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For this issue of the Journal of Rehab R&D Joan Edelstein selected articles in all phases of rehabilitation, from the journals listed below. Accordingly you will find:

Upper limb prosthetics articles 1
Upper limb orthotics 4
Lower limb prosthetics 6
Lower limb orthotics 4
Gait analysis 2
General 5

These have been drawn from:
Prosthetics and Orthotics International 3
American Journal of Occupational Therapy 1
Physical Therapy 1
Archives of Physical Medicine and Rehabilitation 6
Orthotics and Prosthetics 4
Journal of Bone and Joint Surgery 4
American Journal of Physical Medicine 2
Acta Orthopaedica Scandinavica 1


Twenty users of the Otto Bock myoelectric hand were compared with 16 users of cable-controlled hooks. Three of each group were above-elbow amputees. The hand tested is a relatively slow model, intended to conserve energy and allow a lighter prosthesis to be fitted. Faster myoelectric hands are available, resulting from less filtering of the signal or a faster motor, which may produce less reliable operation. Extensive personal, medical, and prosthetic histories were collected, including an activities of daily living questionnaire. The amputees then executed tasks using the prosthesis and their normal arm, including picking up small and heavy objects, simulated feeding, and stacking checkers. Strength of cylindrical grasp, gross dexterity, endurance, and maximum terminal device opening distance were also tested. Sixty per cent of the group had traumatic amputation; 55 percent of the population are male. Conventional prosthesis users had them an average twelve years, compared with an average 1.4 years for myoelectric wear, but there was not significant correlation between the period since fitting, and functional measures. Conventional wearers used the prosthesis an average of 14 hours daily, compared with 9.6 hours for myoelectric wearers. All myoelectric wearers had previous experience with conventional prostheses. Myoelectric wearers scored higher on tests of functional excursion, especially the ability to open the terminal device behind the back and neck. The time needed to complete tasks with a myoelectric prosthesis was approximately twice that with a conventional prosthesis; the conventional wearers required nearly 2.5 times as long to complete tasks, as compared with the normal arm. No significant difference was found between maximum weights which could be grasped. All conventional wearers were able to continue tasks for ten minutes, but a fifth of the myoelectric wearers became fatigued. Most below-elbow amputees preferred the myoelectric prosthesis because of improved functional range with the Muenster fitting and better appearance. Above-elbow amputees were much less accepting of either type of prosthesis.

Knee Flexion During Stance as a Determinant of Inefficient Walking: David Winter (Department of Kinesiology, University of Waterloo, Waterloo, Ontario, Canada) Physical Therapy 63:331–337, March 1983.

Seven young adults without gait problems walked on a walkway at three voluntarily controlled cadences: slow, natural, and fast. They had reflective markers attached to several anatomic landmarks and wore their own footwear. A tracking cart, carrying television and cinematographic cameras, followed each individual. Background wall markers gave a reference so that body coordinates could be scaled as absolute coordinates. Using displacement and linear and angular velocities of body segments, the potential and kinetic energies were calculated and summed to give the total energy of each segment. Summation of all segmental energies yielded the total body energy, showing increases when net positive work was done by muscles, and decreases when negative work was done as muscles absorbed energy. Over a given stride at constant speed, the body returns to the same energy level once; positive work equals negative work; the sum is called internal work. The work cost of walking, defined as work per body mass per distance walked, was correlated with maximum knee flexion during stance.
A significant positive correlation exists between energy cost and maximum knee flexion, but not between velocity and flexion, nor between cadence and flexion. The prediction of Saunders that stiff-legged walking is more energy consuming than normal flexed knee stance does not appear to be true. As knee flexion increased to the normal value, knee, hip, and ankle extensor muscles contracted to limit knee flexion; such activity increases the metabolic cost of walking and quintuples the bone-on-bone forces at the knee and hip, as compared with stiff kneeed gait.


