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Control of Movement for the Physically Disabled

Control for Rehabilitation Technology

With 297 Figures

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Non-powered, yet active orthoses can be used to control some degrees of freedom (e.g., the knee cage provides stability in the lateral direction and allows flexion and extension). Active orthoses can use body power of joints that are non-paralyzed to drive the paralyzed one (e.g., tenodesis brace, Figure 4.88). Bodily powered orthoses use the principles and methods described in Section 4.4.1 on bodily powered prosthetics. Externally powered orthoses use actuators to generate joint movement.

Arm and Hand Orthoses

Arm and hand orthoses are designed to provide the posture and control of movement in the shoulder, elbow, and wrist joint, and in addition for controlling the opening and closing of the hand. The mechanics of the arm is described in Chapter 2.

To allow a person with impairment preventing him/her to move objects against gravity (shoulder and elbow movement), an orthotic device for mechanical guidance of the arm counteracting effect of gravity has been designed. The complexity of the developed prototypes has ranged from simple, passive, counterweighted devices that support the arm against gravity [James and Orr, 1984] to complex, powered exoskeletons with many degrees of freedom [e.g., Vukobratović and Stokić, 1981]. Powered orthosis prototypes suffer from difficulties relating to cost, power consumption, weight, user acceptance, and the control interface.

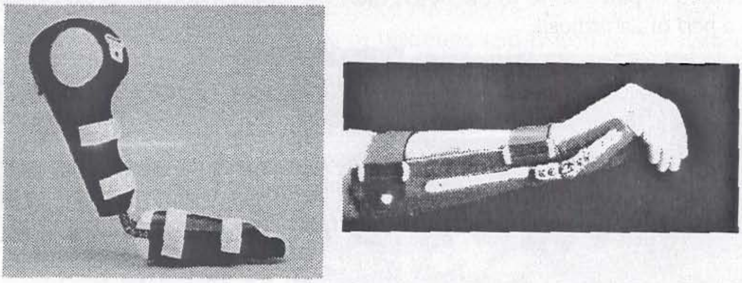

Fig. 4.89: Orthosis to provide control and support to arm joints. Many similar designs, most from prefabricated modules and thermoplastic material (formed to fit the body shape) with Velcro bands are used mostly as temporarily relief from stress and prevention of injury.

Two examples of commercially available orthosis with articulated, however, non-powered joints are shown in Figure 4.89. The two representative elbow and wrist orthoses, which can be locked in a position or where the joint angle can be limited to a certain range, follow the principle of modularity. Prefabricated elements are made in different sizes, and the custom made interface from thermoplastic or composite materials to fit the user. Both orthoses have a spring stainless steel lateral and medial joint constructions, which are attached to acrylic and silastic lamination.

The wrist fixation orthosis is frequently applied in tetraplegic subjects and combined with FES based neuroprosthesis [Peckham and Keith, 1992; Keith *et al.*, 1988].

Lower-Limb Orthoses

Varieties of mechanical orthoses have been used in the past, and they continue to be used. They vary from a Swivel Walker, which is almost like a standing frame with a limited facility to move from place to place, to Ankle-Foot Orthosis (AFO).

Ankle-Foot-Orthosis (AFO)

Only one orthotic system of this variety has been described for ambulatory use in paraplegics [Lyles and Munday, 1992]. The Vannini-Rizzoli Stabilizing Limb Orthosis (V-RLSO) comprises a pair of polypropylene orthoses shaped to fit the lower legs. An orthosis on each side partially encloses the lower leg from about 2 cm below the lower pole of patella to tips of the toes. A pair of specially designed leather boots fit over the orthoses.

The V-RLSO has a flat rigid sole externally, and the inside of the orthosis is 10 to 15 degrees plantar flexed in order to shift the ground reaction in front of the ankle and knee during standing. When using this system, the subject controls static equilibrium by maintaining hips and knees in extended position and head held high. During ambulating, the subject can shift the center of mass forward and then with the help of walking aids is able to move the unweighted foot forward in a pendulum fashion by shifting the upper torso slightly towards the side and forward.

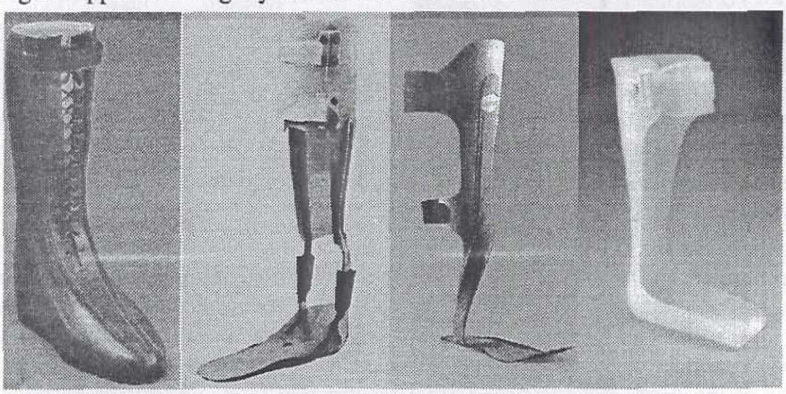

Fig. 4.90: Four types of Ankle Foot Orthosis (AFO). V-RLSO (left), two types of composite material based orthoses (middle two panels), and an AAFO with a spring mechanism for the ankle joint (left).

Most AFO systems are made out of light composite materials (Figure 4.90). Plastic AFO can support the shank allowing limited dorsiflexion, yet it can be made very stiff in order to immobilize the joint. Andrews *et al.* [1987, 1989] introduced an orthosis proven effective for standing, and it is named floor-reaction orthosis (FRO). The FRO forces the shank into a five degrees of flexion, bringing the hips forward; thus, the center of gravity is in front of the center of pressure. This leads to a so-called "C" posture, which is intrinsically stable and does not need the knee extensors for standing. The C posture may lead to a knee joint injury; the knee cage could protect the knee.

Knee-Ankle-Foot-Orthosis (KAFO)

Knee-Ankle-Foot-Orthoses, as the name suggests, provides stabilization of the knee along with stabilization of the ankle and foot. This is achieved by application of three forces. The first is applied in front of the knee preventing it from buckling under the body weight, and the other two forces are applied in an opposite direction to that of the first, *i.e.*, posteriorly, one at the upper posterior thigh and the other at shoe level.

In the past, KAFOs were extensions of below knee double iron orthoses. Traditionally they were attached at the lower end to a firm lace up shoe through a stirrup or a round caliper. The ankle joint provided medio-lateral stability; it was similar to traditional AFO, in that, it had a posterior stop to prevent plantar flexion but allowed dorsiflexion. A knee joint, with or without lock, was added at the level of the knee. Most KAFOs used for paraplegic locomotion have knees with locks so that they are locked during standing or walking, but could be unlocked during sitting. The simplest lock is a drop lock, which is unlocked after sitting down. A bale or Swiss lock provides easier locking and unlocking mechanism. The most essential bands on a KAFO are posterior calf and upper thigh bands. A large variety of other combinations of thigh and calf bands has been described. However, Lehmann and associates [1969, 1976] showed that most of these additional bands are often unnecessary. Some KAFOs are fitted with pelvic bands in order to help stabilize the pelvic girdle. Theoretically these orthoses are Hip-Knee-Ankle-Foot variety (HKAFO) and should not be classed as KAFOs (Figure 4.91).

Scott [1971] described several modifications of traditional KAFOs, making them lighter in weight and giving more stable base with improved balance to the user. Supporting bands were discarded except a thigh band and an anterior rigid patellar tendon strap. The knee was offset eccentrically, and the ball locks have been retained. The shoe plate was extended beyond the metatarsal heads to provide better support. A solid ankle was set at 5 to 10 degrees of dorsiflexion. This KAFO is one of the commonly prescribed orthosis for paraplegics.

Ambulating with knees locked in full extension in a swing-through or a swing-to fashion requires lifting the center of mass up at every step. This increases the energy cost of walking, however, it is an acceptable trade-off with the increased amount of safety due to locked and stable knees. Locked knees allow the subject to lean backwards to stabilize hips. Leaning backwards, with locked knees, places the center of mass of the trunk behind the hip joint resulting in tightening of the anterior hip capsule providing internal stabilization of the hip.

Swivel Walker was established as a means of ambulating for severely disabled children by Motlock and Elliott [1966] in early sixties. Rose and Henshaw [1973] have described an updated and somewhat improved version. The Swivel Walker allows humans with spinal cord injury at thoracic and cervical levels up to C6 to maintain the upright posture without the use of any other walking aid. The device consists of a rigid body structure that provides stability to hips, knees, and ankles via a frame with three points of fixation. The extrinsic stability is provided by a pair of swiveling footplates mounted beneath the body frame. These feet plates provide mobility. The center of mass of the system projects slightly forwards of the foot-plate bearing center; hence, when inclined to one side it causes progression of the center of mass due to the gravity and ground reaction force. Walking is achieved by alternately sideways rocking of the trunk so that one footplate is lifted off the ground, the frame then automatically swivels forward on the contralateral footplate. This process is repeated cyclically with the opposite leg. The orthosis is available in standard sizes and designed in modules that can be interchanged.

The Swivel Walker provides some independence in getting around, the walking is extremely slow, and the energy consumption very high. Humans can walk only on an even level surface. Transferring to upright posture needs to be carried by helpers.

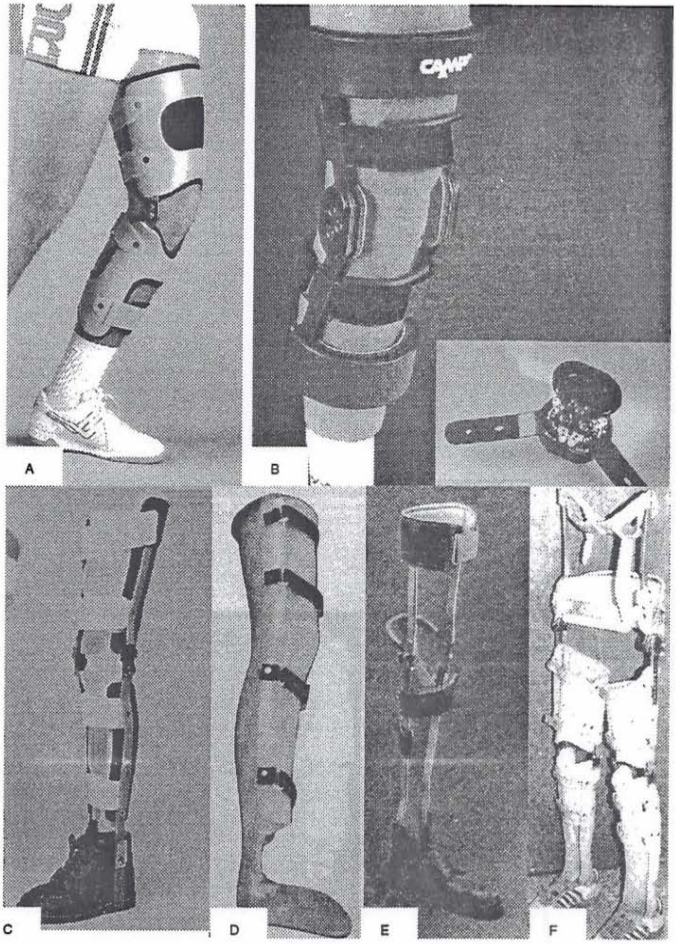

Fig. 4.91: Different types of the knee-ankle orthoses (A and B), Knee-Ankle Foot Orthoses (KAFO) (C, D, and E), and Hip-KAFO (HKAFO) orthoses (F).

Following a thorough analysis of requirements of reciprocal walking for humans with paralyzed legs, Rose [1979] evaluated a system comprising specially designed hip guidance articulations. The orthosis was initially called the *Hip Guidance Orthosis*, yet it is now known as the *ParaWalker*. The orthosis has a rigid body brace that is attached to two KAFOs via the specially designed hip joints (Figure 4.92, left). The KAFOs are in minimal abduction at the hips in relation to the trunk. The orthosis has three support points. A support point on the chest is provided by a leather chest strap. A support point on the buttocks is provided by a polypropylene band attached on either side directly to the joint housing. It has low friction hip joints with flexion/extension stops. A release mechanism for the stops is provided to allow the user to sit down. The KAFOs stabilize the knees and ankles. The knee of either side is held in extension by an anterior strap at the level of upper end of the Tibia reacted by the posterior thigh band and a vertical extension on the rear of the shoe plate. These two together provide the third support point for the main orthosis. The orthotic knee joints are normally locked during walking, but have release mechanisms to enable user to sit down. A shoe plate incorporating a rocker sole is fixed to the either KAFO with an appropriate amount of dorsiflexion.

