Rehabilitators, Robots, and Guides: New Tools for Neurological Rehabilitation

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1 Introduction

Millions of people in the United States suffer from movement disabilities as the result of neurological injury and disease. Their rehabilitation is labor intensive, often relying on one-on-one, manual interactions with therapists. For many disorders, it is unknown which types of therapeutic manipulations best promote recovery. In addition, patient evaluation is often done subjectively, making it difficult to monitor treatment effects. This chapter demonstrates how appropriately designed machines might be brought to bear on these problems.

The chapter is a cooperative effort from research groups at M.I.T., the Palo Alto Veterans Affairs Medical Center, the Rehabilitation Institute of Chicago, and the University of California, Berkeley, each of which is pursuing new devices for rehabilitation. Each group has focused its initial efforts on stroke, the leading cause of major disability in the United States (American Heart Association, 1992). Other potential target disorders exist, including traumatic brain injury, cerebral palsy, and Parkinson’s disease. In this chapter, stroke rehabilitation serves as a specific application illustrating what may be a more general opportunity.

1.1 Overview of Stroke Rehabilitation

More than 300,000 people suffer a stroke each year in the United States; three million stroke victims are alive today (Gresham et al. 1995). Motor deficits persist chronically in approximately one-half of these people (Gresham et al. 1979). Damage to neural areas responsible for controlling movement and concomitant disuse and persistent abnormal posture of the impaired limb cause a host of centrally and peripherally-based sensory motor impairments (Figure 38.1). Common impairments are decreased passive range of motion, weakness (Gandevia 1993), hyperactive reflexes (Thillmann et al. 1993), and incoordination, manifest in part as an inability to independently coactivate muscles (Devaux et al. 1993). Patients commonly experience spontaneous recovery, but are also treated with extensive physical and occupational therapy to enhance recovery.

This poststroke therapy is labor intensive, often relying on one-on-one interactions with a trained therapist several hours a day. Therapy extends for at least one or two months poststroke. Typical therapeutic activities include manual manipulations of the patient’s limbs, either with the patient passive or attempting movement (Figure 38.1). A therapist may stretch a patient’s limb into different positions to help improve passive range of motion, or move it through specific patterns thought to help reduce hyperactive reflexes. Sometimes the therapist will guide or support the limb as a patient attempts to activate desired muscle combinations or to achieve a desired movement or task. Motor activity is also sometimes facilitated by resisting movement, or by applying tactile, thermal, or proprioceptive sensory input to the limb (for a summary of different techniques, see Devaux 1987, Trombly 1995). Patient progress is often evaluated subjectively, with the therapist making hands-on or visual judgments about a patient’s movement ability. Sometimes standardized, coarsely discretized scales are used to improve objectivity (Gresham et al. 1995).

1.2 Use of Technology in Stroke Rehabilitation

Currently, technology is rarely used in stroke rehabilitation. Goniometers are sometimes used to measure range of motion of joints, and one degree-of-freedom (DOF) dynamometers to measure strength. There is a lack, however, of clinically viable, instrumented systems for measuring stroke-specific disturbances and determining how they contribute to reduced functional ability. For therapeutic manipulation, passive counterbalancing systems, such as mobile arm supports or overhead slings, are sometimes used to relieve the weight of the arm and thus allow the weakened patient to engage in self-initiated movement. Also, one DOF actuated systems, such as motorized dynamometers, are sometimes used for range of motion exercises or in active-assisted exercises, in which the patient tries to follow the motion of the machine in an attempt to develop volitional control of movement. There is a lack however of systems that replicate the complex spatial patterns of force and movement often used in manual therapy after stroke.

More sophisticated devices that would attach to a patient’s limb, measure movement and force generation, and apply therapeutic patterns of forces could make a significant impact on stroke rehabilitation management. Such machines could poten-
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Table 38.2: How rehabilitation machines might improve the cost efficiency of rehabilitation.

- Automatic repetitive aspects of therapy.
- Facilitate home care (teletherapy; patient at home, therapist at remote site).
- Make group sessions more effective.
- Help target therapy more accurately.
- Help determine appropriate "dosing" and duration for therapy.
- Increase therapy "dosage" where appropriate (i.e., more repetitions, more hours per day).

