Abstracts of Recent Articles

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For this issue of RR&D Journal, Joan Edelstein again sought articles in all phases of rehabilitation engineering, from a wide number of current journals. Accordingly, you will find:

Upper limb prosthetics articles 1
Upper limb orthotics 2
Lower limb prosthetics 7
Lower limb orthotics 6
Spinal orthotics 2
Gait analysis 4
Seating and wheelchair 2

These have been drawn from:

Prosthetics and Orthotics International 5
Physical Therapy 4
Archives of Physical Medicine and Rehabilitation 3
Orthotics and Prosthetics 3
Journal of Bone and Joint Surgery 3
American Journal of Physical Medicine 3
Acta Orthopaedica Scandinavica 2
Journal of Biomechanics 1


Measurement of transcutaneous oxygen tension appears to be the most promising technique for amputation level selection. The procedure is non-invasive, does not employ radioactive isotopes, and appears to reflect the capacity of the local circulatory system to deliver oxygen to the skin.

Thirty-seven patients presented various indications for amputation, including gangrene, non-healing ulcers, infection, failure of previous amputation, and resting pain. Twenty-four were 60 years or older, twenty-one were smokers. Two-thirds of the population had diabetes. Skin at the measurement sites (upper chest, mid thigh, mid leg, and dorsal foot) was shaved. The stratum corneum was removed by cellophane tape stripping, and the underlying skin heated to 45 degrees C. Heating provides significant local cutaneous vasodilation for maximum increase of blood flow. Readings from a commercially available probe were recorded at the end of 20 minutes of heating. Whenever distal pulses were detectable with Doppler ultrasonic flowmeter, segmental systolic blood pressures were also measured, and converted to an ischemic ratio by dividing the pressure by that of the brachial artery.

Amputations were performed at the below-knee level based only on clinical criteria, especially skin temperature, hair growth, pulses, rate of capillary filling, and bleeding at the operative site. All patients received immediate postoperative plaster dressings.

Two factors were significantly different in the seven patients whose below-knee wounds failed to heal. These patients were heavier smokers and had zero or very low transcutaneous PO₂. A majority also had failure with previous vascular surgery. All patients with PO₂ above 40 mmHg had primary healing at the below-knee level. Ischemic ratios did not prove a reliable predictor for healing. Seven of the 10 patients in whom pulses could not be obtained healed.

Transcutaneous PO₂ measurements appear to be superior to other methods in showing the capacity of the local circulation for delivery of oxygen to the skin.


The pneumatic Airleg, developed in 1975, overcomes several drawbacks of plaster immediate postoperative dressings: plaster prevents easy wound inspection, plaster removal and replacement requires skilled personnel, and plaster fosters damaging piston action as the stump shrinks. The Airleg, available in above- and below-knee models, consists of an inner air sac and an outer quadrilateral fiberglass shell with a metal shank and prosthetic foot. The air sac is worn 24 hours a day, removed only for wound inspection. The outer shell is applied for ambulation. Pressure is ordinarily 12 mmHg, and is increased to 25 to 30 mmHg for ambulation. The Airleg suits immediate-postamputation ambulation, as well as early ambulation when a soft compressive dressing has been applied, and ambulation in the presence of open infection or ulceration.

Three case histories illustrate the types of clinical benefit of the Airleg. As compared with other semirigid dressings, such as the Unna boot and the air splint with or without an external frame, the Airleg provides constant measurable pressure and is a total contact shock absorber allowing the patient to bear as much weight as is comfortable. The shell and foot improve stability and increase weightbearing. The bulkiness of the shell and imposition of a locked-knee gait have not interfered with training with the permanent prosthesis.

Running patterns of 10 active unilateral amputees were recorded cinematographically. They ran at self-selected constant speeds ranging from a jog to near their maximum. Six subjects ran at speeds ranging from 3 to 8 meters per second. Three others achieved periods of non-support from the intact foot, but not from the posthesis. One amputee who had traumatic brain damage had two periods of double support, more characteristic of walking than of running.

The trend suggests greater reliance on step rate for discrete increases in speed than for non-amputee runners. At a given speed the prosthetic step rate was greater and the step length less than intact counterparts. The intact step accounted for approximately 54 percent of the stride length. Duration of non-support during the intact step did not change much over the range of speeds. Non-support duration was positively related to speed for the prosthetic step but nearly constant for the intact step.

Thigh-knee kinematics of several amputees were comparable to those of a skilled runner. Amputees with little or no running experience had patterns similar to walking, characterized by restricted range of knee flexion of the intact limb during its forward swing, so the intact foot was always close to the ground. Thigh position was similar to that recorded in walking. Intact take-off did not occur until after knee flexion had been initiated. A second major pattern was a thigh-prosthetic foot angle equal to or greater than 180 degrees maintained during at least a portion of single support on the prosthetic foot.

