Visual Feedback Manipulation for Hand Rehabilitation in a Robotic Environment

Bambi Roberts Brewer

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Carnegie Mellon University
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Abstract

The ultimate goal of this research is to manipulate visual feedback in order to make robotic therapy more effective than traditional human-assisted and robotic rehabilitation. Toward this end, this thesis examines the limits and effects of visual feedback distortion, a type of visual feedback manipulation, in a robotic environment. Visual feedback establishes a metric of performance for a given task, and visual feedback distortion is defined as a change in this metric such that a change in performance is required to cause the same visual response. In order to design a therapeutic program utilizing visual feedback distortion, it is crucial to identify the amount of distortion that is imperceptible. Thus, the first experiment measured the amount of distortion that is imperceptible (the Just Noticeable Difference, or JND) for age-matched, unimpaired subjects and motor-impaired subjects. The second experiment showed that vision dominates kinesthetic feedback in a robotic environment and that gradual visual distortion beyond 1 JND increases force production and movement distance within a single experimental session. Finally, the third experiment examined the effects of distortion of movement error during a two-finger coordination task. The results of this experiment showed that error distortion is effective in directing a subject's attention to a specific finger but does not improve terminal performance. These foundational experimental results were used to design a game-like therapeutic paradigm incorporating visual feedback manipulation. In initial tests with three chronic stroke and traumatic brain injury patients, all patients followed the visual feedback manipulation to levels of performance above that predicted by their initial assessment at each session. Furthermore, all patients showed functional improvements after participation in the study. Visual feedback manipulation has shown promise for therapy in a robotic environment; more work is needed to further explore the ramifications of visual feedback manipulation for robotic rehabilitation and to spread this technique to clinical practice.

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Chapter 1: Introduction

Almost one million strokes and traumatic brain injuries occur each year in the U.S. [1, 2]. Most survivors require rehabilitation to address motor deficits resulting from the stroke or brain injury; for instance, 50% of stroke survivors exhibit some hemiparesis six months post-stroke [1]. Currently, therapists use techniques such as passive movements of joints and active task practice to restore function, and they also teach compensatory techniques to increase functional independence. Robotic therapy has been proposed as an addition to traditional rehabilitation to enable new types of therapeutic exercise and to increase the amount of therapy available to each patient.

Robotic therapy has been shown to increase strength, range of motion, and possibly functional performance in stroke patients. Such improvements are hypothesized to result from robotic therapy's focus on intensive, repetitive movements. Intensive practice increases use of an impaired limb [3] and encourages cortical reorganization [4], which can improve the ability of the patient to control the limb. In addition, robotic therapy can use visual feedback delivered on a computer monitor to transform repetitive practice into an engaging game. For instance, Boian et al. [5] use a flight simulation game to make ankle exercises more fun for patients. However, visual feedback in rehabilitation also has the potential to manipulate the patient's perception of therapeutic exercise. Visual feedback in a robotic rehabilitation environment can be used to control the information received by a patient about his or her performance, and this control of visual feedback can be used to address psychological variables that may influence therapy.

Patients may not strive for difficult goals in therapy due to entrenched habits [3] or personality variables such as low self-efficacy or a fear of failure [6, 7]. Visual feedback manipulation can be used to encourage patients to move beyond an established level of performance. The ultimate goal of this research is to use manipulations of the visual feedback given to a patient to make robotic therapy more effective than traditional human-assisted therapy and previous robotic rehabilitation applications. Toward this end, this thesis examines the limits and effects of visual feedback distortion, a type of visual feedback manipulation, in a robotic environment. A flowchart outlining the work presented here is shown in Figure 1.1. Though visual distortion could be applied to almost any task, we focused on tasks designed involving fine motor control in the hand.

Visual distortion refers to visual feedback that establishes a metric of performance for a given rehabilitation task and then gradually changes this metric such that improved performance is required to cause the same visual response. Different types of visual distortion are possible, and two are considered here: goal distortion and error enhancement. Goal distortion means that the subject is shown a visual display of a quantity such as distance or force, and while he or she aims for a particular point on this display, the performance required to reach that point is gradually changed. For example, gradually increasing the amount of joint extension required to reach the top of a visual feedback bar might encourage a subject to extend the joint farther without realizing that he or she is doing so. Error enhancement means visually magnifying the subject's error in relation to a given target performance. For instance, if the subject is attempting to move the arm along a specific trajectory, deviations from that path could be visually magnified.

The first set of experiments conducted for this thesis addressed the limits of visual distortion with unimpaired subjects. For a therapeutic program involving distortion to be most effective, patients must not detect the visual distortions as they interact with the robot. When a subject believes visual information to be unreliable, he or she begins to rely more on the kinesthetic sense [8], which could reduce the influence of visual distortion on a patient's perception of therapeutic exercise. With this in mind, the first objective was to measure the Just Noticeable Difference (JND) for produced force and movement distance/finger position in unimpaired subjects. The JND of a physical dimension is the smallest percentage change in the dimension that can be reliably perceived. The force and position JNDs of unimpaired subjects provide a lower bound on the amount of distortion that is imperceptible.

For distortion to affect performance during rehabilitation, patients must not only fail to detect the distortion, but they must rely primarily on visual feedback when performing tasks in our robotic environment. Previous research has demonstrated that when vision and kinesthesis concerning a movement are in conflict, the subject's perception of the movement tends to be dominated by the more precise feedback source, which is often, but not always, vision [9]. Further experiments with unimpaired subjects were conducted to show that vision dominates kinesthetic feedback in a robotic rehabilitation environment for a simple, one-finger task and that gradual visual distortion beyond one JND can be used to manipulate force production and movement distance within a single experimental session. These experiments were conducted with subjects who performed within the limits of their ability.

To determine the effects of distortion on performance of a more complicated task, a difficult learning task involving coordinated movement of the index finger and thumb was studied. Distortion influenced the division of attention between fingers but did not affect terminal performance.

We used these foundational experimental results to design a game-like therapeutic paradigm incorporating visual feedback manipulation. This rehabilitation protocol focused on improving extension of the index finger and bringing the thumb into a position opposing the index finger to establish a functional pinch and release useful for manipulating small objects. Hand function has been shown to be important in predicting patient ability to carry out self-care activities [10], but the majority of the previous applications in robotic therapy have focused on the arms or legs. Two types of visual feedback manipulation were considered, visual distortion and visual progression. Visual progression refers to veridical visual feedback emphasizing and encouraging gradual improvements in performance. In initial tests with three chronic stroke and traumatic brain injury subjects, all subjects followed the visual feedback manipulation to levels of performance above that predicted by their initial assessment at each session. Furthermore, all subjects showed functional improvements after participation in the study.

Chapter 2 reviews the background for this work. Chapter 3 describes the experimental environment used for much of the foundational work. Chapter 4 details the experiments we used to measure the Just Noticeable Differences for force and distance/position. Chapter 5 relates how vision dominated kinesthesis for a one-finger force or distance production task, while Chapter 6 discusses the effects of distortion during a more complicated two-finger task. Chapter 7 outlines our preliminary therapeutic work. Chapter 8 describes the game-like therapeutic paradigm we designed to investigate the use of visual feedback manipulation in rehabilitation. Chapters 9-12 contain initial tests of this paradigm with three chronic stroke and TBI subjects. Chapter 13 summarizes the contributions of this dissertation.

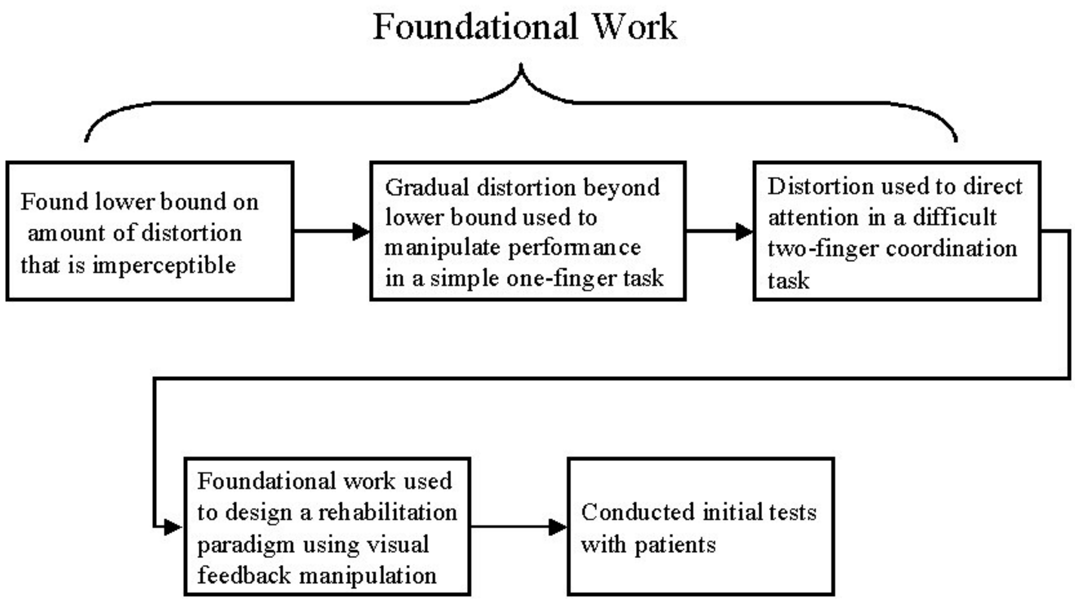


Figure 1.1. Flowchart outlining the work described in this thesis.

Chapter 2: Background

2.1. The Just Noticeable Difference

The Just Noticeable Difference, or difference threshold, has been considered in psychology since the work of E.H. Weber in 1834 ([11], cited in [12]). Gescheider [12] describes three methods traditionally used to measure this quantity for a given physical variable: the method of constant stimuli, the method of limits, and the method of adjustment. In the method of constant stimuli, the user is presented on each trial with a standard stimulus and one of a number of comparison stimuli. There are usually 5-9 comparison stimuli, and they are presented in random order. The experimenter calculates the percentage of trials on which each comparison stimulus is judged larger than the standard stimulus and designates the JND as the difference between the standard stimulus and the comparison stimulus that was judged larger 75% (or 25%) of the time. In the method of limits, the subject also compares a standard stimulus and a comparison stimulus on each trial, but the comparison stimulus starts at a value much lower than the standard and is made gradually larger from trial to trial. As the comparison stimulus becomes gradually larger, the subject says that it is equal to, then greater than, the standard. After the comparison stimulus reaches a value clearly greater than the standard, the experimenter begins making the comparison stimulus gradually smaller. This process is repeated several times, and the JND is calculated as half of the average difference between the value of the comparison stimulus at which the subject transitioned between judging the comparison as "equal" and judging it as "greater" and the value at which the subject transitioned between "equal" and "less." In the method of adjustment, the subject is allowed to adjust a comparison stimulus until he or she judges it equal to a standard. This is repeated a number of times, and the JND is calculated as the standard deviation of the subject's responses.

A more recent method of measuring the JND relies on signal detection theory, and presents the subject with trials consisting of two stimuli that are the same or differ by a constant amount. Using some assumptions about the distribution of sensory responses corresponding to each stimulus, the JND can be calculated based on the proportion of hits (trials on which the stimuli were different and the subject said "different") and false positives (trials on which the stimuli were the same and the subject said "different"). This method is outlined in Pang, Tan, and Durlach [13]. We chose this method because it allows an accurate calculation of the JND using a small number of distinct stimuli.

Many muscle groups in the arm and hand tested under various conditions have been found to have a force JND in the range of 7-10%. The force JND is usually represented as a percentage (or Weber fraction) because it has been found to be a constant fraction of the starting force [12, 13]. For instance, Brodie and Ross [14, 15] measured a JND of 10% for weights actively lifted in the palm by a movement of the forearm about the elbow and a JND of 8% for weights shaken up and down in the hand. Jones [16] found a JND of 7% for forces produced by the elbow flexor muscles. Pang et al. [13] measured the JND for pinching force over a range of base forces, displacements, and velocities. The JND was roughly 7% over all these variations. A force JND experiment closely related to ours was conducted by Raj et al. [17]. This group measured a JND of 11-12% for weights lifted by the middle finger about the MCP joint. All of these experiments were conducted with young subjects.

In terms of distance or position discrimination for the hand, Durlach et al. [18] and Ernst and Banks [9] measured the Just Noticeable Difference in the distance between the index finger and the thumb for young subjects. Durlach et al. obtained a value of 1-2 mm and found that the JND did not obey Weber's law. Thus, the JND for distance/position should not necessarily be expressed as a percentage. Ernst and Banks obtained a value of 3.3 mm for the finger span JND.

The effects of age on the JNDs for force and position have not been previously measured. However, age has been found to decrease the tactile sensibility of the hand as measured by two-point discrimination, the distance by which two contact points must be separated before they are perceived as distinct [19, 20]. In addition, researchers interested in other dimensions have found increased JNDs in elderly subjects. For instance, Fitzgibbons and Gordon-Salant [21] measured the Just Noticeable Difference for young and elderly subjects for the rate of auditory pulses in a sequence and for a single interval between two pulses. They found that the JNDs for elderly subjects were roughly twice those of young subjects. Similarly, Shinomori et al. [22] found that age increased the JND for wavelength for some optical channels. Gescheider et al. [23], on the other hand, found no significant difference in the JNDs for young and elderly subjects for vibrotactile amplitude. However, the test amplitudes used for each subject were expressed in terms of that subject's sensory threshold, and the thresholds for elderly subjects were significantly larger than those of young subjects. The effect of stroke or TBI on the JND for the hand for force and position has not been well researched, but stroke has been found to decrease a patient's ability to discriminate joint position on the impaired side [24].

2.2. Sensory Integration and Visual Dominance

Parameter estimation methods from statistics have been used to study how the brain performs sensory integration and how dominance of a given sensory input can result. Humans have been shown to integrate sensory cues in a way that approximates the statistically optimal solution [25]. Ernst and Bank [9] showed that subjects in the presence of an intersensory discrepancy combine the discordant inputs by weighting each using the reciprocal of its variance. This type of integration minimizes the variance of the final intermodal estimate that the subject uses for task performance. Thus, inputs with low variance tend to influence perception more than high-variance inputs. For example, vision tends to dominate perception in height discrimination because it provides more precise information for this task than kinesthesis [9]. Van Beers, Sittig, and Denier van der Gon further showed that the weight given to each sensory cue can vary with direction if precision is dependent upon direction. Sober and Sabes [26] showed that the weights given to two sensory modalities can differ depending upon the calculation being performed. Different calculations are performed in different reference frames, and Sober and Sabes suggest that changes in the precision of each modality across reference frames could account for their results. However, modality precision does not account for all aspects of sensory integration. Drawing attention to a particular modality can increase the influence of that input [27], and learning can also change the weighting of perceptual cues [28]. Previous studies indicate that multisensory integration involves a variety of areas in the brain, including the superior colliculus [29], as well as the anterior congulate cortex, the inferior parietal lobules, the left dorsolateral prefrontal cortex, and the left claustrum/insular cortex [30].

The dominance of visual feedback in sensory integration has been widely reported. Situations in which vision dominates kinesthetic and tactile feedback include the perception of shape [31], limb position [32], haptic depth [33], and stiffness [34]. However, haptic dominance of vision has been observed for aspects of texture [35]. In addition, haptic dominance has been observed when subjects recognize a discrepancy between kinesthesis and vision [36] or when vision is blurred [8]. We found that vision dominated kinesthesis in our robotic environment, most likely because the visual feedback provided had a resolution higher than that of the subject's kinesthetic feedback. This work is described in Chapter 5.

2.3. Hand Function and Rehabilitation after Stroke or Traumatic Brain Injury

After a stroke, a patient may exhibit a reduced ability to grasp and manipulate objects. These problems may be due to mechanical problems such as reduced range of motion at the joints and spasticity caused by strong stretch reflexes [37, 38]. Spasticity particularly affects the finger flexors, making extension of the fingers difficult for patients. Though stroke patients also exhibit reduced hand strength [39], many patients with sensory deficits actually increase their grip force to avoid dropping objects [40]. The combination of these two factors can cause patients to become quickly fatigued. Stroke patients may have difficulty moving the fingers of the affected hand independently [39, 41, 42]. All of these issues contribute to the fact that even four years after stroke, half of all stroke patients have a non-functional arm [43]. TBI patients may also exhibit spasticity, as well as slow motor responses and sensory deficits [44]. They may have trouble with fine motor tasks, such as picking up small blocks [45]. TBI frequently results in a number of cognitive deficits [46], but only subjects functioning at a high cognitive level were included in the experiments described here.

The goal of hand rehabilitation after a stroke or head injury is to restore the function and independence of patients in activities relevant to self-care, employment, or hobbies.

Self-care activities, or activities of daily living (ADLs), are given particular emphasis. To enable patients to regain functional independence, occupational therapists use two types of techniques. Restorative techniques encourage the patient to relearn how to perform a task as an unimpaired person would. Compensatory techniques, on the other hand, focus on modifying the patient's home environment or teaching the patient a pattern of movement different from that used by unimpaired persons. We will concentrate on restorative techniques here.

Functional improvement with motor relearning is hypothesized to be the result of cortical reorganization and the reversal of habits of nonuse acquired after injury ("learned nonuse") [47]. Literature in occupational therapy states that therapists can best encourage motor relearning through practice of functional tasks, with tasks gradually being made more difficult [48]. For patients who have not reached a level of recovery permitting task practice, therapists may evoke reflexes to help patients regain movement in the impaired hand, or they may passively move and stretch the impaired hand to reduce the effects of hypertone (increased resistance to movement) [49].

Our rehabilitation paradigm was designed for patients who have some active finger flexion and some thumb movement. The goal was for these patients to relearn a pincer grasp. The pincer grasp, also known as a tip grasp, is a pinching grasp used to pick up small objects such as coins or beads [50]. The pincer grasp requires opposition of the thumb and can occur between the pads of the index finger and the thumb, or between their tips (neat pincer grasp), as shown in Figure 2.1.

Occupational therapy has recently begun incorporating results from motor learning to consider what types of practice schedules, contexts, and feedback optimize transfer of motor skills from therapy to the home environment. Flinn [51] reviews that varying the order and the context in which tasks are practiced improves retention of motor skills, and practicing a task as a whole rather than in parts is considered better for a discrete task such as grasping [48]. In addition, the results of Constraint-Induced Movement Therapy show that increasing the amount of task practice by patients positively influences functional performance and cortical reorganization [4, 52]. The use of natural objects and environments is recommended by occupational therapy texts to promote functional improvement [48, 53]. Also, retention of a learned skill has been found to be negatively affected by continuous, high-resolution feedback [51]. For optimal retention of motor skills, subjects must learn to rely on intrinsic feedback, rather than extrinsic feedback from the experimenter [54]. However, the negative effect on retention of frequent extrinsic feedback appears to be less pronounced for stroke patients, though motor learning as a whole seems similar [55]. These results were used in designing the robotic environment used with stroke and TBI subjects (see Chapter 8).

2.4. Previous Work in Robotic Rehabilitation for the Upper Extremity

Traditional human-assisted therapy for the upper extremity based on the principles described in the previous section has been augmented with several robots designed to administer different modes of therapy to the impaired limb. These robots are haptic interfaces, meaning that they allow a patient to interact with a computer by displacing or exerting force on the robot and by feeling forces exerted on them by the robot (force feedback). Patients also usually receive visual information about their performance from a computer monitor or other display. Robotic therapy for the upper limb can focus on movements of the shoulder, elbow, wrist, and/or hand.

Most of the robotic systems that have been used for rehabilitation of the upper limb have been designed help stroke patients relearn to make reaching movements with the arm. This group of robots provides rehabilitation primarily for the shoulder and elbow. The MIT-MANUS is the most widely tested robot of this type (Figure 2.2). It is a 2 DOF robotic arm that consists of a five bar linkage SCARA (selective compliance assembly robot arm) [56]. A stroke patient moves a cursor on a computer screen to a target by moving the endpoint of the robotic arm in a horizontal plane. The robot is designed so that the patient experiences minimal impedance when moving the endpoint of the arm. Force provided by two brushless motors can be used, if necessary, to guide the patient's movement and help him or her to complete the target reach [57]. This mode of operation is known as an active-assisted mode. An extension to the MIT-MANUS now allows about 19" of vertical movement in addition to movement in the horizontal plane [58], and a robot for wrist rehabilitation has also been constructed [59]. During trials of the MIT-MANUS, fifty-six acute stroke subjects received about four hours per week of reaching practice with the robot in addition to a traditional human-assisted rehabilitation program [60]. Forty control subjects received the standard rehabilitation with 1-2 hours per week of exposure to the robot (robot motors were never turned on). Each control patient spent half of this time reaching with the unimpaired arm and half reaching with the impaired arm, assisting with the unimpaired arm when necessary. After the four weeks, robot-treated patients showed greater gains in muscle control and muscle strength than the control patients [60], but increased gains for the robot-treated group were seen only for shoulder and elbow movements like those practiced with the robot. Robot-treated patients did not show greater improvement than controls on a test of functional performance of everyday activities [60], though this was reported earlier for a subgroup of 56 patients [61, 62].

The "Mirror-Image Motion Enabler" (MIME) is another haptic interface designed for rehabilitation of the elbow and shoulder (Figure 2.3). The MIME system uses a 6 DOF industrial robot, a PUMA-560 robot arm [63]. The forearm of the impaired limb rests in a splint that prevents movement of the wrist and hand; the splint is coupled to the robot arm via a 6-axis force sensor. Using the MIME system, a patient can make reaching movements to targets in one of four modes [64]. In the passive mode, the patient relaxes while the robot moves the arm, and the active-assisted mode completes target movements initiated by the subject. In the active-constrained mode, the patient moves in the target direction against a viscous (velocity-dependent) resistive force. The system uses spring-like forces normal to the target direction to encourage the patient to move only in the desired direction. The fourth mode, the bilateral mode, is designed for hemiparetic patients, stroke patients who have one impaired limb and one normal limb. In this mode, the system measures the movements that the patient makes with the unimpaired limb while guiding the impaired arm so that its movements mirror those of the normal arm. Burgar, Lum, Shor, and Van der Loos [64] hypothesize that mirror-image movements will stimulate activity in neural pathways important to stroke recovery. As a test of the MIME system, 13 chronic stroke subjects received 24 one-hour reaching sessions with the robot, while 14 control subjects devoted an equal amount of time to practice of daily tasks with the impaired arm [63]. The robot-treated group showed significantly greater gains in shoulder and elbow strength and reach extent. At a six-month follow-up, the experimental group showed a significantly larger improvement on a test of independence in daily tasks.

The Assisted Rehabilitation and Measurement Guide (ARM Guide) is a simpler, relatively inexpensive system designed to provide therapy to improve reaching performance after stroke (Figure 2.4). This system has also been used to study various factors that impair reaching in stroke patients [65]. The ARM Guide features a linear track that can be adjusted in two dimensions to allow massed reaching practice in a chosen direction. The patient makes 1 DOF reaching movements with the forearm strapped to a splint that moves along this track. The ARM Guide can operate in either a passive or an active-assisted mode, with the necessary force provided by a chain drive and motor attached to the forearm splint [65]. When a group of seven patients who practiced reaching movements with ARM Guide for 24 one-hour sessions were compared to seven controls who practiced reaching for the same amount of time without the robot, the two groups showed comparable improvements in the time to complete functional tasks and the straightness of reaching movements [66]. The robot group also showed an improvement in trajectory smoothness, possibly because the robot imposed a smooth trajectory during practice.

The last robotic system for providing shoulder and elbow rehabilitation that will be discussed is the GENTLE/s system. In this system, a commercially available 3 DOF robot, the HapticMASTER from FCS Robotics, is attached to the wrist via a gimbal with 3 passive DOF [67]. This system suspends the upper limb from a sling to eliminate gravity. The system can provide therapy in three possible modes: passive, active-assisted, and an active mode similar to the active-constrained mode of the MIME system [68]. When compared to task practice with the arm suspended in the sling, the GENTLE/s system seems to yield a greater rate of improvement for some physical variables such as shoulder range of motion and the ability to perform certain arm movements.

Moving away from systems for the rehabilitation of the shoulder and elbow, Hesse, Schulte-Tigges, Konrad, Bardeleben, and Werner [69] have constructed a robot designed to train pronation/supination of the forearm (twisting of the forearm about its long axis) and flexion/extension of the wrist (back and forth movement of the hand). Both of the patient's arms interact with the device. This haptic interface can move both limbs passively, or the unimpaired limb can be used to move the impaired limb in a mirror-like fashion, similar to the bilateral mode of the MIME system. This device also has a bilateral mode in which the impaired arm must overcome an initial resistance before it follows the path of the unimpaired arm. The robot has been tested with 12 hemiparetic patients. The therapy reduced excessive tone (tightness of the muscles) in the impaired wrist and fingers, but these effects were not sustained at a three-month follow-up. This may be because patients spent less time interacting with the system than in comparable studies.

Robotic systems for hand rehabilitation have not yet been widely investigated. Boian et al. have constructed a system that provides force feedback to the hand via a Rutgers Master II haptic glove [70]. The Rutgers Master II uses four pneumatic actuators to apply force to the thumb and the index, middle, and ring fingers [71]. This device also contains infrared and Hall Effect sensors that can be used to estimate the posture of the hand. Position information for the hand is also measured by the Cyberglove, a sensor glove available from Immersion Corporation. The Cyberglove contains 18 bend sensors that measure the position of a variety of joints in the fingers and hand. These two devices are used for various virtual reality exercises designed to improve a patient's range of motion, strength, and speed, as well as his or her ability to move the fingers independently. The Rutgers Master II and the Cyberglove are used for different exercises, rather than simultaneously. The system has been tested with four stroke patients, and improvements in range of motion and independence of movement were seen. These preliminary results also indicate corresponding improvements in patients' performance in a reaching and grasping task.

This brief literature review shows that a variety of robotic techniques have been applied to the problem of upper extremity rehabilitation. These approaches have shown improvements in physical parameters such as strength and range of motion, though few significant functional improvements have been shown. The positive effects of robotic rehabilitation may be largely due to the massed practice paradigms that have been used in these studies. For example, the control condition of the ARM Guide study demonstrated that equivalent amounts of practice without a robot could yield similar physical gains.

Even when robotic rehabilitation does not yield greater gains than intensive rehabilitation with a human therapist, augmenting the limited number of human therapists with robotic devices can allow more patients to receive enough therapy to reach optimal outcomes. In addition, robotic devices could enable rehabilitation to occur in the home with remote supervision by a human therapist. It is, however, crucial to explore ways in which robots can facilitate new types of therapy not feasible for a human therapist to provide. One such example is the bilateral mode seen in the MIME system. This thesis work explored the use of visual feedback to manipulate patients' performance, and this precise control of visual feedback is not possible without a robotic environment.

As mentioned in the previous section, the robotic environment described in Chapter 8 worked with a subject's index finger and thumb to restore a functional pincer grasp. Despite the importance of grasping and manipulation in daily activities, little robotic work has been done for hand rehabilitation. In addition, few robotic studies have examined how visual feedback delivered during rehabilitation affects patient outcome, though the possible entertainment value of visual feedback has been noted [5]. This thesis goes farther to discuss how the information given to patients visually during rehabilitation affects therapy. This work concerning distortion and visible progression is an important first step in exploring how visual feedback can be used to improve robotic rehabilitation.

2.5. Transfer after Learning in a Virtual Environment

For rehabilitation in a robotic or virtual environment to be useful, the skills learned in that environment must transfer to real-world tasks. This issue is particularly important given the emphasis in occupational therapy on functional objects and tasks and the lack of functional improvement seen in many studies of robotic rehabilitation. However, Rose et al. [72] found that for a simple sensorimotor task, training in a virtual environment was as good as or better than training in the real-world environment when both groups were tested for retention in the real-world environment. In addition, Todorov, Shadmehr, and Bizzi [73] used a virtual environment to train subjects to execute a difficult table tennis shot by following an expert's trajectory. Subjects trained in the virtual environment performed the real-world task better than subjects trained with an equivalent amount of real-world practice. Holden et al. [74] demonstrated that two stroke subjects were able to learn reaching trajectories in a virtual environment and transfer those skills to real-world reaching tasks. Because these studies show that motor learning can transfer from a virtual to a real environment, even for stroke patients, we expect that skills learned in our robotic environment will transfer to real-world tasks.

2.6. Motivation for Visual Feedback Manipulation in Rehabilitation

An estimated 20-25% of stroke victims exhibit a phenomenon called "Learned Nonuse," in which cortical damage causes patients to develop movement deficits by learning to move only within certain limits [3]. With rehabilitation and cortical reorganization, patients relearn larger and more powerful movements to a certain extent, but the habit of decreased movement may prevent them from regaining their original mobility and strength. Because patients do not recognize that they are capable of greater movement, they may perform below their capacity in rehabilitation. Visual feedback manipulation provides a way of asking patients to move beyond their learned limits.

The two visual feedback manipulations considered in this thesis were chosen based on two possible models of how patients set and pursue goals. Patients may set themselves challenging goals and increment these goals as they improve, or patients may prefer to practice at a constant level of difficulty below their actual level of ability. This section will discuss three theories that describe these two possibilities: Bandura's theory of self-efficacy, Atkinson's theory of achievement motivation, and Dweck's theory of goal orientation.

The term "self-efficacy" refers to "the belief one has in being able to execute a task successfully" [75]. In the 1980's, Albert Bandura examined the effect of self-efficacy on task motivation and performance [6, 76, 77]. In particular, Bandura and Cervone [6] investigated the effects of self-efficacy on goal choice in an experiment that measured the goals set by subjects after they received feedback concerning their success or failure in a target task. Each subject in this experiment was given a goal of increasing the effort exerted in a physical task by 50% relative to the subject's baseline value. Among subjects who were told that they had fallen slightly short of this goal, 25% set themselves a lower goal for the next attempt, and 30% set themselves a higher goal. Among subjects who were told that they had slightly exceeded the goal, 50% set themselves a higher goal for the next attempt, while 35% actually set themselves a lower goal. Bandura and Cervone found that subjects' goals were predicted by self-efficacy. Subjects who were confident in their ability to meet or exceed the goal set high goals for themselves even after failure, but subjects with low self-efficacy lowered their goals even after success. Self-efficacy has also been shown to predict the effort expended by subjects [6, 76] and performance adaptability [78], though it is difficult to separate the effects of self-efficacy from those of past performance [79].

Atkinson proposed his theory of achievement motivation in the 1960's. He claimed that a subject presented with a choice of tasks would choose a task to maximize achievement motivation, defined as $(M_S - M_{AF})P_S(1-P_S)$, where P_S is the probability of success [7]. M_S and M_{AF} are the motive to achieve success and the motive to avoid failure, respectively, two constants that depend on the subject's individual personality. A subject for whom $M_S > M_{AF}$ would choose a task with $P_S \approx 0.5$, a task that is not easy but which he or she can realistically expect to accomplish with effort. If this subject succeeds several times at the chosen task, thus raising P_S for the chosen task as well as for similar but more difficult tasks, the subject will then choose a harder task for which the new value of P_S is approximately 0.5. A subject with $M_{AF} > M_S$, on the other hand, will tend to choose either a very easy or a very difficult task. Because this subject has a high motive to avoid failure, the subject will either choose a task for which success is certain or a task so difficult that he or she will not feel bad about failure. If the subject succeeds at an easy task, the increased value of P_S will make the subject find that task even more attractive. The subject will continue to practice the easy task rather than moving on to more difficult tasks. Atkinson reviews evidence to support these theories.

Elliott and Dweck [80] hypothesized that subjects can pursue two types of goals: performance goals, in which the subject seeks to demonstrate his or her ability, and learning goals, in which the subject seeks to learn a new task. They present evidence that a subject given instructions emphasizing performance goals tended to reject the opportunity to learn new tasks in which he or she might make mistakes. Instead, the subject tended to choose a task with a level of difficulty within his or her perceived limits of ability. Subjects pursuing learning goals, on the other hand, were more willing to choose difficult goals involving the possibility of failure.

