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ELASTIC MATERIALS AS A SOURCE OF EXTERNAL POWER IN ORTHOTICS
A PRELIMINARY REPORT[1]

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Sources of external power for use in orthotics have been limited for many years. Unfortunately, it appears that an 'ideal' source will not be available for a generation or two. What practical alternatives can be offered to a large proportion of today's handicapped population and those of the next generation? The use of elastic materials as a source of external power seems to hold promise as a practical alternative for the interim.

Elastic materials, such as shock cord, while limited in meeting requirements in any sense of the ideal, nevertheless can store and deliver surprising amounts of dependable energy. Furthermore, the material can be adjusted easily, is noiseless, lightweight, commercially available, durable and inexpensive.

The ability to store energy is the most important characteristic of elastic materials for orthotic use. They can be made to produce a continuous force that can be put to work in a manner that introduces a new element to spinal and lower-limb orthotics: *dynamics*. The fact that these dynamic forces can be varied in magnitude and made to serve a variety of purposes allows a versatility that portends exciting possibilities.

Some examples of the possibilities afforded by elastic materials in orthotics are:

*A small dynamic force across the axis of an anatomic joint can provide constant stretching of soft tissue to prevent the development of contractures without inhibiting useful motion.

*A preset dynamic force can be applied across an anatomic joint that has imbalanced musculature crossing it in a manner that simulates the missing opposing 'tone' to prevent involuntary swaying, thus aiding balance.

*A single, preset dynamic force can be introduced so that it can change from an extension moment, through dead center, and then to a flexion moment, or the reverse, in a 'programmed' sequence, as angular changes occur in the anatomic joint to which the force is being applied. Thus, a single force can be used to oppose or assist motion in *two directions* in the same plane.

*A present dynamic force can be increased gradually by allowing its fixed ends to be stretched further apart as an anatomic joint moves. The resulting additional force can be used to act as a 'guide' or control for motion activated by muscle power.

*An easily adjusted dynamic force, in the form of an extension moment, can be used postoperatively to maintain and preserve surgical releases of flexion contractures of the hips or knees.

*When the weight of a body segment is not too great, available elastic materials can substitute for absent muscle power and move the segment through space, especially when gravity is used to assist the desired motion.

*In instances where elastic materials are being used to activate segmental motion, adjustment of the dynamic force can also affect the velocity of a segment's movement.

*The dynamic forces that available elastic materials offer are of sufficient magnitude to be of exceptional usefulness to growing children, primarily because of children's relatively lightweight bodies, in the prevention of musculoskeletal deformities that develop after birth.

By providing dymanic resistive force, elastic materials offer stability and control without inhibiting useful motion. The current state-of-the-art is limited almost exclusively to static control. As a consequence, useful motion is constantly—albeit inadvertently—being subjected to interference. It now appears that longterm protection for the musculoskeletal system, without sacrificing freedom of mobility, is orthotically possible.

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A SPECIFIC CLINICAL APPLICATION— A PRELIMINARY REPORT

A new look at elastic materials as a source of external power was undertaken as part of an ongoing effort to develop a practical orthotic system of ambulation for paraplegics that would enable them to walk with greater ease and confidence than present-day techniques allow. A practical means of ambulation, with a good deal less dependency upon a wheelchair, continues to be our ultimate goal, and with this in mind, it was evident that first priority must be given to finding an easier way for the paraplegic patient to rise from and return to a wheelchair. This portion on the use of elastic materials as a source of external power in orthotics will limit the *specifics* of their use to our experiences in attacking this multifaceted problem.

The most obvious improvement to the current technique for rising and sitting seems to be a system that will enable the paraplegic patient to get to his feet in a quite conventional manner, starting with the knees flexed in a normal seated position. The more the overall problem was studied, the more the "obvious" seemed to be the most practical.

For a paraplegic to rise from his wheelchair in a smoother, less exhausting way it is necessary to compensate for the lack of sufficient height provided by the arm rests (Fig. 1). Folding "crutches" (Fig. 2) that provide the extra height were designed and installed on a wheelchair as permanent components. The pinlocks on the front casters and the rear wheel locks are engaged before rising to stabilize the wheelchair.