Following a brief period after foot-strike when the ankle angle was constant, intact support consisted of ankle dorsiflexion, then plantar flexion through a greater range; the leg rotated backward about the knee. After take-off the ankle dorsiflexed slightly as the leg first rotated backward at the knee, then forward in preparation for foot strike. On the prosthetic side, SACH foot wearers displayed limited dorsiflexion. The Greissinger foot allowed slightly more passive dorsiflexion.

For the nonamputee, step length increase assumes a greater role at lower speed, while step rate dominates high speeds. For amputees, the primary factor at all speeds is step rate increase due to decrease in single support duration. A more mobile prosthetic foot and adequate instruction can enable more amputees to run.


Complete rehabilitation includes recreation. Skiing for amputees was initiated in Europe forty years ago and in the United States in the 1960s. Sports offer psychological and physical benefits by reducing the relative inactivity imposed by the handicap so the amputee can meet the increased exertion to function with a prosthesis. Participation, usually in sports planned specifically for the handicapped, requires that amputees maintain themselves in good physical condition.

Sports prostheses and equipment include water-resistant prostheses for beach walking, showering, and swimming. The prosthesis must be aligned for barefoot or sneaker use. The shank should be hollow with vents at the base to allow water flow into the shank to minimize the tendency of the prosthesis to float. A foot unit adjustable to plantar flexion or swimming flippers eliminate drag when swimming.

Skiers may use special outrigger poles. A detachable knee-shank unit enables the above-knee amputee to walk to the ski area, replace the distal portion of the prosthesis with a pad to lessen the tendency to injury during falls. Golfers are aided by a rotator in the shank which absorbs torque to limit shear stress and smooth follow-through. A bilateral below-elbow amputee was fitted with adapted prostheses to hold clubs. Similar adaptations suit tennis and badminton racquets. A cosmetic hand and sleeve provide a surface for sculling when swimming.

An above-knee dogsled racer changed from an hydraulic to a pneumatic knee unit to participate in frigid weather. A sky diver has a soft socket, rather than a hard one.

Commercially available attachments for bowling, baseball, tools, and pistol firing benefit the upper-limb amputee, as does a chest harness to hold a fishing pole.

Below-knee amputees need no special equipment for basketball, baseball, tennis, handball, and marathon running. Above-knee amputees using fluid-controlled knee units play tennis, basketball, and volleyball. Wheelchair athletes participate in javelin throwing, discus, shot put, weight-lifting, races, archery, and table tennis without special equipment.


Seventeen hemiplegics accustomed to a bichannel adjustable-ankle locking orthosis were fitted with a rigid-ankle polypropylene ankle-foot orthosis which was custom fabricated over a positive mold of the subject’s leg. Three-sixteenths-inch-thick plastic was used. Medial and lateral trim lines were anterior to the malleoli, and the footplate terminated proximal to the metatarsal head. The AFO was aligned in slight plantar flexion. The plastic AFO weighed less than the metal one (250 to 400 grams compared with 800 to 900 grams). An insole footswitch system provided quantitative information regarding heel, hallux, and first and fifth metatarsal head contact time and loading amount. Electrogoniometers measured ankle and knee motion as subjects walked at free velocity. The first test session involved use of the metal AFO. Two or more weeks later, subjects were tested with the plastic orthosis which they had worn exclusively during the interval between tests. All walked without auxiliary aids.

Subjects ranged in age from 23 to 70 years, the average being 53 years. Ten were right hemiplegic. Brace design did not affect the average values for velocity, cadence, and stride length. Paretic swing time, however, was significantly prolonged with the metal orthosis, and double-support time...

Four healthy young adults were fitted with identical blucher shoes. The right shoe had a rigid sole plate with stirrup and Becker adjustable joints; metal uprights and calf band with strap completed the orthosis. Subjects walked over a 100-foot walkway, first with unmodified shoes at 80 to 85 percent of normal cadence. Data was recorded by means of motion picture photography and Kistler triaxial force plate, aided by a gait event marker system which identified heel contact, instability, and difficulties donning the shoe over the AFO.

The orthosis did not change stride duration, nor ratio of stance to swing duration. It increased the duration of heel strike and push off and decreased midstance. A 5-degree plantar flexion stop lengthened heel-strike duration and increased push-off most. A 5-degree dorsiflexion stop had the opposite effect. With the orthosis, during early stance the foot took a relatively long time to rock over heel to foot-flat. During late stance, the heel rose earlier as a result of dorsiflexion restriction. The forward shear peak was significantly less with dorsiflexion adjustment than that generated with an orthosis, and the peak was significantly greater with plantar flexion setting. A shorter step prior to heel contact decreased shear peak magnitude. Limbs were less inclined at heel-strike when walking with dorsiflexion adjustment. Plantar adjustment held the knee back at midstance, resulting in a larger and longer knee extension moment.

