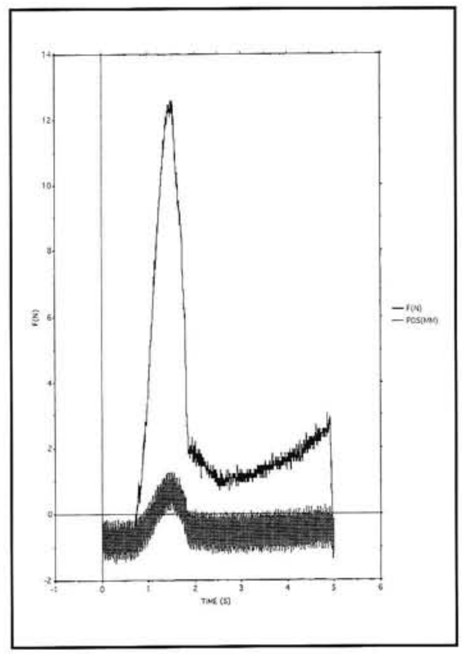

Figure 7B. The corresponding force plot of a standard Scarf osteotomy.

two models are parallel. The relative strength of the traditional Scarf is 12-15N, while the inverted design fails at 55-60N, an impressive three to four-fold increase in mechanical stability.

The Mau osteotomy with screw fixation possesses intermediate stability. At first glance, the design resembles a long Reverse-Kalish cut and appears mechanically sound. Execution of the osteotomy however, requires a very proximal exit point through the plantar cortex. The exit point is more proximal than generally found with the Reverse-Kalish, and there is a much greater moment in this area due to the greater lever arm created. When weight-bearing stresses are applied, all forces are immediately directed to this proximal exit point. Although stable screw fixation attempts to hold the two pieces securely, and prevent any shearing at the interface, gapping at this site is immediately evident. Continued stress will cause continued gapping, and the osteotomy will open distally until failure is seen with fracture at the most proximal stress riser. Although the Mau design is fixed across the plantar osteotomy, the longer lever arm weakens the overall stability of the osteotomy.

Base Wedge Osteotomies

Proximal wedge osteotomies of the first metatarsal have traditionally been treated as non-weight-bearing procedures. Minimal weight-bearing forces applied to proximal hinge osteotomies will rapidly result in failure. The main instability one encounters with these osteotomies results from the break in the proximal plantar cortex. This immediately sets up a longer lever arm, which acts solely from a thin cortical hinge. As weight-bearing forces are applied, even minute distal/dorsal motion of the osteotomy will cause failure of the thin cortical hinge, and point of fixation. The techniques of interosseus wire fixation provide minimal to no added stability.

Any attempt to create or increase compression across the osteotomy will only nominally increase the strength of the fixation. Numerous studies have examined the stabilizing effect of screw fixation using an oblique base wedge osteotomy. The authors have also found that these proximal osteotomies would not be able to sustain full weight-bearing loads in the immediate post-operative period. The additional strength and stability created by compression alone is still easily disrupted by the forces of weight bearing. Careful analysis of the "hinge failure" mechanism suggests that a different mechanical principle is necessary to accomplish significant increases in stability of proximal hinge osteotomies. Splintage and "tension-band" mechanics are employed to resist deformation, and in doing so, the fragile osteosynthesis is protected (Figs. 8A, 8B).

First Metatarsal Arthrodesis

Mechanical stability of the Lapidus arthrodesis is difficult to achieve. Weight-bearing forces acting to displace this arthrodesis are the greatest of any first metatarsal procedure, simply due to the fact that the lever arm is longest here. Again, the tension side of the metatarsal-cuneiform joint is the weakest region. Although internal fixation provides stability across this joint when properly applied, none of the techniques adequately protect against weight-bearing forces. Any attempt to provide compression along the tensile (plantar) surface of the joint will significantly improve its stability against extrinsic forces. The 4.0 mm screws illustrate this principle.

The data illustrates a significant increase in weight-bearing stability when comparing plate fixation with and without an interfragmental compression screw. In addition to providing additional interfragmental compression, the lag screw actually functions as a tension band device, resisting gapping of the plantar aspect of the arthrodesis which is subjected to a weight-bearing load. The metatarsal will actually elevate similarly to a rigid beam effect until failure occurs with screw pull-out.