Our rehabilitation paradigm was designed for chronic stroke and TBI patients, patients at least one year post-injury. These patients have experienced repeated failures in attempts to perform activities that were easy for them before injury. As a result, they may have developed low self-efficacy or a high motive to avoid failure. They may set low goals for themselves and prefer to practice tasks at an undemanding level of performance. Visual distortion may be a way to encourage such a subject to make progress in rehabilitation without directly confronting his or her reluctance to move beyond an established level of performance. On the other hand, patients may focus on the progress they have made since the injury rather than on their functional deficits and may believe strongly in their ability to improve further. In this case, subjects might be willing to strive for difficult goals to improve their function and might be encouraged by visible progression to pursue ever-higher goals. The focus of this thesis was not an investigation of goal setting and pursuit among chronic brain injury patients, and indeed, variation in these characteristics would be expected among patients. The theories discussed here are simply presented to provide the reasoning behind the visual manipulations that we chose to test in a rehabilitation context.

2.7. Previous Work Concerning Feedback Manipulation in Rehabilitation

Various research groups interested in rehabilitation are beginning to investigate the effects of a variety of types of feedback manipulation. A mirror box is one simple type of visual feedback manipulation that has been used with hemiparetic patients; the reflection of their unaffected limb simulated movements of the impaired limb [81]. Another simple type of feedback manipulation is the use of prisms to shift a patient's visual field. Visual distortion via prism goggles has been used with stroke patients to rehabilitate left hemispatial neglect, a condition in which a lesion on the right side of the brain causes patients to ignore stimuli on their left side. When asked to point straight ahead, neglect patients will actually point towards the right [82]. Patients also ignore the left half of pictures or text; when asked to bisect lines, bisections are shifted to the right [82]. Rossetti et al. [82] addressed left hemispatial neglect by asking 16 patients to make reaching movements while wearing prism goggles that shifted the visual field 10° to the right. After a period of adaptation, the goggles were removed and patients were tested. Patients showed decreased neglect on tests of straight-ahead pointing and line bisection. Further work found that the effects of adaptation could last at least four days, even though the adaptation period consisted of only 50 pointing movements [83]. Patients who wore prism goggles twice a day for 2 weeks maintained improvements for five weeks after the end of the distortion sessions [84]. Furthermore, prism adaptation has been shown to improve wheel-chair driving, reading, and other functional abilities in neglect patients [84].

The use of various types of distortion in the rehabilitation of stroke patients has been investigated by Patton and colleagues [85-87]. Patton and Mussa-Ivaldi [85] used a 2 DOF robot to explore a new method to produce desired movement patterns using force perturbations. In an experiment with healthy subjects, participants learned to move the arm in a straight line in the presence of a force field applied by the robot. Then, when the training force was removed, the after-effect of the adaptation caused the subject to shift in the direction of the target movement. In this case, the distortion is the force perturbing the arm from a straight-line path. Patton and Mussa-Ivaldi argue that this method would allow stroke patients to relearn normal patterns of movement in a way that requires less conscious attention or motivation than traditional rehabilitation. A similar experiment with 18 chronic stroke patients found that patients did have the ability to adapt to the distortion and that after-effects of the force training persisted longer in stroke patients [86]. Furthermore, this study found that only distortion that magnified directional errors (that caused an after-effect in the direction of error reduction) improved stroke patient performance during a single session; force perturbations that minimized directional error (similar to the active-assisted or active-constrained modes of some robotic rehabilitation devices) were ineffective. In a similar manner, Emken and Reinkensmeyer [88] found that transiently inducing larger errors speeded learning of a dynamic environment. Wei et al. [87] investigated the effects of a different type of distortion, visual error augmentation. Each subject's error was visually multiplied or increased by a constant offset while he or she made reaching movements with the robot. They found that visual error augmentation speeded learning, while visual distortion via a constant offset both speeded and increased the magnitude of learning.

These results demonstrate that the topic of this dissertation is of great interest to the rehabilitation research community. This study is one of the first to investigate the effects of visual feedback manipulation during a multi-week rehabilitation paradigm and the first to do so in the context of robotic rehabilitation.

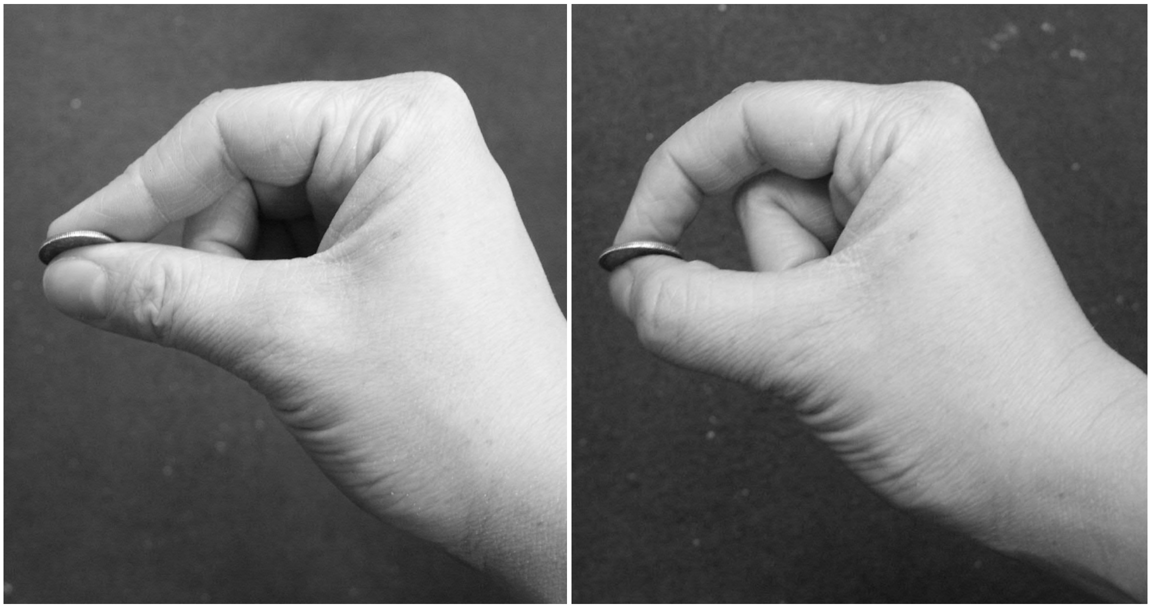


Figure 2.1. The pincer grasp, used for picking up small objects. The goal of our rehabilitation paradigm was to have patients relearn a functional pincer grasp. **(a)** Pad-to-pad pincer grasp. **(b)** Tip-to-tip pincer grasp (neat pincer grasp).

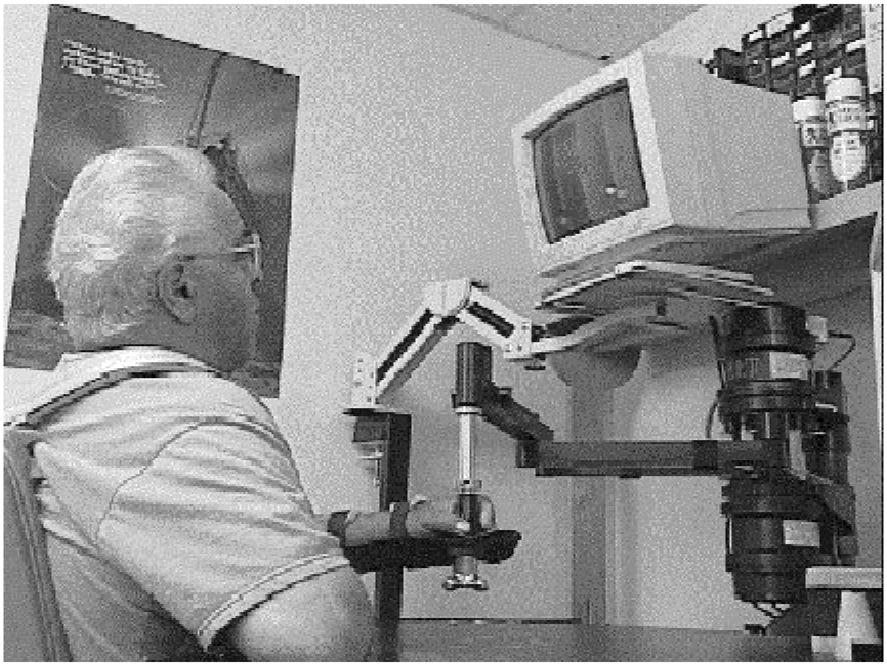


Figure 2.2. The MIT-MANUS.

Figure 2.3. The “Mirror-Image Motion Enabler” (MIME).

Figure 2.4. The ARM Guide.

Figure 2.5. The GENTLE/s system.

Chapter 3. Preliminary Experimental Environment

3.1. Introduction

As described in Chapter 1, our main goal was to determine the limits and effects of visual distortion with unimpaired subjects. In order to do this, we needed an experimental environment in which we could precisely control the visual feedback received by a subject and make accurate and precise measurements of force and position as the subject interacted with the system. This chapter describes the hardware and software details of our experimental environment, as well as the experiments we performed to validate our system.

3.2. Hardware

The experimental environment used in the experiments described in Chapters 4, 5, and 7 consisted of haptic and visual displays, as shown in Figure 3.1. The subject moved his or her index finger against a resisting force while receiving visual feedback on the computer screen. The force feedback was provided by a Premium 1.5 model PHANTOM™ robot (SensAble Technologies, Inc., Woburn, MA). This robot had 3 active degrees of freedom (DOF) and could measure position to within 0.03 mm [89]. The maximum exertable force was 8.5 N, while the largest continuously exertable force was approximately 5 N. This robot was chosen because it was designed to safely interact with human subjects.

We removed the PHANTOM™ stylus provided by SensAble Technologies and designed a custom-made finger cuff shown in Figure 3.2. Our finger cuff adjusted easily to fit a wide range of finger sizes and restrained the finger so that it moved only about the MCP joint. Three pairs of ball bearings gave this finger cuff 3 passive DOF. When combined with the 3 active DOF inherent to the robot, the 6 DOF of our system allowed the finger to move comfortably in any direction within the robot's workspace. The total weight of the finger cuff was 32 g. The finger cuff was tested with subjects ranging in age from 18 to 81, including one motor-impaired individual, and it was light enough to allow every subject to move the finger freely.

In order to isolate the movement of the index finger about the MCP joint and eliminate movements of other fingers or the wrist, we instructed the subject to grasp a post with the remaining fingers and thumb throughout the experiment (Figure 3.1). In addition, the subject's forearm was restrained. We observed that some elderly subjects tended to tilt the hand downward so that movements about the MCP joint did not lie in the horizontal plane. This tilt allowed the path of the index finger to be more easily obstructed by the middle finger. To discourage such tilt, a restraint was placed against the back of the hand (Figure 3.1). A screen concealed the subject's hand throughout each experiment.

3.3. System Software

Our software consisted of a graphics thread that ran at 30 Hz and a haptics thread that ran at 1 kHz. This program was written in Visual C++ and used the GHOST® tool kit provided by SensAble Technologies. This tool kit provided functions to send force commands to the robot and receive position information about the endpoint of the robot from encoders on the motors. The tool kit, however, provided no information about the actual forces generated by the robot; only the commanded forces were known. To calibrate the relative encoders, the tool kit assumed that the robot was initialized at a particular angle, and the software used a Cartesian coordinate frame with the origin located at the endpoint position of the robot during initialization. In this frame, the y-axis was vertical, and the xz-plane was horizontal. All position and force information was given in terms of this Cartesian coordinate frame.

Each subject began a session in our environment by moving back and forth between the two positions marked *O* and *S* in Figure 3.2. Point *O* is the position of the tip of the index finger when the finger was fully flexed, and point $S = (x_S, y_S, z_S)$ is the point at which the subject had extended the finger a Euclidean distance of 65 mm from *O*. The distance of 65 mm was chosen because it was a distance that subjects could comfortably extend the index finger without shifting the hand position. While the subject moved between *S* and *O*, the path of the finger was recorded. The path was an arc between *S* and *O* that lay largely in the *xz*-plane.

In our experimental environment, we created a virtual compression spring and a virtual wall, as shown in Figure 3.2. One end of the virtual spring was fixed at the point marked *O*, and one end was "attached" to the subject's index finger. We chose to make our force simulation a virtual spring because as the subject moved the index finger, the virtual spring provided a continuously varying force that was intuitive for the subject. With the virtual spring, we could examine both force production and range of motion for a subject while varying the relationship between them. We implemented a virtual wall at $x = x_S$ that discouraged the subject from shifting the hand to extend the finger past this vertical plane. The subject could move only between *O* and the virtual wall at $x = x_S$; the values of y_S and z_S were reset each time the subject reached the virtual wall in order to adjust the position of *S* for any small changes in the subject's hand position. The force exerted on the finger was based on the distance that the fingertip moved from *S*. This arc length was approximated by the Euclidean distance of the fingertip from *S*. An average index finger is approximately 100 mm long; for this finger length, the maximum difference between the arc length and the Euclidean approximation is 1.81%. Thus, the Euclidean distance is a good approximation to the arc length. The Euclidean distance between *S* and *O* was the rest length of the virtual spring, and the force experienced by the subject was $F = k\sqrt{(x - x_S)^2 + (y - y_s)^2 + (z - z_S)^2}$, where *k* represents the spring constant and (x, y, z) is the finger position.

In order to apply forces with the robot in a known and controllable direction, we chose to always apply the force tangent to the path of the index finger. Finding this tangent required a model of the finger path. The component of the path that lay in the *xz*-plane (the horizontal plane) and the component that lay in the *y*-direction were considered separately. The horizontal path component was found by approximating *z* as a quadratic function of *x*, $z = ax^2 + bx + c$. The coefficients *a*, *b*, and *c* were computed using the method of least-squares. This least-squares fit was initialized with an array of 333 position data points recorded at the beginning of the experiment. The use of a least-squares fit adjusted the force assignment for each subject's individual finger path, and the least-squares fit was updated throughout the experiment to account for any slight changes that occurred in the finger path. Every time the subject's index finger moved 2 mm in the *xz*-plane, a new data point was added to the position array, and the oldest data point in the array was deleted. Then the least-squares fit was recalculated, yielding a new set of coefficients.

Despite the hand restraint discussed in Section 2.1, the finger's path was not completely confined to the *xz*-plane, and there was marked variability in the *y*-direction between subjects and between trials for a single subject. To model the vertical movement, we assumed that the displacement in the *y*-direction was proportional to the distance moved by the finger in the *xz*-plane. The distance moved by the finger in the *xz*-plane was approximated by the Euclidean distance in the *xz*-plane from *S* to the finger, so y was assumed to be equal to $m\sqrt{(x-x_S)^2+(z-z_S)^2}$ where m is a constant that was found using the point *S* and the position of the finger. This method of approximating the vertical component of the subject's finger path allowed the force assignment to be adjusted for path deviations in the *y*-direction that occurred during a single trial.

Thus, the path $\vec{P}$ of the finger was modeled by

$$\vec{P} = x\hat{i} + m\sqrt{(x-x_S)^2+(z-z_S)^2}\,\hat{j} + \left(ax^2+bx+c\right)\hat{k}\,, \tag{1}$$

where $\hat{i}$ is the unit vector in the *x*-direction, $\hat{j}$ is in the *y*-direction, and $\hat{k}$ is in the *z*-direction. Substituting for z in the *y*-component and taking the derivative, we found that the tangent to the finger path lay along the vector

$$\vec{t} = \hat{i} + \frac{m\left((x-x_S)+(2ax+b)(z-z_S)\right)}{\sqrt{(x-x_S)^2+(z-z_S)^2}}\,\hat{j} + (2ax+b)\hat{k}\,. \tag{2}$$

By normalizing this vector, we found the direction in which the force should be applied. We combined this with the previously calculated force magnitude, and the result was a virtual spring with the force $\vec{F}$ applied tangent to the path of the finger.

3.4. Validation

To test hypotheses in our experimental environment, we needed to know precisely the forces produced by subjects. Rather than add a force sensor to the finger cuff, which would have added unnecessary inertia, we chose to investigate how accurately the commanded force predicted the force actually exerted by the robot on the index finger. We placed a Kistler 9712 quartz Piezotron® load cell against the endpoint of the PHANTOM™ and programmed the robot to execute a sequence of six forces eight times while the load cell recorded the force produced. This procedure was repeated at four arbitrary positions of the robot's endpoint in the *yz*-plane and for forces along the *x*-axis and at 45° to the *x*-axis in the *xz*-plane of the robot. By finding the mean absolute difference between the force measured by the load cell and the nominal robot force, the average absolute error in the force produced by the PHANTOM™ was calculated to be 0.0452 ± 0.0069 N (mean ± standard error). None of our experiments used forces or force differences under 0.4 N. Therefore, we determined that the commanded force of the PHANTOM™ estimated the true force exerted on the finger accurately enough for our experimental purposes.

After verifying the magnitude of the produced force, we assessed the accuracy of the force direction, which was chosen to be tangent to the path of the finger. We assessed the performance of this algorithm using position and force data recorded from 10 young subjects (ages 18-35) and 10 elderly subjects (ages 61-80) selected at random from participants in the JND experiments described in Chapter 4. All subjects had no known neurological or physical problems related to the right arm or hand. A single, arbitrary arc of the finger from S to O was considered for each subject. This arc consisted, on average, of 17 position data points. The y- and z-components of this single arc were fit by polynomials in terms of x. Both polynomials had the same degree n, which was chosen as the smallest degree for which the correlation coefficient between each polynomial and the data it fit was greater than 0.99. The true path tangent for the chosen arc was computed using these polynomial fits, and the angle between the true path tangent and the applied force vector was calculated. The average angular error over the chosen arc was computed for each subject. Two points at each end of the arc were excluded because the derivative of a polynomial fit may not be representative of the true derivative near the edge of a data set.

For the 10 young subjects, the mean angular error between the true path tangent and the applied force vector was $7.85° \pm 1.33°$ (mean ± standard error). The mean angular error for the 10 elderly subjects was 7.96° ± 1.24°. These errors were not significantly different (p = 0.951). The percentage of the applied force that lay along the true path tangent was given by the cosine of the angular error. Combining the young and elderly data, we obtained an overall mean angular error of 7.91°. The cosine of this angle was 0.991, which tells us that only 0.9% of the force was applied in a direction other than the true path tangent. Our algorithm approximated the path tangent extremely well, and it performed equally well for young and elderly subjects.

3.5. Discussion

We constructed an experimental environment consisting of a force-feedback robot and a visual display; this system was accurate enough to provide controllable feedback distortion. Our system was safe because we used a robot designed to interact with humans, and our system tracked movements and exerted forces in three dimensions, giving us an advantage over systems that operate along a single line or plane [65]. Our system also adjusted automatically for movement differences between individuals and for an individual's changes in movement during the experiment. These advantages made our environment appropriate for testing the limits and effects of visual feedback distortion on performance. A preliminary experimental environment is described in [90].

Figure 3.1. The haptic and visual displays that composed our experimental environment. The restraints that isolated the movements of the index finger and the restraint that prevented the hand from tilting are shown. The screen that concealed the hand during experimental trials is not shown.

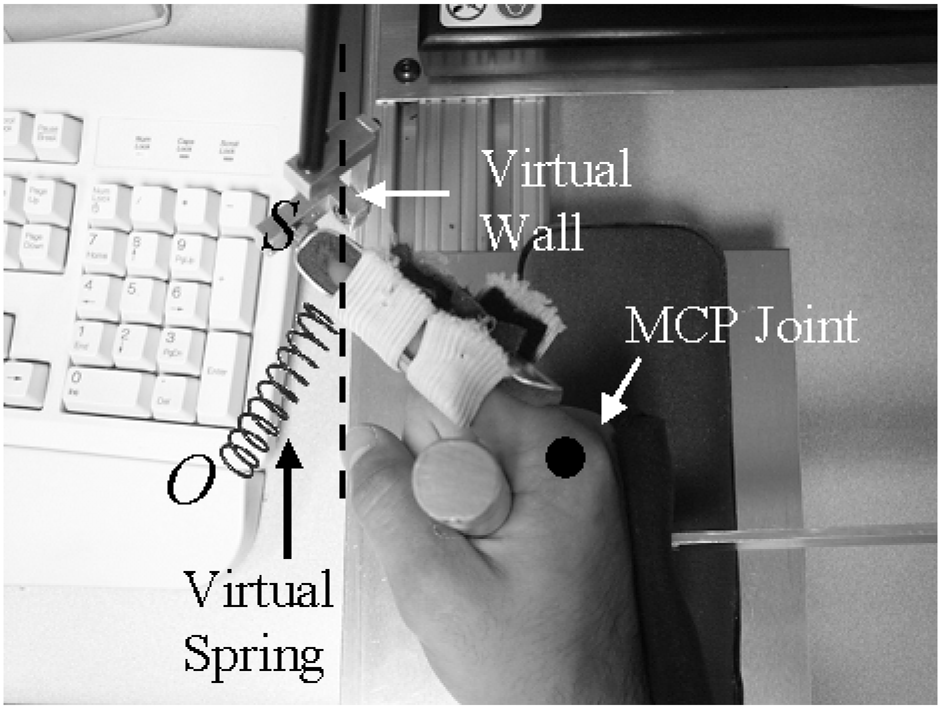


Figure 3.2. Our software simulated a virtual compression spring between *O* and the index finger. A virtual wall was placed at a location 65 mm from *O* (the point *S*). The rest length of the spring was *SO*. The custom-made finger cuff is shown, and the metacarpo-phalangeal (MCP) joint is indicated.

Chapter 4. Perceptual Limits for Distortion

4.1. Introduction

After constructing the experimental environment, our first experiments were designed to determine the amount of visual distortion that would be imperceptible in this environment. For a therapeutic program involving feedback distortion of this sort to be most effective, patients must not detect the visual distortions as they interact with the robot. When a subject believes visual information to be unreliable, he or she begins to rely more on the kinesthetic sense [8], which could reduce the influence of visual distortion on a patient's perception of therapeutic exercise. Because a visual feedback signal on the computer screen is task-specific and not an intrinsic result of performance, disguised progression in this feedback can only be detected if patients (a) map the outcome metric onto intrinsically perceptible results of their actions (e.g., exerted forces or joint angles), and (b) detect changes in the mapping as the performance period progresses. We sought to identify circumstances in which this detection does not occur. The covert nature of the resulting progression means that patients who are physically capable of advancing in rehabilitation will not be prevented from doing so by the perception that they have reached habitual or self-imposed limits. With this in mind and given our target application of hand rehabilitation, our first objective was to measure the Just Noticeable Differences (JNDs) for produced force and movement distance/finger position in young and elderly unimpaired subjects. The JND of a physical dimension is the smallest percent change in the dimension that can be reliably perceived. The force and position JNDs of unimpaired, age-matched subjects give us lower bounds on the amount of distortion that we can expect to be imperceptible. For comparison, we have also measured the JNDs for four motor-impaired individuals.

4.2. Methods

We conducted two experiments to test subjects' perception of force and distance/position. The first experiment measured the Just Noticeable Difference for force, and the second measured the JND for finger position or movement distance. All subjects identified as "young" were between 18 and 35, and all subjects identified as "elderly" were between 61 and 81. All were right-handed with no history of known neurological trauma affecting the right side of the body. No young subject participated in more than one experiment. JND results for four motor-impaired individuals are also reported. P1 and P5 were chronic TBI subjects. P4 and P7 were chronic stroke subjects. For more information on P1, P4, and P7, see Chapters 7, 10, and 11, respectively. P5 was a 19-year-old male who was 2 years post-injury. His injury included diffuse axonal injury with a hemorrhagic component and involvement of the brain stem. His motor abilities were impaired primarily on his left side. He was independent in most daily tasks. He had some impairment in executive function but was cognitively capable of understanding and participating in the experiment. All motor-impaired subjects performed the experiment with an impaired hand (this was the left hand for all except P4). All experiments were approved by the Institutional Review Board of the university, and all subjects gave informed consent.

4.2.1. Force JND Experiment

The force JND experiment was performed to determine the minimum amount of visual distortion of force information that is imperceptible. Twelve young subjects (6 females and 6 males) and 10 elderly subjects (5 females and 5 males) participated in the force JND experiment. Four of the elderly subjects had participated in a previous experiment, but there was no significant difference in JND between the two groups (p = 0.268). All four motor-impaired subjects participated in the force JND experiment. This experiment consisted of 100 trials. A break lasting at least 4 minutes was given every 25 trials. On each trial, the subject sampled two forces. The subject began at the point *S* (see Chapter 3) and flexed the finger against the virtual spring until he or she reached the base force F_0. The subject exerted this force for two seconds and then returned to *S*. Then the subject extended the finger against the virtual spring again until he or she reached a second target force. The second target force was either F_0 or $F_0 + \Delta F$. The subject sampled the second target force for two seconds, then returned to *S* again. After sampling both forces, the subject was asked if the two forces were the same or different and responded by pressing ‘s’ or ‘d’ on the keyboard. After responding, the subject was told the correct answer. The subject then moved on to the next trial. For young subjects F_0 was 1.5 N, and ΔF was 0.3 N. For elderly subjects, F_0 was 2.0 N, and ΔF was 0.7 N. F_0 was 2.0 N for P1 and P5, and their ΔF was 0.8 N. For P4, F_0 was 1.5 N, and ΔF was 0.525 N. P7 experienced two base forces, 1.5 N and 1.25 N, with ΔF values of 1.2 N and 1.45 N, respectively. The reason for this will be explained below. Pang et al. [13] found the JND for force to be a constant fraction of F_0; their analysis, which we follow, also compensated effectively for variations in ΔF. Thus, the different values of F_0 and $\Delta F / F_0$ should not affect the percent JND. ΔF was made larger for elderly subjects and motor-impaired individuals because we anticipated that their JNDs would be larger, and the method of analysis requires that they be able to discriminate F_0 from ΔF at levels well above chance. Each subject was instructed to focus only on the force felt while he or she was in the target window and to ignore other variables such as the position of the finger or the resistance of the robot as the finger moved from *S* to the target force (the spring constant). To gauge whether subjects complied with these instructions, each subject except P1 responded to a post-experiment questionnaire that asked him or her to identify all of the cues used to perform the discrimination task.

A visual display (shown in Fig. 4.1) guided the subject to each target force. The middle box on the display was shaded when the subject was at the target force. When the subject needed to exert more force, the top box was shaded, and when the subject was exerting too much force, the bottom box was shaded. This visual display gave the subject no information about the magnitude of the target force. The visual display was used only to indicate to the subject when he or she was exerting the target amount of force. Each target force was defined by the nominal force plus a window of on either side of this force. For young subjects, the window was 1% of the target force. For elderly subjects and motor-impaired individuals, the window was 1, 2, or 3%, depending upon the results of ten calibration trials that determined the ability of the subject to maintain a constant force. After the subject had stayed within the desired force window for 2 s, the middle box changed color to indicate that the subject should return to *S*.

The spring constant of the virtual spring varied from trial to trial and also between stimuli within a single trial; thus, the distance moved by the finger was not correlated with force. The five possible values for the spring constant were 58.5, 72.0, 85.5, 99.0, and 112.5 N/m.

We computed the Just Noticeable Difference for force using the method described by Berliner and Durlach [91] and Pang et al. [13]. The JND is computed using the proportion p_F of false positives (trials in which the forces were the same but the subject answered 'different') and the proportion p_H of hits (trials in which the forces were different and the subject answered 'different'). The method assumes that for each sampled force, the subject experiences a sensation S_F. For a given force value, the probability density of the random variable S_F is Gaussian with mean μ and variance σ^2. Discrimination between two force values can be described by the sensitivity index d', the difference between the means of the two sensation distributions divided by the standard deviation (assumed to be the same for both force values). The sensitivity index can be found using the equation

$$d' = F_{norm}^{-1}(1 - p_F) - F_{norm}^{-1}(1 - p_H), \tag{4}$$

where F_{norm}^{-1} is the inverse of the cumulative distribution function for a standard normal distribution. The sensitivity index can be normalized by the percent difference between the two forces to obtain $\delta' = d' F_0 / \Delta F$. The JND is defined as the change in force that produces $d' = 1$, which means that the JND as a percent is given by $100/\delta'$. If the subject performs in an unbiased way, the subject will give the correct answer on approximately 69% of the trials if $\Delta F / F_0$ is set equal to the percent JND.

The sensation distributions for F_0 and $F_0 + \Delta F$ overlap. Each subject chooses a particular sensation level *C* as the decision boundary. Forces resulting in sensations larger than *C* are said to be $F_0 + \Delta F$, and forces resulting in sensations smaller than *C* are said to be F_0. The bias β of each subject was computed by finding the ratio of the heights of the distributions at the point *C*. If behavior is unbiased, the point *C* will be where the two normal distributions intersect, and β will be equal to 1. If β is greater than 1, the subject is biased in favor of choosing the response 'same.' If β is less than 1, the subject is biased in favor of choosing 'different.' The bias is given by the formula

$$\beta = e^{\frac{1}{2}\left[\left(F_{norm}^{-1}(1-p_F)\right)^2-\left(F_{norm}^{-1}(1-p_H)\right)^2\right]}. \qquad (5)$$

In addition to computing the JND and the bias for each subject, we were also interested in the effect of the varying spring constant on the subject's discrimination ability. We examined all of the trials in which the spring constant was the same for both forces, and we calculated the percentage of these trials on which the subject responded correctly. We also considered trials in which the spring constants for the two forces were one of the following combinations: 58.5/112.5, 72.0/112.5, and 58.5/99.0 N/m. We computed each subject's percentage accuracy on these trials and compared this to the same spring constant accuracy.

4.2.2. Distance/Position JND Experiment

The distance/position JND experiment was conducted to determine the minimum amount of visual distortion of movement distance or finger position that is imperceptible. Eleven young subjects (2 females, 9 males) and eleven elderly subjects (5 females, 6 males) participated in this experiment. Five of the elderly subjects had participated in a previous experiment, but there was no significant difference between the JNDs of the two groups (p = 0.379). All motor-impaired individuals participated in the experiment. The protocol for this experiment was very similar to the one used in the force JND experiment. On each of 100 trials, the subject sampled two displacements of the finger from *S*. The Euclidean displacement approximated the arc length traveled by the fingertip. Each subject started at *S* and flexed the index finger through a distance of $D_0 = 30\,\text{mm}$. The subject stayed at this displacement for two seconds and then returned to *S*. The subject then moved to a second displacement that was either D_0 (50 trials) or $D_0 + \Delta D$ (50 trials). After sampling the second displacement for two seconds, the subject returned to S and was asked if the two distances were the same or different. After responding, the subject was given the correct answer. ΔD was 6 mm for young subjects and 10.5 mm for elderly subjects and the motor-impaired subjects. The subject was instructed to ignore all cues apart from the distance or position of the finger while he or she was in the target window. As in the force JND experiment, each subject except P1 filled out a post-experiment questionnaire asking which cues he or she had used in the task. The JND and the bias for each subject were computed as described above.

The visual display used in the force JND experiment was used here to direct the subject to the target displacement. Young subjects were required to stay within ± 1% of the target displacement, while elderly subjects and motor-impaired individuals were required to stay within ± 1, 2, or 3% of the target displacement, depending upon the degree of tremor in the subject's movement, which was determined during the calibration procedure.

As in the force JND experiment, the spring constant of the virtual spring varied from trial to trial and between stimuli within a single trial. The five possible values for the spring constant were 32.5, 40.0, 47.5, 55.0, and 62.5 N/m. These spring constants differed from those used in the force JND experiment due to the force limitations of the PHANTOM™ robot. Percentage accuracies were computed for same spring constant trials and different spring constant trials as in 3.1.2, except that the combinations of different spring constants that were considered were 32.5/62.5, 40.0/62.5, and 32.5/55.0 N/m.

4.3. Results

4.3.1. Force JND Experiment

Results for the young and elderly unimpaired subjects can be found in Table 4.1 and Fig. 4.2(a). The mean force JND for young subjects was 19.7% ± 1.85% (0.296 ± 0.0278 N) (mean ± standard error), and the mean force JND for elderly subjects was 31.0% ± 3.99% (0.619 ± 0.0797 N). A t-test showed these two JNDs to be significantly different (p = 0.0137). The mean bias for young subjects was 1.01 ± 0.0467, and the mean bias for elderly subjects was 0.989 ± 0.146. The mean bias was not significantly different from 1 for either group (p = 0.902 for young, 0.943 for elderly). As Table 4.1 shows, the force JND varied widely with individual. Individual JNDs ranged from 12.6% to 31.7% for young subjects and 12.3% to 50.5% for elderly subjects. The results for the motor-impaired subjects were as follows: P1's JND was 46.0% (0.920 N) with a bias of 1.14, P4's JND was 54.8% (0.817 N) with a bias of 1.40, and P5's JND was 24.0% (0.480 N) with a bias of 1.53. We had trouble determining P7's JND. At $F_0 = 1.5\,\mathrm{N}$ and $\Delta F = 1.2\,\mathrm{N}$, he performed no better than chance, i.e. no better than if he had simply guessed 'same' or 'different' without experiencing the forces. At $F_0 = 1.25\,\mathrm{N}$ and $\Delta F = 1.45\,\mathrm{N}$, he answered correctly for every trial except one. These data do not allow us to compute a JND, so all we can say with certainty about P7 is that his JND is somewhere between 80% and 116%.