Inability to prevent 'out-of-phase' anterior rotation of his pelvis when attempting to leave the chair while facing forward is another factor that prevents the paraplegic in achieving a standing position easily. The latissimus dorsi muscles used in raising the trunk inadvertantly cause anterior rotation of the pelvis because their insertions are located on the spinous process of the lower thoracic and lumbar vertebrae. Since the antagonists that normally prevent such out-of-phase pelvic rotation are paralyzed, an increase in forward rotation and an accompanying increase in lordosis are under no restraints other than the range limitations of the joints of the vertebrae and hips.

To stabilize the pelvis and lumbar spine against the forces of the latissimus dorsi and gravity while

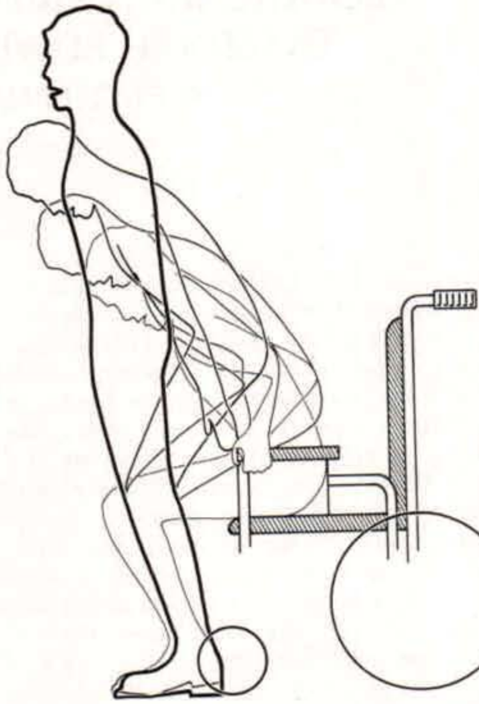

Fig. 1. A schematic view demonstrating the lack of sufficient height of supports for the paraplegic's arms in order for him to lift himself to his feet from the seated position.

Fig. 2. Prototype folding wheelchair crutches: The crutches are designed to be permanent components. Note that the hand grips are turned downward, below the level of the armrests, when not in use.

rising and returning to the wheelchair a modification of a system reported earlier (*1*) was used (Fig. 3). Since paraplegic patients must wear bilateral knee-ankle-foot orthoses (KAFO's), the posterior elastic panels that are attached to the movable pelvic portion of the polypropylene thoracopelvic unit are attached distally to the thigh cuffs of the KAFO's. This arrangement permits the initial, or 'preload', extensor moment to be of greater magnitude because the thigh cuffs are fixed to the uprights of the KAFO's and therefore an excessive buildup of shear forces on the surface of the thighs (of some concern when the thigh cuffs are not attached to fixed points) no longer need be a consideration. However, the characteristics of various elastic materials *do* place a limit on the magnitude of the preload, in that the setting must allow sufficient additional stretching to permit enough excursion of the elastic panels for the patient to sit.

Fig. 3. Closeup of the modified polypropylene thoracopelvic unit with dynamic pelvic extension elastic assists: The elastic is 2" wide and is doubled over nylon rollers attached to the distal ends of the polypropylene quadrilateral thigh cuffs to ensure an even pull throughout. Note the plexidur studs used for the proximal attachments. To don, the patient passes the elastics through the nylon rollers and slips the ends, with their slotted polypropylene pieces, over the studs while lying on her side.

The extensor moment generated by the bilateral elastic panels provides the paraplegic patient with the following functions:

A. While rising from a wheelchair:

1. Prevention of the lumbar region from going into excessive lordosis;

2. A resistive force to out-of-phase, anterior rotation of the pelvis,

3. Substantial assistance to the patient's arms when lifting his body.

B. When standing:

1. Additional stability by preventing involuntary swaying of the trunk in an anterior direction about the hips' axes. Thus a 'back-up' safety feature is added to the technique of 'hanging' upon the iliofemoral ligaments in order to maintain anteroposterior balance.

2. Sensory feedback provided by the intimate fit of the thorax portion of the polypropylene thoracopelvic unit because its upper border extends 2 inches above the level of sensory deficit. Any tendency for the trunk to 'jackknife' is felt in the form of increased pressure as it begins to occur, and the motion is readily checked by his arms and crutches before it is beyond control. Related to this feature is the fact that as the trunk rotates forward, the extensor moment (simultaneously generated by both elastic panels), increases proportionally to angular changes occurring in the hip joints and thereby slows down the trunk's forward rotation. The net result is additional time for the patient to react to the 'reading' he is receiving via the thoracopelvic unit.

