Abstracts of Recent Articles

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Ten normal adults performed the swing-through gait using Canadian crutches while having the left leg suspended. They walked at freely chosen speeds on a 9-meter-long walkway mounted over a force plate.

Vertical ground reaction forces displayed peaks at early and late stance, as in normal walking. Crutches, however, caused an average 16 percent increase in vertical reaction force on the limb. Horizontal forces also duplicated the normal pattern, initially opposite to the walking direction, then reversing to the same direction at push-off. Crutch use increased the force on the limb from the normal 0.15 to 0.35 body weight. Very little horizontal ground reaction force was recorded on the crutches. Thus, major propulsive effort comes from the lower limb rather than the crutches. Height and age of subjects did not affect the results significantly, although men achieved somewhat greater vertical peaks. Most participants placed more load on the right crutch; all were right dominant.

Crutch walking hastened the point in the walking cycle when one leg, unaided by crutches or the other limb, supported the body. Normally, this point is 50 percent of the cycle; with crutches it was 44 percent.

Swing-through gait with one limb increased the horizontal and vertical ground reaction forces which could jeopardize patients with skeletal abnormalities, especially if they could not produce the steady gait which the experimental subjects demonstrated.


Prehension has dynamic and static aspects. The static phases of normal grasp have been analyzed. Conventional classification of grasp as cylindrical, spherical, palmar, lateral, and tip describes only part of prehension, and does not define the grasp of a tool. Prehension may not be determined by the shape of the object, but by the purpose of the action, thus the use of power and precision grips. Other categorizations include variables such as grip components and finger position.

The new classification does not consider specific activities or objects, but is premised on the concept that common patterns exist in finger use and that normal adults use the pattern required by the object’s shape and function, previous habit, and chance.

Ninety-eight objects were held by seven normal adults. The hand was photographed in five directions. Then the object was smeared with ink and again held. The contact areas on the hand were photographed from several directions. Classification of fourteen patterns in four major categories emerged. Power grip includes five patterns; a wide area of the hand, including the palm, contacts the object. Contact is almost entirely on the volar side. The second category is intermediate grip, and has four variations; the palm does not contact the object. The precision grip category has four subclasses; the object is held between the volar aspects of the finger and the thumb pulp. The fourth category is grip involving no thumb, suitable for small light objects held between adjacent fingers. Most objects were fitted into one of these four categories. The remainder involve an intermediate pattern. Subjects grasped some articles using different patterns within a major category or within different categories. The classification can be the basis for describing static prehension by disabled hands; obtaining as many patterns as possible could be a training goal.


Phantom pain is felt as emanating from the amputated portion of a limb. The relation of phantom sensation to phantom pain is unknown. Stump pain is different. Between 0.4 and 50 percent of amputees claim to suffer continual discomfort. The literature review identified 43 treatments purporting to ameliorate phantom pain.

Treatments which depend on physical alterations of the signals from the stump to the brain include supplementation of signals, such as stump percussion, transcutaneous electrical stimulation, relaxation / biofeedback training, phantom / stump exercises, warm baths and massage, ultrasound, dorsal column stimulation, electrical brain stimulation, and acupuncture. Physical alteration of signals may be accomplished by removing some incoming signals by stump surgery, nerve strangulation above the stump end, Novocain (IV) at the stump, procaine blocks, trigger point elimination, Propranolol, sympathectomy chordotomy, rhizotomy, peridural blockage, brachial plexus blocks and lesions, high cervical tractotomy, thalamic surgery, ablation of the post central sensory gyrus, and Carbamazepine.
Another mode of treatment involves alteration of interpretation of pain signals. This may be achieved by physical alteration of interpretive centers of the brain by prefrontal lobotomy and leucotomy, post central topocotomy, midbrain lesions, subcortical parietal lobotomy, anterior cingulate lesions, and electroshock. Chemical alteration of pain stimuli usually attempts to disrupt the anxiety-pain-depression relationship; Chlorpromazine, Propranolol, Carbamazepine, Benperidol, Reserpine, and LSD have been reported as being successful, at least for some patients.

