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A New Orthotic Concept in the Non-Operative Treatment of Idiopathic Scoliosis

A Preliminary Report[1]

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The dynamic orthosis for scoliosis presented here is a departure, in both rationale and design, from current orthotic practice. The rationale for the new system may best be explained by discussing some of the biomechanical considerations from which the rationale was evolved.

From the point of view of basic mechanics, the use of the term "dynamics" when describing scoliosis orthoses can lead to confusion. Newton's first law states "if a body exerts a force on a second body, the second body must exert an equal but opposite force on the first." Since orthoses in current use are static entities, they are limited to returning *precisely* the same magnitude of force acting upon them at any given instant.

When a wearer's trunk is upright, the weight of his or her trunk (i.e., the force of gravity) acts upon the orthosis and the orthosis can only *react* with an "equal and opposite force."

What current scoliosis orthoses do, with varying degrees of effectiveness, (depending upon factors such as type of curvature, severity, age, etc.) is to *hold* a 'prepositioned' realignment from that which the patient initially presented. However, such realignment is restricted by elastic limits, i.e., the range of motion that is present at any particular juncture throughout the length of the spinal column. Once such an orthosis is applied properly and the wearer's trunk is upright, the force of gravity still dictates the course of events, since the orthosis is limited to reacting with equal and opposite force.

Whereas it is impossible for static orthoses to generate "unbalancing" forces, it is possible to make them *alter* the *distribution* of gravity's force acting upon them. For example, whenever possible, it has become common practice to place a symmetrically formed orthosis about the scoliotic patient's asymmetric trunk. It is also common practice to alter the symmetry of the inner surface of these orthoses by attaching protruding pressure pads in order to further alter the distribution of gravity's force. In order to get a patient into such an orthosis, the force required to 'rearrange' the contours of a patient's asymmetric trunk must be supplied either by the patient's own musculature or by a person assisting the patient. Once applied, every square inch of the surface of the orthosis can but return an equal and opposite force to match the force each square inch is receiving.

When the patient is upright, the direction of the force of gravity parallels the spinal column; whereas the reaction force of current orthoses is directed primarily in a horizontal plane *perpendicular* to the column. When a wearer is recumbent, however, both the force of gravity and the reaction force of the orthosis are operative in the same plane, i.e., *perpendicular to the spinal column.* To appreciate gravity's effect when recumbent, visualize the full-length, lateral "C" curve that a normal spine can assume when a person lies on his side in a hammock slung between fixed points. It does not appear unreasonable to assume that when lying upon a flat surface, the spine must yield to gravity's force in the same manner, albeit to a lesser degree. If one accepts this assumption (given the physical makeup of the spinal column), it logically follows that the recumbent scoliotic spine within an orthosis must also yield in the same manner, but to a lesser degree. Whether the patient is in or out of an orthosis, the shoulder girdle and the pelvis are the fixed points between which the recumbent spine is 'slung'. Add the not infrequent presence of transverse rotation, wedge-shaped vertebral bodies, ligament and muscular tightness and/or contractures on the concave side, as well as the shifting of the nucleus pulposus to the convex side of the curvature(s), and very little lateral rotation between *adjacent* vertebrae within a curvature(s) is likely. When one or more of these factors is sufficient to resist gravity's force acting upon the site of the curvature(s), the individual vertebrae within the curvature(s) become "bodies at rest" and lateral motion is limited to the ends of the column. Such hammock-like motion would appear to favor movement of the overall column toward the midline, rather than reduction of the angle(s) of a lateral curvature(s). However, tight and/or contracted musculature on the concave side of a curvature(s) can be particularly troublesome and will be discussed later.

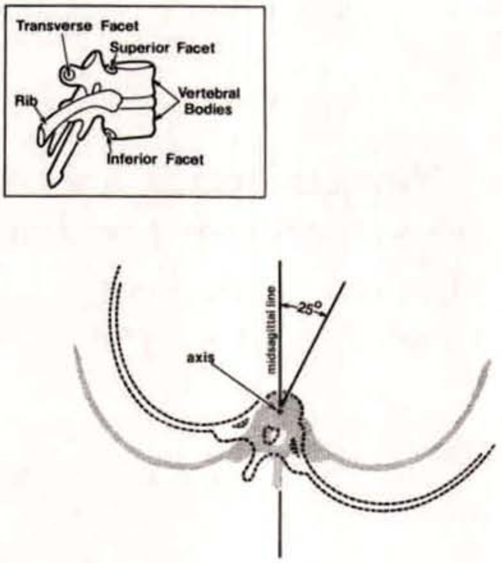

Fig. 1. Horizontal cross-sectional view: Schematic sketch of a thoracic vertebra showing that, beyond 25 degrees of transverse rotation, external force cannot be applied to the posterior aspect of the transverse process on the convex side. A rib's flexibility, coupled with the way it articulates with the vertebrae (as shown in the insert) indicates that it is a relatively weak structure. It is doubtful that a rib can transfer a force vector to a vertebra, in the same direction as it was applied, for the purpose of derotating the vertebra.

Figure 1 illustrates the problem of reversing transverse rotation which accompanies the more severe thoracic curvatures. The schematic drawing shows that the occurrence of transverse rotation in excess of 25 deg to the midsagittal line cannot be derotated by exoskeletal means because force cannot be applied directly to the transverse process on the convex side. The insert illustrates the articulation of a rib with the transverse process of one vertebra and the bodies of two adjacent vertebrae. When a thoracic vertebra is rotated beyond 25 deg, it can be seen that external force applied to the right rib, regardless of its direction, cannot cause the vertebra to derotate. The rib, because of its flexibility, will yield to the external force without affecting the position (as viewed in the transverse plane) of the vertebrae with which it articulates.

Within the last twenty years, fine basic research has been done on the biomechanics of the spinal column and trunk. The data reported in several of these studies have had a strong influence upon both the biomechanical analysis and the rationale upon which the design criteria that is to follow was based.

FUNDAMENTAL STUDIES

Lucas and Bresler reported in 1961 (5) that the self-contained balance of forces of internal pressure within the intervertebral discs, acting against the external binding tension of the ligaments, results in a very stable arrangement between *adjacent* vertebral bodies. However, these investigators also demonstrated that the erect spine, when supported by ligaments alone, will buckle laterally under a compressive force of only four and one-half pounds (2.04 kg), or one-sixteenth of superincumbent body weight. Clinically, this condition may be compared to the state of unconsciousness or a patient with totally paralyzed spinal musculature. (2)

An *in vivo* study by Nachemson and Morris published in 1964 (7) in which they measured intradiscal pressure in the lower lumbar discs, demonstrated that there is approximately 20 percent less pressure upon these discs in the reclining position than when standing. [In context with Lucas and Bresler's findings, it is reasonable to assume that the external tension of the ligaments is also reduced by the same proportion.] In the same study, one subject was given muscle relaxants and general anesthesia; the intradiscal pressure recorded from this subject was similar to that obtained from autopsy specimens.