C. While lowering the trunk into the wheelchair:

1. An increase in magnitude of the extending moment about the hip throughout the descent. The out-of-phase rotation of the lumbar spine and pelvis is being resisted by the extensor moment generated by the bilateral elastic panels.

2. Sensory feedback that enhances control of the rate of descent is received via pressure on the thoracopelvic unit.

The folding crutches and the polypropylene thoracopelvic unit with its pelvic extension assist permitted the patient to rise to a standing position, but, though up, he could not extend his knees fully so as to engage the knee locks of the KAFO's. Consequently, his knees buckled at each attempt to bear weight upon his lower limbs. It was clear that further progress would not be possible without a mechanism capable of automatically extending the knees and locking them in extension.

It was felt that the mechanism would have to extend the tibii fully and automatically, because both arms of the patient are needed for support.

The external power needed would be substantial. Because the weight of the lower limb (including the orthosis), with assistance from gravity, does not produce full extension at the knee joints of a paraplegic patient while his arms support his extended body, a resistive force must be present. The weight of each lower-limb below the knee is calculated to be 1/15 of the body's weight (*2*). The weight of the below-knee portion of each KAFO is about 1 1/2 pounds (.68 kg). A reinforced shoe of the Hush Puppy (R) variety weighs about 1 pound (0.45 kg). For example, the total weight involved, per lower-limb, for a patient who weighs 100 pounds would be about 9.2 pounds (4.1 kg). Ruling out the presence of spasticity in the hamstrings and/or gastrocnemius muscles, or the presence of fixed contractures in either, a 'guesstimate' must be made of the magnitude of force needed to overcome the resistance. Because we do not have information as to the amount of resistive force we are dealing with, clinically, it is necessary to set the initial extension force somewhat arbitrarily. To complete the example given, we chose 6.5 pounds (2.95 kg) (approximately the weight of the lower limb below the knee) as the minimum force that the knee extension assist unit would have to exert upon each lower limb to achieve full extension.

Because of the power limits of available elastic materials to supply an external extension force to the knee joints, an attempt was made to determine just how many *specific* functions that are credited to the quadriceps muscle group would have to be mechanically produced to accomplish the task. An analysis was made of a number of the many functions that the quadriceps muscles normally perform under a variety of conditions. All but four of the functions studied were rejected as either irrelevant to the immediate needs of paraplegic patients or, if a substitute could be delivered, would be impossible for them to manage because of the general limitations of current orthotic development. The four functions of the quadriceps muscles which were felt to be crucial to a solution and therefore would somehow have to be mechanically produced are:

1. The force they generate to resist and/or prevent knee flexion whenever the body's CG moves behind the knee axes when standing. (This force varies from minimal—when checking normal involuntary swaying in the anteriioposterior plane—to much greater magnitudes when serving as a control mechanism for any activity that involves flexing of the knees, during which the body's CG is behind the feet.)

2. Their crucial contribution to the complex control forces that enable the normal individual to lower his trunk into a chair at varying rates of speed.

3. Their active contribution to the lifting of the body from a seated position by extending the thighs about the knee axes.

4. Their ability to extend the lower leg about the knee axis, at varying rates of speed, when the lower leg is not bearing weight.

The rate of acceleration of extension of the lower leg must be regulated, if possible. This is important in order to keep to an absolute minimum the time that the patient must support his full weight upon his arms, once he has achieved full extension of his hips and trunk at the completion of his rise from the wheelchair.

