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CHAPTER 48

REHABILITATION ROBOTICS, ORTHOTICS, AND PROSTHETICS

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48.1. OVERVIEW

One overarching goal drives our research and development activities: to revolutionize rehabilitation medicine with robotics, mechatronics, and information technologies that can assist movement, enhance treatment and quantify outcomes. In this chapter, we present three fronts of this revolution: rehabilitation robotics, orthotics, and prosthetics.

The first and newest approach, rehabilitation robotics, has grown significantly in the last ten years (c.f. special issue of the Journal of Rehabilitation Research and Development, 37:6 of Nov/Dec 2000; International Conference on Rehabilitation Robotics – ICORR 2001 and 2003). Previously, robotics were incorporated into assistive devices to help the physically challenged accommodate their impairment. Rehabilitation robotics, by contrast, fashions a new class of interactive and user-friendly robots that enhance the clinicians' goal of facilitating recovery by not only evaluating but also by delivering measured therapy to patients. Krebs and Hogan review pioneering clinical results in the field, discuss the growing pains of forging a novel technology, and outline the potential for a brilliant future.

Of the other two activities, we will limit our discussion to mechatronic systems. Orthotics and prosthetics may be considered as a category of assistive robotics. While the previous high water mark for mechatronic assistive technology occurred during the Vietnam War decades of 1960s and 1970s, recent advancements in materials, computers, and neuro-connectivity (neuro-prostheses) have reinvigorated research in this field. In fact, the lack of equivalent advances in realm of energy storage represents the only major hurdle preventing the realization of practical versions of Hollywood's fancies such as Star Trek's Commander Data or the Terminator. Durfee reviews pioneering developments in orthotics, Krebs and Hogan review upper-limb prostheses, and Herr reviews lower extremity prostheses. We will also discuss some emerging developments that could render some science fictions into reality.

48.2. REHABILITATION ROBOTICS

Rehabilitation Robotics encompasses an emerging class of interactive, user-friendly, clinical devices designed to evaluate patients and, also deliver therapy. Robots and computers are being harnessed to support and enhance clinicians' productivity, thereby facilitating a disabled individual's functional recovery. This development represents a shift from earlier uses of robotics as an assistive technology for the disabled. The new focus on mechanisms of recovery and evidence-based treatment together with developments facilitating safe human-machine interaction has paved the way for the surge in academic research, which started in early 1990's.

We can group devices into two main categories for the upper and lower extremity. For upper, extremity, Erlandson, (1990) described a patented robotic "smart exercise partner" in which the recovering stroke patient executes general spatial motions specified by the robot. Positive results using that system in a clinical setting were reported (Erlandson et al., 1995). However, patients had to be capable of moving independently or using the contralateral limb to move and guide the impaired limb (self-ranging). Independent movement is also essential to use of Rosen's 3-D controllable brake device (Maxwell, 1990), Rahman's functional upper limb orthosis (2000), and Burdea's pneumatically actuated glove (Merians et al., 2002). Other upper extremity robotic tools differ insofar as they do not require patients to be capable of independent movement; controlled forces can be exerted to move the patient or to measure aspects of motor status such as spasticity, rigidity or muscle tone. Lum (1993, 1995) described the design and application of robotic assistive devices focused on bi-manual tasks to promote motor recovery. More recently, Lum and colleagues (Reinkensmeyer et al., 2000; Burgar et al., 2000, Lum et al., 2002) used a commercial PUMA robot augmented by improved sensors to implement the MIME (Mirror Image Movement Enabler) system, in which the robot moves the impaired limb to mirror movements of the contralateral limb. Harwin (2001) is using another commercial robot (Fokker) to move the impaired arm in the recently initiated European Union sponsored project. While attempts to adapt or re-configure industrial robots for use in rehabilitation robotics appears to be a reasonable approach it suffers from a critical drawback: some twenty years of experience with industrial robots shows that low impedance comparable to the human arm cannot practically be achieved with these machines (intrinsically high impedance machines). In contrast to these approaches other groups have developed robotic technology configured for safe, stable and compliant operation in close physical contact with humans. For example, the MIT-MANUS robot developed in the Newman Laboratory for Biomechanics and Human Rehabilitation was specifically designed for clinical neurological applications and ensures a gentle compliant behavior (Hogan et al., 1995). Other low-impedance rehabilitation devices are Reinkensmeyer's ARM Guide (2000) and Furusho's EMUL (2003). Operationally, these robots can "get out of the way" as needed. They can therefore be programmed to allow the recovering stroke survivor to express movement, in whole or in part, even when the attempts are weak or uncoordinated. Whether this feature is crucial for effective therapy remains unproven but its importance for obtaining uncorrupted measurements of a patient's sensorimotor function has been established unequivocally (Krebs et al., 1998, 1999; Rohrer et al., 2002; Reinkensmeyer et al., 2002).

Evolving lower extremity devices are inspired mainly by gym machines and orthoses rather than by re-configured industrial robots. The best examples are Hesse's Elliptical Gait Trainer, Yaskawa's TEM, and the Lokomat (Hesse et al. 2000; Sakaki et al., 1999, Colombo et al., 2000). Not unlike their upper-extremity industrial robot counterparts, however, these designs suffer from a high impedance drawback. Most are presently being re-designed to modulate their impedance and to afford an interactive experience similar to the low-impedance lower extremities devices under development at MIT in the Newman laboratory and at UC Irvine.