In discussing stabilization of the upright spine, Cotch, in 1975, (2) refers to Lucas and Bresler's study of the behavior of the spinal column to the behavior of elastic rods. Their "Column-end fixation" and four "end support situations", as they relate to a critical loading of the column when vertical, are cited. In the situation where each end of the column is free to rotate but not to deviate laterally, the critical load the spine can support is approximately one-fourth of that which it can support when both ends are fixed, as is the case in the normal vertical situation. [This suggests that the spinal column when recumbent, in or out of orthosis, is placed in the situation in which each end of the column is free to rotate laterally, since gravity's force acting upon the column is perpendicular to its long axis. However, in an intimately fitting orthosis, the 'prepositioning' would be expected to reduce the amount of motion than can occur within the orthosis, because most of the elasticity that is present will have been utilized when donning the orthosis.]

In 1974, Markolf and Morris (6) demonstrated two phenomena exhibited by intervertebral discs: 1. A decrease of the compressive load occurs as a "function of time". This "load relaxation", over long-term intervals of time, results in a constant deformation when acted upon by a given load. 2. A disc under constant compressive force has a tendency to compress with time. This behavior is termed "creep". The rate of creep increases as the force level increases. These investigators feel that these phenomena may be related to the gradual decrease in the length of the spinal column that is known to occur during the day. Since the superincumbent weight resting upon the discs within a given curvature and the equal floor reaction force are perpendicular to the floor, the effect of load relaxation and creep upon the lateral angle of a curvature would tend to increase its angle as the discs compress during the day. The horizontal reaction force provided by an intimately fitted orthosis, being equal to the superincumbent weight, blocks any further lateral rotation of the curvature. The net effect is maintenance of the status quo—a significant achievement by any standards. The question is whether progress beyond this point is possible.

Some Conclusions

From the foregoing analysis, six major conclusions were drawn with respect to the non-operative treatment of idiopathic scoliosis:

1. Current scoliosis orthoses do not appear to affect lateral curvatures beyond the elastic stage.
2. The phenomena of load relaxation and creep exhibited by intervertebral discs seems to be a crucial element to the correction of lateral scoliosis. Assuming that discs respond in the same manner to force applied to them from any given direction, their response to the perpendicular direction of gravity's force, when recumbent, would produce a "looseness" that is essential to the reduction of lateral curvatures.
3. This leaves tight and/or contracted muscles as the only known mechanical element that can block the intervertebral discs' normal response to long-term compressive forces when in the reclining position. A component of dynamic external force, acting directly upon the vertebrae within a curvature during sleep, could provide an equally essential, continuous "loosening" of contracted musculature because these muscles would be incapable of offering dynamic resistance.
4. If conclusions two and three are valid, an orthosis which has a component *capable of generating a dynamic force* directed to act upon the vertebrae within a curvature(s) should be able to effect a change in the relationship of adjacent vertebrae. Since a single vertebra only weighs a few ounces, the magnitude of a dynamic force applied to the site of a curvature does not have to be large to constitute an "unbalancing" force. The involved vertebrae must yield to a force greater than the weight of each individual vertebra to which the force is directed.
5. Floor reaction forces are as destructive as superincumbent weight. Their unimpeded travel from the floor to the site of a curvature must be controlled to permit a shifting of pressure to the convex side when upright. If such a control were applied effeciently, the Heuter-Volkman law encourages the hope that nature would have the opportunity to reverse bony deformations, *if the amount of remaining growth were sufficient to complete the process.*
6. A reasonable expectation of such a

dynamic system would seem to be a long-term, positive influence upon the plastic stage, i.e., growth, could be realized.

DESIGN CRITERIA

The design criteria from which the new system was developed follows:

- Accept the reality that when the wearer's trunk is upright, all that can be expected of any system that could be tolerated is that it is a *holding device,* whether or not the system has dynamic components.
- Therefore, if a system is to have a long-term effect upon an idiopathic scoliosis—i.e., beyond the elastic stage and on into the plastic stage in order to 'guide' growth and thereby effect a reversal of both soft and bony tissue deformations—it must provide *dynamic* forces that are operative while the wearer is *asleep.*
- The destructive elements of gravity upon the scoliotic spine when upright are generally believed to be nil when the body is in a reclined position. With all soft tissue surrounding the spinal column in a relaxed state during sleep, contracted tissue may be expected to yield to long-term dynamic pressure accurately directed to the site of the most severely contracted area, i.e., tissue within the immediate vicinity of the apex of the curve, or curves.
- However, no matter how efficiently the dynamic force is applied to the immediate area about the apex of a curve, it will not be effective unless the rest of the vertebral column is under firm control, in order to prevent diffusion of the force in the form of unwanted lateral shifting of the column as a whole. Reduction of the angle of a curve at the site of its apex should result in an overall *elongation* of the spinal column, not in lateral displacement during the night.
- The system must be adjustable in a manner that permits the overall mediolateral width of the unit to be gradually drawn in to maintain, during the day, the correction (overall elongation) attained during sleep.
- All dynamic forces within the system must be readily adjustable, especially since the forces (both in magnitude and duration) necessary to attain complete correction of a curve of a given degree and/or location, with or without bony deformations, are as yet unknown.
- However the external dynamic force is applied, derotation of curves should be a major, if not the major, purpose for its application. Without positive derotation it will not be possible to affect correction of a given curvature beyond the elastic stage.
- Consideration must be given to the effect of floor reaction forces upon lumbar curves, whether or not the pelvis is involved. Correction attained by a system during nighttime wear cannot be fully retained during the day without a method to manipulate floor reaction forces. Control of floor reaction forces cannot be achieved within the system per se, so it will be necessary to provide control in some form that acts as an adjunct to the system.

THE DESIGN

A polypropylene thoracopelvic cylinder is formed over a modified plaster-of-Paris model of the patient's torso (Fig. 2). The major purpose of the

