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A DYNAMIC ORTHOTIC SYSTEM FOR YOUNG MYELOMENINGOCELES

A PRELIMINARY REPORT[1, 2]

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Before discussing the approach that is in development at the Indiana University Medical Center for improving orthotics for myelomeningocele children, a few remarks about conventional orthoses in general use may serve as a point of reference.

CONTROL OF THE FOOT/ANKLE COMPLEX

The conventional double-upright, below-knee orthosis cannot control the paralyzed foot/ankle complex. Its principal weakness is the shoe. Because leather is a flexible material, especially when wet, and the intimacy of the fit of high-top shoes is decidedly suspect, a shoe cannot check unwanted mediolateral rotations about the subtalar joint of a flail foot. Acknowledgement of the control limitations of shoes led many years ago to the development of tee straps and mediolateral wedging.

Unfortunately, tee straps are also made of leather. The accumulation of scientific data by use of force plates has given us an understanding of the effect of floor reation forces upon foot balance. The validity of placing a flail foot upon an inclined surface, not parallel to the floor mediolaterally, to achieve a 'balanced' distribution of weight through the subtalar joint is questionable (*3*).

CONTROL OF THE KNEE JOINT

The double-upright, above-knee orthosis is generally used to control mediolateral and/or anterioposterior instability of the knee joint. Locks are added when buckling of the knees is the major problem. In any case, these orthoses are too heavy for the small paralyzed child. Also, the small paraplegic child cannot manipulate the locks and when he falls, getting to his feet again, on his own, is decidedly difficult for him.

CONTROL OF THE PELVIS AND TRUNK

Conventional orthotic means of controlling the pelvis and trunk include hip joints (both free and locking), metal pelvic bands and/or spinal orthoses. All of these components inhibit motion to a degree that many clinicians find unacceptable. They are bulky and heavy, and the smaller the child, the greater the effect. There being no other alternatives, these components have been used in varying combinations to serve two major purposes:

1. To prevent the development of hip contractures and excessive lordosis in growing myelomeningocele children.

2. To support the pelvis and trunk in the standing position in order that the very young child can develop a 'sense of balance' prior to attempting ambulation.

Within the past several years, prebalanced, platform orthoses such as the Parapodium (*6*) and the Swivel Walker (*7*) have been developed for young myelomeningocele children. Since platform orthoses have been available for such a relatively short time, I will continue to confine my remarks to what I have been referring to as

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the 'conventional'. Does placing a myelomeningocele child in a static orthotic system, regardless of its design, contribute to his development of a 'sense of balance'? The question remains open.

How successful then has the combining of hip, pelvic, and trunk components been in the prevention of musculoskeletal deformities of the hips and spines of myelomeningocele children? I believe that it is fair to say that success with conventional orthoses in preventing the development of deformities has been, at best, a 'sometime thing'. What then, are the factors that explain such poor results? The following general conclusions were drawn and served to stimulate the developments now underway at Indiana University:

1. *The dilemma that the state-of-the-art of orthotics presents; namely that the cost of longterm protection of the musculoskeletal system of the growing myelomeningocele child is an unacceptable restriction to motion in terms of the immediacy of daily living. Conversely, removal of available means of longterm protection inevitably leads to an increase in deformities, followed by an inevitable decrease in daily activities.*

2. *The difficulty of getting the myelomeningocele child and/or his family to accept a conventional orthosis because it is an additional burden and/or restriction. We are all familiar with the proud parent's descriptions of how 'active' their young child is; "He can get anywhere he wishes by using his arms to slide across the floor, but when in his braces, he moves about very little."*

3. *The longer the growing child of conscientious parents is kept in conventional orthoses, the more obvious it becomes that the orthoses are* not *preventing further deformities and the rational for their continued use becomes less and less defensible.*

4. *The older a myelomeningocele child is when introduced to conventional orthoses, the less likelihood there is that he will wear them and derive advantages from them. Psychological reasons excepted, the crux of the dilemma is a lack of function. Experience in other areas seems to indicate that when a patient is able to function better when wearing an orthosis, he quite willingly accepts the unavoidable annoyances that are a part of wearing one.*

5. *Is cosmesis important in the final analysis? To those who have experienced the bracing of hundreds of patients from childhood through the teenage years and into adulthood, the answer is a very positive* yes, *especially to the individual whose condition will require the assistance of orthotic devices throughout his life. As the individual grows from early adolescence to maturity, appearance becomes his dominant concern, especially when functional improvement is marginal. To be physically 'different' than one's fellows in a private way, (under one's clothing), is a great deal less disconcerting than the* forced *public display (there being no other alternatives), of the particulars of one's physical differences. Those who are able to overcome the* lack *of choice are rare exceptions.*

Our answer to the failure of conventional below-knee orthoses to prevent deformities of the flail foot, once standing begins, is the "solid-ankle" orthosis (Fig. 1) (*2, 5*). The

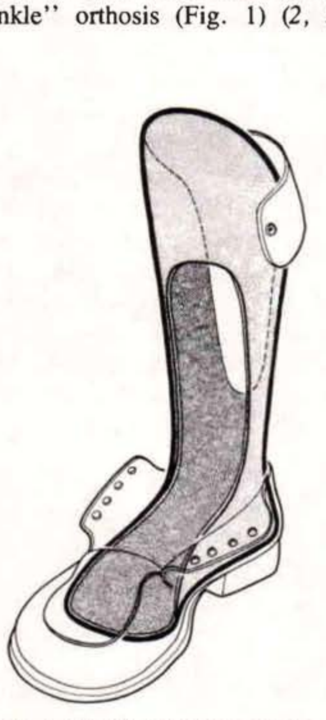

Fig. 1. The "solid-ankle" orthosis.

rationale for the design of the "solid-ankle" orthosis is to achieve balance and gait as close as possible to the bilateral below-knee amputee by eliminating unwanted motion of the flail foot/ankle complex in all planes. During its development, consideration was given to: Control of knee flexion; lightness and durability; its being worn inside the shoe; and that the orthosis should be a single unit to ensure proper donning and function. The shell is molded of polypropylene and lined with Plastazote. The "solid-ankle" orthosis weighs from 6-8 ounces, depending upon the size of the wearer.