The knee moment component due to the vertical force had the major influence on total moment during early heel strike, late push off and midstance, while the component due to fore/aft shear had the most influence during early and late stance. Moments created by the vertical and shear forces partially cancel each other, to the point that total knee moment accommodated by active muscle contraction is reduced, reducing walking effort. Bending during early stance is primarily influenced by shear forces, whereas extension during midstance is generated mostly by vertical ground force. A double-stopped orthosis may increase extension to the point of genu recurvatum. A plantar stop increases knee bending during early stance because of an increase in the magnitude of the flexion moment component due to the forward shear force, in spite of an increase in the extension moment due to the vertical force. A dorsiflexion stop increases the bending moment even more; while the bending moment resulting from shear force is decreased, the subject reduced step length, thus increasing the duration of flexion in early stance. The extension moment in midstance is increased by the plantar flexed orthosis, but not by the dorsiflexed one.


The effects of water temperature, plaster cast thickness, plaster residue in the water, and a pillow beneath a newly applied cast were measured to determine the extent of heat danger from plaster. A Pyrex tube, 12 centimeters in outer diameter, 6 millimeters thick and one meter long, simulated a leg. Pyrex has thermal conductivity close to bone. The tube was filled with water at 36 to 39 degrees C to simulate the homeostatic effects on tissue temperature and dissipation of heat by blood flow. A thermocouple was attached to the inside of the tube, another on the outside, simulating skin, and a third just under the outermost ply of the cast. Standard four-inch stockinette was placed on the tube, followed by a one-ply layer of Webril padding. Fast-setting four-inch plaster was applied with uniform thickness. Eight, 12, and 16 layers of plaster were evaluated. Water temperatures were 24, 32, and 40 degrees C. Tests included clean water and water with 10 percent plaster residue. A number of tests resulted in temperatures that occurred for long enough to cause a second or third-degree burn.

Skin temperature is strongly affected by the number of plies. High enough temperatures are sustained for sufficient durations to cause burns with thicker casts. Increased temperature of the dip water caused progressively higher skin temperatures and greater probability of severe burns. Use of a pillow, which interfered with ventilation, tended to lead to severe burns. Residue in the dip water did not create higher skin temperature.

Subsequent testing with extra-fast-drying plaster showed that temperature was increased to an average 69 degrees C. Peak temperature lasted two to four minutes, but the temperature exceeding 50 degrees C lasted for 15 to 24 minutes, long enough to cause third-degree burn.

Potential sources of burns are dip-water temperatures hotter than 24 degrees C, cast thicker than 8 plies and inadequate ventilation of the newly applied cast. Thicker casts have greater temperature increase because, while the hydration reaction is occurring, heat is evolved. Excess moisture on the surface, however, lowers the temperature due to evaporation. (Newer materials may also cause thermal damage. Scotchcast reaches a peak of 33 degrees C., Deltalite, 37, and Ultracast 39.)

The Burke system designed to improve the function of patients with bilateral arm and hand paralysis consists of four mechanical modules. The shoulder-arm module is a commercially available spring-loaded linkage with proximal ball-bearing housing attached to a standard balanced forearm orthosis bracket. The distal bearing housing contains a receptacle for forearm modules. They are standard balanced forearm orthoses with adjustment mechanism to determine the optimal pivot point of the forearm swivel, and a multi-position elbow dial to stabilize the arm. The wrist module is a custom-made thermoplastic stabilizer splint attached to the forearm trough. Terminal modules are attachments to the wrist support, including a universal pocket, a cup terminal for joystick wheelchair control, and other self-help devices. Key grip orthoses, either wrist-driven or externally powered, have supplanted two and three-jaw chuck prehension orthoses.

Prescription evaluation includes kinesiological and functional assessment and sensory testing with various modules. The system is suitable for patients with bilateral paralysis resulting from lower motor disease, upper motor lesions, rheumatoid arthritis, and primary muscle disease. It is not appropriate for patients with severe spasticity or those who cannot tolerate sitting with the wheelchair backrest at least 45 degrees. A friction damper may make the system beneficial for some individuals with ataxia or other incoordination. Shoulder and arm muscles should be at least poor minus in strength. No active elbow power is required.

The system enables patients to achieve independent wheelchair mobility, mealtime activity, grooming, and writing. It also allows maintenance of shoulder mobility, countering contractures. The system overcomes the problems previously encountered with externally powered exoskeletal orthoses. The present orthosis requires compulsive attention to details concerning evaluation, application, and adjustment of components. Four dozen have been fabricated.