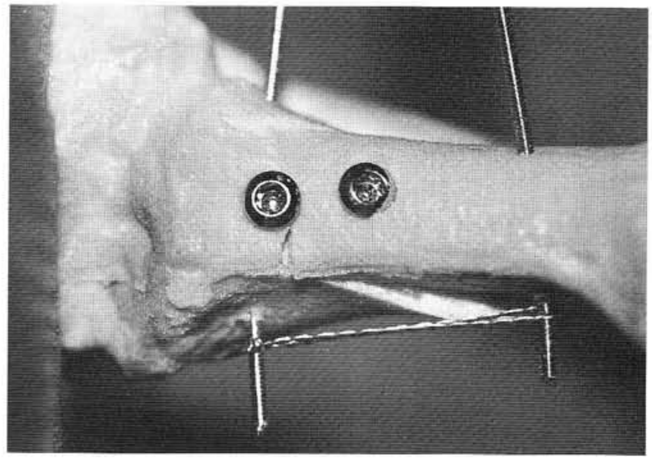

Figure 8A. Close-up of an oblique base wedge osteotomy with a plantar "tension band". Note the convergence of the dorsal wires and the parallelity of the plantar wires as dorsiflexion forces as applied to the metatarsal. The "tension band" maintained supports the fragile cortical hinge of the osteotomy.

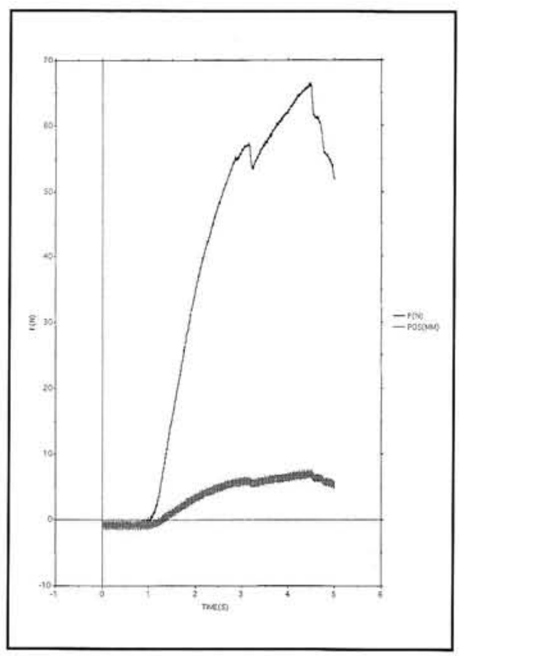

Figure 8B. The force plot for an oblique base wedge osteotomy.

The synergistic effect of the two fixation techniques (plate/axial compression plus splintage and single interfragmental screw) are consistently demonstrated to be superior (Figs. 9A, 9B). In theory, the closer the plate is placed to the plantar surface, the more effectively it will function as a tension band. Due to the level of dissection which would be required to place the plate on the plantar surface of the metatarsal cuneiform joint, this may not be a practical mode of fixation. However, the tension band effect will still exist to some degree with the plate placed on the medial side of the joint, providing that the plate is placed below the neutral axis of the bone.

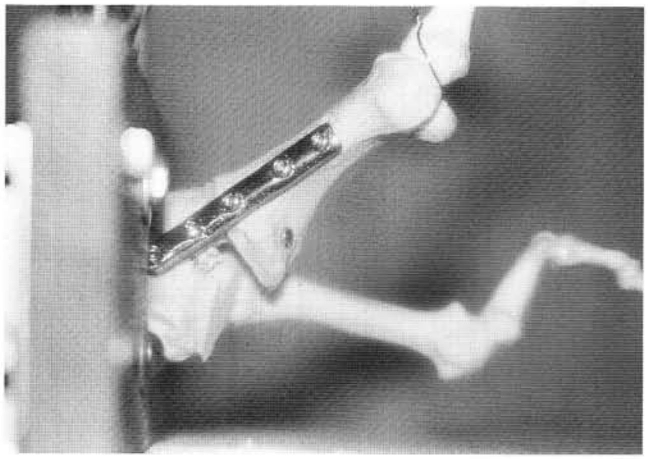

Figure 9A. Model failure of a first metatarsal-cuneiform arthrodesis utilizing a medial 5 hole plate with an interfragmental compression screw. The screw entering plantarly attempts to approximate the a tension band environment important in enhancing stability to this fusion.