The "solid-ankle" orthosis prevents buckling of the knee (*4, 6*) by inhibiting rotation of the tibia about the ankle axis. However, the myelomeningocele child presents further complications that were not readily apparent during the first few years that "solid-ankle" orthoses were used.

The complications arise because of three factors:

1. Stability of the knees depends upon the patient keeping his center of gravity forward of his knee axes.

2. The myelomeningocele child's lack of awareness of floor reaction forces below the level of deficit.

3. The total or partial lack of proprioception in all joints below the level of deficit.

The efficient control of the foot/ankle complex provided by the "solid-ankle" orthosis cannot prevent further deformities from developing in the hip and lumbar regions. (Fig. 2).

It is easier to visualize these children's problems by beginning with a child in the seated position wearing "solid-ankle" orthoses, (it being understood that, though not shown, the child must use his arms for assistance). Note the lordotic posture of the seated child in Figure 2A. Innervated muscles cross the hip joints anteriorly, but there are no muscles posteriorly to check the out-of-phase anterior rotation of the pelvis that occurs as the child descends to the seated postion. Once seated, there being no way for him to 'right' the lumbosacral angle of his pelvis, his lumbar spine must compensate by going into excessive lordosis. It is the only option he has for getting the CG of his trunk

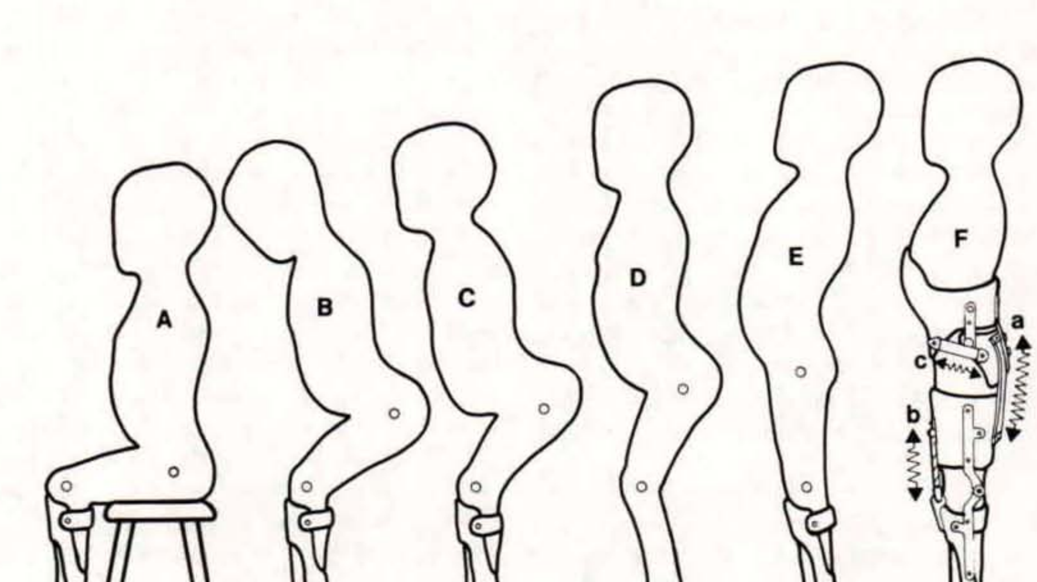

Fig. 2. Sketches to illustrate some of the problems of the myelomeningocele child.

over his buttocks and thus maintain sitting balance.

Figure 2 B shows the child beginning to rise to his feet and at this stage, his CG is well behind his knee axes and feet.

Figure 2C follows the ascent to a more extended position. The quadriceps are firing to extend the femurs about the knee axes. Since the tibiae are 'fixed' by the "solid-ankle" orthoses, the vasti muscles perform efficiently in their normal function of extending the femurs about the knee axes. However, because of the absence of opposition from the gluteus maximus and hamstrings that cross the hip joint, the rectus femoris becomes a powerful *unwanted* rotator of the pelvis in an anterior direction, bringing the trunk forward with it. It is apparent that, were this motion to continue, the youngster's CG would be forward of his feet and he would fall forward. The sketch attempts to catch the instant when the child is beginning to increase his lordotic posture to a much more severe degree — his only option for bringing his CG back over his feet. Unfortunately, his vasti muscles must now work constantly to prevent his femurs from going into further flexion about the knee axes. Unable to rest, they soon tire and the youngster will fall down backwards.

Figure 2D illustrates how it is possible for a child to maintain his balance without orthoses. Flexing his tibiae forward about his ankle axes permits him to bring his knee axes forward of his ankle joints. He is now able to maintain his balance by going into an excessive lordotic posture but with one all-important difference—he is able to keep his CG forward of his knee axes. Thus the burden upon the vasti muscles is dramatically reduced. The relationship of the development of hip and knee contractures and excessive lordosis to anterioposterior balance becomes apparent.