Behavioral alteration of interpretation may be effected by psychotherapy, hypnosis, phantom limb exercises, relaxation and biofeedback, distraction training, and behavior modification. Some methods of altering pain interpretation have no specific mechanism or site, such as Vitamin B12 therapy, radiation, magnetotherapy, and electrosleep techniques.

Most treatments are ineffective after a one year follow-up, although most worked for at least one-third of their recipients temporarily. The placebo effect of treatment is considered. A survey of current treatment techniques is being compiled, which should supplement the literature review.

Double-Stopped Ankle-Foot Orthosis in Flaccid Peroneal and Tibial Paralysis: Evaluation of Function: Justus F. Lehmann, Michael J. Ko, and Barbara J. deLateur (Department of Rehabilitation Medicine, University of Washington School of Medicine, Seattle) Archives of Physical Medicine and Rehabilitation 61:536–541, November 1980.

Eight normal young adults were evaluated while walking along a 100 foot walkway wearing a metal AFO with adjustable anterior and posterior pin stops at the ankle, set at 5 degrees dorsiflexion. Each subject then walked wearing the AFO after having a peroneal nerve block and, finally, with the addition of a tibial nerve block.

Heel strike was not much affected by the blocks. Midstance phase, however, was significantly prolonged by both the peroneal and the combined blocks. This lengthening may be related to reduction in the rate of change with which the location of the vertical ground reaction force moves forward with respect to the ankle. The block prevented the normal quick heal rise and rapid forward motion of the ground reaction, normally achieved by the calf muscles. Calf paralysis forced the wearer to pivot over the end of the sole plate, which is located much farther forward of the ankle. Blocking also prevented normal activity of the toes in push-off. Push-off was briefer when both nerves were paralyzed. Swing phase was not altered.

The rate of change in vertical ground reaction force and maximum forward excursion was not affected significantly by paralysis. Vertical force push-off peak with combined blocks was much less, attributed to loss of active push-off. Forward shear was slightly more. Medialateral shears did not change appreciably. During heel-strike, the plantarfexion moment tending to bend the brace increased with both blocking conditions, because peroneal paralysis prevented active contraction of the dorsiflexors which ordinarily unload the brace structure. The combined paralysis caused a major increase in the dorsiflexion moment at push-off, because the orthosis, rather than the planter flexors, was preventing dorsiflexion.


Six normal subjects, four unilateral below-knee amputees, and two postpolio patients with leg length discrepancy were investigated while using axillary crutches on an 11-meter walkway. They walked at very slow (0.56 meters per second), slow (0.84), medium (1.12) and fast (1.4) speeds, for 5 minutes at each speed. Heart rate was counted and expired air analyzed.

Overall, disabled subjects expended an average of 34 percent more energy than normal persons. Crutch users required 80 percent more energy than controls when walking very slowly, but used only 8 percent more energy than controls at the fastest speed, perhaps because of greater ease in maintaining body balance and walking rhythm. Heart rates were much higher for patients than for normal persons at the three slowest speeds, but were approximately the same for all participants at the fastest speed. Heart rate, which bears a linear relationship to energy expenditure during steady state work, is a clinically convenient indicator of metabolic cost. The optimum walking speed for all 12 subjects was 1.12 meters per second; at this pace, persons with lower limb disability expended 30 percent more energy than normal subjects.


Lower limb support is necessary in normal walking, jogging, and many pathological gaits. The principle of support is the algebraic summation of all extensor moments at the ankle, knee, and hip, which must be markedly positive during stance phase. This principle accounts for the presence of knee flexor activity during stance without knee collapse. Buckling would require simultaneous collapse at the hip and ankle. When one of the three lower-limb joints does not contribute to lower-limb support, the other joints compensate for the non-contributing joint. Kinematic and forceplate data from 30 subjects, of whom 9 had lower limb disorders (hemiplegia, amputation, and knee replacement) were analyzed.