A dynamic, automatic knee extension assist that facilitates full extension of the lower leg and allows the automatic spring knee lock of a KAFO to engage is shown in Figure 4. The illustration also shows the type of KAFO that is used with the new system. A stainless steel Becker automatic springlocking off-set knee joint is modified to enable the patient to put the spring-lock in the disengaged position when standing, a prerequisite to sitting. When he wishes to sit down, the off-set knee joints prevent the patient's knees from buckling as he flips the handles (which are connected to the locking cables) to the disengaged position, one leg at a time. Thus, he has a hand free to support himself on either of the extended crutches attached to the wheelchair. As the patient leans against the anterior panels of the thigh cuffs and the anterior panels of the solid-ankle shells below, (*3,4*), the combination of the mechanical knee center being 1 inch behind the an-

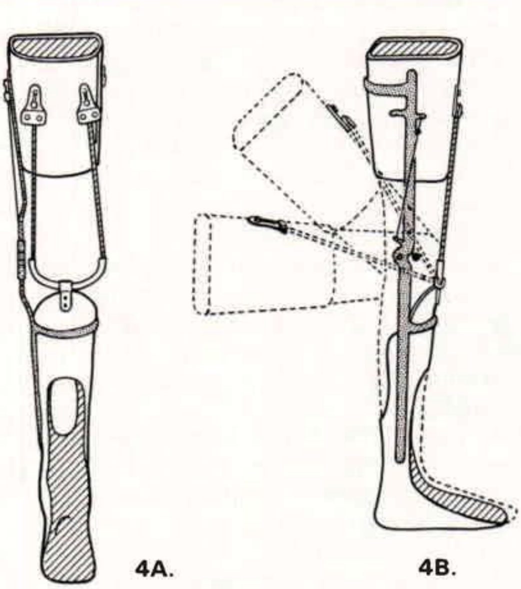

Fig. 4. The dynamic knee extension assist: A. Anterior view showing the unit attached to a KAFO. B. Lateral view illustrating the pivotable action of the knee extension assist unit that shows how the force changes from an extension moment in the upright position to a flexion moment in the seated position. Note the cosmesis of the unit—no part protrudes above the flexed knee. (The solid dot represents the axis of the anatomic knee joint.)

atomic knee center (*5*) and the dynamic extension moment being generated by the looped elastic shock cord, results in a remarkably stable pair of *unlocked* knees.

As the patient lowers himself into the wheelchair, the combined bilateral knee extension force generated by the elastic shock cords offers a substantial resistive force that, in turn, relieves an appreciable amount of the weight that the arms must support. For example, we return to the sample case of the patient who weighs 100 pounds (45 kg). In order to extend each lower limb, it was estimated that each would have to have a dynamic extension force of 6.5 pounds (2.95 kg). Therefore, the total force resisting the weight of the body as it is lowered into the wheelchair would be 13 pounds (5.9 kg)—the sum of both shock cord assists. Since both lower portions of the limbs are resting upon the floor and no motion is occurring below the knee axes, their weight need not be considered as part of the mass being supported by the patient's arms. In the example, the weight of the two lower limbs (exclusive of the KAFO's), totals 13.4 pounds (6 kg). Subtracting 13.4 pounds (6 kg) from the total body weight leaves 86.6 pounds (39.3 kg)—the actual weight that the patient's arms are supporting (exclusive of the thigh and trunk portions of the orthotic system). By dividing the actual weight being supported, 86.6 pounds (39.3 kg), into the total resistive force of 13 pounds (5.9 kg), we find that the arms are being relieved of approximately 15% of their burden. Conversely, the arms are provided the same assistance (about 15%) when lifting the body up from the wheelchair.

Figures 4, 5, and 6 show some details of the components of the system. The knee extension assist unit weighs about 3 oz. (105 gms). The semicircular polypropylene tubing is allowed to swivel so that the force of the shock cord provides an extensor moment in the extended position and a flexion moment in the seated position.

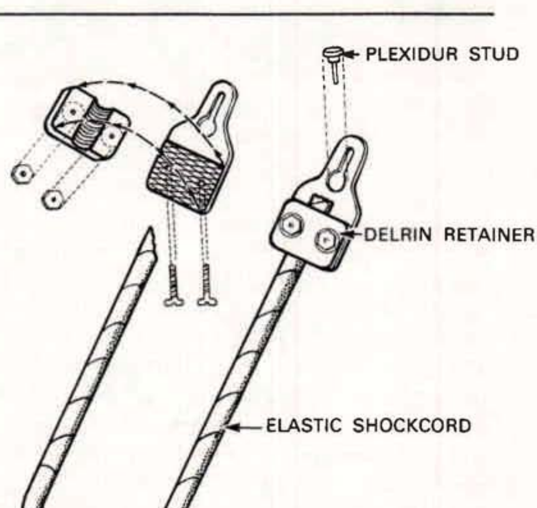

Fig. 5. The dynamic knee extension assist unit: Detail of the components.