For elderly subjects, the percentage of correct responses on trials in which the two forces were experienced with the same spring constant was 79.0% ± 4.40%. The percentage of correct responses for trials with the chosen combinations of different spring constants was 64.6% ± 4.22%. These two values were significantly different (p = 0.0413). For young subjects, the percentage accuracy for same spring constant trials was 71.3% ± 2.83%, while the percentage accuracy on different spring constant trials was 68.4% ± 2.26%. These values were not significantly different (p = 0.474), though 8/12 subjects had a higher percentage accuracy for the same spring constant trials.

On the questionnaires, eleven young subjects said that they based their judgments in the discrimination task on the force that they felt on the finger while the middle box on the visual display was shaded. The remaining subject indicated that he used the sense of effort required to stay in the middle box, which six other subjects also used in addition to the force. Six subjects said that they used the resistance of the robot while moving to the target force (the spring constant) in addition to force or effort, and three subjects stated that they used the position of the finger, despite the fact that all subjects were instructed to ignore these cues.

All ten elderly subjects stated on questionnaires that they used the force that they felt on the finger to perform the discrimination task. Seven said that they also used the sense of effort experienced while in the middle box. Four indicated that they used the resistance of the robot, and two stated that they used the position of the finger. P4 and P5 stated that they used only the force felt on the finger, while P7 used both the force and the resistance of the robot.

4.3.2. Distance/Position JND Experiment

One young subject and one elderly subject were excluded from the analysis because their questionnaires revealed that they had performed the discrimination using only force or the resistance of the robot rather than distance or position. The results for all other unimpaired subjects can be found in Table 4.2 and Fig. 4.2(b). In addition, P7 performed no better than chance, so we could not compute a JND for him (though it must have been higher than 10.5 mm).

We refer to this JND as the distance/position JND because the distance moved and the terminal position of the finger were correlated in the experiment. Subjects could have used either quantity to perform the discrimination task. The percentages give the JND expressed as a fraction of the base distance from the spring origin, and the JND in terms of absolute position difference is given in millimeters. For young subjects, the mean JND for distance/position was found to be 13.0% ± 1.38% (3.99 ± 0.434 mm), and the mean bias was 1.10 ± 0.0601. The mean JND for elderly subjects was 20.7% ± 4.88% (6.32 ± 1.38 mm), and the mean bias was 1.01 ± 0.0511. Neither mean bias was significantly different from one ($p = 0.0912$ for young, 0.902 for elderly). The mean distance/position JND for elderly subjects was not significantly different at the 5% level from the mean JND for young subjects ($p = 0.125$), though there was a trend for the elderly JND to be larger. Individual JNDs for young subjects ranged from 2.27 to 6.08 mm, and individual JNDs for elderly subjects ranged from 3.65 to 18.3 mm. The results for the motor-impaired individuals were as follows: P1's JND was 45.0% (14.8 mm) with a bias of 1.40, P4's JND was 36.4% (13.6 mm) with a bias of 1.55, and P5's JND was 35.6% (11.2 mm) with a bias of 0.991.

The mean percentage accuracy for young subjects on the same spring constant trials was 85.0% ± 3.06%, and the mean percentage accuracy on the different spring constant trials was 72.3% ± 3.85%. These values were significantly different (p = 0.0068). For elderly subjects, the mean percentage accuracy on the same spring constant trials was 89.5% ± 3.02% and the mean percentage accuracy for the different spring constant trials was 73.3% ± 4.13%. These values were also significantly different (p = 0.0013).

The ten young subjects included in the analysis said on the questionnaires that they used the terminal position of the finger to perform the discrimination task. Four stated that they also used the force felt on the finger, and two used the resistance of the robot when moving to the target position. Three subjects utilized the sense of effort required to stay in the middle box.

The ten elderly subjects included in the analysis stated that they used the terminal position of the finger, the distance moved by the finger, or some related variable (e.g., the time required to reach the middle box when moving at a constant velocity). Two also used the sense of effort required to stay in the middle box, three used the force on the finger, and two used the resistance of the robot. P5 stated that he used only the terminal position of the finger, but P4 used the terminal position, the sense of effort required to stay in the middle box, and the resistance of the robot.

4.4. Discussion

4.4.1. Force JND Experiment

The JND for force measured for young subjects in our rehabilitation environment is more than twice as large as the JND of 7-10% measured for this dimension by other researchers (see Chapter 2). There are several reasons that may account for this. First, our environment and the joint tested are unique, which may affect the value of the force JND. Also, our subjects received substantially less training in the discrimination task than those of [13] and [15]. Most importantly, all of the JNDs cited above were measured with all physical dimensions other than the target dimension fixed. These other physical dimension are called background dimensions [92], and the most common way to measure kinesthetic JNDs is to keep the background dimensions constant. Our force JND experiment, however, had two unfixed background dimensions that are related: distance and the stiffness of the virtual spring.

Previous studies have shown that varying background dimensions increases the JND. Tan et al. [93] conducted a series of JND experiments using a four-dimensional tactile display to determine the effects of varying background dimensions. They measured the JND for each of four types of displacement with the other three dimensions both fixed and varying. They found that the JND increased by an average of 260% when the background dimensions were varied. Tan et al. [94] also measured the change in the finger span force JND when the distance moved by the fingers varied. They found that the force JND changed from 6% to 14% for the same set of subjects, an increase of 133% in the JND. In this experiment, only one background dimension was varied, because the force was constant over the displacement; there was no spring constant. Given these studies and that we have two varying background dimensions, we would expect our measurements for the force JND to be approximately twice the value of 7-10% found by previous researchers. We would expect our value for the force JND of young subjects to fall in the range 14-20%, as its value of 19.72% does. The mean accuracy of young subjects was not greater on trials in which the two forces were experienced with the same spring constant. However, 8 out of 12 subjects did better on trials in which the spring constant stayed the same, and accuracy with the same spring constant was significantly greater than accuracy with different spring constants for elderly subjects in the force JND experiments and both young and elderly subjects in the distance/position JND experiment. This supports our hypothesis that the force JND was increased by varying the background dimensions. In the context of our virtual spring simulation, we varied the background dimensions in order to separate force and position cues. In general, we chose to vary the background dimensions in our experiments to increase the amount of visual distortion that is imperceptible, since maintaining that imperceptibility is essential for the use of distortion in therapy.

The mean force JND of 29.82% that we found for elderly subjects was significantly greater than that of young subjects. To our knowledge, we are the first to measure the force JND for elderly subjects. The increase in JND with age is similar to that seen by previous researchers for other dimensions, as reviewed in Chapter 2. The JND we measured for elderly subjects may be larger than that measured for young subjects due to reduced afferent input with age [23] or to effects of aging on the central nervous system, which could include synaptic changes and a decline in the number of nerve fibers [19].

4.4.2. Distance/Position JND Experiment

As mentioned in 3.2.2, finger displacement and position were correlated in the distance/position JND experiment; however, this correlation was not perfect, due to small finger path changes between the two parts of each trial. Because we cannot determine from our experiment which of these cues subjects used to perform the task, we cannot conclusively state whether this experiment measured the JND of displacement or position, and we analyzed the data both in terms of the distance from the origin of the spring (percentages) and in terms of absolute position change (millimeters). Jaric et al. [95] found that subjects could reproduce endpoint position more accurately than movement distance; thus, it is likely that our subjects performed the discrimination task using finger position, not displacement. Because endpoint position was not perfectly correlated with displacement in our experiment, we used nine new young subjects to perform an experiment that controlled the terminal finger position directly. The position JND that we found was not significantly different from that derived in our original experiment.

For convenience, we have expressed both distance and position in terms of the robot coordinate frame. Subjects most likely actually base their judgments on some internal measure of the terminal MCP joint angle. If we assume a 100 mm index finger, our 3.99 mm position JND is equivalent to a JND for MCP joint angle of approximately 2.3°.

At 3.99 mm, the distance/position JND agrees roughly with the JNDs of 1-3 mm measured by previous researchers (see Chapter 2). These previous measurements were for the JND of the distance between the index finger and thumb, but because the discrimination of this distance is based on the position of the index finger and thumb, these results are comparable to the distance/position JND described here. The distance/position JND as described here has not been previously measured by other researchers. Our distance/position JND is slightly larger than previous results, probably due to variations in the background dimensions of force and spring constant, as mentioned above, but the distance/position JND seems to be less affected by the varying background dimensions than the force JND. This contradicts experimental evidence stating that the perception of force is less influenced by distracting position information than position perception is by force distracters [96]. The force JND may be affected more by the varying background dimensions because the difference between the largest spring constant and the smallest spring constant was 54.0 N/m in the force JND experiment but only 30.0 N/m in the distance/position JND experiment, even though the percent change in the spring constant was the same for both experiments. This means, effectively, that the position varied more in the force JND experiment than the force did in the distance/position JND experiment.

We observed a greater difference between the young and elderly force JNDs than between the young and elderly distance/position JNDs. It may be that elderly subjects are more affected by varying background dimensions than young subjects, and the elderly subject JND is more different from the young subject JND in the force JND experiment because the background dimensions varied more in that experiment.

4.4.3. JNDs for Motor-Impaired Individuals

While we cannot draw strong conclusions from only a few motor-impaired subjects, it seems that we can expect the JNDs for stroke and TBI patients to be much larger than those of the appropriate control group. This is similar to the decline in tactile sensibility and position sense noted in stroke patients by other researchers [24]. Based on age, P1 and P5 should be compared to the young subject control group, while P4 and P7 should be compared to the elderly control group. P5's force JND was comparable to that of the young unimpaired subjects, but P1's was much larger. In addition, the distance/position JNDs for P1 and P5 were well outside the range measured for young unimpaired subjects. P4's force JND was larger than the largest JND we measured for unimpaired elderly subjects, and P7's force JND, though its precise value is unknown, was also much larger than the mean force JND for unimpaired elderly controls. P4's distance/position JND was well outside the range of most unimpaired elderly subjects, though there was an outlying control subject with a greater distance/position JND. Given that P5 was the only individual with a JND in the range of most unimpaired subjects, it is worth noting that P5's motor impairment was fairly mild. He had full range of motion with his hand and was able to use it in most daily activities. All other subjects had limited ranges of motion. However, P7 had better functional use of his hand than P4 (see functional test results, Chapters 10 and 11) but had a larger force JND. Though the JNDs were not necessarily correlated with the level of impairment, the tendency of the JNDs of impaired individuals to be larger than those of the controls is encouraging from the point of view of our proposed rehabilitation paradigm, because larger JNDs mean that larger amounts of visual distortion will be imperceptible.

4.4.4. Conclusions

Our reason for measuring the JNDs for force and distance/position was to discover the lower bound on the amount of distortion that is undetectable by a subject, so that we could use imperceptible visual distortion to encourage patients to push harder or move farther during rehabilitation. A distortion between the visual display and the actual value of the force or distance should be imperceptible if it is below the JND for the appropriate dimension. For instance, if we vary the background dimensions of distance and stiffness, we should be able to distort the visual display of force by at least the force JND without subjects detecting the distortion. One might expect that the JND for a given dimension would become smaller as a subject neared his or her maximum level of exertion, which would complicate the use of distortion in rehabilitation. However, Jones [16] showed that the force JND did not decrease, even when subjects were exerting forces that were 85% of their maximum voluntary contraction.

Finally, it should be noted that the above discussion assumes that we vary the background dimensions, changing position and stiffness when force is the target dimension and varying force and stiffness when position is the target dimension. If we choose not to vary the spring constant, so that force and position are correlated, subjects will be able to combine these two sources of information to possibly increase their ability to detect any visual distortion [9].

The force and distance/position JNDs indicate the minimum amount of distortion that should be undetectable by a subject, but because vision is the dominant human sense in most situations [97], we expect that visual distortions well above the limit of the JND may be imperceptible. The next chapter demonstrates this point by describing how we used our measured JNDs to design an experiment to measure the effect of distortion on force production and movement distance. Preliminary results for the experiments described in this chapter can be found in [98, 99], and this chapter appears in [100].

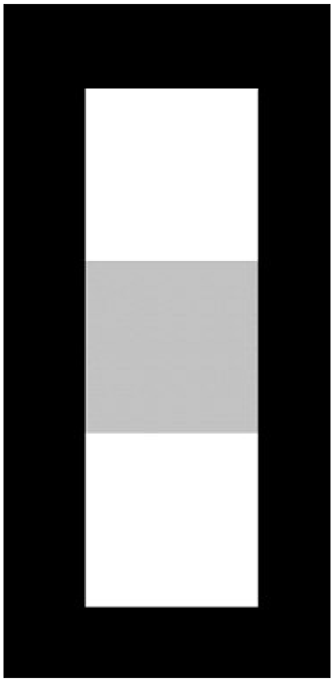


Figure 4.1. The visual display used for the force and distance/position JND experiments. The middle box was shaded when the subject was at the target value for force or distance, respectively.

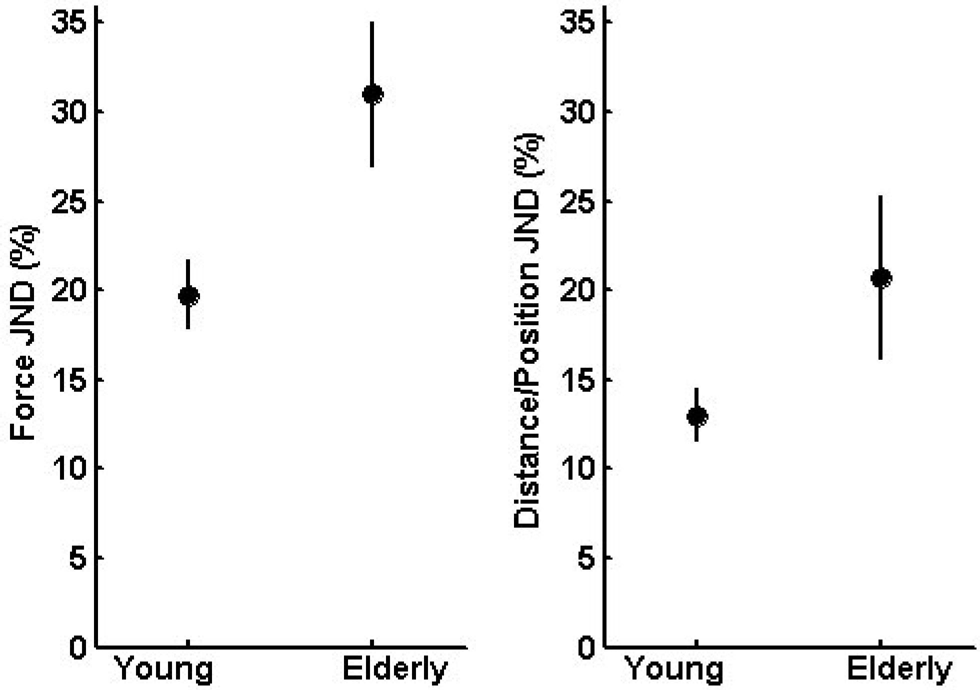


Figure 4.2. (a) Results of the force JND experiment for young and elderly subjects. Mean and standard error for each group are shown. The mean JND for elderly subjects was significantly larger than that of young subjects. **(b)** Results of the distance/position JND experiment for young and elderly subjects. The mean JND for young subjects was not significantly different from that of elderly subjects.

Force JND Experiment

Young Subjects		Elderly Subjects	
JND	Bias	JND	Bias
15.1% (0.227 N)	0.905	12.3% (0.246 N)	0.164
16.9% (0.254 N)	0.869	38.7% (0.773 N)	0.658
30.3% (0.455 N)	0.948	28.2% (0.564 N)	1.04
17.0% (0.256 N)	0.933	33.4% (0.667 N)	0.941
16.3% (0.244 N)	0.964	31.5% (0.630 N)	1.10
22.7% (0.340 N)	0.929	50.5% (1.01 N)	1.41
14.9% (0.224 N)	1.40	42.4% (0.848 N)	1.35
12.6% (0.189 N)	1.06	15.0% (0.300 N)	1.77
19.3% (0.290 N)	0.766	18.0% (0.360 N)	0.541
25.7% (0.385 N)	1.06	39.7% (0.792 N)	0.929
31.7% (0.475 N)	1.14		
14.1% (0.212 N)	1.10		

Mean Young Force JND: 19.7% ± 1.85% (0.296 ± 0.0278 N)

Mean Young Bias: 1.01 ± 0.0467

Mean Elderly Force JND: 31.0% ± 3.99% (0.619 ± 0.0797 N)

Mean Elderly Bias: 0.989 ± 0.146

Table 4.1.

Distance/Position JND Experiment

Young Subjects		Elderly Subjects	
JND	Bias	JND	Bias
20.3% (6.08 mm)	1.32	19.5% (5.85 mm)	1.40
11.7% (3.86 mm)	1.28	19.4% (6.14 mm)	0.937
10.9% (3.40 mm)	0.869	17.0% (5.24 mm)	0.442
16.3% (4.80 mm)	1.12	11.7% (3.70 mm)	1.25
18.2% (6.21 mm)	1.24	12.4% (3.65 mm)	6.11
9.61% (2.89 mm)	1.31	23.0% (7.27 mm)	0.700
7.76% (2.45 mm)	1.03	12.4% (3.95 mm)	6.12
7.29% (2.27 mm)	0.847	14.6% (4.46 mm)	0.567
13.5% (4.17 mm)	0.863	14.0% (4.63 mm)	1.14
14.7% (3.73 mm)	1.14	63.4% (18.3 mm)	0.817

Mean Young Distance/Position JND: 13.0% ± 1.38% (3.99 ± 0.434 mm)

Mean Young Bias: 1.10 ± 0.0601

Mean Elderly Distance/Position JND: 16.1% ± 1.18% (5.01 ± 0.370 mm)

Mean Elderly Bias: 2.01 ± 0.692

Table 4.2.

Chapter 5: Distortion to Manipulate Force Production and Movement Distance

5.1. Introduction

In the previous chapter, we described how we measured JNDs to discover the minimum amount of distortion that should be imperceptible. In this chapter, we discuss how we used that information to investigate the effects of visual feedback distortion in our robotic environment. For distortion to affect performance during rehabilitation, the patient must rely primarily on visual feedback when performing tasks in our environment. Previous research has demonstrated that when vision and kinesthesis concerning a movement are in conflict, the subject's perception of the movement tends to be dominated by the more precise feedback source, which is often, but not always, vision [9]. The experiments described here were conducted to show that vision dominates kinesthetic feedback in our robotic rehabilitation environment for a simple, one-finger task and that gradual visual distortion beyond one JND can be used to manipulate force production and movement distance with a single experimental session. We performed experiments with young and elderly subjects and four motor-impaired individuals to investigate how the effects of visual distortion depend upon age and impairment level.

We designed a task in which subjects set various force or distance goals while the visual feedback corresponding to those goals was gradually distorted. We predicted that a subject would use the visual feedback to set a goal; then, as the visual feedback was distorted, the subject would gradually change the force or distance response in order to maintain approximately constant visual feedback. This in turn would produce new somatosensory feedback and hence cause the subject to recalibrate the association between somatosensory experience and the specified force or distance level. We included trials with no visual feedback to test whether or not the visual distortion would influence the somatosensory representation of the goal even on trials where vision was not specifically present.

5.2. Methods

Subjects identified as young were 18-35 years old, and those identified as elderly were 61-81. All unimpaired subjects had no history of known neurological trauma affecting the right side of the body. All unimpaired subjects performed the experiment with the right hand, which was dominant. Four motor-impaired individuals, P1, P4, P5, and P7, also participated. For more information on these individuals, see Chapters 7, 10, 3, and 11, respectively. All motor-impaired subjects performed the experiment with an impaired hand (this was the left hand for all except P4). Each subject produced forces and distances within his or her limits of ability.

Table 1 contains the name and a brief description of the four experiments and four control experiments conducted. In three experiments, gradual distortion was implemented to encourage force increases (FI) without the subject's knowledge with young unimpaired subjects (FI-Y), elderly unimpaired subjects (FI-E), and the motor-impaired subjects (FI-MI). To illustrate the similar results when movement distance representation was distorted, a young unimpaired subject group with distortion of movement distance was included (DI-Y).

Four control experiments were conducted with young unimpaired subjects: a control with no distortion (CND), a control with no visual feedback (CNF), a control with subjects who were informed of potential distortion between the actual somatosensory feedback and visual representation of force (CI), and a control with distortion designed to encourage force decreases (CD).

All experiments were conducted using the experimental environment described in Chapter 3. In each of the force experiments, the subject was asked on each trial to produce a particular level of force from 1 (smallest) to 5 (largest). The subject flexed the index finger against the virtual spring until the desired force was reached, then pressed the space bar and extended the finger. Subjects were instructed to be as consistent as possible in the force that they produced for each level, though each subject independently chose the magnitude of force corresponding to each level. Before the trials of the experiment began, the subject was allowed to explore the range of force and decide on the magnitude of force that he or she felt should correspond to each level.

The visual feedback bar that was provided to the subject is shown in Figure 5.1. The number of shaded boxes on the feedback bar indicated the magnitude of the force exerted by the subject. The direction of increasing force was vertically downward because this provided a more intuitive mapping between the visual display and the actual movements of the subject's finger as the force was produced. We chose to use a visual feedback bar composed of discrete blocks to draw attention from vision to the force experienced via the kinesthetic sense. Since we cannot expect patients to ignore forces that are difficult for them to produce, we felt that this would make our experiment a better model for a rehabilitation situation.

The visual feedback bar was gradually distorted throughout the experiment by altering the range of force represented by the bar. This distortion was based on the JND for force. The JNDs for the various subject groups (Table 2) were determined in our previous experiments [101]. Each distortion step was accomplished by increasing or decreasing the force represented at each end of the bar by a fraction of the JND multiplied by the original value of the corresponding endpoint. For example, for P1 in FI-MI, the first distortion step changed the range of forces represented by the bar from 0.50 – 2.73 N to 0.62 – 3.36 N; each endpoint was increased by 23.0% (i.e. one half of the force JND of 46.0%). The second distortion step changed the range of force represented to 0.73 – 3.99 N, an increase in each endpoint of one JND relative to the original value. Table 2 describes the number of distortion steps and the force range represented on the initial feedback bar for each experiment. In the highest level of distortion reached in each experiment, the endpoints of the feedback bar were two force JNDs larger or smaller than the initial endpoints of the bar. It is important to note that the appearance of the visual feedback bar did not change; only the range of forces represented on the bar changed. During the exploration of the force range at the beginning of the experiment, the initial, undistorted feedback bar was shown.

Since each distortion step was less than the JND for force, we anticipated that the distortion would be imperceptible, even though the total amount of distortion experienced in one session was larger than the JND. A questionnaire was administered to each subject (except P5) at the end of the experiment to assess whether he or she detected the distortion.

All young subjects experienced a total of 140 trials. Elderly and motor-impaired subjects experienced 100 trials. Each subject had twenty trials for each level of distortion (including 0% distortion). In each block of twenty trials, the subject produced each level of force four times, twice with visual feedback and twice without. The no-feedback trials were included to determine whether the subject's somatosensory representation of the force goal changed over time. On no-feedback trials, the subject did not see the visual feedback bar. The screen showed only the level of force to be produced. Each force level/feedback condition was experienced once with a spring constant of 90.0 N/m and once with a spring constant of 120.0 N/m. The spring constant was varied to increase the amount of distortion that would be imperceptible to the subject, as discussed in the previous chapter. The order of the trials within each block of twenty was randomized for each subject. We measured the effects of the visual distortion on the forces that the subject produced for each force level.

In the three experiments we conducted first (FI-Y, CND, CF), a break was given after 80 trials to prevent fatigue of the subject and overheating of the robot. After the break, the subject was again allowed to explore the range of force, this time while viewing a feedback bar corresponding to the last distortion level experienced before the break (fourth distortion level). Then, ten of the trials for the fourth distortion level were repeated before the trials for the fifth distortion level began. We corrected for the effect of the break by finding the difference between the mean produced force for each force level/feedback condition for the fourth distortion level before the break and the mean produced force for the corresponding force level during the ten repeated trials. We then added this difference to the forces produced during trials of the appropriate force level/feedback condition that occurred after the break. The corrected data were used for all analyses. No break was given in FI-E and FI-MI because these experiments had only 100 trials. No break was given in CI, CD, and DI because earlier experiments revealed that subjects could complete all 140 trials of the experiment without fatigue.

In DI, subjects were asked to flex the index finger through different levels of distance while pressing against the virtual spring. The design of this experiment was similar to that described above, but the spring constants used were 40.0 N/m and 60.0 N/m. Smaller spring constants were used to preclude the possibility that relatively large forces would discourage the subject from increasing the distance moved for each distance level as the feedback bar changed.

5.3. Statistical Analysis

To determine the effects of distortion on force production, we compared the results of FI-Y and CD with CND and CF. The data from each experiment were analyzed as follows: for each force level/feedback condition, the data were divided into seven successive blocks of twenty trials. These blocks corresponded to the seven distortion levels for FI-Y and CD. Linear regression was performed for each subject to find the slope of produced force as a function of block number for each force level/feedback condition. For each force level/feedback condition, we used a t-test to determine whether the mean slope over subjects was significantly different from zero. Because this resulted in ten tests for each experiment (five for CF), an adjustment was made in each set to minimize the false discovery rate [102]. Those linear trends reported as significant were significant at the 5% level after this adjustment was made.

For FI-Y and CD, we also tested the mean slope of produced force versus block number against the slope that would be predicted if the subject relied only on vision (total visual dominance). If f_0 is the mean force produced for a given force level in the with-feedback condition with 0% distortion, the slope predicted by visual dominance for that force level can be expressed as $\frac{(\text{total percent distortion})f_0}{\text{number of blocks - 1}}$, the change in the force represented at the initial visual location of f_0 divided by the change in the block number. Testing the mean slope against that predicted by visual dominance resulted in another set of ten tests for each experiment, and as before, an adjustment for multiple tests was made.

Linear trend analysis was used to test for a change in the slope with force level for each feedback condition. A result was taken to be significant if p was less than 0.05. To compare the different experiments, we focused on the slopes of the average regression lines for force level 5. T-tests were used to test for differences in slope between different experiments for a given feedback condition. We chose to use level 5 since the distortion was implemented as a percentage, which means the effects of the distortion should be greatest for force level 5. T-tests were also used to test for differences in slope between the with-feedback condition and the no-feedback condition for force level 5 in each experiment. The results of the tests between experiments and between feedback conditions were called significant if p was less than 0.05.

Next, the results of FI-Y, CI, FI-E, and FI-MI were compared to investigate how instruction, age, and impairment affected the use of distortion to increase force production. The analysis for these experiments was identical to that described above, except that produced force was considered as a function of the percent distortion rather than block number in order to allow us to compare experiments where the amounts of distortion differed.

Finally, the data from DI were analyzed to assess to the effect of distortion on movement distance. The distance data for each distance level/feedback condition were divided into seven blocks, one for each distortion level that was used, and analyzed as described above.

5.4. Results

5.4.1. Effects of Distortion on Force Production

Figure 5.2 shows the results of FI-Y (distortion to increase force production). The visual distortion of the feedback bar caused a significant increase in the forces produced for force levels 3-5 in both the with-feedback and the no-feedback conditions (mean slope of produced force versus block number significantly greater than zero). For force levels 1 and 2, only force level 1 in the no-feedback condition had a significant upward trend. The mean slope was not significantly different from that predicted by visual dominance for any force level/feedback condition. In addition, the slope of produced force versus block number was linearly related to the force level for both feedback conditions ($p < 0.001$ for both feedback conditions). The maximum increases in produced force occurred for force level 5. The produced force increased by 30.0% in the with-feedback condition and 33.0% in the no-feedback condition. When the with-feedback and the no-feedback conditions were compared for force level 5, the slopes for produced force as a function of block number were not significantly different ($p = 0.288$).

In CD (distortion to decrease force production), significant downward trends in produced force with block number were observed for every force level in the with-feedback condition and for force levels 2-4 in the no-feedback condition (Figure 5.3). The mean slope of produced force versus block number was significantly different from that predicted by visual dominance only for force level 1 in the no-feedback condition. The slope of the mean regression line was linearly related to the force level in the no-feedback condition (p = 0.0027), but not in the with-feedback condition (p = .129). However, the slope of the mean regression line for force level 5 in the with-feedback condition was not significantly different from that of the no-feedback condition (p = 0.768).

The slope of produced force as a function of block number was not significantly different from zero for any force level/feedback condition in the CND control (no distortion, Figure 5.4). Because of this, we did not perform trend analysis for the slope as a function of force level. The mean slope for force level 5 for the with-feedback condition was not significantly different from that of the no-feedback condition (p = 0.768). T-test comparisons between CND and FI-Y revealed that the average slope of produced force with block number was significantly different for these two experiments for both feedback conditions for force level 5 (p = 0.0041 with-feedback, 0.0196 no-feedback). When CD was compared to CND for force level 5, the results were also significant for both the with-feedback (p = 0.0177) and the no-feedback (p = 0.0334) conditions.

There was a significant upward trend in produced force with block number only for force level 4 in CF (no visual feedback), though the force appeared visually to increase with block number (Figure 5.5). The slope of produced force with block number was not linearly resulted to the force level (p = 0.0640). The mean slope for force level 5 for CF was compared to the no-feedback data of both CND and FI-Y. It was not statistically different from the mean slope of either data set (p = 0.241 and 0.645, respectively). Comparisons of the force level 5 data for FI-Y, CD, CND, and CF are found in Figure 5.6.

5.4.2. Influence of Instruction, Age, and Impairment on the Effects of Distortion

In CI (distortion to increase force production; young subjects instructed about possible distortion), there was a significant upward trend in force as a function of distortion for force levels 3-5 in both feedback conditions and for force level 1 in the no-feedback condition (Figure 5.7). There was no significant difference from the slope predicted by visual dominance for any force level/feedback condition. The slope of force versus distortion was linearly related to the force level in both feedback conditions (p = 0.0027 with-feedback, 0.0113 no-feedback), and there was no significant difference between the slope of the regression line for force level 5 in the no-feedback condition and the with-feedback condition ($p = 0.378$). Distortion caused an increase of 36.2% in the with-feedback condition and 26.4% in the no-feedback condition. For force level 5 in each feedback condition, a t-test was performed between the mean slope of force as a function of distortion for CI and the mean slope for FI-Y (Figure 5.9). Neither t-test was significant ($p = 0.418$ with-feedback, 0.273 no-feedback).

The slope of produced force with distortion was significantly greater than zero for all force levels for FI-E (distortion to increase force production, elderly subjects) in the with-feedback condition (Figure 5.8). For force level 5, distortion increased the produced force by 72.5%. Mean slopes significantly greater than zero were seen only for force levels 1, 2, and 5 in the no-feedback condition. The mean slope was not different from that predicted by visual dominance for any force level/feedback condition. The slope of force versus distortion was linearly related to the force level for elderly subjects in the with-feedback condition ($p < 0.001$) but not in the no-feedback condition ($p = 0.156$). For force level 5, the difference in the slope between the with-feedback condition and the no-feedback condition was close to significance ($p = 0.0668$). When a t-test was used to compare the mean slope of produced force with percent distortion in the with-feedback condition to that of FI-Y, no significant difference was found ($p = 0.707$, Figure 5.9). However, in the no-feedback condition, the result was close to significance ($p = 0.0557$).

Figure 5.10 shows the forces produced for level 5 as a function of distortion for each motor-impaired individual. Qualitatively, we can see an upward trend with distortion in the with-feedback forces produced by each subject. For example, visual distortion of 92.0% increased P1's produced force for force level 5 by 97.7% in the with-feedback condition. P4 and P7 also appear to have an upward trend in no-feedback force as a function of distortion, while such a trend seems to be lacking for P1 and P5.