Figure 2E shows the posture that can be achieved with "solid-ankle" orthoses and posterior heel wedges when the orthoses are formed so that the feet are held in a neutral position with respect to the tibiae. The posterior wedge brings the knee and hip axes forward of the ankle axes. It is especially significant to note that the hip joints are *more forward* than either the knee or ankle joints. If these children's feet were cast in dorsiflexion, the result when standing, would be to force them into the posture shown in Figure 2D. When the feet are placed in equinus as in the technique used with poliomyelitis patients (*8*), the hip and knee axes are maintained directly over the ankle axes. The resulting vertical alignment of these joints makes anterioposterior balance of the trunk about the hip axes extremely precarious due to the ever present involuntary rotation of the trunk, i.e., 'swaying' in the anterioposterior plane. Were his joints aligned vertically, an increase in lordosis in the lumbar region would place his CG behind his feet and he would then have to flex his trunk forward about his hip axes in order to keep his CG over his feet. The forward placement of the hip joints, as illustrated, brings the mass of the trunk forward of the knee axes, thereby adding to their stability. The additional stability of the knee joints allows a relatively mild but essential increase of lordosis to bring the body's CG back over the base of support *without having to flex the hip joints to achieve balance*.

When "solid-ankle" orthoses are first applied, the result is very gratifying. Unfortunately, two or three years later, it is apparent that the battle to prevent hip contractures and increased lordosis is being lost. Although prevention of deformities to the foot/ankle complex and knee joints has been successful by eliminating the need for flexion in the ankles and knees to maintain balance, without his gluteus maximus muscles the only option the child has to maintain his balance is excessive lordosis. As time passes, the vertical load upon the lumbar spine is bound to increase the lordotic curve and without the opposing stabilizing force of the gluteus maximus muscles, the pelvis will be forced to rotate anteriorly. Growth, and the inevitability of bony deformation locks in the deformities.

Figure 2F shows portions of the present design of the dynamic orhtotic system that has been developed for the myelomeningocele child. The components shown are for the L3 to L4 level. This system consists of the following bilateral components; "solid-ankle" shells, single lateral aluminum uprights with offset knee joints, polypropylene quadrilateral thigh cuffs and the thoracopelvic unit with its pivotable pelvic portion. Dynamic forces (generated by elastic materials), are applied to the system as the zig-zag lined arrows indicate. The letter 'a' identifies the pelvic extension assist; 'b' the knee extension assist; and 'c' the part designed to keep the lower paraspinal muscles on stretch. Note that no rigid members cross the hip joints and that the system is free of locking devices.

Figure 3 shows details of the design of the polypropylene thoracopelvic unit. The anterior view (Fig. 3A) shows the opening with a Velcro closure. Above and below the Velcro strap are .050-in. thick stainless steel strips that fit into nylon pieces that are slotted to receive them. This feature ensures that the portions on both sides of the opening cannot slide up or down or bulge in or out with respect to each other, thus preserving the integrity of the fit and also preventing any 'sawing' action upon insensitized portions of the patient's skin. The flattening of the abdomen is the same as used in the Milwaukee-Brace technique (*1*). The lateral views (Fig. 3B, C) also demonstrate the pivotal action of the pelvic portion and the sliding of

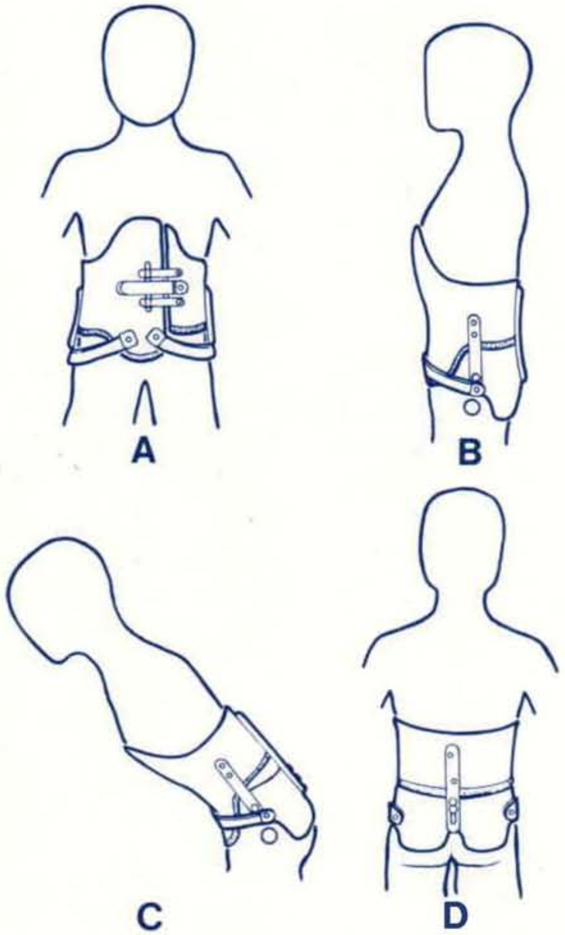

Fig. 3. Polypropylene thoracopelvic unit: A, anterior, B and C, lateral, and D, posterior view.

the slotted posterior polypropylene upright that allows the posterior of the thoracic portion to slide away from the pelvic portion to accomodate for approximately 1/3 of the elongation of the back as it flexes forward. Note the constant pressure that the pelvic portion is able to maintain upon the buttocks. The posterior view (Fig. 3D) shows the position of the slotted posterior upright, which serves as a stop to prevent excessive lumbar lordosis, when standing or sitting. Contrary to conventional spinal bracing, the *thorax* is used as the *foundation* from which the pelvis is controlled. The entire thoracopelvic unit weighs from 12 to 18 ounces, depending on the size of the child.