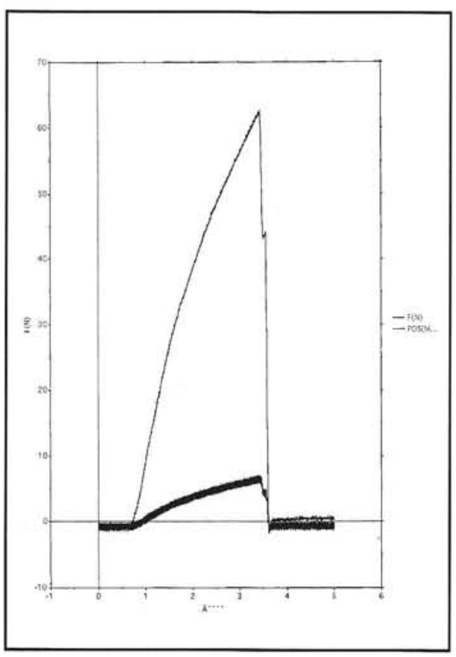

Figure 9B. Force plot of the model failure of a first metatarsal-cuneiform arthrodesis.

Bone Models

The selection of an appropriate bone model for this study deserves mention. Although cadaveric specimens have generally been accepted as the ideal model in mechanical testing experiments, the variability (age, length, width, bone density) between cadaver specimens is also known to be significant. In this study, the goal was to minimize the variability from specimen to specimen in order to focus all attention on the geometric issues. By simplifying the experimental conditions accurate comparisons of the relative stability of each configuration could be studied. As testimony to this technique, reproducibility of the results were found to be high as exemplified by the load required for failure, and the fracture pattern of the unaltered (no osteotomy and no fixation) bone model. This reproducibility was also evident in the osteotomy/fixation configurations examined. Despite the relatively small sample sizes used, the patterns of failure and loads at failure were consistent in each group.

The bone models do posses shortcomings in that they lack a medullary canal. Unlike real bone, foam models are essentially anisotropic. Therefore, the troughing effect, which can occur when metatarsal osteotomies are performed, was never observed. The added effect of impaction of the distal metatarsal on the metatarsal shaft was performed to some extent with the foam models, but not as effectively as with cadaveric specimens. Also, foam models do not have the viscoelastic material characteristics of natural bone. Consequently, they are less responsive to the loading rate then comparable experiments performed with cadaveric bone.

However, these factors do not impact on the value of the research. This project was designed to eliminate this multitude of variables in order to focus on issues which the surgeon can control. Specifically, the study has established how each of these osteotomies and fixation configurations perform relative to each other, under the most optimal conditions. This technique, and the information gathered will provide a basis for future cadaveric studies.

SUMMARY

Strengths

After analysis of the data and the failure patterns of the individual test models, certain strengths and weaknesses of osteotomy design and fixation technique can be concluded. Strengths that can contribute to a successful end result include proper knowledge of the following:

Mechanical design of the osteotomy. Particular care should be taken to consider the position of stress risers, and intrinsic stability of the fusion or osteotomy site.

Selection of internal fixation. Selection must not only include the particular device, but also the position it is used in. The study indicates that compressive and tension band devices work well when loaded under tension in the proper direction. However, no mechanical advantage is gained by using a good fixation device in an inappropriate location.

Postoperative management. Controlled post-operative management is essential. In particular, when dealing with more proximal osteotomies, or with proximal arthrodeses, care must be taken not to exceed the capabilities of the fixation device selected. The decision to allow the patient to ambulate is multi-factorial, and prudent consideration will include an assessment of the patient's weight, activity level, compliance level, the quality of bone and fixation observed intra-operatively, the type of surgical procedure, signs of postoperative progress, and ultimately the surgeon's best clinical judgement.

Weaknesses

Certain weaknesses worth noting were also identified in various configurations throughout this study. These are:

Mechanisms of osteotomy failure. Failures always resulted from the application of vertical loads. However, the mechanism of failure was variable from combination to combination. However, several distinct trends emerged.

1."Chevron" type osteotomies require fixation across the plantar arm of the osteotomy. Without fixation in this location, plantar gapping and dorsal rotation of the capital fragment was consistently seen. This finding correlates well with previous studies[2] (Fig. 10).

2. Proximal "hinge" osteotomies and arthrodeses consistently fail by plantar gapping. This results from an osteotomy configuration in which there is essentially no resistance to tension on the plantar aspect of the osteotomy. In this case, the only resistance to this tensile load is the osteotomy hinge. This is extremely ineffective due to the minimal bone stock located beneath the neutral axis, and the large moment arm of the first metatarsal which tends to rotate the distal segment around the bone hinge. In the case of an arthrodesis, where no hinge exists, the situation is intensified. Here, only the fixation is able to resist distal metatarsal elevation.

