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THE STIFFNESS RESPONSE OF SEVERAL
EXTERNAL FRACTURE FIXATION DEVICES

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THE STIFFNESS RESPONSE OF SEVERAL EXTERNAL FRACTURE FIXATION DEVICES

By

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ABSTRACT

THE STIFFNESS RESPONSE OF SEVERAL
EXTERNAL FRACTURE FIXATION DEVICES

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The purpose of this research was to study the stiffness response of several external fixateurs of varying geometric designs. The fixateurs were applied to excised canine tibias and the system subjected to an axial deflection. The maximum stiffness of the system was obtained from the linear portion of the load/deflection curve. The results of these experimental tests indicated that variations in geometry greatly effect the stiffness response of the fixateur. The axial stiffness could be effectively increased by increasing the number of connecting bars, angling the fixateur pins or using a bilateral configuration.

An analytical analysis was initiated in an attempt to provide a predictive model for the various fixateurs. The linear response of this model did not imitate the nonlinear response of the experimental device. However, the results from the analytical model accurately duplicated the trends in stiffness response associated with the various geometries.

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TABLE OF CONTENTS

LIST OF FIGURES

Figure | Page

LIST OF TABLES

I. INTRODUCTION

For years, the treatment of long bone fractures has utilized numerous methods and devices, but the development of any new device requires an extensive understanding of the mechanical behavior of the components of the instrument. The present acceptance of biomechanical engineering principles in the field of orthopaedics, and the increased knowledge of fracture healing have made this understanding an easier task to accomplish. The combined knowledge of surgeons and engineers has led to the development of a greater number of orthopaedic appliances for use in fracture reduction and trauma management. One such application is the external bone fixation device.

First invented by Clayton Parkhill, M.D., in 1897 (14), the external fixateur has undergone numerous improvements and modifications over the years. Useful for both humans and animals, the external fixateur allows for soft tissue management as well as rigid skeletal fixation. The fixateur applied externally also permits freedom of joint motion above and below the fracture site which encourages circulation and minimizes muscle and bone atrophy. This type of device has been used in a variety of anatomical locations including the femur, tibia, humerous, ulna, radius, and pelvis. The external fixateur may be employed for the treatment of open fractures, infected nonunions, stabilization of bone fragments, osteotomies, and even for limb lengthening. Such versatility results from the numerous geometrical configurations available for these devices. The form of fixation applied depends largely on the location of the fracture, the nature of the disunion, and the amount of soft tissue involvement.

The design of any external fracture fixation device must satisfy several criteria. The application of the fixateur should be easily managed and allow for later adjustment should it become necessary. The materials used in the frame must be biocompatible with the biological environment they are subjected to, and management of soft tissue injuries should be simple and unobstructed. Durability and variability are important properties of the fixateur, required to handle numerous types of fractures in many different locations. If possible, early patient mobility is also desired. The most important criterion though is high rigidity, or stiffness, of the overall fixateur system.

During rehabilitation of the patient, the external fixateur is subjected to various loading conditions and these must be taken into consideration in fixateur design and pin placement. The fixateur should be able to prevent excessive tensile forces at the fracture site causing disunion, and yet allow the transmission of compressive forces across the fracture thereby promoting healing. This can only be accomplished by varying fixatuer geometry for different types of fractures. For trauma management in long bone fractures, fixateur configurations can be catagorized into five geometric groups - unilateral, bilateral, quadralateral, triangular, and circular. These five groups are based upon the number of connecting bars and the shape of the resultant frame structure.

The mechanical behavior of these devices varies widely. Therefore, extensive experimental work is needed to determine the mechanical properties of each fixateur design. Although such experimental testing is important and highly effective in evaluating the mechanical performance of each device studied, it suffers from many disadvantages. The testing procedure itself is extremely time consuming since the number of geometrical variables is large. These include the number and location of the transfixing pins, their diameter and the direction of orientation. The diameter of the connecting bars can also be varied, as well as their number, arrangement, and the distance located from the axis of the bone. Several non-geometric parameters can also be altered, including pin and frame material and the method of loading. Simple experimental methods provide no information on the internal stresses in each component in the system or at the complex pin-bone interface.

Therefore, based on known mechanical parameters and basic structural analysis, theoretical models have been developed to predict the behavior of such external fracture fixation devices. These models allow rapid study of the effects of variation of the numerous parameters. Finite element methods and computer simulation have been utilized to provide overall stiffness data as well as data on the internal stresses developed in the system.

II. SURVEY OF LITERATURE

Since its first application in 1894, the external fixateur has been modified and utilized in numerous ways and many published reports have discussed its behavior. In 1897, Parkhill, himself, reported on its use in 14 human cases (15). Shortly after, in 1902, Dr. Albin Lambotte of Antwerp, Belgium independantly designed and successfully installed an external fixateur. In documenting his use of the external fixation device, Lambotte states: "In numerous cases, I could avoid, thanks to the fixateur, amputations that seemed inevitable." (13) A half-frame fixateur was brought into clinical use in 1937 by a Pennsylvania veternarian, Otto Stader (17). The unilateral fixateur design has been in constant use in veternary medicine since that time. The external fixateur was further developed by Roger Anderson, M.D., however, his design lacked the ability to adapt to different types of fractures or to provide a variety of configurations (1).

In veternary practice, in the mid 1940's, the Kirschners redesigned the clamps utilized by Anderson and the Kirschner splint became the most commonly used fixateur for animals. This popular splint used two pins clamped securely to a single connecting bar. For those cases where additional stiffness was needed, a second bar was added (Figure 1), but this configuration tended to be extremely bulky and expensive.

In human medicine, many surgeons altered the existing designs to suit their needs. During the second World War, a Swiss surgeon, Raoul Hoffmann, modified and used Lambotte's fixateur. The hardware for this

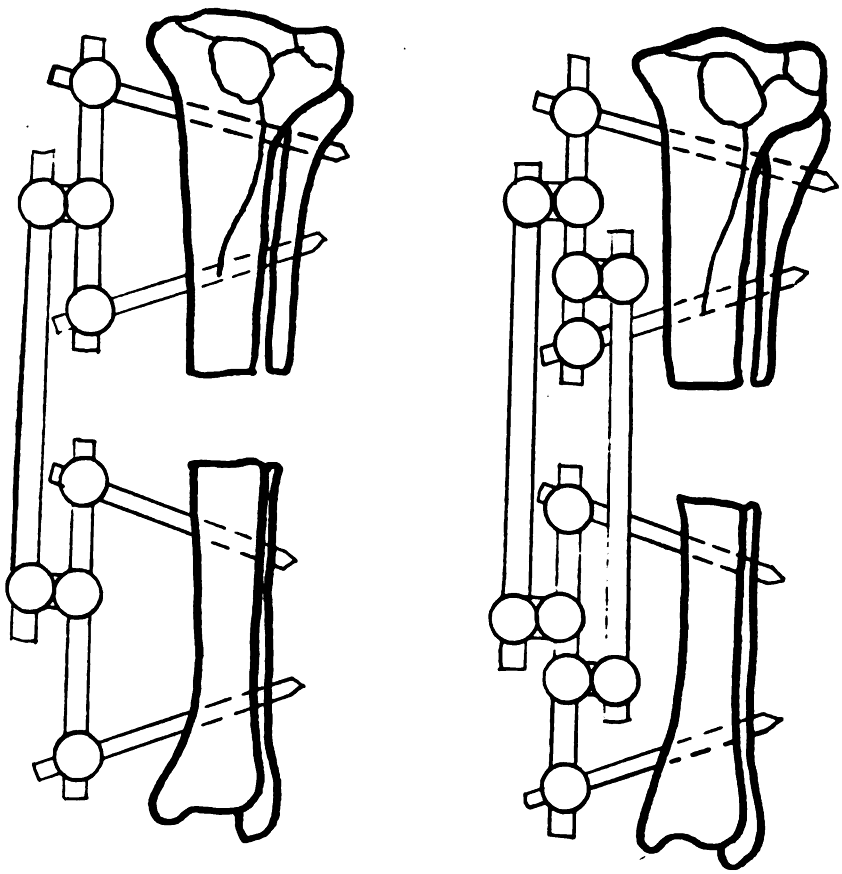

FIGURE 1. Original Kirschner Splints

new fixateur became known as the Hoffmann device. Frequent pin track infections and fracture disunions during the war, resulting from most types of external fixation, discouraged the use of the fixateur for treatment of fractures. The trial of the Haynes and Anderson fixateurs, during the same period, exposed the need for an increased study of the mechanical basis for external fixation. Initial investigations into the engineering concepts of external fixation renewed interest in this form of treatment.

Preliminary biomechanical analysis of several fixateurs led to the realization that the behavior of each device depended greatly on the design and method of application. The Hoffmann half-frame was subjected to much experimentation in the early 1970's. J. Vidal was one of the first to study and modify this fixateur. With his addition of a quadralateral frame and compression screws, the device became known as the Hoffmann-Vidal apparatus and is now widely used in numerous orthopaedic applications. F. Burney, R. Bourgois, and M. Donkerwinkle studied the properties of the Hoffmann half-frame using a mathematical model developed on the basis of clinical observation (6). Upon examination of the numbers of pins used and their placement, they concluded that at least two pins must be used in each bone clamp, the clamps should be tightened as close to the bone as clinically possible and that two bars more than doubled the rigidity of the system. Further, their experimental analysis showed complete agreement with these statements.

The Hoffmann-Vidal quadralateral fixateur was analyzed by E.Y. Chao, Ph.D., B.T. Briggs, M.D., and M.T. McCoy, M.D. using both experimental and theoretical methods in an attempt to quantitate the basic mechanical properties of the fixateur (8). The experimental analysis was performed with 21 varying configurations and utilized five static loading conditions. In studying the fixateur stiffness at the fracture site, in the various loading modes, they found anterior-posterior bending to have the lowest value. The overall stiffness of the fixateur could be increased by increasing the number of pins or the pin diameter. Decreasing the pin length, moving the connecting bar closer to the bone, or decreasing the pin separation in each segment also helped to increase the fixateur rigidity. The same group performed theoretical analyses using a two-dimensional finite element model of the device. Significant disagreement was seen between the theoretical and experimental results. This was claimed to be due to the fact that a two-dimensional model was being used to examine a three-dimensional system. Chao and Briggs (4), and Chao and K. An (7), also analyzed the Hoffman-Vidal apparatus. Besides altering the geometrical parameters of the system, which resulted in the same conclusions as before, they varied the materials used for the fixateur components. Of the two materials tested, stainless steel and titanium, it was reported that use of titanium pins and titanium frames decreased the overall stiffness 41%. This would be expected since titanium ($E = 16.5 \times 10^6$ psi) has an elastic modulus 41% lower than stainless steel ($E = 28.0 \times 10^6$ psi). However, if stainless steel pins are used with a titanium frame, reduction in rigidity is minimal and the overall weight of the fixateur is decreased considerably.

In the mid 1970's G. Hierholzer and A. Chernowitz attempted to specify by fracture type the tubular system of the ASIF (Association for the Study of Internal Fixation) (12). Type I (unilateral) was proposed for use in an open fracture or closed comminuted fracture of the thigh. A Type II fixateur (bilateral) was suggested for fractures with bony support at the fracture site. To bridge a large distance or to fix a small metaphyseal fragment, they recommended the Type III (triangular) design. F. Behrens and K. Searls also analyzed the ASIF frame (Type II), reporting on the mechanical and clinical shortcomings of the bilateral design for tibial fractures (3). Despite its popularity, they found that full pins, also called through and through, can often cause compartment syndromes and injuries to the anterior tibial artery. Impalement of muscle is inevitable in this case and, many times, this leads to a permanant decrease in ankle motion. Behrens and Searls also noted that the two connecting bars often interfere with trauma management and that full weight bearing was rarely allowed until complete healing had occurred. They concluded that most of these shortcoming could be overcome with the proper unilateral design.

The Oxford fixateur utilizes a unilateral configuration in an attempt to overcome the disadvantages of the bilateral designs. In 1979, M. Evans, J. Kenwright, and K. Tanner examined the factors that contribute to the deflection of Oxford system (11). They found that the bending of the pins could contribute to over 1/2 the total deflection. Deformation of the bone at the site of pin insertion also added to the total axial deflection. Many authors have analyzed this highly complex pin-bone interface region. Since little experimental data has been aquired concerning this area, theoretical models have been developed to analyze the stress distribution at the pin insertion site. Chao and An developed such a model utilizing finite element methods and computer simulation (7). Based on this three-dimensional model, the highest compressive stresses were found to be located at the outer portion of the bone cortex directly above the pin. Chao and An also applied cyclic bending loads to fixater pins of various diameters to determine their fatigue properties. The larger pins were found to have both higher yield strengths and higher fatigue strengths under similar loading modes. The larger diameter pins were shown to increase the system's rigidity, however, Chao and An suggested that they may also increase the stress concentration in the bone near the pin insertion site.

Most of the previously mentioned fixateur designs are used solely for human applications. In veternary practice, the most commonly used configuration is the unilateral design. Wade Brinker, D.V.M., and Gretchen Flo, D.V.M., reported on its clinical use in 1975 (5). They established a modified procedure utilizing two pins in each bone segment connected by a single bar. They also found this fixateur design to be extremely useful in conjunction with other methods of fixation including intramedulary pins, lag screws, and orthopaedic wire. In 1982, Erick Egger, D.V.M., investigated the static strength of six different configurations utilizing the Kirschner equipment (10). He found the double clamp configuration (original Kirschner design) to be the weakest of all those tested. The triangular configuration was found to be the strongest design, yet its clinical use is limited by its obstruction of soft tissue injuries.

III. EXPERIMENTAL METHODS

In order to study the compressive strength of the fixateur, in vitro studies were carried out on fresh canine tibial bones. The bones were frozen immediately after removal from the animal and stored until testing could be performed. Before testing, each tibia was placed in a bath of warm water until completely thawed. If the bone was not completely thawed before potting, a pocket of moisture would form around the bone ends and allow unwanted motion during testing. All soft tissues were removed from the proximal and distal ends and small nails were inserted radially around these ends to aid in gripping. The articulating surfaces were allowed to dry slightly, since the potting material would not adhere to moist surfaces. The potting substance, Devcon Plastic Steel* (SF), was formed by mixing 3 parts epoxy with two parts hardener until a soft putty resulted. This putty was placed into a plastic form and the proximal end of the tibia embedded into it. The axis of the bone could then be lined up perpendicular to the base and the distal end clamped to hold its position (Figure 2). After approximately fifteen minutes, the putty had hardened sufficiently enough that the potting form could be inverted and the distal end of the bone set in the same manner. When completely hardened, the putty provided excellent gripping of the bone ends.

*Devcon Corp., Denver, Mass. 01923

FIGURE 2. Bone Potting Form

Thus potted, the bone was clamped into a specially designed pinning form with the proximal end to the left and tuberosity pointing up. The fixateur pins could then be inserted into the shaft of the tibia. The pins tested were the 2.5 mm and the 4.0 mm Sythes External Fixateur No. 395* and the 1/8 inch Kirschner Splint**. Pin guides were drilled so that all pins could be placed in the same plane and at a constant angle to the axis of the bone. Pins were inserted into the bone at either 72.5° or 90° to midline. When the connecting bar is attached, any pins not in the same plane tend to prestress both the bar itself and the other pins in the system. Thus lined up, the guide was clamped to the pinning form to create a rigid system. Four seperate guides were drilled; two for the unilateral configurations and two for the bilateral configurations. The 2.5 mm pins required one guide while the 4.0 mm and 1/8 inch pins utilized another pin guide. The length of each bone sample determined the pin size used. Bones measuring 7.5 inches to 8.0 inches used the 2.5 mm pins and any bones longer than 8.0 inches used either the 4.0 mm pins or the 1/8 inch pin.