The schematic drawing in Figure 4 demonstrates the specific functional advantages that a myelomeningocele child realizes from the force generated by the elastic pelvic extension assist. To clarify the drawing, the elastic is not shown, but is represented by the zig-zag arrow from the proximal edge of the pelvic portion (a) to the distal edge of the quadrilateral thigh cuff (b) to illustrate its attachment points as well as its line of force. The system shown is designed for the youngster who has both medial and lateral hamstrings. Because stability of the knees is not a problem, neither the offset knee joints nor the knee extension assist units are necessary. However, the lack of innervation to the gluteus maximus muscles does leave us with the problem of preventing progressive lordosis. It is interesting to note that youngsters with the full complement of hamstring muscles do not develop *fixed* flexion contractures of the hips but do adopt a posture of hip flexion which appears to be an accomodation, for purposes of balance, to the excessive lordotic posture that they cannot avoid.

Fig. 4. Schematic lateral view of the orthotic system for the myelomeningocele child who has stable knee joints.

This system is based upon the premise that in these cases involuntary swaying of the pelvis is occurring constantly about the hip joints in the anterioposterior plane. It is assumed that tone in the abdominal muscles and anterior musculature crossing the hip joints is adequate to control involuntary rotation of the pelvis in the *posterior* direction. Because innervated anterior and posterior musculature crossing the knee joints is virtually complete, it is further assumed that both anterior and posterior tone is also present. The stability of the foot/ankle complex within the "solid-ankle" orthoses rules out involuntary anterioposterior swaying at either the ankle joints or at floor level. With these 'givens' present, the paralysis of the gluteus maximus muscles takes on a more subtle significance, i.e., the missing element to trunk balance appears to be the absence of the relatively light activity commonly referred to as 'tone', *not* the great power that these muscles are capable of generating. If then, a low magnitude of force is adequate to maintain equilibrium of the normal trunk, elastic materials seem to be a practical means of introducing dynamic control of trunk balance. The key seemed to be to apply the dynamic force in a form that would mimic the action of the missing gluteus maximus muscles, i.e., in the form of an extension *moment* about the hip joints. All signs seem to indicate that whenever the gluteus maximus muscles are paralyzed involuntary sway about the hip axes in an anterior direction is the *trigger* that sets off excessive lordosis. The formula for determining how much force should be applied is based upon free-body analysis and has been published (*4*).

The trunk and leg motions depicted in Fig. 4 are to scale in order that one may relate various degrees of motion about the hip axes to the actual positions of segments of the body as they can be clinically observed. As an example, let us take the child who weighs 30 lbs. and whose trunk length (measured from the center of the hip joint to the top of the head) is 20 inches. A preload of four pounds of force (which would be generated by bilateral elastics, each one set for two pounds of force), would allow the child to voluntarily flex his trunk forward five degrees about the hip axes and would automatically extend it for him within the five degree range. Ten degrees of forward flexion would require a preload to each of the bilateral elastics of slightly more than four pounds to automatically return the trunk to a balanced position of extension. For 15 deg. of forward flexion a 6.25 lb. preload setting to each of the bilateral elastics would be needed to automatically return the trunk to equilibrium. The preload settings allow motions of the arms and head (which affect the position of the CG of the trunk), without upsetting trunk balance. The elastic pelvic extension assist, then, provides three functions simultaneously:

1. It maintains trunk balance.

2. It places soft tissues that cross the hip joints anteriorly on constant stretch to prevent flexion contractures.

3. It provides a "safety zone" within which the child may move his arms and head and/or flex his trunk without fear of jackknifing.

The child can overpower the extension moment at will, simply by flexing his trunk beyond the preset safety zone. As he flexes forward, he increases the distance between the attachment points at either end of the elastics and, in so doing, automatically increases the force of the elastics. The extra force then becomes a control mechanism for motion initiated by the child. For example, as the child lowers himself into a chair, the extension moment about the hip axes will automatically increase and by the time he is seated, the extension force will be 2 to 2.5 times greater than its preload setting. Conversely, the child has assistance rising from a chair from this extra force which automatically diminishes as he extends his trunk. Thus, the mechanical pelvic extension assist appears to be a reliable, albeit limited, replacement for some essential functions of the normal gluteus maximus muscles.

The young wearer is now placed in a learning situation. He *must* be an active participant in the control of his body, i.e., he must understand that voluntary movements of his trunk within the "safety-zone" are totally within his control, whereas *involuntary* movements beyond the safety range mean a loss of balance. It is felt that this learning process can be likened to that of a normal child, but with obviously narrower margins of safety to be mastered. The feeling is that the dynamic environment presented by the orthotic system makes it possible for the child to actually develop a 'sense' of balance and confidence. This last statement is based upon the fact that the intimate fitting of the thoracopelvic unit serves as a simple, but efficient, feedback system (in the form of alterations of pressure) to the wearer. He now can be aware of the occurence of motion when it *begins* instead of when it's too late for him to maintain control.

There are disadvantages. Elastic materials, as a source of external power, present strict limitations upon orthotic design. The most important to this discussion are:

1. The limits of available power in relation to the need.

2. The practical implications of specific characteristics of elastic materials, e.g., increased elongation means increased power,

whereas the limits of elongation mean limits to freedom of motion. To illustrate: The elastic material selected to assist pelvic extension must have a stretch range sufficient to permit activities such as sitting or climbing stairs and still deliver an amount of power that substantially aids, rather than inhibits, such activities.

3. Due to the need for constant force for purposes of balance, attention must be given to shear forces upon the skin of the thighs and lower trunk, especially upon areas that are insensitive. This consideration, when coupled with the characteristics of elastic materials, has the most profound influence upon the system's design.