5.4.3. DI: Distortion to Increase Movement Distance

A significantly positive slope for movement distance as a function of percent distortion was found for every distance level/feedback condition except distance level 1 in the with-feedback condition. No slope was significantly different from that predicted by visual dominance. The slope was proportional to distance level for both feedback conditions ($p < 0.001$ with feedback, $p = 0.0040$ no feedback). No difference between the feedback conditions was found in the slope of movement distance versus distortion for distance level 5 ($p = 0.768$) (Figure 5.11).

5.4.4. Questionnaires to Assess Detection of Distortion

Questionnaires revealed that while eight of the FI subjects reported that "the same feeling of force occurred at different positions on the visual display on different trials," six of these subjects attributed the apparent change to fatigue, the changing spring constant ("resistance to motion") of the robot, or to other causes. Similarly, eight subjects in CD stated that the mapping between the visual display and force appeared to change during the experiment, but only two suspected that we were manipulating the display. Eight subjects in CND also stated that the display mapping varied, and two of these subjects suspected deliberate distortion. For DI, seven subjects stated that the visual feedback bar appeared to change, but only one suspected distortion. In CI, six subjects stated that the force represented on the visual feedback bar changed during the experiment. However, only one subject correctly stated that reaching the end of the bar became progressively more difficult throughout the experiment. One subject stated that reaching the end of the bar required progressively less force, and four stated that the force required was sometimes greater and sometimes less. Subjects in CI felt that they were consistent in the force they produced for each level.

Five elderly subjects answered in the affirmative when asked whether the forces represented on the bar changed during the experiment, but only one correctly identified the direction of the distortion. Most subjects stated that they were consistent in the force that they produced for each level. P1 also did not notice the distortion and stated that she was very consistent in the force that she produced for each level. Both P4 and P7 stated that it got harder to reach the bottom of the bar throughout the experiment. P4 said she was not very consistent in the force that she produced, while P7 felt he was fairly consistent.

5.5. Discussion

5.5.1. Dominance of Visual Feedback Occurs in Our Robotic Environment for Different Ages and Different Levels of Motor Ability

The visual-feedback data from our experiments indicate that when asked to produce a consistent force or distance, subjects set a visual goal for each level and respond in such a way as to keep the visual goal constant, even when this requires them to adjust the relationship between the somatosensory feedback and a given force or movement distance level. In this sense, visual feedback dominated cutaneous/kinesthetic feedback in our robotic environment. When the visual feedback was gradually distorted so that the subject had to exert more force to reach the same point on the visual feedback bar, we observed significant upward trends in the produced force for young subjects, elderly subjects, and the motor-impaired individuals, even though these subjects were attempting to be consistent in the forces that they produced. The upward slope of force production with distance for each subject group and each force level was not significantly different from the slope that would be expected if the subjects' had relied only on vision. We were also able to use distortion to decrease the force produced and to increase the distance moved about the MCP joint. We were even able to use distortion to increase force production after informing subjects of the possibility of distortion.

In addition, our results also show that distortion can provide a well-controlled way to encourage specific increases in movement distance or force production. Distortion in these experiments was implemented by increasing the force represented at each endpoint of the feedback bar by a percentage. This means that the position of a larger force on the visual feedback bar changed more than the position of a smaller force. This is shown in Figure 5.2 by the lines showing the predictions of visual dominance. The slope of produced force versus block number (distortion) was predicted to increase as a function of force level. We observed this trend in the slopes we measured for produced force versus distortion in the FI experiments and the slope of movement distance versus distortion in DI-Y. Larger absolute amounts of distortion led to larger absolute changes in force production or movement distance. The percent change in force or distance with distortion was approximately constant across force levels, as indicated by the fact that the force and distance responses did not differ significantly from those predicted by total visual dominance.

We used a discretely segmented visual feedback bar (Figure 5.1) in an attempt to draw attention away from vision to the kinesthetic sense, but the discrete visual feedback bar used in these experiments did have, in general, a finer resolution than the kinesthetic sense of the hand. The resolution of the kinesthetic sense is given by the force JND, which has been measured to be a constant percentage of the reference force under consideration; we measured the force JND to be 19.7% for young subjects in our environment. As an example, halfway through FI-Y and CI, the range of forces shown on the visual feedback bar was 0.572-3.99 N. The resolution of the 25-unit discrete display was 0.137 N, and the percent resolution of the display varied from 3.43% at 3.99 N to 23.9% at 0.572 N. Thus, for most of the force range shown on the visual feedback bar, the resolution of the visual display was finer than the force JND for young subjects (and the JND may be larger at the low end of the force range). The dominance of vision in the experiments described here can be attributed to the finer resolution of the visual display compared to the kinesthetic sense, which is consistent with the work of Ernst and Bank [9] and other work concerning sensory integration (see Chapter 2). Ernst and Bank's theory predicts that vision would not be completely dominant; it states that the haptic input should contribute to the overall perception of force or distance according to the inverse of its variance. However, due to the high variability of our data and the fact that visual resolution is generally much finer than haptic resolution, we cannot measure this small haptic contribution, and the trends of produced force with distortion in our work are indistinguishable from what we would expect if vision was completely dominant. If we use a visual display with a resolution at least as fine as that of the discrete display used here, we can expect that the perception of force and distance by a patient in our experimental environment will be dominated by vision rather than kinesthetic feedback.

While our with-feedback data support the dominance of vision in these experiments, two anomalies must be discussed. In FI-Y and CI, the slopes for produced force versus block number (distortion) for force levels 1 and 2 were not significantly different from zero, though they were also not significantly different from the slopes predicted by visual dominance. This is because the slopes predicted by visual dominance were small for these low force levels. We cannot statistically distinguish these small slopes from zero due to a lack of power. The slope of produced force versus distortion for distance level 1 in DI was not significantly different from zero for the same reason. The second curiosity in our data is that in CD, the slope of produced force versus block number did not become significantly more negative with increasing force level, as visual dominance would predict. This is primarily due to force level 5, for which the slope was almost significantly different from what would be predicted by visual dominance. Subjects tended to decrease their produced forces less with distortion than was predicted. We speculate that this may be because subjects felt that it should not be "too easy" to produce a level 5 force, so the tendency to continue producing an effortful force may have countered the visual distortion somewhat.

5.5.2. Visual Dominance Occurs Even at High Levels of Distortion

What makes our work distinct from that of Ernst and Bank [9] is that visual dominance was maintained at large levels of distortion. A sudden distortion of two JNDs would be easily noticeable, and a subject who detected the distortion might lose confidence in the visual display, possibly reducing the amount of visual dominance. But because we increased the distortion in steps of less than one JND, subjects did not notice the distortion and did not discredit the visual display. Instead, they increased the forces they produced by 30% or more. This was particularly true of the stroke and TBI subjects. P1 increased her level 5 force by 97.7%, and the other motor-impaired subjects demonstrated similarly large increases. To encourage patients to move farther and push harder in therapy, progressively larger amounts of distortion will be needed. These results show that we can use such progressive distortion without eliminating visual dominance in our environment.

5.5.3. Effect of Informing Subjects of the Possibility of Distortion

One practical issue that arises when considering the use of distortion in a long-term rehabilitation paradigm is what happens when the patient suspects distortion. It is easy to imagine a slip in conversation, a computer glitch, etc. that might reveal the possibility of distortion to a patient. It might be supposed that such a suspicion would reduce or eliminate any positive effects of distortion. This concern is alleviated by the results of CI. When subjects were informed of the possibility of distortion, they often claimed to recognize the distortion, but they were unable to identify the direction of the distortion. The results from these subjects did not differ materially from those of subjects who were not told that the display might be distorted. Welch and Warren [32] review experiments showing that knowledge of distortion did not reduce visual dominance when the amount of distortion was small. Our results extend that research, in that knowledge of possible distortion did not reduce visual dominance in a situation with large distortions where the distortions were reached gradually. This result is crucial for the use of visual distortion in rehabilitation. When considering the use of distortion during rehabilitation protocols extending over weeks or months, it is unrealistic to expect that subjects will never suspect distortion. This experiment indicates that such suspicion should not affect the results of the distortion.

5.5.4. Effect of Visual Distortion on Internal Somatosensory Representation of Movement Goal

Trials with no visual feedback were included in each experiment to determine the extent to which visual distortion manipulated the somatosensory representation of the goal. In all experiments with young subjects, subjects maintained the upward/downward trends in movement force/distance encouraged by distortion even on the trials with no visual feedback. These results indicate that the somatosensory representation of the goal corresponding to each force or distance level was affected by the visual distortion. As a subject maintained a constant visual goal on the visual-feedback trials, he or she gradually adjusted the somatosensory representation of the goal to match the somatosensory input experienced at the visual target. This gradual recalibration resulted in an upward/downward trend in produced force or distance even on the trials with no visual feedback.

We observed fewer significant upward trends on the no-feedback trials for elderly subjects (and 2 of the motor-impaired subjects), though the slope of produced force versus distortion was never significantly different from that predicted by visual dominance. Larger amounts of distortion did not lead to larger increases in force for these trials, and in addition, the slope of produced force with distortion tended to be smaller for elderly subjects than for young subjects. We believe that this may be due to the large discrepancy between the with-feedback force and the no-feedback force for each force level for elderly subjects at the beginning of the experiment (0% distortion). For 0% distortion, the mean level 5 no-feedback force was 1.01 N higher than the mean level 5 with-feedback force for elderly subjects. This difference was only 0.336 N for young subjects. At maximum distortion (towards the end of the experiment), the difference between the mean level 5 no-feedback force and the mean level 5 with-feedback force was 0.545 N for young subjects, but it was only 0.0228 N for elderly subjects.

The overestimation of the force in the no-feedback condition for elderly subjects at the beginning of the experiment recalls the work of Cole [103], who found that elderly subjects use greater grasp forces than young subjects, possibly to compensate for decreased sensation. The larger difference for elderly subjects between the no-feedback and the with-feedback forces at the beginning of the experiment is also attributable to the larger force JND of elderly subjects. Throughout the experiment, the elderly subjects produced no-feedback forces that they thought were consistent with their with-feedback forces. As they increased the with-feedback forces to match the visual distortion, the elderly subjects had to increase their no-feedback forces by only a small amount, because the no-feedback forces started at a higher level. The visual feedback may have changed elderly subjects' somatosensory representations of the force goals, but we cannot observe that change due to the initial force discrepancy. It is interesting that young subjects maintained the discrepancy between feedback conditions throughout the experiment. Jones [16] reported a similar overestimation phenomenon for forces well below the maximum voluntary force. We speculate that younger subjects may overestimate because they want to make sure they reach the force they produced with visual feedback. The force discrepancy may not be maintained in elderly subjects because they become more easily fatigued. The differences in performance on the no-feedback trials is of interest because practice of tasks without feedback enhances retention of learning [51]. However, because the differences between with- and no-feedback trials for elderly subjects were most likely due to the initial force discrepancy, rather than to a persistent tendency to produce less force in the absence of visual feedback, these differences should not have an effect on the use of distortion to encourage increases in force production or movement distance.

Given the large JNDs of the motor-impaired subjects, we might expect that their performance would be similar to that of the elderly subjects. Like elderly subjects, motor-impaired subjects tend to exhibit decreased sensation and compensate by increasing grip force [40]. However, of the four motor-impaired subjects, only P1 had an initial no-feedback force for level 5 much greater than the initial with-feedback force. P1's no-feedback data did lack an upward trend, but at least two of the motor-impaired individuals did exhibit upward trends in their no-feedback data. More work with motor-impaired subjects is necessary before their data can be compared with data from the young and elderly unimpaired subjects.

5.5.5. Performance in the Absence of Visual Feedback

Because elderly subjects tended to produce more force in the no-feedback condition, it might be proposed that task practice without visual feedback might be more effective than using distortion to encourage improvements in rehabilitation tasks. This idea might also emerge from the apparent increases in force production seen in Figure 5.3 for CF. Subjects seemed to overestimate forces in the absence of visual feedback, possibly to ensure that they produced a force at least as large as that they previously produced for that force level. While simply eliminating visual feedback may seem appealing, the upward trends in force in CF were generally not significant, and the increase in force production in the absence of visual feedback was not predictable or controllable in the way that the increases seem with visual distortion were, thus making visual distortion more useful for application to rehabilitation.

5.5.6. Conclusion

These experiments show that we can use visual distortion to effectively manipulate force production and movement distance for unimpaired and motor-impaired subjects performing well within the limits of their ability. In addition, even when subjects suspect distortion, the distortion still influences their performance. This gives us good reason to believe that distortion may be a helpful addition to rehabilitation for our target patient population: patients who choose to maintain a steady level of performance below their level of full ability, rather than pursuing more challenging goals. However, some research with weightlifters suggests that distortion may result in better performance even for a subject performing at the limit of his or her ability [104-106]. This effect has been measured for a weightlifter's one-repetition maximum. The next chapter describes experiments designed to examine whether distortion affects learning of a more complicated motor task when subjects are performing at the limit of their ability and the distortion is present for a large number of trials. Preliminary results for the experiments described in this chapter can be found in [99, 107].

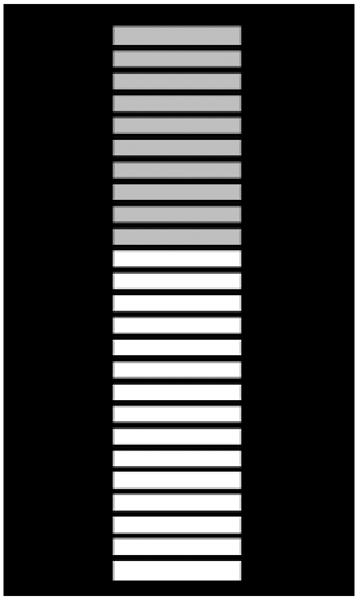


Figure 5.1. The visual display used for the experiments described in this chapter. The number of shaded boxes indicated the magnitude of the force exerted by the subject. The direction of increasing force was vertically downward. Visual distortion was accomplished by changing the range of force represented on the feedback bar.

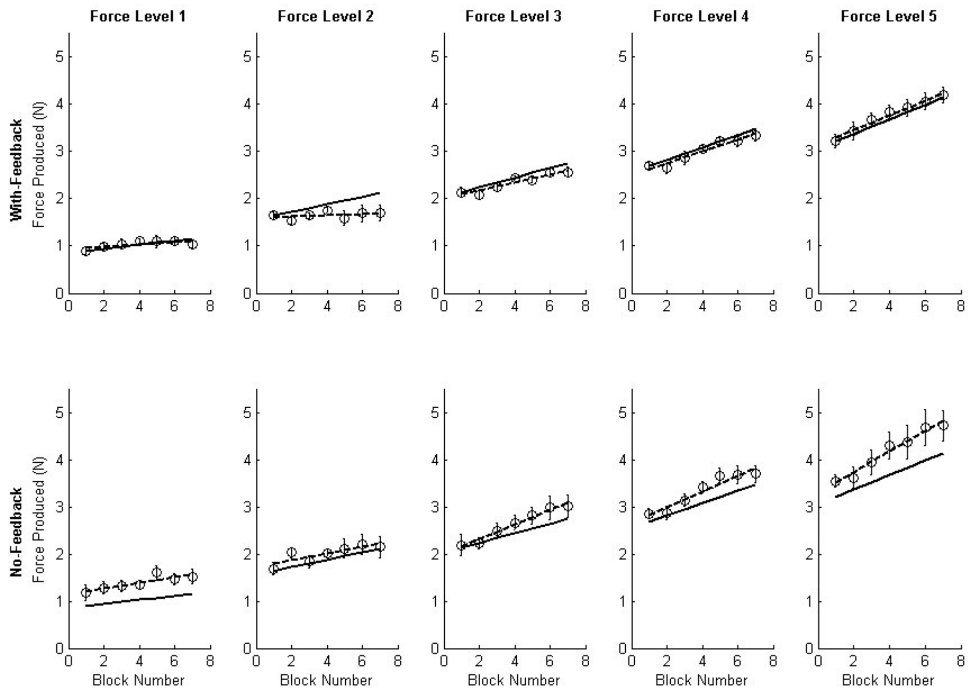


Figure 5.2. The results of experiment FI-Y (distortion to increase force production, young subjects). The mean produced force (circles) is plotted as a function of the block number for each force level/feedback condition. The with-feedback condition is shown along the top, and the no-feedback condition is shown along the bottom. Column 1 shows the data for force level 1, column 2 the data for force level 2, etc. Error bars indicate the standard error. The dotted line in each graph is the mean regression line for that data set. The solid line in each plot shows the trend in force predicted by total visual dominance. The slope of produced force with distortion was significantly different from zero for force levels 3-5 in both feedback conditions and for force level 1 in the no-feedback condition. No slope was significantly different from that predicted by visual dominance.

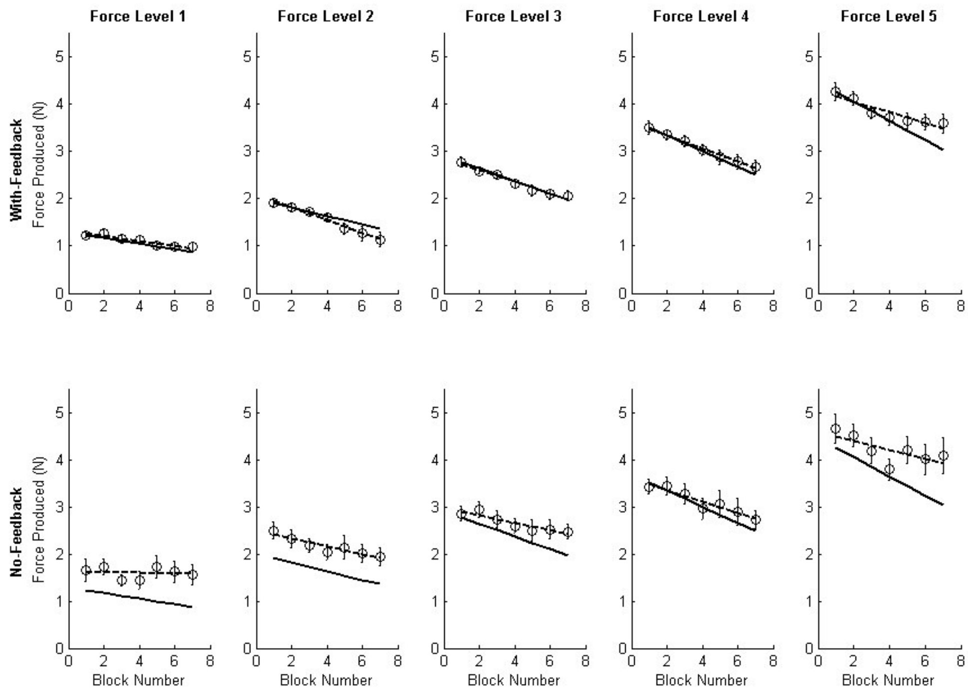


Figure 5.3. Results of experiment CD (distortion to decrease force production). All plots and symbols are analogous to those in Figure 5.2. Significant downward trends for force as function of block number were found for every force level in the with-feedback condition and for force levels 2-4 in the no-feedback condition. The slope of force versus block number was significantly different from that predicted by visual dominance only for force level 1 in the no-feedback condition.

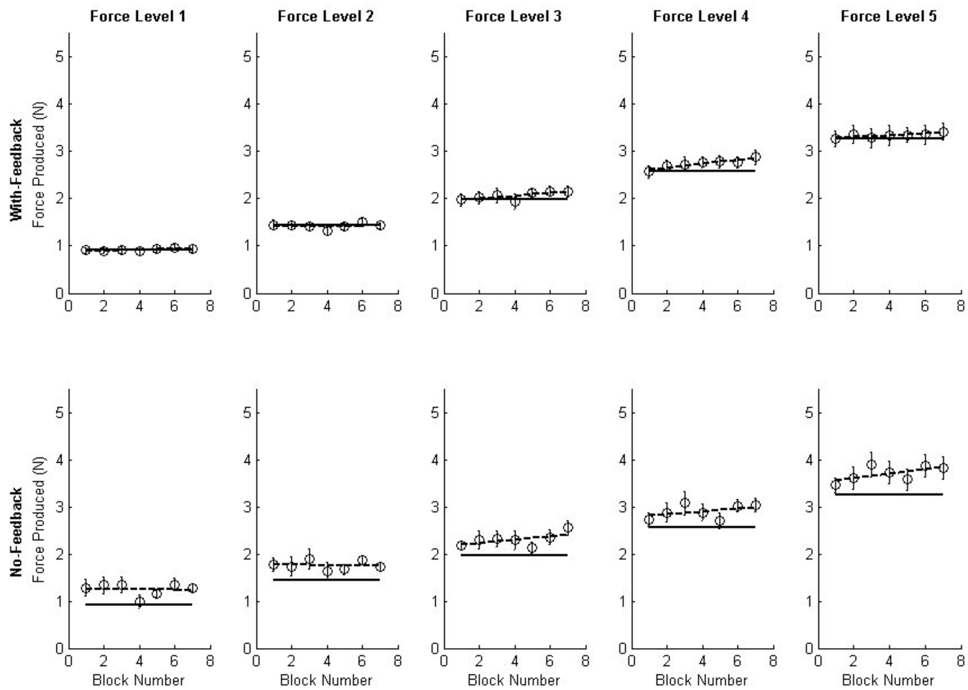


Figure 5.4. The results of experiment CND (no-distortion control). All plots and symbols are analogous to those shown in Figure 5.2. The slope of produced force versus block number was not significantly different from zero for any force level/feedback condition.

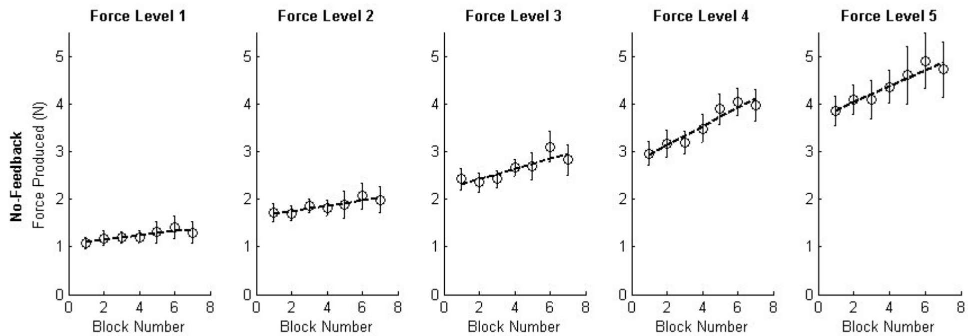


Figure 5.5. The results of experiment CF (no-feedback control). The circles show produced force as a function of block number for each force level. Error bars indicate standard error, and the dashed line in each plot represents the mean regression line. The slope of produced force with block number was significantly different from zero only for force level 4.

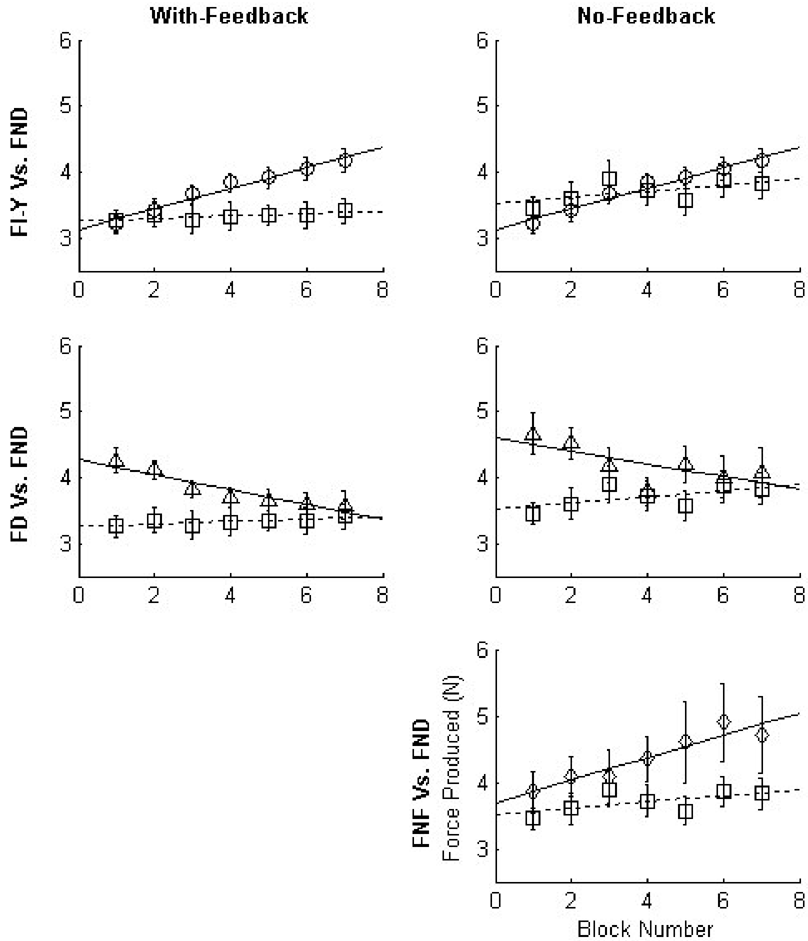


Figure 5.6. Effects of distortion on force production. The top row shows a comparison of FI-Y (circles, solid line) and CND (squares, dotted line) for force level 5. The left column shows the with-feedback condition, and the right shows the no-feedback condition. The slope of produced force with distortion was significantly different between FI-Y and CND for both feedback conditions. The second row shows the comparison of CD (triangles, solid line) to CND (squares, dotted line). The slope of force with distortion was also significantly different for CD when compared to CND for both feedback conditions. The bottom row compares CF (diamonds, solid line) to the no-feedback condition of CND (squares, dotted line). No significant difference was found between CF and CND. All plots show block number along the *x*-axis and the force produced along the *y*-axis, as demonstrated for the CF vs. CND plot. Bars represent standard error.

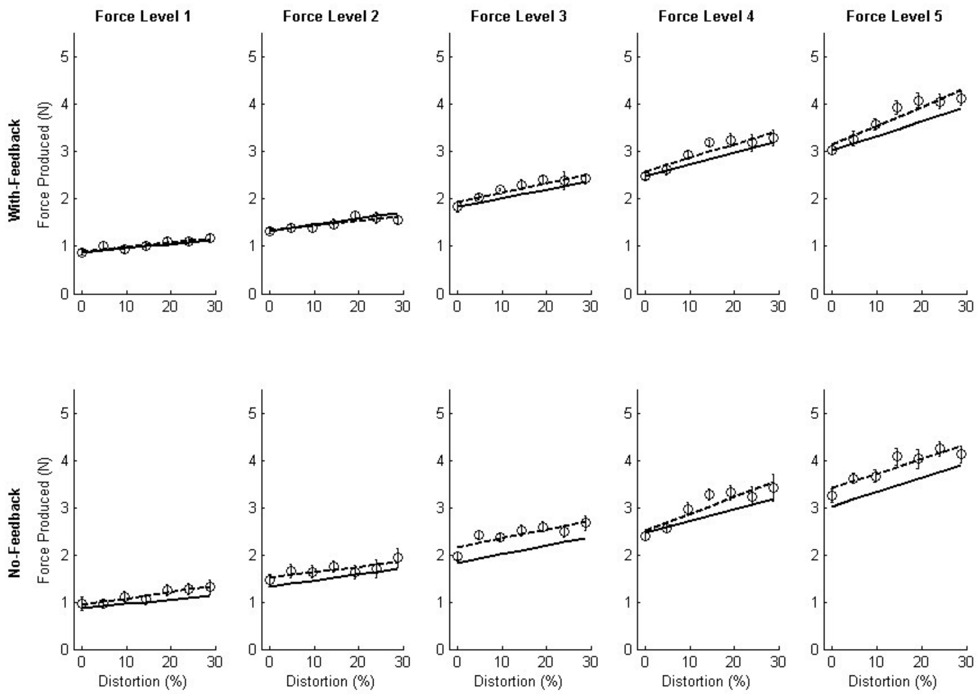


Figure 5.7. The results for experiment CI (distortion to increase force production, young subjects informed of the possibility of distortion). All plots and symbols are analogous to those shown in Figure 5.2, except that force is plotted as a function of percent distortion, rather than block number. The slope of produced force with distortion was significantly different from zero for force levels 3-5 in both feedback conditions and for force level 1 in the no-feedback condition. No slope was significantly different from that predicted by visual dominance.

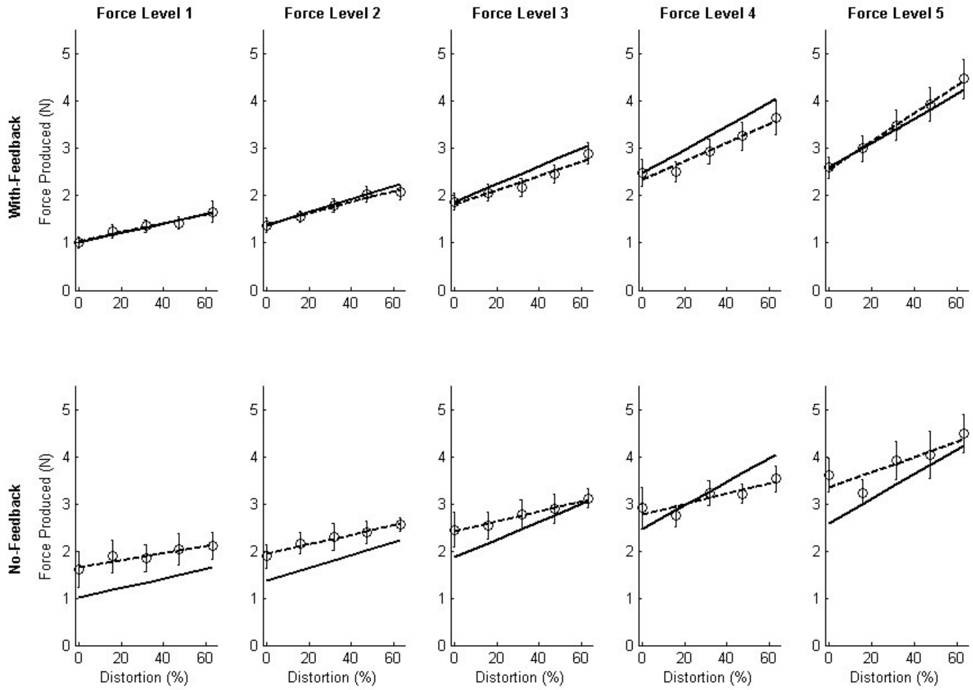


Figure 5.8. The results for experiment FI-E (distortion to increase force production, elderly subjects). All plots and symbols are analogous to those shown in Figure 5.7. The slope of produced force with distortion was significantly different from zero for all force levels in the with-feedback condition and for force levels 1, 2, and 5 in the no-feedback condition. No slope was significantly different from that predicted by visual dominance.