The pins were inserted through the guide and into the shaft of the tibia using a combination low speed power drill/hand chuck. The drill was used to make the initial perforation and the hand chuck was used to complete the pinning and to duplicate actual sugical procedure. Pins inserted solely by the power drill tended to loosen rapidly during testing. To minimize the drilling during the testing procedure, the maximum number of pins were inserted while the sample was clamped to the

*Synthes Ltd. (U.S.A.), P.O. Box 529, Wayne, PA

**Kirschner, P.O. Box 459, Aberdeen, Maryland 21001 pinning guide. The connecting bar/bars were also attached while the bone was held fixed and the clamps were tightened securely. (Figure 3) This rigid form reduces the geometric variables of the fixateur by keeping both the placement of the pins and the distance from the bone to the first connecting bar a constant between specimens.

After all clamps were secured, the system was removed from the pinning device and a 1/2 inch section of bone was cut away at a midpoint between the two center pins. This was done to ensure that no contact occurred at the fracture, since fracture site compression tends to increase fixateur stiffness. Also, this amount approximates a condition of a markedly comminuted fracture, one of the primary uses for external fixation. To simulate actual joint motion, special grips were designed to allow the fixateur to bend in the direction of least resistance. The ends of the grips used ball and socket joints to approximate in vivo motion. The testing machine used was an Instron* servohydraulic materials testing machine. The actuator was mounted in the upper crosshead to reduce mechanical noise and vibration. Load and stroke data were monitored and stored on a Nicolet digital oscilloscope, coupled to a mini floppy disk drive that allowed for permenant data storage.

For consistency in protocol, each fixateur configuration was compressed the same distance at a constant displacement rate, 0.01 inch/ sec. The displacement was then reversed to return the system to its initial unloaded condition. A deflection of 0.2 inch (5.08 mm) was chosen after ten consecutive tests were completed on the known weakest configuration. Maximum deflections tests started at 0.1 inch and

*Model 1331, Instron Corp., Canton, Mass.

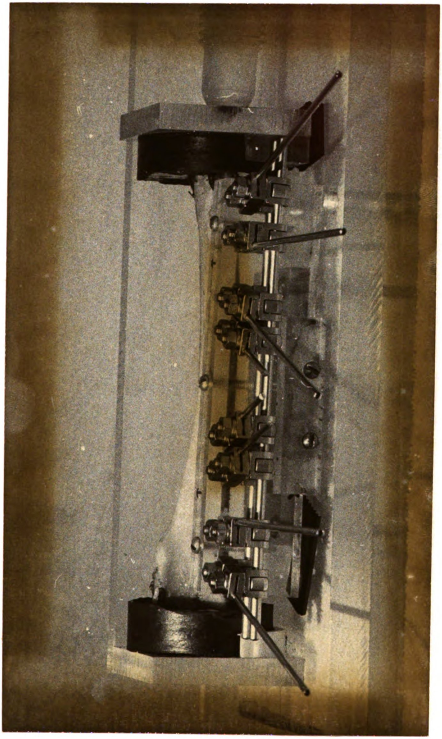

FIGURE 3. Bone Pinning Form

were increased until the slightest plastic deformation could be seen in any of the fixateur members after unloading. The 0.2 inch was well below the point of plastic response.

The test protocal began by filing any loose edges from the potted ends of the bone and then inserting the sample tightly into the grips. The entire system was then placed between the upper crosshead and the load cell mounted in the lower crosshead. Next, the actuator was lowered until contact was made between the grips and the test machine. (Figure 4) A 'no-load' condition was established by monitoring the load trace on the oscilloscope. After this initial state was attained, a 35 mm camera was placed perpendicular to the plane of the fixateur, at approximately four feet from the machine. A series of three photographs were taken for each test run; one before the test was started, another when the actuator reached 0.2 inch and a third at the end of the unloading step.

For the unilateral fixateurs, two tests were run for each configuration to ensure reproducibility. The bilateral designs required four test runs for each configuration. Two were run with the system oriented in the test machine as stated before and two were run with the plane of the fixateur oriented perpendicular to the frame of the machine (Figure 5). These extra tests were performed to observe the motion of the fixateur perpendicular to the plane of the device. It was observed throughout the testing procedures that bending occurred only in the plane of the fixateur for the unilateral configurations. On the other hand, bending could be seen in planes both parallel and perpendicular to the plane of the fixteur for bilateral configurations. Testing of the

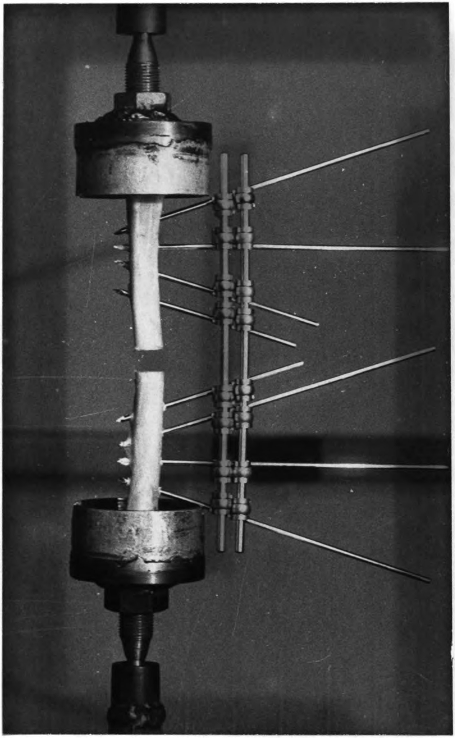

FIGURE 4. Normal Testing Arrangement

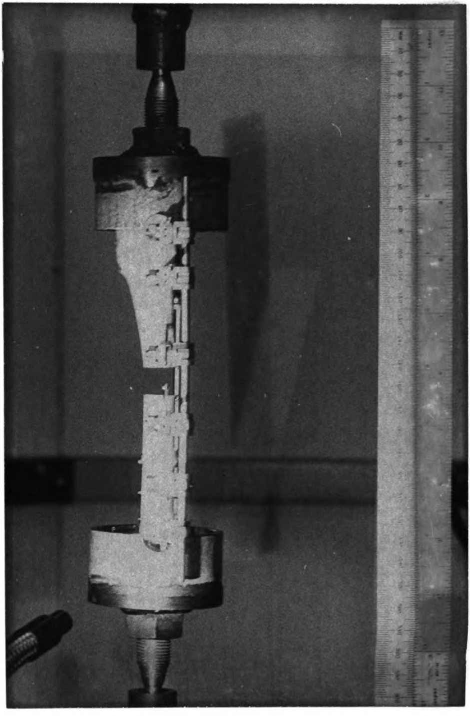

FIGURE 5. Perpendicular Testing Arrangement

unilateral fixateur utilized ten seperate configurations; three with a single connecting bar, six with two parallel connecting bars and one with two connecting bars oriented perpendicular to each other (Figure 6). The bilateral testing was performed on five seperate configurations (Figure 7). For the sake of comparison, a small number of tests were run utilizing bone plates. These plates were attached to the bone after potting and then removed again so that a section of bone could be removed at the midpoint of the shaft. The plate, either a 3.5 mm broad (DCP*) or a 4.5 mm narrow (DCP*), was then reattached and testing proceeded as before.

After each series of runs, the specimen was removed from the machine and the configuration was changed by carefully removing pins or by increasing or decreasing the number of bars. The new system was then replaced into the testing machine and the compression process repeated. Removed pins and bars were checked for any signs of plastic deformation, but none was seen in any of the tests run. An effort was made to utilize new pins and connecting bars in each test series to avoid the effect of any fatigue properties. Due to expense and availability, though, a few tests were run using previously tested pins, but no variations in results occurred.

Load and deflection data were collected as raw voltages on the Nicolet digital oscilloscope and stored on floppy disks for each test run. At the completion of each test series, the data was recalled and the peak load on the sample, corresponding to a compression of 5.08 mm, was calculated and recorded. The data from the disk was then transferred to a computer file and stored on a PDP 11/23 computer for

*Synthes Ltd. (U.S.A.), P.O. Box 529, Wayne, PA

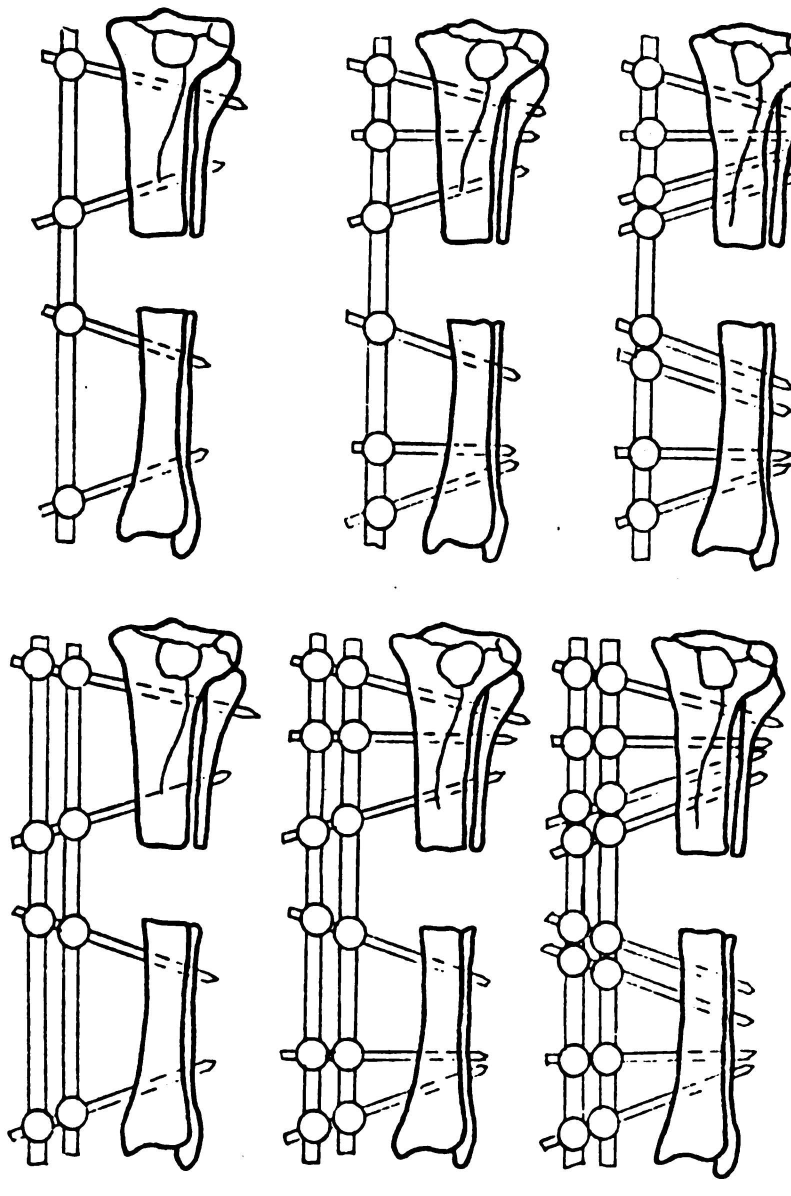

FIGURE 6. Unilateral Configurations

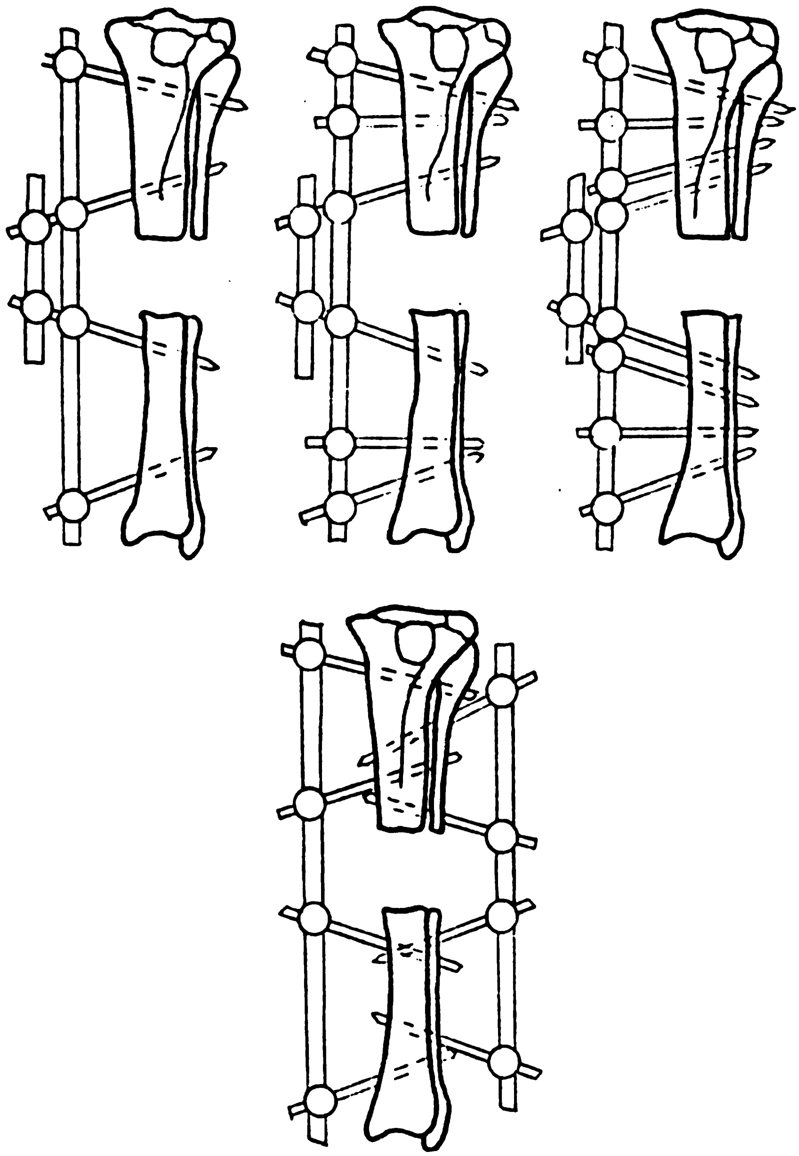

FIGURE 6. (cont'd)