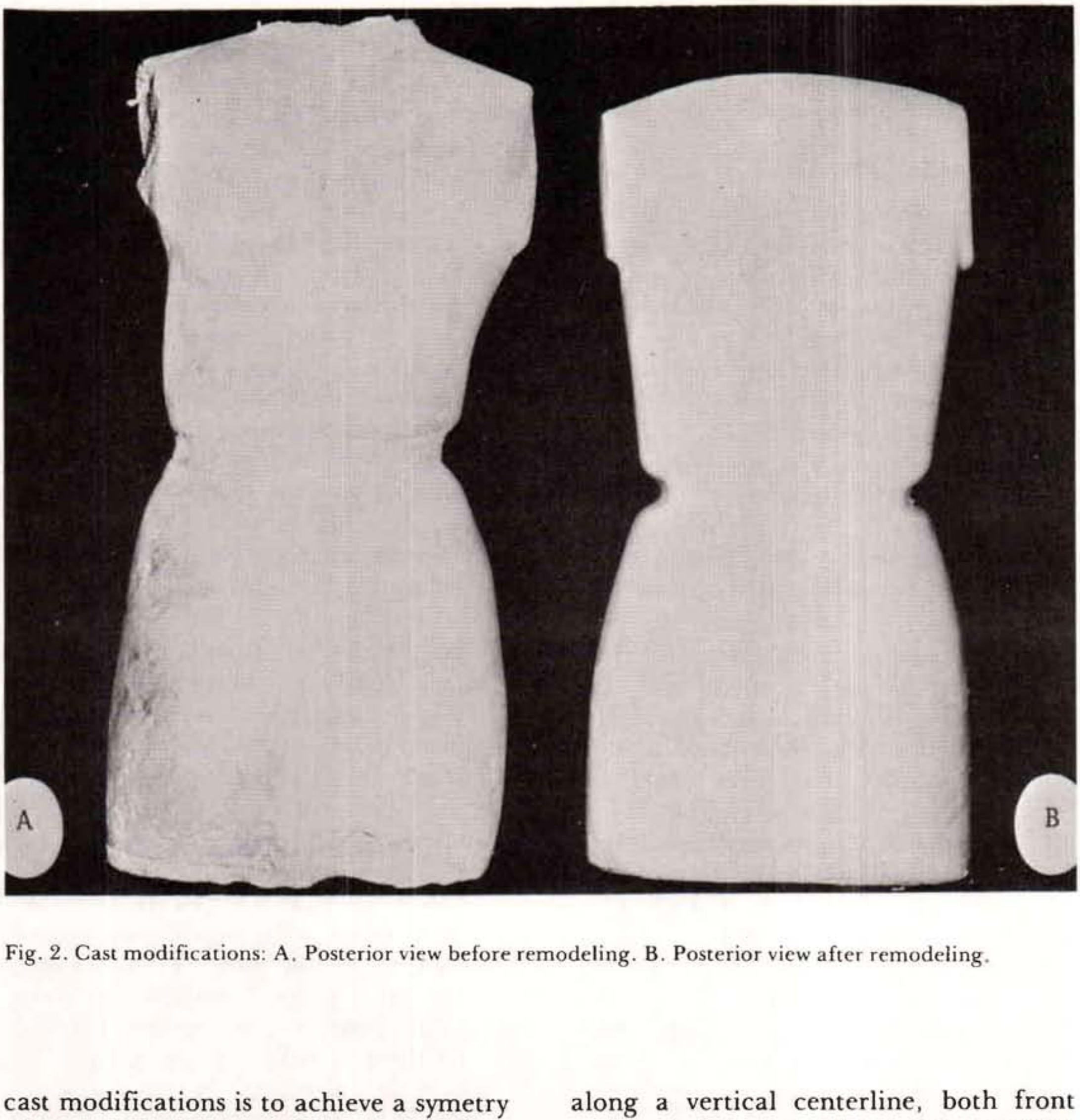

Fig. 2. Cast modifications: A. Posterior view before remodeling. B. Posterior view after remodeling.

cast modifications is to achieve a symetry of form to both sides of the model. Two important features of the "Milwaukee Brace" (1) are retained – the flattening of the abdomen and the indentations at waist level (Fig. 3) because they are efficient means of partially unweighting the lumbar region of the spinal column.

The plastic thoracopelvic form is lined with a closed-cell polyethylene foam and the waist indentations filled with Silastic elastomer. The thoracopelvic unit is cut along a vertical centerline, both front and back, into symmetrical left and right halves (Fig. 4). On a spinal X-ray film of a patient, a line, perpendicular to the floor, is drawn through the center of the body of the vertebra that forms the apex of the lumbar curve. Another line, parallel to the first, is drawn through the center of the vertebra that forms the apex of the thoracic curve. The distance between the two lines is then divided in half and that amount is cut off of both halves,

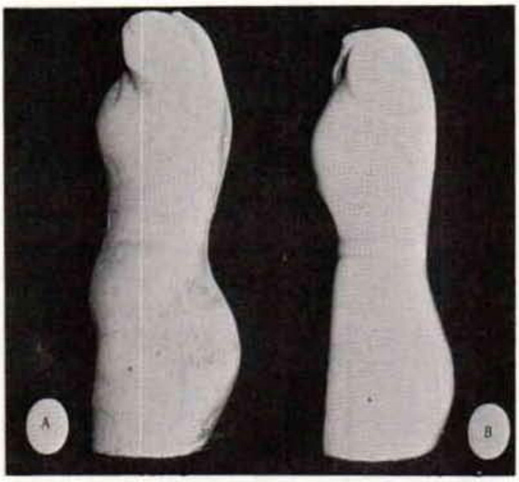

Fig. 3. Cast modifications: A. Lateral view before remodeling. B. Lateral view after remodeling. Note that an optimum amount of lumbar lordosis is retained.

along the center line, both front and back (Fig. 5) leaving a gap between both halves equal to the distance between the parallel lines drawn on the X-ray so that the two halves can be drawn together as the curves are reduced, thereby maintaining during daytime wear the reduction achieved by dynamic forces applied to the apices of the curves during the hours of sleep. Full correction of mediolateral alignment would be achieved when the mid-sagittal line passes through the center of each and every vertebral body. The two halves are joined by mounting slotted receptacles along the midline, both front and back, of one half of the unit and aluminum alignment bars that slide into the receptacles on the other half. Velcro straps prevent the two halves from sliding apart.

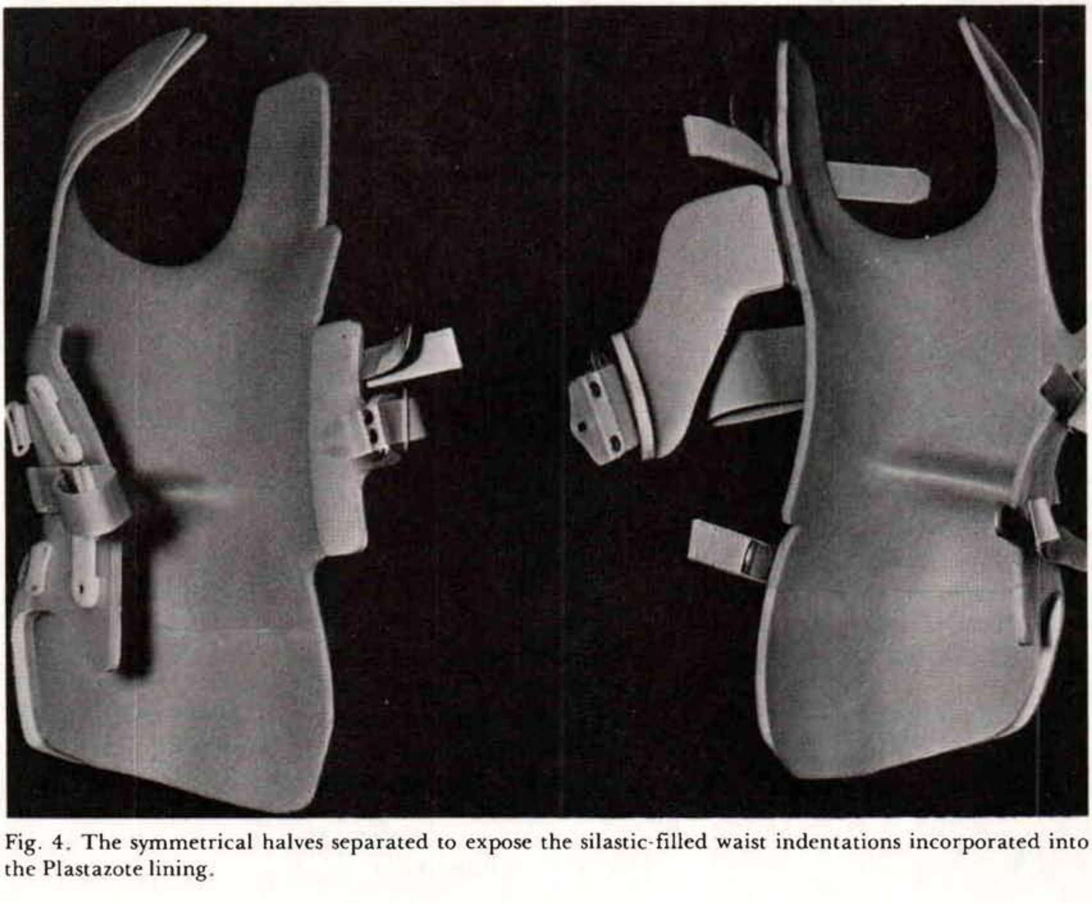

Fig. 4. The symmetrical halves separated to expose the silastic-filled waist indentations incorporated into the Plastazote lining.