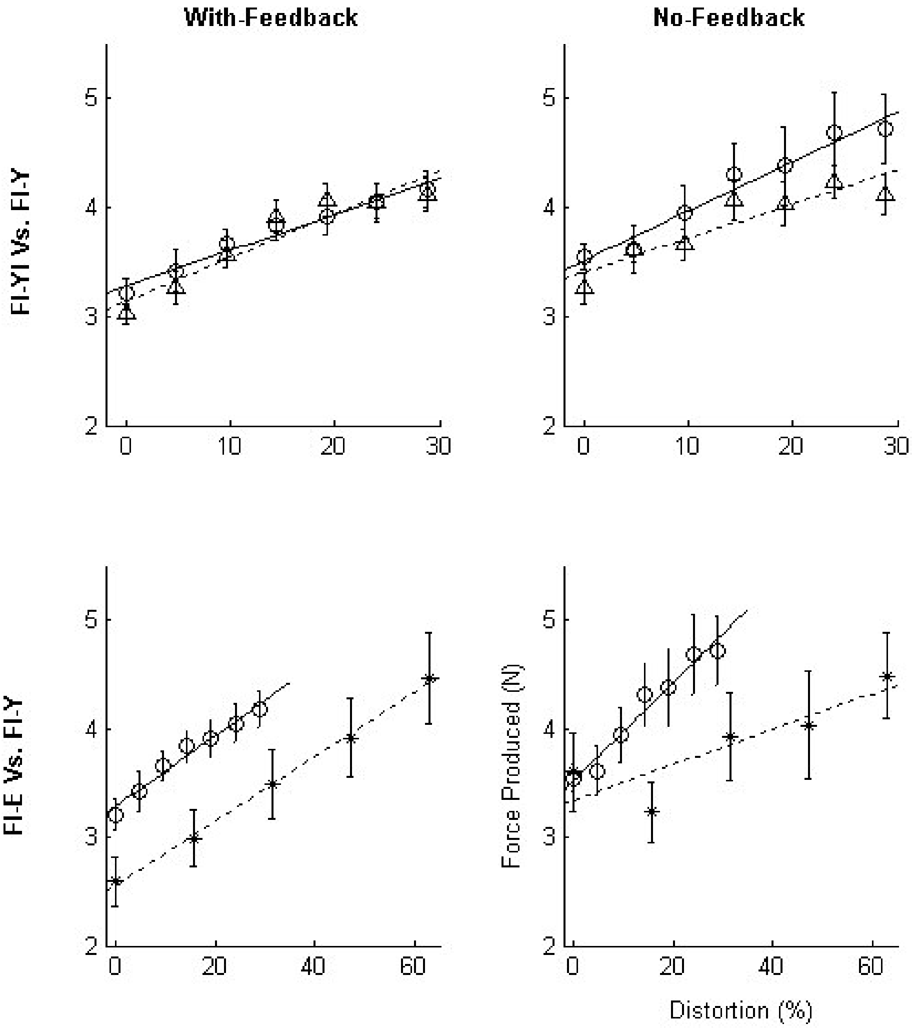


Figure 5.9. Influence of instruction and age on the effects of distortion. The top row shows a comparison of CI (triangles, dotted line) and FI-Y (circles, solid line) for force level 5. The left column shows the with-feedback condition and the right column shows the no-feedback condition. No significant differences in the slope of force with distortion were found between FI-Y and CI. The bottom row shows a comparison of FI-E (asterisks, dotted line) and FI-Y (circles, solid line). There was no significant difference between FI-Y and FI-E in the with-feedback condition, but the difference in slope in the no-feedback condition was close to significance (p = 0.0557). The x-axis in all plots represents the percentage distortion, and the y-axis represents the produced force, as shown in the lower-right plot. Bars represent standard error.

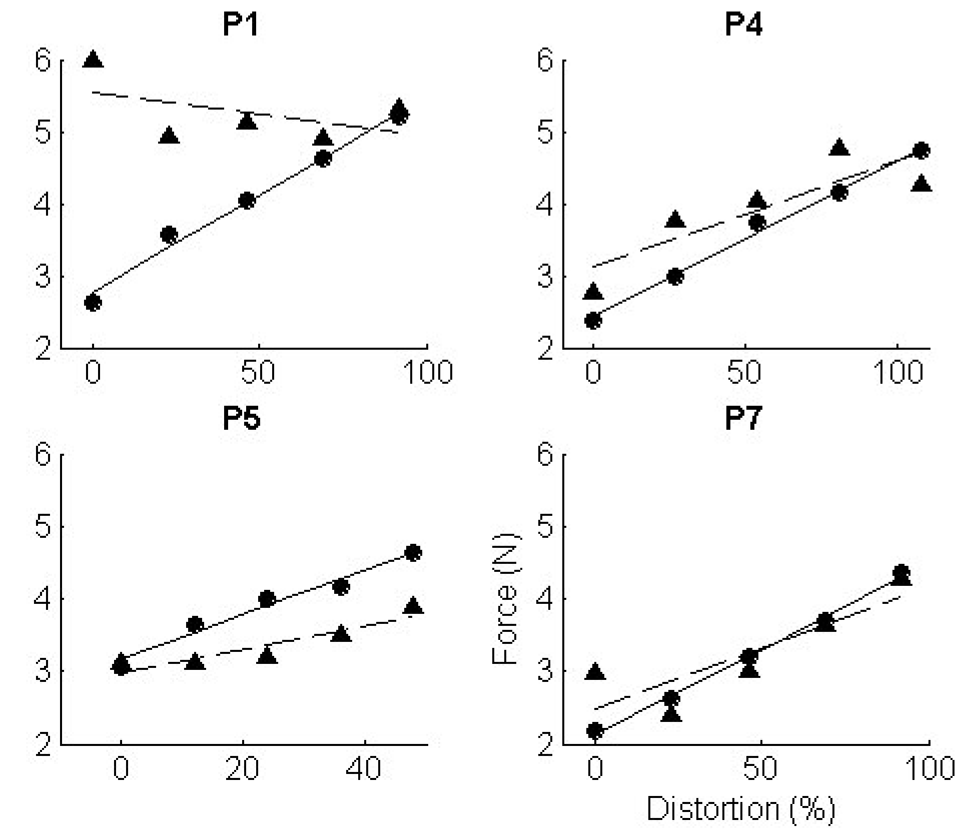


Figure 5.10. The results of FI-MI (distortion to increase force production, motor-impaired subjects) for force level 5. Each plot corresponds to a motor-impaired individual. The *x*-axis indicates the percent distortion, while the *y*-axis represents force, as shown in the plot for subject P7. Each point represents the mean of two trials for a given percentage distortion. Circles represent with-feedback trials, and triangles represent no-feedback trials. Regression lines are shown.

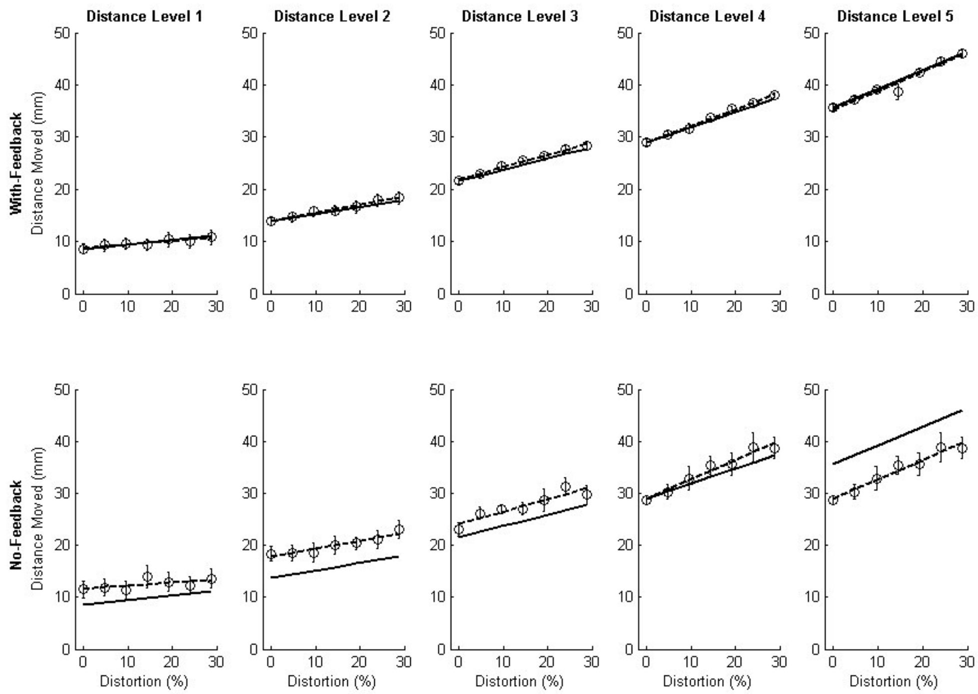


Figure 5.11. The results of experiment DI (distortion to increase movement distance). The mean distance moved (circles) is plotted as a function of percent distortion for each force level/feedback condition. The with-feedback condition is shown along the top, and the no-feedback condition is shown along the bottom. Column 1 shows the data for distance level 1, column 2 the data for distance level 2, etc. Error bars indicate the standard error. The dotted line in each graph is the mean regression line for that data set. The solid line in each plot shows the trend in distance predicted by total visual dominance. A significant positive trend in movement distance with distortion was found for every distance level/feedback condition except distance level 1 in the with-feedback condition. No slope was significantly different from that predicted by visual dominance.

Experiment	Description	Subject Group
FI-Y	Distortion to increase force production	young
FI-E	Distortion to increase force production	elderly
FI-MI	Distortion to increase force production	motor-impaired
DI-Y	Distortion to increase distance production	young
CND	Control with no distortion	young
CNF	Control with no visual feedback	young
CI	Control with subjects instructed about possible distortion	young
CD	Control with distortion to decrease force production	young

Table 1. The name and a brief description of each of the experiments designed to determine the effects of distortion on force and distance production.

Experiment	# of Subjects	Female/Male	JND	# of Trials	Distortion Levels	Beginning Limits	Break
FI-Y	10	3/7	14.4%*	140	7	0.5 - 3.49 N	Yes
FI-E	6	3/3	31.5%**	100	5	0.5 - 3.22 N	No
FI-MI	1	1/0	Varied #	100	5	0.5 - 2.73 N	No
DI-Y	10	5/5	14.4% ##	140	7	5.00 - 38.83 mm	No
CND	10	3/7	N/A	140	none	0.57 - 4.00 N	Yes
CNF	10	N/A	N/A	140	none	N/A	Yes
CI	9	4/5	14.4%*	140	7	0.5 - 3.49 N	No
CD	10	3/7	14.4%*	140	7	0.64 - 4.50 N	No

* 14.4% was an estimate of the young subject force JND made from preliminary data. This JND was later measured more accurately to be 19.7%.

** 31.5% was an estimate of the elderly subject JND made from preliminary data. It was very close to the value of 31.0% measured later.

The JND for each subject can be found in Chapter 4. We used P1's force JND for P7 because we were unable to precisely compute P7's JND.

The young subject distance JND was initially assumed to be the same as the force JND. This JND was later measured to be 13.0%.

Table 2. Characteristics of each of the experiments described in Table 1.

Chapter 6: Enhanced Visual Error in a Coordinated Pinch Task

6.1. Introduction

In the experiments described in the previous chapter, subjects were free to set the target force or distance specified by a particular numerical magnitude. The purpose of the distortion was to change the force or distance goal corresponding to each level. The present work used a different distortion paradigm that we call "error enhancement." According to this paradigm, subjects were asked to work at a demanding task with an objective performance criterion. Departures from ideal performance were displayed as errors, which were distorted to appear larger. We chose to examine the effects of distortion on the learning of a difficult two-finger coordination task. A coordinated pinch task was chosen because coordination of multiple fingers is important for many activities of daily living. As a subject learned the coordination task, we presented feedback about his or her error in performance. Distortion was implemented as an artificial inflation of this error. We first investigated whether error enhancement would improve performance of either finger or of both fingers relative to controls. Some evidence to this effect for one limb comes from Patton et al. [7], who showed that magnifying visual error by physically displacing the arm's trajectory resulted in smoother and straighter trajectories in stroke patients. We further asked whether distortion of the feedback corresponding to a single finger would improve the performance of just the distorted finger. This might occur, in particular, if the feedback induced subjects to focus their attention on the distorted finger. Alternatively, it is possible that trying to improve just one finger at a time provides a superior training method relative to attempting to coordinate both, in which case transfer between the fingers could be observed. We also looked at whether distorting the feedback for both fingers would improve the performance of both relative to controls who received no distortion.

6.2. Experimental Environment

The robotic environment used in this experiment is shown in Fig. 6.1. Two Premium 1.0 PHANTOM^{TM} force-feedback robots (SensAble Technologies, Inc.) were used. Each robot had 3 active degrees of freedom and a position resolution of 0.03 mm [89]. To each robot was attached a standard finger cuff made by Sensable Technologies, Inc. The subject placed the index finger in one finger cuff and the thumb in the other. The remaining fingers grasped a post to keep the hand stationary throughout the experiment. The subject sat with the arm flexed at the elbow and the forearm horizontal.

Each experiment began with a set of calibration steps. These steps were also repeated after each break given during the experiment. During these calibration steps, the experimenter placed her hand over the subject's and gently moved the index finger and the thumb into the appropriate positions. The subject began with the index finger and the thumb flexed and touching. Next, the experimenter extended the index finger and thumb and then flexed them together again. The subject then extended the finger and thumb until he or she reached two virtual walls simulated by the robots. The wall that the subject extended the thumb against was defined by a plane normal to the line segment in the *xz*-plane that connected the flexed position and the extended position of the thumb found during calibration (Figure 6.2(a)). The plane passed through this line segment at the point $(0.2x_{tf} + 0.8x_{te}, 0.2z_{tf} + 0.8z_{te})$ in the *xz*-plane, where (x_{tf}, z_{tf}) is the flexed position of the thumb and (x_{te}, z_{te}) is the extended position. The plane defining the virtual wall for the index finger was defined similarly. The force simulation corresponding to each virtual wall was a stiff virtual spring. When the subject's finger was beyond the plane defining a virtual wall, a force proportional to the distance of the finger from the plane was exerted by the robot. The direction of this force was normal to the plane defining the virtual wall and in the direction opposing further extension of the finger. The virtual walls were implemented to provide a position of partial index finger and thumb extension from which to start each trial. The walls were not placed at positions of full extension because a small shift in hand position might then prevent the subject from being able to reach the position required to start a trial.

Information gathered during the calibration steps was also used to place a virtual object between the index finger and the thumb. The walls of this object were defined by two planes normal to the line segment in the *xz*-plane connecting the extended position of the thumb with that of the index finger (Figure 6.2(b)). The wall closest to the thumb passed through a point on this line segment defined by $(x_{tfp}, z_{tfp}) - \frac{1}{2} w\hat{d}$, where w is the experimenter-defined object width, $\hat{d}$ is the unit vector corresponding to the vector connecting the extended position of the thumb in the xz-plane with that of the index finger, and (x_{tfp}, z_{tfp}) is the flexed position of the thumb projected onto $\hat{d}$. The wall closest to the index finger passed through the point $(x_{ifp}, z_{ifp}) + \frac{1}{2} w\hat{d}$, where (x_{ifp}, z_{ifp}) is the flexed position of the index finger projected onto $\hat{d}$. During the experiment, the subject moved the index finger and thumb only between the virtual object and the virtual walls that defined the starting position for each trial. The experimenter moved the subject's hand during the calibration steps to ensure that the experiment was initialized for each subject in such a way that the subject had sufficient range of motion for the thumb and index finger between the starting position and the virtual object. Preliminary experiments revealed that when subjects were simply told to flex and extend the finger and thumb, they moved the thumb very little, and the starting position for the thumb was too close to the virtual object for successful completion of the experiment.

6.3. Methods

Sixty-one subjects between 18 and 35 years of age participated in this experiment. No subject had a history of known neurological injury. All gave informed consent and performed the experiment with the dominant right hand.

During this experiment, each subject learned to move the index finger and the thumb in a particular target pattern while receiving visual feedback corresponding to each finger. On each trial, the subject started with the index finger and thumb extended. The visual feedback seen by the subject during each trial is shown in Fig. 6.3(a). The small white rectangle at the top of the screen represented the virtual object implemented using the force-feedback capabilities of the robots. The circles to the left and right of the white rectangle represented the thumb and the index finger, respectively. The large bar on the left represented the distance from the starting position of the thumb to the edge of the virtual object. The bar on the right represented the same for the index finger. The shaded portion of each bar represented the normalized position of the corresponding finger between the starting position and the virtual object (0 at the bottom of the bar and 1 at the top).

The line crossing both bars showed the subject the target movement pattern for the thumb and the index finger. The line started at the bottom of the bars, and as soon as the subject crossed the line with the shaded portion of either bar, the line began to move. On every trial, the line moved for 8 seconds according to the equation $L = 0.5 + 0.95\frac{\Delta t}{8} + 0.2375\sin\left(\frac{4\pi\Delta t}{8}\right)$, where Δt is the time in seconds that has passed since the beginning of the trial and L is the normalized position of the line along the bars (Fig. 6.3(b)). The subject's goal was to learn to keep the tops of the shaded portions of the bars as close to the line as possible. For each trial, we computed the mean absolute difference between the normalized position of each finger and the target line; we call this the trial absolute error for each finger. The trial absolute error for each finger was displayed to the subject after each trial. As an incentive to maintain subject attention to the task, each subject was told that he or she would receive a candy bar if the error over the entire experiment was below a predetermined threshold. The subject was not told the threshold.

Visual distortion was implemented as a magnification of the error between the thumb and/or index finger position and the position of the target line. When the visual feedback of the index finger was distorted, the signed error between the index finger and the target line was computed in real time and then increased by 20%. The height of the shaded bar on the screen at each time point was obtained by adding the modified error to the height of the target line. Thus, the height of the shaded bar no longer corresponded faithfully to the position of the subject's index finger. The numerical error for the index finger given to the subject at the each of each trial was also increased by 20%. The feedback for the thumb was distorted in the same way. We measured the effect of different distortion modes on learning as measured by the trial absolute error.

Three experimental conditions were considered in addition to a control condition in which the visual feedback was not distorted. Ten subjects participated in each condition, except for the ITB (index-thumb-both) condition, which had eleven subjects. No subject participated in more than one condition. In the ITB (index-thumb-both) condition, each subject experienced 80 training trials in which there was no distortion. The subject then encountered a block of 40 trials in which the visual feedback for the index finger was distorted, followed by a block of 40 trials in which the visual feedback for the thumb was distorted. The experiment concluded with a block of 40 trials in which the visual feedback for both the index finger and the thumb was distorted (scale change relative to original feedback). The TIB (thumb-index-both) condition was similar, except that the block of trials with distorted thumb feedback occurred before the block of trials with distorted index finger feedback. In the B (both) condition, the initial learning period of 80 trials was followed by a single block of 40 trials in which the visual feedback for both the index finger and the thumb was distorted. Breaks were given frequently to prevent subject fatigue. Post-experiment questionnaires revealed that subjects did not detect the distortion. All but four subjects stated that they tried to move the finger and the thumb in a coordinated way.

After the first 20 trials, one random trial out of every 20 was a no-feedback trial. On the no-feedback trials, the normalized position of each finger was not shown; the screen appeared the same, except that no part of the visual feedback bars was shaded. These trials were included to assess whether subjects could reproduce the target movement without feedback about finger position.

In these experiments, the subject's attention was divided between the index finger and the thumb. To examine how this affected our results, we also ran a control experiment in which each subject tried to learn a target pattern using only the thumb. This experiment was similar to that described above. Only one robot was used, and the subject only saw feedback corresponding to the thumb. The target pattern of movement shown by the line was the same. Two conditions were considered in this experiment. One group of ten subjects experienced 80 trials with no distortion, then 40 trials with 20% distortion, followed by another 40 trials with no distortion. Another group of ten subjects experienced 120 trials with no distortion, followed by 40 trials with 20% distortion.

6.3. Results

6.3.1. Learning of the Target Task

Data from the control subjects who used both the index finger and the thumb were examined to determine whether learning occurred for the target task. The trial absolute error (the mean absolute difference between the normalized index finger position and the position of the target line) for the index finger was added to that of the thumb for each trial. This sum was averaged over trials 21-40 and over subjects to obtain the mean total absolute error for these trials, which was compared to the mean total absolute error for trials 181-200 using a t-test. Data from no-feedback trials were excluded from this analysis. The first twenty trials were not used for this comparison because some subjects misunderstood the task during the first few trials. The mean total absolute error for trials 21-40 was significantly different from that for trials 181-200 ($p < 0.001$). The learning trend in the control data can also be seen in Fig. 6.4. When the mean absolute error for the index finger was compared to that of the thumb, the thumb error was significantly larger for trials 21-40 ($p = 0.0068$), but not for trials 181-200 ($p = .128$).

To investigate the reasons for these decreases in error, we compared the mean absolute difference (computed over trials and over subjects) between the normalized position of the index finger and the normalized position of the thumb during trials 21-40 and trials 181-200. The difference in position was less for the later trials ($p < 0.001$), which means that subjects learned to move the thumb and the index finger in a more coordinated fashion during the experiment. We also considered the lag between the position profile of each finger and the position of the target line. The lag for each trial and finger was computed by finding the difference in time that maximized the correlation between the finger position and target line position. Essentially, this indicates the time period by which the subject's response trailed the target movement. The mean lag (computed over trials and over subjects) for trials 21-40 was not significantly different from the lag for trials 181-200 for either the index finger or the thumb ($p = 0.775$ for index, 0.501 for thumb). Subjects learned to keep the normalized position of each finger closer to the target line during the experiment, but the mean lag of each finger remained the same (Fig. 6.5).

Data from the control subjects were also used to examine whether learning in the with-feedback trials transferred to improvements in performance on the no-feedback trials (Fig. 6.6). Subjects had much larger errors on the no-feedback trials throughout the experiment. Even at the end of the experiment (trials 181-200), the mean total absolute error for the no-feedback trial was significantly larger than that for the with-feedback trials ($p < 0.001$). However, the difference between the mean total absolute error on the second and last no-feedback trials was close to significance ($p = 0.0542$), indicating a trend toward a learning effect in overall error. The first no-feedback trial was excluded because despite instructions, many subjects were confused when the first no-feedback trial occurred. The mean absolute difference between the normalized position of the index finger and that of the thumb decreased from the second no-feedback trial to the last ($p = 0.0240$), showing an improvement in coordination of the finger and thumb on the no-feedback trials.

6.3.2. Effects of Distortion

To assess the effects of the distortion conditions, the data for each ITB, TIB, and control subject were divided into four blocks: trials 41-80, trials 81-120, trials 121-160 and trials 161-180. The first forty trials were regarded as practice and excluded, because we wanted to examine how distortion affected performance after subjects had acquired a level of mastery of the task. For each subject, the trial absolute error for each finger was averaged over each block. Next, for each finger and each subject, the mean absolute errors of blocks 1-4 were divided by the mean absolute error for block 1 to reduce variability due to individual differences in baseline error. We used this error measure, which we will call the normalized error, for our analysis. The normalized error measured performance improvement for each subject as a proportion of the baseline error. The data for subjects in condition B were analyzed similarly using only two blocks, trials 41-80 and trials 81-120.

To determine whether distortion affected terminal performance, we compared the mean normalized error in block 4 for each distortion condition to the control. None of these comparisons was significant (ITB: $p = 0.0669$ for index, 0.689 for thumb; TIB: $p = 0.144$ for index finger, 0.991 for thumb). Similarly, the mean normalized error for block 2 of condition B was compared to mean data from block 2 for the control condition; there was no significant difference for either finger ($p = 0.704$ for index finger, 0.835 for thumb).

Though distortion did not affect subjects' final performance in the task, it did appear to change the shape of the learning curve, as seen in Fig. 6.7. To investigate the significance of these apparent changes, we conducted two types of comparisons. First, we compared experimental and control subjects using the normalized error for each finger in blocks 2 and 3, in which the error of a single finger was distorted (the finger whose error was being distorted changed between blocks). A difference would indicate an effect of distortion. The normalized index finger error for ITB subjects in block 2 did not differ significantly from that of control subjects ($p = 0.235$), but the ITB index error in block 3 was significantly larger than the control error ($p = 0.0128$). The normalized thumb error for ITB subjects did not differ from the control error in either block, though the comparison for block 2 approached significance ($p = 0.113$ for block 2, 0.425 for block 3). No significant differences were found between TIB and control subjects.

The second comparison we made between groups was with respect to the change in normalized error when we changed which finger error was distorted (between blocks 2 and 3). A significant difference in this measure would indicate an effect of distortion different from the learning effect seen in the controls. For the index finger of ITB subjects, the difference between the mean block 2 error and the mean block 3 error was significantly smaller than for control subjects ($p = 0.0034$). For the thumb, the error difference was significantly larger for ITB subjects than for controls ($p = 0.0046$). These results tell us that the learning curve for each finger changed differently from block 2 to block 3 for ITB subjects when compared to control subjects. Specifically, distortion did not make the corresponding finger perform better, but it did negatively affect the performance of the finger whose error was not distorted. Corresponding changes were not observed for TIB subjects.

6.3.3. Control Experiment Using Only the Thumb

Data from each subject in the thumb-only control experiment were divided into three blocks (trials 41-80, trials 81-120 and trials 121-160) and analyzed as above. No significant differences were observed between the group that received distortion in block 2 and the group that received distortion in block 3.

6.4. Discussion

6.4.1. Learning of the Target Task

We examined learning in a challenging coordinated pinch task. We found that the average time by which the subject trailed the target movement remained constant while subjects learned to reduce their mean absolute error. Increased coordination of the index finger and thumb in normalized distance space was a product of this task, as subjects had to track the target line with both fingers simultaneously. Subjects stated that they performed the task by concentrating on moving the index finger and the thumb together, though they had to move the two fingers at different rates because the index finger and the thumb did not necessarily start at the same distance from the virtual object.

The results of this experiment show that subjects can learn to control even performance variables that are not clearly related to daily tasks. While humans do have to consider the position of the index finger and thumb in daily life, subjects in this experiment were required to move the thumb more than they naturally tended to do when grasping objects, and they had to track a target with an arbitrary sinusoidal path. Latash et al. [108] found that subjects could not learn to overcome certain patterns of coordination among the fingers. The results of this experiment show that the typical pattern of coordination between the index finger and the thumb can be modified through learning.

Learning in this coordination task transferred partially to the no-feedback case. In this task, subjects learned a particular pattern of movement as they tried to minimize visual error. Because emphasis was placed on the visual error, it is no surprise that a subject's error was greater when executing the learned movement without visual feedback of position. However, the mean total absolute error in the no-feedback case did seem to decrease as the experiment progressed, and subjects learned to move the index finger and the thumb in a more coordinated manner on the no-feedback trials. More transfer to the no-feedback case might have been observed if more no-feedback trials had been included. For effective learning and transfer of a motor task to occur, subjects must learn to use internal cues rather than relying on extrinsic feedback [51, 109].

6.4.2. Effects on Terminal Performance

One of our hypotheses for this experiment was that visual enhancement of error would improve the terminal performance. The improvement might be specific to the distorted finger, or decomposing the task might have a more generally beneficial effect across the fingers.

Although distortion affected ongoing performance while individual finger error was distorted, it did not improve terminal performance when both finger errors were distorted together at the end of the experiment. In the experiment involving both fingers, there was no difference between the performance of control subjects and that of experimental subjects receiving distorted visual feedback for both fingers (B condition and the last block of ITB and TIB). Also, in the thumb-only experiment, distortion had no effect on the normalized error of subjects. These results are similar to those of [110], which reported that error augmentation using a multiplicative gain did not improve terminal performance in a reaching task. Error augmentation through a constant offset was found to be more effective [110], but that type of distortion would not be relevant for the task we considered.

The results of this experiment appear to contrast with those of our previous work, in which goal distortion was used to increase force production and movement distance over the time-course of a single session [111]. Some of these differences may be explained by considering the differing natures of the two tasks. Subjects participating in previous experiments were asked to move only one finger while concentrating on their force/distance production. The change in their goals under distortion appears to reflect a re-mapping between kinesthetic feedback and the visual display. Subjects in this experiment, on the other hand, were learning a complex coordination task with distortion implemented as error enhancement.

In general, a subject participating in a previous experiment was asked to use one finger to produce specified forces or distances well within his or her voluntary maximum. In this experiment, on the other hand, subjects were asked to perform a coordinated two-finger task that required them to follow a specific distance profile, challenging their motor coordination to the point where they made involuntary errors. Distortion may have failed to improve the terminal performance of a subject because the subject was already performing at the limit of his or her ability.

6.4.3. Attentional Effects of Distortion

We believe that the primary effect of distortion in this experiment was to shift attention between the index finger and the thumb, with a corresponding decrement in performance resulting for the unattended digit. This is particularly clear for the ITB condition. When the index finger error was distorted (block 2), the thumb error tended to be higher than that of controls because subjects concentrated attention on the index finger. When distortion shifted from the index finger to the thumb (block 2 to block 3), ITB subjects showed an increase in the index finger error (countering the general learning trend) and a steep decrease in thumb error, in contrast to the trends for the control and TIB groups. In block 4, the error for both fingers was distorted for ITB subjects, and they divided their attention between the two fingers in much the same way as control subjects.

Although the TIB group more closely mimicked the controls, they too showed a steep decline in index error and counter-learning trend in thumb error when distortion shifted from the thumb error to the index finger error (block 2 to block 3). The lack of significant differences between TIB and control subjects can be explained by the fact that the trial absolute error for the thumb was significantly larger than that of the index finger for control subjects at the beginning of the experiment. This result is similar to those of Smeets and Brenner [112], who found that the path of the thumb was more variable than that of the index finger during a reaching and grasping task. This may be because the thumb has more degrees of freedom than the index finger. Both control and TIB subjects saw the thumb error as larger in block 2, and both groups focused attention on minimizing that error. The thumb error was not significantly different from the index finger error for control subjects at the end of the experiment. Thus, as the experiment proceeded, control subjects may have divided the attention more evenly between the index finger and the thumb, which may parallel the shift in attention of TIB subjects to the index finger error and then to both fingers.

6.4.4. Implications for Patient Work

This experiment has several implications for rehabilitation. First, we learned that it is possible to change the coordination pattern between two fingers for unimpaired subjects and we hope to do the same for patients with abnormal coordination skills. Though the specific task used here may not be task-oriented enough for rehabilitation, a similar task could be used to train patients to bring the index finger and thumb together in a coordinated way for pincer grasp. Also, this work demonstrates that we should provide opportunities for patients to practice the target task without visual feedback to prevent dependence on the visual display. It should be noted that for patients whose coordination difficulties reflect a selective decrement in a particular digit, the present paradigm could guide their attention to that locus and hence could direct rehabilitative action to the point of greatest need. Finally, in this experiment, distortion may have failed to improve terminal performance because subjects were performing at the limit of their ability. As discussed in Chapter 2, this may also be the case for some patients who choose to pursue challenging goals. On the other hand, some patients may be unable to motivate themselves in this way, and distortion may be beneficial for these patients.

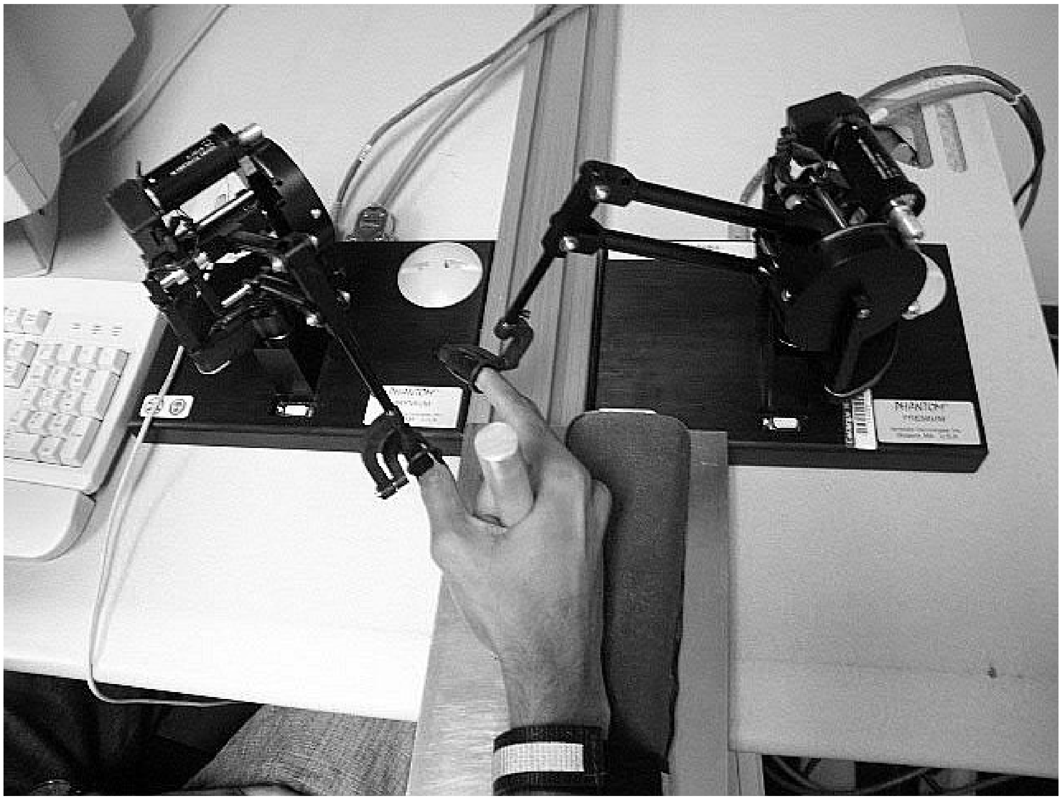


Figure 6.1. The experimental environment. The subject placed the thumb in a finger cuff attached to the left robot; the index finger was placed in a finger cuff attached to the right robot. The robots were used to simulate a virtual object between the index finger and thumb.