A basic support curve was demonstrated with double peaks similar to the normal ground reaction curve. The support varied very little among the normal and disabled subjects. For example, an elderly woman with knee replacement walked with knee flexion throughout stance; she compensated with a large hip extensor moment and greater than normal plantar flexor ankle moment. Joggers revealed strong knee extensor activity with normal ankle and strong hip flexor activity.

Gait analysis should therefore examine the total limb, rather than focussing on a particular joint, since one joint compensates for lack of support at another joint, probably on a neurophysiological basis.

Investigation of the resistive force, moment, and torque data in the shoulder complex on three live subjects was conducted while they were in a restraint system. The sternoclavicular joint is saddle-shaped with concave and convex curvatures. The clavicle is separated from the sternum by a meniscus. The joint displays three degrees of freedom while not featuring mating bones, is surrounded by a fibrous capsule, and has three degrees of freedom. The glenohumeral joint is a ball-and-socket design. The glenoid cavity has a smaller surface than the humeral head; its inferior capsule is loose.

In addition to ligaments, the shoulder complex is maintained by three groups of muscles: scapulohumeral (rotator cuff, deltoid, and teres major); axioascapular, from the torso (trapezius, rhomboids, serratus anterior, and levator scapulae); and axiohumeral (pectoralis and latissimus dorsi).

Quantifying the resistive force, moment and torque values required knowing the relative motion of the arm with respect to the torso and the values of the three-dimensional force and moment vectors responsible for moving the arm. The realistic shoulder model which emerged from this investigation allows computation of the rate of change of the slopes of curves representing torques and moments near the boundaries of the test regions. In such a manner, one can evaluate the distribution of the passive resistive force and moments on the bony and soft structures of the region. Biological materials, especially soft tissues, display nonlinear viscoelasticity and may be assumed to be modeled similar to the Kelvin viscoelastic material, with additive elastic and viscous forces.

Future research should therefore emphasize determination of the velocity-dependent viscous component of the reaction force and moments at the shoulder so that an even more realistic model can be developed.


The human knee incorporates a complex geometric ligamentous structure which has been explored for its three-dimensional load transmission characteristics. Dynamometers which could measure all orthogonal components of applied force and moment were attached to the femur and tibia of four in vitro human knee specimens. They were loaded to near failure. Anterior-posterior and medial-lateral tibial displacements were applied; for the latter, knees were tested at full extension and 30 degrees flexion; for the former, knees were extended.

Lateral load of 650 N was carried primarily by the posterior cruciate ligament (36 percent) augmented by the medial collateral (22 percent), posterior oblique (21 percent) and less than 10 percent by other ligaments. A medial load of 690 N was sustained mostly by the anterior cruciate ligament (52 percent) and lateral collateral ligament and popliteal tendon (30 percent). Another knee exhibited greater restraint by the medial collateral ligament to lateral displacement while the anterior cruciate protected most against medial displacement. Results were similar when knees were loaded while flexed. That the cruciates restrict both medial and lateral movement is logical since their geometry dictates that the posterior cruciate tense when the tibia moves laterally and that ligament again tightens at the extreme of medial displacement as it wraps around the medial femoral condyle. The medial collateral ligament caused the medial meniscus to move during medial tibial displacement to reduce resistance caused by the meniscus. Without the medial ligament, the meniscus resisted medial tibial displacement.

The cruciates are the primary load-bearing elements for anterior and posterior tibial displacement; the anterior cruciate carried 87 percent of the load during anterior tibial movement and the posterior ligament carried 93 percent of the load during posterior movement in one specimen tested.
Ground Reaction Forces in Distance Running: Peter R. Cavanagh and Mario A. Lafortune (Biomechanics Laboratory, Pennsylvania State University, University Park, Pennsylvania) Journal of Biomechanics 13:397–406, 1980.