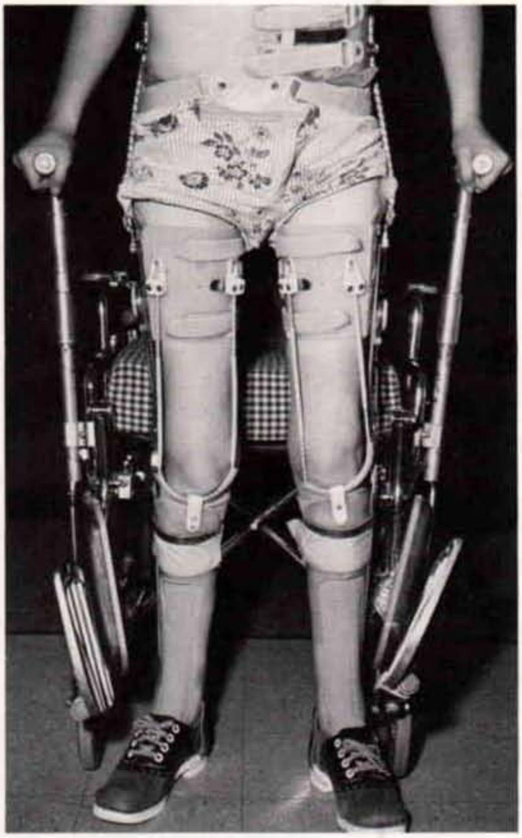

Fig. 6. The dynamic knee extension assist unit; Close-up showing the units on a patient's KAFO's.

Figure 7 shows a patient rising from her chair while using the folding crutches, the thoraco-pelvic unit with extension assists, the knee extension assists, and the dual locking, single, lateral upright KAFO's with offset knee joints. The patient is a 16-year-old, T10 level, traumatic paraplegic with a surgical fusion of L2-3 and with hip flexion contractures of 15 degrees on the right and 10 degrees on the left. It is 3 1/2 years since her accident. The patient weighs 115 pounds. Note the degree of pelvic control and how the dynamic knee extension assists change from a flexion to an extension moment as she rises.

ADVANTAGES TO THE PARAPLEGIC PATIENT

As the patient rises from his wheelchair the shock cord provides a gradually diminishing flexion moment for approximately the first 25–30 deg. of his ascent. This insures that his lower legs will remain under him until he gets enough of his weight over them to the point that they will not slide out from under him as he is rising.

For the remaining 60 deg. to full extension the shock cord provides an extension moment of increasing magnitude, with the minimum at approximately the 35 deg. point. The axis of the offset knee joints being 1 inch behind the anatomic knee axes add to the magnitude of the extensor moment as the patent's thighs press against the anterior panels of the quadrilateral cuffs. The pressure of the thighs is transferred down the uprights to the mechanical axes and increases the radius perpendicular to the extension force generated by the shock cord. The net effect is additional assistance to the lifting of the trunk and greater stability of the lower legs as the trunk is lifted up and over them.

When the patient is seated the force generated by the shock cord is constant. Therefore, the femurs are placed in compression. The pelvis is pressed against the femoral heads by the force generated by the elastic panels of the thoraco-pelvic unit and the condyles are being pressed toward the pelvis by the force of the shock cords. These two forces, combined with the thoraco-pelvic unit, provide trunk stability while seated. It should be noted that when the offset knee joints are fully extended the force of the shock cords is absorbed by the uprights and does not increase the vertical load on the lower limbs when standing.