Accompanying the vigorous development of rehabilitation robotic devices is an equivalent growth in clinical evaluations of device performance. Results include new insights into the recovery mechanisms for a variety of conditions, and into the rehabilitation techniques that best engage those mechanisms. Areas of research focus include not only stroke, but also motor deficits associated with diverse neurological, orthopedic, arthritic conditions. A number of studies demonstrated the exciting opportunities and benefits of integrating robotic technology into patients' daily rehabilitation program (e.g., Aisen et al., 1997; Burgar et al., 2000; Colombo et al., 2000; Fasoli et al., 2003 and 2004; Ferraro et al., 2003, Hesse et al., 2000; Krebs et al., 1998, 2000; Lum et al., 2002; Reinkensmeyer et al., 2000 and 2002; Volpe et al., 1999, 2000, and 2001). This chapter presents only our own results to delineate the potential of the technology and future directions for rehabilitation robotics. To date, we have deployed three distinct robot modules in collaborating clinical institutions[1] for shoulder-and-elbow, wrist, and spatial movements.

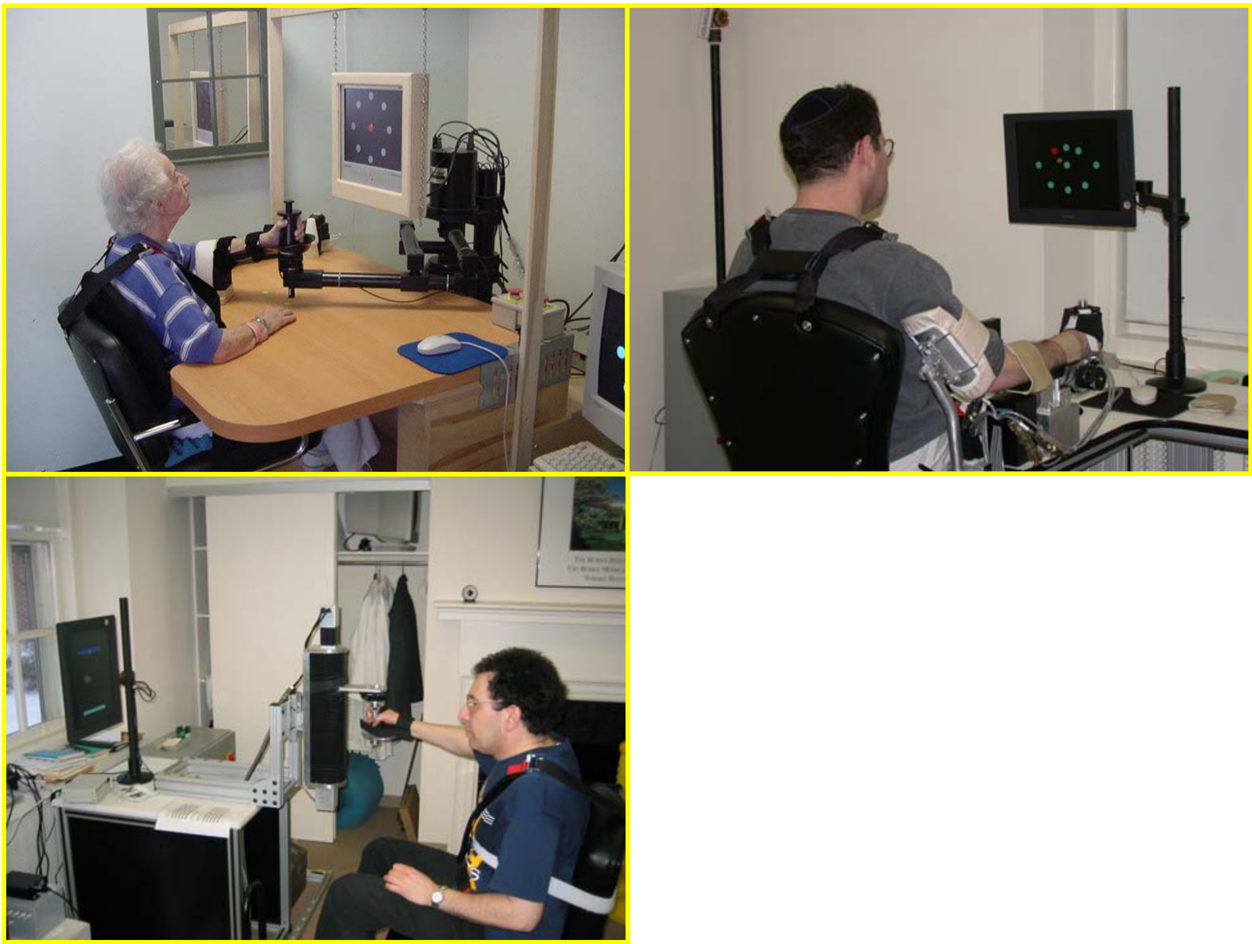

Figure 48.1. Rehabilitation robot modules during clinical trials at the Burke Rehabilitation Hospital (White Plains, NY). The figure shows on the top row the shoulder-and-elbow robot (MIT-Manus) and the wrist robot, and on the bottom row, the anti-gravity spatial module.

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[1] Hospitals presently operating one or more MIT-MANUS class robots include Burke (NY), Spaulding (MA), Helen Hayes (NY), Rhode Island (RI) Rehabilitation Hospitals, and the Baltimore (MD) and Cleveland (OH) Veterans Administration Medical

Volpe (2001) reported results of robotic training with 96 consecutive inpatients admitted to Burke Rehabilitation Hospital (White Plains, NY) who met inclusion criteria and consented to participate. Inclusion criteria were diagnosis of a single unilateral stroke within four weeks of admission to the study; the ability to understand and follow simple directions; and upper limb weakness in the hemiparetic arm (i.e. a strength grade of 3/5 or less in muscle groups of the proximal arm) as assessed with the standardized Medical Research Council battery. Patients were randomly assigned to either an experimental or control group. The sensorimotor training for the experimental group consisted of a set of "video-games" in which patients were required to move the robot end-effector according to the game's goals. If the patient could not perform the task, the robot assisted and guided the patient's hand. The sensorimotor training group received an additional 4-5 hours per week of robot-aided therapy while the control group received an hour of weekly robot exposure.