The system allows reciprocal walking for subjects with thoracic level lesion using crutches. The rigid body brace of the ParaWalker can resist the adducting moments about the stance hip in the coronal plane. A push on the crutch to tilt towards the stance side allows the subject to clear the swing leg. As soon as the swing leg is clear off the ground, the gravity causes it to move forward through pendulum action causing flexion at the hip. This is possible because the center of mass of the leg lies behind the hip joint axis when it is in extension. The flexion is limited by stops on the hip joints. When the swinging leg contacts the ground, it is in flexion. Force is required to raise the trunk upwards and pull it forwards during stance. The force has to be generated by the user, through the same crutch, which tilts him sideways.

Reciprocating Gait Orthosis (RGO). The principle of linking hips together to allow reciprocal movement of both hips in HKAFO was first developed in the late 1960s. At the Ontario Crippled Children's Center in Toronto, Canada, Motlock and his colleagues described a mechanism based on the use of gears [Motlock, 1992]. In England, David Scrutton [1971] described the use of a pair of Bowden cables to control the motion of orthotic hip joints in a reciprocating brace with poly-planar hip hinges used for spina bifida children.

The reciprocating gait orthosis, which is a refinement of the original design described by Scrutton, was jointly developed by the Louisiana State University Medical Center and Durr-Fillauer Medical Inc. [Douglas *et al.,* 1983]. The orthosis includes trunk support, a pelvic assembly and bilateral KAFOs. The thoracic support consists of extensions of upper side members of hips up-to or just above the level of the xiphoid process of sternum with 5 centimeter wide adjustable anterior and posterior encircling straps (Figure 4.92, right). The extension prevents uncontrolled collapse of the trunk and yet allows the user to move upper extremities freely. The rigid pelvic assembly consists of a pelvic band covering the gluteal and sacral areas with special thrust bearing hip joints coupled together with a cable and conduit. The KAFOs have posterior offset of the knee joints with drop locks on the lateral sides. The AFOs and thigh sections are made of Polypropylene. The AFOs are each reinforced in the ankle area with an insert made of a composite material to assure resistance against dorsiflexion. The most important feature of the system is the cable coupling of the left and right leg movement. The coupling provides hip joint stability during standing by preventing simultaneous flexion of both hips; yet, it allows flexion of one hip and simultaneous extension of the other when a step is taken. The cables can be disengaged to sit down.

Solomonow and his associates [1989, 1997a and b] described several limitations of the original orthotic system which were encountered during its usage and suggested modifications to the system. In their experience most subjects were able to doff and don the orthosis independently, but were unable to stand up without assistance. Many paraplegics have some degree of Hamstrings contracture and hence their knees cannot reach full extension. Since the automatic knee locks operated only in full extension, these subjects were unable to remain upright. Another major concern was the inability to negotiate ramps or driveways with mild to moderate inclinations.

A ratchet knee joint was incorporated into the design. This knee joint extends freely, but locks in any position short of full extension. This prevents the knee from flexing during the process of standing up even if it is not fully extended and supports the subject's weight. Force applied to the knee at heel strike of the first step brings it into full extension. The ratchet knee joint has a ball type lock which can be operated to allow free flexion to enable the subject to sit down.

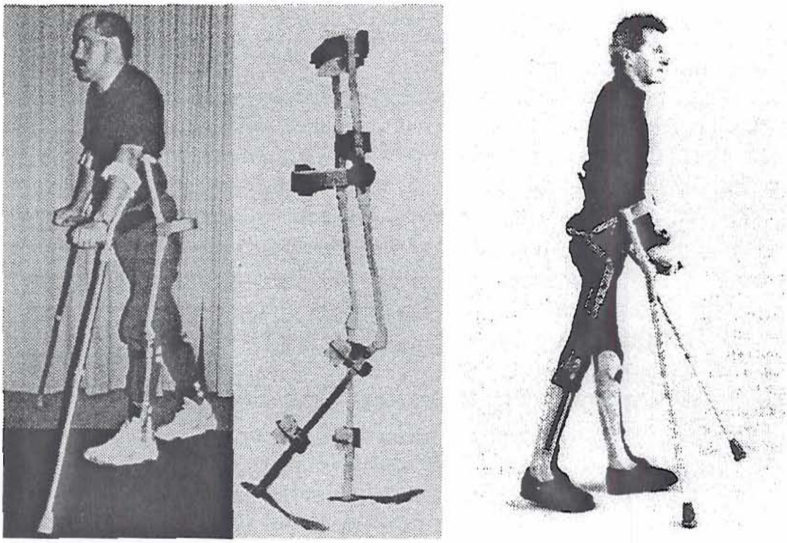

Fig. 4.92: Hip-Guidance Orthosis (HGO) – left, and Reciprocating Gait Orthosis with the Bowden cable connecting the hip joints (middle and right).

A different hip joint within RGO has two locking positions. One is the standard position at full hip extension to attain normal upright posture. The other position allows 20 degrees of flexion from the previous position. This allows the subject to move the center of mass of upper trunk forward so that while walking on the ramp it would remain in front of the support area preventing loss of balance posteriorly.

Whittle [1989] studied orthotic gait of subjects using the RGO and the ParaWalker as a part of comparative biomechanical assessment of these two orthoses. He found similar walking velocities for both orthoses. The total excursion of the leg in the sagittal plane (flexion/extension) was significantly greater with the RGO. However, the stride length was similar for both orthoses, because in the ParaWalker gait, the smaller leg excursion was compensated with greater pelvic rotation in the transverse plane. In addition, during the ParaWalker gait there was significantly greater abduction of hip on the swing side and smaller adduction of hip on the stance side, making it easier to use with the crutches. The subjects were unable to use crutches with the RGO. The force through the legs was a little higher with the ParaWalker. The force through the arms was probably higher with the RGO.

Advanced Reciprocating Gait Orthosis (ARGO). Although the crossed Bowden cables are a simple and effective way to produce reciprocal hip joint movement, they may not be the most efficient mechanical coupling. Since the cables are secured only at each end, some of the energy associated with the active hip extension is wasted in the unwanted cable friction. In addition, since a cable needs to be in tension to transmit large forces effectively, only half of the system is being used at a time. A modified version of RGO is called Advanced Reciprocating Gait Orthosis (ARGO). The hip joints are modified and are interconnected using a single Bowden cable encased in a tube in an attempt to reduce friction. Its major advantage over the RGO is that it allows the user to rise from sitting position and to sit down from standing position with relative ease. The user can stand up directly from normal sitting position with flexed knees without having to extend them manually before commencing to stand up as is necessary with the RGO as well as with the ParaWalker. This is achieved by mounting a compressed gas strut on the thigh side link on each side providing a knee extension moment to augment standing and control hip flexion during sitting down. The hip and knee joints on each side are connected via a knee lock actuating cable so that the hip mechanism releases the knee lock.

During the development of the ARGO, Jefferson and Whittle [1990] tested a subject using the ParaWalker, RGO, and the ARGO in parallel. They found similar motion in pattern and in magnitude between the RGO and ARGO. In the ARGO, however, the pelvis appeared to be shortly stationary at a particular instant during the gait cycle enhancing its jerky movement pattern. The walking with the ParaWalker showed marked differences from that of the other two. It showed greater smoothness before and after movements. The subject's legs remained more or less parallel to each other in the coronal plane allowing better ground clearance.

Motlock [1992] suggested another modification to the RGO known as the Isocentric RGO; the Bowden cables have been replaced with a centrally pivoting bar and tie rod arrangement.

Powered Orthoses

Attempts to design externally powered skeleton to be used to "carry" the subject and move his/her legs following the nature-like pattern characterize the early seventies,

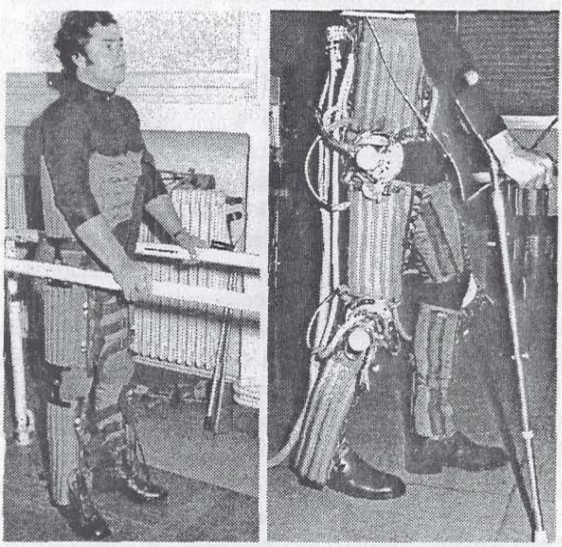

Fig. 4.93: The modular non-actuated (left) and powered (right) exoskeleton (AMOLL) used for training and exercise of the walking. Courtesy of Dr. Pierre Rabishong, France.

when bipedal robots became popular [e.g., Hristić *et al.*, 1974, 1981; Vukobratović, 1974, 1975]. This development resulted with heavy, rigid, cosmetically unacceptable, and inappropriate assistive systems for paraplegics. The so-called *soft orthosis* has been introduced by the company Aerozur, Paris; long tubes that could be filled with compressed air created "high" trousers reaching mid-thoracic level. Once the tubes were blown up, the subject had erect posture. The pressure applied to skin, inability for skin to breath through the costume, need to depressurize the system every time a subject wanted to sit, and essential use of the compressor every time he/she wanted to stand up made the system obsolete.

The modular brace that inherited compressed air tubes, but only to form modules around the shank, thigh, and trunk (Figure 4.93), provided soft interface at points of contacts with the body, yet used metal joints [Rabishong *et al.*, 1975]. This orthosis has been implemented for the gait training in the rehabilitation institution using a master-slave control; the therapist walked with one orthosis (master), while the other (slave) followed the kinematics and carried the paraplegic subject.

A modular orthosis, where the textile cuffs (trunk, thigh, shank) connected the mechanical joints, has been designed for muscular dystrophy subjects and tested in a small number of subjects [Hristić *et al.*, 1981].

The powered Self-Fitting Modular Orthosis (SFMO) was introduced [Popović *et al.*, 1979, Tomović *et al.*, 1978; Popović, 1990a] to be a part of a hybrid assistive system [Schwirtlich and Popović, 1984, Popović and Schwirtlich, 1993].SFMO was designed based on the following principles: 1) modularity; 2) self-fitting to the body; 3) soft interface; 4) self-centering of orthosis joints to the leg joints during walking; and 4) partial powering (Figure 4.94). The modules have been designed to allow the integration into a lateral hip-knee-ankle orthosis, and integration of the left and right side. The integration has been accomplished by wearing modified types of jeans. The SFMO allowed a selection of the joints with a mechanical brake, electrically controllable brakes, and powered joints by so-called cybernetic actuator [Popović M and Popović, 1983]. The SFMO has been evaluated; the results show that the standing and walking, when using passive SFMO, are comparable with functioning with conventional orthoses, however, the use of the powered SFMO was assessed as too complicated and unacceptable for paraplegic subjects.

The advantages of the SFMO are the small weight, modularity, and over all self-fitting. The modularity proved to be a favorable feature; during the treatment modules could be added or removed depending on the needs.

4.3.6 Hybrid Assistive Systems

In order to minimize problems because of the muscle fatigue and to increase the safety, a combination of a mechanical orthosis and FES was suggested. The resulting orthosis is called *Hybrid assistive system or Hybrid orthosis* [Tomović *et al.*, 1973]. The support, stability of the joints, and constraint to unwanted motion of the joints are provided by the mechanical component of the orthosis, while FES provides propulsion. Schwirtlich and Popović [1984] suggested a hybrid orthosis, which consisted of the SFMO and surface electrodes FES to provide the knee extension and swing of the leg during walking (Figure 4.94, left). A four-channel stimulator was used. The user initiates steps by triggering flexion, and this is followed by sensory driven knee extensor activity. The SFMO uses brakes activated by a micromotor. The shoe insole switches, inclinometers, and joint angle sensors provided necessary information for automatic, rule-based control described in Chapter 5.