Five months after injury an 18-year-old man with C4 complete quadriplegia was dependent in electric wheelchair propulsion. Manual muscle testing revealed good-plus upper trapezius function and Poor-minus middle and lower trapezius, deltoid and biceps function was trace. He was initially placed in a chin-controlled wheelchair. The control box in front of his face was considered detrimental to the patient’s self-image. After two weeks, during which time he engaged in an exercise program to strengthen the shoulder girdle and upper limb musculature, he changed to a variable speed right-hand control. The stronger right arm was placed in a balanced forearm orthosis. An elbow orthosis was set at full extension and the wrist orthosis was set at neutral position. The BFO trough was connected to the toggle switch of the wheelchair hand control. One week later the patient had gained sufficient control so that the elbow orthosis could be discarded. After another two weeks he no longer needed the BFO. A T-stick was placed on the toggle switch with the hand secured to it by a Velcro strap. One month later the wrist orthosis and Velcro strap were removed, and two weeks afterward, the T-stick was eliminated. The toggle switch was lengthened to increase leverage, then was reduced to standard size as the patient improved control. The patient lifted his arm onto the armrest by shifting weight to the opposite side and contracting the ipsilateral upper trapezius, biceps and deltoid. Subsequent training on the left side proceeded rapidly, resulting in independence in wheelchair operation in 3 weeks. Followup revealed that the patient eventually learned to feed himself independently with a wrist orthosis. He is also learning to propel a manual wheelchair using rims with pegs.


The Krusen Limb Load Monitor was designed to provide auditory feedback whenever weight exceeding a predetermined magnitude is placed on a force plate inserted in a shoe, thus aiding patients to increase unilateral weight bearing or walk with more equal bilateral weight distribution and stance times. The clinician can examine a voltage signal on an oscilloscope or strip-chart recorder. This enables measurements of the loading and temporal elements of stance. Total cost for two LLMs with three sizes of shoe inserts and a quality two-channel strip-chart recorder is approximately $2500. Recordings take less than fifteen minutes, and analysis another quarter hour. The study was intended to determine the reliability of the LLM by comparing its output with that of a stationary force platform. Data were obtained from three healthy adults.

Separate trials of 10 steps revealed slight differences between LLM and platform for all measures except time to maximum force at push-off. One subject had a wider range of temporal values, while another demonstrated a slightly wider range for loading measures. The LLM and platform are thus not measuring precisely the same values for each temporal and force characteristic of stance. LLM values are an average 3.6 milliseconds less than platform for total stance time measures. Sources of errors are difficult to explain, inasmuch as data analysis suggests that a consistent bias is not operating within or across subjects. A source of variable error may be related to the fit of the shoe insert within the shoe. The subject with the best fit showed the smallest mean differences between LLM and platform values.

Other investigators have shown that median stance duration for normal persons is 630 milliseconds. The percentage error of the LLM is only 0.57 percent of stance time, not of clinical importance. Percentage loading error is 11 percent which may be clinically relevant.
Energy Expenditure During Ambulation with Ortho Crutches and Axillary Crutches: Catherine Hinton (Saint Thomas Hospital, Nashville, Tennessee 37204) and Karen Cullen (Brigham and Women's Hospital, Boston, Massachusetts 02115) Physical Therapy 62:813–819, June 1982.

Thirteen healthy men were each provided with a pair of wooden double upright axillary and a pair of aluminum single upright Ortho crutches, suitably adjusted. Subjects practiced nonweight-bearing with each type. Three electrocardiographic electrodes were placed on each subject's chest to monitor heart rate via telemetry. Blood pressure was measured with a sphygmomanometer attached to the arm. Oxygen consumption was calculated from expired air collected in a portable calorimeter. Subjects walked for 11.5 minutes unassisted, then with crutches. Trial order was randomized. They ambulated at self-selected velocity on a level 300 feet long circular hallway.

Mean velocity for unassisted ambulation was 66 meters per minute. With crutches, velocity slowed to approximately 43 meters per minute with no significant difference between crutch type. Oxygen consumption was significantly greater with crutches than with unassisted ambulation when judged by time and also by distance. Consumption during the first 2.5 minutes of walking was significantly greater with the axillary crutches. Consumption during later portions of walking was not affected by crutch type. Heart rates were also greater when subjects walked with crutches. Additional significant differences were found between crutch types at the beginning and concluding periods of walking; at both times the axillary crutch was associated with higher heart rate. Blood pressure was higher when individuals used crutches, although the type of crutch did not affect the results.

Possible explanations for the differences in energy consumption include differences in work required to perform isometric hand grip, stabilize the scapula or thorax, and balance, lift, or extend the elbows during push-off.


Thirty-one subjects each had crutches fitted and gait instructions. They walked across a force plate which measured horizontal crutch reaction force. They also stopped with the crutch at the walking angle at which time a plumb line was dropped to enable calculation of the angle of the crutch with the horizontal plane. The investigators present an experimental and analytical technique for determining dynamic loads exerted at the crutch handles and at the axilla.

The peak dynamic load at the hands varies from a minimum of 1.14 to a maximum of 3.36 times body weight, averaging 1.84. The walker thus must support an additional, time-varying inertial load arising from the acceleration and deceleration of his body during swing-through. The peak dynamic horizontal force on the axilla is much smaller, averaging 7.5 percent. No correlations were found between forces at the hands and axilla and factors such as height, weight, swing-through time period and crutch angle.

Such data can be useful in selecting axillary crutches with sufficient padding to minimize pressure and predetermining whether a patient is strong enough to use crutches.