Amputation of 320 limbs was performed in 310 arteriosclerotic patients with a mean age of 70 years. Hospital mortality was 18 per cent, with average stay of 68 days. Of the survivors, 116 were above-knee amputees, 48 through-knee, and 101 below-knee amputees. Prosthetic fitting was attempted in all who had previous walking capacity. Fifty-six per cent of the entire group were able to walk with a prosthesis. The result is significantly related to age and level of amputation. Sixty-eight per cent of through-knee and below-knee amputees used prostheses well. Only 9 per cent of patients admitted from nursing homes were successfully fitted. Success rates of 77 per cent were achieved for those discharged to their home, and 76 per cent for those sent to a rehabilitation institution. The most successful amputation level in relation to fitting was through-knee, with a rate of 97 per cent, as compared with 83 per cent for below-knee and 61 per cent in above-knee. The risk of death during hospitalization is highest for above-knee, and equal for through- and below-knee. Through-knee amputation should be considered an alternative to the below-knee level in feeble patients with poor muscular balance, as most can walk on a prosthesis, if necessary with a knee lock. Failure of below-knee fitting may be explained by the infrequent use of a conventional below-knee prosthesis with knee lock. Cosmetic considerations are less important in geriatric patients.


Eight male unilateral below-knee amputees were evaluated wearing a single axis foot and a SACH assembly with temporary patellar tendon bearing prostheses with cuff suspension. As they walked 6 meters on level flooring they were photographed simultaneously from lateral and frontal perspectives. Measurements included vertical displacement and velocity of the center of mass; hip, knee, and ankle angles at heel strike, foot flat, mid stance, heel off, toe off, acceleration, mid swing, and deceleration. The percentage of time spent in stance, swing, and double support phases was calculated, as well as step and stride lengths. Six angular comparisons proved statistically significant, although in five instances, the difference was 4 degrees or less. The clinically important comparison was ankle angle at foot flat, which showed a difference of 6.5 degrees between the two foot types. The single axis foot permits more planter flexion and dorsiflexion than SACH. The amount of time spent in each phase of gait was approximately the same regardless of type of foot. Interchanging prosthetic feet thus did not appear to have a significant overall effect on gait.


The major problems encountered in sports are running and walking long distances. Ten physically active amputees were photographed running at self-selected constant speeds. Ground reaction forces acting on the prosthesis were two to three times body weight. Several amputees were unaware they could run; one ran 40 yards in 5 seconds after coaching. The major problems were maintenance of an excessively straight knee on the prosthetic side during heel contact, reducing shock absorption and overstressing the knee, hip, and spine, and restricted range of motion of the intact knee and hip during swing phase. Normal foot function during running requires pronation and supination to allow the foot to roll on the lateral border after ground contact. Prosthetic feet do not have an adequate amount of this motion. Running prostheses should be planar flexed so that the runner’s weight can be centered over the ball of the foot during push off. Rubber latex sleeves minimized pistoning while the wearer ran. An energy storing ankle-foot design is needed. The new foot incorporates a fiberglass leaf spring mechanism to aid push off. The spring stores and releases the energy of gravitational compression. A rubber deflection bumper is angled 22 degrees. An extension limitation cable just anterior to the bumper allows the foot to compress but not extend. The new design is being evaluated through laboratory strength testing, running analysis with force plates, and patient responses. Amputees noted initial difficulties with slow walking; the foot tended to throw the leg forward. Running and ramp and stair ascent were easier with increased stride length and push off force. The foot increased activity levels to include easier running and jumping with more comfort.

Seven male traumatic unilateral above-knee amputees walked with both a prosthesis with one of several types of constant friction knee unit and one with a hydraulic knee, either Henschke-Mauch or Dupaco. Interrupted light photography recorded displacement patterns of multiple body segments. At slow speed no significant differences exist between gait with either type knee unit, with regard to velocity, cadence, and stride length. At free and fast speeds, amputees walked faster with hydraulic units, increasing cadence, rather than stride length. At all speeds, amputees walked slower than normal controls, regardless of types of prosthesis, and had longer prosthetic swing phase than sound swing phase, especially with constant friction units. Step length with hydraulic components was 2 to 4 cm shorter than with constant friction units. Type of knee unit did not affect stride width or foot angle, although the prosthetic foot usually was in less toe-out than the sound foot.

Peak knee flexion in early swing increased notably when they walked faster, although flexion was subnormal with hydraulic units and excessive with constant friction. With constant friction at the end of swing phase, some amputees showed small hip extension excursion followed by flexion before heel strike. With either type unit, amputees were in less dorsiflexion during stance and more plantar flexion at toe-off, and most exhibited vaulting during fast walking. During fast walking with constant friction, ipsilateral elbow excursion was notably more subnormal than with hydraulic components. Heel rise during fast walking was markedly excessive with constant friction, but within normal range with hydraulic units.