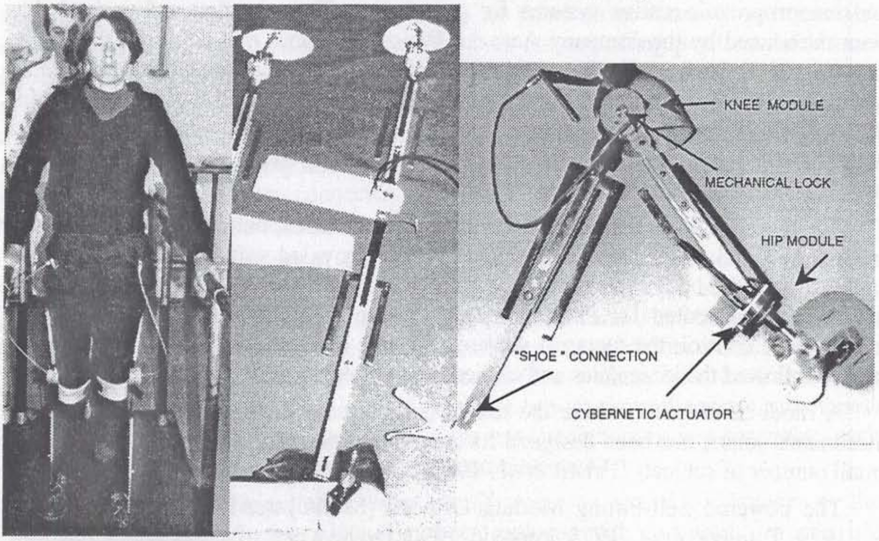

Fig. 4.94: An active, partially powered Self-Fitting Modular Orthosis (middle and right) designed for usage in Hybrid Assistive Systems (left). The cybernetic actuator shown at the hip is applicable for all joints if necessary (ankle, knee or hip) [Popović, 1981].

Andrews and Bajd [1984] suggested two variants of hybrid orthoses. One consisted of a combination of a pair of simple plastic splints used to maintain the knee extension and a two-channel stimulator per leg. One channel stimulated Gastro-soleus muscles, and the other provided flexor withdrawal response. Since the knee was held in extension only dorsiflexion of the foot and flexion of the hip were obtained. The other hybrid system comprised the KAFO and two-channel stimulation per leg. The quadriceps and common peroneal nerve were stimulated on each side. The quadriceps stimulation caused knee extension, and the peroneal nerve stimulation produced synergistic flexor response. The mechanical brace incorporated knee joint locks, which were remotely controlled by a solenoid actuator or a Bowden cable. Andrews [1986] also described a short leg orthosis in combination with FES. A Floor Reaction Orthosis (FRO) was fabricated using high-density polypropylene in a usual fashion, except that the ankle joint was set in approximately 5-degree plantar flexion. This has the advantage of being able to stabilize the knee joint when the ground reaction vector passes anterior to the knee joint axis. The FRO cannot stabilize the leg when the vector passes through or behind the knee joint axis. The FES control system was added to react appropriately, whenever a destabilizing situation arose, and to activate Quadriceps muscles. A sensor was incorporated in the calf strap of the FRO to feedback the status of the ground reaction vector. Andrews and associates [1988] suggested a different way of using the FRO in conjunction with FES. This consisted of a rigid ankle foot orthosis, a multichannel stimulator with surface electrodes, body-mounted sensors, a 'rule based' controller, and an electro-cutaneous display for supplementary feedback. The finite state controller reacted automatically to destabilize shifts of the ground reaction vector by stimulating appropriate anti-gravity musculature to brace the leg. The system also featured a control mode to initiate and terminate flexion of the leg during forward progression. A simple mode of supplementary sensory feedback was used during the laboratory standing tests to assist the subject in maintaining the posture.

In the cases of Reciprocating Gait Orthosis and Hip Guidance Orthosis used in combination with FES, other control methods were used. In the case of RGO the inbuilt mechanism to provide propulsive forces was not adequate. Solomonow and associates [1989] found that ambulating with the modified RGO at the Louisiana State University (LSU), so-called LSU RGO (Figure 4.95), was associated with high energy cost, and most subjects were unable to stand up without assistance. Considerable arm strength was required for the subject to attain an upright position by pressing down on the handlebars of their walker. Most subjects were unable to generate such a force. To reduce the stress over upper extremities and to reduce the energy requirements during walking an electrical stimulation system was designed. Surface stimulation of Rectus Femoris and Hamstrings muscles was used. The stimulation electrodes were incorporated in a plastic polymer cuff.

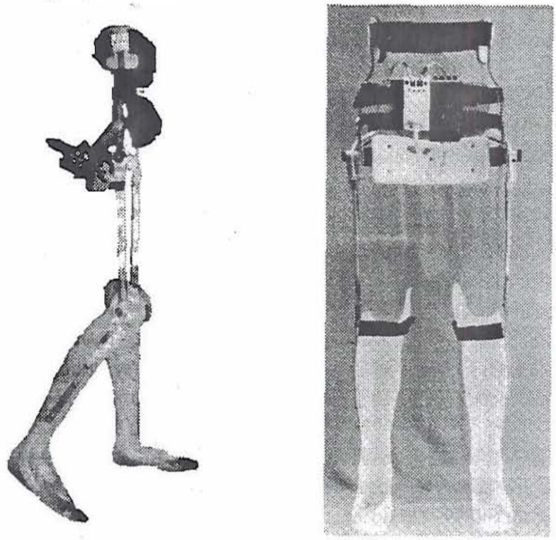

Fig. 4.95: The LSU RGO based hybrid system for restoring standing and walking. The system comprises an RGO, an electronic stimulator, and surface electrodes. FES is added to RGO to enhance movement generated with the upper part of the body, thereby decrease the energy requirements for walking.

The electrode cuffs were secured on the thighs with Velcro straps. A four-channel stimulator was worn on a belt. Finger switches were mounted on the handle bars of the walker. The subject controlled the stimulation himself. To stand up without assistance the subject stimulated both Quadriceps and both Hamstrings simultaneously. The Quadriceps extended the knees, and the Hamstrings extended the hips pushing the subject in an upright position. During walking the subject simultaneously stimulated the right Quadriceps to produce right swing and left Hamstrings to produce left forward push, then again simultaneously stimulated the left Quadriceps and right Hamstrings to produce left swing and right forward push, respectively. Phillips [1989] described a similar system; in addition to the stimulation of Hamstrings, the stimulation of ipsilateral Gluteal muscles was also suggested to improve hip extension of the stance leg. Four two-channel stimulators were used. The entire bulk of Quadriceps was stimulated using three-electrode configuration with two channels of one stimulator for each leg. The other two stimulators were used to stimulate Hamstrings and Gluteal muscles on each side.

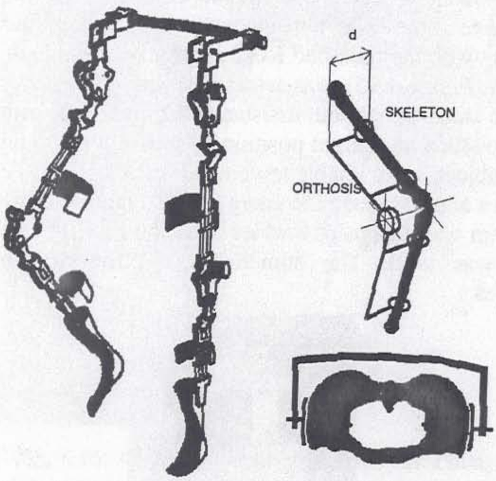

Fig. 4.96: The sketch of the Controlled-Brake Orthosis (CBO), an orthosis with brakes in the knee and hip joints (left), and model of parallel action of the orthosis and the skeleton (right). Adapted from Goldfarb and Durfee, 1996, © IEEE.

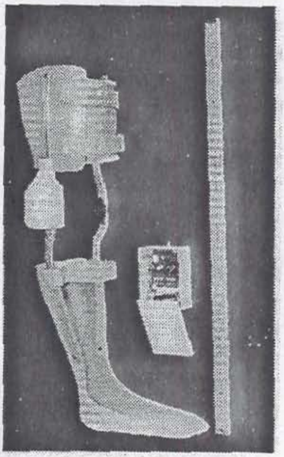

Fig. 4.97: The Dynamic Knee Brace System (DKBS) and the control circuitry. A wrap-spring clutch mechanism is controlling the knee joint state at any angular position. Adapted from Irby *et al.*, 1999, © IEEE.

A research group in the Netherlands is currently working on the development of Modular Orthosis with multichannel Surface Electrical Stimulation (MOSES) system. It is proposed that the orthotic component of the system will be of HKAFO variety with a modular construction. The trunk support will be removable to facilitate wheelchair usage and transfers to and from wheelchairs to other places in a sitting position. In addition, the knee joint mechanism will provide automatic unlocking of the joint during the respective swing phase while remaining locked during the respective stance phase. Actuation of this mechanism will be provided from the hip on the same side. This is in an attempt to reduce the energy requirements of locomotion and at the same time improving kinematics of locomotion. Electrical stimulation of the stance side hip extensors will provide propulsion. The system is currently undergoing clinical evaluation [Hermens *et al.*, 1994].

An important element in hybrid assistive systems is the possibility of controlling joints of the orthosis. Goldfarb and Durfee designed and evaluated a magnetic particle brake for FES-Aided walking [1996].

The design of the system considers the feature that has been neglected by most other researchers; the fact that the external skeleton operates in parallel with the body skeleton (Figure 4.96) and that the movement of the orthosis is not identical to the movement of the body. The orthosis will generate forces that will be transferred to the body through the visco-elastic connections, and the orthosis will move relatively to the body even during the quite standing, *i.e.*, swaying around the vertical posture [Popović, 1981].

Irby *et al.*, [1999a and b] suggest a wrap-spring clutch to be used within a KAFO system (Figure 4.97). The wrap-spring clutch is a principle that is proven in transmission of rotational movement. The effects of friction when a flexible body is surrounding a non-movable body are amplified in the exponential manner; thus, a small force at one end can hold a large force at the other end. This design allows minimization of the device, and over all small energy consumption when compared with other brake systems used in orthotics. There are two braking mechanisms used for the SFMO: a spring loaded pin brake that can be released by hand or micro motor, and the use of cybernetic actuator where the braking is applied using the telescopic mechanism (ball-screw).

4.3.7 Issues Impeding the Effective Use of Neuroprostheses

Restoring movement by activating the paralyzed neuro-musculo-skeletal structures is one of the promising methods, especially when combined with extensive neurorehabilitation, which is goal oriented intensive therapy. The usage of neuroprosthesis allows many humans with disabilities to improve their quality of life. NP is an external system that interfaces the preserved bodily functions, and provides necessary drive to the paralyzed structures. The technology available for NP is improving in parallel with the body of knowledge how and what to assist; thus, interface to natural mechanisms becomes more appropriate. There are some neuro-musculo-skeletal issues that has to be dealt with in the future in order to make NP suitable and applicable for many more users and their daily independent life. Available NP in many cases do not provide adequate function because of the inherent problems to the neuro-musculo-skeletal system when exposed to external activation after the injury: muscle fatigue, reduced net joint torque when generated by NP in comparison with the torque activated by CNS in able-bodied subjects, modified reflexes, spasticity, joint contractures, osteoporosis, and stress fractures [Commar *et al.*, 1962].

NP activate synchronously motor units at frequencies above the physiological values typical for natural control, thereby cause muscle fatigue [Bigland-Ritchie *et al*, 1979; Bigland-Ritchie and Woods, 1984; Kralj and Bajd, 1989; McNeal *et al.*, 1989; Solomonow *et al.*, 1983]. In principle, physiological sequence of stimulation of the muscles can be generated [Baratta *et al.*, 1989], yet it has not been integrated in the available NP. Control schemes are still not good enough to limit the duration of stimulation and thereby minimize the chance for muscle fatigue to occur. The modeling and methods that can overcome the issues related to muscle fatigue are discussed in Chapter 5.