Ground reaction forces, changes in the center of pressure distribution, and changes in velocity of the center of mass were documented as 17 young adults ran on a force platform wearing Etonic Km 506 running shoes; they ran at a steady speed of 6 minutes per mile. The surface of the platform was covered with a foam surface. The imprint of the chalked sole so that foot placement could be measured. All subjects contacted the floor initially on the posterior two-thirds of the shoe. Most were rearfoot strikers, having an average placement angle of 10 degrees abduction. Pressure was initially at the rear lateral border of the shoe and moved rapidly toward the midline, then anteriorly. Many had centers of pressure outside the shoe outline because placement was preceded by scuffing when forces greater than threshold exist between foot and floor; many running and walking gaits have a different placement angle of 10 degrees abduction. Midfoot strikers made contact at 50 percent of shoe length with the foot angled 5 degrees. Their centers of pressure migrated posteriorly and medially, then more anteriorly; they had triple the peak-to-peak amplitude of mediolateral force as compared with rearfoot strikers. Rearfoot strikers had two anterioposterior force peaks, while midfoot strikers produced vertical force component curves showing a different pressure outside the shoe outline because placement was preceded by scuffing when forces greater than threshold exist between foot and floor; many running and walking gaits have a different placement angle of 10 degrees abduction. Midfoot strikers made contact at 50 percent of shoe length with the foot angled 5 degrees. Their centers of pressure migrated posteriorly and medially, then more anteriorly; they had triple the peak-to-peak amplitude of mediolateral force as compared with rearfoot strikers. Rearfoot strikers had two anterioposterior force peaks, while midfoot strikers produced vertical force component curves showing merging with steady force increase in the midfoot. The results are specific to the conditions of footwear, ground surface, running speed, and gradient. Only the right foot was studied.

Runners do not strike on the heel, rather the foot is supinated and the hip adducted, so initial contact is on the lateral border of the shoe. Changes in anteroposterior component of velocity were similar among all runners. The range of peak values for the vertical component of ground force was considerable; some ran at the same speed while exerting forces 30 percent lower than others; peak force is negatively correlated with contact duration. The shoe should protect the foot from impact injury in an area from the heel to 60 percent of the shoe length, especially for the midfoot group. Different shoes may be needed for each running style. Since braking and propulsion occur between 60 and 80 percent of shoe length, the outsole should resist slip from forces applied in anterior and posterior directions here. Outsole wear is caused by relative motion between sole and ground, not where maximum forces occur. Forces during distance running are double that of slow walking. Midfoot strikers are at particular risk from impact injuries. Common patterns of motion, forces, and phasic muscle activity were identified in 10 normal young men as they climbed stairs. During ascent from one step to the next, the hip and knee flex and the ankle plantar flexes slightly. Knee extensors and plantar flexors contract from foot-strike through midstance. The soleus contracts from foot-strike to midstance and the gastrocnemius from midstance to just before toe-off. From midstance toward toe-off, the hip and knee extend and the ankle plantar flexes. Biceps femoris and tibialis anterior become active from just before toe-off until midswing. From midswing to foot-strike, the hip and knee extend, while the ankle moves from maximum dorsiflexion toward plantar flexion. No muscle activity was recorded between mid-swing and foot-strike. Movements are different when ascending from the floor to the first step. Descent from one step to the next requires hip and knee flexion and maximum ankle dorsiflexion. During swing phase, flexion decreases and the ankle plantar flexes. The biceps femoris is active during early swing. Tibialis anterior contracts during midswing and gastrocnemius at foot strike. At the lower step, the hip and knee are slightly flexed and the ankle plantar flexed. As the limb moves toward midstance, the hip extends, accompanied by hip extensor activity. The knee extendors are active through most of stance, and the plantar flexors contract during early stance to balance the dorsiflexion moment which is present. From midstance to toe-off the hip remains extended, while the knee starts to flex, controlled by knee extensors. Maximum dorsiflexion occurs just prior to toe-off. Maximum hip flexion occurs during swing while ascending. The knee flexes most acutely at swing during descent. During stance while descending the knee flexes nearly 69 degrees when going from step to step as compared with 29 degrees while going from step to floor. Hip and knee moments were greater during climbing than during level walking. Ankle motion and moments were similar on ascent, descent, and level walking.