Figure 4 also illustrates the functional advantages that the dynamic pelvic assist offers the myelomeningocele child in activities such as walking and climbing stairs. The contralateral limb is shown in 25 deg. of hip flexion, just prior to heel-strike, and again in 60 deg. of hip flexion, as if the foot had just been placed upon a step. As the limb in swing phase flexes forward, the distal attachment point, 'b', of the elastic on the posterior panel of the quadrilateral cuff draws away from the proximal attachment 'a'. This elongates the elastic as it conforms to the changing relationship between the thigh and the buttock and thereby increases its force. The increase in force from distal point 'b' to points 'c' and 'd' would be approximately 1.25 lbs. and 3.25 lbs., respectively, in addition to the preload which is constant. Using the previous example of a preload of 2 lbs. per elastic, the additional force of 1.25 lbs. amounts to a force equal to 35 percent of the trunk's actual weight acting to stabilize the pelvis and trunk on the stance phase side just before heel-strike. As the child who weighs 30 lbs. begins to lift himself up a step, the force to stabilize his pelvis and trunk is equal to 48 percent of the weight of his trunk.

Of equal importance, is the fact that the child receives in addition to these increased stabilizing forces, the benefits of reciprocation during such activities. This result is achieved because, although the pelvic panel allows free rotation in an anterior direction, the total force of the elastic upon the buttock on the swing-phase side (which is now actively resisting execssive anterior pelvic rotation, via the panel), transfers to the buttock on the stance phase side during mid-stance. Also, as the child raises the limb in stance-phase off the floor to lift himself up a step, the force transfers back to the buttock on the side of the weight-bearing limb now upon the step above him. Furthermore, he receives additional force with which to further stabilize his trunk, by flexing his trunk forward, which further increases the stretch upon both elastics, a motion that is 'in phase' with the normal sequence of events during the two activities described.

Previously, the problem of shear forces was mentioned as having a strong influence upon the system's design. Figure 5 shows the addition of free hip joints with posterior stops at 180 deg. Regretfully, the addition of hip joints is the best answer, to date, for two more complex orthotic challenges that some myelomeningocele children present.

First, youngsters with lesions above L3 to as high as T10: When anterior musculature that crosses the hip axes is also paralyzed, it is not possible for these children to stop their trunks from involuntarily rotating in a *posterior* direction as well as in the anterior direction. The free hip joints with posterior stops attached to the thoracopelvic unit prevent posterior rotation of the pelvis beyond neutral position about the hip axes, while providing the patient control of anterior pelvic rotation, without locking devices, by virtue of the dynamic pelvic assist. In order that the thoracopelvic unit may be worn separately, when deemed desirable and to make donning of the total system easier for parents and/or the child, a simple quick-release has been devised to disconnect the above-knee orthoses. The enlargment in the upper left of Figure 5 shows the precontoured polypropylene bar being removed from its slotted receptacle. Note how the polypropylene bar is springing back to its precontoured sharp radius which acts as a spring lock when passed through the slot and forced to press against the much more gradual radius of the thorax portion of the unit.