Spasticity. Redundant muscle groups can not be externally controlled in the way identical to biological, that is central nervous system functions have not be successfully cloned so far. A central nervous system injury results in modified reflexes [e.g., Stein and Capaday, 1988], so numerous unexpected situations may occur, resulting in inappropriate antagonist contraction. In addition, the CNS changes are responsible for the reorganization in tonic and phasic properties of different muscle groups, *i.e.*, spasticity [Stefanovska, *et al.*, 1989; Dimitrijević and Nathan, 1971]. Some subjects with paralysis cannot benefit from NP because it is impossible, or extremely difficult to create functional movement because of the spasticity.

The definition of spasticity is a subject of diverse opinions. A frequently used definition is that of Landau [1980] which includes: 1) decreased dexterity, 2) loss of strength, 3) increased tendon jerks, 4) increased resistance to slower passive muscle stretch, and 5) hyperactive flexion reflexes (flexor spasms). Knutsson [1985] described almost unlimited inter individual variation in subjects with spastic paresis; therefore a detailed description of each particular subject's motor dysfunction is required. Studies applying refined biomechanical and electrophysiological measures have revealed a significant change in the passive properties of the spastic subjects [Herman, 1968; Thilmann *et al.*, 1991; Sinkjær *et al.*, 1993; Sinkjær and Magnussen, 1994]. Based on such observations, the idea that changes in the intrinsic muscle properties are largely responsible for spastic hypertonia has been accepted by some researchers [Dietz *et al.*, 1981; 1995; Hufschmidt and Mauritz, 1985]. Other investigators conclude, however, that the major cause of spastic muscle hypertonus is the widely accepted pathological increase in the stretch reflex activity [e.g., Ashby *et al.*, 1987]. Here, a description obtained by analyzing different peripheral factors responsible for the "muscle tone" to passive stretch in subjects with their muscles relaxed or active is given to document changes in the organization of movement after central nervous system injury.

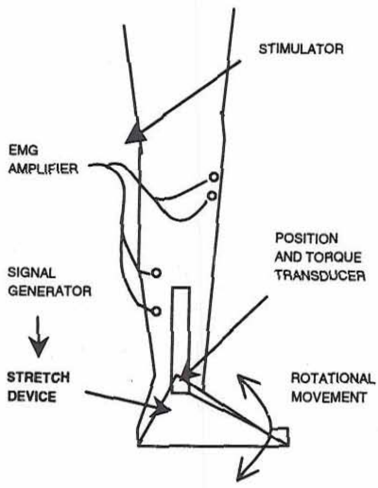

Fig. 4.98: A schema of the setup for studying spasticity during sitting, standing and waling in the plantar and dorsiflexors. For details see Sinkjær, 1997.

The resistance can be divided into: 1) an increase in the passive stiffness of tendons, joints, or muscles [Lowenthal and Tobis, 1957; Herman, 1968], 2) an increase in the intrinsic stiffness of the contracting muscle fibers [Dietz *et al.*, 1981], and 3) an increase in the stiffness mediated by the stretch reflex [Ashby *et al.*, 1987; Thilmann *et al.*, 1991].

To study the importance of the different components, a brief passive stretch can be applied to the isometric muscle of interest during different voluntary activities in able-bodied and spastic hemiplegics. These studies should be performed by imposing well-defined angular displacements (Figure 4.98) around a joint. The measured forces and/or changes in electrical activity of the muscle [Gottlieb and Agarval, 1970; Hunter and Kearney, 1982; Allum and Mauritz, 1984; Sinkjær *et al.*, 1988] can then provide information to understand the pathology, and resolve the problems that it causes.

The increase in ankle joint torque in the spastic and contralateral plantar flexors of a hemiplegic subject after imposing passive dorsiflexion was measured. The total torque increment is the sum of the reflex-mediated torque and the non-reflex mediated torque. The non-reflex torque increment was measured during a continuous electrical stimulation of the tibial nerve innervating the ankle plantar flexors. The electrical stimulation abolishes the stretch reflex [Sinkjær *et al.*, 1988; Toft *et al.*, 1991]. The net

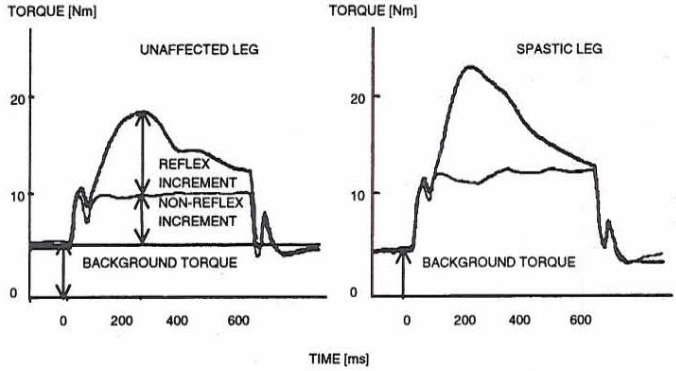

Fig. 4.99: Total and non-reflex torque responses to a stretch of the plantar flexors in a hemiplegic subject: A) unaffected, contralateral leg; B) affected, ipsilateral leg. The non-reflex torque was measures when stimulating the tibial nerve at 10 Hz. This stimulation generated plantar flexion (contraction) and the stretch reflex suppression. Adapted from Sinkjær, 1999, with permission.

reflex torque obtained when the non-reflex torque has been deducted from the total torque exceeded the non-reflex torque with nearly two times in the contralateral (unaffected) and spastic leg (Figure 4.99). The reflex net torque and the non-reflex net torque in the spastic leg both exceeded the ones in the contralateral leg. The non-reflex muscle component part of the torque opposing the stretch stems from the properties of the collagen tissue (passive properties) and the contractile apparatus in the stretched muscles (intrinsic properties). The sum of the passive and intrinsic properties is termed the non-reflex properties. The passive stiffness at a joint is reported to be several times bigger in spastic subject than those in able bodied subjects depending on the joint, joint position, subject group, and applied method [Thilman *et al.*, 1991; Sinkjær and Magnussen, 1994; Malouin *et al.*, 1997; Mirbaghei *et al.*, 1998]. Sinkjær and Magnussen [1994] found an increase in passive stiffness of up to 40 times in the spastic ankle during fast passive dorsiflexion, and an increase of about 100 percent in the passive stiffness of the contralateral leg.

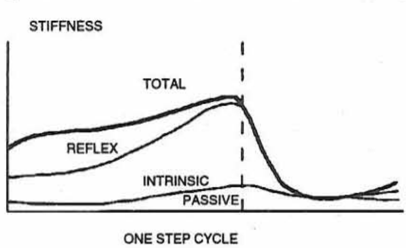

Fig. 4.100: A sketch of the passive, intrinsic and reflex mediated stiffness during walking. See text for details. Adapted from Sinkjær, 1999, with permission.

When the passive stiffness was measured in a more acute stage [Malouin *et al.*, 1997], an increase in passive stiffness of about 39 percent has been found in the affected leg three months after stroke. These increases are all with respect to ankle passive stiffness in age-matched able-bodies subjects. Malouin *et al.*, [1997] found a low correlation between the increased passive stiffness and factors such as the range of movement, the Ashwort score, and the Fugl-Meyer lower extremity motor score. This indicates that the early changes in the mechanical response to stretch in plantar flexors occur without regard to the level of disability. The changes in passive stiffness may be due to changes in collagen tissue, tendons, joint capsules, and the muscles possibly leading to clinically observable contractures [Lowenthal and Tobis, 1957; Herman, 1968; Hufsehrnidt and Mauritz, 1985].

Given the different Young's module of muscle and tendon, the large proportional length (six to eight times), and the small cross-sectional area (about 140 times) of the tendon, the changes in the Achilles tendon have most likely led to the observed increase in stiffness [Thilmann *et al.*, 1991]. This does not exclude that changes in other collagen tissues [Toft *et al.*, 1989a and b], or muscle fibers themselves [Dietz *et al.*, 1986] add to the increased passive stiffness.

Changes in the muscle fibers should be reflected in the intrinsic stiffness (Figure 4.100). Sinkjær *et al.* [1993] found a significant increase in the intrinsic stiffness of the ankle dorsiflexors in spastic multiple sclerosis subjects and observed an insignificant increase in the hemiplegics [Sinkjær and Magnussen, 1994]. These findings are consistent with physiological, morphometrical, and histochemical investigations demonstrating changes of the muscle fibers, which are specific to the spastic muscle [Dietz *et al.*, 1986; Edström, 1970]. The prolonged twitch contraction times in the spastic muscles [Dietz and Berger, 1984] are consistent with increased force production as demonstrated in potentiated muscles [Burke *et al.*, 1976; Sinkjær *et al.*, 1992]. If the muscles were relatively immobilized, the subjects would probably be particularly susceptible to this type of potentiation.

Although the reason for the pathological changes in the mechanical properties of the contractile elements of the spastic muscles as well as alternations of the connective tissue (e.g. in the tendon) is still speculative, the marked increase in passive and intrinsic stiffness reported by several authors suggests that a peripheral non-reflex mediated input can importantly contribute to spastic muscle tone.

Reflex mediated mechanical muscle responses in the active muscle, the stretch reflex, make a large contribution to the total mechanical stretch response in able-bodied subjects [Allum and Mauritz, 1984; Sinkjær *et al.*, 1988; Toft *et al.*, 1991]. Rack *et al.*, [1984] have shown that during a maintained muscle contraction in spastic hemiplegic persons, the mechanically measured stretch reflex is within the normal range of the reflex stiffness in able-bodied subjects, but in the upper end of the normal range. In moderately spastic multiple sclerosis subjects, the reflex mediated stiffness in the ankle plantar flexors was unchanged compared with control subjects [Sinkjær *et al.*, 1993]. In spinal cord injured (SCI) subjects, a system identification method that separated non-reflex and reflex contributions [Kearney *et al.*, 1994] showed increased non-reflex stiffness as well as increased reflex stiffness in the passive and weakly contracted ankle extensors [Mirbagheli *et al.*, 1998]. The entire above subject groups had clinical signs of spasticity when measured by the Ashwort scale [Ashby *et al.*, 1987].

The lack of pronounced increases in the muscle tone (reflex mediated stiffness), which was found in the active spastic muscle above [Rack *et al.*, 1984; Sinkjær *et al.*, 1993; Sinkjær and Magnussen, 1994], may also be due to too small stretches. Only at higher stretch velocities and when using larger amplitudes as in the clinical bedside examination, the increased stiffness might express itself. Powers *et al.* [1988] used stretches that are comparable with the stretches used in the clinical situation, yet they did not find increased total muscle responses in spastic elbow muscles.

Increased muscle tone could be caused by an increase in the non-reflex properties (the passive and intrinsic muscle components), but also to other mechanisms. In the situations described above, the subject was asked to maintain a constant contraction, and this might introduce a situation in which the able-bodied subjects set the stretch reflex in a facilitated state. It is, however, more inhibited in the clinical situation, in which the subject is asked to relax. This leads to the hypothesis that the spastic subjects are unable to inhibit the mechanically strong stretch reflex due to an impaired (descending) control in the relaxed "clinical" situation. The increased stretch reflex in the relaxed spastic muscle and at weak precontractions can be caused by reduced postsynaptic and presynaptic inhibitions, and/or changes in postactivation depression [Nielsen *et al.*, 1995]. These are all inhibitory mechanisms, which are believed to be important to make the muscle relax in able-bodied subjects. As the muscle in able-bodied subjects is made increasingly active, these inhibitions are removed, and the reflex stiffness expresses itself fully, as it has already done in the spastic subject's relaxed "uninhibited" muscle.

EMG measured stretch reflex responses. Several possible pathways might be involved in explaining the threshold change of the stretch reflex without changing the gain such as absent reciprocal inhibition [Crone, 1993; Crone *et al.*, 1994], a changed intrinsic regulation of transmitter release from the Ia afferents in spastic subjects [Nielsen *et al.*, 1995], changes in γ-motoneuron activity [Matthews, 1959], and changes in the activity in descending motor pathways of the brainstem [Feldman and Orlowsky, 1972] that affect the postsynaptic inhibition [Matthews, 1959]. Since the stretch reflex is much less sensitive to presynaptic inhibition than the H-reflex [Morita *et al.*, 1998], it is less clear if a decreased presynaptic inhibition [Delwaide, 1973; Faist *et al.*, 1994; Stein, 1995] can explain a shift in threshold.