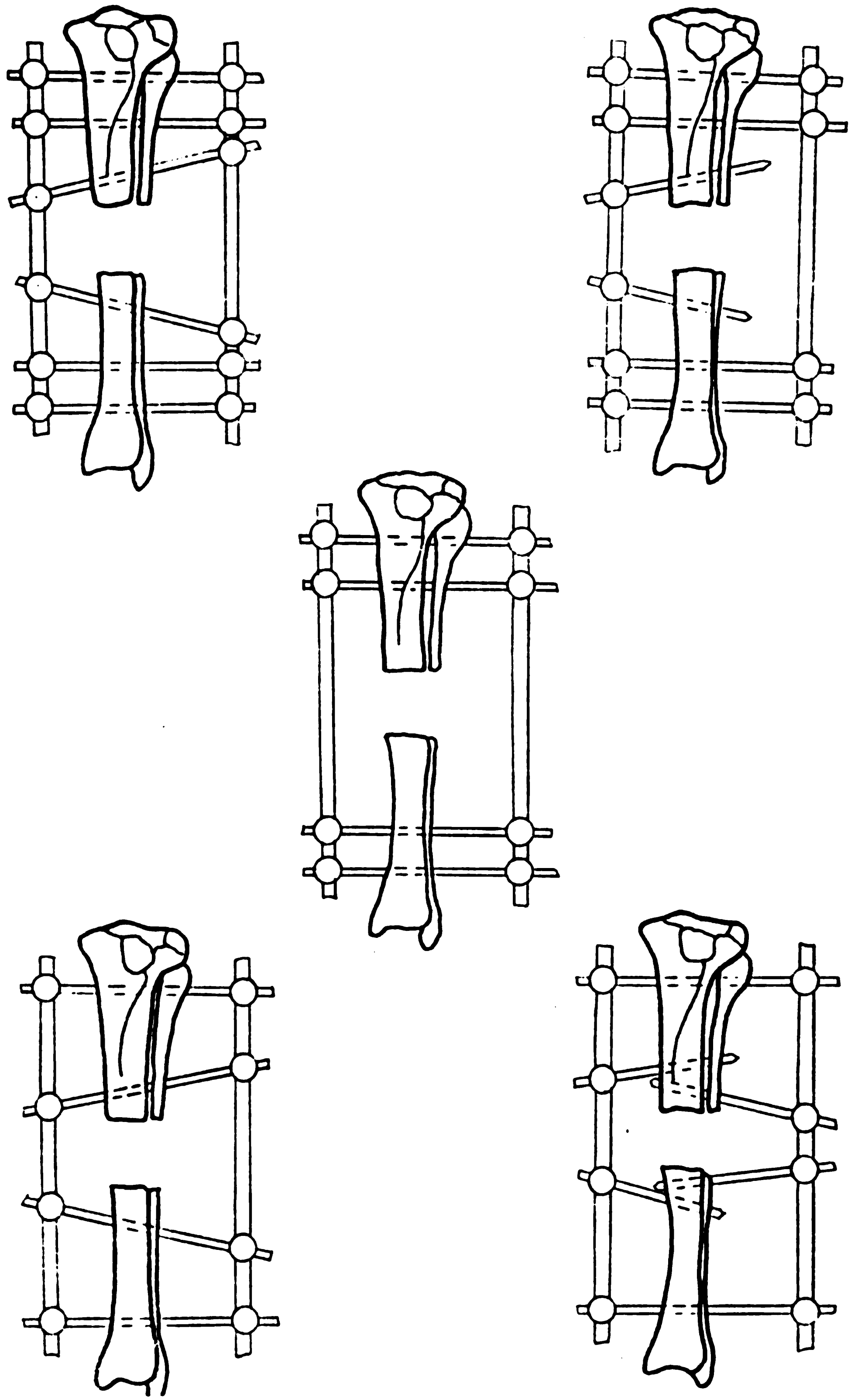

FIGURE 7. Bilateral Configurations

later analysis. A computer program was written to convert the data from voltages to equivalent mechanical values, expressing time-load-stroke as seconds-Newtons-millimeters. This same program was used to analyze the data and determine the maximum axial stiffness for the fixateur system during compression. The maximum stiffness was found by calculating the slope of the force/displacement curve generated by the loading data. The curve was analyzed by windowing a set of data, calculating the stiffness and then moving the window to the next data set and repeating the process. Each stiffness value was then compared to the previous one calculated to obtain the maximum value. As shown in Figure 8, a typical loading curve is nonlinear, and maximum stiffness occurs in the initial portion of the curve.

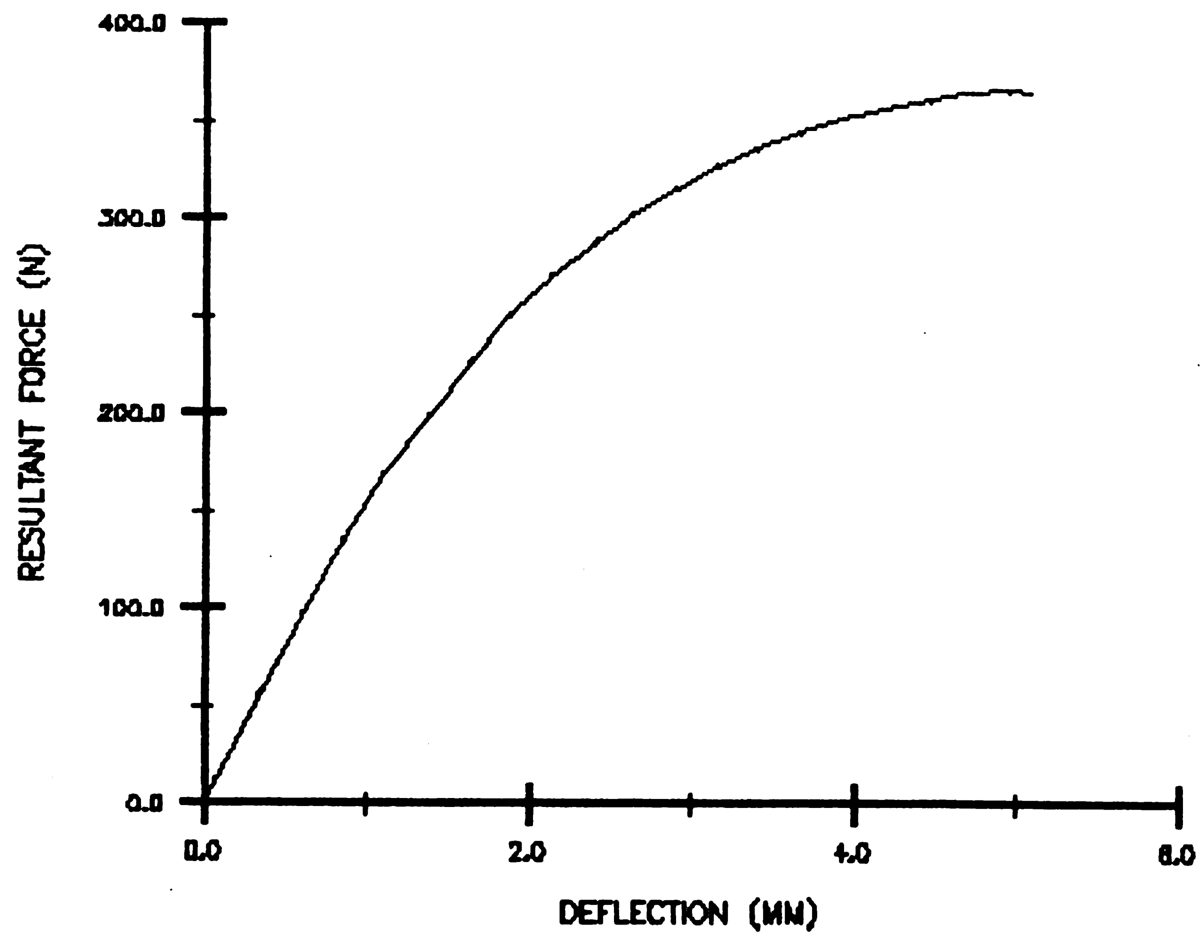

FIGURE 8. Typical Load-Deflection Curve

IV. EXPERIMENTAL RESULTS

Presentation and comparison of the results is difficult as fifteen different configurations were tested for each of three seperate pin sizes. These configurations will be grouped into unilateral and bilateral categories for easier examination.

Within the unilateral category, pin size and number were varied, as well as the number of bars and bar-pin attachments. Figures 9 and 10 compare the variation of connecting bar number and manner of attachment for the unilateral configurations utilizing two pins in each bone segment, for pin diameters of 4.0 mm and 2.5 mm, respectively. It can be seen from these figures that an increase from one to two bars does not greatly increase the stiffness if only the mid-pin is clamped. This procedure does not seem to have beneficial value. However, if all pins are clamped to the additional bar, the axial stiffness of the structure is increased by 195% for the 4.0 mm pin size and 100% for the 2.5 mm pin size. Figure 11 summarizes this trend and includes the 1/8 inch (3.175 mm) pin size data.

To assist in comparison with previous works, a single set of tests were performed on the two original Kirschner designs. (See Figure 1) The maximum axial stiffnesses of these splints are shown on Figure 12, along with the other configurations utilizing two 1/8 inch (3.175 mm) Kirschner pins. Both original splints showed a lower stiffness than any of the modified designs. The two-plane configuration produced