1.3 Overview of New Techniques

Research is just beginning on meeting these goals. This chapter describes initial work on "bimanual rehabilitators," a "teaching guide," "MIME," and MIT-MANUS, four devices for neurological rehabilitation of the arm (for descriptions of postural, locomotor, and other devices the reader is referred to Flores (1992), Wing et al. (1993), Siddiqi et al. (1994), and Erlanson (1995)). We emphasize that the blueprint for the ideal machine is not obvious from currently available information. Thus, the designs presented in this chapter are reasoned guesses at what might constitute useful machines. They have required distinct design choices regarding manipulation techniques, movement tasks, and actuator, sensor, and software configurations (Table 38.3). One goal is to give concrete examples of ways to approach these choices. The reasoning behind each of the resulting devices, along with illustrative data demonstrating some of each device's capabilities, are presented. The other major goal is to show how these devices are beginning to generate new ideas, information, and potential therapeutic techniques for neurological rehabilitation. The illustrative data in each section shows some of these results. In addition, the concluding section outlines several basic scientific questions raised by these devices, and possible directions for future technological innovation.

2 Bimanual Rehabilitators

2.1 Design Choices

This section presents devices called "bimanual rehabilitators," which were built to test the applicability of simple machines to stroke rehabilitation. Feasibility experiments with the devices demonstrate how a common therapeutic technique used in stroke rehabilitation can be implemented by the devices, at least for temporarily weakened, unpaired subjects. This suggests that machine implementation of some therapy techniques might not require complex devices.

Bimanual rehabilitators measure and assist bimanual control—coordination of the hands to achieve a task. The machines are task specific, and operate under simple feedback control laws. Applied to a hemiplegic stroke patient they would potentially allow the more able hand to train the less able one. Initial tests of the machines (Lum et al. 1993a; Lum et al. 1995) confirmed that such task-specific machines, operating under simple control laws, could provide adaptive assistance, performing only that part of the task that a weakened person could not perform.

Bimanual control is important for activities of daily living (Washam 1973). Imagine tying shoes, buttoning buttons, or washing dishes without the coordinated use of two hands. Bimanual coordination is so common that people have dedicated motor control systems devoted to two-hand activities: the two hands work together using different control patterns than either one would use separately (Reinkensmeyer et al. 1992); each hand "knows" by feedback mechanisms what the other hand is doing (Lum et al. 1993b); under some circumstances, each hand "knows" by feedback mechanisms what the other is about to do (Lum et al. 1992). Stroke generally affects one hand more than the other, so that bimanual activities are curtailed, but one hand can still provide motor intention, and perhaps training, for the other. For all these reasons, rehabilitation of bimanual control seems useful and possible.
2.2 Illustrative Data

Two bimanual rehabilitators were built and tested with unimpaired subjects. The Hand-Object-Hand device (Lum et al. 1993a) consists of two handles on a table top, each free to move about an axis coincident with the subject's wrist (Figure 38.3A). Potentiometers attached to the two axes measure position of the hands, and a force transducer mounted in a pencil-like manipulandum measures the grasp force between the outstretched fingers of the subject's hands. A motor under one handle can produce an external torque on one hand. The chief motivation for this device configuration was that it allows performance of bimanual tasks, with the possibility of powered assist for one hand, in a simplified, controlled setting.

In one preliminary experiment, the motor was driven by a simple proportional feedback controller regulating the position of the motor-driven hand. The subject was instructed to squeeze the object between the outstretched fingers. A blood pressure cuff was inflated on the upper arm of the motor-driven hand to occlude blood supply. After a few minutes, ischaemia in this hand and forearm caused a reduction in force and sensation, a crude model of hemiparesis. As the subject's muscles produced less force, the device contributed more by stabilizing the weakened hand, allowing the subject to continue to perform the squeezing task (Figure 38.3B).

In another experiment, subjects were asked to move the object back and forth, maintaining a grasp force adequate to avoid dropping the object. Again, ischaemia was used to create a reduction in force and sensation. Instead of hand position, grasp force was fed back to the motor such that the device assisted the subject in transporting the object back and forth, even though one hand lacked strength and sensation to accomplish the task.