Muscle and cutaneous reflexes are highly modulated during locomotion in an adaptive manner within each phase of the step cycle [Crenna and Frigo, 1987; Duysens *et al.*, 1990]. This modulation is often lost or seriously reduced in subjects with spasticity. To get a good understanding of how increased muscle tone relates to more functional motor tasks, it becomes, therefore, important also to investigate the "non-reflex muscle component" together with the spinal/central integration of afferent inputs during more functional motor tasks such as walking.

The increased passive stiffness and intrinsic muscle stiffness found in sitting spastic subjects are also present during walking. Based on indirect force measurements of Achilles tendon in hemiparetic subjects, Berger *et al.* [1984b] demonstrated that the non-reflex stiffness was increased during walking. They suggested that spastic hemiparetic subjects could develop a larger muscle tension during walking at a lower level of the neural drive compared to able-bodied control subjects. This was seen as a benefit since it facilitates the subject to support body weight during the stance phase of walking in spite of the inability to activate the soleus muscle. This might be correct, but it should be remembered that a major part of the increase in non-reflex stiffness is caused by an increased passive stiffness, which will impair dorsiflexion in swing [Sinkjær *et al.*, 1996a and b] This becomes even more prominent in spastic subjects because they have a decreased voluntary drive to the ankle dorsiflexors. In addition, spastic subjects often cocontract their muscles during walking [Knuttsons and Richards, 1979], which further increases the non-reflex stiffness.

The reflex modulation that takes place during walking in able-bodied subjects [Capaday and Stein, 1986] is often lacking in spastic subjects [Fung *et al.*, 1990; Yang *et al.,* 1991; Sinkjær *et al.,* 1995, 1996a and b]. The impaired reflex modulation has been interpreted to increase the muscle stiffness because of a disrupted, supraspinal control of the stretch reflex [Fung and Barbeau, 1994; Capaday, 1995].

Phasic Response to External Perturbation - Spasms. In addition to the increased muscle stiffness in persons with injuries of CNS very strong firing of the muscles has been observed. These sudden muscle contractions are often triggered by some peripheral input (e.g., touching the skin at the leg, moving the leg passively, moving the foot passively, transferring). The tetanic contraction of muscles leading typically to simultaneous bilateral extension has been documented [Kralj and Bajd, 1989]. In most paraplegic and tetraplegic subjects both legs (hips, knees, plantar flexion) will extend generating a painful and fatiguing pattern. The movement can be so strong that it "catapults" the body from the chair. This spasm is obviously centrally mediated and peripherally triggered. Experiments with standing of paraplegic subjects showed that these bilateral spasms are enhanced in low thoracic lesion (T11-T12), yet not so common in higher thoracic and incomplete cervical lesion subjects.

In children with cerebral paralysis, it was found that during a period of minutes or longer they developed strong tonus of extensor muscles (e.g., hip extensors, knee extensors). The extension can be so strong that it sometimes prevents a child from sitting in the chair or prevents his/her walking with assistive systems. Some of the children are treated by a dorsal root rizotomy that is the dorsal roots innervating the extensors in the legs are cut. This leads to unrepairable denervation of leg extensors, which is not necessarily positive. Pharmaceutical treatments can decrease to some extent the spasm, yet they interfere with some other behavior and are not always effective. The prolonged extension suggests that a higher central input that inhibits the extension is missing. Many connections are changed or missing after CNS injury developing very individual behavior; thus it is very difficult to generalize the motor changes, but spasms occur because the input of the higher CNS centers is missing to the part of the spinal cord below the lesion.

In most subjects with paraplegia the spasm can be eliminated by a simple manipulation: strong stretching of the one of the muscles that is contracted (e.g., Gastrocnemius m. and Soleus m.) would stop the spasm. It has been noted in paraplegic subjects that when treated with functional electrical stimulation to stand, the first standing trial is accompanied with spasm. Loading of the toes, adjusting the amount of electrical charge that is delivered, and over all pushing the hips forward, that is stretching the calf muscles, will stop the firing. When a subject stretches his leg extensors by going into a so-called "C" posture during the first stand; the spasms are not likely to repeat in standings to follow during the same session. Extension spasms do not occur during walking, however, clonus, that is a repetitive flexion-extension movement at the ankle joint with the frequency of about 2 to 5 Hz, does. The clonus is the consequence of previously described changes of the reflex response to peripheral input.

Joint contractures reduce the range of movement; thus, compromise the FES activated functional movements [Perry, 1981]. Our own experimental studies have been faced with a change in the musculo-skeletal system, which results in an inability for functional movement [Popović *et al.,* 1991d]. The term functional contracture has been introduced to describe this change. It relates to the lack of performance because one joint movement restricts the movement of the neighboring joint. This is caused by increased tonus of muscles and shortening of the biarticular muscle-tendon systems. A clinical measurement of the range of joint movement does not include the testing of this behavior.

Osteoporosis [Lukert, 1982], normally found in SCI subjects, may compromise the use of legs for support and may lead to stress fractures [Rafil *et al.*, 1982] if used inappropriately. In the literature, there are some speculations that patterned electrical therapy can decrease osteoporosis [Phillips *et al.*, 1984], but other results with chronic subjects have been negative [Leeds *et al.*, 1990]. It might still be possible to prevent osteoporosis with chronic stimulation, if applied immediately after the onset of injury.

Energy efficiency of walking

Energy cost of paraplegic walking assisted by available orthoses or neuroprostheses, which restrict lower limbs to most rudimentary mechanical function and the propulsive forces are generated by the musculature of the upper body and extremities is intrinsically very high. The energy requirements of paraplegic walking using long leg braces and crutches have been studied in the past by various researchers. The pioneering study by Gordon [1956] showed that energy consumption when ambulating in a swing-through fashion is at least 3.5 times, sometime as high as 5.5 to 8 times of the basal requirements of the subjects who participated in the study. A study by Huang and associates [1979] showed that paraplegics consumed three times greater oxygen during walking than when they were at rest. A study by Chantraine and associates [1984] showed that when paraplegics are allowed to ambulate at their comfortable speed the energy consumption was lower in subjects who used their long leg braces regularly than in those who used braces sporadically. It was directly related to the level of spinal cord lesion. However, the gait velocity was directly related to brace use, *i.e.*, the velocity was higher in subjects with longer periods of regular use; and inversely related to the level of lesion, *i.e.*, walking velocity was slower in subjects with higher level lesions. Merkel and associates [1984] studied energy expenditure of paraplegics standing and walking using two types of KAFOs (Scott-Craig KAFO and Single Stopped long leg KAFO). They found that the energy cost of paraplegic locomotion with either variety of orthoses is about 5 to 12.8 times that of normal locomotion [Blessey, 1978]. Miller and associates [1984] measured energy requirements of paraplegic locomotion during negotiating turns, stairs and ramps using the same two varieties of KAFOs. They found that the energy consumption of paraplegic locomotion during negotiations architectural barriers was approximately the same as the one measured in able-bodied walkers (control group), yet the energy cost was as much as 15 times more than that of controls.

Table 4.3: Measures of the walking efficiency as function of the LEMS (lower extremity muscle score). Adapted from Waters *et al.*, 1994, with permission.

	LEMS 0 - 40 %	LEMS 40 - 60 %	LEMS 60 - 90 %
O_2 RATE [ml/ kg min]	15.2±3.4	13.2±1.6	14.6±3.0
O_2 RATE INCREASE	158±1114	110±60	49±35
O_2 COST [ml/ kg m]	0.76±0.61	0.51±0.25	15.2±3.4
VELOCITY [m / min]	30.5±15.6	31.4±11.4	57.5±12.3

Waters and coworkers [1988a and b, 1992, 1994] provided a comprehensive analysis of the efficiency of walking (e.g., energy cost, energy rate, heart rate, blood pressure) as function of the measurable preserved functioning. They suggest that the classification on the lower extremity muscle score (described in Chapter 3) is highly correlated with velocity, cadence, oxygen (metabolic energy) cost and rate, axial peak loading of the legs and the ambulatory motor score. The study that included 36 paraplegic subjects [Waters *et al.,* 1994] shows that there is a high correlation (Table 4.3) between the lower extremity muscle score (LEMS) and the energy efficiency.

All of the above mentioned studies concluded that paraplegics with high or mid thoracic level lesion are probably incompatible for walking. In all studies, subjects walked using a form of KAFO, and there was no trunk support provided during walking, therefore, energy was also utilized to maintain the upright posture. Nene and Patrick [1989] studied the energy cost of locomotion for thoracic level paraplegics using the ParaWalker, which gives adequate support to the trunk and provides better stability than KAFOs. They found that the energy consumption with the ParaWalker was about 3.5 times the resting level of the subjects. The energy cost was 4.9 times more than that of normal walking; this was in contrast to the findings for paraplegics with thoracic level lesion reported in the previous studies. They deduced that the efficiency of the ParaWalker lies in the intrinsic stability provided by the orthosis. Surface electrical stimulation of stance side Gluteal muscles helped to reduce the energy cost of locomotion by approximately 6.5 percent to 8 percent in the rather small study group of 5 subjects. In a study comparing energy requirements of ambulation using the RGO alone and hybrid RGO, Hirokawa *et al.,* [1990] found approximately 16 percent reduction in the energy cost throughout the full range of walking speeds when RGO was combined with electrical stimulation of the thigh muscles. They concluded that at slow speeds the energy cost of ambulation using RGO was less than that using KAFOs. Winchester and associates [1993] examined the energy cost of ambulation using a standard RGO and using a modified RGO, *i.e.,* Isocentric RGO. They found that with the standard RGO subjects' oxygen uptake was 14.2 ml/(kg min) (mean) and the mean velocity was 12.7 m/min; with the Isocentric RGO the oxygen uptake was 13.0 ml/(kg min) (mean), and the mean velocity was 13.5 m/min (the differences were statistically not significant).

Marsolais and Edwards [1988] reported energy requirements of walking using FES alone and compared those to using KAFOs in only 3 subjects. During FES walking energy consumption was about 59 to 75 percent of maximal aerobic power of the subjects. There was no increase in energy consumption when the walking speed was increased. The energy cost equaled that of KAFOs. At speeds approaching 25 m/min FES walking energy consumption was similar to that of KAFOs. They inferred that at speeds between 25 m/min and 36 m/min.

The ability to measure the efficiency by hart rate or some related measurable could be important, since measuring oxygen consumption is difficult and cumbersome. In the past it has been shown that it was possible to establish walking performance of subjects by monitoring speed and heart rate [Stallard *et al.,* 1978; Stallard and Rose, 1980]. MacGregor [1979,1981] described a method of combining these two parameters to produce a single index called the Physiological Cost Index (PCI). The PCI is a measure of the cardiovascular stress, but it is not directly related to the metabolic energy consumption. Nene and Jennings [1992] measured the PCI of paraplegic locomotion using the ORLAU ParaWalker. In a study group of 16 subjects mean PCI was 3.11 heart beats per meter (b/m). It ranged from 1.47 b/m to 4.76 b/m. Bowker and associates [1992] reported a mean PCI value of 5.04 b/m for a group of

28 RGO users. Their subject group consisted of subjects with paraplegia due to varied pathology.

A paraplegic should be able to use the orthosis completely independently. This includes doffing and donning; transfers, *i.e.*, standing up, walking and sitting down again. The subject must also be able to negotiate commonly encountered architectural barriers such as gentle slopes, curbs, steps, *etc.* Transfers in and out of wheelchairs, cars and other means of transport are also essential for everyday life of a paraplegic. All presently available KAFOs can be donned, and doffed by the subjects independently.

Rosman and Spira [1974] surveyed the use of orthoses (KAFO) in 51 paraplegic subjects after their discharge from the hospital for 16 years in total. They used four groups based on the level of lesion (T1-T6, T7-T11, T12-L1, and L2-L5). The over all conclusion is that most subjects with the lesion above T11 abandon the use of an orthosis even for standing, some subjects occasionally use them for standing and only exceptionally for walking, and that about one third of the L2-L5 subjects use the assistance for walking.

Subjects with severe spasticity and contractures have some difficulty in doffing and donning the ParaWalker and the RGO [Moore and Stallard, 1991; Solomonow *et al.*, 1989]. The RGO had knee joints with automatic knee locks, which caused difficulty for some subjects. These have been modified and the newer generation of RGO has ratchet knee joints [Solomonow *et al.*, 1989]. The HGO users need to extend and lock their knees before they are able to stand up. A special assistive device can be made for individuals who have difficulty with locking their knees.