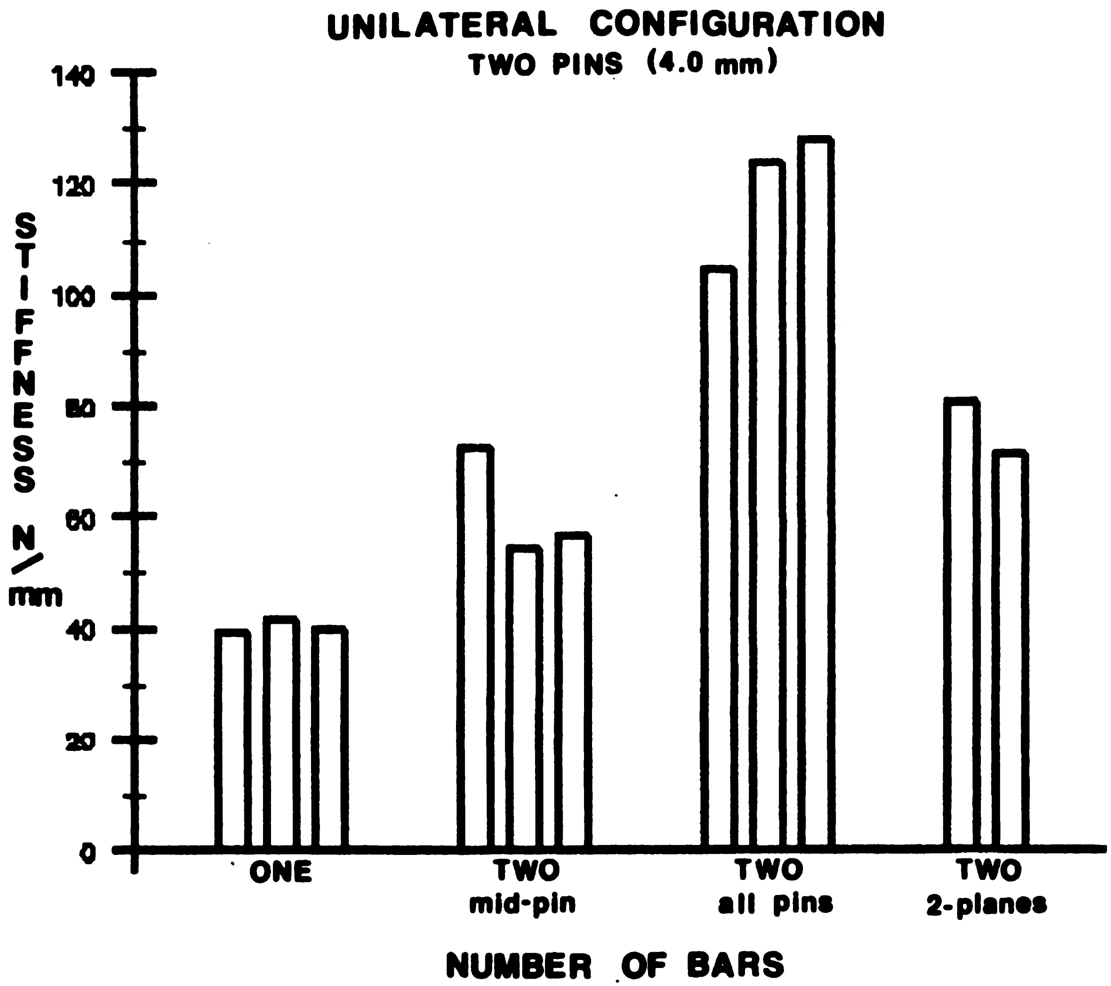

FIGURE 9. Stiffness Response with Varying Bar Number - 4.0 mm Pins

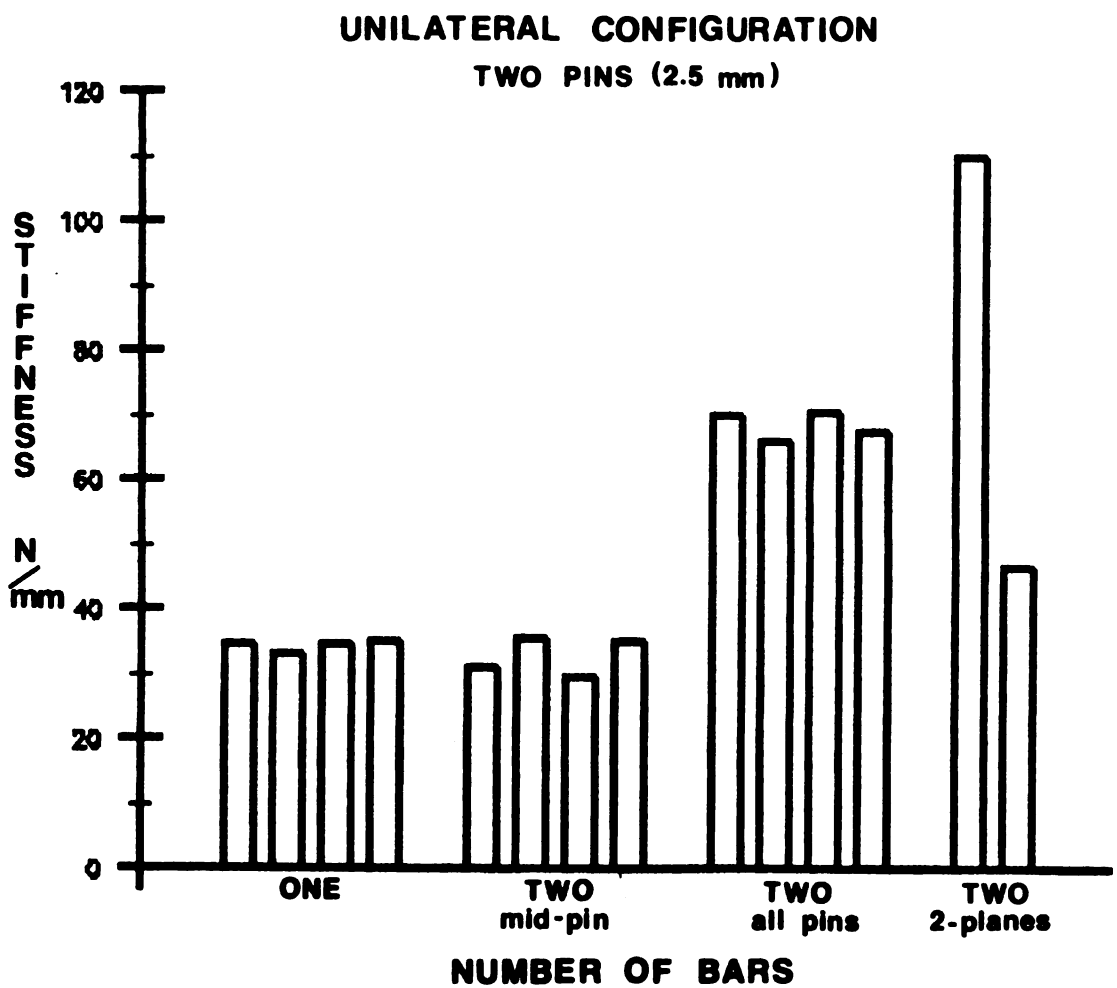

FIGURE 10. Stiffness Response with Varying Bar Number - 2.5 mm Pins

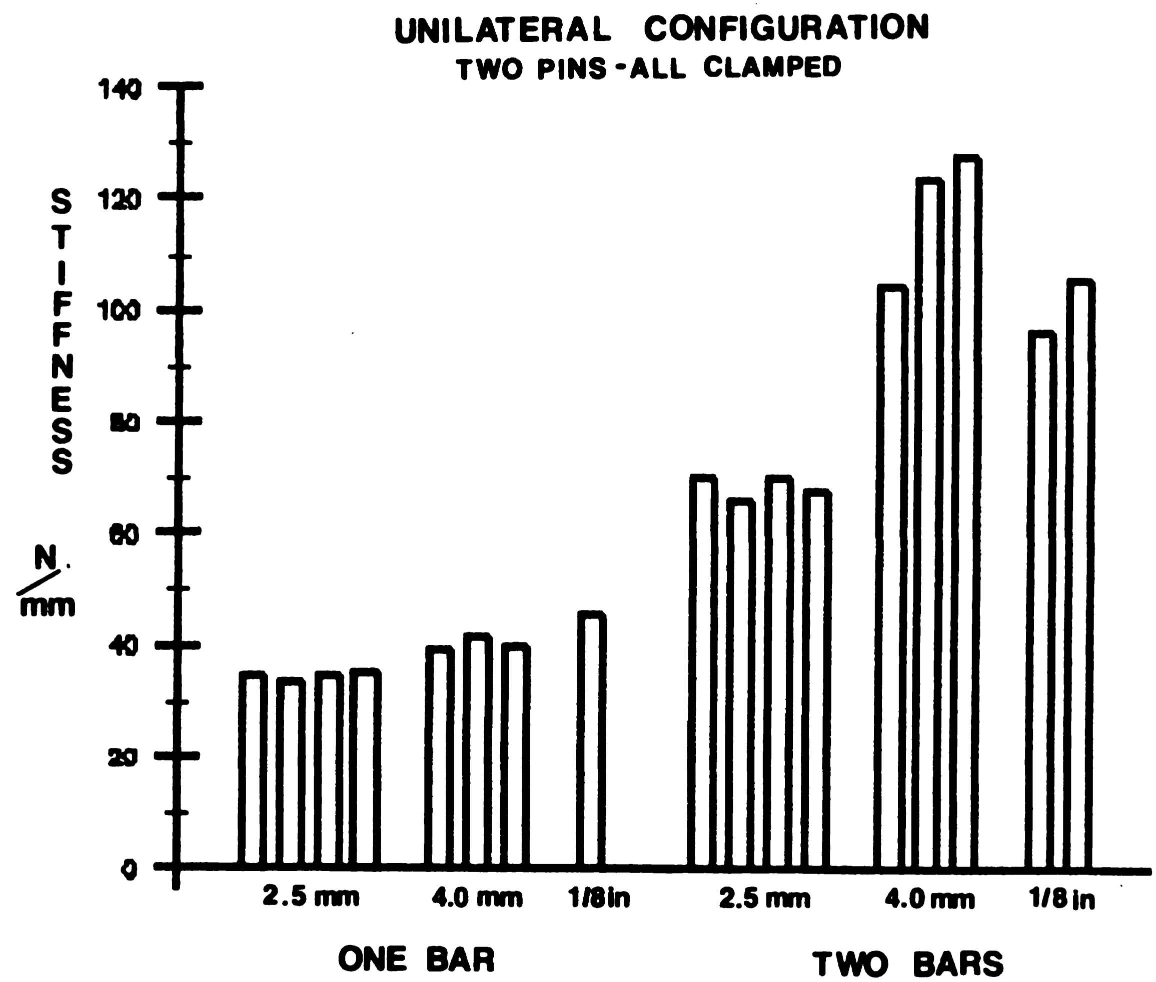

FIGURE 11. Stiffness Response with Varying Bar Number and Pin Size

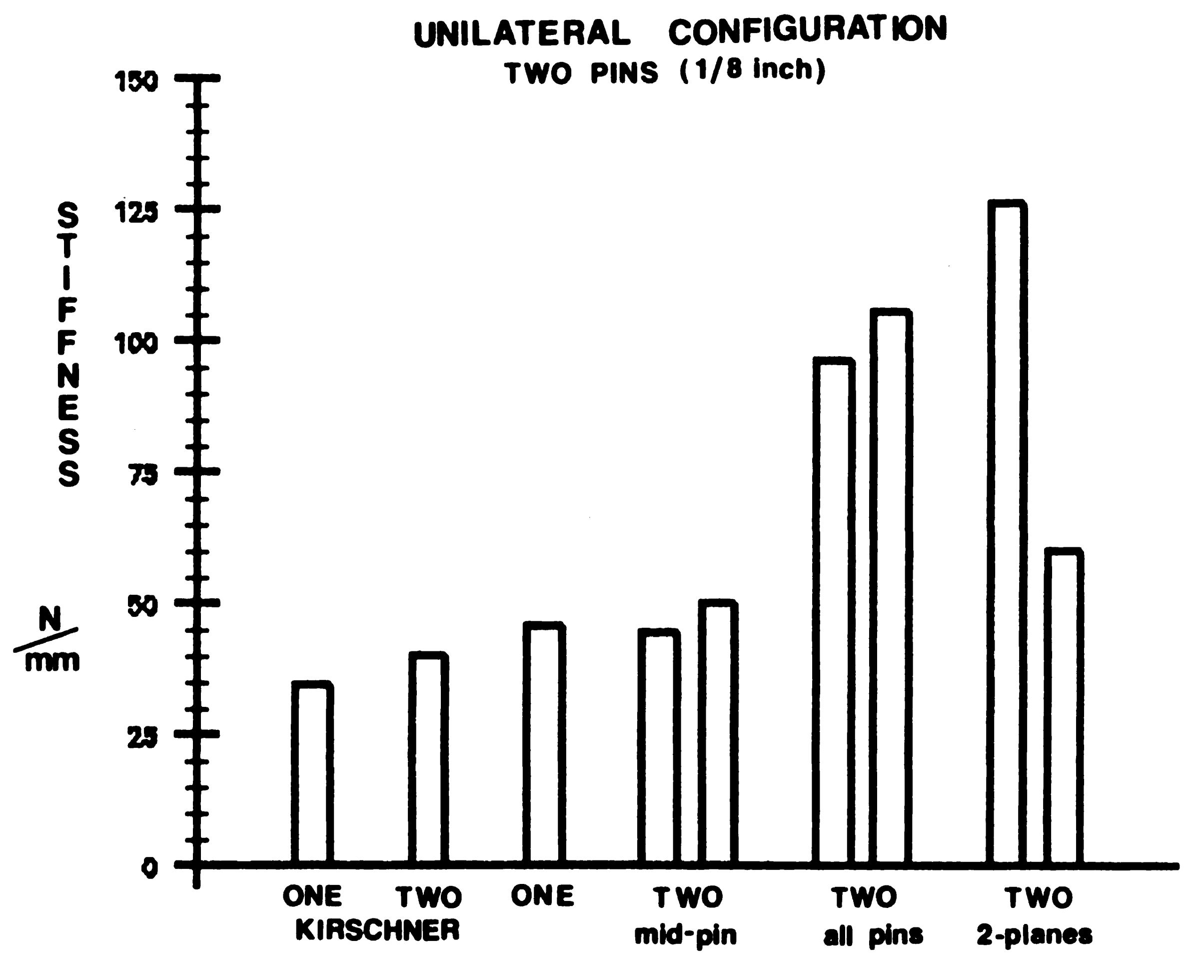

FIGURE 12. Stiffness Response with Varying Bar Number - 1/8 inch Pins

inconsistant results for all three pin sizes and comparison of this arrangement with the others is difficult. The scatter in the data is due to the variability of the pin placement in the plane orthogonal to the pinning form plane.

Figures 13 and 14 compare the variation of axial stiffness with an increase in the number of pins attached to a single connecting bar for the 4.0 mm pins and the 2.5 mm pins, respectively. These figures indicate that a minimal increase in stiffness is obtained by increasing the number of pins in each bone segment. Therefore, there is no mechanical advantage to the additional tissue invasion by increasing the number of pins beyond the typical value of two. Figure 15 shows the change in stiffness with pin number increase for the configurations utilizing two connecting bars with all pins doubly clamped. Both the 2.5 mm diameter pins and the 1/8 inch (3.175 mm) diameter pins showed a definite increase in stiffness with increasing pin number, typically from 44 to 102 percent.

Tables 1, 2, and 3 summarize the maximum stiffness and maximum load values obtained from analysis of experimental data. It should be noted that the maximum load and the maximum stiffness do not occur at the same point on the load/deformation curve. The stiffness of the system decreases with increasing loads and deformations due to the nonlinear bending component of the axial loadings in the pins and bars.

Within the bilateral category, five seperate configurations were examined for each of three different pin sizes. Figure 16 and 17 compare the change in stiffness with the varying frame designs for the 4.0 mm diameter pins and the 2.5 mm diameter pins, respectively. By