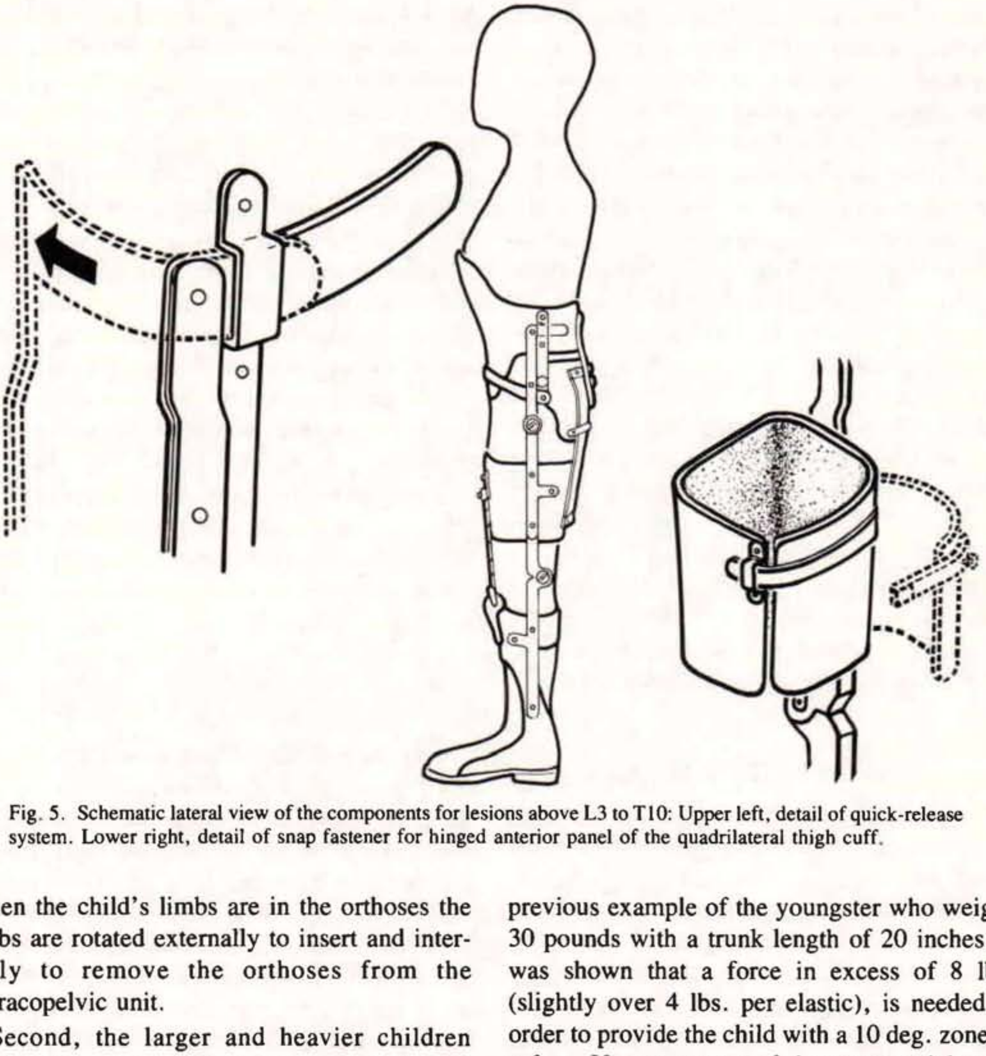

Fig. 5. Schematic lateral view of the components for lesions above L3 to T10: Upper left, detail of quick-release system. Lower right, detail of snap fastener for hinged anterior panel of the quadrilateral thigh cuff.

When the child's limbs are in the orthoses the limbs are rotated externally to insert and internally to remove the orthoses from the thoracopelvic unit.

Second, the larger and heavier children and/or the very young who may need additional help until they learn to use the as yet untried muscles that they do have, and the youngsters who have 5 to 10 degrees of hip flexion contractures *prior* to bracing: All of these cases present a common problem. They need greater amounts of force to keep their trunks upright without resorting to hip locks.

HIP CONTRACTURES

The factors that reflect the limitations upon design, when hip contractures are present, can be demonstrated by referring to Figure 4. In the previous example of the youngster who weighs 30 pounds with a trunk length of 20 inches, it was shown that a force in excess of 8 lbs. (slightly over 4 lbs. per elastic), is needed in order to provide the child with a 10 deg. zone of safety. If a youngster of the same weight and trunk length presents 10 deg. hip contractures, the same force would be required just to prevent the trunk from flexing further forward. However, this would be of little advantage to the child because his trunk balance would remain precarious due to the lack of a 'safety' zone. To provide him with a minimum safety zone of 5 degrees, it would be necessary to preset the force of each elastic to 6.2 lbs. for a total force of pelvic assist of over 12.4 lbs. If a 10 deg. safety zone were attempted, the total pelvic assist force required would be 16.4 pounds. The effect of one characteristic of elastic materials, i.e., an increase in elongation results in an increase in force, cannot be ignored. Since, when seated, the initial setting or preload is known to increase by a minimum factor of 2, a 10 deg. safety zone setting would then be equal to a minimum of 32.8 lbs., well over the child's total weight. Due to this child's size, the length of elastic is limited to the distance from point "a" to point "b". To obtain such high preload settings from short lengths of elastic requires stretching the elastic so close to its limits of elongation that there would be insufficient stretch left to allow for activities such as sitting. There is no choice than but to reduce the preload setting and we must settle for a safety zone of 5 deg. or less. Such a slim margin of safety, especially when the trunk cannot achieve an upright posture, is too difficult for a small child to master. As we begin to understand the magnitude of the forces that these children's muscularly imbalanced bodies must contend with, the importance of dynamic bracing *before* contractures develop becomes apparent.

OVERWEIGHT CHILDREN

Overweight children without hip contractures, present less of a dilemma, yet their excessive weight does require design considerations that relate to shear forces. For example, a child with the same trunk length of 20 in., who weighs 40 lbs., requires a total preload setting for a 5- or 10-deg. safety zone of 6 and 11.1 lbs. respectively. The child with a preload setting for a 10 deg. safety zone, when seated, will have a total force of 22.2 lbs. acting between the thoracopelvic unit and the quadrilateral thigh cuffs. Although the weight of his trunk and the stretch length of the elastics may be sufficient to allow him to sit down easily, a downward force of 22.2 lbs. upon the thoracopelvic unit and an upward force of 11.1 lbs. upon each quadrilateral thigh cuff translates to shear forces upon the insensitive tissue of these areas that are too high to risk. A single lateral upright with a free hip joint, with or without a posterior stop, as the case requires, must be added bilaterally to eliminate the unwanted shearing effect. The free hip joints do not inhibit the function of the dynamic pelvic extension assist.

THE BEGINNERS

The very young, who are neither overweight nor have hip contractures, may require the addition of free hip joints with 180 deg. posterior stops in order to increase their safety zone to the maximum that the elastic materials will allow without risk of introducing unwanted shear forces to insensitive tissue. The hip joints can be removed later because the preload force maintaining the safety zone can be reduced to a smaller range once the child gains control. Also, as the child grows taller and heavier, longer elastics can be used and greater forces are available to assist him without restricting freedom of motion.

A FURTHER REFINEMENT

A quick release to separate the knee joints and "solid-ankle" portions of the KAFO's from the quadrilateral thigh cuffs is now in development, so as to make one system serve as both a day and bedtime orthosis. It is expected that the thoracopelvic unit, with its free hip joints extending to the quadrilateral cuffs, can be utilized as a night orthosis and the KAFO's added for daytime use. This design is intended for the youngster who has 5- to 10-deg. hip contractures *prior* to bracing. The idea is that hip contractures may be reduced during sleep when the dynamic pelvic extension assist would not have to contend with the force of gravity. Also, the corrective force acting about the hip axes would have an additional advantage because the anterior muscles and other tissues being stretched would be relaxed during sleep. Thus, contracted tissue could be placed on stretch around the clock. It is hoped that, with this approach, *time* and *growth*, two factors that have been so destructive in the past, may be brought under control for the myelomeningocele child.