Although patient groups were comparable on all initial clinical evaluation measures, the robot-trained group demonstrated significantly greater motor improvement (higher mean interval change ± sem) than the control group on the impairment scales (see Table 48.1). These gains were specific to motions of the shoulder and elbow, the focus of the robot training. There were no significant between-group differences in the mean change scores for wrist and hand function, although there was a trend favoring the robot group. Likewise there were no significant differences in Functional Independence Measure (FIM) performance, used to indicate changes in the level of disability. Use of the FIM to indicate changes in the level of disability that may be associated with robotic therapy is not without limitations. Although the FIM is a well-established, reliable, and valid measure of basic activities of daily living, many of the self-care tasks that comprise the motor subscale of the FIM can be independently accomplished by using only the "unaffected" upper limb (Dodds et al., 1993; Granger et al., 1993). Therefore, a definitive

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conclusion about the relationship between changes in upper limb motor impairment, increased motor function, and reduction in disability is not possible with the FIM.

Between Group Comparisons: Final Evaluation Minus Initial Evaluation	Robot Trained (N = 55)	Control (N = 41)	P-Value
Impairment Measures (± sem)			
Fugl Meyer shoulder/elbow(FM-se) max/42	6.7 ± 1.0	4.5 ± 0.7	NS
Motor Power (MP) max/20	4.1 ± 0.4	2.2 ± 0.3	<0.01
Motor Status shoulder/elbow (MS-se) max/40	8.6 ± 0.8	3.8 ± 0.5	<0.01
Motor Status Wrist / Hand (MS/wh) max/42	4.1 ± 1.1	2.6 ± 0.8	NS
Disability Evaluation			
Functional Independence Measure (FIM upper) max/42	32.0 ± 5.0	25.5 ± 6.5	NS

Table 48.1. Mean Interval Change in Impairment and Disability Measure for Inpatients (significance $p < 0.05$). For all evaluations higher scores indicate better performance. Motor Power was only evaluated for shoulder and elbow movements.

Long after stroke, patients still suffer a variety of impairments dependent on the size and location of their individual stroke. Fasoli (2002, 2004) reported results of robotic training described at Spaulding Rehabilitation Hospital (Boston, MA) with 42 consecutive community dwelling volunteers with stroke, who responded to information they had obtained from media sources about the robotic training experiments. These patients received the same sensorimotor robotic training used with inpatients (see above). Prior to engaging in robotic therapy, these patients were assessed on three separate occasions to determine baseline function and to establish a within subject control. The primary outcome measures were the Fugl Meyer, Motor Status Scale for the shoulder and elbow, and the Motor Power score. Our baseline analyses revealed no statistically significant differences among any of the pre-treatment clinical evaluations, indicating the stability of chronic motor impairments in this subject group (Fasoli et al., 2002, 2004). However, after robotic training we found significant reductions in motor impairment of the hemiparetic upper limbs shown in Table 48.2. Results indicated statistically significant increases in each primary measure: Fugl-Meyer Test, Motor Status Score for shoulder and elbow, and the MP. Clinically, subjects reported greater comfort when attempting to move their hemiparetic limb, and were better able to actively coordinate shoulder and elbow movements when reaching toward visual targets during robotic therapy. This result has demonstrated that task specific robotic therapy can improve upper limb motor abilities and reduce chronic motor impairments, on average, 6 months after stroke. Others have obtained similar results (Burgar et al., 2000; Kahn et al., 2001; Lum et al., 1993, 1999, 2002; Reinkensmeyer et al., 2000; Shor et al., 2001).

Between Group Comparisons: Final Evaluation Minus Initial Evaluation	Robot Trained (N = 42)	P-Value
Impairment Measures (± sem)		
Fugl Meyer shoulder/elbow(FM-se) max/42	3.4 ± 4.0	<0.01
Motor Power (MP) max/40	2.1 ± 1.9	<0.01
Motor Status shoulder/elbow (MS-se) max/40	1.4 ± 2.0	<0.01

Table 48.2. Comparison of Mean Interval Change for Outpatients at Spaulding Rehabilitation Hospital (n=42 – sensorimotor protocol). For all evaluations higher scores indicate better performance. Motor Power was only evaluated for shoulder and elbow movements.

A critical aspect of low-impedance rehabilitation robots that often escapes clinicians is that these devices can be programmed to deliver a vast range of different kinds of therapy. For example, the sensorimotor training mentioned earlier can be described in lay terms as a "hand-over-hand" approach. In prior work Hogan (Flash and Hogan, 1985) has shown that normal reaching movements may be accurately described as "optimally smooth" in the sense of minimizing mean-squared jerk. While executing point-to-point movements, the robot controller uses this minimum-jerk profile as the reference hand path. The parameters of these minimum-jerk reference trajectories were obtained from therapists performing the same task at comfortable speed for training. During sensorimotor training, the robot uses a fixed impedance controller while following this fixed minimum-jerk reference trajectory.

To exemplify the potential of delivering different kinds of therapy, let's compare the results of Table 48.2 with those from an innovative robotic therapy modality developed in our lab based on motor-learning models. It applies a performance-based progressive algorithm and modulates the parameters of an impedance controller according to patient's performance (Krebs et al., 2003). This approach appears particularly suitable when considering that different stroke lesions can lead to quite different kinematic behavior during reach. For example, one patient might make rapid but poorly aimed movements, while another might aim well but move slowly. The novel feature of the performance-based progressive algorithm is that it can guide the hand of the patient that aims poorly without holding him/her back and assist the slow patient in making faster movements. Ferraro reported results of performance-based robotic training at Burke Rehabilitation Hospital with consecutive community dwelling volunteers with stroke, who responded to information they had obtained from media sources about the robotic training experiments (Ferraro et al., 2003). This protocol has the same enrollment criteria and lasted the same amount of time and number of sessions as the one reported by Fasoli (2002, 2004), but the outcome of the performance-based protocol represent a significant improvement over the sensorimotor one (see Table 48.3). In fact, if we exclude the severe strokes from the analysis of both protocols, persons with moderate stroke receiving the performance-based protocol improved twice as much as the ones receiving the sensorimotor protocol for both impairment and disability measures (Krebs et al., 2004 manuscript in preparation).