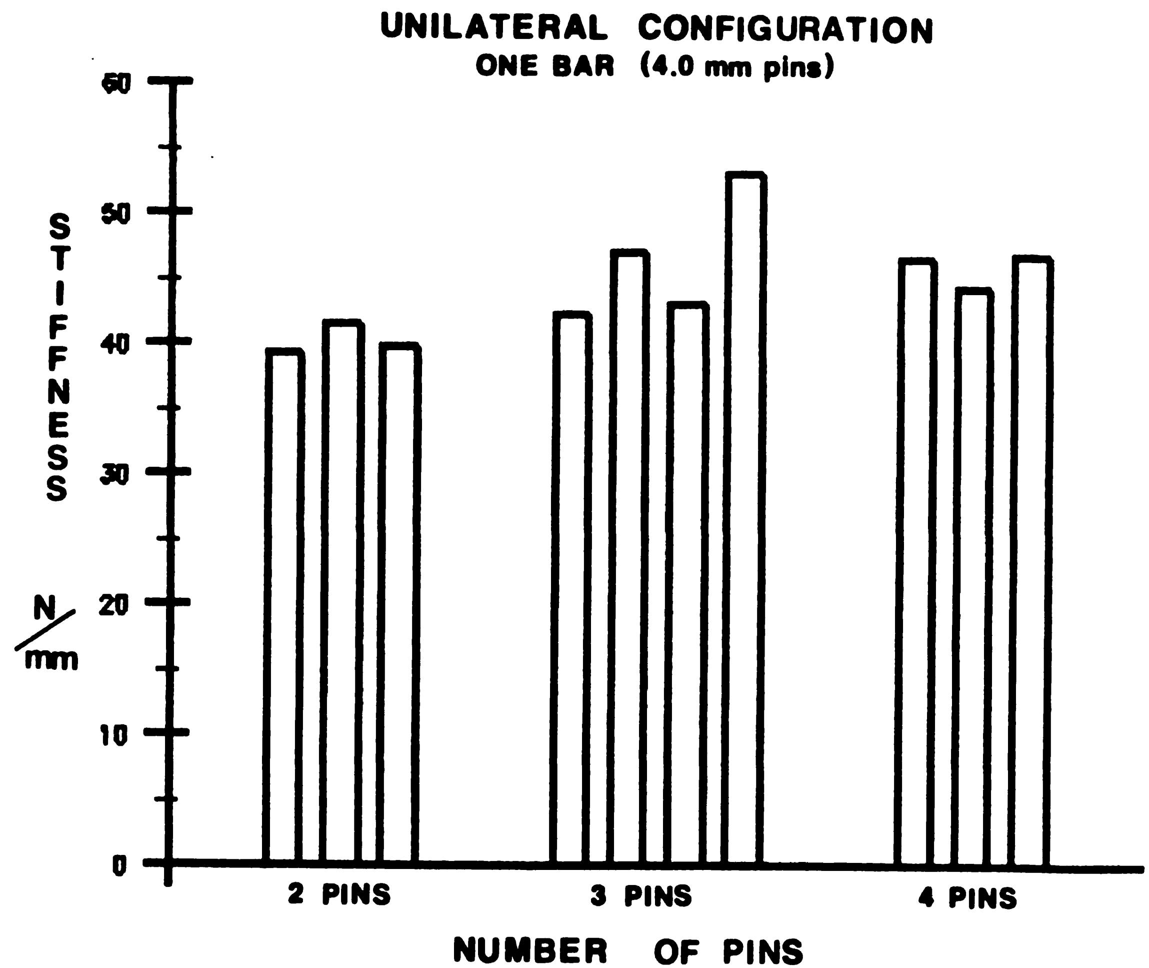

FIGURE 13. Stiffness Response with Varying Pin Number - 4.0 mm Pins

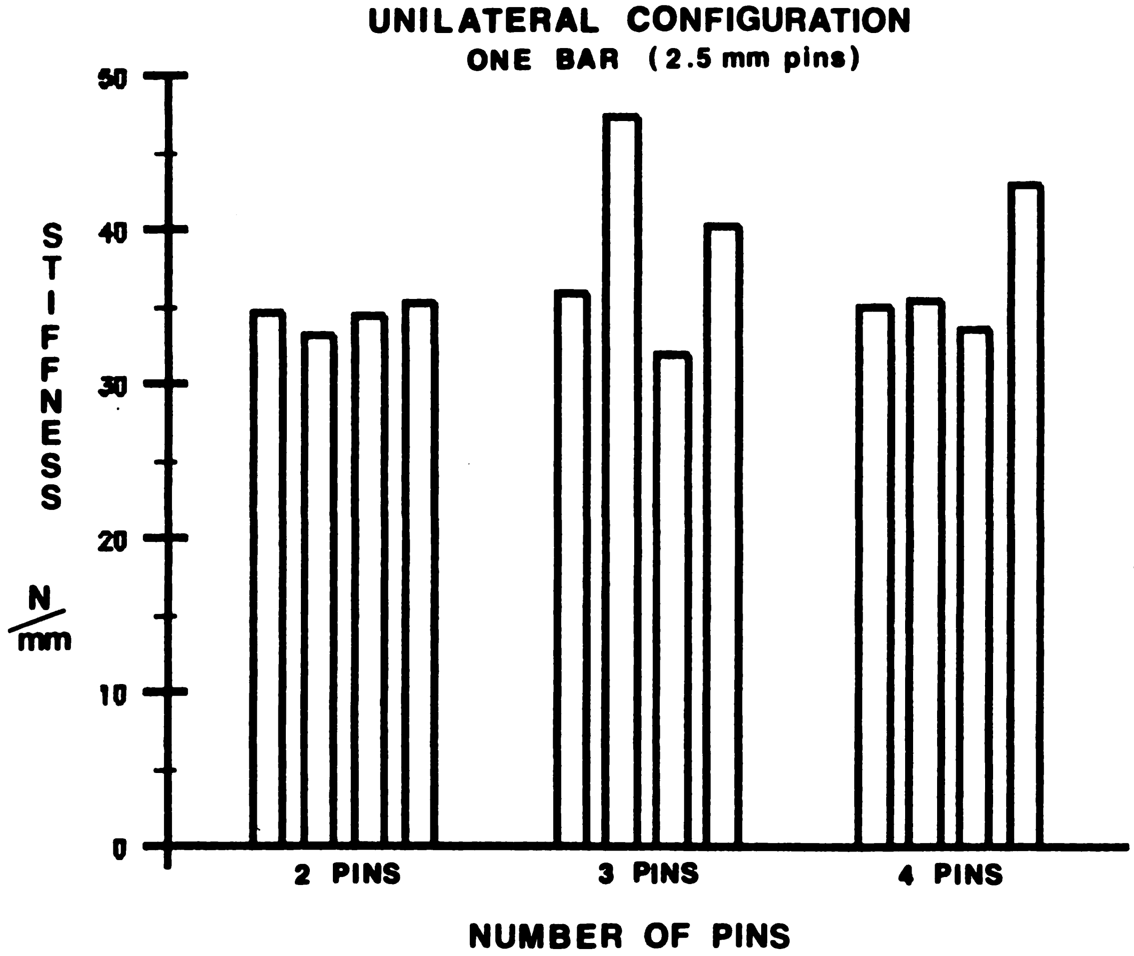

FIGURE 14. Stiffness Response with Varying Pin Number - 2.5 mm Pins

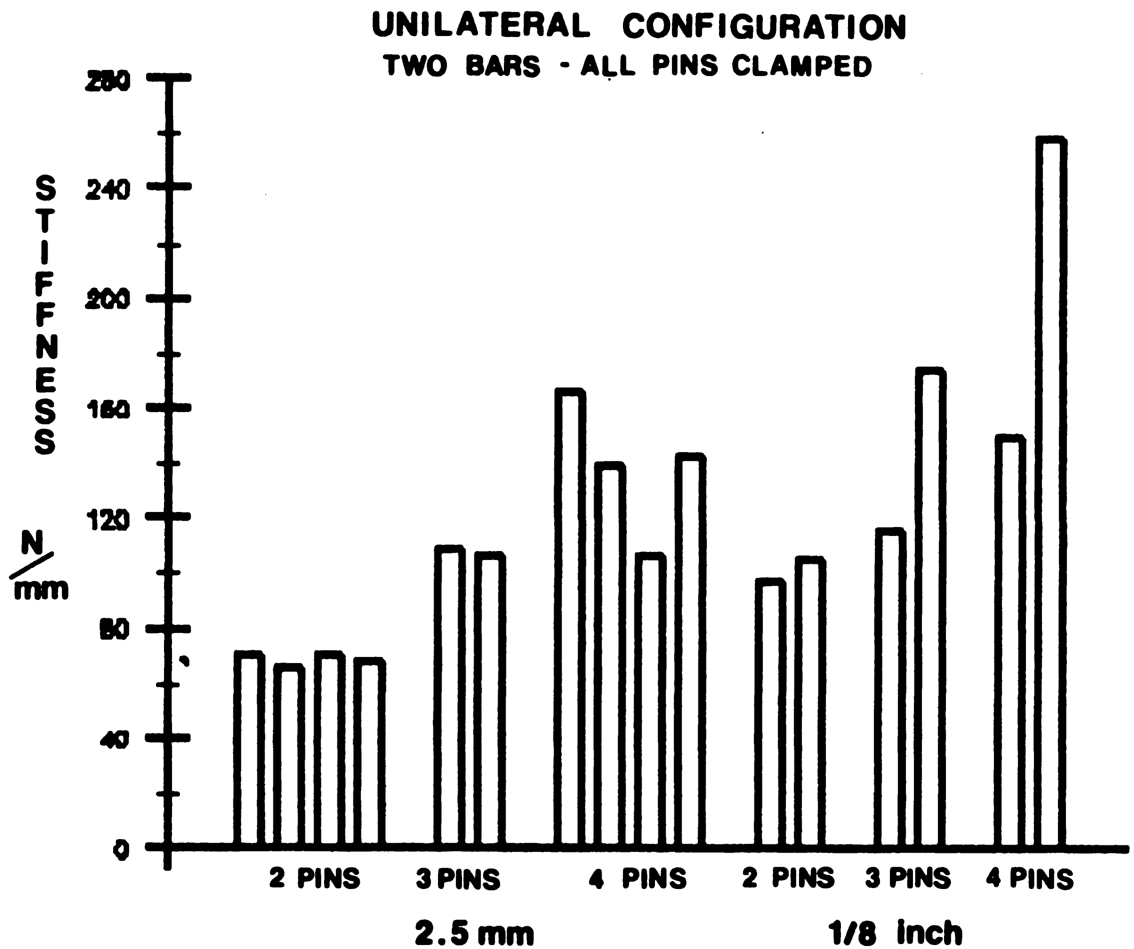

FIGURE 15. Stiffness Response with Varying Pin Number and Pin Size

TABLE 1. Unilateral Data for the 4.0 mm Pins

	One Bar			Two Bars All Pins Clamped			Two Bars Mid-Pin Clamped			Two Planes
Date Tested	4 Pins	3 Pins	2 Pins	4 Pins	3 Pins	2 Pins	4 Pins	3 Pins	2 Pins	2 Pins
5/20										
Load (N)	—	137.22	—	561.78	—	311.36	266.40	—	191.71	237.30
Stiffness (N/mm)	—	84.44	—	244.15	—	104.70	103.70	—	72.65	80.68
6/1										
Load (N)	125.88	121.43	116.10	—	—	208.17	—	—	98.08	—
Stiffness (N/mm)	46.64	47.00	39.30	—	—	124.05	—	—	54.10	—
6/15										
Load (N)	128.10	129.22	118.99	—	—	—	—	—	—	—
Stiffness (N/mm)	44.33	43.04	41.58	—	—	—	—	—	—	—
9/8										
Load (N)	136.33	123.43	125.88	407.88	—	310.92	162.35	171.47	165.91	195.05
Stiffness (N/mm)	46.92	61.77	39.74	179.70	—	127.65	63.84	70.66	56.63	71.41

TABLE 2. Unilateral Data for the 2.5 mm Pins

	One Bar			Two Bars All Pins Clamped			Two Bars Mid-Pin Clamped			Two Planes
Date Tested	4 Pins	3 Pins	2 Pins	4 Pins	3 Pins	2 Pins	4 Pins	3 Pins	2 Pins	2 Pins
5/23										
Load (N)	69.39	66.16	58.94	300.24	—	138.78	88.30	—	51.74	259.10
Stiffness (N/mm)	35.10	35.81	34.49	166.30	—	70.27	37.36	—	31.29	109.85
6/8										
Load (N)	99.86	91.85	86.96	345.61	283.34	203.05	120.99	105.42	93.85	—
Stiffness (N/mm)	35.44	47.39	33.26	139.10	109.25	66.13	39.04	33.04	35.79	—
6/24										
Load (N)	59.83	53.60	49.60	213.95	—	109.64	—	78.95	52.05	—
Stiffness (N/mm)	33.53	31.99	34.45	106.55	—	70.40	—	43.39	29.40	—
9/9										
Load (N)	103.15	99.41	94.97	320.26	234.64	224.18	121.88	115.20	106.31	153.24
Stiffness (N/mm)	40.98	42.25	35.19	142.75	106.00	67.81	39.00	36.25	35.04	46.69

TABLE 3. Unilateral Data for the 1/8 inch (3.175 mm) Pins

	One Bar			Two Bars All Pins Clamped			Two Bars Mid-Pin Clamped			Two Planes
Date Tested	4 Pins	3 Pins	2 Pins	4 Pins	3 Pins	2 Pins	4 Pins	3 Pins	2 Pins	2 Pins
5/25										
Load (N)	125.43	112.09	--	330.49	267.32	210.84	143.67	132.55	115.20	309.58
Stiffness (N/mm)	49.09	43.64	--	149.00	115.70	96.50	55.02	46.81	44.78	126.30
9/12										
Load (N)	187.71	165.47	148.57	569.12	436.93	330.27	214.62	194.83	175.25	171.25
Stiffness (N/mm)	55.21	50.02	45.90	258.60	174.50	105.61	63.29	59.62	49.90	60.40

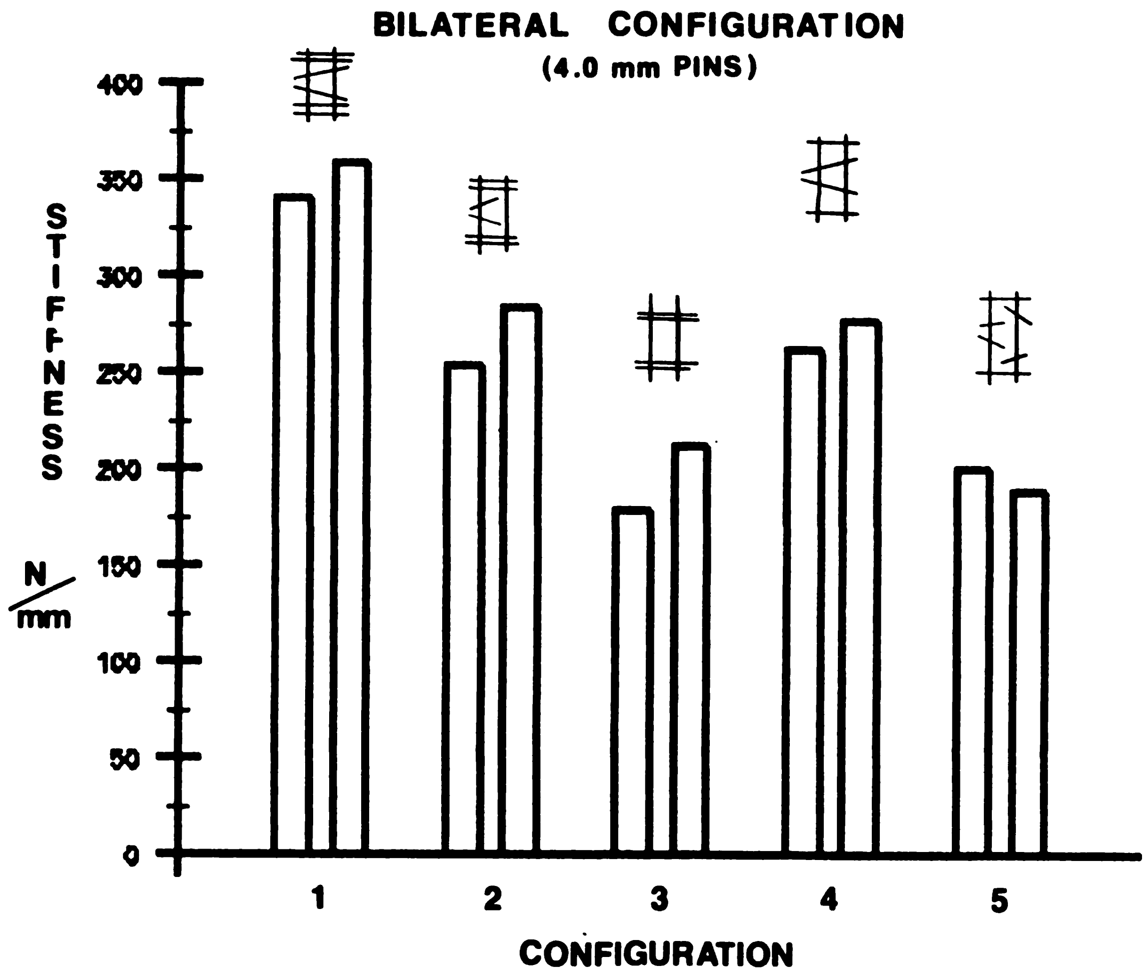

FIGURE 16. Stiffness Response with Varying Bilateral Configuration - 4.0 mm Pins

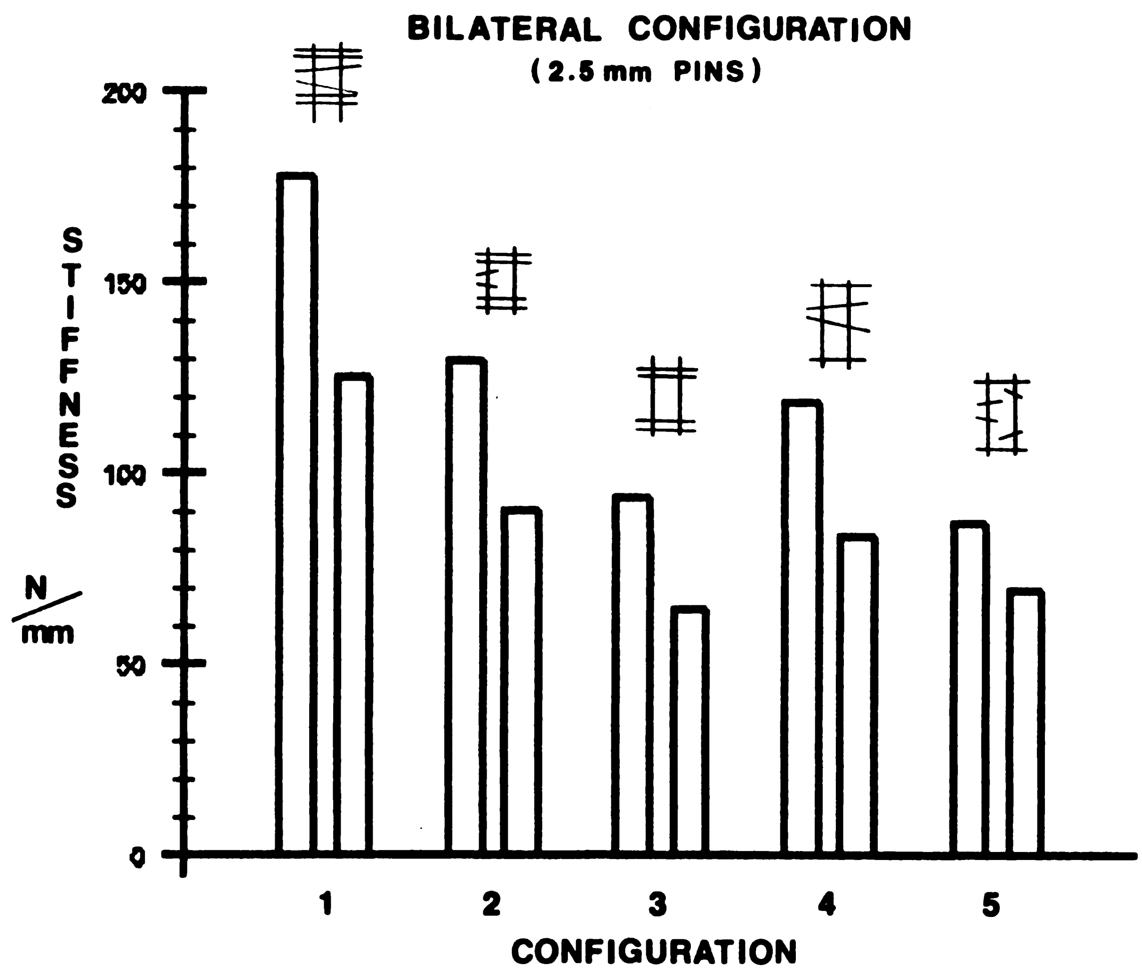

FIGURE 17. Stiffness Response with Varying Bilateral Configuration - 2.5 mm Pins

referring to these figures and Figure 7, it can be seen that as the number of full, or through and through, pins is increased from one to three, the axial stiffness increased 80% for the 4.0 mm pin size and 94% for the 2.5 mm pin size.

The two configurations utilizing only two full pins, i.e., #3 and #4, demonstrate the effect of angling the pins 72.5^{o} to the axis of the bone. The design composed of one pin connected perpendicular to the bone and one pin connected 72.5^{o} to the axis of the bone shows a higher axial stiffness than the design with both pins connected perpendicular to the bone axis. The configuration using 4.0 mm pins showed this increase to be 37% while a 28% increase was obtained for the 2.5 mm pin size. The data used for analysis of the bilateral configurations is given in Tables 4, 5, and 6 for the 4.0 mm, 2.5 mm, and 1/8 inch (3.175 mm) pins, respectively. Again, the maximum load and maximum stiffness values do not occur simultaneously due to the nonlinear response of the system.

There are numerous fixatuers, both internal and external, used in the treatment of long bone fractures. To assist in comparison with previous works, a single Stader splint, composed of two pins clamped together and attached to a single connecting bar, was tested in the same manner as described before and achieved a maximum stiffness of 110.0 N/mm. Several trials were also run on two sizes of bone plates, a 3.5 mm broad and 4.5 mm narrow. The data for these tests is presented on Table 7.

TABLE 4. Bilateral Data for the 4.0 mm Pins

Date Tested	Configuration Number				
	1	2	3	4	5
9/17					
Load (N)	487.80	425.52	284.34	446.43	432.05
Stiffness (N/mm)	340.27	253.97	179.20	262.27	199.93
10/2					
Load (N)	465.04	408.55	301.69	436.57	369.93
Stiffness (N/mm)	358.90	282.88	211.30	276.48	188.70

TABLE 5. Bilateral Data for the 2.5 mm Pins

Date Tested	Configuration Number				
	1	2	3	4	5
9/14					
Load (N)	368.00	337.60	164.13	313.58	325.15
Stiffness (N/mm)	177.37	129.57	93.64	118.40	86.71
9/30					
Load (N)	228.36	232.41	133.55	237.38	230.63
Stiffness (N/mm)	125.02	90.34	64.40	83.59	69.44

TABLE 6. Bilateral Data for the 1/8 inch (3.175 mm) Pins

Date Tested	Configuration Number				
	1	2	3	4	5
9/28					
Load (N)	535.24	633.40	267.10	476.53	—
Stiffness (N/mm)	293.70	329.15	243.55	246.27	—

TABLE 7. Bone Plate Data

Thickness (mm)	Maximum Load (N)	Stiffness (N/mm)
4.5	478.61	537.00
4.5	765.28	551.00
3.5	1146.70	914.35

A general summary of the data collected in this study is given in Figure 18. This figure compares the average stiffness values for all configurations utilizing two 4.0 mm pins in each bone segment, as well as average results from the Stader splint and the bone plates. The stiffness of the bone plates were substantially higher than any of the external fixateurs.