Of the 1100 patients with lower limb amputation admitted during the past twelve years, 103 subsequently required contralateral amputation. Postoperatively, the amputation limb was placed in a soft or plaster dressing, depending on the level of amputation. Most patients were discharged in a wheelchair. Following wound healing, one to three months after surgery, they were readmitted for prosthetic training, usually for two to four weeks.

The average interval between amputations was 2.8 years, and the median was 1.2 years. No correlation exists between interval and rehabilitative outcome. The only bilateral above-knee or through-knee amputees able to walk with prostheses were traumatic amputees. A fourth of AK/BK amputees walked with prostheses and auxiliary aids; half were able to transfer from the wheelchair with prostheses. No correlation exists between sex, different levels of amputation, or extent of systemic disease and success of rehabilitation. Eighty per cent of patients with bilateral BK or Syme’s amputation were prosthetically rehabilitated; none were bedridden. Those with one or two Syme’s amputations did better than the bilateral BK amputees. Age, etiology, systemic involvement and levels of amputation are important predictors of rehabilitation outcome, rather than patient motivation.


Six bilateral dysvascular below-knee amputees, aged 45 to 70 years, were compared with eight nonamputees of similar age. All amputees had completed prosthetic rehabilitation and could walk for at least 5 minutes; none had pain, edema, or ulceration. Tests were conducted on two consecutive days at the same time each day. On the first day, subjects walked 40 meters at their natural speed with any accustomed aids. On the second day, they performed on a stationary wheelchair ergometer fitted to the individual’s wheelchair. They propelled the wheelchair the equivalent distance walked the previous day.

The average velocity of the control group during ambulation was 63 meters per minute compared with 40 for the amputees, a 64 per cent performance. With the wheelchair, the mean velocities of amputees and nonamputees were similar, and amputee ambulation and wheelchair velocities were similar. Able-bodied persons walked 48 per cent faster than they propelled a wheelchair. Mean resting heart rate showed no significant difference between groups or activities. Amputees had a 26 per cent higher heart rate than controls during ambulation. During wheelchair propulsion, both groups had similar heart rates. Oxygen consumption during ambulation was 157 per cent more for amputees than for wheelchair propulsion. Wheelchair propulsion is physiologically less demanding for elderly bilateral below-knee amputees than prosthetic ambulation, and allows one to be mobile at a greater velocity. The wide divergence in results among patients highlights the need for individual evaluation with regard to mobility training.

Energy Expenditure in Hip Disarticulation and Hemipelvectomy Amputees: Farhad Nowroozi, Mario Salvanelli, and Lynn Gerber (University of California Irvine Medical Center, Orange, California) Archives of Physi-
Eight hip disarticulates and ten patients with hemipellectomy were compared with 11 able-bodied individuals. The amputees used prostheses, had no existing tumor nor clinically evident cardiovascular or pulmonary disease, nor chemotherapy or radiation therapy for the tumor nor clinically evident cardiovascular or pulmonary disease. Subjects walked with prosthesis and any accustomed ambulatory aid. After resting they walked with two crutches, using the swing-through gait.

Walking speeds were significantly slower as compared with able-bodied controls. Crutch ambulation was significantly faster than prosthetic walking. Oxygen consumption during rest and comfortable walking speed was not significantly different. Amputees had higher resting pulse rates, but all returned to preexercise rates within five minutes of terminating exercise. Energy cost was not significantly different when comfortable speeds were compared; crutch walking had slightly lower consumption than prosthetic walking. When oxygen consumption is expressed per unit of distance worked, the greater efficiency of unimpaired individuals is evident. Hip disarticulates spent 82 per cent more energy than controls, and hemipellectomy patients spent 125 per cent more. The energy cost of crutch walking was 45 per cent more than walking by the control group. Hip disarticulates walked at 61 per cent and hemipellectomy patients at 51 per cent of normal speed. Amputees could not maintain fast walking very long.