The other major factor, which can hamper independence, is the ability to perform toilet functions while using the orthosis. Use of either the ParaWalker or RGO certainly causes major problems in this area. Use of KAFOs, due to hips being unconstrained, does not cause serious inconveniences.

Any walking system used for paraplegic walking must have a very high level of reliability, but at the same time in case of any unforeseeable failure the user must not come to any physical harm. All mechanical systems currently in use are fairly robust and safe. Stand alone FES systems fail to provide any mechanical safety in the case of failure of electronic components during walking. With percutancously implanted electrodes the rate of electrode failure is high thereby necessitating re-implantation of the electrodes. Additionally, the electrode insertion sites acting as portals for entry for infecting organisms remain a risk.

Technical and technological issues relate to the NP-neuromuscular interface and the biocompatibility of a NP. The NP-neuromuscular interface is realized with electrodes. Currently, several types of electrodes are in use or under development. The least invasive is transcutaneous electrical stimulation with surface electrodes, while other techniques use percutaneous or implanted electrodes. Percutaneous or implanted electrodes are applied to muscles, close to the motor point, to the nerve directly. The use of other types of electrodes is in its developmental phase. The use of implanted electrodes and stimulators; that is totally implanted neuroprostheses is progressively being used in many more subjects. The efficiency of the energy transfer between an implant and the external unit has to be improved, specially for NP that use rather sophisticated microcontroller, implanted sensors with amplifiers, and multichannel devices. An interface device that can be adapted for various users allowing them to control the NP is most probably the critical issue. The technology of using the brain interface is very appealing, yet it is in its infancy. Plausible use of implanted brain interface is another plausible command link, however, it is rather invasive, and at this stage does not fulfil the cost to benefit ratio.

The practicality of the system relates to the cosmesis, ease of donning and doffing, safety and reliability. These elements determine how easy a system could be accepted by potential candidates for using it, and over all the difference that the device is bringing in the quality of life of users.

The control issues being essential for improving the functioning of the NP are discussed in Chapter 5.

4.4 Artificial Legs

An artificial extremity is a morphological and functional replacement of a human limb [Muilenburg and Wilsson Jr, 1996]. Artificial extremity compensates for the disability caused by amputation of a limb or its part. The differences between the upper and lower extremity prostheses come from the dissimilarity in functions, which they have, to replace and/or compensate. An artificial leg should support the body weight (standing) and mainly perform so-called cyclic movement (e.g., walking, running). The arm prosthesis needs to endow with extremely complicated goal-directed movement (manipulation) in order to allow hand grasp. The states of the art of the leg, arm, and hand prostheses are presented to indicate what novelties are needed for improved function.

4.4.1 The Lower-Limb Prosthesis

An amputation results in a disabling condition. With modern prostheses and treatment methods, when the musculature is good, the circulation is adequate, and there is an absence of excessive scarring, however, the unilateral amputees can do many of the things, which they could before amputation. The objective of this section is to describe lower-limb prostheses available to users, including persons who have had a knee disarticulation, hip-disarticulation, or hemipelvectomy.

During the past few years the International Standards Organization (ISO) for prosthetics and orthotics has developed a standard method of describing amputations and prostheses that is being adapted worldwide. They adopted the term “transfemoral” in place of "above-knee" to identify an amputation between the knee and hip joint. This term has been selected to avoid confusion with disarticulation at the hip and amputations through the pelvis. The term

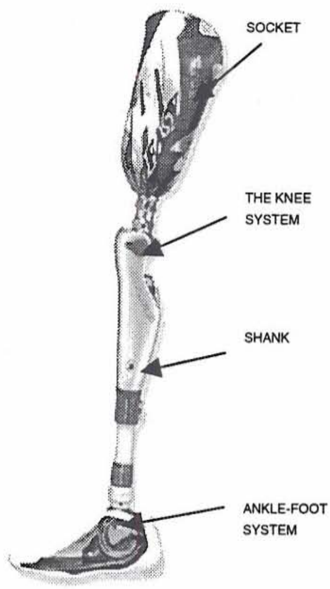

Fig. 4.101: Four main components of a transfemoral prosthesis (C-Leg® System): the socket, the knee, the shank, and the ankle-foot complex (www.ottobock.com)

"transtibial" instead of "below-knee" describing an amputation below the knee joint has been adopted accordingly. The ISO is also planning to adopt the term "stump" referring to that part of the limb that is left after amputation.

The lower-limb prosthesis has to duplicate the behavior of the missing portion of the leg. The biomechanical requirements for a lower-limb prosthesis have been summarized by Wagner and Catranis [1954] and refined by Radcliffe [1980].

The prosthesis must support the body weight of the amputee in a manner similar to the normal limb during the stance phase of level walking, on slopes, soft or rough terrain. This implies that the prosthesis provides for "stability" during weight bearing. The stability in this context refers to prevention from sudden or uncontrolled flexion of the knee during weight bearing. It is obvious that an "unstable" knee in the prosthesis can lead to dangerous situations and possible injury for the amputee.

The support of the body weight has to ensure that undesirable pressures of the amputation stump are excluded or that gait abnormalities due to painful contact between the stump and socket are prevented. The analysis of biomechanical factors, which influence the shaping, fitting, and alignment of the socket is a problem of itself. If the fitting has been accomplished in a manner, which allows the amputee to manipulate and control prosthesis in an active and comfortable manner, the socket and stump can be treated as one single body.

The third requirement placed upon the prosthesis is that it duplicates as closely as possible the kinematics and dynamics of normal gait. The amputee should be able to walk with an essentially normal appearance over a useful range of walking speeds associated with typical activities for normal persons of similar age. The latter requirement has received a great deal of attention in recent years, and fully integrated systems, so-called self-contained, active, transfemoral prosthesis, are being incorporated into modern rehabilitation. The self-contained principle implies that the artificial leg contains the energy source, actuator, controller, and sensors.

The prostheses cannot be unambiguously classified into passive and active ones. Such solutions as the polycentric knee mechanism [Cappozzo *et al.,* 1980], the polycentric knee mechanism with hydraulic valve [Radcliffe, 1980], or the transfemoral prosthesis with friction type brake [Aoyama, 1980] satisfy some of the above performance requirements. Logically controlled transfemoral with the hydraulic valve represents a further bridge between purely passive and fully controllable assistive devices [Turajlić and Drakulić, 1981; James, 1983]. All o f the abo ve mentioned prostheses satisfy the stance phase requirements as well as the minimum power consumption principle. However, the amputee is not able to flex/extend the knee in the stance phase once it has been flexed. Extension in the swing phase requires additional metabolic energy. Gait asymmetry has to be minimized.

4.4.2 Transfemoral Prosthesis

The transfemoral prosthesis has four major parts: the socket, the knee system, the shank, and the foot-ankle system. A variety of sockets, knees, shanks, feet, and ankles are available and can be combined to produce prosthesis that meets the needs of each individual amputee.

Figure 4.101 shows an example of the transfemoral prosthesis with an advanced ankle-foot complex, hydraulic actuator at the knee joint capable of controlling the knee position throughout the walking cycle, and an endoskeletal shank (3C100 C-Leg® System, Otto Bock Orthopedic Industry, Germany).

The Socket

The socket is the basis for the connection between the user and the prosthesis. It always provides the means of transferring the weight of the amputee to the ground by way of the rest of the prosthesis. The shape of the socket is critical to comfort and function. The socket must not restrict circulation, yet it cannot be loose. Most sockets for above-knee prostheses cover the entire stump. There are several designs available to take maximum advantage of the muscles in the stump of the individual amputee for control of the prosthesis and for transferring the weight of the amputee to the floor.

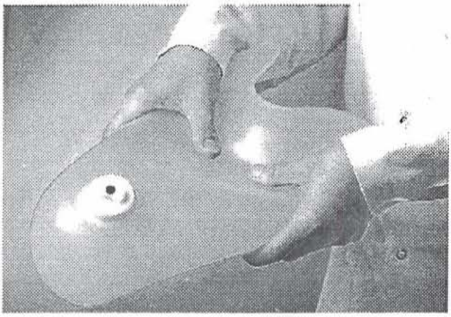

Fig.4.102: Flexible suction type socket for a transfemoral prosthesis

Most sockets are made of a rigid plastic, but some amputees prefer a flexible socket supported by a rigid frame because comfort during walking and sitting seems to be improved.

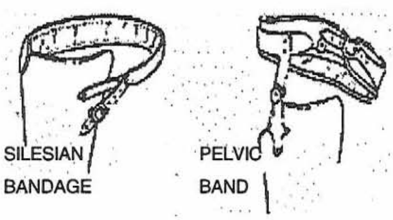

Fig. 4.103: Systems for fixing of a "loose" socket to the stump

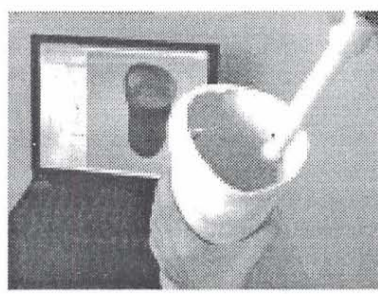

Fig. 4.104: CAD of the socket with the TracerCAD system (http://www.tracercad.com)

For most subjects, the prosthesis can be held in place by "suction" or a vacuum, provided by a close fit between stump and socket (Figure 4.102).

This is known as a suction socket. Nothing is worn between the stump and socket. When circulation is marginal or precarious, a looser fit should be provided. A loose fit requires use of soft, yet stable interface to fill in the "loose" contact. This is done with "socks" of wool worn over the stump. In such a case, it is necessary to add a system that will hold the socket in place (e.g., a Silesian Bandage, Figure 4.103).

To design a socket needs a lot of attention, and in many cases remains only the skill of a prosthetist. The stump of the leg changes its form (e.g., gaining or loosing weight, muscle atrophy) throughout the lifelong application of the prosthesis. Therefore, it is necessary to facilitate fast and adequate manufacturing of a socket.

Computer Aided Design (CAD), or as it was originally called Computer Aided Socket Design (CASD), is a technique, which uses computerized three-dimensional image of the stump and numerical machines to automatically fabricate the socket.

The first phase for automatic fabrication of the socket relates to creating a three-dimensional image of the stump. Since the stump is unique for each user, and because of the individual biomechanical features (e.g., aligning of the thigh vs. the pelvis, Figure 4.104), it is essential to include expert knowledge gained through experience in automatic design of the sockets.

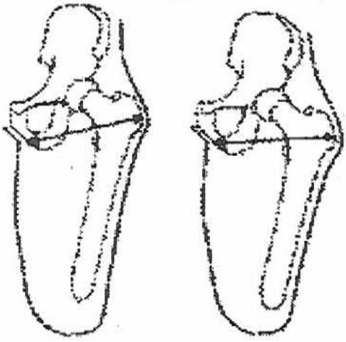

Fig. 4.105: Two similar, yet biomechanically different stumps.

TracerCAD® system (Figure 4.104) has introduced an example of direct and interactive expert intervention system, and it allows taking precise measurements with instantaneous and lifelike images of the process by using a "pen". The "pen" allows a prosthetist to mimic a touch with the stump that imitates the natural sweep of the fingers. The advantage of the system is that it enables clinical modifications directly on the subject. The method must include skeletal fit technology, not only the three-dimensional surface image. The example (Figure 4.105) shows two stumps having the same distance between the Anterior-Medial aspect of the Ramus and the Greater Trochanter, illustrated by the arrow, but different orientations; the long axis of the stump ensures that the force is transmitted in a direction which will not cause hip pain and possible injury.

The Knee System

The above-knee amputee needs a stable support while walking; thus, the prosthesis must have a knee joint that will allow only controlled buckling as he rolls over the artificial foot during the stance phase of walking. The same knee joint has to provide flexion during the swing phase to allow ground clearance and natural-like movement of the leg. Most prostheses, by engineering terms of automatic control, are under-powered systems. This means that at least one degree of freedom in the system does not have the actuation capabilities to control the movement at all times.