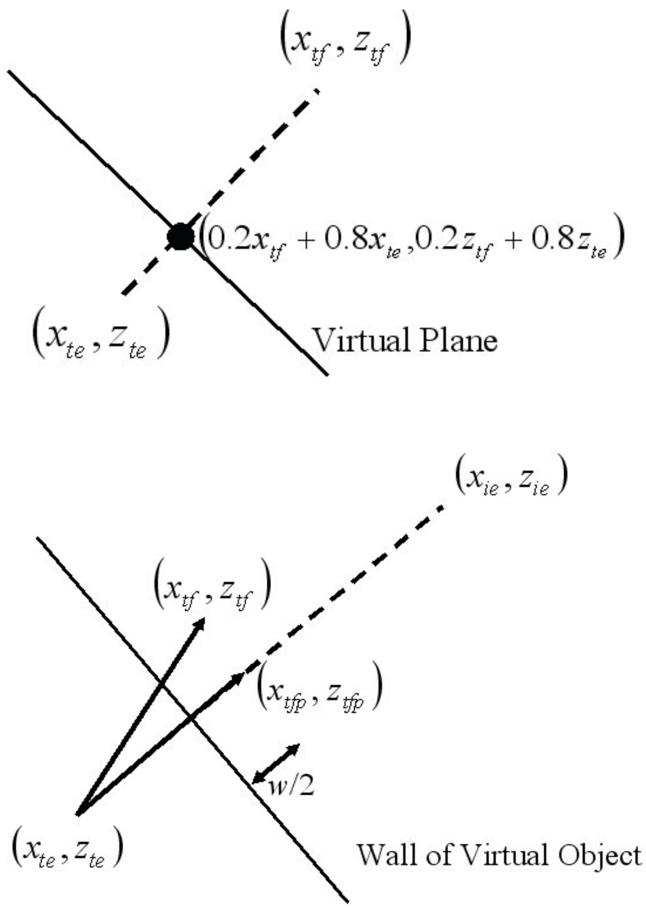


Figure 6.2. **(a)** The starting position of each trial is indicated to the subject by two virtual walls implemented on the robot, one for the index finger and one for the thumb. This schematic shows how the virtual wall for the thumb was determined. The wall was perpendicular to the line connecting the extended position of the thumb (x_{te}, z_{te}) and the flexed position of the thumb (x_{tf}, z_{tf}). The virtual wall was required to pass through the point shown. The virtual wall for the index finger was defined analogously. **(b)** During the experiment, the subject moved the index finger and the thumb between the virtual walls that defined the starting position and two virtual walls that defined an object. This schematic shows how the object wall closest to the thumb was determined. The plane defining this virtual wall was perpendicular to the line segment connecting (x_{te}, z_{te}) with (x_{ie}, z_{ie}), the extended position of the index finger. The wall passed through this line segment at a distance of $\frac{1}{2}w$ from (x_{tfp}, z_{tfp}), the projection of (x_{tf}, z_{tf}) onto the line segment.

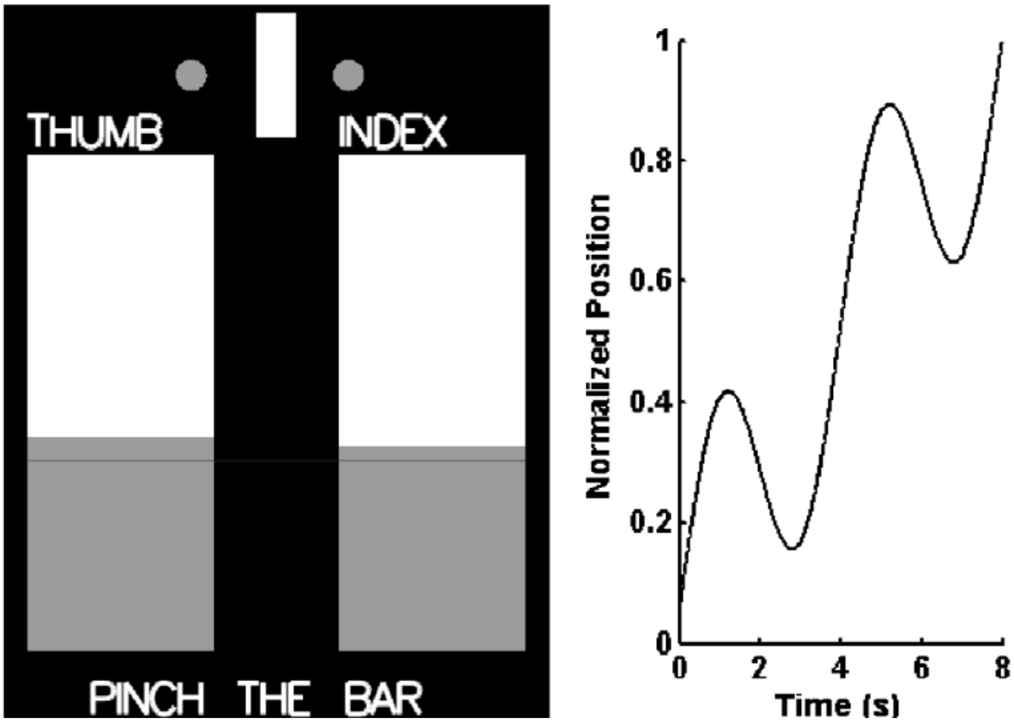


Figure 6.3. **(a)** The visual feedback seen by the subject during a trial of the experiment. The line crossing both bars moved the same way on each trial, and subjects tried to move the index finger and thumb in such a way that the tops of the shaded rectangles tracked the line. **(b)** The normalized position of the target line as a function of time. When the normalized position was equal to 0, the target line was at the bottom of the visual feedback bars; when the normalized position was 1, the target line was at the top of the feedback bars.

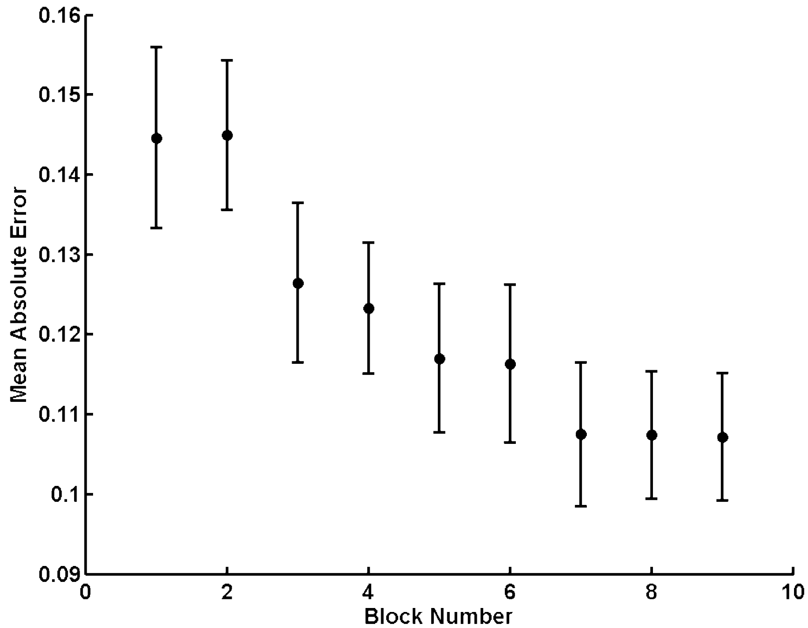


Figure 6.4. Learning over time. Data from the control subjects were divided into blocks of twenty trials. Block 1 corresponds to trials 21-40, block 2 to trials 41-60, etc. A decrease in the total mean absolute error as a function of block number occurred as the experiment progressed.

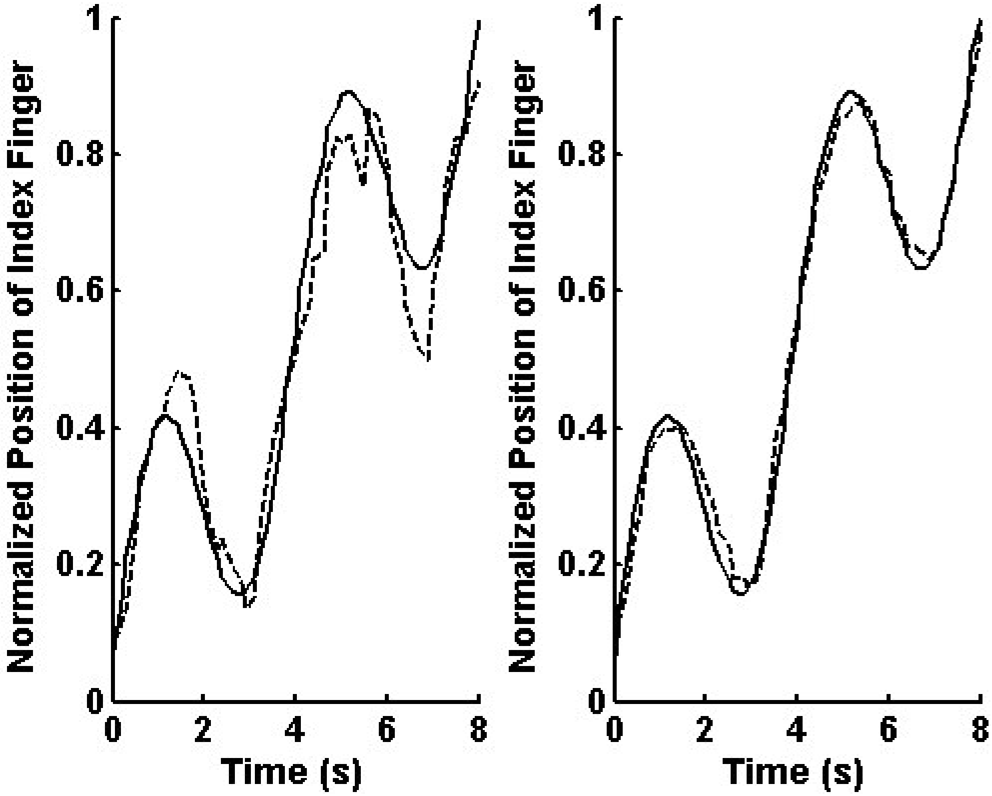


Figure 6.5. **(a)** Performance of a single subject on trial 21. The solid line represents the normalized position of the target line as a function of time, and the dashed line represents the normalized position of the index finger. **(b)** Performance of the same subject on trial 200. The subject reduced the error of the index finger relative to the target line, but the lag between the target line and the path of the index finger remained approximately the same.

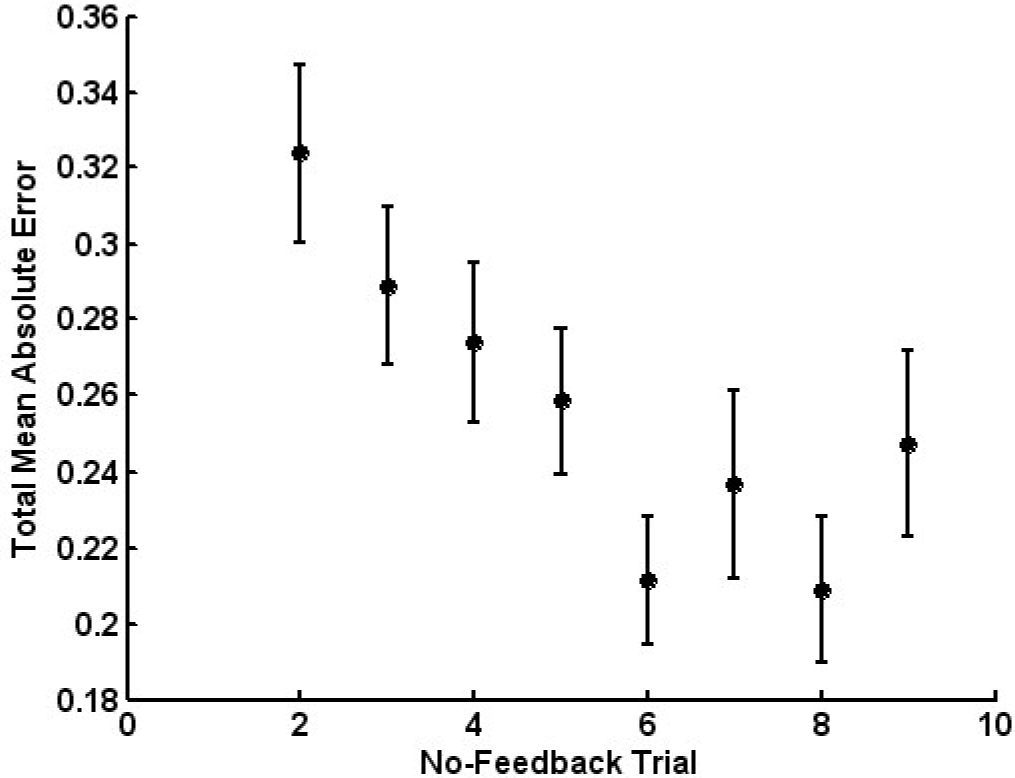


Figure 6.6. Results of the no-feedback trials for control subjects. Data from the first no-feedback trial were excluded because many subjects misunderstood the task (see text). The total mean absolute error was significantly larger for no-feedback trials than for the with-feedback trials, but a decrease in total absolute error and an increase in coordination did occur during the experiment for the no-feedback trials.

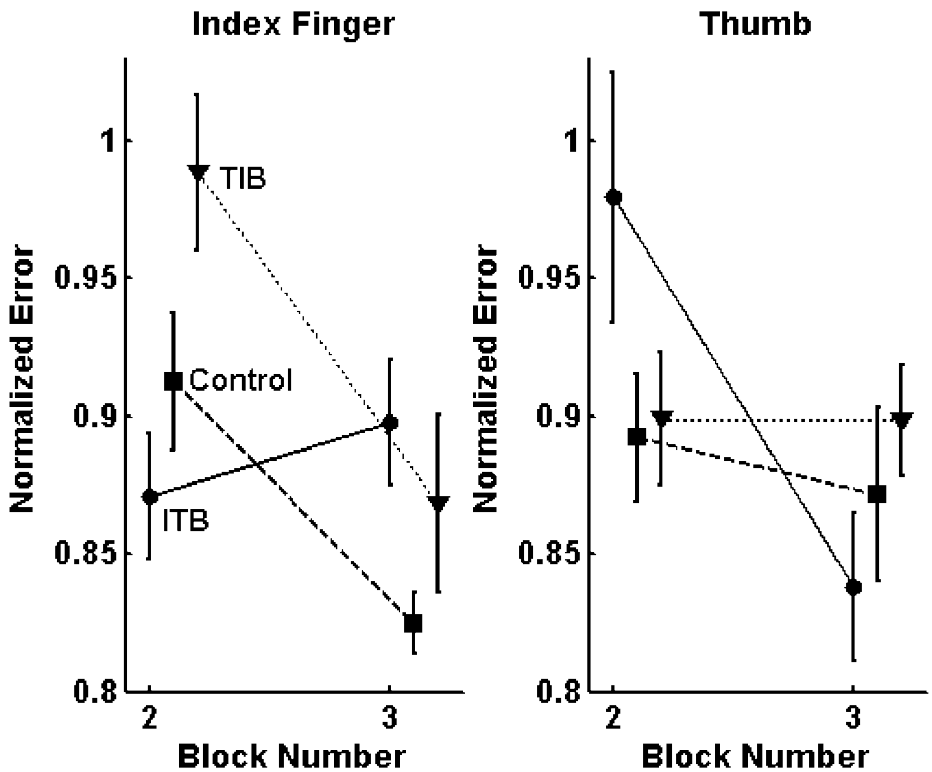


Figure 6.7 The effects of distortion on learning of the coordination task. Squares represent control subjects, circles represent subjects in the ITB condition, and triangles represent subjects in the TIB condition. **(a)** The normalized error for the index finger. The change from block 2 to block 3 for the ITB condition was significantly different from that of the controls. **(b)** The normalized error for the thumb. Again, the change from block 2 to block 3 for controls was significantly different from that for subjects in the ITB condition. No differences were found between control and TIB subjects.

Chapter 7. Distortion in a Rehabilitation Paradigm

7.1. Introduction

The experiments discussed in the previous chapters provided information on the limits and effects of visual distortion. However, before performing a lengthy experiment with multiple subjects to examine the effects of visual feedback manipulation in therapy, we conducted a preliminary test of hand rehabilitation for a single subject in a robotic environment. We designed and implemented a six-week rehabilitation paradigm to assess the potential of our robotic environment to improve hand function.

7.2. Subject Description

P1 was a 34-year-old female who sustained a closed head traumatic brain injury 8 years prior to her participation in the study. Her injury included injury to the brain stem and resulted in bilateral impairment. She required minimal assistance of another person in most daily activities. She could not speak clearly and used a communication device, but she was cognitively unimpaired. She could ambulate using a cane or walker. P1 performed the experiment with her left hand, which she had limited ability to move. She had no voluntary movement in her right hand.

7.3. Methods

Subject P1 participated in a six-week rehabilitation paradigm using the experimental environment described in Chapter 3. She received approximately two hours of rehabilitation per day, three days per week. During each rehabilitation session, she played a custom-designed Hangman game using the robot, choosing letters in an attempt to guess a word indicated by a number of blanks (Figure 7.1). A mapping between the alphabet and the force exerted by P1 allowed her to move between letters by flexing the finger against the virtual spring. P1 selected a letter by remaining within the force bounds of that letter for two seconds. That letter then disappeared from the alphabet listing. Either the chosen letter appeared in the word, or another segment of the Hangman appeared. P1 then extended her finger before selecting the next letter. The vowels were placed at the high-force end of the force mapping so that P1 had to regularly exert challenging forces. She won a round of the game by selecting all of the letters in the word before the seven segments of the Hangman figure appeared. To increase the speed at which P1 performed the rehabilitation task, she was given a clue for each word. A break was given every three words. To ensure that P1 exerted force only about the MCP joint, her wrist was immobilized by using a wrist splint and placing a restraint against the back of the wrist. Her index finger was placed in a cast of Aquaplast^{TM}-T to prevent flexing of the proximal and distal interphalangeal joints.

The distortion used in this experiment was similar to that described in Chapter 5. In the experiments with distortion that encouraged increased force production, the forces represented at the ends of the visual feedback bar were gradually increased. Here, the force limits of the mapping from force to letters were incremented systematically from day to day and within a single day. Figure 7.2(a) shows the limits of the mapping at the start of each rehabilitation session, as well as the maximum values attained by these limits during the session. During a rehabilitation session, the distortion of the force limits of the mapping was based on P1's force JND of 46.0%. For the second and third words of a set of three, each of the limits of the force mapping was increased by one fourth of the JND multiplied by the original value of the corresponding endpoint. The limits of the force mapping after a break were one fourth of the JND less than the limits before the break. This was done to minimize the subject's perception of the distortion, which might have been more obvious immediately after a break. This two up-one down approach is shown in Figure 7.2(b).

For each force mapping with limits F_{low} and F_{high}, the consonants corresponded to a range of force $\left(F_{low}, F_{low} + \frac{21}{26}\left(F_{high} - F_{low}\right)\right]$, and the vowels corresponded to a range of force $\left(F_{low} + \frac{21}{26}\left(F_{high} - F_{low}\right), F_{high}\right]$. Each subrange was itself subdivided using the number of remaining vowels or consonants to determine the interval of force corresponding to each letter. These intervals became larger as the subject chose letters, but the dividing point between the consonants and the vowels stayed the same throughout the word. This was done so that P1 could not select low-force consonants to reduce the amount of force necessary to reach the vowels. F_{low} was always greater than or equal to 0.5 N in order to provide P1 with a window of low-force in which she could rest between letters without accidentally selecting the first consonant.

At the beginning and the end of each of the eighteen rehabilitation sessions, P1's maximum MCP joint force was measured over a period of five seconds. These measurements allowed us to examine changes in P1's performance during the rehabilitation paradigm. Before and after undergoing the six-week rehabilitation program, P1 was assessed by an independent occupational therapist using the Ashworth Scale, the Action Research Arm Test (ARAT), and the Arm Motor Ability Test (AMAT). The Ashworth Scale was used to assess hypertone (excessive resistance to movement) in the thumb and fingers. The ARAT consists of a series of manipulation tasks performed with the hand and fingers. Each task is given a score from 0 to 3; the maximum possible score on the ARAT is 57. The AMAT consists of thirteen tasks, some of which are divided into component movements. Each subtask is timed and given a score from 0-5 for both functional ability and quality of movement. For more information on the content and validity of these tests, see [113-117] as well as Appendix B. The therapist also measured P1's range of passive and active motion for each finger, as well as the time she required to complete a nine-hole peg test. The distance between the tip of her index finger and the distal palmar crease (DPC) with the MCP joint and the distal and proximal interphalangeal joints fully flexed was also measured before and after rehabilitation.

7.4. Results

Figure 7.3 shows the increase in P1's maximum MCP joint force with time (first two days' data were not included because the wrist was not adequately restrained). This figure shows the joint-force data taken at the end of each rehabilitation session; the maximum forces measured at the beginning of each rehabilitation session could not be analyzed because some of the data were lost. A regression line for maximum force as a function of time is also shown in Figure 9. The slope of that regression line is 0.0951 N/day, which was significantly different from zero (p-value = 0.0013).

P1's clinical improvement was most apparent on the range of motion test. Her active range of motion for the MCP joint of the index finger increased from 40° to 68°, and her passive range of motion for that joint increased from 70° to 88°. In addition, an increase of 15° (from 75° to 90°) was seen for the passive range of the MCP joint of the middle finger. A 15° improvement (85° to 100°) was also seen for the active range of the proximal interphalangeal joint (PIP) of the middle finger, and the passive range of that joint increased from 90° to 100°.

The Ashworth scale measurements revealed that the hypertone in P1's thumb decreased from an Ashworth 2 to an Ashworth 1 during the course of the experiment. The hypertone of the fingers as a group remained constant at Ashworth 2, but the index finger considered alone changed from Ashworth 2 to Ashworth 1. The distance between the tip of the index finger and the DPC (distal palmar crease) with all finger joints flexed decreased from 0.5 cm to 0 cm. The time required for the nine-hole peg test was approximately the same before and after the intervention (2 min, 40 s before; 2 min, 39 s after), but after the intervention, the therapist noted much less use of the wrist to compensate for lack of finger movements.

An increase of 3 points was seen on P1's total score for the ARAT (47 points to 50 points), but this increase was not greater than the minimal clinically important difference, which we assume to be 5.7, 10% of the total range [117]. The minimal clinically important difference on a test of function is the smallest improvement on the test that indicates an important change in the patient's ability to carry out functional tasks. P1's mean functional ability score on the AMAT increased from 2.70 to 2.85, while her mean quality of movement score increased from 2.63 to 2.78. We take the minimal clinically important difference for the AMAT to be 0.5, 10% of the possible range for these means, which means that these increases were not clinically important.

7.5. Discussion

Throughout this experiment, we measured improvement in P1's performance of the target task, production of force about the MCP joint of the index finger. More importantly, an independent occupational therapist measured improvement using a variety of clinical measurements and scales. The largest improvements were seen for low-level motor variables, such as range of motion and hypertone. Interestingly, improvements were seen not only for the index finger, but also for the middle finger and thumb, which were not directly involved in the intervention. It was noticed that P1 often moved her other fingers and her thumb in concert with her index finger during the rehabilitation sessions. We suspect that the secondary improvements seen for the thumb and the middle finger were due to these movements.

While some improvement was seen on clinical scales of function, these improvements were not clinically important. This lack of important functional improvement was expected due to the nature of the intervention. The AMAT measures the ability to perform functional activities like cutting meat and putting on a sweater. Since our intervention addressed only the MCP joint of the index finger, it would be unreasonable to expect a large change on this clinical scale. The ARAT measures the ability to pick up small objects, but many joints besides the index MCP are important for this type of task. In addition, the test does not always specify which fingers must be used, and P1 performed most of the test using the middle finger and the thumb. The goal of this preliminary experiment was to determine whether practice of an activity in a robotic environment could cause a clinically noticeable improvement in hand capabilities, not to address the details involved in the rehabilitation of fine motor control.

P1 practiced the force-production task primarily within the range of her ability, the domain in which we expect distortion to be effective. Distortion was used in this experiment, but because there were no control subjects, no conclusions about the use of distortion can be drawn without further experiments. While we changed some characteristics of this experiment before working with other motor-impaired subjects, the results of this preliminary test were positive overall. The Hangman game maintained P1's interest in the experiment throughout the six-week period. In addition, P1 did not seem to notice the distortion, which hints that we may be able to use distortion over a long period of time without a subject's knowledge. Most importantly, this preliminary experiment was one of the first applications of robotic therapy to the hand, and this protocol resulted in noticeable physical gains for P1, as discussed above.

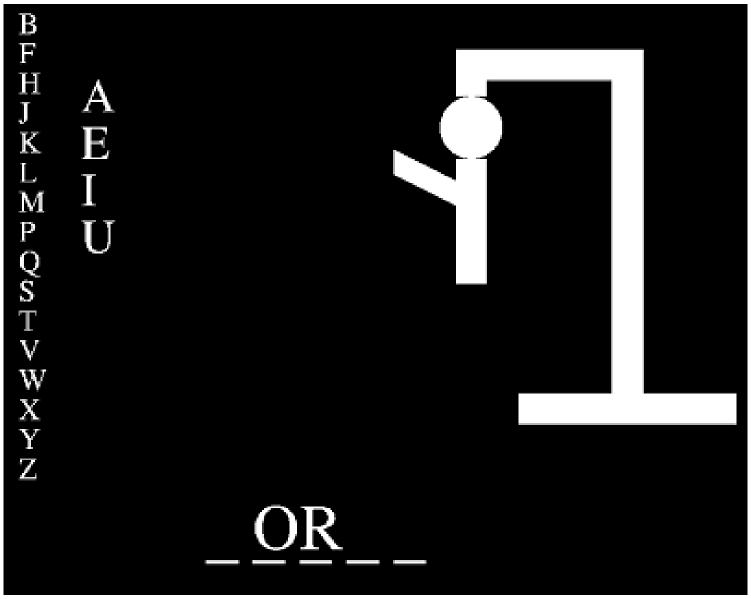
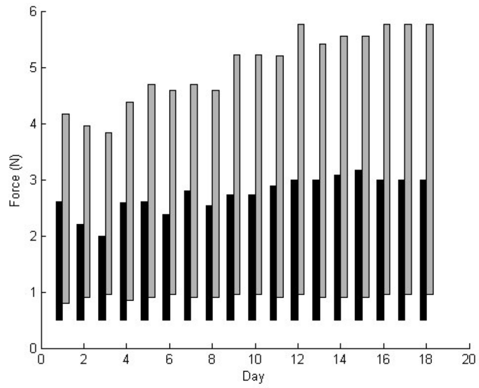


Figure 7.1. The visual display used for the Hangman game. The force exerted by subject P1 was mapped to a letter of the alphabet, with vowels requiring more force. P1 chose letters in an attempt to complete the word indicated by the blanks before all seven segments of the Hangman appeared.



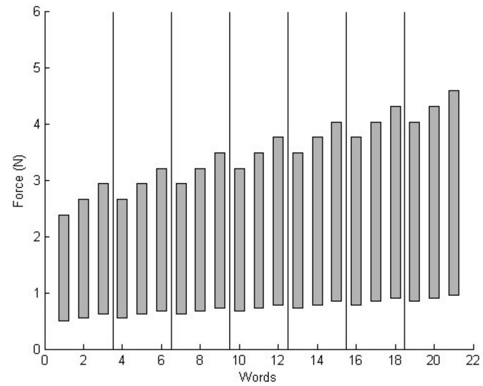


Figure 7.2. The force-alphabet mapping for the Hangman game played by P1changed from day to day and was distorted during each rehabilitation session to encourage P1 to produce more force. **(a)** The range of force mapped to the alphabet at the beginning of each rehabilitation session (black bars), and well as the maximum limits this range attained during the rehabilitation session (gray bars). **(b)** The range of force mapped to the alphabet for every word of the sixth rehabilitation session. The black lines denote breaks. The force limits of the alphabet mapping increased between breaks; the force limits decreased for the word immediately following a break.

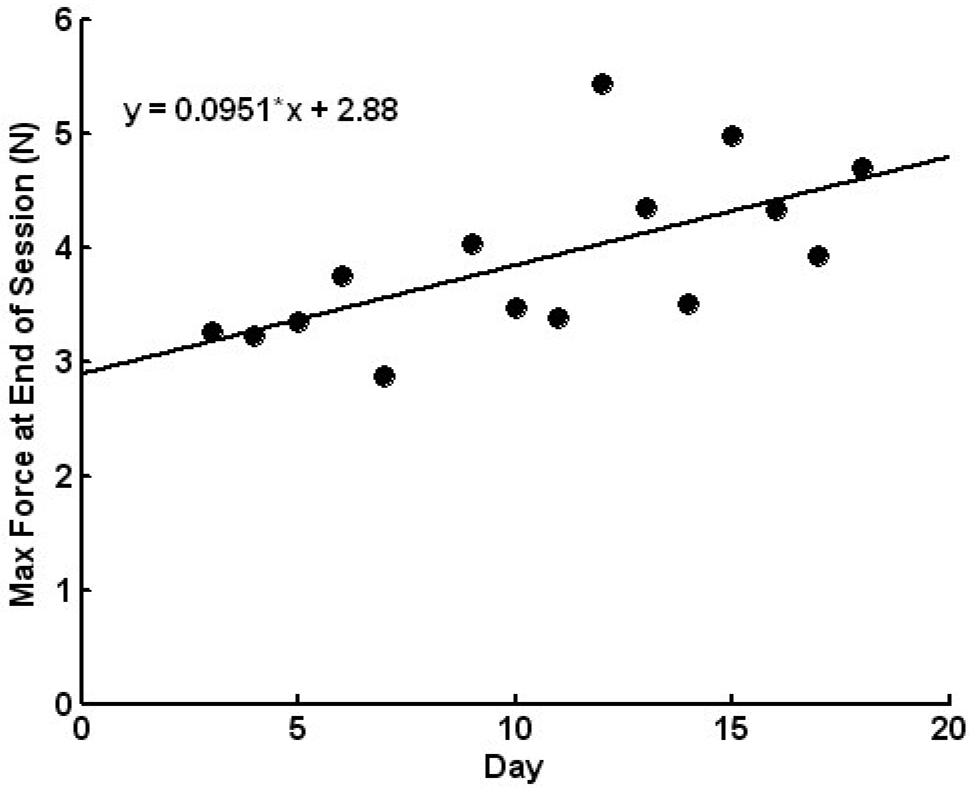


Figure 7.3. Subject P1's maximum force about the MCP joint increased throughout the rehabilitation period. The positive slope of this regression line was significantly different from zero.

Chapter 8. A Robotic System to Test the Effects of Visual Feedback Manipulation on Performance in Rehabilitation

8.1. Introduction

In this chapter, we discuss the rehabilitation protocols that we designed to test whether visual feedback manipulation during rehabilitation results in performance better than that observed during rehabilitation sessions in which the visual feedback is not manipulated. This work built upon the foundational work that we conducted to investigate the limits and effects of visual distortion. However, the psychological literature, described in Chapter 2, indicates that visual progression might also prove useful for rehabilitation. In addition, visual progression is the appropriate control for visual distortion. Subjects experiencing visual progression can be asked to advance at the same rate as subjects in the visual distortion condition; then, the only difference between the two groups is that the visual distortion group is not aware that that they are being challenged to improve their performance.

Our rehabilitation paradigm was designed for stroke and TBI subjects who had some active finger flexion and some thumb movement. The inclusion/exclusion criteria for participants can be found in Appendix A. Our goal was for these subjects to relearn a functional pincer grasp. The pincer grasp, also known as a tip grasp, is a pinching grasp necessary for picking up and manipulating small objects, including pills and finger foods [50]. It is also an important component of the grip used when writing. The pincer grasp can occur between the pads of the index finger and the thumb, or between their tips (neat pincer grasp), as shown in Figure 8.1. Pinching requires the subject to flex the index finger about the metacarpophalangeal (MCP) and the proximal interphalangeal (PIP) joints and to bring the thumb into a position opposing the index finger by medially rotating and flexing the thumb about the carpometacarpal (CMC) and metacarpophalangeal (MCP) joints [118] (Figure 8.2). Flexion of the distal interphalangeal (DIP) joint of the index finger and the interphalangeal (IP) joint of the thumb can also be involved when the pinching grasp is a tip-to-tip pinch rather than a pad-to-pad pinch. In our rehabilitation paradigm, subjects practiced bringing the index finger and the thumb together in a coordinated way to form a pincer grasp and extending the fingers in a coordinated way to release the grasp. This was done in the context of a Hangman game similar to that described in the previous chapter.