These initial results encouraged us to build a more practical device that could provide adaptive assistance during an activity of daily living. The Bimanual Lifting Rehabilitator (Figure 38.4A) was designed to measure and perturb movements during the lifting of a large object, such as a cafeteria tray or large pot. The device has a handle and force transducer for each hand attached to one rigid link. A second link is connected to this one through a one DOF bearing, and to a motor. The subject attempts to lift the link by the handles, without tilting it. A potentiometer connected to the bearing measures tilt, and tilt is regulated using a simple control law (Lum et al. 1995). If the object begins to tilt, the motor corrects, assisting the impaired hand.

In an experimental test of the device, it was demonstrated that the device can substitute for assistance from the rehabilitator (solid). The third row overlays the normal left hand force (dotted), left hand force while receiving assistance from the rehabilitator (solid), and the motor force (dashed, near zero). Averages of 30 trials. (Adapted from Lum et al. 1995, © 1995 IEEE.)

The left hand of a subject (Figure 38.4B). Using only the right hand, a subject could lift the object with only a small tilt (in an experiment with a cafeteria tray affixed to the device, a large soft drink did not spill). When an unimpaired subject lifted the tray with both hands, the device produced no force. Thus, a simple control law provided appropriate assistance (either full or none) in a bimanual task used in activities of daily living.
2.3 Discussion

These preliminary experiments show how bimanual rehabilitators might replicate adaptive assistance for bimanual therapy. Applied to stroke, the devices could give a measure of the amount of assistance needed in tasks of daily living, providing a means to track improvement and provide motivation to patients. Also, adaptive assistance is a common therapy technique that allows weak and uncoordinated patients to practice functional movement patterns. The simple devices used here have the potential to quantitatively evaluate such assistance as a physical rehabilitation technique. Likewise, bimanual therapy could be rigorously tested using the devices.

The control laws used to implement assistance for movement were simple proportional feedback controllers. In principle, similar assistance could even be provided for some tasks by purely mechanical and purely passive devices. That the bimanual rehabilitators require few actuators would make them more economical. However, the task specificity of the devices is a potential weakness: a therapist might need many such devices if there is little transfer of learning between tasks.

3 Reaching Guide

3.1 Design Choices

The device presented in this section, called a "reaching guide," makes use of a passive constraint to guide one-handed reaching attempts (Figure 38.5). The motivation for this design was that therapists often provide a strategic support or constraint for the arm during patient-initiated movement. The reaching guide is suited to explore the extent to which such guidance is therapeutic. Also, it was hypothesized that the range of motion along the constraint, as well as the extent and direction of guidance forces provided by the constraint during movement, might be used to quantify recovery.

Reaching is fundamental to many activities of daily living and requires sophisticated multimuscle coordination (e.g., see Chapters 25–31). It thus seemed a good target task for a therapeutic device. In addition, because reaching movements typically follow nearly straight-line trajectories, a simple linear bearing can be used to passively steer movement along the desired trajectory. However, the device has to meet several challenges because there is inherently safe and potentially inexpensive. In the event that actualized assistance is deemed desirable, the guide incorporates a motor for moving the arm along the linear constraint (Figure 38.5B).

For the reaching guide, a linear bearing (LB in Figure 38.5A) is attached to a computer-controlled brake (CB). The bearing can thereby be locked at different elevation angles (θ) in the vertical plane. Movement along the constraint, and the orientation of the constraint, are measured with two optical encoders (OE), one attached to a chain drive. Contact forces and torques against the constraint are measured by a 6-axis force-torque sensor (FT) (For brevity only one force component—the mediolateral contact force shown schematically in Figure 38.5B—is considered here). The device also incorporates a computer-controlled motor (M) and second brake not used in this study but which could be used to give assistance or resistance to reaching along the guide.

3.2 Illustrative Data

As an initial test of the device, we compared guided reaching movements by unimpaired and hemiplegic stroke subjects (see also Reinkensmeyer et al. 1999). We were interested in whether the maximum reach a subject could achieve along the guide at different angles (called the "workspace boundary"), and the contact forces generated against the guide, could be used to quantify movement impairment after stroke.

Six unimpaired subjects and three hemiplegic stroke subjects were tested. The three hemiplegic stroke subjects had sustained hemispheric strokes at least one year prior to testing and were qualitatively classified as mildly, moderately, and severely impaired by an experienced physical therapist. The subjects had no obvious sensory impairment and showed no evidence of neglect of the impaired side.