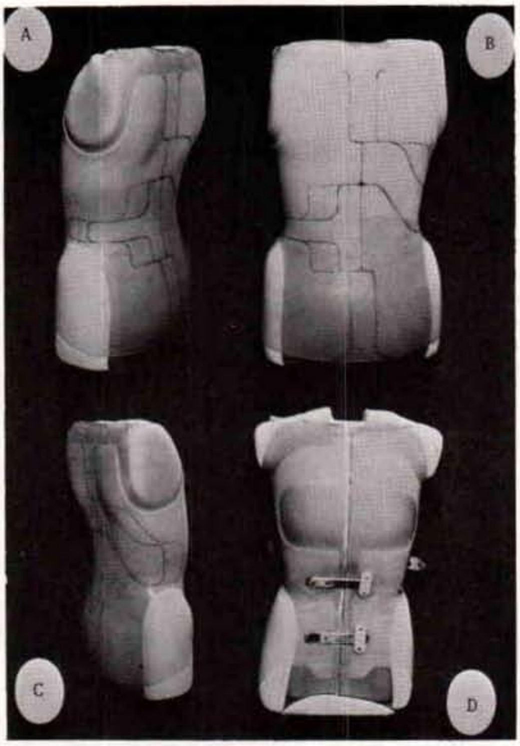

Fig. 5. Plan of the orthosis is drawn on the polypropylene form. The dotted lines indicate the amount of material that is to be cut out of the polypropylene surrounding the pressure pads to allow the two halves of the orthosis to come together. A. Left lumbar pad. B. Posterior view of the pressure pads. Note the cutouts along the center line, above and below the outlines of the pressure pads. C. Thoracic pad. D. Anterior view with the space along the center line cut away and with the aluminum bars and their recepticles attached.

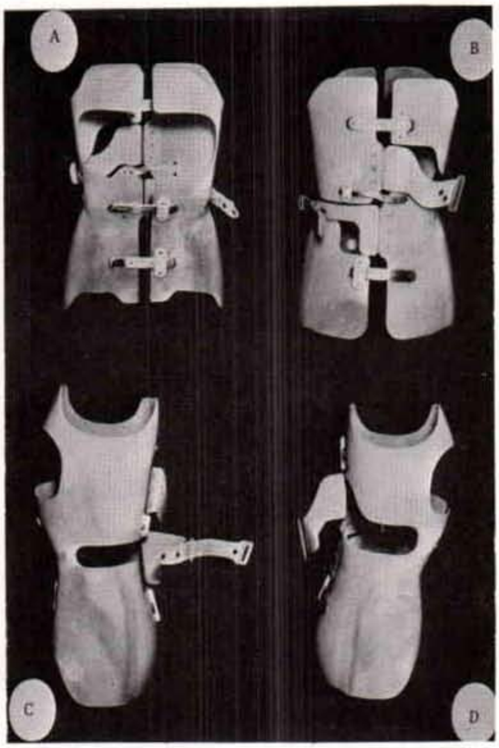

Fig. 6. The finished polypropylene form before the strapping and the lining are added: A. Anterior view. B. Posterior view. C. Lateral view with the left lumbar pressure pad with its elastic strap attached. D. Lateral view with the thoracic pressure pad swung back. (Note the gap along the center line and the cutouts around the pressure pads to allow both halves to slide together.)

The size and location of the corrective pressure pad(s) are determined from the X-ray film. The posteriomedial edge of a pressure pad is attached to the half of the unit that is contralateral to it by a polypropylene hinge (Fig. 6). Woven elastic strapping, three or four layers thick, is located so as to cross over the length of the outer surface of the pressure pad in a horizontal line. Figure 7 illustrates the type of force that can be generated and how a single force can be used to perform more than one function simultaneously.

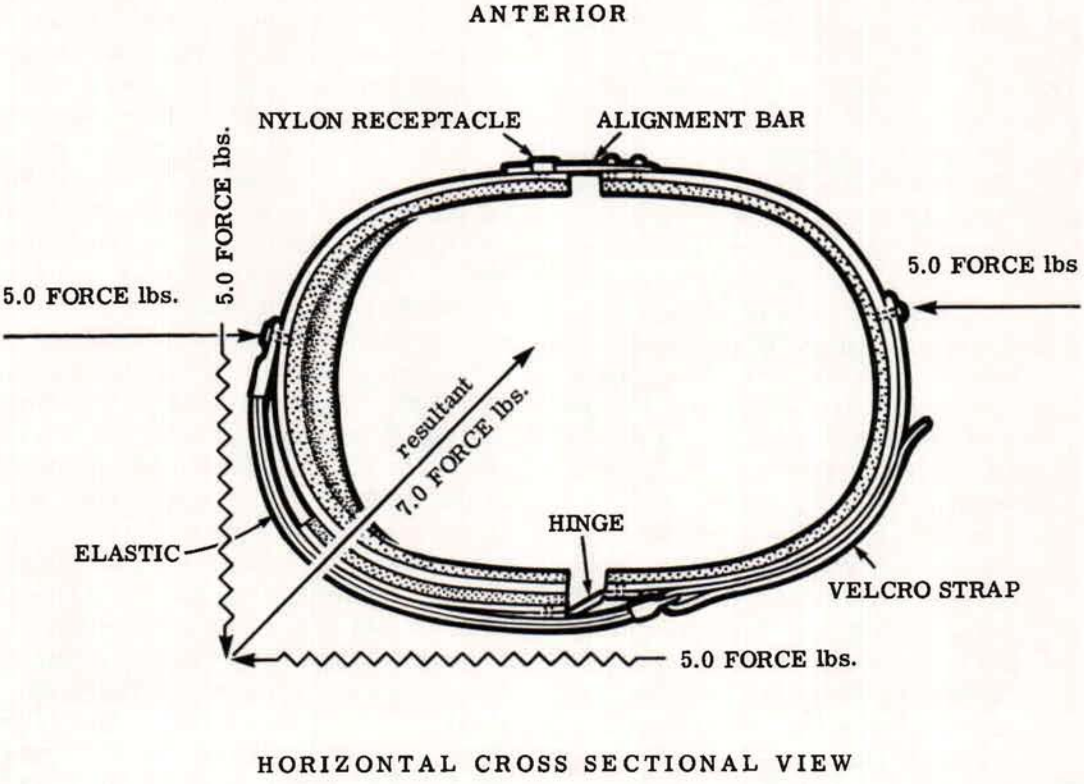

Fig. 7. Schematic sketch of the application of dynamic forces.