8.2. Experimental Environment

The robotic environment for this task included two PHANTOM™ Premium 1.0 robots and is shown in Figure 8.3(a). The environment is configured for the right hand in Figure 8.3, but it was possible to arrange it to work with the left hand as well. The index finger was coupled to one robot, and the thumb was coupled to the other. The custom-made finger cuff used to link each finger to its robot had 3 passive DOF and consisted of a tube of elastic attached to curved piece of metal bent about the tip of the finger (Figure 8.3(b)). This cuff kept the finger firmly coupled to the robot while allowing the subject to bend the joints of the finger. The finger cuff used in the experiments described in Chapters 4,5, and 7, on the other hand, consisted of a metal tube that extended the length of the index finger, immobilizing the proximal and distal interphalangeal joints. The previous finger cuff distributed the force exerted by the robot across a greater portion of the finger; thus, the same force experienced with both cuffs might be perceived as less with the previous cuff. However, this should not affect our results, as the rehabilitation paradigm described in this chapter focused on range of motion rather than force. The subject wrapped all fingers except the index finger and thumb around a metal post. To ensure that the subject used only the fingers to pinch and extend, the forearm was strapped to the table, and the subject wore a wrist brace. An additional restraint was placed against the dorsal side of the wrist to further discourage the subject from bending the wrist.

Performance in grasp formation and release was measured by the distance between the end of the index finger and the end of the thumb. We denote this measurement by d_{it}. The subject tried to minimize d_{it} when pinching and maximize it when extending the fingers to release. It should be noted that each robot tracked the approximate center of the finger coupled to it. Thus, d_{it} actually represents the distance between the fingers plus half the width of each finger.

8.3. Calibration Program

Each rehabilitation session began with a short program to measure three parameters used to calibrate the Hangman game to the subject's range of motion and degree of tremor. First, the subject was asked to pinch the index finger and the thumb together as much as possible and to hold that position for five seconds. During this period, the computer measured an average minimum value for d, which we will call d_{min}. The program then measured a maximum value d_{max} for d_{it} after instructing the subject to extend the index finger and thumb as much as possible. The value of d_{max} was also averaged over a period of five seconds.[1]

The next part of the calibration program was designed to calculate *n*, which represents the number of letters presented to the subject as choices in the Hangman game. To select a letter in the Hangman game, the subject had to remain within an interval of d_{it} values corresponding to that letter, and we found that some subjects were unable to select a letter when the range of motion was subdivided among all 26 letters. In this part of the calibration program, the subject saw a screen containing a white box bisected by a blue line (Figure 8.4). The blue line represented a distance of $\frac{(d_{max} - d_{min})}{2}$, and the white box indicated a window of 5 mm on either side of this distance. d_{it} was shown on the screen as a green ball. The ball moved to the left as d_{it} decreased and to the right as d_{it} increased. The subject was asked to keep the green ball as close as possible to the blue line. One second after the green ball entered the white box, the program began to measure d_{it}. It recorded d_{it} for 3 seconds and then calculated the standard deviation *s* of the distance between the thumb and index finger. The following function was then used to calculate *n*:

[1] For the two patients whose data is presented in Chapters 9 and 10, the maximum and minimum values of d_{it} were not averaged over a five second period. They were simply the maximum and minimum values reached by each patient as she was asked to pinch the index finger and thumb together and move them apart. We later decided that having the patient hold the maximum and minimum for a short period of time would yield a more stable measurement.

$$n = \text{floor}\left(\frac{d_{max} - d_{min}}{ks}\right). \qquad (1)$$

In this equation, *k* is a constant that we set to 4. We came up with this equation for *n* by assuming that the subject would be able to stay within ± 2 standard deviations of the mean distance value corresponding to a given letter. This equation assumes that *s* is a constant throughout the subject's range of motion. This is not necessarily true, but the value for *n* obtained from this approximation seemed to work fine in practice.

8.4. Hangman Game

After completing the calibration program, which took less than a minute, the subject began to practice pinching and extending in the context of the Hangman game. The subject saw a screen like that in Figure 8.5(a), with some number of blanks indicating the number of letters in the word to be guessed. The *n* letters from which the subject chose while trying to guess the word were shown across the top of the screen. The subject selected different letters by changing the distance between the index finger and thumb; we denote the range of distances mapped to the list of letters by (d_{mapmin}, d_{mapmax}). This list was subdivided so that a subinterval corresponded to each of the *n* letters, as explained below. When the subject was within a subinterval corresponding to a given letter, that letter was highlighted. The subject moved to the left by pinching the fingers together and to the right by extending the fingers. The subject selected a letter by remaining within that letter's subinterval for 2 seconds. The letter then appeared in the target word, or a new segment of the Hangman appeared (Figure 8.5(b)). After a letter was selected, it appeared gray in the letter list and could be selected again until the subject moved on to a new target word. A round of Hangman ended when the user selected all of the letters in the target word (winning the round) or when all seven segments of the Hangman appeared (losing the round) (Figure 8.6).

The list from which the subject selected letters always contained the five vowels, 'A,' 'E,' 'I,' 'O,' and 'U.' These vowels could be shown on either the right or left side of the screen (or both). For instance, if the subject could pinch the fingers together but had difficulty extending the fingers, we places the vowels on the right so that the subject had to practice extension to play the game successfully. The consonants in the list of n letters were shown in alphabetical order. This list, of course, contained all the consonants needed to spell the target word. Other consonants were chosen randomly by the computer.

A vertical purple line appeared in the center of the row of possible letters. This line represented a window about a neutral distance defined by the data measured in the calibration program as $d_{neut} = \frac{(d_{max} - d_{min})}{2}$. The window represented by the purple line was defined by $d_{neut} \pm w$, where w is a constant equal to 0.75 mm (Figure 8.7). After selecting a letter, the subject had to return to this window before he or she was allowed to select another. The program signaled this to the user by highlighting letters in green when a letter could be selected and in red when the subject had to return to the neutral window (letters that had already been selected were always highlighted in a shade of gray lighter than that in which they otherwise appeared). The requirement of returning to the neutral window between letters was implemented to increase the number of times the subject pinched and extended the fingers while playing the Hangman game. Originally, we also thought that subjects could use this window as an opportunity to pause and rest without selecting a letter. Subjects did not use the window in this way, probably due to its small size. Instead, subjects chose to rest on letters that they had already selected.

As mentioned above, the range (d_{mapmin}, d_{mapmax}) was subdivided among the number of letters from which the subject chose (Figure 8.7). This range was divided by the neutral window into $(d_{mapmin}, d_{neut} - w)$ and $(d_{neut} + w, d_{mapmax})$. The interval $(d_{mapmin}, d_{neut} - w)$ was subdivided into $n_1 = \text{floor}\left(\frac{n}{2}\right)$ subintervals corresponding to the leftmost n_1 letters in the list. The interval $(d_{neut} + w, d_{mapmax})$ was subdivided into $n_2 = n - n_1$ subintervals corresponding to the remaining letters. When $d_{it} > d_{mapmax}$, the rightmost letter was highlighted, and when $d_{it} < d_{mapmin}$, the leftmost letter was highlighted.

As the subject flexed and extended the finger, he or she had to resist a small amount of force. This force was included to provide an additional source of feedback to the subject's nervous system with the goal of encouraging reorganization in the brain. This force was particularly important in extension, because resistance to extension has been shown to improve the active range of motion of stroke patients more than resistance to grasp or ballistic extension [119]. The magnitude of the force was determined by $\sqrt{|d_{it} - d_{neut}|}$. To obtain the magnitude of the force exerted against each finger, this factor was multiplied by a scaling factor that differed for flexion and extension (in general, subjects could resist more force in flexion than in extension). In flexion, the scaling factor was equal to $\frac{F_f}{\sqrt{d_{neut} - d_{mapmin}}}$. In extension, the scaling factor was equal to $\frac{F_e}{\sqrt{d_{mapmax} - d_{neut}}}$. F_f and F_e are constants. Either F_f or F_e (or both) could be set equal to zero if the subject was unable to play the Hangman game with force feedback. Otherwise, F_f was set equal to 2 N, and F_e was set to 1 N. The direction of the force was determined differently for different subjects and is discussed in the appropriate case study.

In designing this robotic rehabilitation protocol, we addressed the motor learning issues discussed in Chapters 2 and 6. Task variation was accomplished by the fact that different letters required differing amounts of grasp and release. After each word, the subject was asked to pinch and extend the fingers in the absence of feedback. The subject was asked to hold the minimum pinch and maximum extension for 2 seconds each.[2] The goal of these trials was to encourage transfer of tip grasp skills to ADL activities by ensuring that the subject could perform the target task without feedback from the computer. In addition, after sessions 4, 7, 10, 13, and 16, the subject was asked to complete the Grip and Pinch subtests of the ARAT (see Appendix B). These subtests allowed the subject to practice picking up various objects using the skills practiced in the Hangman game. The administration of this subtest of the ARAT also gave us the opportunity to examine the effects of the robotic protocol on real-world performance.

[2] The minimum pinch and maximum extension positions were not held for 2 seconds each for the subjects described in Chapters 9 and 10.

8.5. Implementation of Visual Distortion and Progression

Each subject who participated in the experiment received 18 rehabilitation sessions, 3 per week for 6 weeks. Each session lasted 90 minutes. Breaks lasting at least 3 minutes were given every three words. In addition, the program could be stopped and restarted anywhere, so the subject was free to take additional breaks as necessary. In our rehabilitation protocol, each subject received either visual distortion or visual progression in odd-numbered sessions. There was no visual feedback manipulation in even-numbered sessions. Visual distortion was implemented in this robotic environment by manipulating the range of distance (d_{mapmin}, d_{mapmax}) that was mapped to the list of letters from which the subject selected. For each rehabilitation session, the initial values of d_{mapmin} and d_{mapmax} were based on the values d_{min} and d_{max} measured by the calibration program at the beginning of that session. When the session began, d_{mapmin} was equal to $1.2d_{min}$ and d_{mapmax} was equal to $0.8d_{max}$. These proportions of the maximum and minimum distances were chosen as limits that normal subjects could reach comfortably and repeatedly. During sessions in which no distortion was present, d_{mapmin} and d_{mapmax} remained constant throughout the session.

During sessions with distortion, d_{mapmin} and d_{mapmax} were made gradually smaller and larger, respectively. This distortion encourages subjects to move the index finger and thumb closer together when pinching and further apart when extending. The subject had to gradually improve his or her performance to reach all of the letters in the list. The amount of distortion experienced during a rehabilitation session was determined by the minimum and maximum distances between the index finger and thumb measured at or before the beginning of the first rehabilitation session, which we will call $d_{initmin}$ and $d_{initmax}$. These distances were compared to estimates of what normal maximum and minimum values for the subject would be. The theoretical minimum $d_{theomin}$ for each subject was 15 mm, which was approximately the distance measured by the robot between the fingertips in a pad-to-pad pinch (this value was nonzero because the system measured the distance between the centers of the fingers). The maximum theoretical value $d_{theomax}$ for each subject was obtained either using the unimpaired hand, or, in the case of bilateral impairment, from the geometry of the hand, assuming a maximum angle of 130° between the index finger and thumb.

To calculate how much d_{mapmax} should be distorted within a single session, we assumed that the subject always followed the distortion, and that gains were maintained from session to session. From these assumptions, the maximum extension reached by the subjects during the *m*th distortion session is given by $d_{initmax}x^{m}$, where x represents the maximum distortion experienced in the session. We then calculated $x_{ext} = \left(\frac{d_{theomax}}{d_{initmax}}\right)^{\frac{1}{5}}$, the distortion that would be required in each session for the subject to reach his or her theoretical maximum extension in 5 distortion sessions. We calculated x_{ext} based on 5 distortion sessions instead of 9 to account for the fact that the subject did not necessarily follow the distortion or maintain gains from session to session. While we recognize that one or both of our assumptions may not be completely true, we used this method as a principled way of choosing the degree of distortion that was experienced by each subject.

The variable x_{ext} represents the total amount of distortion experienced during a single rehabilitation session, but this total was reached via a series of steps, similar to the method described in the previous chapter. Words were presented in sets of three, with a break given between each set. For the second and third words of a set of three, d_{mapmax} was increased by one distortion step. After the break, d_{mapmax} was decreased by one step. This method was followed until the fourteenth word was reached. For the fourteenth word and all subsequent words, d_{mapmax} was equal to $x_{ext}d_{max}$ (d_{mapmax} was also equal to $x_{ext}d_{max}$ for word 12, but the break between word 12 and word 13 decreased d_{mapmax} for word 13). Thus, d_{mapmax} went from $0.8d_{max}$ to $x_{ext}d_{max}$ in 5 distortion steps. We made each step of equal size. Thus, d_{mapmax} was equal to $0.8d_{max}$ for the first word, $\left(0.8+\frac{x_{ext}-0.8}{5}\right)d_{max}$ for the second word, etc. There were 24 words for each rehabilitation session, but we expected that the subject would not reach all of them. We designed the environment such that the subject first reached the maximum distortion at the twelfth word because preliminary results indicated that 12 words was a reasonable number to expect a subject to complete in a 90-minute session.

The distortion applied to d_{mapmin} was analogous to the described above for d_{mapmax}. Rehabilitation sessions involving visual progression were identical to distortion sessions except that a scale of distance in millimeters was shown along the bottom of the list of letters from which the subject chose (Figure 8.8). Each multiple of 10 mm between d_{mapmin} and d_{mapmax} was shown. When d_{mapmin} and d_{mapmax} were made more extreme, this scale was altered to alert the subject to the change. The only difference between the distortion protocol and the progression protocol was that visual progression subjects knew that the display is changing. Each protocol provided a control for the other. For example, if improvements were seen for visual distortion subjects but not for visual progression subjects, then imperceptible distortion could be concluded to have caused these improvements.

As visual progression and distortion encouraged subjects to improve their performance, we had to ensure that no subject was required to move beyond his or her physical abilities. To avoid the possibility of harm to the subject, he or she always had the option of asking the experimenter to contract the bounds of the d_{it} range by one distortion step. This was done by pressing the 'x' key on the keyboard (the subject could do this without the experimenter's help if he or she had a unilateral impairment). The implementation of this feature also allowed us to examine the effects of distortion and visual progression on the level at which subjects chose to perform.

8.6. Measurement of Effects

The effects of distortion and visible progression were explored in several ways in our paradigm. Performance in the target task of pincer grasp and release with the robots was measured by the values for d_{min} and d_{max} calculated by the calibration program at the beginning of each session; for comparison purposes, we also ran the calibration program at the end of each rehabilitation session.[3] We also considered the values for d_{it} measured while the subject is playing the Hangman game, as well as the number of times the subject chose to contract the bounds of the d_{it} range. To allow us to analyze Hangman data taken with the same target answer, the word 'DAISY' was included as words 1, 7, 12, and 24 in each rehabilitation session. When the target answer was 'DAISY,' the word was shown below the answer blanks on the screen; the subject simply had to select the letters indicated. As mentioned above, the Grip and Pinch subtests of the ARAT were administered once a week during the six-week protocol.

Before and after the six-week period, each subject was evaluated by an occupational therapist. This therapist measured the passive and active ranges of motion of all joints in the thumb and fingers and measured the hypertone of the joints of the arm and fingers using the Ashworth Scale. She then administered the AMAT and the ARAT (see Appendix B). Of the tests administered by the therapist, the ARAT was the one most closely related to the tasks practiced in our robotic environment. In addition to the therapy tests, a questionnaire was administered before the six-week protocol to assess how the subject responded to challenging activities. It included the following questions:

> People seem to differ in how they approach challenging activities. Some people prefer to stick to what they know they can do, while others like to jump to something more difficult than they can do right now. Let's call the first group the 'sure things' and the second group the 'bring-it-ons.' If we put these on a 1-10 scale, representing the liking for challenge, where the sure things are 0 and the bring-it-ons are 10, where would you put yourself?
>
> People also differ in what they like to do after accomplishing a difficult task. Some people prefer to practice that task for a while to make sure they can do it, while others want to move on to a new, more difficult task. If 0 is continuing to practice, and 10 is immediately moving on to a new task, where would you put yourself?

[3] The calibration program was not run at the end of each session for the subjects whose data is presented in Chapters 9 and 10.

As discussed in 2.1, a full examination of all psychological variables that may influence the effect of distortion or visual progression was beyond the scope of this work. The questionnaire was administered just to obtain a preliminary idea of the variables that may affect the use of visual distortion and progression in rehabilitation and how these quantities varied in our subject population.

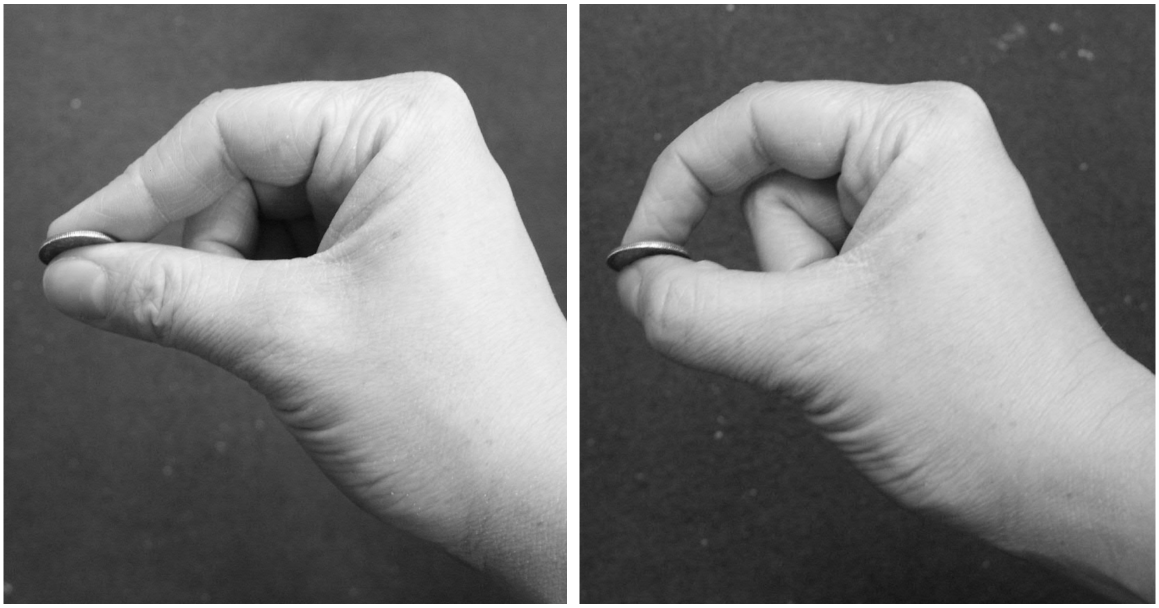


Figure 8.1. The pincer grasp, used for picking up small objects. The goal of our rehabilitation paradigm was to have subjects relearn a functional pincer grasp. **(a)** Pad-to-pad pincer grasp. **(b)** Tip-to-tip pincer grasp (neat pincer grasp).

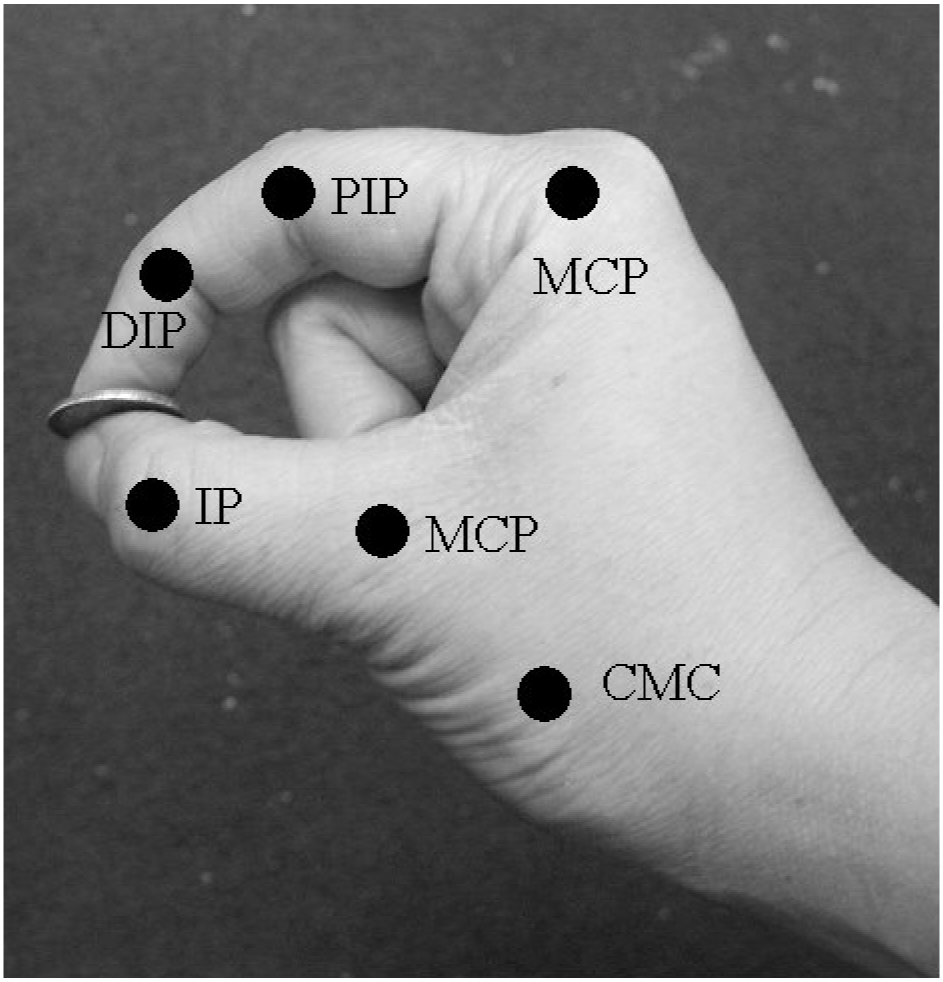


Figure 8.2. The joints of the index finger and thumb that are involved in a pincer grasp. The pincer grasp shown here is a tip-to-tip pincer grasp, also known as a neat pincer grasp. Each joint of the index finger and the thumb that is involved in this grasp is denoted by a labeled black dot. The abbreviations used are as follows: CMC = carpometacarpal joint, MCP = metacarpophalangeal joint, IP = interphalangeal joint, PIP = proximal interphalangeal joint, and DIP = distal interphalangeal joint.

Figure 8.3. **(a)** The robotic environment with which the subject interacted in our rehabilitation protocol. Two PHANTOM^{TM} robots were used, one coupled to the index finger and one to the thumb. The subject grasped the metal pole with the remaining fingers to keep the hand stationary. The subject's wrist was restrained. **(b)** A close-up of the custom-made finger cuff used to couple the index finger to the robot.

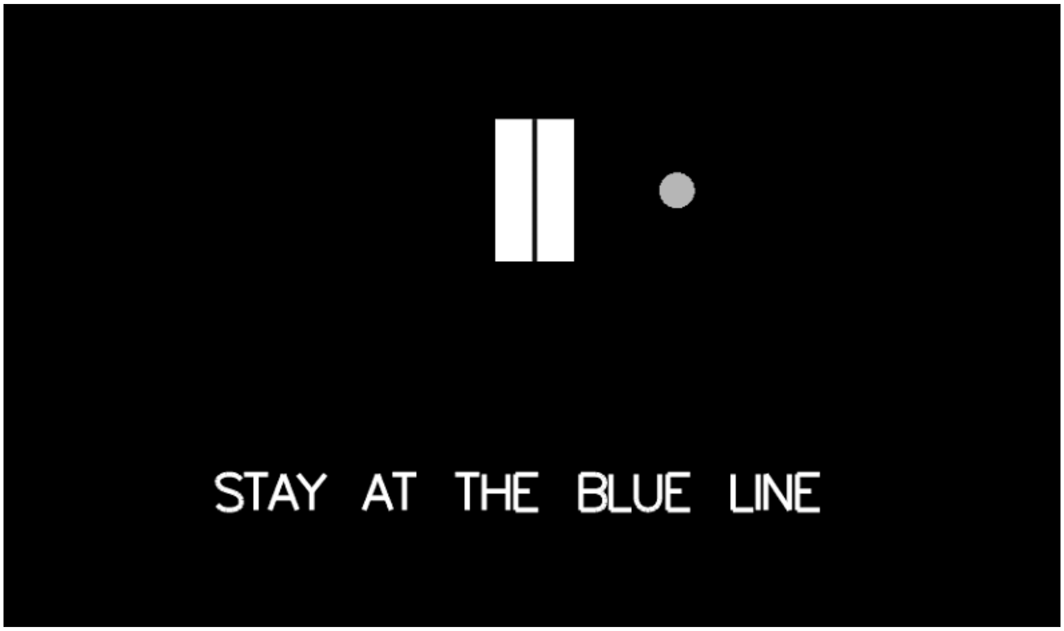


Figure 8.4. An example of the screen shown during the final part of the calibration program run at the beginning of each rehabilitation session. The ball represented the distance between the subject's index finger and thumb. The subject brought the fingers closer together to move the ball to the left and separated them to move it to the right. The subject was instructed to keep the ball as close as possible to the blue line centered in the white box. This part of the calibration program measured the tremor in the subject's movements as he or she tried to maintain a constant distance between the fingers.

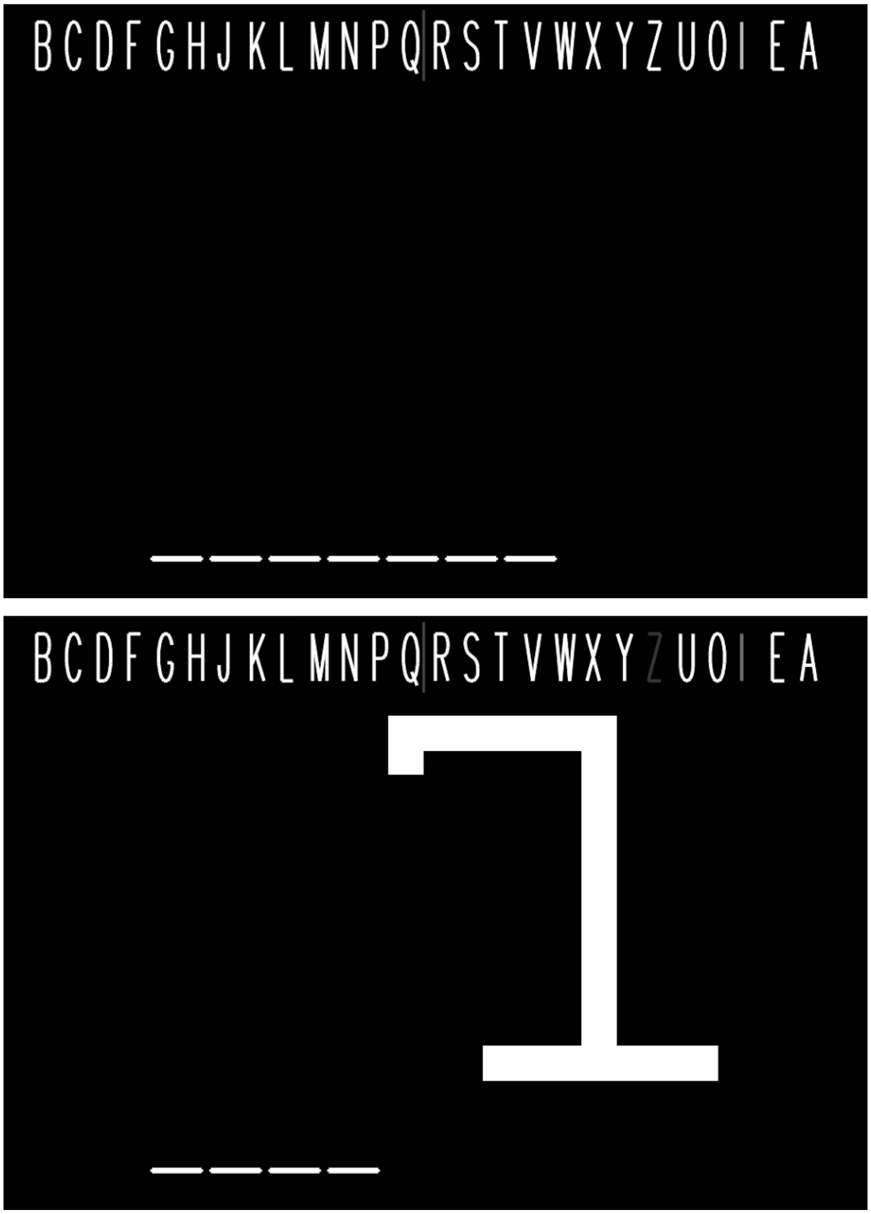
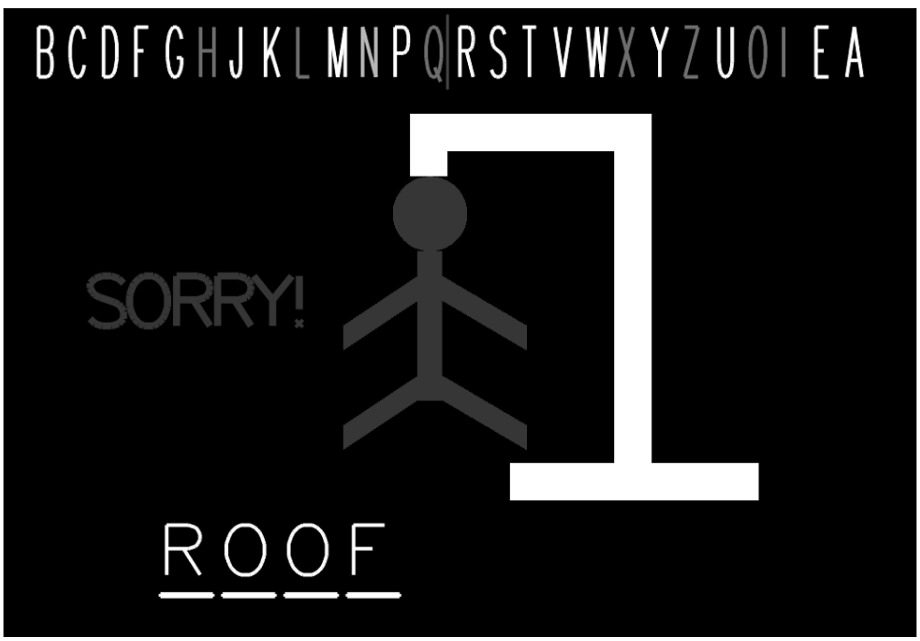


Figure 8.5. (a) An example of the appearance of the screen at the beginning of a new word in the version of the Hangman game seen by subjects in the distortion condition. The list from which the subject chose letters was shown horizontally across the top of the screen. The subject moved between letters by changing the distance between the index finger and thumb. The bar in the center of the list of letters denoted the neutral window to which the subject returned after selecting a letter. The target word was denoted by the blanks at the bottom of the screen. **(b)** The appearance of the screen after the subject selected an incorrect letter. One segment of the Hangman is shown.



(b)

Figure 8.6. (a) An example of the screen when the subject won a game of Hangman. **(b)** An example of the screen when the subject lost a game of Hangman.