To measure the workspace boundary, subjects lay supine on an adjustable patient table with the torso along the edge of the bed. The supine position was chosen initially for comfort, but other positions such as sitting and standing are also possible. The height and location of the bed were adjusted to align the approximate center of the head of the numerus, determined by palpation, with the axis of rotation of the guide. The guide was then locked at a series of elevation angles (θ = 0°, 60°, 10°, 50°, 20°, 40°, 30°, 20°, 10°, 50°, 10°, 60°) [see Figure 38.5]. At each angle, the subject started with the elbow flexed against a backrest and reached out as far as possible, "smoothly and steadily," eight times. Instructions were given not to push to the right or left against the guide, or against the top or bottom of the guide during reaching.

Using this technique, we found that unimpaired subjects reach very consistently to the same maximum reach (avg. SD = 0.5 cm.). Also, the average difference in workspace boundary location for the left and right arms of unimpaired subjects is about 2.0 cm. This number is a sort of minimum "resolution" for workspace measurement with the guide, taking into account the inherent variability between left and right arms, and variability in aligning a subject with the guide.

For the stroke subjects, the repeatability of the maximum reach was approximately the same or decreased only slightly: the average standard deviations of the maximum reach for the hemiplegic arms were 0.5, 2.5, and 1.8 cm, compared with 0.5, 0.7, and 1.1 cm for the corresponding contralateral arms. This suggests the subjects reliably achieved their maximum reach, even after stroke.

Also, for the three stroke subjects, the average differences in workspace boundary location between the hemiplegic and contralateral arms were 3.8, 13.7, and 25.8 cm (Figure 38.6), exceeding the 2 cm device "resolution" found with unimpaired subjects. For all three subjects, the boundaries for the hemiplegic arms were shrunken particularly in the humeral flexion.

Finally, even though all subjects were instructed to minimize contact force against the guide, they generated substantial non-zero medio-lateral contact forces during reaching (Figure 38.7). The patterns of force differed between unimpaired and hemiplegic arms: all unimpaired arms generated a laterally-directed force during reaching (average magnitude 13.5 N). The hemiplegic arms began the movement pushing laterally but changed to pushing medially as they reached outward along the guide (i.e., during elbow extension and shoulder flexion). This spatial pattern of contact force is consistent with the clinically observed abnormal "extension synergy" in which the shoulder is involuntarily adducted and internally rotated during efforts at elbow extension (Brunnstrom 1970; Reinkensmeyer et al. 1999).

3.3 Discussion

By accurately measuring range of movement and multiaxial force generation, the reaching guide provides a means to objectively assess functional ability. A key idea in this work is that the forces normal to the desired direction of movement along a constraint contain valuable information about a patient's movement impairment. Measuring such constraint forces allows quantitative assessment of abnormal synergistic control. This is important because currently there are few objective methods available to analyze impaired arm coordination in the clinic.
4 MIME

4.1 Design Choices

This section discusses a prototype robotic device, called MIME ("Mirror Image Movement Enhancer"), capable of implementing two commonly used therapeutic techniques: passive and active assisted movements. Normally, these techniques are applied by a therapist who moves the paretic limb as the patient either remains passive, or actively attempts to contribute to the movement. MIME can substitute for the human therapist during such exercise tasks.

MIME uses two standard mobile arm supports that limit movement to the horizontal plane, and a 6-DOF robot arm that applies forces and torques to the paretic forearm through the support (Figure 38.8). Use of the arm supports for support against gravity allowed the robot arm to be relatively small, while still allowing performance of coordinated shoulder and elbow tasks. Movements of the arm supports can be controlled with programmed forearm position and orientation trajectories (for which experimental data is presented here), or by a position feedback control system that slavishly tracks the trajectories to the movements of the contralateral limb. Optical encoders on the joints of the arm supports measure the position and orientation of the forearms, and a 6-axis force/torque transducer measures the forces/torques applied to the paretic limb.