Frontal and horizontal moments were also reported.


The Stationary Attachment Flexible Endoskeleton artificial foot conforms to the shape and action of the human foot. The bolt block and keel are encased in a soft foam cover. Its design is based on study of the human foot where motions are dictated by the shape of articular surfaces and ligamentous restriction. The tarsals and metatarsals are arched, very strong when rigid, but requiring additional support when flexible. Ligaments, especially the long plantar and the plantar aponeurosis, provide trusses. The arch is stable along its long dimensions only. A series of arches bound by ligaments increases mediolateral stability.

The two feet form a dome, each foot being a half-dome, a stable structure. The medial arch is highest. The foot can function when ligaments are intact. Electromyographs indicate that muscles are inactive until the load reaches 181.4 kg. Intrinsic muscles do not respond until midstance; thus, during the first half of stance when the foot is subjected to 120 percent of body weight, the weight is borne by bones and ligaments. The half-dome concept is the basis for the prosthesis.

During late stance the foot must be a semi-rigid lever as the leg rotates externally as much as 29 degrees on the fixed foot. The arch absorbs rotation at the subtalar joint which becomes a motion and torque converter like two 45-degree
beveled gears, so that the forefoot maintains floor contact. At early swing the foot untwists. This motion is not attained when transverse rotations are absorbed in the prosthetic shank. This action is demonstrated by an amputee who wears a rotator; unless he suppresses some rotation, the torque developed in the shank will rotate the foot so violently it will strike the contralateral leg. Thus, rotation should occur within the foot. The foot should have a subtalar joint and a toe break with a plantar fascial strap from the toe area to the heel. It must be all plastic without mechanical joints.

The anterior third of the SAFE foot has a toe break carved on the plantar surface. The middle third is dome-shaped. The posterior third has a bolt block beveled 50 degrees to act as a subtalar joint. The block is made of resin surrounded by polyurethane elastomer. The resulting keel has two Dacron straps acting like plantar fascia. Alignment is like that with single-axis feet. Initial experience with 33 feet has been very successful.

A New Material in Orthotics and Prosthetics: Melvin Stills and A. Bennett Wilson, Jr. (Division of Orthopedics, University of Texas Health Sciences Center, Dallas, Texas) Orthotics and Prosthetics 34:29-37, September 1980.

Although thermosetting plastics have been used in prosthetics since at least 1952, the laminates fail quickly in lower-limb braces when subjected to repeated bending loads. The first reference to thermoforming sheet plastics, such as polypropylene, in lower-limb orthotics appeared in 1968. Now polypropylene and polyethylene are used extensively; although orthotists continue to seek better materials. Surlyn is a thermoplastic used primarily for golf ball covers. Its physical characteristics are similar to polypropylene, but it is transparent and more flexible. Surlyn is DuPont's trademark for ionomers, resins consisting of polymers derived from ethylene / methacrylic acid copolymers, which possess many characteristics found in olefin polymers. Thermo-vac is the trade name given by United States Manufacturing Co. to the Surlyn useful in prosthetics and orthotics. The material is formed by heating it to from 200 to 500 degrees F, depending on the amount and type of forming to be done and the thickness of the material. It may be hand or vacuum formed. It can be shaped over a wet, cold cast. Case reports represent the more than 200 upper-limb, lower-limb, and spinal orthoses fabricated from Surlyn. It is especially suitable for body jackets and upper-limb orthoses where flexibility is sought.

Optimizing the Function of Geriatric Amputees: Joan Erback Edelstein (New York University Postgraduate Medical School, New York) Physical and Occupational Therapy in Geriatrics 1:21-41, Fall 1980.

More than 36,000 Americans over the age of 65 are lower-limb amputees. Typically, this is a below-knee amputee, male, with peripheral vascular disease. His physical status is compromised by one or more cardiovascular disorders in the remaining leg and in the brain, heart, and abdomen. Amputation which occurs prior to cerebrovascular accident and ipsilateral amputation, especially right-sided, offers the best functional prognosis. The deleterious effects of debility are many. Psychosocial factors influence function, especially financial worries, sudden dependence, and social atrophy. Clinicians should anticipate such reactions and orient the prospective amputee and his family to reality, to reverse or reduce situational depression.