SUMMARY

A preliminary report is presented on the potential benefits of elastic materials as a source of external power for orthotic use in general. In particular, a description of the experiences, to date, of an ongoing effort to devise a practical orthotic system of ambulation for paraplegics demonstrates the application of dynamic forces for two specific activities that were given first priority; getting up from and sitting down into a wheelchair. An orthotic system has been devised for these two activities that consists of four newly-developed components:

*A pair of folding crutches that are permanently attached to the wheelchair;

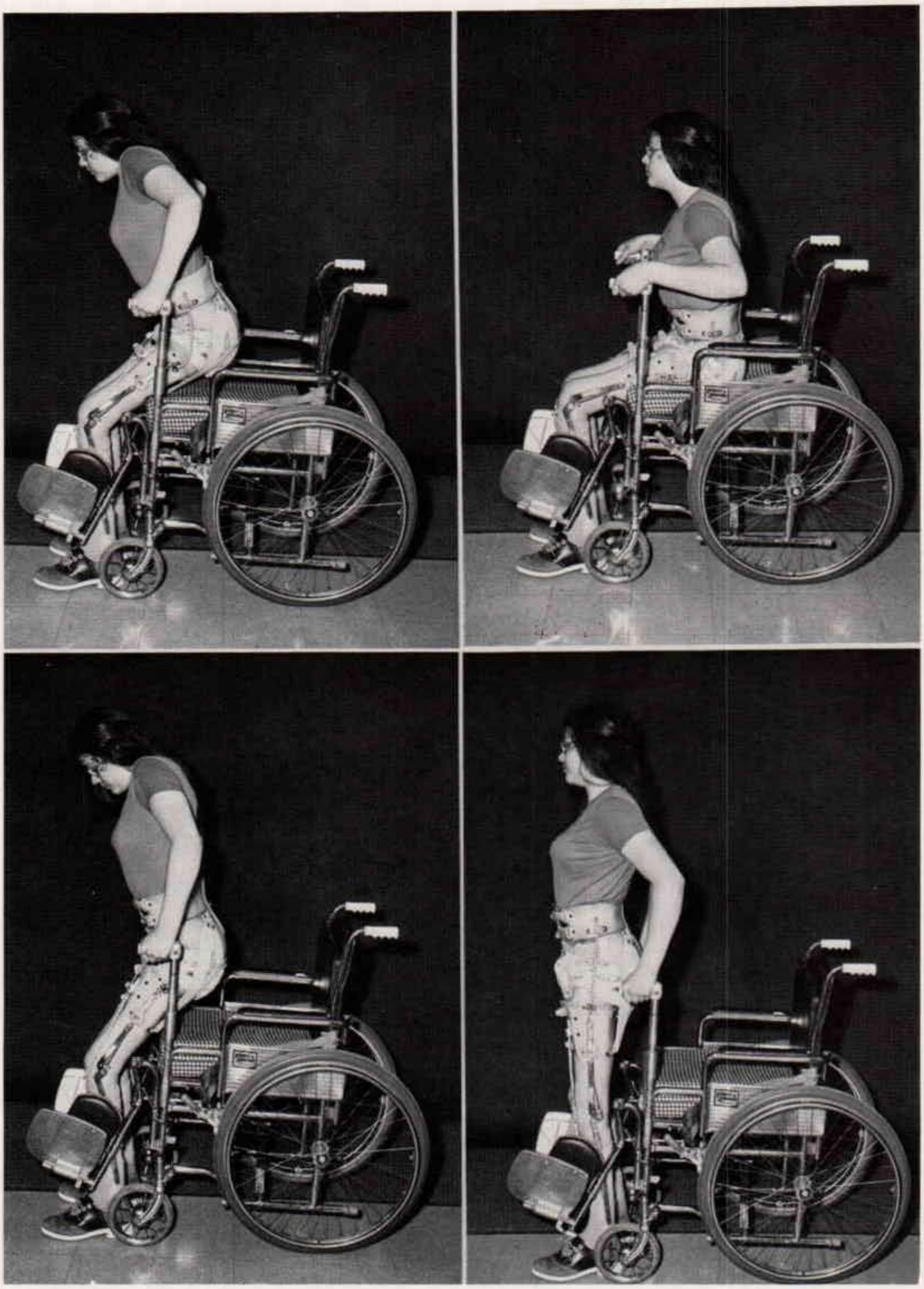

Fig. 7. Patient rising from specially equipped wheelchair.

*A modified use of the polypropylene thoracopelvic unit with dynamic pelvic extension assist that was previously developed for low-level myelomeningocele patients.

*A dynamic knee extension assist unit that automatically extends the lower limbs before the patient lowers himself to the floor, after having achieved an upright position;

*The use of a single lateral, spring-locking, offset knee joint for the patient's KAFO's. The spring-locks are modified so that they can be locked individually in the disengaged position in preparation for sitting down into a wheelchair.