To visualize the multiple functions that the dynamics of the system are capable of delivering, this illustration shows a setting of 10 force pounds for a left lumbar curve. Let us say that this patient also has a compensatory right thoracic curve. Let us set the elastic acting upon the thoracic pressure pad over the curve's apex for 15 force pounds (it being reasonable to expect that the greater force will be needed to overcome the resistance of the multiple jointings of the rib cage to the thoracic vertebráe). Using the same method of calculation as shown, the resultant force acting upon the thoracic curve would be 10 1/2 force pounds, as compared to the seven force pound resultant for the lumbar curve shown in the sketch. Figuring the resultant force (as pictured) to be four inches lateral to the pad's hinge, a moment of 28 inch pounds would be acting to derotate at the site of the apex of a lumbar curvature. The 10 1/2 force pound resultant — also four inches lateral to the thoracic pressure pad's hinge — would produce a moment of 42 inch pounds to derotate. The magnitude of the horizontal vectors acting to draw the unit's two halves together would now be the *combined* force of both elastic straps generating 12 1/2 force pounds per side. The effect upon a given curvature is threefold:

1. Because the pressure pad is hinged along its medioposterior edge, the resultant vector (as illustrated) is converted to a dynamic moment that *derotates* the curvature.
2. Due to the placement of the elastic strap, the same resultant force generated by it is used to apply a dynamic inward thrust to the pressure pad which acts to *reduce* the lateral angle of the curvature at its apex.
3. The same force that is generated by the elastic strap (or straps) is utilized dynamically to draw both halves of the unit together, thus preventing the force being directed to an apex from being dissipated by unwanted sideward 'shifting' of the overall column. The overall effect is a *dynamic* three-point pressure system acting upon a curvature during sleep.

With such an interplay of dynamic forces, it seems possible to effect a continuous, positive influence upon the plastic stage for any period of time desired. It should be noted that, while the same magnitude of external force(s) that acts upon a curvature(s) during sleep is continuous during the day, the superincumbent weight, being greater, overpowers the external dynamic force. The Velcro straps convert the system to an efficient 'static' cylinder that resists the destructive elements of gravity's forces during the day. Thus, the system functions efficiently and automatically, both day and night, without conscious effort on the part of the wearer.

An explanation of the treatment of the chest area with the dynamic orthosis is needed to complete the description of its design. The choice of a modified version of conventional subclavicular extensions (in preference to a solid front) was made at the very beginning of the design's development (Fig. 5D). The open-chest feature allowed for two important functions, i.e., it provided freedom for derotation of involved thoracic vertebrae and chest expansion for ease of breathing. A receptacle and sliding aluminum bar joined the two extensions in the same manner as they are used to join the rest of the system together. A Velcro strap completed the original assembly. However, the Velcro strap proved to be too restrictive because it inhibited lateral expansion of the thorax. While the open-chest feature was thought to be adequate to accommodate for the normal tidal volumn of quiet breathing, the blocking of lateral expansion of the thorax appeared to interfere with the sigh reflex (2). To correct this important deficiency of the design, the Velcro strap across the subclavicular extensions was replaced by an elastic strap. The latter is set with one-to-two force pounds of "preload" which the sigh reflex can easily overcome.

The cushion heel lift (Fig. 8) that is used as an adjunct to the system which makes it possible to maintain correction of lumbar curvatures against the destructiveness of floor reaction forces. Figure 8A shows a left lumbar curve with an oblique pelvis. The hip on the concave (or right) side has been drawn upward with apparent shortening of the right lower limb. The large arrow above the figure represents the weight of the child's body above the level of the hip axes which is known to be approximately 50 percent of a person's total body weight (3, 4). The two smaller arrows under each heel represent floor reaction forces. It can be assumed that in normal balanced posture, both limbs are sharing the full weight of the body equally. The weight of the trunk borne by the spinal column is transferred to the pelvis, which is supporting the spinal column. The burden is equally divided and continues downward to the floor via the skeletal structures of the lower limbs. By the time contact with the floor has been made, the weight of the pelvis and each limb has been added to the vertical load that each foot is carrying. (It must be borne in mind that the leg length discrepancy is only apparent.) Thus, the child must compensate by keeping her left knee bent in order to lower her right leg so that it can contact the floor. By this compensatory flexion of the left knee, both lower limbs bear an equal share of the body's weight and overall mediolateral balance is restored. Unfortunately, such a 'no choice' com-

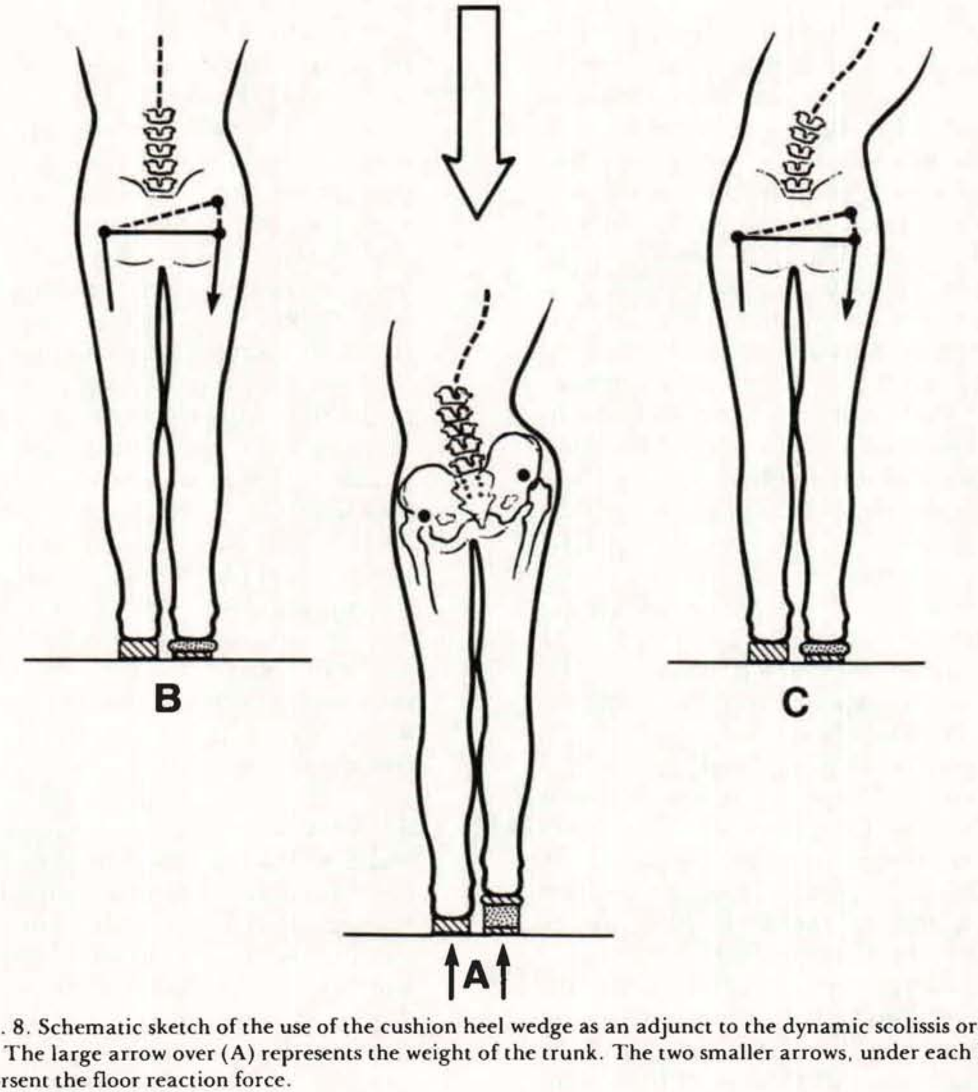
CONTROL OF FLOOR REACTION FORCES: THE CUSHION HEEL

Fig. 8. Schematic sketch of the use of the cushion heel wedge as an adjunct to the dynamic scolissis orthosis: The large arrow over (A) represents the weight of the trunk. The two smaller arrows, under each heel reprsent the floor reaction force.