This approach is different than the other approaches presented in this chapter in that it uses a commercially available industrial robot, and a standard clinical mechanism—the mobile arm support. The design thus builds on established technologies from the robotics and rehabilitation industries, thus reaping all the many previous design iterations that went into these technologies. For example, in producing an exoskeleton capable of supporting or assisting the arm, or use of existing technologies to control a safe system with general movement capability quickly and cost-effectively.

As a first step in evaluating the utility of MIME, we investigated whether improved performance in active-assisted movements (as measured by the interaction forces/torques during robotic assistance) correlates with improved functional recovery of the paretic limb (as measured by an upper extremity Fugl-Meyer examination—a standard clinical exam) in hemiplegic stroke patients (see also Lam et al. 1999). If robot assisted measurements are to yield insight into the mechanisms of lost function, it is important to understand how those measurements correlate with functional ability. Also, it is important to establish that improved performance ability of a robot exercise task correlates with improved functional recovery, if practicing that exercise task is intended to have functionally significant, therapeutic effects.

4.2 Illustrative Data

Each test session began with an upper extremity Fugl–Meyer examination intended to gauge the subject’s arm function with an accepted clinical scale (Fugl-Meyer et al. 1975). Subjects were then strapped into a modified wheelchair, and their forearms were secured in the arm support. Six different point-to-point reaching movements were then tested (Figure 38.9), first with the subject passive and the robot moving, and then with the subject actively moving along with the robot.

The desired movement trajectory for the robot was based on the forearm trajectories of unassisted,

![Figure 38.7. Mediolateral contact force during reaching. Top row shows spatial averages of mediolateral contact force for three hemiplegic subjects during eight reaches with the guide locked at 0 degree elevation. Bottom row shows contact forces for three unimpaired subjects.](image)

![Figure 38.8. MIME](image)

![Figure 38.9. Top view of the five movement types. Movements 1a, 2a, 3a start from the dashed positions and end at the solid arm positions. Movements 1b, 2b, 3b are in the reverse directions.](image)
2.5 second duration, movement trajectories measured from three uninjured subjects. Those trajectories were ensemble averaged across subjects to yield six "normal" trajectories. Those trajectories were then programmed into the robot and used as a reference trajectory for a joint-based PID controller supplied with the robot.

For each desired point-to-point movement, the subject was first instructed to remain passive as the robot moved the limb in the programmed trajectory five times. The position and orientation of the forearm, and the interaction forces/torques between the parietal arm and the robot were sampled and stored.

Next, the subject was instructed to voluntarily contribute to movement by pushing the robot with approximately one pound of force as it moved along the pre-programmed trajectory. After each trial, the subject was given knowledge of his average force level in the direction of movement and encouraged to increase or decrease his effort accordingly. Ten active-assist trials were collected.

For each desired point-to-point movement, we found the passive force/torque profiles were very consistent. Thus, to estimate the passive force/torque level at each point during the movement, the passive profiles were ensemble averaged. For each active trial, the forces/torques voluntarily generated were then estimated by subtracting the average passive profile from the active profile.

From this estimated voluntary profile we derived several parameters for characterizing performance. The "force magnitude" was defined as the magnitude of the average force vector generated during movement. The "force direction" error of the parietal limb was calculated as the angle between average force vectors generated from the left and right arms. "Work efficiency" was estimated by calculating the work done by the parietal limb during movement, and normalizing it by the "potential work," defined as the work that would have been produced during a trial if at each instant the force magnitude were directed precisely in the desired movement direction, and the torque magnitude were oriented precisely in the desired direction of rotation. Finally, the "normal force per unit work" was defined as the average force magnitude normal to the movement scaled by the positive work done during that movement. The normal torque per unit work was defined similarly.

All of these parameters were calculated for each active trial, averaged over the movement duration, averaged across trials, and finally averaged across movement types to reach a scalar value that reflected the subject's average performance during the session. Averaging across movement types was done to compare with the Fugl-Meyer (FM) score, which reflects overall limb ability.