Patients with cerebral palsy, muscular dystrophy, spina bifida, and traumatic head injury often cannot maintain the upright position in an ordinary chair. Commercially available seating with adaptive parts may not meet specific needs. Goals for seating include comfort, orthopedic support, functional positioning, durability, socially acceptable appearance, and economy. The current approach for custom fabrication was initiated by Scimedics, Anaheim, California in collaboration with the Demonstration of Service Modalities Project of the University of Alabama. It produces a coated, rigid polyurethane support which can be fabricated in local treatment facilities by physical therapists with an initial one-time equipment cost of $150 for a portable air compressor, vacuum gauge, and latex tubing, and an additional cost of $50 to $150 per chair for a large rubber weather balloon, polyurethane beads, epoxy resin and hardener, and vinyl latex paint.

Fabrication begins with determination of the patient's special seating needs, such as a short thigh support for the shorter leg of the individual with length discrepancy. Epoxy resin and hardener are mixed and poured into the balloon which has been filled with beads. The tubing is attached to the balloon and the vacuum pump which pulls vacuum equal to 0.5 inches of mercury. The balloon is placed in the chair and is hand molded to approximate the needed support features. The patient is then seated and the support's contours are modified. When desired support is attained, vacuum is set to 1 inch of mercury. The patient is removed and the pump discontinued after five to six hours by which time the epoxy will have cured. Eight coats of thickened paint finish the support.

Fifty-one supports have been fabricated and evaluated over a two-year period. They have been molded to school chairs, wheelchairs, infant seats, and other seats. No support has broken, nor was there difficulty in transferring the patient into or out of the seat. No patient experienced skin breakdown. The seats can be adjusted to accommodate growth by cutting and repainting the mold.


Investigators quantitatively mapped pressure distribution of 65 normal subjects and patients using 64 special transducers on an aluminum plate. Subjects sat on the scanner so that ischial tuberosities were equidistant from the center line. Hands were supported lightly in front of the torso, and
feet hung free. All subjects were evaluated at two consecutive readings, with results reproducible to within 3 percent. The normal individual sits with even pressure over both ischial tuberosities, both thighs and sacrum. Each tuberosity sustains approximately 18 percent of the load, each thigh 21 percent, and the sacrum bears 5 percent. The remaining pressures were distributed evenly throughout the sitting area. The center of pressure is in the midline in front of the perineum.

Case reports of three patients with uneven pressure distribution are presented. A young adult traumatic T10 paraplegic developed scoliosis with loss of lumbar lordosis. He revealed posterior weight shift, excessive loading on the lower side of the pelvic obliquity, and three times the normal sacral pressure. A full thickness sacral decubitus ulcer developed on the lower side. A spastic diplegic adolescent revealed a typical distribution pattern, with weight borne posteriorly by the sacrum, one tuberosity, and the contralateral thigh. A woman with right hemipelvectomy showed pressures toward the left and anteriorly onto the thigh.

Pressure sores are caused by shear and/or compression forces, especially the latter. The amount of pressure is proportional to the weight transferred to each point of contact between the body and the resting surface and the duration of loading and is inverse to the area of contact. Poor trunk balance, particularly when complicated by fusion of an incompletely corrected scoliosis and pelvic obliquity, aggravates the risk. The seated paraplegic needs lumbar lordosis to shift body weight forward for balance and to distribute pressure to the thighs.

Computerized pressure scanning may prove valuable in the assessment and management of sitting problems.

**PUBLICATIONS FROM FOREIGN SOURCES**

Papers originating in overseas laboratories and institutions.

**Level Selection in Leg Amputation for Arterial Occlusive Disease: A Comparison of Clinical Evaluation and Skin Perfusion Pressure:** P. Holstein (Department of Clinical Physiology, Bispebjerg Hospital, Copenhagen, Denmark) Acta Orthopaedica Scandinavica 53:821–831, 1982.

A prospective study over a 2 1/2-year period was conducted with all 99 patients referred for measurement of skin perfusion pressure (SPP) before amputation. Most surgery was at the below-knee level. SPP is the external counter pressure just sufficient to stop washout of intradermal depot of radioactive isotopes 10 cm distal to the knee. Measurements were made pre- and post-operatively on both legs. Surgical level was selected according to the general condition of the patient, extent of any skin lesions, level of temperature demarcation, presence of pulsations, and bleeding at operation. All below-knee amputations had plaster rigid dressings. Most patients walked with crutches without distal prosthetic components. Nearly all patients with SPP below 20 mmHg failed to heal at the below-knee level, primarily because of deep necrosis. Half the patients with SPP of 20 to 30 mmHg had to be revised from below-knee to above-knee level. Nearly all patients with SPP above 30 mmHg healed below the knee. All patients selected for primary through- or above-knee amputation who survived the operation healed.