pensation is unable to 'right' the oblique pelvis. Let us say that the patient weighs 100 pounds. With the 50-pound trunk resting upon the pelvis and the total body weight of 100 pounds pushing back in the form of a floor reaction force of 50 pounds through each lower limb, the net effect is to 'lock in' the pelvic obliquity. Under such circumstances, it can be seen that it is impossible to apply external forces directly to the spinal column via the trunk that could overcome the enormous mechanical advantage of the floor reaction force.

The cushion heel manipulates floor reaction forces by putting these forces to work to help correct that which they have wrought. The effect is achieved by adding, for example, a half-inch *solid* heel lift under the heel of the limb on the convex side of a curvature and a *cushion* heel lift of a half-inch, plus an additional thickness equal to the pelvic obliquity, under the 'short' limb on the concave side. Figure 8A shows the child bearing her full weight upon her fully extended left leg which has a solid heel lift under it. The sketch attemps to show the subject the instant before she begins to shift half her weight onto the right limb with a cushion heel lift under it. A rapid sequence of events occurs, as follows:

1. As weight is received by the cushion heel lift under the 'short' limb, the cushion absorbs a portion of it and thereby causes a delay in that portion of weight reaching the floor.
2. The fact that both hip joints have a normal range of motion makes it possible to manipulate floor reaction forces in a way that forces the oblique pelvis into a position parallel to the floor. Whereas, a portion of the weight being placed upon the cushion heel lift under the 'short' right limb is delayed for an instant on its way to contact with the floor, the full amount of the weight passing downward through the left limb reaches the floor *without* delay, therefore, a floor reaction force of equal magnitude travels up instantaneously through the left limb to the left hip. As the floor reaction force reached the level of the hip on the left side, the slight delay due to absorption by the cushion (only a very short period of time is necessary) has automatically reduced the magnitude of the floor reaction in its upward travel through the right or 'short' limb. The weight of the trunk, pelvis and right leg may be said to be 'falling free' during the instant delay. Their combined weight overcomes the weight of the left limb forcing rotation of the pelvis about a sagittal axis about the left hip joint.
3. The cushion heel compresses and becomes firmer as the amount of weight upon it rapidly increases and results in a proportional increase in the floor reaction force travelling up to the right hip. However, the floor reaction force will continue to be somewhat less in magnitude than it is in the left limb because the initial delay cannot be overcome until the cushion heel has 'bottomed out'. At the instant the cushion heel has compressed into as firm a platform as the one under the left heel, only then will the floor reaction force under the right heel be equal, in *timing* and *magnitude,* to the left side.
4. As resistance to the descending trunk and pelvis builds in the right limb—due to the rapidly increasing floor reaction force travelling up the the hip joint—the heavier weight from above will cause the necessary degree of rotation to occur about the right hip joint to complete the levelling of the pelvis, *before the*

floor reaction force under both feet become equalized.

The weight of the patient determines the durometer of the cushion material to be used. The material must not 'bottom out' before the pelvis is parallel to the floor. It must also be thick enough to allow the amount of 'drop' necessary to level the pelvis. The equalization of the floor reaction forces 'locks' the pelvis into its level position. This result occurs automatically whenever the patient is standing. A unilaterally cushioned seat can be provided which will, in the same manner, level the pelvis while in the sitting position.

What effect can the levelling of an oblique pelvis be expected to have upon a lumbar curvature? It depends on the tightness and amount of foreshortening of a musculature and other soft tissue on the concave side of the curvature at the time the cushion heel is applied. Figure 8B is a schematic illustration of what may be expected when the lumbar curvature is mild and flexible. It seems reasonable to expect that an intimately fitted *passive* plastic cylinder, with the assistance of a cushion heel lift, should maintain the spine in balance. The schematic drawing (Fig. 8C) shows the lateral shift of the entire trunk to the concave side that results when the musculature and other soft tissue on that side are tightly contracted and the pelvis is leveled without an orthosis. Obviously, the dynamic scoliosis system cannot release the contracted soft tissues as soon as it is applied. Time will be needed for its dynamics to achieve correction beyond whatever small amount of elasticity may be present. The tightness of the contracted tissues makes an immediate levelling of the pelvis, as shown in Figure 8C, unacceptable. Therefore, when the dynamic system is first applied, the right hip will be much as shown in Figure 8A, but with a slight compression of the cushion heel lift. It is in such cases that the cushion heel lift is an essential adjunct to the dynamic scoliosis system. For as the dynamic system begins to reduce and derotate the curvature over the long term, the cushion heel lift allows the hip on the concave side to become parallel with the ground gradually as the contracted tissues respond to the system's dynamics. The manipulation of the floor reaction forces is the same as previously described, except that instead of a level pelvis being attained in a matter of seconds (as was true of the case shown in Figure 8B,) the process may take weeks—or if necessary, months—to complete. Throughout the necessary time span, instead of floor reaction forces 'locking in' the pelvic obliquity during the day and thereby reversing any correction of the curvature obtained by night, the weight of the "semi-suspended' right limb is used to place all soft tissue on the concave side on stretch through the day.

Two major improvements to the original design were added later:

First, removable side pads were added. The dynamic drawing together of the two halves of the unit maintains the correction achieved by the pressure pads during sleep. Reduction of the angle of the curve(s) results in elongation of the spine and thereby also reduces the mediolateral width of the thorax. It became apparent that a way had to be found to retain the rigidity between the rib cage and the pelvis (which is vital to the system) and still accommodate for the variance in mediolateral width between the correcting spinal column and the relatively 'fixed' diameter of the pelvis.