Between Group Comparisons: Final Evaluation Minus Initial Evaluation	Robot Trained (N = 29)	P-Value
Impairment Measures (± sem)		
Fugl Meyer shoulder/elbow(FM-se) max/42	5.6 ± 0.9	<0.01
Motor Power (MP) max/40	3.3 ± 0.7	<0.01
Motor Status shoulder/elbow (MS-se) max/40	2.7 ± 0.6	<0.01
Disability Evaluation (± sem)		
Functional Independence Measure (FIM) max/42	0.9 ± 0.7	NS
Disability Evaluation for Persons with Moderate Strokes (± sem)		
Functional Independence Measure (FIM) max/42	3.0 ± 0.6	<0.01

Table 48.3. Comparison of Mean Interval Change for Outpatients at Burke Rehabilitation Hospital (n=29 – performance-based protocol). For all evaluations higher scores indicate better performance. Motor Power was only evaluated for shoulder and elbow movements.

In conclusion, research in rehabilitation robotics has so vibrantly evolved during the last decade that it is now feasible to mingle robots with humans, supporting or participating in therapy activities. We envision a short-term future with a range of natural extensions of the existing rehabilitation robots including novel modules suitable to provide therapy for the fingers, for the ankle, and for over-ground walking. Since it appears that we can influence impairment and disability, the next step is to determine how to tailor therapy to particular patient needs and maximize the stroke survivor's motor outcome. Results to date are statistically strong and reproducible by different groups in different clinical settings, but the reported results are also arguably functionally modest. For the near-term future, we envision a range of clinical and neuroscience-based training paradigms addressing both functional abilities and impairment, harnessing the patient's potential to its limits with just movement-based therapy. It is not far-fetched to predict that by decade's end, we will have a range of rehabilitation robots at home and in the clinic, operating much like a gym. While movement-based therapy may lack the glamour of cure, for the long-term future we envision the combination of rehabilitation robotics working synergistically with novel pharmacological, neuro-regeneration or tissue-regeneration agents to achieve results equivalent to a "cure."

48.3. ORTHOSES

Orthoses are passive or powered external devices that support loads, or assist or restrict relative motion between body segments. The word orthosis is derived from Greek for making or setting straight, and is a general term that encompasses bracing and splints. Orthotics plays an important role in the rehabilitation of patients with motor impairments. Orthotics include devices for the neck, upper limb, trunk and lower limb that are designed to guide motion, bear weight, align body structures, protect joints or correct deformities. Unlike prostheses that replace a body part, orthoses are designed to work in cooperation with the intact body, and either control or assist movement.

Common types of lower limb orthoses include foot orthosis (FO) shoe inserts for correcting ankle and foot deformities, ankle-foot orthoses (AFO) for correcting foot drop, functional knee orthoses (KO) for athletic injuries, hip abduction orthoses for limiting range of motion, long leg knee-ankle foot orthosis (KAFO), and full length hip-knee-ankle-foot orthoses (HKAFO) for standing and gait stability. Trunk and neck orthoses include thoracolumbosacral orthoses (TLSO) for correcting scoliosis, lubrosacral orthoses (LSO) for stabilizing low back fractures, elastic trunk supports for preventing back injuries during lifting, and the common cervical orthoses (neck braces) for whiplash injuries or muscle spasms. Upper limb devices include shoulder and elbow slings for weight support during fracture healing, balanced forearm orthoses (BFO) for feeding assist, and an array of wrist, hand, and wrist-hand orthoses to position the joints or assist in activities of daily living.

Orthoses apply forces to resist or transfer motions and loads. For example, a knee brace restricts motion to the sagittal plane to protect a knee from off-axis forces following surgery or injury (Edelstein and Bruckner, 2002; Seymor, 2002). A hand orthosis can be constructed to provide elastic resistance to finger extension, thus enhancing a strengthening program following stroke (Fess and Philips, 1987). Plaster casts and wrist immobilization splints are orthoses that restrict all motion. In these cases, orthoses act very much like mechanical bearings whose purpose is to restrict motion in some dimensions and allow frictionless motion in others. A sling orthosis transfers loads from one part of the body to another. For example, a simple, single strap shoulder sling off loads the weight of the arm from the ipsilateral shoulder joint and applies it to the contralateral scapula and clavicle, thus preventing ipsilateral shoulder subluxation in those with rotator cuff injuries or paralysis. Foot orthoses are carefully designed to shift load bearing forces on the bottom of the foot from one area to another, typically to reduce pain or to off load pressure ulcers. Other orthoses transfer motion. A prehension orthosis contains a linkage that couples the wrist to the fingers. Extension of the wrist causes the fingers to close enabling patients with spinal cord injury at C6 to grasp objects (Edelstein and Bruckner, 2002).

The fit of orthosis is critical as they must carry loads without interfering with normal skin and tissue function (Fess and Philips, 1987). Of particular concern is excess pressure, particularly over bony prominences, that can lead to pressure ischemia and eventually skin ulcers. This is a particular problem when fitting patients with neuropathies or spinal cord injury who lack conscious sensation. A basic principle for the design and fitting of an orthosis is to spread the load over as large an area as possible. Avoiding bony prominences means the body attachment point for an orthosis is largely over soft tissue. This brings out an inherent tradeoff that continually challenges orthosis designers. The ideal orthosis should be rigidly anchored to the skeleton, while practical orthoses have considerable motion with respect to the skeleton because of the soft tissue that lies between.