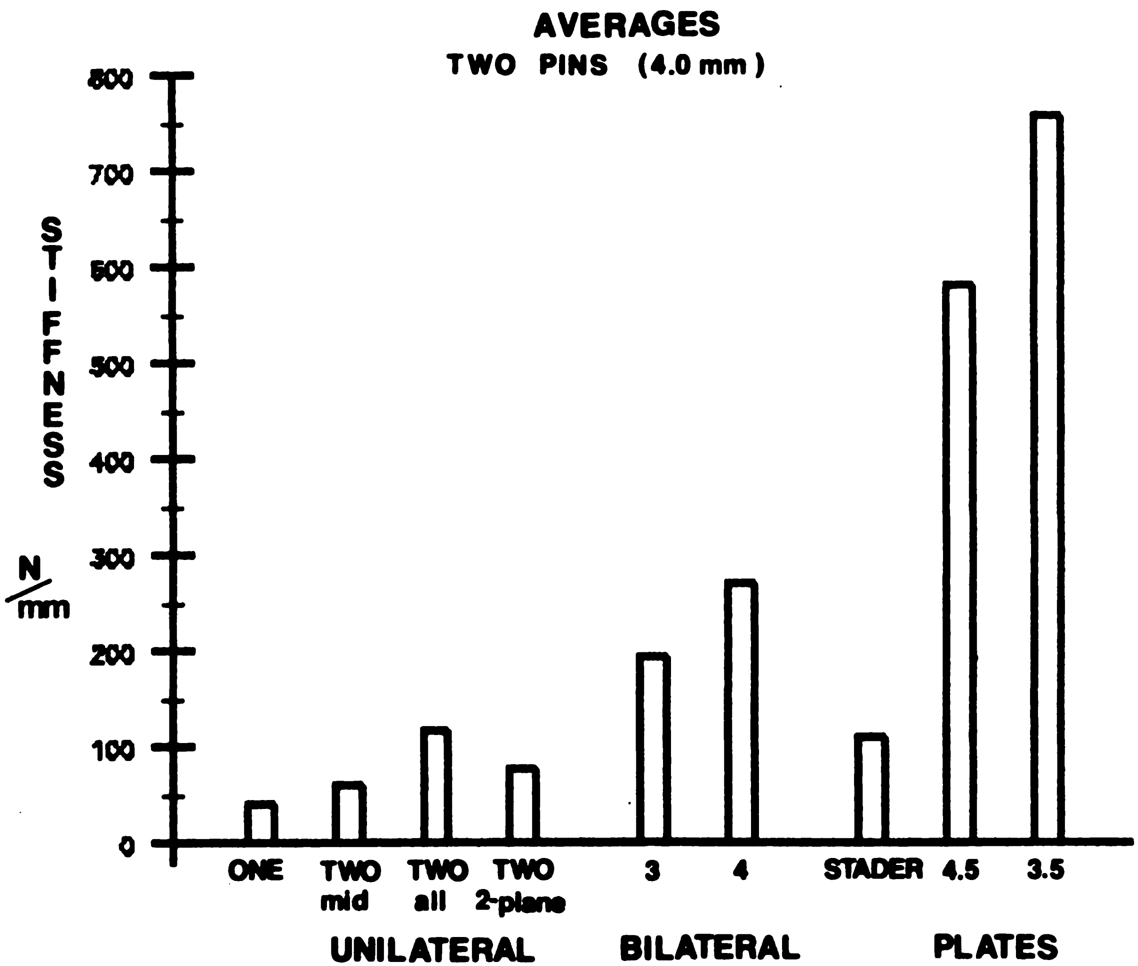

FIGURE 18. Summary of Stiffness Response with Varying Configurations

V. ANALYTICAL METHODS AND RESULTS

Though experimental testing supplies necessary information on both the behavior of the fixateur and the reaction of the bone during axial compression, it suffers from a number of disadvantages. The actual testing procedure is quite time consuming due to the number of variables, both geometric and nongeometric, that can be altered in numerous combinations. Also, this experimentation only provides stiffness and deflection data for analyzing the fixateur performance. The procedure yields no data on the internal stresses each component is subjected to. For this reason, theoretical models are developed, often utilizing computer simulation and finite element methods.

Theoretical analysis for the previously described fixateur configurations was performed by utilizing the finite element program ANSYS (2). This computer program allows for examination of several classes of engineering problems with static, dynamic, elastic, plastic or creep analysis. The ANSYS program uses the wave-front direct solution approach for the system of linear equations developed by the matrix displacement method. Results are achieved with both high accuracy and a minimum of computer time.

For the problem at hand, a static analysis was utilized to solve for the displacements and forces attained in the fixateur during axial compression. All joints were considered rigid since the pins were clamped tightly to the connecting bar and firmly inserted into the bone by means of the hand chuck, the bone was cemented into the potting material and the pot set securely into the grips. For the unilateral configurations, a two-dimensional frame structure was developed, as shown in Figure 19, since all elements remain in the x-y plane throughout the testing procedure. A total of 30 nodes and 42 elements made up the configuration utilizing four pins and two connecting bars, each pin doubly clamped. To examine the other geometric designs, elements were simply dropped from the model. Two-dimensional elastic beam elements were used for each component of the system. This element allows for tension, compression, and bending loadings. It has three degrees of freedom at each node: translations in the x and y direction and rotation about the z-axis. The component properties for the unilateral configurations are listed in Table 8 along with the element numbers associated with each.

A three-dimensional frame structure was developed for the bilateral configurations, since motion was observed in all three planes during testing. The model created is shown in Figure 20. The initial bilateral model consisted of a total of 24 nodes and 32 elements representing three full transfixing pins (configuration #1). Again, elements were simply removed to analyze the various other geometric designs. Three-dimensional elastic beam elements were used to model the structural components. This type of element allows for six degrees of freedom: translations in the x, y, and z directions, as well as rotations about all three axes. The component properties for the bilateral configurations and the element numbers associated with each are given in Table 9.

For both the two dimensional and three dimensional analyses, each element is defined by the location of its two endpoints, or nodes.

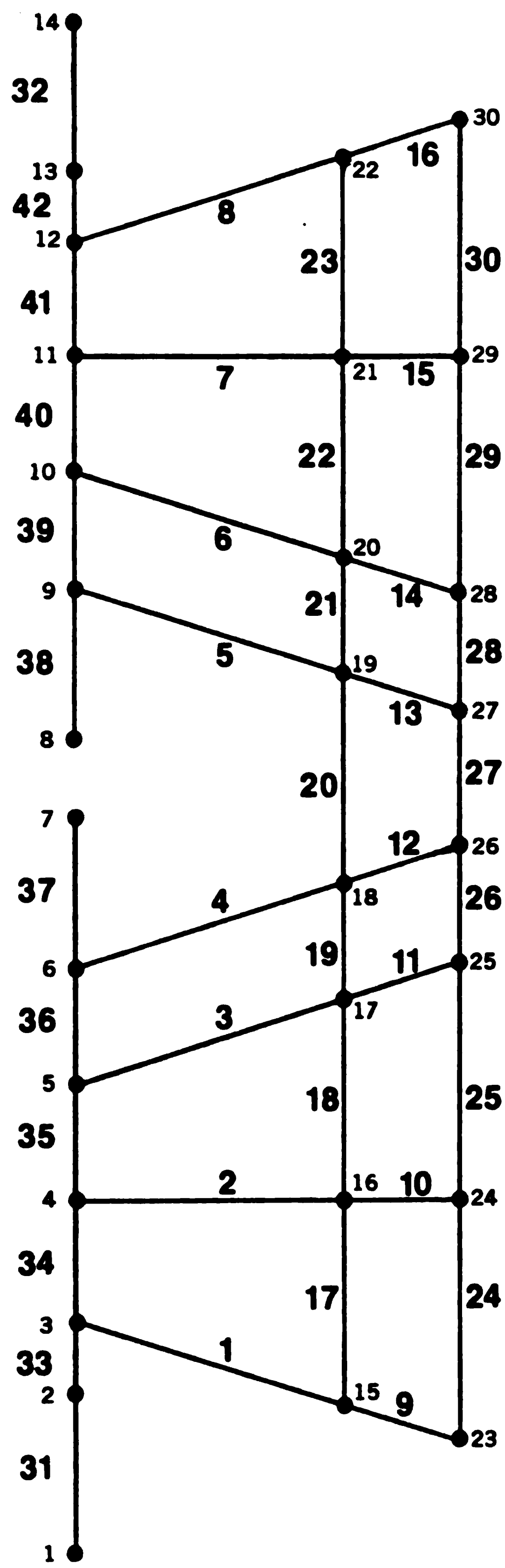

FIGURE 19. Node and Element Numbering for Unilateral Configurations

TABLE 8. Parameters Used in Two-Dinensional Finite Element Model

Component	Element	Area (mm^2)	Izz (mm^4)	Diameter (mm)	E ($x10^5 N/mm^2$)	DENS ($x10^{-6} kg/mm^3$)
4mm Pins	1-16	12.57	12.57	4.0	2.0	7.89
2.5 Pins	1-16	4.9	1.92	2.5	2.0	7.89
1/8 inch Pins	1-16	7.92	4.99	3.175	2.0	7.89
4mm Bar	17-30	12.57	12.57	4.0	2.0	7.89
3/16 inch Bar	17-30	17.35	23.95	4.76	2.0	7.89
Bone	33-42	varies	varies	varies	0.2	2.0
Grip	31-32	same as bone	same as bone	same as bone	2.0	7.89

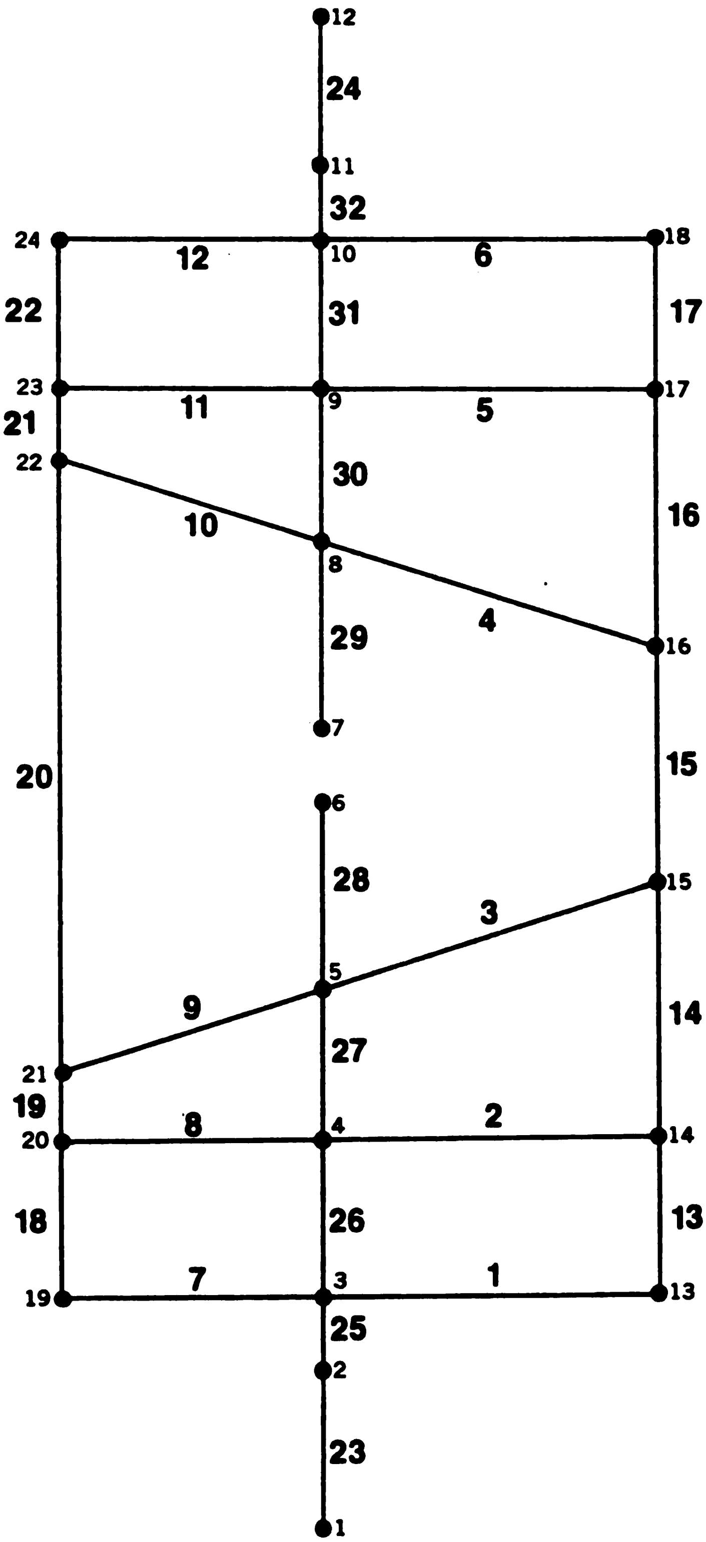

FIGURE 20. Node and Element Numbering for Bilateral Configurations

TABLE 9. Parameters Used in Three-Dinensional Finite Element Model

Component	Element	Area (mm^2)	Iz,Iy (mm^4)	Diameter (mm)	E ($x10^5 N/mm^2$)	DENS ($x10^{-6} kg/mm^3$)
4mm Pins	1-12	12.57	12.57	4.0	2.0	7.89
2.5 Pins	1-12	4.9	1.92	2.5	2.0	7.89
1/8 inch Pins	1-12	7.92	4.99	3.175	2.0	7.89
4mm Bar	13-22	12.57	12.57	4.0	2.0	7.89
3/16 inch Bar	13-22	17.35	23.95	4.76	2.0	7.89
Bone	25-32	varies	varies	varies	0.2	2.0
Grip	23-24	same as bone	same as bone	same as bone	2.0	7.89

These geometric parameters were obtained from the photographs taken during the actual experimental testing. The photos were developed into 8 1/2" x 11" pictures. A reference frame was generated by placing the y-axis at the midline of the two grips and the x-axis then placed perpendicular to this line at its midpoint. The cross section of bone was assumed circular and located symmetrically about the y-axis. Nodal points were chosen along the bone at the intersection of the transfixing pins and the vertical axis, along the connecting bar at the points where the pins are clamped to it, at the junctions between the bone and the grip, and at the points where the grip articulates with the test machine. The pictoral values were then converted to actual measurements by means of a conversion factor calculated through the measurement of a known distance. Constraints were placed on the system by preventing any translation of the nodal points at the junction between the grips and the test machine. The 'loading' was achieved by displacing nodal point #14, in the unilateral configuration, or #12, in the bilateral configuration, a distance of 0.2 inch (5.08 mm) in the negative y-direction.

The postprocessing capabilities for the ANSYS program include sorting, printing, and plotting of results from any analysis. Reaction forces at the constrained nodes, nodal displacements and component nodal stresses may be presented as tabulated values or as geometrical plots. For this analysis, the reaction forces at the constrained nodes are of primary interest. It was found that the maximum load exhibited by the fixateur varied linearly with the applied deflection. Therefore, the overall axial stiffness of each design can be calculated by dividing the reaction force by the total axial deflection (5.08 mm). Table 10 lists

TABLE 10. Unilateral Data from the Analytical Model

Bar Number	Pin Number	Maximum Load (N)	Axial Stiffness (N/mm)
1/8 inch			
Two (all pins)	4	1204.0	237.01
	3	920.8	181.26
	2	588.2	115.79
Two (mid-pin)	4	339.2	66.77
	3	301.7	59.39
	2	242.8	47.80
One	4	235.9	46.44
	3	214.5	42.22
	2	178.4	35.12
2.5 mm			
Two (all pins)	4	986.9	194.27
	3	753.6	148.35
	2	486.3	95.73
Two (mid-pin)	4	244.4	48.11
	3	220.6	43.43
	2	175.7	34.59
One	4	182.5	35.93
	3	168.4	33.15
	2	142.4	28.03
4.0 mm			
One	4	217.8	42.87
	3	202.4	39.84
	2	191.7	37.74

the maximum loads and axial stiffnesses exhibited by a representative group of unilateral fixateurs. As can be seen for all three pin sizes, when a single connecting bar is used, increasing the number of pins in each bone segment has little effect on the axial stiffness of the system. For the configurations utilizing two connecting bars, if all pins are doubly clamped, increasing the number of pins in each segment markedly increase the rigidity of the device. For both the 2.5 mm pin size and the 1/8 inch (3.175 mm) pin size, this increase is 104% as pin number is incremented from two to four.