The simplest way to achieve walking with an under-powered prosthesis is to use mechanical friction about a bolt that connects the socket (the thigh) and the shank. The bolt must be located behind the line determined by the ground reaction force at all times during the stance phase of walking. If the axis of the knee rotation comes in front of the line of the ground reaction, the knee joint becomes unstable and a locking mechanism is instrumental.

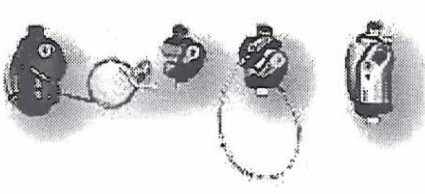

Fig. 4.106: Single-axis knee joint mechanism with a friction mechanism and a mechanical lock in fully extent position (http://www.Ottobock.com)

The mechanical friction system, which may be a simple adjustable braking mechanism, is needed to keep the shank from swinging forward too fast as the user swings the artificial leg. The principal limitation of a single-axis, constant friction design of the knee system is that appearance of walking is "normal" at only one speed of walking for a given setting of friction; hence, the amputee must be careful while walking, especially if walking on uneven surfaces.

A great deal of effort has been spent over the years developing knee systems which overcome the limitations of the single-axis, constant friction knee. Many designs have been somewhat successful, yet the constant friction knee system is still in use (Figure 4.106). The second level of complexity in knee systems is the use of a weight-actuated brake with constant friction. Two bolts are used at the knee so that when one pivots about the other when the amputee is standing, the force of the body weight engages a brake that keeps the knee from buckling.

To allow the amputee to vary his speed of walking, a number of hydraulic devices are available. In the simplest system, the piston is attached to a pivot in the thigh section of the prosthesis behind the knee bolt, and the cylinder is attached to a pivot in the shank (Figure 4.107). Because of the way oil acts when forced through a small hole, the amount of resistance required for a given velocity of walking is provided automatically.

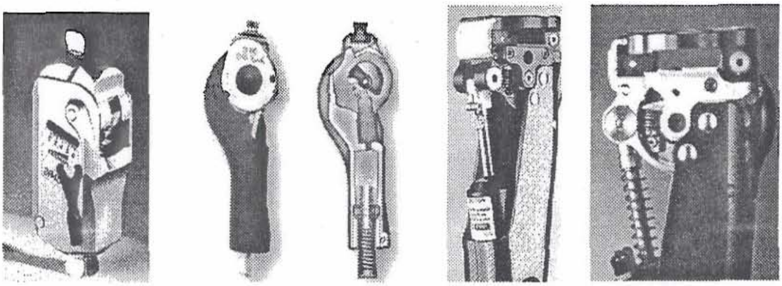

Fig. 4.107: Single-axis mechanisms for the knee joint assembly. 3R45 Modular Knee Joint, 3R80 Modular Rotary Hydraulic Knee Joint, 140 degree Flexion ESK Superior Knee Stabilization and Pneumatic Swing Phase Control (from the left to the right respectively) (http://www.Ottobock.com and http://www.endolite.com)

3R45 Modular Knee Joint is an ultralight, low profile, hydraulic single axis knee. It allows a wide range of flexion and extension resistance giving a higher subject activity level. Swing phase resistances are independently adjustable. The 3R80 Modular Rotary Hydraulic Knee Joint is the rotary hydraulic device that is responsive to the cadence of walking. It allows up to 135 degrees of flexion and adjustable resistance to flexion and extension during the swing. The new Endolite Stabilized Knee (140 degree Flexion ESK) incorporates many enhanced features: independent control of swing and stance, 140 degrees of knee flexion, high load capacity, rapid and reliable fitting. This knee can be equipped with various hydraulic and pneumatic actuators offered by Endolite. The novelty in the design of actuators is servo PSPC (Pneumatic Swing Phase Control), which allows variable knee joint angular velocity; hence, adaptation to the speed of walking within limits.

To provide better control of the transfemoral prosthesis during standing and the stance phase of walking, mechanical linkages between the socket and shank that provide a moving center of rotation have been introduced (Figure 4.108). Such designs are known as polycentric knees. Used originally for the knee-disarticulation case,

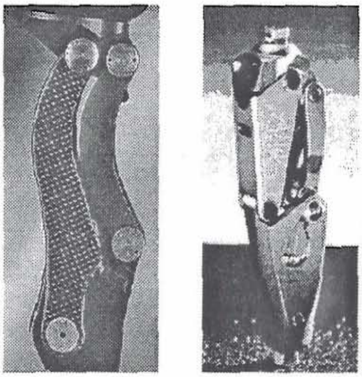

Fig 4.108: Polycentric knee mechanisms for the knee. Four-bar linkages provide movement of the center of rotation with respect to the thigh and shank following closely the movement of the center of rotation in the biological counterpart. The 3R60 EBS Knee, Otto Bock Industries, Inc. (left) and the Slim Profile 4-Bar Knee Disarticulation, Blatchford, Endolite (right) (http://www.Ottobock.com and http://www.endolite.com)

polycentric knees are now also used in prostheses for higher levels, especially when stability at heel strike is desirable. The swing phase control may be either mechanical friction or hydraulic resistance. The one limitation of the polycentric design is that the range of motion about the knee may be restricted to some degree, but not enough for it to be objectionable to most users. The 3R60 EBS Knee has unique new polycentric features imitating the human knee by providing up to 15 degrees of cushioned stance flexion.

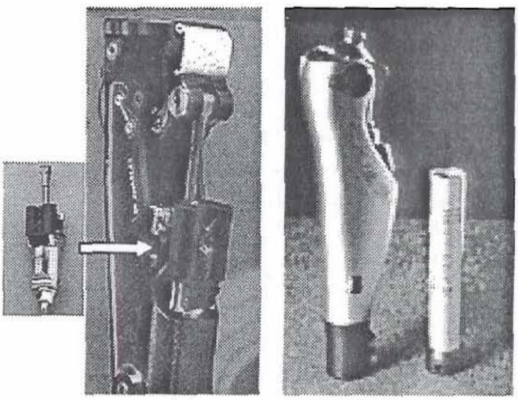

Fig. 4.109: The IP+ (http://www.endolite.com) (left) and 3C knee joint (http://www.Ottobock.com) (right) systems allow microcomputer intelligent control of leg behavior during walking.

Two hydraulic cylinders, one to influence the stance flexion, the other to control the swing phase, offer a more natural walking. The controlled knee buckling during the stance phase is especially effective for walking on uneven terrain. This polycentric knee allows flexion up to 150 degrees. The Slim Profile 4-Bar Knee Disarticulation can be equipped with different hydraulic or a pneumatic actuator providing great flexibility in providing a subject with the preferred solution for him/her. The linkages limit the knee flexion to 125 degrees. Complex carbon fiber composite used for the links ensures smooth profiles, lightweight, and durability.

The Intelligent Prosthesis Plus (IP+) automatically adjusts the swing of the knee to match the individual amputee's walking speeds. This control ensures better walking symmetry, and as shown in clinical tests involves reduced cognitive and cardio-pulmonary stress. T he most complex knee systems of those available are those which control both the swing and stance phase with a single hydraulic cylinder (Figure 4.109).

The 3C100 C-Leg System represents the first and only commercially available microprocessor controlled hydraulic knee with swing and stance phase regulation (Figure 4.109). The operation of the system includes sensors detecting the loading of the leg and rotation in the knee joint. The program stored in the self-contained microcomputer uses sensory information and a sequence of logic operations, similar to reflexes in the natural control to provide control of both swing and stance. The 3C-knee joint can be combined with a variety of shank and foot assemblies. The same system permits the velocity of walking to be varied at will, being appreciated by many active amputees.

The Ankle-Foot Complex

The ankle-foot complex has to provide shock absorption at the heel strike, provide stability and rolling over the sole during the stance of walking, allow adaptation for uneven terrain and small obstacles, and possibly store the energy during the loading phase, which it will return to the leg during the push-off phase of the stance. The ankle joint has to allow some medial and lateral rotation, as well as eversion and inversion.

Artificial feet currently available can be divided into two classes: articulated, and non-articulated. Most of the non-articulated feet are available with toes molded in to provide a very realistic appearance. Articulated feet have moving joints generally requiring maintenance and are heavier than most of the non-articulated feet. Articulated feet may have one or more joints. The single-axis foot (one-joint) provides for ankle action that is controlled by two rubber bumpers, either of which can be changed to permit more or less motion as needed. It is often used to assist in keeping the knee stable. A multi-axis foot is often recommended for people who have to walk on uneven surfaces because it allows some motion about all three axes of the ankle. The simplest type of non-articulated foot is the SACH (solid ankle-cushion heel) foot. The keel is rigid.

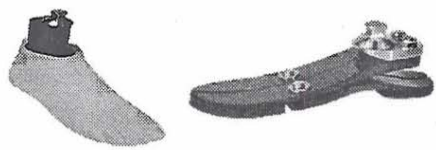

Fig. 4.110: The K2 Sensation (left) and Allurion (right), low profile multi-axial feet. (http:// www.flexfoot.com)

Ankle action is provided by the soft rubber heel, which compresses under load during the early part of the stance phase of walking. The rubber heel wedges are available in three densities: soft, medium, and hard.

The SAFE (solid ankle-flexible-endoskeletal) foot has the same action as the SACH plus the ability for the sole to conform to slightly irregular surfaces and thus makes it easier for the amputee to walk over uneven terrain. Feet of this type make walking easier because of the flexibility, and they are sometimes called "flexible keel" feet.

In recent years, there has been a proliferation of new designs for artificial feet (Figure 4.110). Most are capable of absorbing energy in a "flexible" keel during the "rollover" part of the stance phase of walking and springing back immediately to provide push-off, or assistance in getting the toe off of the ground to start the swing phase of walking. These designs are often called "dynamic response" feet.

Most of the systems used today are multi-axis feet with a limited range of movement. The ankle joint is not powered. The researchers have investigated the ankle joint articulation, but both the control and power requirements came as arguments against the usage of these assistive systems. A realization of an articulated ankle joint was with the coupled ankle-knee joint movement within transfemoral prosthesis (The WLR-7, Waseda leg, Japan).

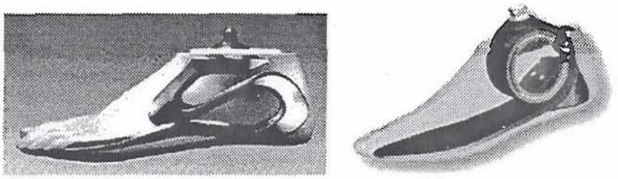

Fig. 4.111: The 1D25 Dynamic Plus Foot (left) and C-foot (right) fully articulated systems (http://www.Ottobok.com) providing shock absorption and storing of energy.

The K2 Sensation foot (Figure 4.110, left) provides multi-axial range of motion cushioning each step while helping to maintain control over barriers as raps or small obstacles like curbs. The full-length toes level, a flexible keel, and an integrated multi-axial design offer a smooth transition from heel strike to toe off that is absent in some other feet. This means added comfort and relief for those that may be struggling with dysvascular complications. The Allurion, low profile, dynamic response from Flexfoot (Figure 4.110, right) is designed for the amputees with long stump of knee units. The compact design of the Allurion provides all the function that an active amputee would expect. A carbon fiber is used for strength and durability, and it can be integrated with transtibial or transfemoral prosthesis through a titanium male pyramid, being standard today. The Carbon Active Heel provides shock absorption for the overall health and comfort. This heel harnesses heel strike energy and transfers that energy into the foot module enabling forward progression, which ensures decreased metabolic energy cost and rate for walking.

Fig. 4.112: The Endolite Multiflex Foot and Ankle (http://www.endolite.com)

The 1D25 Dynamic Plus Foot (Figure 4.111, left) is designed for amputees who want a highly functional foot. It offers increased greater mobility to the amputee. The improved S-shaped spring offers enhanced physiological rollover during stance phase of walking. The stored energy returns to the forefoot at toe-off, therefore, further improving the initiation of swing phase. The 1C40 C-Walk Foot (Figure 4.111, right) provides true comfort. A carbon fiber foot combines comfort with dynamic response and multi-axial rotation. The C-element cushions the foot for a comfortable heel strike. The stored energy is released to facilitate a smooth rollover to mid-stance. As the fore foot loading increases, the C-spring element loads again along with the base spring. This energy is returned at toe-off to assist in initiating swing phase.