Clinical signs of poor circulation were present in some patients. Skin temperature demarcation line was found in five cases. The skin area above the demarcation constitutes a borderline zone regarding healing. Four patients had primary above- or through-knee amputation, and the fifth did not heal at the below-knee level. Cyanotic limbs had low SPP, confirming severe ischemia. Ulcerations, however, did not necessarily indicate ischemia; their appearance regarding granulation tissue, necrosis, localization, size, and depth demonstrate tissue viability. Pulsation in the popliteal artery signifies an almost 90 percent chance of healing below the knee. Vigorous bleeding at the operation helps to maintain many below-knee levels. Healing of below-knee amputations was significantly correlated with the SPP in diabetics and non-diabetics; infection rate was approximately equal in both groups. Wound healing failed in above half of the infected below-knee amputations. No significant difference in healing rate could be attributed to differences in surgical technique (sagittal vs. long posterior flap).

**Measurement of Maximal End-Weight-Bearing in Lower Limb Amputees:** B. M. Persson and E. Liedberg (Department of Orthopaedic Surgery, Lund University Hospital, Sweden) Prosthetics and Orthotics International 6:147–151, December 1982.

Sixty-nine amputees were evaluated with regard to the amount of force each could exert with the end of the amputation limb on a spring weighing scale. The average of the group was 67, the average body weight was 64 kilograms.
The average maximal end-weight-bearing was 13 kilograms. Amputation through the hip, knee, and ankle had much higher end-bearing tolerance than transmedullary amputations. Above-knee amputees tolerated 13.7 kilograms, compared with 11 kilograms for below-knee amputees. Ischemic below-knee amputees had a mean of 10.7 kilograms, compared with the average 13.9 kilograms tolerated by other amputees. Men, heavier patients, and those who were active walkers had significantly higher tolerance. Diabetic patients, lacking protective sensibility, tolerated more end-loading. Individuals with stump pain had less tolerance than those with phantom pain. End-bearing tolerance may thus be an objective method to differentiate pain problems and to refine socket distribution of pressure during casting. Tolerance was not statistically related to pointed or rounded stump contour.

Through-knee amputations tolerate more loading than Syme, and through-hip more than through-knee. Disarticulations at all levels tolerate several times higher loading than transmedullary levels do.

End-loading tolerance did not increase significantly with duration of amputation. The total contact socket should distribute weightbearing differently in individual individuals, allowing one to load at least body-weight on the prosthesis without pain or skin disturbance.


Five controls (men) and four amputees were tested while walking at various speeds in a laboratory equipped to yield precise stereophotogrammetric data from markers on the acromial processes and iliac crests. Compared with normal gait, amputee gait has larger amplitude of all rotations, the upper body rotates more and moves asymmetrically. The pelvis moves abnormally, elevating, rather than dropping, on the side of the swinging leg. Such motion can be correlated with the passive prosthetic ankle and reduced efficiency of abductors on the amputated side. Hip hiking gains clearance for the swinging prosthesis. During stance on the prosthesis, the sound hip elevates and the trunk bends toward the prosthesis to make equilibrium easier, thereby decreasing abductor effort. Among normal subjects pelvic horizontal rotation is highly variable. Amputees display forward pelvic rotation during midstance. Forward movement of the normal hip during prosthetic stance is opposed by knee stability problems, reduced or absent muscular efficiency, and prosthetic ankle passivity. Absence of active prosthetic push-off reduces pelvic rotation during stance phase on the sound leg.

Horizontal rotation is also altered in the amputee because the prosthetic ankle prohibits pushing the hip forward during deploy, and prosthetic knee stability requires that hip extension on the amputated side during early stance be inhibited.

Differences in segmental rotations, especially at pelvic level, are apt to be correlated with increased metabolic costs and increased mechanical load on the spine. Harmonic components analysis also suggests the need for active prosthetic knee and ankle mechanisms to ensure normal hip movement. A knee-ankle mechanism which could dorsiflex the foot during knee flexion would reduce hip elevation during prosthetic swing and a knee equipped with an energy recovery mechanism could perform knee flexion-extension during early stance to reduce knee stability problems and allow improvement of horizontal pelvic rotation.


The Yugoslavian underknee peroneal stimulator (FESE-L2) was used on rehabilitated hemiplegics. The device consists of a small stimulator unit attached to an elastic knee support. External electrodes are fixed in the elastic and are placed over the tibialis communis in the popliteal fossa and behind the fibular head over the peroneal nerve. The stimulator is controlled by a heel switch in the shoe. Heel-lift switches the stimulator on until heel strike occurs or a duration of 3 seconds passes, whichever is less. The stimulator has a fixed delay time and provision for variable delay. The first delay occurs at the start of the stimulus, allowing plantar flexors to continue activity. If invertor and plantar flexor spasticity persists strongly, stimulus delay can be omitted. The second delay, at the end of stimulus trigger is not adjustable; it allows the dorsiflexors to be activated from heel strike until foot flat.

Kinetic and kinematic data were collected on a normal subject, three patients not receiving FES was eleven with FES. Television cameras recorded frontal and lateral displacements and a Kistler force platform recorded postural sway measurements of vertical and shear forces and their moments were made on the force platform as subjects stood with eyes open and then closed.