Dynamic splinting facilitates hand rehabilitation by providing low amplitude force over a prolonged period to influence the synthesis of new tissue, keeping tissues in constant mild tension. Sterling Bunnell had designed plaster splints with outriggers many of which were low profile with force lines directed parallel to the splint base. For easy manufacturing the custom molded plaster bases were sacrificed for metal, necessitating high outriggers. Low temperature plastics allow rapid application of low profile splints. The line of pull of force to stretch a joint should be perpendicular to the axis of the long bone that is the distal articulation of the joint in question. The outrigger end is as close as possible to the point of force application, allowing it to be of minimal length. The outrigger directs the line of pull, keeping it close to and parallel with the splint base. The finger loop is attached to string carried over or under the outrigger, and a rubber band is attached to the string. The band is secured to the base by hooking it over a small metal hook on the base.

Bases are made of polycaprolactones. Brass welding rod is well suited for outriggers hooks, loops, and pulleys. Loops are leather, vinyl, or plastic. Nylon monocord and stationer's rubber bands complete the assembly. Outrigger and base stability are obtained with short outriggers and close anatomic conformity of the base. Force application should be specific and changed as the patient's needs alter. Uniform low pressure distribution is easier to obtain with low profile designs than with the traditional lumbrical bar. A rubber band provides more constant tension the longer distance it is stretched. Low Profile orthoses are less cumbersome than previous designs involving high outriggers.


The Triceps-Supination Orthosis was designed for a C6 quadriplegic with elbow flexion contracture and forearm limitation. It extends the elbow while permitting active flexion, pronation, and supination. During rehabilitation, the patient had six inhibitive serial casts to obtain more elbow extension. The casts applied neutral warmth with prolonged stretch in a submaximal range. Each cast increased range approximately 9 degrees, although
gains were not maintained following cast removal. The orthosis is intended for individuals with contracture or spasticity. Active elbow extension of minus 45 degrees and passive pronation and supination are prerequisites for fitting. The orthosis will extend the elbow to minus 10 degrees depending on the severity of spasticity, and will permit full forearm motion. High patient motivation is also necessary for best use of the device.

The orthosis incorporates mechanisms permitting adjustment of the amount of tension applied to the mechanical elbow and radioulnar joints. Rotation of a spring alters the amount of tension. For forearm motion, rod end bearings imitate radial motion. Distally, the rod end bearings are riveted to a forearm band and distally to a polypropylene wrist cuff. Two threaded steel rods rotate in the sleeve of the proximal rod end bearing. Distally, the rods are attached statically to the rod end bearings. A torsion spring over the radial rod passively advances the hand into pronation. In addition to increasing elbow and forearm mobility, the orthosis aids wheelchair propulsion, feeding, and light hygiene.


A self-administered questionnaire survey was conducted of all men between the ages of 18 and 55 who were patients of a family practice facility during a 3-year period. Subjects rated current or past back pain as none, mild, discomforting, distressing, horrible, or excruciating. Items dealt with symptoms, medical care, occupational requirements, use of motor vehicles, recreation, and time lost from work. Thirty per cent of the respondents never experienced pain; 46 per cent had moderate pain, and 24 per cent had severe symptoms. The median age for each of the three groups was 33 years. Three per cent had had surgery. Men with moderate or severe pain were much more likely to be cigarette smokers than those without pain. The most important prognostic variable was repetitive weight lifting of 20 kilograms or more. Patients who used jackhammers or other vibrational equipment were more likely to have pain, as were professional drivers. Current and past sports activity was not appreciably different among the groups, although those who were cross-country skiers during adolescence were more likely to develop moderate pain in adulthood.

Bed rest and medication were the most common treatments, although a significant number had physical therapy and back supports. The economic consequences of low back pain can be extrapolated to $11 billion annual lost wages.


Two of every three people will suffer from low-back pain eventually. Of the 1.25 million Americans who injure their backs annually, more than 65,000 qualify for permanent disability. Failure to understand the physiological and psychological complexities of pain accounts, in part, for the low success rate of treatment. Pain may be a means of dealing with guilt, especially for those who believe that they do not deserve happiness. The pain-prone patient utilizes pain to avoid even more unpleasant feelings. Others use pain for stimulating affection and controlling the environment. Chronic pain may be an acceptable means of blaming failures.