In Figure 7, left, the patient is shown releasing a knee-lock and locking it in the disengaged position in preparation for sitting. The shock cord at lateral midline, with its proximal attachment to the thorax portion of the thoracopelvic unit and its distal attachment to the lateral panel of quadrilateral thigh cuffs, provides mediolateral stability of the trunk without the use of metal bars. In figure 8, right, the patient is shown lowering herself into the specially equipped wheelchair.

The rationale for these developments is described in detail. It should be emphasized that these developments are the results of an *ongoing* project. Their preliminary use is reported at this early stage because it is believed that their apparent potential value can readily be clinically tested and evaluated by others and, if accepted, speed up their refinement and further define applications for general orthotic use.

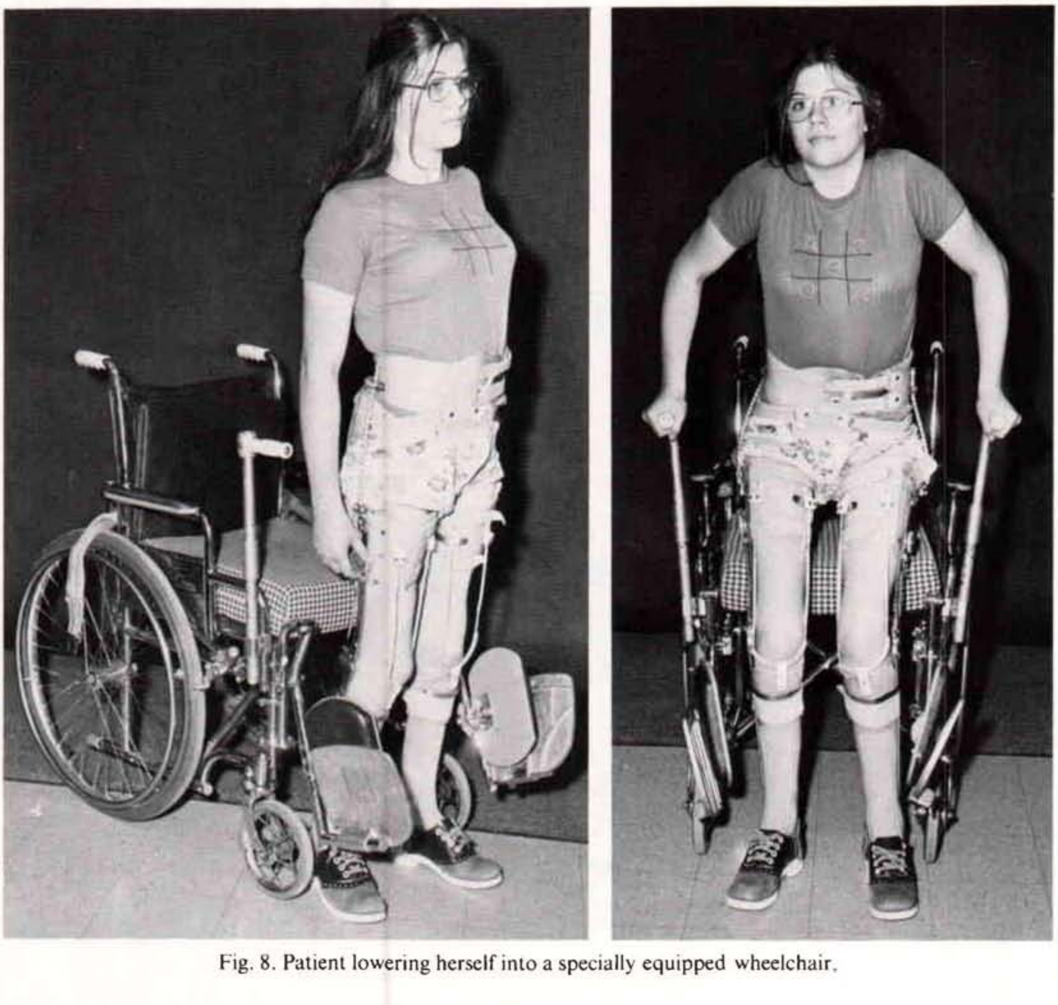

Fig. 8. Patient lowering herself into a specially equipped wheelchair.

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