Using these techniques for three uninjured and seven stroke subjects, we found that the more severely impaired subjects produced higher force magnitudes than less impaired subjects, but these forces were often misdirected relative to the direction of movement, and relative to the force directions generated by their contralateral limb during the same movements (Figure 38.10). There was a negative correlation between the force magnitude error during a trial versus the FM score (p < 0.02), with the most impaired subjects (FM scores of 14-17) producing forces of approximately 10 N, and unimpaired arms producing forces of 6 N. The force direction error of the parietal limb also was negatively correlated with the FM score (p < 0.01), with the directional errors of the most impaired subjects approximately 70 degrees, and those for the unimpaired arms less than 30 degrees. Finally, the stroke subjects showed a positive correlation between the work efficiency of the parietal limb versus the FM score (p < 0.02), and negative correlations between the normal forces and torques per unit work versus the FM score (p < 0.02 for forces, p < 0.01 for torques).

4.3 Discussion

The experiments with MIME quantified impaired coordination after stroke. During active-assist movements, stroke subjects pushed harder against the robot than unimpaired subjects, but with greater directional errors, creating larger normal forces and torques. They ended up doing increased work. All of these parameters were strongly correlated with the FM score, suggesting that robot-assisted performance has functional relevance.

These results, coupled with the positive results reported from clinical trials of MIT-MANUS presented in the next section, are motivating for a clinical test of the therapeutic efficacy of exercise with MIME. The MIT-MANUS study showed that increased robotic therapy resulted in increased upper limb recovery, an exciting and potentially groundbreaking result. We plan to build upon this result by testing if robotic therapy is better than traditional upper limb therapy, and if the active aspect of the therapy is necessary. A test group will exercise each day with our device in addition to their normal therapy, while a control group will receive additional traditional upper limb therapy. A second control group will practice the same movements as the test group in a passive device that constrains the endpoint motion to one direction while providing various levels of loading in that direction, similar to the linear reaching guide presented in the previous section. This will allow us to determine if an actuated system is absolutely necessary for therapeutic efficacy, or if a passive system would suffice.

5 MIT-MANUS

5.1 Design Choices

As described in the introduction, neurorehabilitation is labor intensive, relying on therapy and evaluation procedures that are typically administered by a single clinician working with a single patient; indeed, this is true of much of the practice of clinical neurology. Labor-intensive procedures suggest that robotics and information technology may be used to improve quality, enhance documentation and reduce cost, and that has been the vision guiding the development of MIT-MANUS. However, at the outset of this research there was no firm evidence that the manipulation of patients' limbs that dominates current neurologic rehabilitation practice (and that might be aided by robotics) has a positive effect on recovery from brain injury. In short, the kind of assistance a robot could provide might not matter.

In a pilot clinical trial, we used MIT-MANUS (Hogan et al. 1995) a robot designed for neurological applications to assess whether manipulation of the impaired limb influences motor recovery in hemiparetic stroke patients (Krebs et al. 1998). Unlike most industrial robots, MIT-MANUS (Figure 38.11) is configured for safe, stable, and compliant operation in close physical contact with humans. This is achieved using impedance control, a key feature of the robot control system (Hogan 1985). Its computer control system modulates the way the robot reacts to mechanical perturbation from a patient or clinician and ensures a gentle compliant behavior. MIT-MANUS can move, guide, or perturb the movement of a subject's or patient's upper limb and can record motions and mechanical quantities such as the position, velocity, and forces applied. The machine was designed to have a low intrinsic end-point impedance (i.e., be backlash driveable), with a low and nearly isotropic inertia and friction, and be capable of producing a predetermined range of forces and impedances. We emphasized this design requirement primarily to facilitate control of robot impedance and to ensure that the robot's intrinsic dynamics would be minimally encumbering to the
These twenty sequential patients with a single CT-verified stroke hospitalized on the same acute rehabilitation ward at the Burke Rehabilitation Hospital (White Plains, New York) were enrolled in either a robot-aided therapy group (RT, N = 10) or a standard therapy group plus "sham" robot-aided therapy (ST, N = 10). Both groups were comparable in age, physical impairment, and time between onset of the stroke and enrollment in the trial (mean age: RT 58.5, ST 63; mean admission to rehab in weeks since stroke onset: RT 2.8, ST 3.2). All patients were blinded to the treatment group and were assigned to the same blinded clinical team. Both groups received conventional therapy; the RT group received an additional 4 to 5 hours per week of robot-aided therapy typically for 6.5 weeks consisting of peripheral manipulation of the impaired limb correlated with audiovisual stimuli, while the ST group had an hour weekly robot exposure. The sensory-motor training for the RT group consisted of a set of "video-games." Patients were required to move the robot end-effector according to the game's goals, which included drawing circles, stars, squares, diamonds, and navigating through graphical windows. If the patient could not perform the task, the robot assisted and guided the patient's hand.