Preoperative physical management includes evaluation, especially joint mobility and skin condition. Early postoperative mobilization has a positive effect on postoperative illness and mortality and long-term survival. Contractures are to be prevented by exercise, prone lying and immediate fitting. A wheelchair is usually necessary and must be selected carefully. A temporary prosthesis with a biomechanically rational socket prevents loss of confidence which might result from prolonged immobilization, and is an excellent means of assessing potential for a definitive prosthesis.

A prosthesis does not suit all amputees, such as those with Class IV cardiac disease. The prosthesis should be light in weight and adjustable in socket size. The patellar tendon-bearing socket with liner, cuff suspension, and SACH assembly prevail. Plastic total-contact quadrilateral sockets are usual for the above-knee prosthesis, although the socket may need to be modified to accommodate soft tissue atrophy. Single-axis constant sliding friction knee units predominate. Prostheses for bilateral amputees may be somewhat shorter than the unilateral device, with rather firm heels to prevent a backward fall. Stubbies benefit some bilateral above-knee amputees.

Prosthetic training emphasizes balance and weight transfer. Advance planning is necessary to insure that the older amputee can function well at home; installation of bannisters, for example, may be required. Predicting functional outcome is a major responsibility, and is based on amputation level and other physical and emotional characteristics. Rehabilitation time is prolonged by administrative delays. Energy consumption and walking speed are influenced by amputation site and age.


The amount of energy required by paraplegics to propel a wheelchair was compared with their needs when walking, to determine why so many discontinue walking. Eleven adults with spinal cord lesions at or below T11 were evaluated. All had normal upper-limb and good upper-abdominal strength. Most had impaired proprioception at the hips, knees, and ankles. Only one had any lower limb contractures, namely 15 degree plantar flexion contractures. Ten had no spasticity or pain. Ten wore standard KAFOs with knee locks and ankle joints fixed in dorsiflexion; the other subject wore a pneumatic orthosis (Orthowalk). Eight used wheelchairs primarily.

Heart rate, respiratory rate, step frequency, and expired air were investigated as they circled a 60.5-meter level track while clutch walking and while propelling a wheelchair. Although the energy cost for propulsion was higher for the three who customarily walked, the metabolic demand was less for them than for walking, and they wheeled faster than the other subjects. Habitual walkers displayed similar heart rates and lower respiratory quotients than chair users. When tested at free walking speeds, the walkers had velocities and stride lengths more than double that of the eight who normally used chairs, and had significantly lower heart rates.
Results of heart rate determinations, oxygen consumption per unit of time and per unit of distance, and respiratory quotients confirm that walking is much more demanding than wheeling. Since the average walking speed of paraplegics is less than half that of normal subjects and the rate of oxygen uptake is increased 50 percent, net oxygen uptake for paraplegics is six times that for normal persons. The one who wore the pneumatic orthosis fared no better than the others. Chair use by paraplegics approximates the energy cost for walking by normal subjects.


Cervical fusion is performed for rheumatoid, metastatic, or traumatic instability or pain of the neck and for cervical osteotomy for correction of deformities in ankylosing spondylitis. Early mobility is desirable to maintain general mobility. The spine is supported (for 12 weeks until the graft has united) in a halo-shoulder brace, and then the spine is protected in a mandibular-shoulder brace while the graft consolidates. Special problems of the arthritic are the short neck due to vertebral collapse; small chin; asymmetrical neck, shoulder, and upper torso; prominent shoulder girdle; thin skin; excessive sweating; and posterior midline scar. The spondylitis case poses particular problems as does the cancer patient. The orthoses are custom made to overcome particular difficulties. They are light and offer broad contact with the shoulder and neck. While immobilizing the neck, they keep the mouth, larynx, and arms free. The orthoses are easily removable.