The most widely used self-administered questionnaire is the Minnesota Multiphasic Personality Inventory. Its 550 items separate into ten clinical psychological scales and three validity scales. In low-back pain, the most important indication of somatic fixation is abnormal elevation of hypochondriasis and hysteria. Scores of patients with functional pain were significantly higher than those with organic findings on the scales of hysteria, depression, hypochondriasis, psychopathic deviance, psychoasthenia, and schizophrenia. Scores can aid prediction of who should have success with orthopedic treatment; low hypochondriasis, depression, and hysteria scores were found among patients achieving good treatment results. Litigation patients scored significantly on those scales. The inventory also elucidated other personality characteristics, particularly chronic anxiety with resultant tendency toward compulsiveness which predisposes such persons to conversion symptoms by which they resolve psychological conflicts.

The Cornell Medical Index, 195 items, is moderately helpful in understanding spinal pain; low-back patients have a greater tendency to adopt an invalid self-concept than do arthritics. The Middlesex Hospital Questionnaire and the Eysenck Personality Inventory are other self-administered tests pertinent in differentiating patients likely to have functional pain. The Melzack Pain Questionnaire consists of 102 words relating to pain, enabling subjects to describe pain; the experience of pain seems to originate from sensations of tissue damage and from the effect of pain on one’s mood. Those with organic pain used fewer words to describe pain than those with functional disorder. The Mooney Pain Diagram permits subjects to draw the site of pain; those with anatomically confusing patterns tend to have histories of chronic pain. Routine use of psychological testing is warranted to predict treatment outcome.

Existing motions must not be restricted unnecessarily. Orthotic axes should match body axes as perfectly as possible to avoid shear stresses. When the orthosis is attached to the shoe, effective positioning of the stop is difficult and appearance is compromised. The sandal creates spatial problems with shoe size, but offers clearer proportions with regard to the entire orthosis. The shoe and foot support must be level to the floor, regardless of foot deformity. A low lateral trim gives less hold on the outside. The positive model of the foot should be taken weight bearing, or the heel of the model be flattened and plaster added to give the calcaneus room. If the foot can be corrected actively or passively, the orthosis should be aligned so that the plumb line of the leg and orthosis coincide with the line of the foot. Support should never lie medial to the foot. Joints should be on the sagittal plane, regardless of the amount of toe-out.

Overcorrection of knee valgus or varus limits leg extension. Avoid free play between the orthosis and leg, especially when hip muscles are weak. If one joint is blocked, the two adjacent limbs form a rigid lever whose mechanical importance lies in the line connecting the two free ends: the line corresponds to the longitudinal axes of both limbs, rather than the anatomical axis. An ankle-foot orthosis with blocked dorsiflexion thus aids the patient with quadriiceps paralysis and active hip musculature. A knee lock requires that the dorsiflexion stop be reduced considerably because knee and foot movements should not be blocked simultaneously. The dorsiflexion stop should never exceed 90 degrees; otherwise, the knee will hyperextend and the patient will experience difficulty in stance transition. To increase knee stability, lengthen the foot support, rather than extending the dorsiflexion angle. A soft plantar-flexion stop prevents toe drag without exerting a knee flexion moment.


The original small portable FES unit, reported in 1961, was supplanted in Yugoslavia by larger portable stimulators which divided the gait cycle into 16 parts of equal duration with individually adjustable intensities of stimulation. Many hemiplegics, however, cannot maintain the regular cadence needed for the system. The present system is a nonportable battery-operated multichannel universal control system, by which FES can be timed independently of cadence. The sequence and strengths of FES can be adjusted to a wide range of gait variations. The unit is less bulky, less expensive, and less complex than comparable portable units with wide operational options.

Patterns of logic triggered by the heel, ball, and toe switches on each foot are formed on the console by externally plugging into five kinds of electronic units: (i) an inverter causes delays determined by a timer; (ii) flip flops are turned on and off by separate positive pluses; (iii) a one-shot timer is triggered by a positive pulse to produce a positive pulse for an adjustable duration; (iv) OR gate allows convergence of logic sequences into a single FES channel; and (v) AND gate is triggered if all sequences are producing positive pulses simultaneously. The logic is always gated through an AND gate to produce interrupted direct current with potential frequency range of 20 to 100 cps. This protects the patient against continuous surges of direct current.