Figure 9 shows two removable pads that are incorporated into the sides of the pelvic portion, between the lining and the outer polypropylene form. They are made of one-eighth to one-fourth-inch-thick firm, closed-cell polyethylene foam. Selection of the thickness of these pads is dependent upon the severity of the curve(s) and the age of the patient. As the spinal column elongates, the pads can be

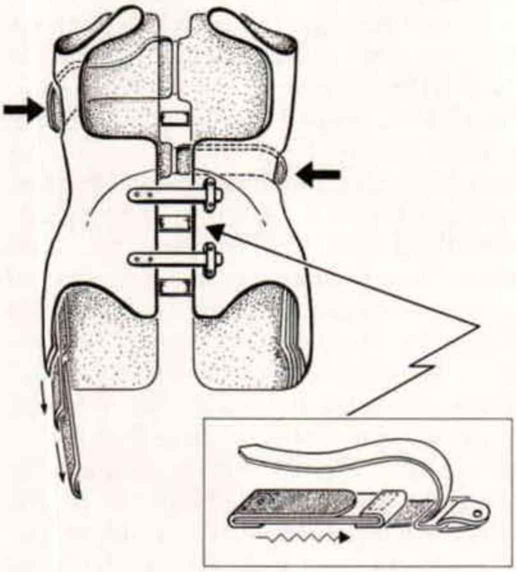

Fig. 9. Schematic sketch of the removable, lateral pelvic pads with the elastic 'balancing' strap shown in the insert.

removed in any order desired, thus ensuring a continuous, positive influence along the thoracic spinal column. When correction has been achieved, the systen is worn as a prophylactic during the remaining period of bony growth. Also, during the prophylactic period the removable pads can be utilized to accommodate the maturing pelvis as its mediolateral width broadens.

Second, an anterior elastic 'balancing' strap was added. The greater resistance that a more rigid anatomic structure presents initially to the dynamic force being applied to the pressure pad(s) caused the two halves of the unit to migrate around to the center of the back, where their medioposterior edges butted against each other. This butting stopped any further drawing in of the two halves and, consequently, blocked any further correction by the pressure pad(s). In short, the system became a *static* holding device both *night* and day.

An elastic 'balancing' strap, as shown in the insert in Figure 9, was added to the abdominal portion at approximately waist level. The tension of this strap is set to approximate the force(s), as previously described, acting to draw both halves of the unit together. This elastic does not increase the overall force being applied to the two halves. Its function is to eliminate rotation of the two halves about the trunk, by directing the force(s) acting upon the two halves toward the midsagittal line of the trunk, thus restoring the dynamics to the system.

SUMMARY OF DESIGN CRITERIA

Static orthoses are limited to reacting with an equal and opposite force to the weight of the trunk. As a consequence, the magnitude of force required to manipulate, or 'preposition', a curvature must be applied before and/or during the donning of these orthoses. A new dynamic scoliosis orthosis has been developed that utilized components that generate dynamic forces that act upon a curvature(s) while the wearer is asleep. The rationale and design criteria from which the orthosis was developed is described in considerable detail. The components and the function of the dynamic orthosis have been described in full.

RESULTS

To date, twenty-two idiopathic scoliosis patients and one eight-year old female with hypophosphatasia have been fitted

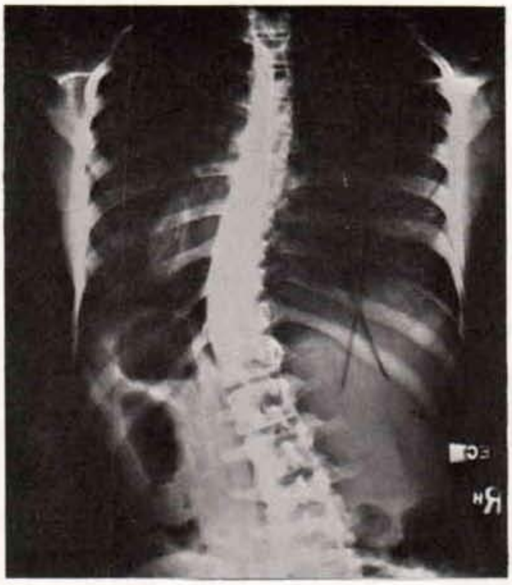

Fig. 10. Patient D.H.: Left lumbar of 32 degrees with a pelvic obliquity and a right thoracic of eight degrees. A dynamic pressure pad was directed to the site of the lumbar curve with a setting of 10 force pounds. A dynamic pressure pad was not used for the mild thoracic curve. This film was taken on March 13, 1976. The patient was then 14 years and one month old.

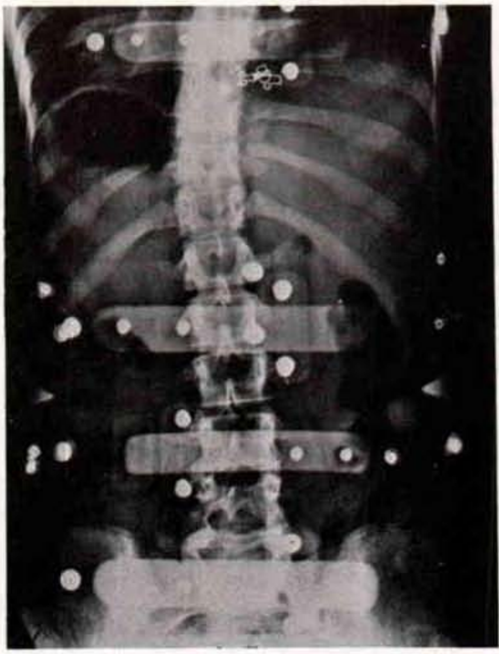

Fig. 12. Film was taken August 1, 1977. This orthosis which is the same D.H. wore in Figure 11 was an early prototype and did not have the improvements referred to in Fig. 13.

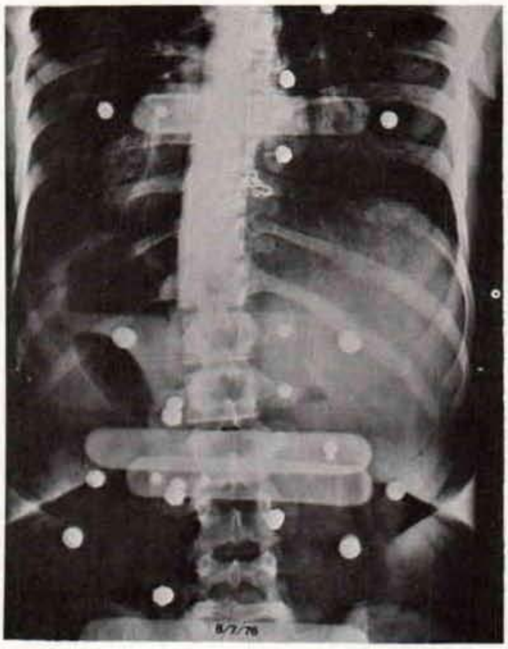

Fig. 11. Film taken August 7, 1976, three weeks after D.H. received her dynamic orthosis, which replaced an earlier one which did not function properly.

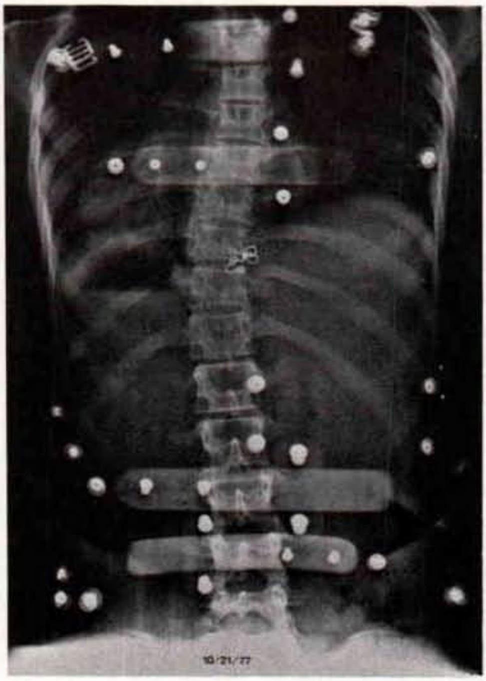

Fig. 13. Film was taken October 21, 1977, three months after she had received a new orthosis that has the removeable pelvic side pads, the 'balancing' strap and the elastic strap joining the subclavicular extensions. Patient had outgrown the pelvic portion of her previous orthosis.