Orthotic interventions are prescribed for patients with orthopedic or neurologic impairments (Nawoczenski, 1997). Orthopedic impairments result from chronic musculoskeletal disorders or acute musculoskeletal injuries, including athletic injuries. Ankle taping for athletes is a simple, custom fit orthosis to limit motion in the subtalar joint (Hemsley, 1997). The most common neurologic impairments where orthotic approaches are considered include traumatic brain injury, stroke, and spinal cord injury (Zablotny, 1997).

Lower limb orthotics prescribed for those with neurologic impairments have the function of restoring or improving gait (Zablotny, 1997). Typical objectives for walking orthotics are to establish stable weight bearing, to control the speed or direction of limb motion, or to reduce the energy required to ambulate. Simple AFOs are used to correct the foot drop that is a common byproduct of stroke, while KAFOs can improve gait for those with progressive quadriceps weakness. The challenge with designing and prescribing more involved orthotics is to generate an energy-efficient gait (Waters and Yakura, 1989). Walking with bi-lateral KAFOs and crutches requires five times the energy per meter as normal gait, while gait velocity is about one third normal. Wheelchairs are a faster, more efficient means of travel, and with the recent explosion in mobility device design and improved accessibility of buildings, wheelchairs are generally the device of choice for those with severe neurological impairments.

Several HKAFOs have been developed as paraplegic walking systems (Miller, 1997). There are two major walking systems. The first is the Hip Guidance Orthosis (HGO) that locks the knee joint but has freely moving hip and ankle joints (Major et al., 1981; Butler et al., 1984; Stallard et al., 1989). The second is the Reciprocating Gait Orthosis (RGO) that links opposite joints so that extension of the hip on one side leads to flexion on the contralateral side (Jefferson and Whittle, 1990). Although these systems can restore rudimentary gait for some people with spinal cord injury, the energy cost, and the size, weight and unwieldiness of the hardware has resulted in limited use (Stallard et al., 1989; Whittle et al., 1991).

Functional electrical stimulation (FES) is another means of providing assisted gait to those with spinal cord injury. In an attempt to overcome the limitations of FES and orthotic walking systems alone, hybrid systems that combine FES and orthotics have been developed. Several studies have shown improved gait speeds and lower energy consumption when FES and the RGO are combined (Solomonow et al., 1989; Hirokawa et al., 1990; Petrofsky and Smith, 1991; Isakov et al., 1992; Solomonow et al., 1997). Others have combined stimulation with the HGO (McClelland et al., 1987; Nene and Jennings, 1989); an enhanced AFO (Andrews et al., 1988), the Hybrid Assistive System (Popovic et al., 1989; Popovic 1990; Popovic et al., 1990), the Case Western hybrid system (Ferguson et al., 1999; Marsolais et al., 2000), the Strathclyde hybrid system (Yang et al., 1997; Greene and Granat, 2003), and the Controlled Brake Orthosis shown in Figure 48.2 (Goldfarb and Durfee 1996; Goldfarb et al., 2003).

The addition of power to an orthosis enables the external device to move a limb actively. Most commonly, powered orthoses are designed for use by individuals with spinal cord injury to restore modest function to a paralyzed extremity. Powered exoskeletons have a long and rich history, starting from the first robotics arm built in the 1960's at Case Western Reserve University, and the early powered walking machines pioneered by Tomovic and colleagues (Tomovic et al., 1973). For example, an EMG controlled battery powered hand orthosis can provide grasp for C5 quadriplegics (Benjuya and Kenney, 1990), while more ambitious multidegree of freedom, upper extremity exoskeletons have been tested in the laboratory (Johnson et al. 2001).

Powered lower limb walking systems have also been designed. Popovic developed a powered knee joint (Popovic et al., 1990), as did Beard (Beard et al., 1991). A hydraulic powered, five degree-of-freedom walking assist device was developed by Seireg and colleges in the 1970's (Grundman, 1981). More recently, Beleforte and colleagues have created the pneumatic active gait orthosis (G. Belforte, 2001) and Hiroaki and Yoshiyuki have developed the Hybrid Assistive Leg, a battery and DC motor driven powered exoskeleton (H Kawamoto, 2002).

Lab based powered exoskeletons have been developed for non-rehab, human power amplification applications. The most famous is the 1965 Hardiman whole body exoskeleton developed at the General Electric Research and Development Center in Schenectady, NY (Rosheim, 1994). Since then, a variety of human amplification systems have been developed for civilian and military applications, but none have made it out of the laboratory (Kazerooni and Mahoney, 1991; Rosheim, 1994; Kazerooni, 1996; J. Jansen, 2000; Neuhaus and Kazerooni, 2001).

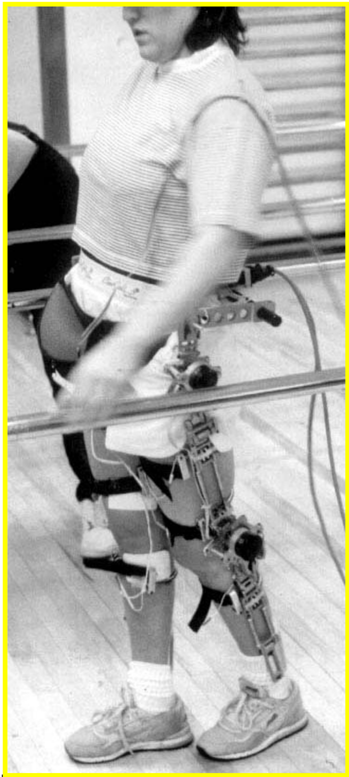

Figure 48.2. Laboratory version of a hybrid orthotic/FES system to enable rudimentary gait for some individuals with complete spinal cord injury at the thoracic level. The Controlled Brake Orthosis combines surface electrical stimulation of the lower limb muscles with an orthotic structure containing computer controlled brakes that regulate stance and swing phase trajectories (Goldfarb and Durfee, 1996, Goldfarb et al., 2003).