Most patients treated with FES showed either marked or barely assessable improvement. It is not clear, however, how patients who would benefit from FES can be identified. Although hemiplegics are clumsy, they learned to don the device quickly and satisfactorily. Postural sway measurements may reflect progress more accurately than instrumented gait analysis which tends to exhibit great variability from step to step.


Complete rehabilitation requires that a potential ambulator be able to stand up without the help of another person. Functional electrical stimulation has proven an efficient means of strengthening disused atrophied muscles of spinal cord injured patients, preventing further muscle atrophy, improving blood flow, and preventing contractures. By means of a two-channel electrical stimulator, a paraplegic can achieve more than one hour of security by locking the knee joints through stimulation of both quadriceps. The present study involved developing a standing-up procedure in which only the knee extensors needed stimulation. The patient will assist the FES by lifting himself with his arms.
Torques in the hip, knee and ankle joints of a healthy subject were measured by means of stroboscopic photography and force plate measurements. The hip-joint moment increases rapidly to a maximum 160 Nm and must be compensated for by both hip extensors. The knee moment also increased rapidly to two Nm and must be counterbalanced by both knee extensors. Much lower torque occurs in the ankle joints, for which plantar flexors are responsible. When the trunk is leaned far forward, maximum hip-joint moment remains unchanged, while knee torque decreases to 160 Nm and at the ankle, first the dorsiflexors pull the knee forward, then plantar flexors compensate for the external joint moment.

The paraplegic compensates for hip torque by use of the arms and trunk muscles. As the values of ankle moment are small, no stimulation need be applied. FES is reserved for knee extensors. Knee joint forces during standing-up with arm support are half that without the help of the arms, thus reducing the amount of stimulation required.

Nine complete paraplegics have completed a muscle training process lasting 2 to 3 months, and received a two-channel FES for home use. Five were able to stand for more than one hour. Four could stand for a few minutes only before fatigue of electrically stimulated muscles became noticeable. For rising from the wheelchair at home, the patient uses a piece of solid furniture at one hand and crutch in the other. He turns on the stimulator, which has a delay mechanism, grasps the support, then rises with the help of knee extensor stimulation. A special supporting frame was built for standing. A delay between turning off the stimulator and the end of stimulation permits secure sitting.


Twenty-five of 128 patients fitted with myoelectric prostheses derive little or no use from them. Early fittings were with the Russian Arm as modified by the Rehabilitation Institute of Montreal; most current fittings were with Otto Bock 6-volt systems. The rejection rate of 9 percent is much lower than that reported for trained and untrained Swedish amputees, 44 percent and 77 percent respectively. Each patient in the current series possesses a cable-operated and a myoelectric prosthesis. Patient ages range from 29 to 67, the average being 41 years. Six were fitted bilaterally. Most were injured on the dominant side. Nearly all are male.

Limited users all wore the myoelectric prosthesis only on weekends. Half wore it passively, emphasizing its cosmetic value in social situations. During the week, the duration of hook use was similar for limited users and for non-users of myoelectric prostheses. The mechanical hand was seldom used, except on social occasions, and appears to be obsolete with the advent of the electric hand. All subjects rejected myoelectric prostheses at work; 84 percent used the hook at work, depending on the cleanliness of the environment, manipulative skill required, force requirement, and degree of public exposure.

Eighty percent of the limited users expressed a strong concern with appearance, while a third of the non-users stated this. Nonusers noted that the hook was as good looking as the myoelectric hand. Some were dissatisfied with glove color and the small, feminine, hand size. Two patients with transcarpal amputations had considerable length discrepancy when wearing the myoelectric prosthesis. Most patients were very aware of the heaviness of the myoelectric prosthesis; small actual weight differences were perceived as greater because of differences in suspension. Most also complained of elbow discomfort. Limited users
prefereed the comfort of the myoelectric prosthesis because of absence of a harness.

The most common reason for not using the hand was fear of damaging the prosthesis or its glove. Others complained of instability of the socket for heavy work; fluctuations in amputation limb size affected Muenster socket fit.

Relative contraindications to myoelectric fitting are bilateral amputation; lack of concern with appearance of harness discomfort; need for prosthetic strength, precision and durability; and transcarpal or metacarpal amputation.

**Bracing and Supporting of the Lumbar Spine:** S. Schroeder, and others (Department of Orthopaedic Surgery, University of Bonn, West Germany) Prosthetics and Orthotics International 6:139-146, December 1982.

Low back pain affects 80 percent of all persons during their lifetime, most of whom return to work within 3 months. Five percent, however, never return to work. The syndrome includes lumbosacral strain, facet syndrome, herniated or degenerative disc, spinal stenosis, and unstable functional unit, as well as spinal degeneration and fatigue, spondylitis, and tumor. Most cases result from degenerative disc disease. The motion segment consists of a disc, the adjacent vertebral bodies, and ligamentous capsule. Each vertebra can be divided at the posterior border of the body into an anterior supporting and a posterior motion control element. Abdominal muscles comprise four-fifths of the abdominal circumference.