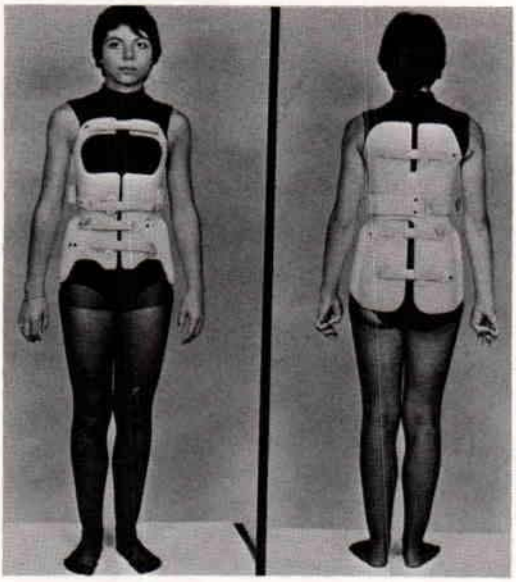

Fig. 14. Patient D.H. wearing her dynamic scoliosis orthosis. Note the narrowness of the midline openings in front and back which relates to correction achieved.

with the new orthotic system. The patients are representative of a variety of curvatures. Curvatures with an apex at T8 are the highest that have been attempted. The majority of those patients have been wearing a dynamic orthosis for less than a year. There is much to be learned about the skillful use of the new orthosis as a tool in the non-operative treatment of scoliosis. It is still too soon to evaluate the effects of its long-term use. However, results to date have been gratifying and it would appear that the concept of the use of dynamic forces that are operative while the patient sleeps is valid. The progress of one patient is shown in Figures 10-14.

Footnotes

[1] Portions of this report were presented at the 1977 World Congress for Prosthetics and Orthotics in New York City, May 27-June 2, 1977; the National Assembly of the American Orthotics and Prosthetics Association in San Francisco, October 25-29, 1977; the Fourth Annual Roundup Seminar of the American Academy of Orthotists and Prosthetists in Orlando, January 19-21, 1978; the University of Kentucky seminar "Orthotics and Biomechanics" in Clarksville, Indiana, February 2-4, 1978, and educational seminars sponsored by the American Academy of Orthotists and Prosthetists in Region VI (December, 1976), Region V (October, 1977), and Region IV (November,

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References

(1) Blount, W.P. and J.H. Moe, *The Milwaukee Brace*. Baltimore: Williams and Wilkins Company, 1973.

(2) Cotch, M.T., *Biomechanics of the thoracic spine*. Chapter 16, in: *Atlas of Orthotics*. St. Louis: C.V. Mosby Company, 1975.

(3) Dempster, W.T., *Space requirements of the seated operator* (WADC TR 55-159), Wright-Patterson Air Force Base, Ohio, 1955-A

(4) Glancy J. and R.E. Lindseth, *A dynamic orthotic system to assist pelvic extension: a preliminary report.* Orthotics and Prosthetics, 27 (1): 3-9, March, 1975.

(5) Lucas, D.B. and B. Bresler, *Stability of the ligamentous spine,* Biomechanics Laboratory, University of California. Technical Report No. 40, San Francisco, January 1961.

(6) Markolf, K.L. and J.M. Morris, *The structural components of intervertebral disc; a study of their contribution to the ability of the disc to withstand compressive forces,* J. Bone and Joint Surgery 56-A:675-687, June, 1974.

(7) Nachemson, A. and J.M. Morris, *In vivo measurements of intradiscal pressure: discometry, a method for the determination of pressure in the lower lumbar discs,* J. Bone and Joint Surgery 46-A:1077-1092, July, 1964.

Additional References

Arkin, A.M.jk and J.F. Katz, *The effects of pressure on epiphyseal growth,* J. Bone and Joint Surgery 38-A:1056-1076, 1956.

Edmonson, A.S. and J.T. Morris, *Follow-up study of Milwaukee brace treatment in patients with idiopathic scoliosis,* In: *Clinical Orthopaedics,* No. 126, M.R. Urist (editor-in-chief). Philadelphia: J.B. Lippencott Co., pp. 58-61

Enneking, W.F. and P. Harrington, *Pathological changes in scoliosis,* J. Bone and Joint Surgery 51-A:165-184, 1969.

Glancy, J., *Dynamic control of abnormal M-L motion of the os calcis: the cushion heel wedge—a possible solution,* Orthotics and Prosthetics 27 (1):28-33, March, 1973.

Gregersen, G.G. and D.E. Lucas, *An in vivo study of axial rotation of the human thoracolumbar spine,* J. Bone and Joint Surgery 49-A:247-262, March, 1967.

Harrington, P.R., *The etiology of idiopathic scoliosis,* In: *Clinical Orthopaedics,* No. 126, M.R. Urist (editor-in-chief). Philadelphia: J.B. Lippincott Co., 1977. Pp. 17-25.

Lumsden, R.M. and J.M. Morris, *An in vivo study of axial rotation and immobilization at the lumbosacral joint,* J. Bone and Joint Surgery 50-A:1591-1602, December, 1968.

Mellencamp, D.D., W.P. Blount and A.J. Anderson, *Milwaukee brace treatment of idiopathic scoliosis; late results,* In: *Clinical Orthopaedics,* No. 126, M.R. Urist (editor-in-chief). Philadelphia: J.B. Lippincott Co., 1977. Pp. 47-57.

Moe, J.H., *A critical analysis of methods of fusion for scoliosis: an evaluation in two hundred and sixty-six patients, Ibid,* pp. 4-16.

Nash, C.L., Jr. and J.H. Moe, *A study of vertebral rotation,* J. Bone and Joint Surgery 51-A:223-229, March, 1969.

Nordwall, A., *Studies in idiopathic scoliosis; relevant to etiology, conservative and operative treatment,* Acta Orthopaed. Scand. Supplementum No. 150, Munksgaard, Copenhagen, 1973.

Park, J., S. Houtkin, J. Grossman and D.B. Levine, *A modified brace (Prenyl) for scoliosis,* In: *Clinical Orthopaedics,* No. 126, M.R. Urist (editor-in-chief). Philadelphia: J.B. Lippincott Co., 1977. Pp. 67-73.

Prime, F.J. and P.A. Zorab, *Respiratory function in scoliosis,* In: *Scoliosis,* P.A. Zorab (editor). Springfield, Ill.: Charles C Thomas, 1969.

Watts, H.G., J.E. Hall and W. Standish, *The Boston system for the treatment of low thoracic and lumbar scoliosis by the use of a girdle without superstructure,* In: *Clinical Orthopaedics,* No. 126, M.R. Urist (editor-in-chief). Philadelphia: J.B. Lippincott Co., 1977. Pp. 87-92.

Winter, R.B. and J.M. Carlson, *Modern orthotics for spinal deformities, Ibid.* pp. 74-86.