48.4. PROSTHETICS

48.4.1. Upper-Limb Amputation Prostheses

Despite advances in technology, progress in the development of effective upper-limb amputation prostheses has been modest. This may be partly due to irregular interest in their development, which tends to correlate with major wars. Mann (1981) shows an 1866 Civil War era below-elbow prosthesis with eating utensils and a hook. World War II saw the development by Northrop Aircraft cable-operated “body-powered” arm prosthesis. Mechatronic or “externally-powered"[2] prostheses were introduced in the Vietnam War decades. Unfortunately, the resulting devices offer limited benefits. Studies suggested that as few as 50% of upper-limb amputees use any prosthesis at all, versus 75% for lower-limb amputees (LeBlanc, 1973). Unilateral amputees (with one sound arm) overwhelmingly find that a prosthesis offers too little cosmetic or functional benefit to offset its discomfort and inconvenience. LeBlanc (1991) estimated that only 10% of prosthesis users in the U.S. (5% of the upper-limb amputee population) operate externally-powered devices. Thus half of all upper-limb amputees do not use a prosthesis and nine-tenths of those who do use the body-powered type, whose basic design is essentially unchanged in over half a century.

Yet mechatronic prostheses hold substantial promise. It is difficult to transmit significant power from the body to the prosthesis without severely compromising comfort; externally-powered devices avoid this problem. It is difficult to control multiple degrees of freedom (e.g., elbow, wrist, thumb, fingers) by recruiting other body motions; mechatronic prostheses may be controlled by bioelectric signals, which can be obtained from a large number of nerves or muscles including, in principle, those originally responsible for controlling the functions of the lost limb. The concept of using bioelectric signals to control a mechatronic device is due to MIT's Norbert Wiener who proposed it in his well-known work *Cybernetics* (Wiener, 1948). In founding cybernetics, the "study of automatic control systems formed by the nervous system and brain and by mechanical-electrical communication systems" he had "the idea several years ago to take an amputated muscle, pick up the action potentials, amplify this and make motion of it." Mann implemented this idea in an above-elbow amputation prosthesis controlled by EMG[3] from muscles in the residual limb (Rothchild and Mann, 1966; Mann, 1968), which led to

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[2] A "body-powered" prosthesis is powered by the amputee, e.g., bi-scapular abduction (shoulder rounding) for elbow flexion; mechatronic prostheses are typically powered by batteries "external" to the amputee.

commercial products including the Boston Arm (Jerard et al. 1974; Liberty Mutual, Inc.; Liberating Technologies, Inc.) and the Utah Arm (Jacobsen et al. 1982; Sarcos, Inc.; Motion Control, Inc.). EMG control has been applied to other upper-extremity motions including wrist rotation and grasp and methods for simultaneous control of many degrees of freedom have been proposed (Jerard & Jacobsen 1982).

Given the potential advantages of EMG control, the continued overwhelming preference for the body-powered system is remarkable. Although generally lighter and less expensive, a body-powered prosthesis is also considerably less comfortable, mostly due to the harness needed to transmit body motion the prosthesis; its lifting ability is extremely limited; it is less cosmetic than mechatronic models as it requires unnatural body motions for its operation; and independent operation of multiple degrees of freedom is difficult to impossible (LeBlanc, 1991). A clue to this puzzle may be found by studying how an arm amputation prosthesis is used.

Motion control and sensory feedback

One abiding concern is that EMG control may restore “forward path” communication between the central nervous system and the peripheral (bio-)mechanical system but does not restore “feedback” sensory communication from periphery to center. However, whether continuous feedback is essential for unimpaired movement control remains unclear with recent evidence suggesting that neural commands are substantially “pre-computed” from learned internal models (Shadmehr and Mussa-Ivaldi, 1994). Furthermore, substantial feedback information is available through mechanical interaction with the socket and harness that secure the prosthesis to the amputee. Doeringer & Hogan (1995) compared motor performance of unilateral amputees using a body-powered prosthesis with their performance using their unimpaired arm on a series

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[3] “EMG” refers to the ElectroMyoGram, more correctly “myoelectric activity”.

of elbow motion tasks. Motion control with the prosthesis was remarkably good. For the more active and experienced prosthesis users, eyes-closed positioning ability was indistinguishable from unimpaired arm performance.

Interaction control

Most functional tasks for an upper-limb prosthesis are *contact tasks* involving kinematically-constrained motions (e.g. opening a drawer, wiping a surface, etc.) or mechanical interaction (e.g. wielding a tool, cooperating with the other arm, etc.). For unilateral amputees (who constitute about 95% of the arm amputee population) a prosthesis will mostly serve as the non-dominant arm, e.g. holding or steadying objects while they are manipulated by the unimpaired limb. Controlling interaction is thus a fundamental requirement for arm prosthesis function. However, most mechatronic prostheses have been designed using motion control technology (Klopsteg et al. 1968, Mason 1972, Jerard et al. 1975, Jacobsen et al. 1982). Unfortunately, robotic experience has shown that motion controllers are poorly suited to controlling interaction.