Toe contact initiates FES to the dorsiflexors. As the toe leaves the floor, the rising pulse from the inverter signals the start of a flip flop unit’s “on” phase which is terminated by heel contact. Alternatively, the duration of FES can be regulated by timers. Application of FES to calf muscles briefly just after foot-flat and for a longer period from foot-flat to toe-off improves gait. Other logic circuits are also described and diagrammed.


Gait characteristics of ten subjects were evaluated as they walked with five different ankle-foot orthoses (AFOs). The subjects included six hemiplegics and four nondisabled persons. All hemiplegics could walk without a cane and had some dorsiflexion limitation. All wore laced Oxford shoes. The orthoses were the TIRR(Engen) corrugated polypropylene, Teufel polycarbon posterior leaf spring, and three versions of the Seattle polypropylene orthosis. One was trimmed just anterior to the malleoli, another terminated just posterior to the malleoli, and the third was trimmed farther posterior. A self-aligning electrogoniometer measured plantar and dorsiflexion, and a gait event marker system recorded the occurrence of heel strike, toe strike, heel-off, and toe-off. The anteriorly trimmed Seattle AFO restricted motion more than the Teufel. All motion comparisons among the various Seattle trimlines were significantly different; with each trim, the AFOs became more flexible. The Engen orthosis was not significantly different from any Seattle AFO, except that it allowed less peak dorsiflexion than the most posteriorly trimmed Seattle AFO. No difference existed between peak plantar-flexion values at heel strike for control or paretic subjects. During push-off, the peak dorsiflexion values were significantly lower for hemiplegics wearing all braces except the Engen. The Teufel was the most flexible AFO. The anteriorly trimmed Seattle orthosis provided the best substitute for push-off by restricting dorsiflexion; it also resisted plantar flexion the most during swing phase.

Knee instability may be straight lateral or straight medial, demonstrated by widening of the joint space with varus or valgus stress exerted on the knee at full extension or some flexion. Straight anterior and posterior instabilities are illustrated by drawer signs. Rotary instabilities involve posteriomedial, anteromedial, anterolateral, and posterolateral instabilities. Instabilities are often combined. The shape of the femoral condyles, tibial plateau, and intercondylar eminence contributes to stability. The slight concavity of the plateau controls tibial excursion and produces some straight stability during weight bearing. Menisci, muscles, and ligaments also aid stability.

The Lenox Hill orthosis protects nonoperatively treated patients against recurrent stress during activity. The orthosis also increases static stability of postoperative patients. It may be fitted by taking a positive mold before surgery so it will be ready for application on cast removal in 6 weeks. Alternatively, the mold may be taken during the first cast change 2 weeks postoperatively, allowing for better accommodation of atrophy. If athletic activity is allowed, the orthosis is worn up to 1 year. With the knee extended, valgus deviation is resisted by lateral leg pads above and below the knee and the medial knee disc. With the knee flexed, anteroposterior tibial excursion is resisted by the pretibial bar, derotation strap, distal knee loop, and circumferential rubber, all opposed by the circumferential rubber above the knee. A side joint stop and nonelastic popliteal strap resist hyperextension. The orthosis also resists rotary instability by lateral leg pads, medial knee disc, derotation strap, and the circumferential rubber above and below the knee.


A prospective study of all consecutive patients diagnosed as having isolated complete tear of the medial collateral ligament was conducted. The mechanism of injury was valgus stress to the lateral aspect of the distal thigh or proximal leg. All patients had stress testing under general anesthesia within six days of injury and all had arthroscopic evaluation to ascertain that there were no other intra-articular injuries. Twenty-four patients seen early in the study had surgical repair of the ligament. All patients were placed immediately in a toe-to-groin plaster cast. The surgical patients retained the cast for six weeks while walking with minimum weight-bearing; the cast was then removed and active exercises initiated. The twenty-seven patients seen later in the study had the cast changed to a fiberglass hinged cast brace at two weeks; the orthosis allowed 30 to 80 degrees of flexion. All patients had similar rehabilitation, concentrating on quadriceps, hamstring, and hip flexor power. No appreciable difference existed in age or athletic ability between the surgical and non-surgical groups.