An increase in fixateur stiffness is also shown when the pin number is held fixed and a second bar is added to the system. If only the mid-pin is clamped to this second connecting bar, this increase is minimal and therefore not beneficial. Yet, if all pins are clamped to both connecting bars, the increase in stiffness is marked. For both the 2.5 mm pin size and the 1/8 inch (3.175 mm) pin size, this increase in the rigidity of the unilateral design is 236%.

Table 11 lists the maximum loads and stiffnesses for several analytical cases utilizing the bilateral configurations. A marked increase in axial stiffness can easily be seen when the pin diameter changes from 2.5 mm to 4.0 mm. This increase ranges from 306% for configuration #1 to 406% for configuration #3. Also, as the number of full pins is increased from two to three, the associated axial stiffness increases. The 2.5 mm pin size shows increase of approximately 111%, while an increase of approximately 81% can be seen for the 4.0 mm pin size. When only two full pins are used, the configuration with one pin angled at 72.5° to the axis of the bone (#4) shows a slightly higher stiffness than the design with both pins set perpendicular to the bone.

TABLE 11. Bilateral Data from the Analytical Model

Configuration	Maximum Load (N)	Axial Stiffness (N/mm)
2.5 mm		
1	1527.0	300.59
2	852.9	167.89
3	606.5	119.39
4	688.5	135.53
1	1456.0	286.61
2	783.7	154.27
3	611.8	120.43
4	703.7	138.52
4.0 mm		
1	6321.0	1244.29
2	3634.0	715.35
3	3084.0	607.09
4	3274.0	644.49
1	5786.0	1138.98
2	3755.0	739.17
3	3086.0	607.48
4	3275.0	644.69

This increase is only 12% for the 2.5 mm pins and 6% for the 4.0 mm pins. Therefore, if two through and through pins must be used, it would be advantagous to use a design such as configuration #4 for increased rigidity.

VI. CORRELATION OF RESULTS

The development of a theoretical model, based on the concepts of structural analysis, endeavors to predict results duplicating those observed during actual experimentation. This type of model allows geometric and material properties to be altered without the time and expense requirements necessary for an experimental analysis. In this study, an attempt was made to simulate the reactions of unilateral and bilateral external fixateur frames subjected to axial loading. Two-dimensional and three-dimensional finite element models were utilized to observe the resultant force developed in the structures due to the applied displacement.

In general, the trends exhibited during the experimental testing were reinforced by the results obtained from the theoretical models. As the number of pins and/or the number of bars are increased, in the unilateral and bilateral configurations alike, the axial stiffness of the external fixateur device increases. The bilateral configurations showed a consistantly higher maximum stiffness than the unilateral configurations for both the finite element model and the experimental trials. Geometric variations, therefore, yielded similar results in both analyses.

The static analysis used in the analytical model to determine the resultant force, assumed that the fixateur's response to any applied loading condition would be linear. This assumption was verified by imposing displacements, varying from -5.08 mm to +5.08 mm, on several configurations. The reaction force at node #1, constrained in both the x and y directions, is shown plotted versus the applied deflection in Figure 21 for a single unilateral frame utilizing two pins, doubly clamped to two connecting bars. This graph demonstrates that the load varies linearly with the applied displacement, regardless of whether the system is in tension or compression.

As mentioned previously, this linear response permits the maximum stiffness to be calculated by dividing the maximum load by the peak deflection. Tables 12 and 13 compare the stiffness values calculated from the theoretical model with the maximum stiffnesses obtained from identical experimental tests for several unilateral and bilateral configurations, respectively. This representative group of values demonstrates the variability between the two methods of analyses. For the unilateral configurations, the error between the theoretical and experimental stiffness values range from 0.4% to 41.2%. A greater range of error can be seen in the bilateral data. The minimum difference observed between the theoretical and experimental values was 14.5%, while the maximum error shown was 265.7%.

It should be noted here that the analytical models developed for analysis of the external fixateurs are only first approximations. The decision to attempt an analytical analysis was made long after the experimental testing had begun. The theoretical models assume the bone segments to be circular solid beams. Yet, in vivo, the bone segment is actually a tube of cortical bone surrounding the medullary cavity. The pins are inserted through a layer of compact bone, through the marrow cavity, essentially hollow, and then pierce the other layer of compact bone. Since the effective pin length used in the analytical model, from

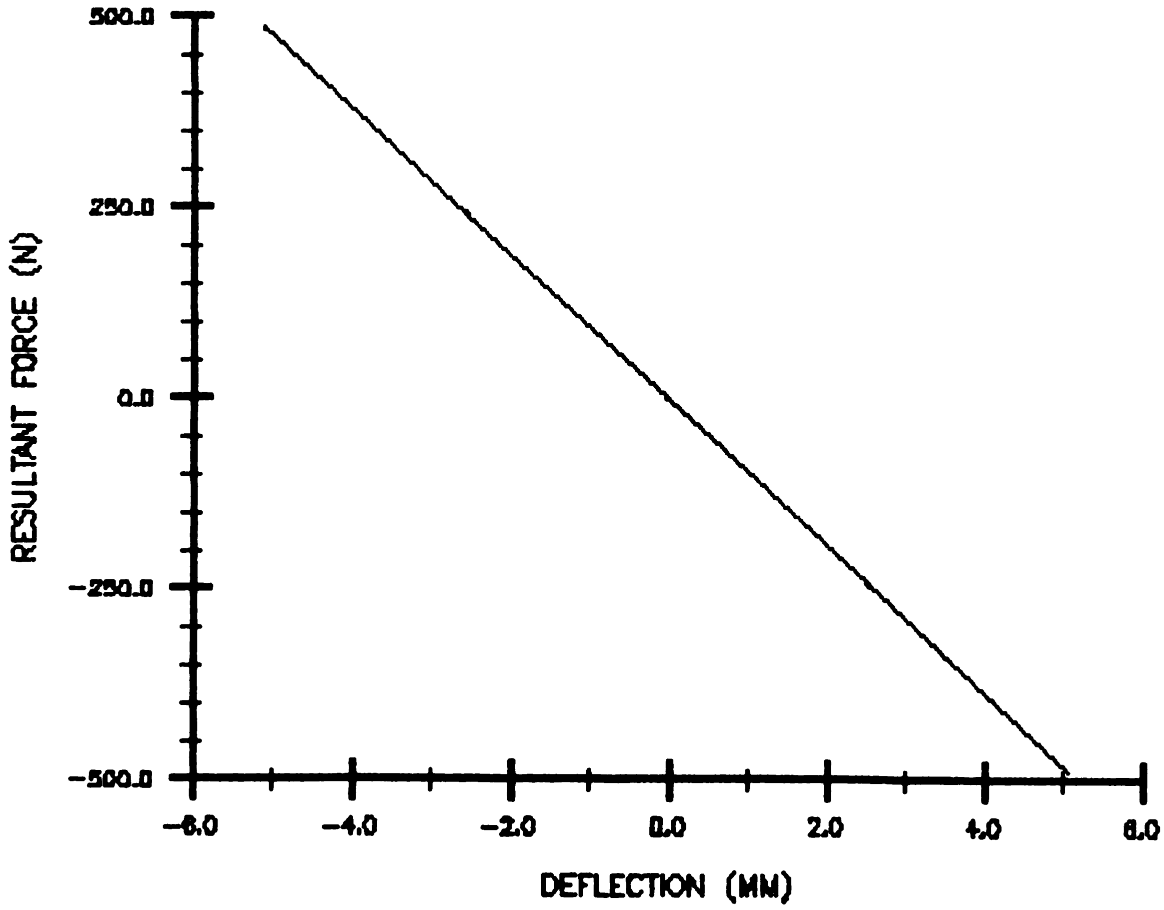

FIGURE 21. Linear Response of Analytical Model

TABLE 12. Unilateral Stiffness Data
Analytical vs. Experimental

Bar Number	Pin Number	Analytical Stiffness (N/mm)	Experimental Stiffness (N/mm)
1/8 inch			
Two (all pins)	4	237.01	258.60
	3	181.26	174.50
	2	115.79	105.61
Two (mid-pin)	4	66.77	63.29
	3	59.39	59.62
	2	47.79	49.90
One	4	46.44	55.21
	3	42.22	50.02
	2	35.12	45.90
2.5 mm			
Two (all pins)	4	194.27	142.75
	3	148.35	106.00
	2	95.73	67.81
Two (mid-pin)	4	48.11	39.04
	3	43.43	33.04
	2	34.59	35.79
One	4	35.92	42.98
	3	33.14	40.25
	2	28.03	35.19
4.0 mm			
One	4	42.87	44.33
	3	39.84	43.04
	2	37.74	41.08

TABLE 13. Bilateral Stiffness Data
Analytical vs. Experimental

Configuration	Analytical Stiffness (N/mm)	Experimental Stiffness (N/mm)
2.5 mm		
1	300.59	177.37
2	167.89	129.57
3	119.39	95.29
4	135.53	118.40
1	286.61	125.02
2	154.27	90.34
3	120.43	64.40
4	138.43	83.59
4.0 mm		
1	1244.29	340.27
2	715.35	253.97
3	607.09	179.20
4	644.49	262.27
1	1138.98	358.90
2	739.17	282.88
3	607.48	211.30
4	644.69	276.48

the connecting bar to the midpoint of the shaft of the bone, is not an exact representation of this actual condition, the axial and bending forces acting at the pin-bone interface are only approximations. Further investigations into a three-dimensional, solid element model of the bone is anticipated, but is beyond the scope of the current study.

It was discovered during the testing that as the geometry of the system became more complex, i.e., the number of pins or the number of bars is increased, the load-deformation curves became more nonlinear. This is demonstrated in Figure 22 and 23 for the unilateral configurations. Figure 22 shows the load-deformation curves, from both experimental and theoretical analyses, for the frame design utilizing four 2.5 mm pins clamped to a single connecting bar. Figure 23 shows similar curves for a configuration utilizing the same number of pins, but doubly clamped to two connecting bars. It can easily be seen that the second curve deviates from the theoretical line to a greater extent than the first curve. The same type of deviation can be seen in the bilateral configurations. Figure 24 shows the load-deformation response of configuration #4, utilizing two 2.5 mm through and through pins, for both experimental and theoretical analyses. Figure 25 demonstrates a similar response for configuration #1, utilizing three 2.5 mm through and through pins. Again it can be seen that as the geometry of the system becomes more complex, the experimental curve deviates from the linear response to a greater extent.