A comparison with natural motor behavior is informative. Natural arms are compliant, yielding on contact with objects, which makes them less sensitive to the disturbances caused by contact (see Hogan, 2002 for a review). In contrast, a typical mechatronic prosthesis responds little, if at all, to external forces. Furthermore, the natural arm’s compliance is under voluntary control (Billian & Zahalak 1983, Humphrey & Reed 1983). Tensing muscles makes the arm less compliant and serves to stabilize it against disturbances. Relaxing the muscles makes the arm more compliant and allows it to accommodate external constraints.

The advantages of the natural arm’s behavior may be conferred on a machine using impedance control (Hogan 1979, 1985), which has proven effective for robot control, especially for contact tasks (see Hogan & Buerger, 2004 for a brief review). Impedance control has been applied to an

EMG-controlled mechatronic arm prosthesis, partially mimicking the natural arm's behavior (Abul-Haj and Hogan 1990): the response to external forces varies with co-activation of agonist and antagonist muscles while differential activation generates motion.

Amputee performance of contact tasks

A study of amputees performing simple contact tasks showed the importance of interaction control. A computer-controlled arm amputation prosthesis (see Abul-Haj and Hogan 1987 and Figure 48.3.) was programmed to emulate (1) a body-powered prostheses in free-swing mode; (2) the Boston Elbow and (3) the NY elbow (two mechatronic prostheses which control elbow velocity from the difference between EMG of remnant arm muscles); and (4) an impedance-controlled prosthesis. The main difference between these cases is that the velocity-controlled prostheses, (2) and (3), are unresponsive to external forces whereas the other two, (1) and (4), move easily under external forces. Unilateral amputee subjects used each prosthesis to turn a crank mounted in a vertical plane at three speeds (low, medium and fast, approximately 2, 4 and 6 rad/sec). In almost all cases, motions of prosthesis and crank were similar, the crank handle moving with a smooth, unimodal speed profile, indicating that all prostheses provided comparable motion control. In contrast, the forces exerted on the crank were significantly different in each case. Computation of the power produced by the prosthesis motor showed that for the impedance control system, prosthesis output power was always positive. For the body-powered prosthesis in free-swing mode power was zero. For the Boston Elbow and the NY elbow control systems, output power was *negative* for a significant portion of the task. Positive output power from the impedance control system means that it always assisted motion; amputees consistently rated this prosthesis easiest to control. Negative output power means that the Boston Elbow and NY elbow prosthesis were behaving as *brakes* and *impeding* performance of the task, not assisting it (Mansfield et al. 1992; Krebs et al., 1999).

Control and communication are the barriers

The way a mechatronic prosthesis responds to forces (its mechanical impedance) is clearly an important factor determining its usefulness. Unfortunately, most available mechatronic prostheses are unresponsive (they typically have high impedance) and this may be the main reason why, after decades of development, most amputees prefer not to use them, despite their apparent functional advantages such as independent control of different motions (e.g., elbow and terminal device).

Providing an amputee with the ability to adjust impedance (as in the natural limb) yielded superior performance and seems a promising way to improve upper-limb prostheses. However, it requires additional control signals to assess the user's intent (e.g., both reciprocal *and* co-contraction of antagonist muscles). For limited motions EMG may be suitable; activity of two muscles such as the remnant biceps and triceps may be used to command the elbow. However, the ideal mechatronic arm, assuming that one could be built, is not limited to elbow movement, but must also assess the user's intent to drive the wrist and fingers. Recent work on brain-machine interfaces (Nicolelis et al., 1995) shows that in principle the required information may be obtainable directly from the brain, though advances in technology for implantable neuro-electric recording in the periphery may be more practical (e.g., Bions – Chapter 33). New ways to communicate and more natural control strategies could reinvigorate research and open a plethora of possibilities to turn Hollywood's mechatronic fancies into reality.

Figure 48.3. Prosthesis Emulator

48.4.2. Lower-Limb Amputation Prostheses

Dissipative Knees and Energy-Storing Prosthetic Feet.

Today's prosthetic knees typically comprise a hydraulic and/or pneumatic damper that dissipates mechanical energy under joint rotation (Popovic and Sinkjaer, 2000). In these devices, fluid is pushed through an orifice when the knee is flexed or extended, resulting in a knee torque that increases with increasing knee angular rate. To control knee damping, the size of the fluid orifice is adjusted. Although for most commercial knees the orifice size is controlled passively when weight is applied to the prosthesis, some contemporary knee systems use a motor to actively modulate orifice size. For example in the Otto Bock C-Leg, hydraulic valves are under microprocessor control using knee position and axial force sensory information (James et al., 1990). Actively controlled knee dampers such as the C-Leg offer considerable advantages over passive knee systems, enabling amputees to walk with greater ease and confidence (Dietl and Bargehr, 1997; Kastner et al., 1998).

Today's prosthetic ankle-foot systems typically employ elastomeric bumper springs or carbon composite leaf springs to store and release energy throughout each walking or running step (Popovic and Sinkjaer, 2000). For example, in the Flex-Foot Vertical Shock Pylon System, carbon composite leaf springs offer considerable heel, toe and vertical compliance to the below-knee prosthesis, enabling leg amputees to move with greater comfort and speed. Although considerable progress has been made in materials and methods, commercially available ankle-foot devices are passive, and consequently, their stiffnesses are fixed and do not change with walking speed or terrain.