Mean-driven software allows the clinician to choose different values of impedance (very soft, soft, medium, hard, very hard). However, we opted to use the same soft range (100 N/m, 2 N·sec/m) throughout the trial primarily because of patients' pretrial preferences and to minimize any risk of exacerbating joint or tendon pain. For this trial, the impedance controller was implemented using a nonlinear position and velocity feedback structured to produce constant isotropic end-point stiffness and damping. Coupled to our highly back-driveable design, the stability of this controller is extremely robust to the uncertainties because of physical contact (Hogan 1988; Colgate and Hogan 1988).

The training for the RT group was similar to the RT group, except that half of the one hour session consisted of playing the video games with the unimpaired arm and half the session with the impaired arm while the robot passively supported the arm and provided visual feedback. If the patient could not perform the task, he/she used the unimpaired arm to assist and complete the game, or the clinician assisted. The purpose of this "sham" robot therapy was to give the minimum extra "therapy" to the control group yet (1) provide all patients with comparable motivation and attention (2) blind patients and clinicians to which treatment group a patient was in, and (3) familiarize the control group with the machine so that robot-aided evaluation would be feasible and meaningful. All patients were evaluated by the same blinded therapist with standardized clinical assessment procedures and robot-measured assessment of movement kinematics. The standard assessment procedures included measurements of functional status (FIM—Functional Independence Measure) (Dodds et al. 1994), upper extremity motor impairment (F-M—Fugl-Meyer Upper Extremity Subscores, MSS—Motor Status Score, MP—Motor Power) (Fugl-Meyer et al. 1975; Aisen et al. 1995), as well as pain and shoulder-hand syndrome.

Results of this pilot clinical trial indicated that robot-assisted therapy positively influenced recovery of patients. Mean baseline FIM, F-M, MSS, and MP at admission indicated no statistically significant differences between the control and experimental groups. Nonetheless, at discharge the experimental group outperformed the control group in all the clinical assessments of the motor impairment involving shoulder and elbow (limb segments exercised with the robot-aided therapy) (Table 38.4). There was a clear trend in the F-M and MP scores favoring the experimental group (p < 0.20 and p < 0.10), and a statistically significant improvement in the MSS for shoulder/elbow (p < 0.05) (Aisen et al. 1996). At the same time, the measurement of functional status (FIM) indicated no statistically significant difference between groups (p = 0.98). We believe that this may be the result of the non-specific nature of the FIM score which broadly assesses overall functional recovery and the fact that both groups had the same standard therapeutic experience aiming at functional recovery. There was no significant difference in frequency between groups for pain in joints and tendons, or shoulder-hand syndrome. Even more encouraging, the MSS results' time history suggests that the control group did not improve after 5 weeks from the stroke onset, while five patients in the experimental group continued to improve up to 8 week's poststroke.

5.3 Discussion

For a number of reasons (most prominently the small sample size) these results should be inter-
and technological innovation. The answers, along with the answers to cost, safety, and ease of use questions not discussed here, will play a key role in determining the ultimate pattern of use of this technology.

6.1 Questions Regarding Improved Measurement with Machines

The scientific basis for improved measurement with machines is an important area for future research. Experiments in this chapter suggest that guiding and assisting forces, including those normal to the desired direction of movement, contain useful information. What is the diagnostic and predictive value of these parameters? Other quantitative methods have been developed for measuring weakness (Gandevia 1993), spasticity (Rymer and Katz 1994), and abnormal synergies (Beer et al. 1995; Dewald et al. 1995) in stroke patients. Can these methods be incorporated into machine design? What is the minimum set of measures needed to adequately characterize impairment and recovery?

A chief motivation for measurement with machines is increased sensitivity in gauging functional recovery. Improved sensitivity could help clinicians to better judge patient progress and the efficacy of different therapy techniques. However, none of the devices in this chapter has yet demonstrated functional measurement at higher resolution than is possible with current clinical scales. There is good reason to question the feasibility of high resolution measurement in neurological rehabilitation: human voluntary movement is inherently variable, impaired movement even more so. To what extent is improved resolution even possible? Can machine-based measurement do better than clinical scales?