Lumbar bracing is mostly for degenerative spinal disease. Braces are either corrective or supportive, although only supportive orthoses are used for low back pain. Nearly all the braces prescribed at the clinic were elastic Tigges or Bauerfeind nonatrophic lumbar supports. Lindemann semielastic and Hohmann rigid-elastic overbridging braces were prescribed for fewer than one-fifth of the patients. Lumbar supports replace the physiological function of the abdominal wall, relieving the weight-bearing spine, and reduce flexion and extension. Supports do not reduce axial rotation nor lateral bending.

The Tigges and Bauerfeind supports are effective only below L3-4, offering only minimal control of flexion and extension. They are indicated for lumbar osteochondrosis, postoperative control, slight instability, and muscle strain. The Lindemann corset supports the spine below T12, controlling all motions minimally. It is indicated for lumbar osteochondrosis, postoperative control, slight instability, and muscle strain. The Lindemann corset supports the spine below T12, controlling all motions minimally. It is indicated for lumbar osteochondrosis.


Forty healthy adults participated in gait testing on treadmill equipped with force plates which could measure the vertical, transverse, and sagittal forces of each foot separately. Temporal events, such as stance phase and double support time, and step and stride lengths were also calculated. The external work of the gait was calculated from the ground reaction forces and velocities. All persons were tested with bare feet at a fixed speed, 1.11 meters per second. Measurements were recorded for one minute after two minutes of adaptation. Subjects were compared with regard to age and sex. Elderly women had significantly shorter strides than elderly men. Younger women also had shorter strides than young men. Total ataxia was significantly greater among younger women than with young men. Ataxia measured by representing the force curves in each of the measured directions as mean curves "shaded" by one standard deviation. The average standard deviation was taken as the measure of gait unsteadiness. Although statistically significant, the actual value of ataxia is of no clinical significance.

Although other studies have reported significant differences between stride lengths of young men and women and between elderly men and women, the differences reflect height differences. If stride length is expressed as a percentage of height, no difference is found; the percentage is constant at 58 to 59 percent. Similarly no differences are found in stance phase, double support time, or step length when expressed as a percentage of stride length. External work was similar among younger and older persons and among men and women. Gait differences in older persons reported in other studies reflect changes influenced by arthritis and the very high proportion of subjects taking one or more drugs. In the current study, all subjects were healthy and not taking medication. When walking speed is kept constant, no clinical difference in ground reaction parameters are found attributable to age or sex.


Eight men who had chronic low back pain and had been wearing lumbar supports for at least 3 months were compared with 10 healthy men. Subjects moved within a pelvic constraint frame whereby two strings held by a belt intersected at the T12 spine. String length was recorded electrically to provide a two-dimensional record of movement. Intra-abdominal pressure was measured by means of a catheter with a pressure transducer inserted 15 cm. into the rectum. Thermistors measured skin temperature under the support and on the thorax. Subjects were tested wearing (i) no support, (ii) a semi-elasticated narrow corset with padded increase radicular pain because of increased venous blood flow through the intraspinal canal venous plexus, compressing the irritated nerve root.
lumbar insert and rigid anterior section; (iii) a narrow fabric corset with posterior stays; (iv) a long fabric corset extending to the thorax with posterior stays; (v) a leather-covered steel brace with uprights and anterior abdominal pad; and (vi) a polythene jacket.

All supports warmed the lumbar skin, especially those with padding. Several subjects commented that the plastic jacket provided a cooling funnel which reduced added warmth. All orthoses reduced movement in normal subjects. Rigid supports were very restrictive. Fabric ones were less restrictive, especially the long corset. Narrow supports prevented movements by impinging on the pelvis and thorax. Lateral stays reduce lateral movements; front stiffening restricts anteroposterior movement. All supports raised intraabdominal pressure in all postures, but the increases were significant only in four instances: elasticated support when walking; and when subjects sat, the plastic jacket, long fabric, and rigid brace. No significant inter-support differences exist. Pressure increases make orthoses more effective at spinal load relief than forces normally generated by the transversus and obliques which produce a disadvantageous mechanical moment. Spinal supports tended to reduce peak intraabdominal pressure levels. This may be due to transmission of axial load from pelvis to thorax directly through the supports, thus reducing lumbar spinal load which is the stimulus for the intraabdominal pressure reflex.

Spinal supports influenced the patient group in a manner similar to the normal group, but the effects were modified by pain. The long fabric support had no influence on mobility. Narrow fabric supports had some effect, but movement was less than for healthy individuals. The rigid brace had the most effect, indicating the support, rather than pain, was dominant. Orthoses raised intraabdominal pressure a similar amount in the patient group as compared with the normal group, but patients had lower resting pressure, indicating loss of abdominal muscle tone. With supports the patient group had higher pressures than normal subjects, indicating the patients were using the orthosis to increase pressure.