Both orthoses are constructed on a plaster model made, if possible, with the patient seated. Bone prominences are marked, the patient is casted, and the positive cast has extra reliefs over the prominences. The halo-shoulder brace is made of 3-mm Plexidur lined with 6-mm Plastazote for the back and chest sections. The halo has two anterior and two posterior connecting rods. The mandibular-shoulder brace supports the ramus of the mandible while keeping the thyroid cartilage free. It reaches to a point just above the xiphoid process. The occipital part supports the head just behind the ears and terminates at T4. It is also made of Plexidur lined with Plastazote.

Nine patients have used both braces successfully. Case reports drawn from a total of 13 patients are presented. The halo-shoulder brace weighs 1200 grams and the mandibular-shoulder device weighs only 360 grams.


The Danish Amputation Register was founded in 1972 to study the amputation problem. The register consists of voluntary reports from hospitals and prosthetists. It records data on approximately five thousand new amputations and prostheses. Data on survival is derived by biannual comparison with the Danish Central Citizen Register. This report is based on 2,029 persons with dysvascular amputations. No significant differences were found between diabetics and non-diabetics.

Ipsilateral reamputations were done on 10 percent of all amputees within a month of the initial surgery, on 15 percent after two months, and on 16 percent after three months. After six months, the percentage of reamputation is 19, accounting for most of the reamputations during the 4-year study period. Most ipsilateral reamputations are due to uncontrolled postoperative complications. Contralateral amputations were performed on 12 percent of the population during the first year, and by the fourth year, 44 percent became bilateral amputees.

Survival is similar for diabetics and non-diabetics. The greatest risk of death is encountered within 3 months after amputation. At 6 months postoperatively, the chance of survival is about the same as in the normal population. After 4 years, 22.5 percent of the amputees died.

These findings suggest that most ipsilateral reamputations were sequel to unrealistic primary amputation level and/or postoperative complications, such as infection. Contralateral amputation is a relentlessly increasing risk for all dysvascular patients, which the present study confirmed in its agreement with statistics published previously by other investigators. This study demonstrated that mortality after 6 months is not greater than for the normal population, and that very significant numbers of patients live long enough to lose the second leg.


The principle of this orthosis is to support the arm with a movable splint attached to the arm and held in balance by a pneumatic system. The orthosis assists whatever active movements the arm can make, assuming the muscles to be at least poor grade (able to move the arm if gravity is eliminated). If the hand is uninjured, it can be directed in space and rendered usable by the orthosis. The orthosis consists of a harness, an articulated splint, and a pneumatic system which raises and counterbalances the arm. An aluminum harness is fitted to the trunk, over the shoulders, and is held by a belt and straps. A steel spherical articulation is attached to the harness at the scapulohumeral joint for the arm splint. The splint has two metal parts, one for the upper arm and one for the forearm. They are hinged at the elbow, and strapped to the limb. The forearm segment is cylindrical, able to turn on its long axis, for pronation and supination. The pneumatic lifting system consists of a cylinder fixed vertically to the harness at the waist by a universal joint. Inside the cylinder is a piston attached to a vertical rod. A reservoir of compressed air is made from a coiled tube 1.5 meters long fixed to the back of the harness and connected to the upper end of the cylinder.

Air is hand-pumped into the reservoir. When pressure is sufficient, it forces the piston down and lowers the posterior part of the splint, while raising the anterior end. Air pressure balances the weight of the arm. If the elbow is extended, the arm's center of gravity moves distally, thereby adding to the resistance leverage, and requiring variation in piston force.
A pulley at elbow level rotates by forearm flexion and extension. This pulley is linked to another at the shoulder, so that rotation of one joint turns the other. Passive ranges of movement permitted by the orthosis are extensive. The orthosis has been used by 18 patients with a variety of disabilities. It caused no pain and was well accepted. Patients needed no training. With the orthosis, nearly all users increased shoulder and elbow mobility and a third increased strength. Patients improved in daily function also. The orthosis weighs 4 kilograms. It can be used both temporarily during early rehabilitation and permanently.