Clinical results were graded by objective physical examination, subjective responses relating to activities, and performance of functional activities. Those with cast-bracing treatment scored better with regard to per cent of patients achieving good to excellent results (90 percent versus 88 percent) and speed of rehabilitation (11 weeks versus 15 weeks).

The study demonstrated the value of early protected mobilization in a cast brace for ligament injuries of the knee. Rehabilitation was facilitated without any apparent negative effects on the final stability of the knee. Both groups achieved minimum knee laxity of no functional significance. No obvious advantage was gained by surgical intervention.


An Ace elastic support brace and Lenox Hill derotation brace were activated during overground running, using the CARS-UBC electrogoniometer, on six young adults, all of whom had unilateral knee surgery for repair or removal of deranged structures at least 10 months prior to testing. All had worn the Lenox Hill orthosis and maintained daily regimens of jogging, bicycling, swimming or soccer.

Subjects ran with the electrogoniometer in place in a 20 meter area at 3.35 to 3.58 m/s pace. The derotation brace reduced maximum knee flexion 11 per cent during swing phase and maximum external rotation to a significant extent. No marked differences existed between orthoses as measured by maximum flexion during stance or maximum internal rotation. Restraint on dynamic flexion was considered undesirable, but not significant in altering the total gait pattern. Subjects responded to the decreased restraining effect of the elastic support by producing greater flexion values in stance and swing than were obtained as the surgical limb was measured with no brace. The elastic support may have influenced the subjects’ ability to move the joint through greater ranges. The elastic device tended to equalize internal and external rotation parameters of the surgical limb. Since the healthy limb demonstrated greater external rotation values, the effect of the elastic support was to increase external rotation so it approximated the healthy limb performance. The derotation brace reduced external tibial rotation of the surgical limb 31 per cent and internal rotation 22 per cent, whereas the elastic support increased external tibial rotation and did not reduce any motion parameter.

Nine healthy individuals and seven with lower limb disabilities were evaluated while propelling an Everest and Jennings Universal model wheelchair with and without an arm cranked Unicycle attachment. The arm cranked version was arranged to control for the weight it would have added to the basic wheelchair. Participants rolled over smooth and carpeted surfaces at various speeds.

Able-bodied and disabled participants exhibited the same general responses to all wheelchair combinations. Handrim propulsion elicited significantly higher energy costs, in spite of the disabled persons' chronic exposure to wheelchairs requiring the upper body propulsion system of handrim operation. Arm cranking requires lower pushing force because the rolling resistance is greatly improved when the small front casters of the conventional wheelchair are raised off the floor by the larger wheel of the Unicycle. Cranking employs asynchronous movements of the upper limbs, rather than synchronous arm movements in handrim use. Asynchronicity may use inherent neural pathways of reciprocal innervation while also permitting continuous force application, rather than intermittent thrusts used during handrim stroking. Cranking allows propulsion whether the person pushes or pulls, and involves greater muscle mass than does handrim stroking.

Addition of gear drive variability to wheelchair design allows better matching of propulsion system force application to the physique of the user. High and medium gears are best for arm cranking, but low gear may be essential for traversing ramps and rough flooring and for feeble patients.


Conventional materials do not permit easy adjustment with adequate strength. A structural node and beam matrix was developed forming a strong enclosing or supporting structure. The matrix is an array of small components which can be linked, shaped, and locked. It has a wide range of rigidities and is adjustable in stiffness and shape. The matrix is suitable for cerebral palsy seating. The system, however, can only be contoured easily to cylindrical-conical shapes. Also, the locking force of the matrix is opposed by loading forces acting along the neutral axis of the structure, so very high friction is needed at the nodes to give secure locking. The system uses the I-beam principle, where structural beams are separated from the neutral axis; rods are separated by nodes. Loading puts one beam in tension and the other in compression, creating a strong structure. Nodes can be positioned at any point along the beam, giving continuous adjustment and complex shapes. The system is also applicable to lower and upper limb and spinal bracing, and may eventually replace plaster for many orthotic applications. Further refinements will include use of different surfacing elements under different loading conditions, and use of insertable modules for functions such as load measurement, feedback, and attachment of accessories such as hinges. The matrix approach takes advantage of mass production for producing standard components which can be assembled without special tools or facilities. The resulting appliances are lightweight and well ventilated.