Patient-Adaptive Prosthetic Knee System

Using state-of-the-art prosthetic knee technology such as the C-Leg, a prosthetist must pre-program knee damping levels until a knee is comfortable, moves naturally, and is safe (Dietl and Bargehr, 1997; Kastner et al., 1998; James et al., 1990). However, these adjustments are not guided by biological gait data, and therefore, knee damping may not be set to ideal values, resulting in the possibility of undesirable gait movements. Still further, knee damping levels in such a system may not adapt properly in response to environmental disturbances. Recently, an external knee prosthesis was developed that automatically adapts knee damping values to match the amputee's gait requirements, accounting for variations in both forward speed and body size (Herr et al., 2001; Herr and Wilkenfeld, 2003). With this technology, knee damping is modulated about a single rotary axis using magnetorheological (MR) fluid in the shear mode, and only local mechanical sensing of axial force, sagittal plane torque, and knee position are employed as control inputs (see Figure 48.4). With every step, the controller, using axial force information, automatically adjusts early stance damping. When an amputee lifts a suitcase or carries a backpack, damping levels are increased to compensate for the added load on the prosthesis. With measurements of foot contact time, the controller also estimates forward speed and modulates swing phase flexion and extension damping profiles to achieve biologically realistic lower-limb dynamics. For example, the maximum flexion angle during early swing typically does not exceed 70 degrees in normal walking (Inman et al., 1981). Hence, to achieve a gait cycle that appears natural or biological, the knee controller automatically adjusts the knee damping levels until the swinging leg falls below the biological threshold of 70 degrees for each foot contact time or forward walking speed.

To assess the clinical effects of the patient-adaptive knee prosthesis, kinematic gait data were collected on unilateral trans-femoral amputees. Using both the patient-adaptive knee and a conventional, non-adaptive knee, gait kinematics were evaluated on both affected and unaffected sides. Results were compared to the kinematics of age, weight and height-matched normals. The study showed that the patient-adaptive knee successfully controlled early stance damping, enabling amputee to undergo biologically-realistic, early stance knee flexion. Additionally, the knee constrained the maximum swing flexion angle to an acceptable biological limit. In Figure 48.4, the maximum flexion angle during the swing phase is plotted versus walking speed for a unilateral transfemoral amputee using the non-adaptive, mechanical knee (top plot, filled diamonds) and the patient-adaptive knee (bottom plot, filled diamonds). In both plots, the subject’s sound side leg is shown (open squares), along with reference data from unimpaired walkers (standard error bars). For the amputee participant, the non-adaptive, mechanical knee produced a maximum flexion angle that increased with increasing speed, far exceeding 70 degrees at the fastest forward walking speed, whereas the patient-adaptive knee gave a maximum flexion angle that was less than 70 degrees and agreed well with the unimpaired, biological data. These results indicate that a patient-adaptive control scheme and local mechanical sensing are all that is required for amputees to walk with an increased level of biological realism compared to mechanically passive prosthetic systems.

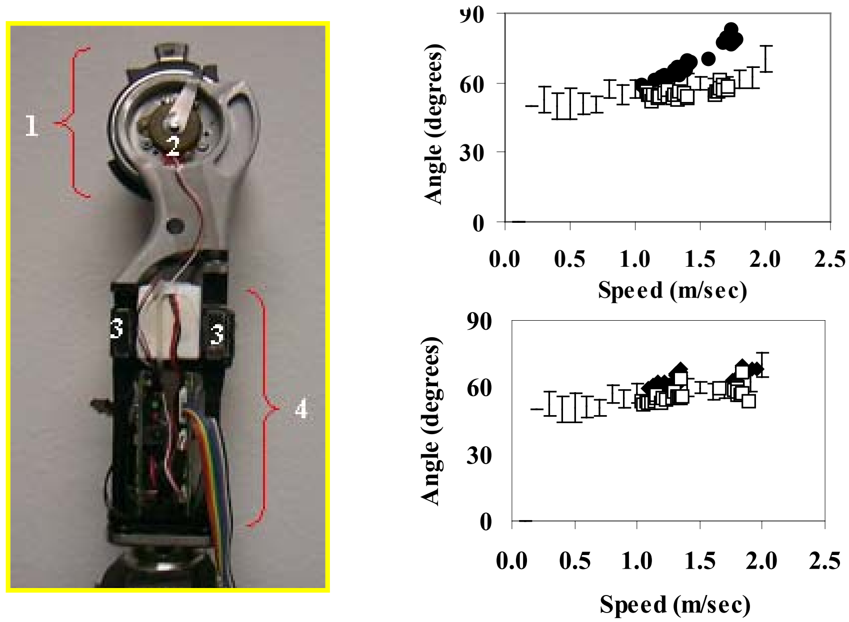

FIGURE 48.4. An external knee prosthesis for transfemoral amputees. The damping of the knee joint is modulated to control the movement of the prosthesis throughout each walking cycle. The prosthesis shown on the left comprises magnetorheological brake (1), potentiometerangle sensor (2), force sensors (3), and battery and electronic board (4). The right plots show the maximum flexion angle during the swing phase versus walking speed. The patient-adaptive knee affords a greater symmetry between affected and unaffected sides.

New Horizons for Lower-Limb Prosthetic Technology: Merging Body and Machine

Society is at the threshold of a new age when prostheses will no longer be separate, lifeless mechanisms, but will instead be intimate extensions of the human body, structurally, neurologically, and dynamically. Such a merging of body and machine will not only increase the acceptance of the physically challenged into society, but will also enable individuals suffering from leg amputation to more readily accept their new artificial appendage as part of their own body. Several scientific and technological advances will accelerate this mergence. An area of research of considerable importance is the development of improved power supplies and more efficient prosthetic actuator designs where both joint impedance and mechanical power generation can be effectively controlled in the context of a low-mass, high cycle-life, commercially viable prosthesis. Another critical area of research will be to combine local mechanical sensing about an external prosthetic joint with peripheral and/or central neural sensors positioned within the body. Neural prostheses such as the Bion (Loeb, 2001 – see Chapter 33), combined with external biomimetic prosthetic systems, may offer important functional advantages to amputees. The fact that only EMG or local mechanical sensors were employed in prosthetics imposes dramatic limitations in the system's ability to assess user intent. In the advancement of prosthetic systems, we feel that distributed sensory architectures are research areas of critical importance.

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