Pitfalls of Internal fixation. (Stress risers) It is important to realize that the placement of an internal fixation device in bone may create a point whereby a defect is initiated. Care must be taken to understand that the size of the defect created by the fixation device is not the sole determiner for that device's potential to create a stress riser (Figs. 11A, 11B). A small, but highly angular defect concentrates stress in a small area, and may be more likely to lead to failure. Smooth surfaces within the holes may reduce the chance that a stress riser will be created from a fixation device. Thus, when comparing a threaded device (screw) to a smooth device (K-wire), there are further considerations beside the diameter of the device. One must also consider the fact that the threads of the screw will score the surface they are embedded in, which could lead to a stress riser.

The diameter of the fixation device is crucial when a large portion of the bone stock is removed. This creates structural weaknesses in the bone

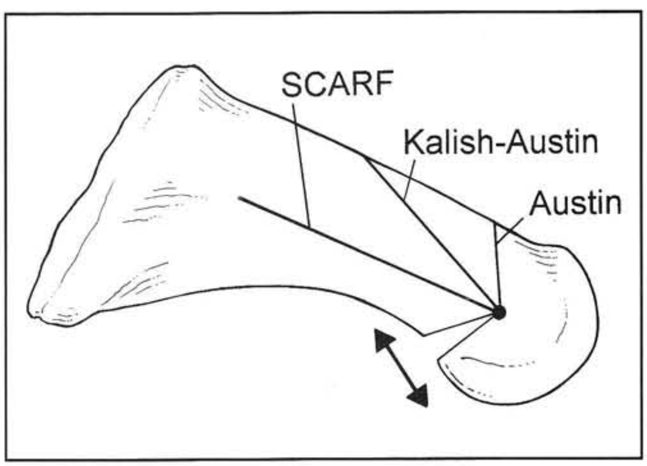

Figure 10. The plantar gapping and subsequent rotation of the capital fragment in osteotomy designs where the plantar arm is left unfixated.

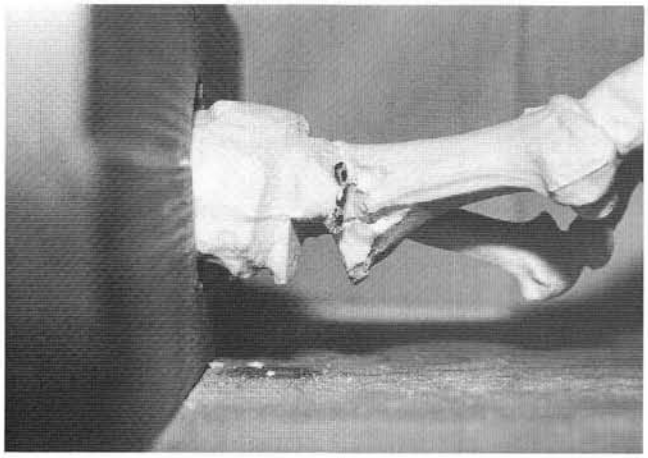

Figure 11A. An example of model failure illustrating the role of stress risers in providing weaker areas during applied forces to the metatarsal.

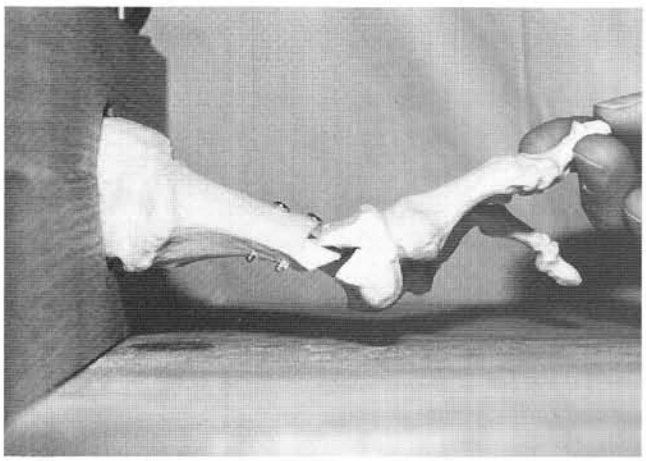

Figure 11B. An additional example of stress riser failure occuring at the most distal fixation device after rotation of the capital fragment.

beam. Thus, when loaded, stress will concentrate in this region because it is the least rigid, and failure will be most likely to begin here. It must also be remembered that the deformation of the metatarsal results from a cantilevered load. Therefore, the more proximal the bone defect is, the greater the bending moment on that point, and the greater the chance of failure when the load is applied distally.

Disruptive Force. Even the best method of fixation combined with the best arthrodesis or osteotomy procedure may fail as a result of early postoperative ambulation on the surgical site. By carefully selecting the osteotomy or arthrodesis site, and fixation modality, and then matching this to the patient, the chances for success will be greatest.

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