The Endolite "Muliflex" concept provides a unique foot and ankle combination capable of excellent performance (Figure 4.112). Torsional movement (medial and lateral rotations), inversion and eversion of the ankle give comfort and control customized to suit the individual's requirements. The walking performance is enhanced by careful design of the toe break. The break is reinforced with a polymer strip. A keel manufactured from a new lightweight, long fiber composite material, has been incorporated. The foot design provides increased heel cushioning. The Multiflex ankle-foot complex provides "natural"-like rollover during the initial and mid-stance phases, a facilitated push-off due to the energy stored in the foot. All features of the Multiflex concept lead to decreased cardio-vascular stress, and metabolic energy cost and rate. The system can be directly integrated with a standardized endoskeletal shank.

The Shank

The primary purpose of the shank is to transfer the vertical loads caused by the weight of the amputee to the foot and onto the floor (Figure 4.113). Two types of shank systems are available: 1) the Crustacean or exoskeletal, which transfers the forces through the outside walls of the hollow shank which is shaped like a leg; 2) and the or pylon, or endoskeletal, where the forces are carried through a central structure, usually a tube, and the esthetics of provided by a foam covering.

Fig. 4.113: The central, endoskeletal (left) and the Crustacean, or exoskeletal (right) shanks connected to the knee joint system.

Fig. 4.114: Modular design of ankle-foot system (K2 Sensation) including the titanium pyramide for connecting it with the pylon (http://www.flexfoot.com)

Each design has advantages and disadvantages. The endoskeletal systems offer the most life-like appearance and "feel", but require more care to maintain. The crustacean design is suitable for heavy duty. Most endoskeletal parts are designed for moderate or light duty, but heavy-duty systems are available.

Another advantage of some of the endoskeletal systems is that knee units of great complexity can be introduced as the amputee becomes more proficient or his functional needs change.

Fig. 4.115: Three different types of Flexfoot pylons (http://www.flexfoot.com)

The skeletal part of the prosthesis replacing the shank is designed as a rather rigid skeleton, or a flexible structure. The flexible solution of the skeletal part allows the user not only to walk and stand safely, but also to participate in sports and other intensive bipedal activities (e.g., dancing, hoping). The rigid skeleton, more frequently used, provides weight bearing and stability.

The Flexfoot support for the pylon promotes whole body comfort by providing lightweight, high strength, non-corrosive, exceptional energy storage and return system for support of the body (Figure 4.114). This construction minimizes the additional weight to the limb and offers maximum durability (Figure 4.115). Faster walk with less efforts is provided allowing lively, energy efficient functions.

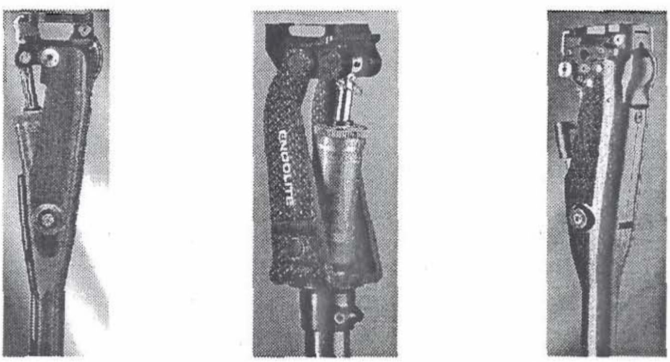

Fig. 4.116: The Uniaxail Knee Chassis (left), the 160 Hi-Activity Chasis, and the 160 Universal Shin Range pylon-knee mechanism housing (http://www.endolite.com)

The Stanceflex Uniaxial Knee Chasis (SFEUK) integrates the housing for the hydraulic or pneumatic actuator with the pylon (Figure 4.116). The construction allows 115 degrees of flexion and is primarily used with a CaTech hydraulic device that controls stance flexion. The whole structure absorbs the heel impact; it is energy efficient because it can give the energy back that was stored during loading of the leg.

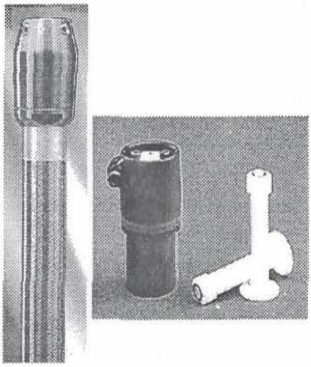

Fig. 4.117: The TT pylon and the demountable torque absorber (http://www. endolite.com)

The 160 HI-activity (Figure 4.116) chassis is a tough, long life, low maintenance pylon-knee actuator housing. It provides 120 degrees of knee flexion and is frequently applied with a CaTech cylinder that gives good swinging control. This uniaxial device includes super strong needle roller bearings for better support of high stresses. The 160 Universal Shin Range (Figure 4.116) is an integrated shank and knee-actuator housing for the transfemoral prosthesis. The system allows 130 degrees of knee flexion and is typically used with IP+ actuator. The endoskeletal structure is very convenient for use of both continuous and discontinuous cosmesis.

The TT pylon incorporates a vertical shock pylon and torque absorber in one compact package (Figure 4.117). A maximum movement of 1.5 cm ensures smooth vertical deceleration at heel strike and an improved push-off with any foot system. This is possible because of the high performance spring that is made in three grades to meet individual needs of amputees. Torsional resilience is controlled by a thermo-plastic torsion rod, available in three grades (firm, medium, and easy). The maximum rotation is 30 degrees in both medial and lateral directions. The amount of rotation is determined by external torque applied to the knee, while the foot is loaded during stance. This feature is extremely valuable for sports requiring twisting of the body (e.g., tennis, golf).

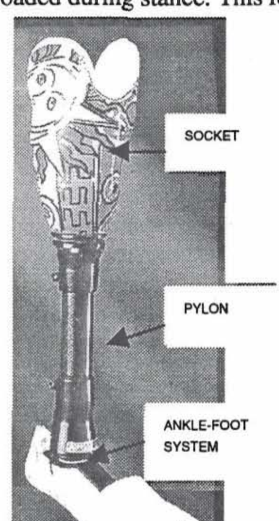

Fig. 4.118: The Endolite Below Knee System composed of three components: the suction type socket, the pylon and the foot-ankle system (http://www/ endolite.com)

The demountable torque absorber allows up to 45 degrees of rotation in either direction (medial and lateral) from the neutral position. The rotation is resisted by an advanced plastic torsion element, which can be charged to give a stiffer or softer response to loading. The device contributes to the comfort of the prosthesis by limiting shear forces transmitted to the stump.

4.4.3 Transtibial Prosthesis

The basic difference between transfemoral and transtibial prostheses is that the later does not have any joints (Figure 4.118).

The transtibial prosthesis consists of three major components: the socket, the pylon, and the ankle-foot system (Figure 4.118). No transtibial prosthesis includes a mechanism to replicate the ankle joint, but does include elastic visco-elastic elements to absorb energy at heel impact, to store energy during the loading phase, "give back" energy during the push off, and to allow limited dorsi- and plantar flexion. The transtibial prosthesis allows limited inversion-eversion and medial-lateral rotation of the shank during the stance phase of walking.

The ankle-foot and socket considerations for the transfemoral prosthesis hold for the transtibial prosthesis.

The prosthesis for the subject with Syme's amputation is similar to the below-knee prosthesis except that the socket also serves as the shank (Figure 4.113, right). Because of the short space between the end of the stump and the floor, a special type of foot, usually a modification of one of the popular designs, has to be used. The prosthesis has the shape of the stump; hence, no other provision for suspension is necessary.

Two types of sockets are in general use: the plastic socket with an expandable liner and the plastic socket with a medial opening. Both types were designed for easy entry, yet they take advantage of the shape of the stump to provide suspension.

4.4.4 Hip-Disarticulation and Hemipelvectomy Prosthesis

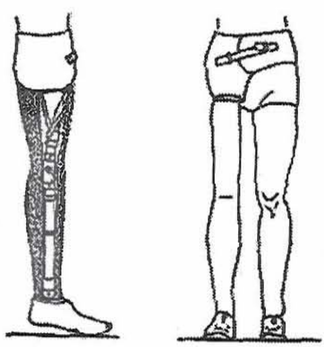

Fig. 4.119: Sketch of the hemipelectomy prosthesis.

Most of the components designed for above-knee prostheses are suitable for amputees who have lost function about the hip due to amputation just below the hip joint, at the hip joint (hip-disarticulation), or hemipelvectomy (when half of the pelvis has been removed). To provide good control of the leg, the artificial hip joint is placed on the front of the socket rather than opposite the anatomical hip joint; an arrangement that provides better control of the prosthesis (Figure 4.119).

For these prostheses, the socket is either made of laminated plastic or a thermoplastic, and the construction is usually modular that is, pylon or endoskeletal, because this type of construction results in a relatively lightweight prosthesis.

The hemipelvectomy prosthesis presents an added problem to the prosthetist because there is no ichial bone present to aid in weight bearing.

All of the instructions presented about use of the above-knee prosthesis apply equally to the hip-disarticulation and hemipelvectomy prostheses.

4.4.5 Preparatory Prosthesis

Fitting prosthesis to the stump as soon after surgery as possible helps to combat edema. A preparatory prosthesis is frequently used for several weeks or months until the stump has stabilized before the "permanent" or definitive prosthesis is provided.

The socket of the preparatory prosthesis may be made of either plaster-of-Paris or a plastic material and is attached to an artificial foot by a lightweight tube or strut, a pylon. When indicated, a suction socket is used. Most pylons are designed so that the alignment of the foot with respect to the socket can be changed when it is needed.

A belt about the waist is usually used to help keep the preparatory prosthesis on the stump properly. At least one prosthetic sock is worn between the socket and stump to provide for ventilation and general comfort. Prosthetic socks are used to prevent skin abrasion and to provide ventilation.

Regardless of the functions provided by the most sophisticated, mechanical devices, the most important factors in the usefulness of an artificial leg are fitting of the socket and alignment of the various parts with respect to the body and to each other. Fitting and alignments are difficult procedures that require a great deal of skill on the part of the prosthetist and a great deal of cooperation on the part of the amputee. During fitting and alignment of the first prosthesis, it is necessary to train the amputee in the basic principles of walking. The fitting affects alignment and *vice versa*, both affect comfort and function.

4.5 Artificial Hands and Arms

Pointing, reaching, and grasping allow humans to function independently and do many things that make them productive and enjoy many tasks of everyday life. Humans position the hand for all of the function by using the body and arm. Here we only discuss the arm. The shoulder joint provides for three degrees of freedom (flexion and extension, adduction and abduction, and medial and lateral rotation of the humerus), the elbow and forearm carry two degrees of freedom (flexion and extension, and pronation and supination of the forearm), and the wrist allows two degrees of freedom (dorsal and volar flexion, and radial and ulnar deviation). These seven possible rotations allow a human to bring the hand in the desired position and orient it in the appropriate direction. Total arm prosthesis would be a system with minimum six degrees of freedom and the grasping device. It will interface the trunk at the level "above" the shoulder. There is no total arm prosthesis available at this time. *Transhumeral prosthesis* is a replacement of the portion of the upper arm, the elbow, lower-arm, wrist, and the hand. It needs to functionally replace the positioning system of the elbow and wrist, in addition to allowing the grasping. Movement of the shoulder joint (e.g., moving the trunk and scapula) also helps the positioning of the hand. *The hand prosthesis* is a system that replaces a portion of the forearm, the wrist, and the hand. We will not discuss partial hand and finger prostheses.

4.5.1 Transhumeral Prosthesis

Transhumeral prosthesis interfaces the stump via a socket. Designing a socket follows the principles described in Section 4.4.1. The transhumeral prosthesis has to provide positioning of the hand and grasping, transport of the item grasped, and its eventual utilization. Grasping and positioning of the hand should not be separated, because there is a strong interaction between the type of the grasp and the approach trajectory that is the path of hand moving between the initial position and the target. In addition, once the object is grasped, the transport of the object is highly dependent on the object (e.g., cup of coffee, spoon, and fork).

The Elbow Joint

The artificial elbow joint should provide two degrees of freedom. One degree of freedom corresponds to the numeral rotation, while the other is for the forearm flexion and extension (Figure 4.120). The three joints to the right in Figure 4.120 are for endoskeletal designs, while the left one is for exoskeletal applications.