CONSIDERATIONS FOR A VOLUNTARY HIP RELEASE MECHANISM

The preceding discussion may well prompt a question as to why the system does not provide a means for the wearer to voluntarily release the elastic tension of the pelvic assist. While it must be conceded that a greater preload could be applied and thereby increase the range of the safety zone if a release were available to disengage before sitting, the decision *not* to provide a release is a matter of 'tradeoffs'. A release would deny these children valuable automatic assistance and/or control in getting up and down from a chair, play position or fall. Furthermore, as has been discussed, without hip joints, the amount of increase to the preload is limited strictly by the necessity of avoiding unacceptable levels of shear forces. However, the importance of keeping the dynamics of the system fully automatic, without the need for conscious effort on the part of the young children for whom the system is designed, in my view, far outweighs all other considerations.

The enlargement in the lower right hand corner of Figure 5 shows details of the polypropylene snap fastener that secures the hinged anterior panel of the polypropylene quadrilateral cuff. Its purpose is three-fold:

1. To ensure that there is no slackness in the fit of the cuffs, especially in the anterioposterior plane, which would affect the accuracy of feedback being received from the system via the thoracopelvic unit.

2. To stabilize the anterior panels of the cuffs against the strong pull of the knee extension assist since the cuffs serve as the proximal attachment points for these units.

3. For ease of removal and donning and to ensure that the cuffs are worn in the same manner each and every time they are applied, because of their importance to the accuracy of feedback.

Figure 6 demonstrates how the knee extension assist unit is designed to 'program' its pivoting action so that the line of force generated by the elastic shockcord changes from an extension moment about the knee and mechanical axes, through both centers as the knee is flexed and drops below both centers to become an efficient knee *flexor* in the seated position. The advantage to the wearer is an assistive force equal to 15 percent of the weight of the trunk and thighs to aid in both rising from and descending into a chair. The flexion moments stabilize the lower legs, when rising from a chair, until the point when a sufficient amount of the body weight is above them to take over and also it prevents the occurrence of unwanted extension of the lower legs when seated. Figure 7 shows a detailed view of the dynamic knee extension assist unit.

Experience to date with the dynamic orthotic system indicates that it is technically feasible to improve the effectiveness of the orthotic contribution to the overall management of myelomeningocele children. From a biomechanical point of view, the key appears to be *prevention* by introducing dynamic external forces sufficient to maintain balance and control, in a manner that enables innervated muscles to develop to their full potential in both strength and function without causing deformities. It can be demonstrated that a surprisingly low magnitude of force is sufficient to maintain balance about a joint that is muscularly imbalanced and which has a normal range of motion. Also, as has been shown, adding a mild increase to the balancing force across a mobile joint provides a range or 'zone of safety' that allows movement within the safety range without jeopardizing the patient's overall balance. However, even a mild degree of contracture, e.g., about the hip joints, brings kinetic forces into play that add to the dimensions of the problem of maintaining balance with sufficient freedom to perform daily activities. Early application of a dynamic system would seem to suggest the direction toward prevention of musculoskeletal deformities that develop after birth.

SUMMARY

A lightweight, dynamic orthotic system has been developed for the myelomeningocele

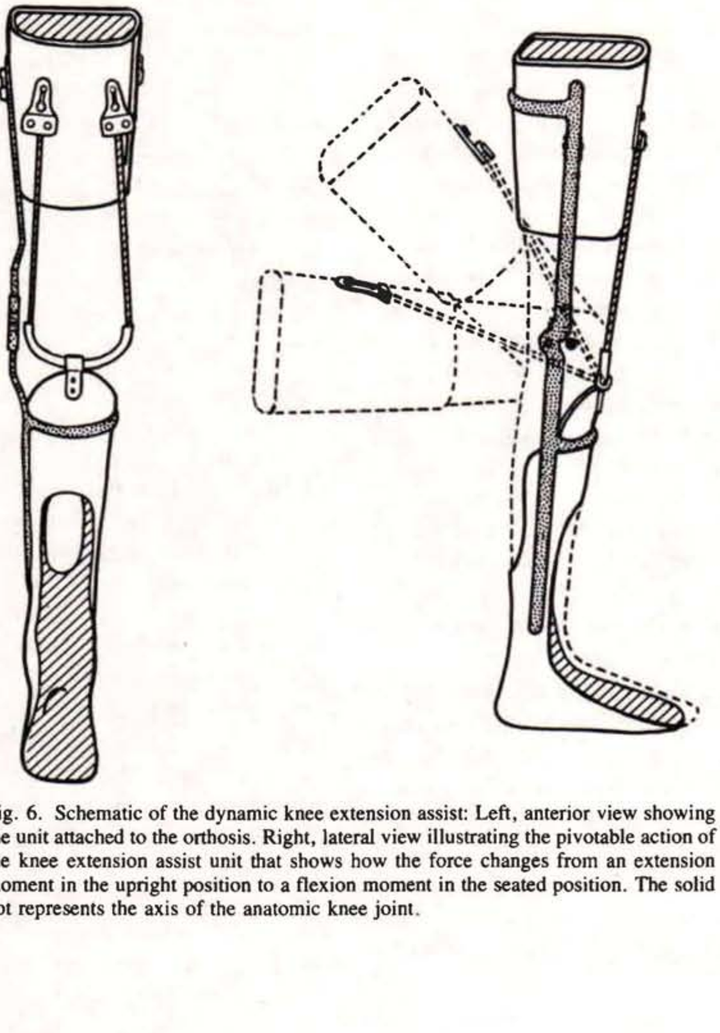

Fig. 6. Schematic of the dynamic knee extension assist: Left, anterior view showing the unit attached to the orthosis. Right, lateral view illustrating the pivotable action of the knee extension assist unit that shows how the force changes from an extension moment in the upright position to a flexion moment in the seated position. The solid dot represents the axis of the anatomic knee joint.