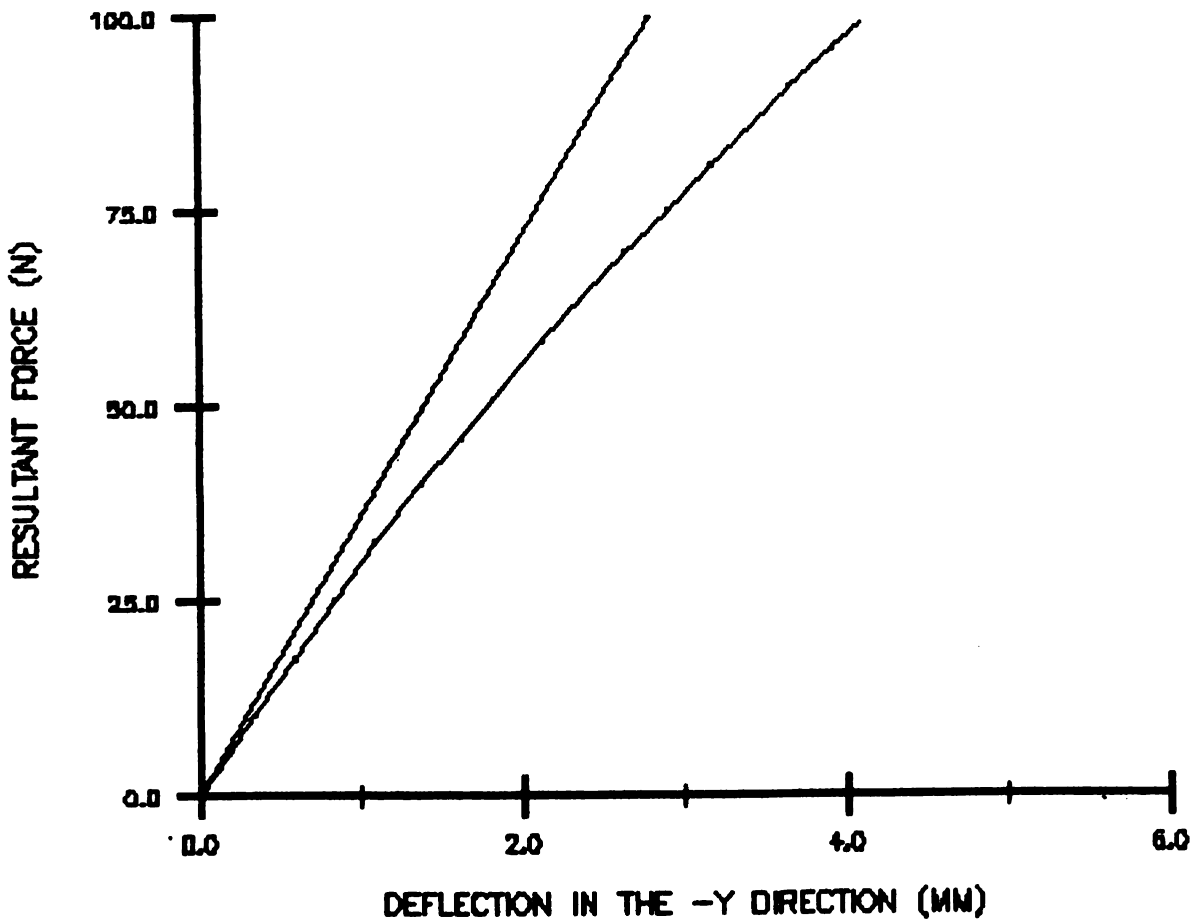

FIGURE 22. Analytical vs. Experimental Response - One Bar

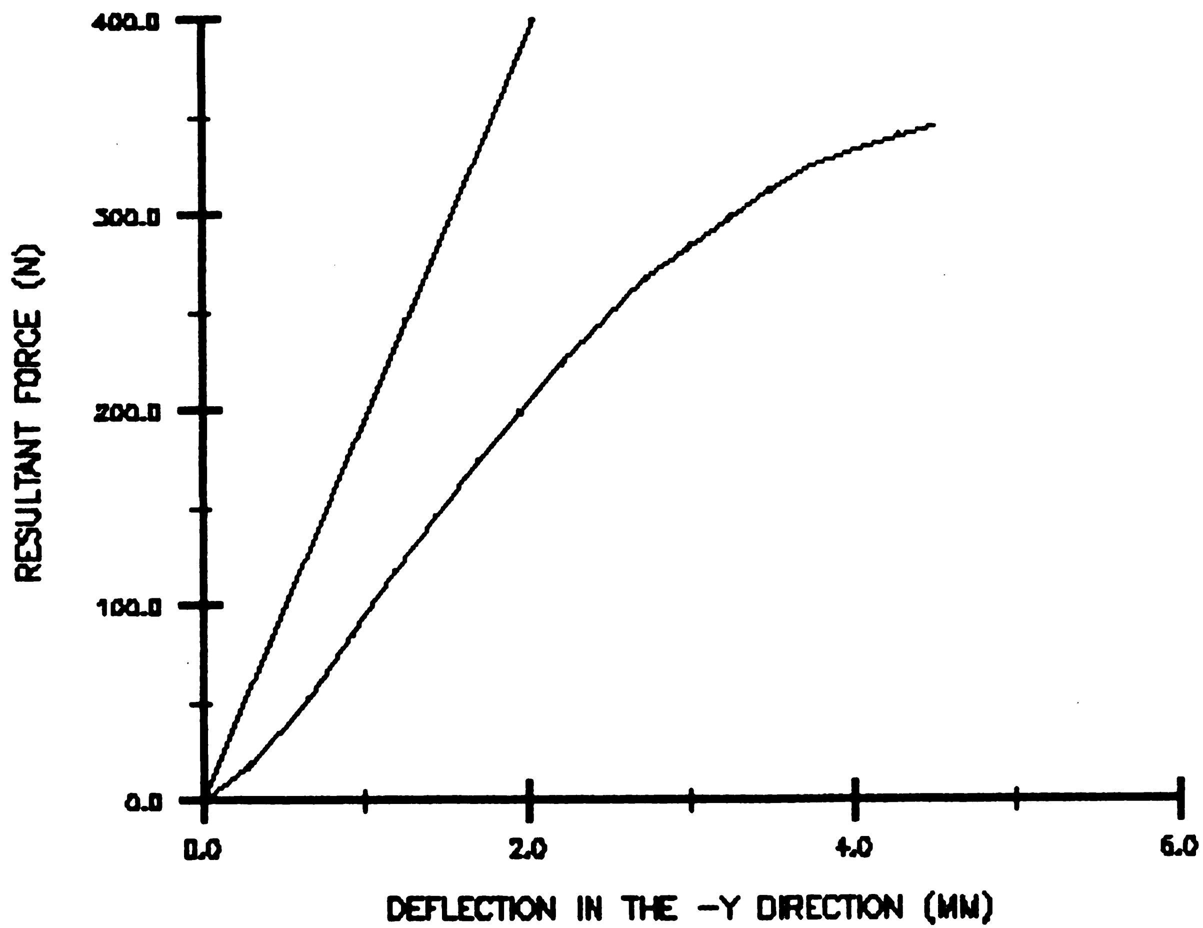

FIGURE 23. Analytical vs. Experimental Response - Two Bars

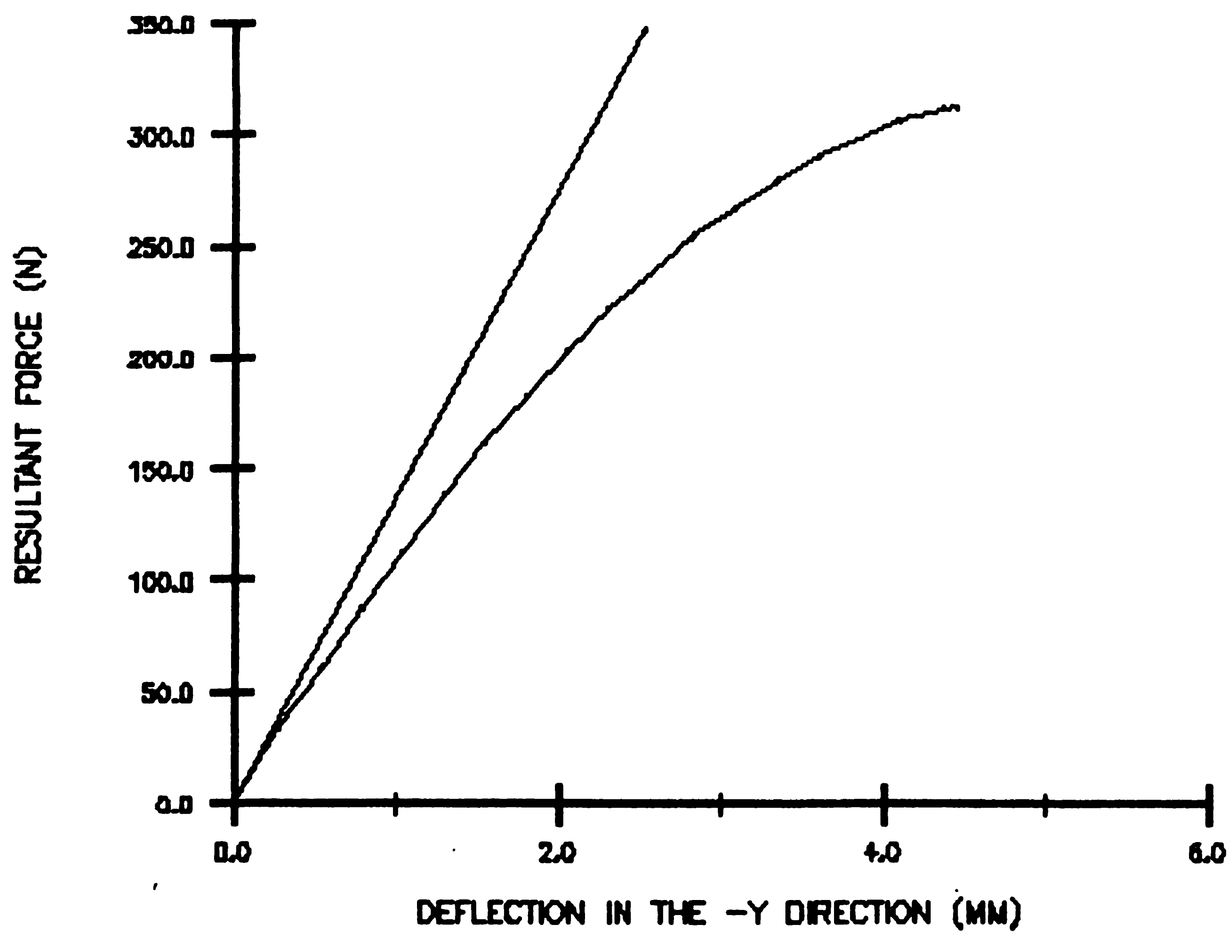

FIGURE 24. Analytical vs. Experimental Response - Two Pins

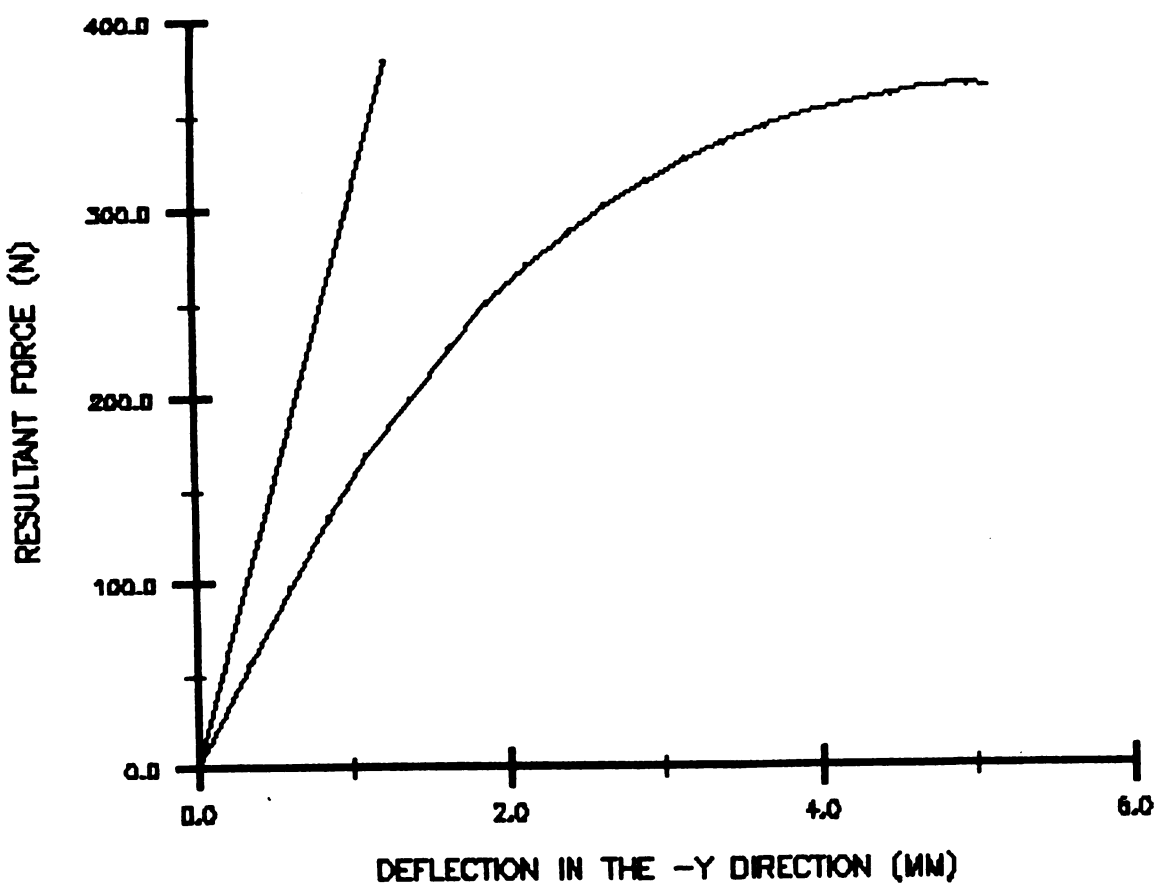

FIGURE 25. Analytical vs. Experimental Response - Three Pins

VII. CONCLUSION

As stated frequently in the literature, variation in the geometric parameters of the external fixation device has a great effect on the frame's stiffness response to any applied loading. The changes in axial stiffness caused by varying the configuration of the fixateurs examined in this study closely follow the trends reported on by all previous authors. The increase in pin size showed a definite effect on the rigidity of the fixation device, for both the unilateral and bilateral frames. An increase of as much as 187% was observed when the pin diameter was increased from 2.5 mm to 4.0 mm.

For the unilateral designs, the relation between the pin number and the stiffness of the system is highly dependent on the number of connecting bars utilized. If only a single connecting bar is to be used, increasing the number of pins perforating the bone does little to affect the rigidity of the frame. A higher stiffness response can be obtained, in the unilateral designs, without any increased soft tissue involvement, by the addition of a second connecting bar, attached parallel to the first. When this additional connecting bar is clamped solely to the mid-pin, the increase in stiffness is again minimal and therefore not profitable. However, if this second bar is securely clamped to every pin transfixing the bone, the stiffness of the fixateur increases greatly. This arrangement was shown to increase the rigidity of the system by as much as 195% for the two pin arrangement. If an even higher axial stiffness is required for fracture healing and a unilateral device is to be used, increasing the number of pins doubly clamped to both connecting bars effectively increases the rigidity of the system. Three pins raise the stiffness 44% and four pins increase the stiffness approximately 102%. This advantage of increased rigidity must be weighed against the clinical shortcomings created by the additional invasion of soft tissues, to arrive at the best fixateur frame configuration to suit the needs of the specific fracture type.

The bilateral configurations showed a consistantly higher axial stiffness than the unilateral configurations tested. A maximum increase of 325% could be seen in the stiffness response of the bilateral designs when compared to unilateral frames utilizing the same number of pins. Within the bilateral category, it was shown that by increasing the total number of full, or through and though pins, the axial stiffness of the fixateur can also be effectively increased. If two full transfixing pins are to be solely utilized in the bilateral frame, the axial stiffness of the system can be increased approximately 33% by inserting the mid-pin at an angle of 72.5° to the axis of the bone. Again the increased stiffness factors must be compared to the possible damage created by inserting the full fixation pins through the tissues and vessels not previously invaded by the unilateral designs.

The ulitization of an analytical model can effectively decrease the time and expense necessary for an experimental analysis. The finite element model used in this analysis accurately duplicated the trends observed from the experimental testing. Varying the geometric parameters in the computer model produced similar variations in stiffness as developed during experimentation. For both analyses, axial stiffness could be increased by increasing bar number or by increasing pin number, if two bars are used and all pins are doubly clamped, in the unilateral configurations, or by using one of the bilateral configurations.

The ANSYS static analysis utilized in this study was shown to produce a linear response approximating the linear portion of the load/deformation curve observed during the experimental testing. The stiffness response developed by the static analysis more closely represented the response of the elementary fixateur designs. This was due to the fact that the experimental load/deformation response became nonlinear as the system's geometry became more complex, i.e., the number of pins and/or bars are increased.

In conclusion, it was determined that the analytical model developed in this study predicts the change in stiffness associated with a specific modification of a fixateur's geometry. Therefore, any alteration in the configuration of an external fixation device can be analyzed, quickly and with confidence, without the drawbacks of an experimental analysis. Although the nonlinearity of the fixateurs' stiffness responses have yet to analyzed, it is the subject of future studies to develop a predictive model to incorporate these properties.

BIBLIOGRAPHY

BIBLIOGRAPHY

1. Anderson, R., "An Ambulatory Method of Treating Fractures of the Shaft of the Femur," Surg. Gynecol. Ob., Vol. 62, pp. 865-873, 1936.

2. ANSYS-Engineering Analysis System User's Manual, Swanson Analysis Systems, Inc., Houston, PA, 1979.

3. Behrens, F., and K. Searls, "Unilateral External Fixation Experience with the ASIF 'Tubular' Frame," Current Concepts of External Fixation of Fractures, Ed. by H. Uhthoff, pp. 177-183, 1982.

4. Briggs, B.T., and E.Y. Chao, "The Mechanical Performance of the Standard Hoffmann-Vidal External Fixation Apparatus", Journal of Bone and Joint Surgery, Vol. 64-A, No. 4, pp. 566-573, 1982.

5. Brinker, W.O., and G.L. Flo, "Principles and Application of External Skeletal Fixation," Veternary Clinics of North America, Vol. 5, No. 2, pp. 197-207, 1975.

6. Burny, F., R. Bourgois, and M. Donkerwolkcke, "Elastic External Fixation: A Biomechanical Study of the Half-Frame," Concepts in External Fixation, Ed. by D. Seligson and M. Pope, pp. 67-78, 1982.

7. Chao, E.Y., and K.N. An, "Biomechanical Analysis of External Fixation Devices for the Treatment of Open Bone Fractures," Finite Elements in Biomechanics, Ed. by R.H. Gallagher, B.R. Simon, and J.F. Gross, pp. 195-222, 1982.

8. Chao, E.Y, B.T. Briggs, and M.T. McCoy, "Theoretical and Experimental Analysis of the Hoffman-Vidal External Fixation System," External Fixation - The Current State of the Art, Ed. by A.F. Brooker and C.C. Edwards, pp. 345-370, 1979.

9. Chao, E.Y., and M.H. Pope, "The Mechanical Basis of External Fixation," Concepts in External Fixation, Ed. by D. Seligson and M.H. Pope, pp. 13-39, 1981.

10. Egger, E.L., "Static Strength Evaluation of Six External Skeletal Fixation Configurations," Veternary Surgery, Vol. 12, No. 3, pp. 130-136, 1983.

11. Evans, M., J. Kenwright, and K. Tanner, "Analysis of Single-Sided External Fracture Fixation," Engineering in Medicine, Vol. 8, No. 3, pp. 133-137, 1979.

12. Hierholzer, G., and A. Chernowitz, "External Fixation, Tubular ASIF Set," Current Concepts of External Fixation of Fractures, Ed. by H. Uhthoff, pp. 75-82, 1982.

13. Lambotte, A., "Chirurgie Operatoire des Fractures," Paris, Masson et Cie, 1913.

14. Parkhill, C., "A New Apparatus for the Fixation of Bones After Resection and in Fractures with a Tendency to Displacement," Transactions of the American Surgical Association, Vol. 15, pp. 251-256, 1897.

15. Parkhill, C., "Further Observations Regarding the Use of the Bone Clamp in Ununited Fractures, Fractures with Malunion, and Recent Fractures with a Tendency to Displacement," Ann. Surg., Vol. 27, pp. 553-570, 1898.

16. Pope, M.H., and M. Evans, "Design Considerations in External Fixation," Concepts in External Fixation, Ed. by D. Seligson and M. Pope, pp. 109-135, 1982.

17. Stader, O., "A Preliminary Announcement of a New Method of Treating Fractures," North American Veternarian, Vol. 18, pp. 37-38, 1937.