6.2 Questions Regarding Therapeutic Manipulation with Machines

Besides machine-based measurement, the scientific bases of therapeutic manipulation are also a key area for future research. Encouragingly, the clinical trial of MIT-MANUS showed that providing additional therapy with a manually assisting robot can improve recovery. This suggests that more therapy can be better, and that machines might help provide this extra therapy. Future research will need to clarify what components of the additional therapy—the peripheral manipulation, the attempts by patients to activate descending pathways, or the enhanced sensory stimulation, to name a few potential candidates—are essential. What are the key components of therapy and how are they best implemented?

It is also important to better understand the nature of transfer from machine therapy to everyday function. In motor learning research with unimpaired subjects transfer of learning between tasks has been found to be small (Schmidt 1988). However, the experiments with MIME and MIT-MANUS suggest that transfer may exist between planar arm movement and overall arm function. What is the nature of this transfer? Could exercise with simple, task-specific devices like the bimanual rehabilitators generate functional transfer? To what extent do performance improvements extend beyond the exercised limb and the specific action used in therapy? What are the mechanisms of transfer and how are they best exploited?

6.3 Questions Regarding Further Technological Innovation

Finally, there is an opportunity for further understanding of the interaction of movement. In designing new devices, one must ask: How do they differ? How will their effects be modified? How do they affect movement? What is the optimal design for a given task? These questions have been answered in part by the experiments described in this chapter, but further research is still needed. The development of new devices will require careful consideration of the underlying mechanisms of movement and the goals of therapy.
Other therapeutic modalities might be combined with manual manipulation. Would tactile stimulation make a difference? What about neuro-muscular stimulation (the focus of Section IX of this book)? Are the emerging "virtual environment" technologies applicable? They may hold the promise of providing tools to address cognitive deficits as well as sensory-motor deficits. Telerobotic technology might enable remote monitoring and intervention (see also Chapter 37). Portable and even wearable sensors, actuators, and information processors (e.g., built into "smart fabrics") are technologically feasible. Can home-based self-therapy be effective?

These are a few of the questions that will help determine the ultimate pattern of use of machine technology in neurological rehabilitation. Answering these questions will ultimately require two types of effort. One is to learn by doing: build the technology and try it out. Much of our work is in this stage now, and more research along these lines is needed. It is also essential to understand the outcomes. Scientific analysis complements design synthesis, and often the two are synergistic. We suggest, however, that this chapter has begun to provide support that rehabilitators, robots, and guides will provide a new medium for both rehabilitation innovation and science.

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References

Commentary: Adaptive Rehabilitators, Robots, and Guides
Dava J. Newman

The devices and data presented in this chapter highlight promising new results and systems for neurological rehabilitation. Most importantly, machine-based techniques were shown to have a significant impact on stroke rehabilitation through different machine designs ranging from simple control and geometry to complex multidegree-of-freedom (DOF) devices. The devices were reported to replicate complex motions, quantify measures of impairment, and even aid recovery according to functional and clinical measures. In the conclusions section the authors pose the question whether their results (and devices) will impact neurological rehabilitation. I agree with them that this is the essential question and very difficult to answer. I will use the three categories (measurement, therapy, and technological innovation) proposed for future research directions as the basis for my commentary. As suggested, the potential exists for improved measurements using machines and information technology. I would like to suggest that comprehensive experimental studies are necessary to iden-
I recommend underwater therapy as a potential for musculoskeletal rehabilitation, and this technique could be coupled with machine rehabilitation. Recovering patients will find relief in the lighter loading and increased mobility of underwater submersion.

Additional technological assessment tools include: noninvasive kinematic analysis (e.g., limb or whole-body motions), piezoelectric force sensors, shape memory alloys to provide resistive or actuation forces, and tactile stimulation (see also Chapter 37). There are many commercially available systems to consider. As alluded to in the chapter, future technological advances might allow us to wear “smart suits”, envisioned as lightweight garments with sensors, actuators, and computers woven into the fabric. These technologies could revolutionize rehabilitation methods.

Finally, I believe that the patients’ acceptability of the machines as advantageous rather than as intrusive should be given serious consideration.

Reference