child. The system is modular in design in order that a variety of components can be 'plugged' into it to meet the multiple needs that each level of lesion presents. The system is fully automatic in that it provides the wearer the following functions, without conscious effort on his part: overall body balance parameters with freedom of movement for daily activities; dynamic forces that act reciprocally during gait; assistance and/or control to and from a chair, play position and in climbing stairs; assists pelvic and knee extension without locking devices. The system has additional uses, e.g., the dynamic pelvic extension assist components can be used post-operatively to maintain surgical releases of hip flexion contractures and/or as bedtime orthoses to reduce mild hip contractures during sleep. Also, the knee extension assist components can be used in either manner for knee flexion contractures. The overall design as well as the rationale for its use is described in detail.

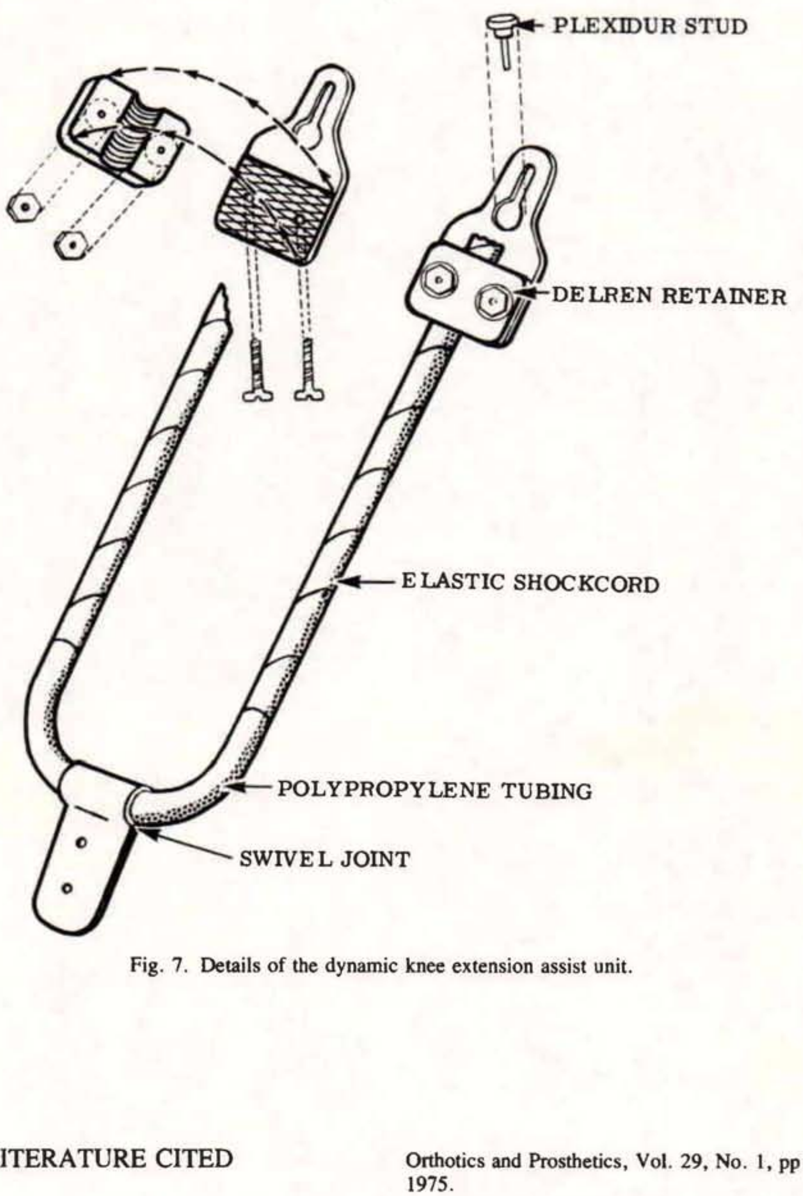

Fig. 7. Details of the dynamic knee extension assist unit.

LITERATURE CITED

1. Blount, W. P., *Bracing for scoliosis*. Orthotics Etcetera, Sidney Licht ed., Elizabeth Licht, New Haven, 1966.

2. Glancy, J. and R. E. Lindseth, *The polypropylene solid-ankle orthosis*. Orthotics and Prosthetics, Vol. 26, No. 1, pp 12–26, March, 1972.

3. Glancy, J., *Dynamic control of abnormal M-L motion of the os calcis: The cushion heel wedge – a possible solution*. Orthotics and Prosthetics, Vol. 27, No. 1, pp 28–33, March, 1973.

4. Glancy, J. and R. E. Lindseth, *A dynamic orthotic system to assist pelvic extension: A preliminary report*, Orthotics and Prosthetics, Vol. 29, No. 1, pp 3–9, March, 1975.

5. Lindseth, R. E. and J. Glancy, *Polypropylene lower-extremity braces for paraplegia due to myelomeningocele*, The Journal of Bone and Joint Surgery, Vol. 56-A, No. 3, pp 556–563, April, 1974.

6. Ontario Crippled Children's Centre, *Orthotic research project, annual report*, December 15, 1969.

7. Rose, G. K. and J. T. Henshaw, *A swivel walker for paraplegics: medical and technical considerations*. Biomechanical Engineering, Vol. 7, No. 9, pp 420–425, October, 1972.

8. Saltiel, J., *A one-piece laminated knee locking short leg brace*. Orthotics and Prosthetics, Vol. 23, No. 2, June, 1969.