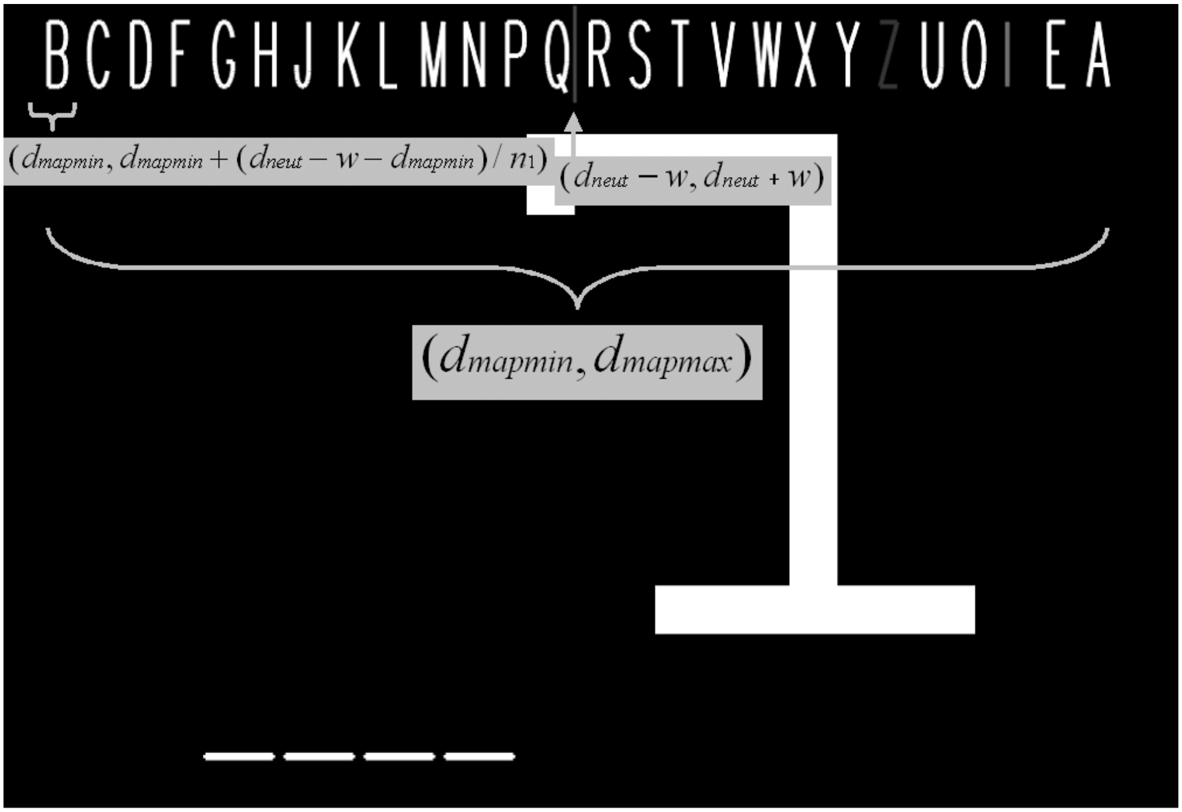


Figure 8.7. During the Hangman game, the distance between the subject's index finger and thumb was mapped to the horizontal list of letters shown across the top of the screen. (d_{mapmin}, d_{mapmax}) represents the range of distance mapped to the list of letters. This range was subdivided among the letters shown and the neutral window, represented by a line. The subject had to return to the neutral window after selecting each letter. The subrange of distance that defined this window was $d_{neut} \pm w$ (see text). As an example, the subrange of distance corresponding to the letter 'B' in the above figure was $\left(d_{mapmin}, d_{mapmin} + \dfrac{d_{neut} - w - d_{mapmin}}{n_1}\right)$, where n_1 is the number of letters to the left of the line representing the neutral window.

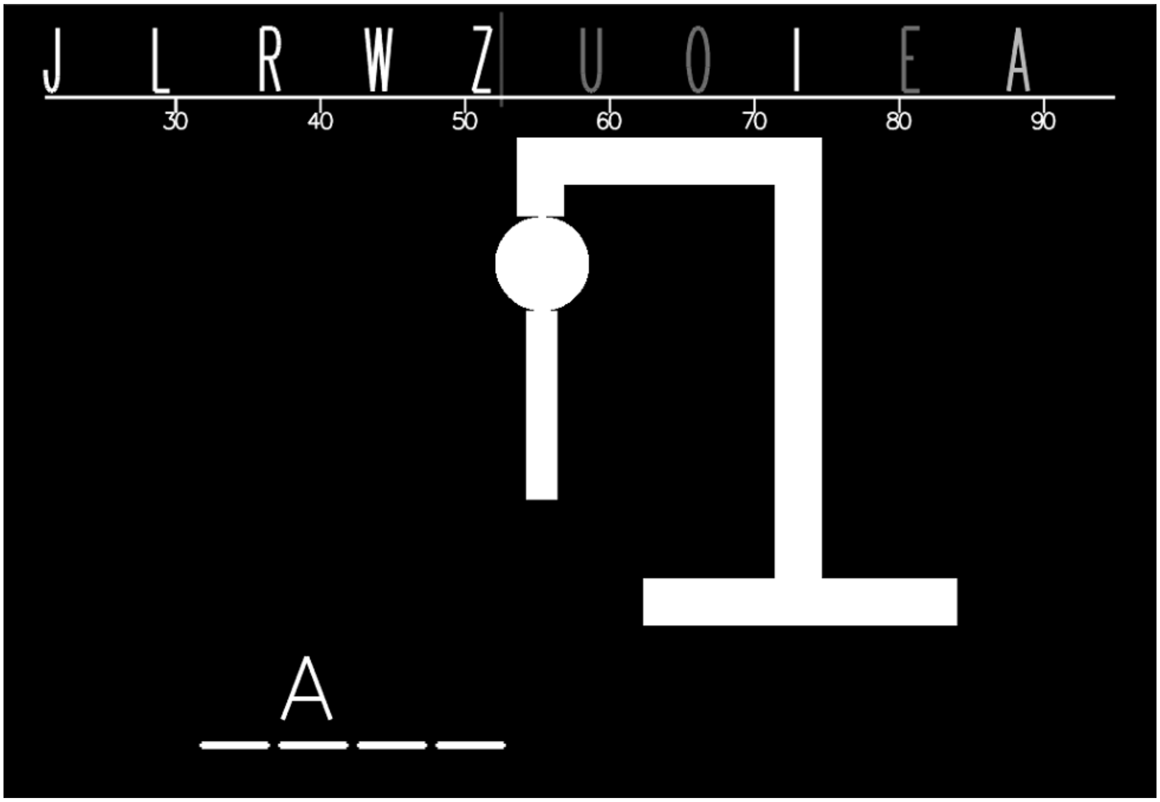


Figure 8.8. An example of the appearance of the Hangman screen for subjects in the visual progression condition. The Hangman game was identical as to that used in the distortion condition except that progression subjects saw a distance scale (in mm) below the horizontal list of letters.

Chapter 9. Case Study: Subject P3

9.1. Subject Description

Subject P3 was a female, 25 years old at the beginning of the study, who suffered a traumatic brain injury 2 years prior to her participation. Her injury included diffuse axonal injury and possibly anoxia. She suffered a number of complications after her injury, including focal seizure disorder and respiratory failure. P3 met all the inclusion/exclusion criteria for TBI subjects. She could not speak, but could use hand or head signals to answer questions with 'yes' or 'no.' Conversations held in this way and her ability to play the Hangman game indicated that her intellect was intact. P3 was severely physically impaired. She participated with her right hand, which had limited movement. She also had limited movement of her right arm and her head. Apart from these areas, she had little or no active movement. She required maximal assistance in all activities of daily living. Prior to the therapeutic sessions, P3 could, in general, pinch her fingers together such that the distance between the index and thumb was less than 15 mm (the minimum pinch distance that could be consistently measured in the robotic environment). Thus, we focus here on her performance in extension. Data from session 12 (of 18) were excluded from analysis because a programming error prevented sufficient data from being collected on this day.

Subject P3 experienced the distortion protocol during her six weeks of participation. The amount of distortion was computed as described in the previous chapter. During a distortion session, the upper bound of the distance mapped to the list of letters changed from 80% of the maximum extension measured by the calibration program to 113% of the calibration maximum. Most experimental sessions were conducted by an undergraduate assistant who did not know on which days the visual display was distorted or the details of how the distortion was implemented during a single day. Because P3 could not verbally ask for the limits of the display to be contracted, the experimenter observed her performance, and if she appeared to be having trouble reaching the letters at the ends of the display, the experimenter offered to make these letters easier to reach. P3 did not experience force feedback during the protocol, because she was unable play the Hangman game while resisting any force.

9.2. Results

During the six-week protocol, P3 always followed the distortion. P3 was consistently able to reach the letters she needed to play the game, and she never responded affirmatively when asked if she wanted the outermost letters to be made easier to reach. Therefore, relative to the maximum extension distance measured at the beginning of each rehabilitation session, P3 reached greater levels of extension on days in which she experienced distortion. This can be seen in Figure 9.1, which shows the proportion of time in each session for which d_{it} for P3 was greater than 75% of the maximum measured by the calibration program. Using the non-parametric Mann-Whitney test, the median proportion was found to be significantly greater for sessions with distortion ($p = 0.021$).

The effect of distortion on P3's performance can also be seen by examining her maximum extension distance per word across sessions (Figure 9.2). For each of the first nine words (P3 completed at least nine words in each session), we calculated the maximum extension distance used to select a letter during that word, averaged over the 2 s interval necessary to select a letter. We then averaged these distances across sessions to find how the maximum extension distance evolved as a function of time during a rehabilitation session. Though the data are noisy due to variations in the letters selected for each word and the two-up-one-down nature of our implementation of distortion, an upward tendency can be observed in the maximum extension distance for distortion sessions. This tendency was confirmed by linear trend analysis ($p = 0.020$). There was no significant linear trend for the first nine words of sessions with no distortion ($p = 0.79$).

As mentioned in Chapter 8, the P3's maximum extension distance was measured by the calibration program at the beginning of each rehabilitation session. No significant slope in this distance was found for the six-week participation period ($p = 0.94$; Figure 9.3). The calibration program also measured the standard deviation in the distance between the thumb and index finger as the subject tried to keep a visual cue representing this distance near a particular point on the screen. A marginally significant positive slope in the standard deviation was found over the duration of P3's participation ($p = 0.060$; Figure 9.4(a)). However, though the standard deviation was used to determine the number of letters shown on the screen, the number of letters shown was not allowed to decrease from session to session. By the third week of the six-week period, P3 was playing the game with the entire alphabet shown on the screen (Figure 9.4(b)). However, there was a significant upward trend with session in the time it took P3 to select each letter ($p < 0.001$) (Figure 9.5).

After each word in the Hangman game, P3 was asked to pinch and extend her fingers in the absence of visual feedback. For each session, we averaged the maximum extension distances measured during the no-feedback periods (P3 completed at least 8 no-feedback periods in all sessions). There was a highly significant downward slope in the mean maximum extension distance over the six-week period (p < 0.001; Figure 9.6(a)). The Mann-Whitney test indicated that there was no significant difference between sessions with distortion and sessions without distortion ($p = 0.67$).

We also looked at the maximum no-feedback extension as a function of time within distortion sessions, since the data collected for normal controls in Chapter 5 suggested that the distance produced in the absence of feedback might increase as the distance produced within the Hangman game increased. There was no linear trend for maximum no-feedback extension over the first 8 words within a session ($p = 0.32$; Figure 9.6(b). However, when we compared the first no-feedback maximum extension of each session with the first with-feedback maximum extension (the first distance used to select 'A'), the no-feedback extension was found to be significantly higher ($p = 0.0019$). The first no-feedback maximum extension was not significantly different from the calibration maximum ($p = 0.49$). By the end of the session, however, the last no-feedback maximum extension was marginally smaller than the last with-feedback maximum extension for sessions with distortion ($p = 0.098$) but not for sessions without distortion ($p = 0.64$).

To investigate whether P3's passive range of motion improved over the participation period, we found the sum of the maximum angles measured in flexion at the three joints of each finger. We then did the same for the minimum angles measured in extension at the three joints of each finger; this data can be found in Table 9.1. The process was repeated for P3's active range of motion for each finger (Table 9.2). No improvements from pretest to posttest were noted in either the passive or the active range of motion. The angle between the thumb and the index finger was measured both in the plane of the palm and perpendicular to the plane of the palm. The angle measured in the plane of the palm increased from 55° to 95° from pretest to posttest. However, the angle measured perpendicular to the plane of the palm decreased from 70° to 55°. The Ashworth measurements for the elbow, index finger, and thumb remained the same from pretest to posttest. Ashworth measurements for the wrist and the fingers as a group increased from 1 to 2.

The results of the ARAT were very positive for P3. During the six-week rehabilitation protocol, P3 was given the Grasp and Pinch subtests of the ARAT once a week. These results, as well as her scores on these subtests before and after the six-week period, are plotted in Figure 9.7. There was a significant upward trend in P3's scores with time (p = 0.023). P3 scored a 26/57 on the ARAT before her participation in the rehabilitation protocol and a 32/57 after her participation. If we again take a clinically important difference to be 10% of the maximum possible score [117], this is a clinically important improvement for P3. On the other hand, P3's mean Functional Ability score on the AMAT declined from a pretest value of 2.11/5 to a posttest value of 1.61/5, a clinically important decrement. P3's mean Quality of Movement score on the AMAT declined from 1.72/5 to 1.50/5. Before her participation, P3 ranked herself as 5/10 in her desire to approach challenging activities and 4/10 in her desire to move onto a harder activity after accomplishing a difficult task.

At a six-month follow-up visit, the range of motion of P3's index finger appeared to be slightly improved (Tables 9.1 and 9.2). However, the hypertone in P3's hand appeared worse, with the index finger and thumb increasing from Ashworth 1 to 2, and the fingers as a group increasing from Ashworth 2 to 3. P3 scored 31/57 on the ARAT at her follow-up visit. Her mean Functional Ability score for the AMAT at the follow-up visit was 2.00/5. Her mean Quality of Movement score was also 2.00/5.

9.3. Discussion

Throughout the six-week protocol, P3 was consistently willing and able to follow the distortion used. This indicates that she was capable of extending her fingers farther than she did during the short calibration program at the beginning of each session. Based on the data collected in this experiment, we cannot know the reason P3 performed below her actual capacity during the calibration. Her responses to our informal questionnaire suggest that the psychological issues discussed in Chapter 2 may be a contributing factor; before participation, she seemed ambivalent about choosing to pursue challenging activities. Also, P3 had spent two years almost completely paralyzed, which almost certainly reduced her confidence in her ability to improve her physical performance. It is also possible that P3 performed below her ability in the calibration program because she found it uninteresting. Regardless of its cause, P3's underperformance in calibration and willingness to follow distortion has important implications for rehabilitation in general, and for robotic rehabilitation in particular. Robotic rehabilitation systems must be calibrated for each individual, and the activities a patient is asked to perform are based on this calibration. This case study suggests that a patient's performance during calibration may not connote his or her actual abilities and that distortion can be used to guide the patient towards more challenging activities.

That P3 performed below her actual capacity during calibration is also implied by the standard deviation used to determine the number of letters shown on the screen. There was a marginally significant upward trend with session for the standard deviation, possibly caused by P3 becoming bored with the calibration program. Despite the fact that the standard deviation increased, P3 successfully played the Hangman game with more letters as the six-week period progressed, finally playing with all 26 letters shown on the screen (Note: Before her participation in the experiment, P3 attempted the Hangman game with all 26 letters, and was unable to play successfully). However, the upward trend with session in search time per letter indicates that P3 did find it more difficult to select letters when more letters were presented on the screen. This upward trend could also reflect the increased cognitive load of playing the game with more letters.

Over the six-week participation period, we observed a significant downward trend in the mean maximum extension measured during the no-feedback periods presented throughout the Hangman game. The cause of this trend is unclear; one possibility is that it was due to increasing boredom with these interruptions of the Hangman game. Though the cause is unknown, the fact that a downward trend occurs in the absence of feedback is a good argument for the use of visual feedback in robotic rehabilitation, and for the use of visual distortion in particular.

We also examined the no-feedback data to determine whether there was an upward trend with time during distortion sessions. In Chapter 5, we discussed how visual feedback distortion affected the performance of young unimpaired subjects, even during no-feedback trials. This trend did not occur for elderly unimpaired subjects. In this respect, P3's performance was similar to that of our elderly subjects. No significant upward trend in no-feedback performance occurred during distortion sessions. Interestingly, this may be explained using the same argument we employed for the elderly subjects in Chapter 5. P3's no-feedback maximum extension was significantly higher than her with-feedback performance at the beginning of the experiment. Thus, she did not have to increase her no-feedback extension in order to follow the distortion. We chose the starting point of the distortion as 80% of the calibration maximum because, based on normal subjects, we expected a subject to require considerable effort to repeatedly produce 80% of his or her initial maximum. Due to fatigue, we expected that the maxima measured during the no-feedback periods would be less than the maximum measured during calibration. Thus, it is somewhat surprising that we found the first no-feedback maximum to be significantly greater than the first with-feedback maximum and that the first no-feedback maximum was not significantly different from the calibration maximum. Again, this indicates that P3 was capable of extending more than she did during the calibration program. If her performance during the calibration program had been her absolute maximum, she would have found it difficult to repeat that performance.

The distortion increased the upper bound of the distance range mapped to the alphabet to 113% of the calibration maximum over the first twelve words, and P3 sometimes completed as many as 15 words in a session. Figure 9.8 shows, for each session, P3's last no-feedback maximum extension and her last with-feedback maximum extension (the last distance used to select 'A'). In general, after the first session of the experiment (during which her motivation was likely highest), P3's last no-feedback maximum was less than her last with-feedback maximum. By the end of the session, she was usually reaching higher levels of extension with the distortion than in the absence of feedback. We should also note that while she had to hold the with-feedback maximum extension for two seconds to select a letter, P3 was not required to hold the maximum extension during no-feedback periods. After analyzing these data, we modified the protocol so that the no-feedback maximum extension also had to be held for 2 seconds.

Regardless of the effects of distortion, we expected practice of the extension task to result in an upward trend in the maximum extension distance measured by the calibration program, similar to the increase in force we observed for P1 over her six-week participation in our earlier experiment (Chapter 7). We observed no such upward trend. We thought that this might have been because P3 did not have to hold her maximum extension for any length of time during the calibration program, whereas P1 had to hold her maximum force for approximately 5 seconds. For this reason, we altered the calibration program so that subjects had to hold their minimum pinch and maximum extension for 5 seconds each.

The results of the occupational therapy tests given to P3 were quite mixed. In part, we found that this was because our occupational therapist (though highly recommended) did not always administer the tests in a well-standardized way. This was true even of the range of motion measurements, and we believe it to be more a reflection of current practice in occupational therapy than an indication of the quality of our therapist. In addition, interpretation of the therapy test results is difficult because due to unavoidable delay, the pre-participation tests were given 3 months before P3 experienced the rehabilitation protocol. No clear improvements were seen in range of motion immediately after the six-week protocol, and P3's Ashworth scores remained the same for the index finger and thumb. On the other hand, there did appear to be some improvement in the index finger at the six-month follow-up, despite the increase in hypertone noted at that visit. P3's scores on the AMAT declined from pretest to posttest evaluation, though her performance at the follow-up visit was similar to her original score. However, we saw a clinically significant improvement on the ARAT, the therapy test most closely related to the pinching and extension task practiced in the Hangman game. This was also the most standardized test, as the therapist was unfamiliar with this test and followed a precise protocol for how it was to be given. We noted a significant upward trend in P3's score on the subtests of the ARAT that were administered each week during the rehabilitation protocol. It could be argued that the improvements we noted on the ARAT were a result of P3 having been given parts of the test multiple times, but we find it unlikely that a few minutes of task practice each week would cause such a large improvement in performance. P3's improvement on the ARAT was maintained at her six-month follow-up. It should be noted that during the 6 months between the experiment and the follow-up, P3 obtained a communication device that she operated using her fingers. This practice may have helped her maintain the gains we observed immediately after the 6-week protocol.

In sum, this case study indicates that distortion can be used with a brain-injured subject to improve her performance within a single session. P3's underperformance in the calibration program and the downward trend observed in her no-feedback data offer reasons why distortion may be a positive addition to robotic rehabilitation protocols. We were encouraged by the positive results of the protocol on P3's performance in the practical pinching and grasping skills examined by the ARAT; these results indicate that work in our robotic environment can lead to functional gains. This case study led us to make some changes in the protocol to obtain more reliable data from future subjects. Clearly, the results obtained here were for a single subject and cannot be generalized, but this case study provides a positive first indication of the possible results of our rehabilitation paradigm.

Passive Range of Motion

Finger	Pretest Extension	Pretest Flexion	Posttest Extension	Posttest Flexion	Follow-up Extension	Follow-up Flexion
Index	H 20	280	H 10	275	H 40	320
Middle	H 30	300	H 10	265	H 55	283
Ring	H 18	290	H 40	245	H 65	285
Small	H 20	275	H 75	270	H 50	180

Table 9.1.

Active Range of Motion

Finger	Pretest Extension	Pretest Flexion	Posttest Extension	Posttest Flexion	Follow-up Extension	Follow-up Flexion
Index	H 10	239	H 10	225	H 20	260
Middle	H 10	251	H 10	225	H 45	240
Ring	H 15	258	H 40	220	H 25	235
Small	H 15	250	H 60	173	H 5	200

Table 9.2.

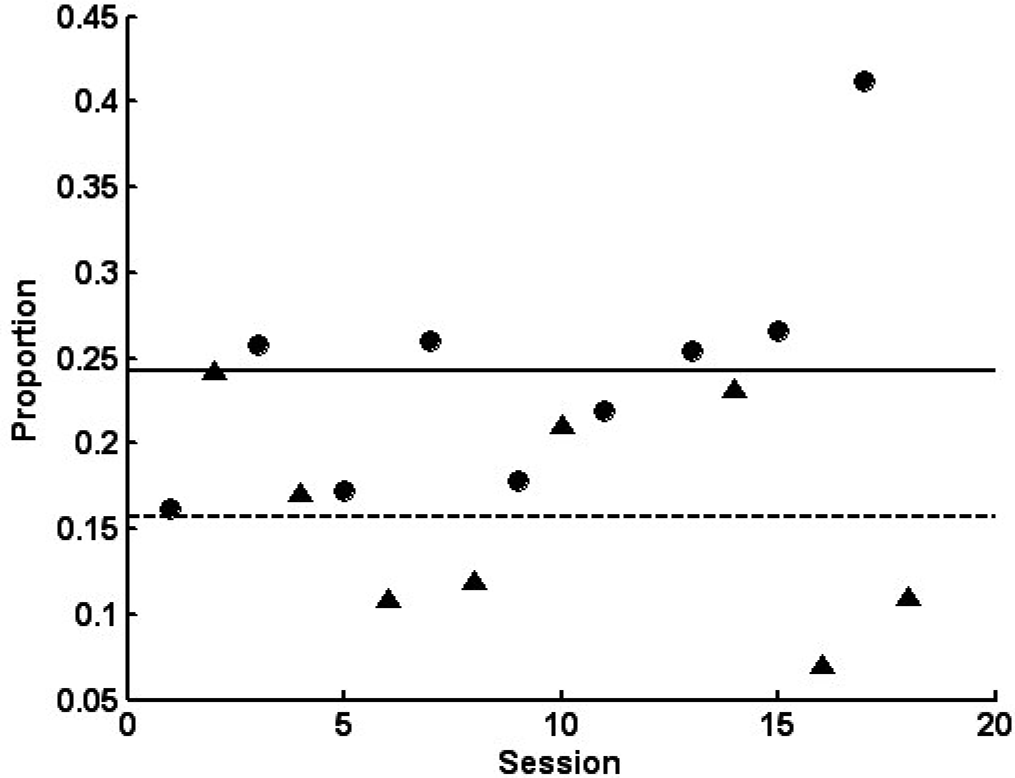


Figure 9.1. Relative to the maximum extension distance measured at the beginning of each rehabilitation session, P3 spent more time at greater levels of extension on days in which she experienced distortion. The *x*-axis represents session number, and the *y*-axis represents the proportion of each session during which P3's extension was greater than 75% of the maximum extension measured by the calibration program at the beginning of the relevant session. Circles indicate sessions with distortion, and the solid line represents the mean of these sessions. Triangles represent sessions without distortion, and the dashed line indicates the mean of the no-distortion sessions. The proportion of each session in which P3's extension was greater than 75% of the maximum calibration extension was significantly greater for distortion sessions.

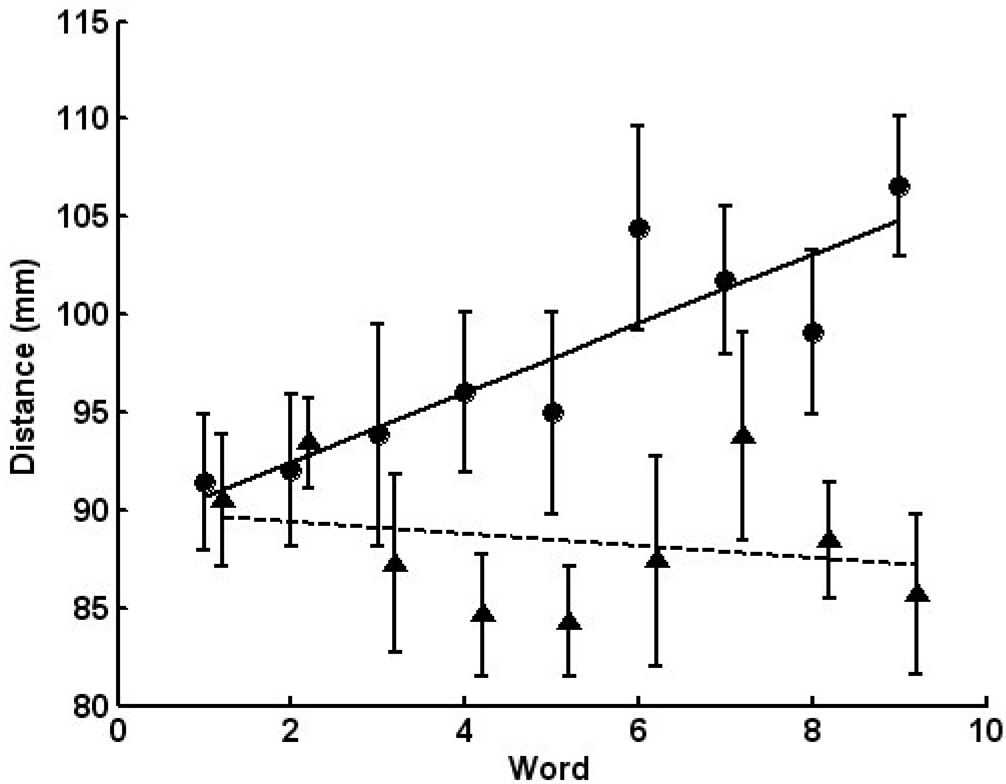


Figure 9.2. P3 consistently followed the distortion, causing an upward trend in her extension during distortion sessions. Circles represent the mean over distortion sessions of the maximum extension distance used to select a letter in each word. Triangles represent the same quantity for no-distortion sessions. No distortion data is slightly horizontally offset from distortion data for clarity. Regression lines for each data set are shown. All error bars represent standard error. A significant linear trend in extension distance with word number was observed for distortion sessions, but not for no-distortion sessions.

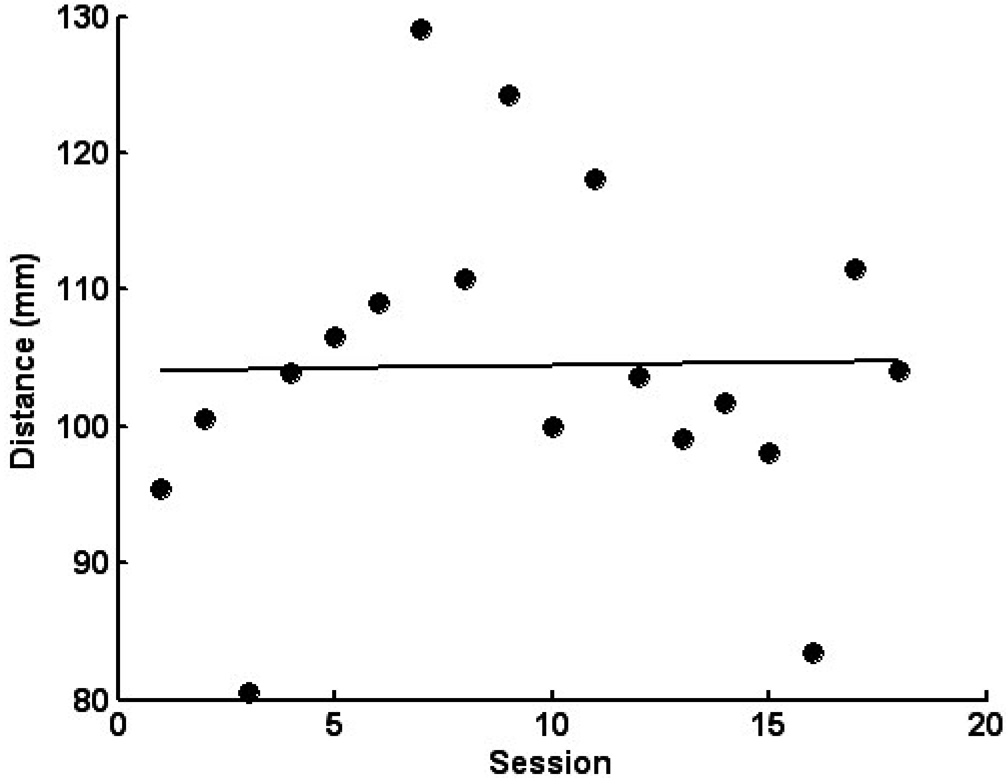


Figure 9.3. The maximum extension distance measured for P3 by the calibration program at the beginning of each rehabilitation session. A regression line is shown. The slope of this line was not significantly different from zero.

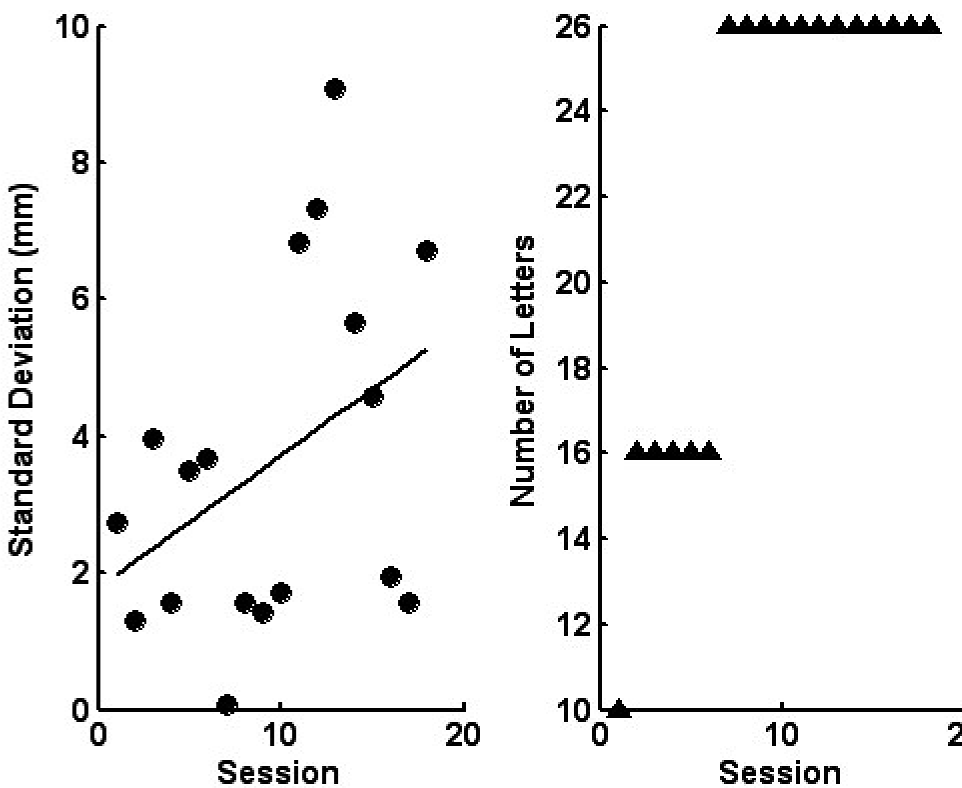


Figure 9.4. **(a)** For each session, the calibration program measured the standard deviation in the distance between the thumb and index finger as P3 tried to keep a visual cue representing this distance near a particular point on the screen. This data was used to determine the number of letters shown on the screen, with the caveat that the number of letters shown was not allowed to decrease from session to session. A regression line is shown for the standard deviation data as a function of session number, and this line had a marginally significant positive slope. **(b)** The number of letters from which P3 selected for each session of Hangman. Although P3's standard deviation data had an upward tendency, she was capable of playing the Hangman game with more letters as the six-week period progressed.

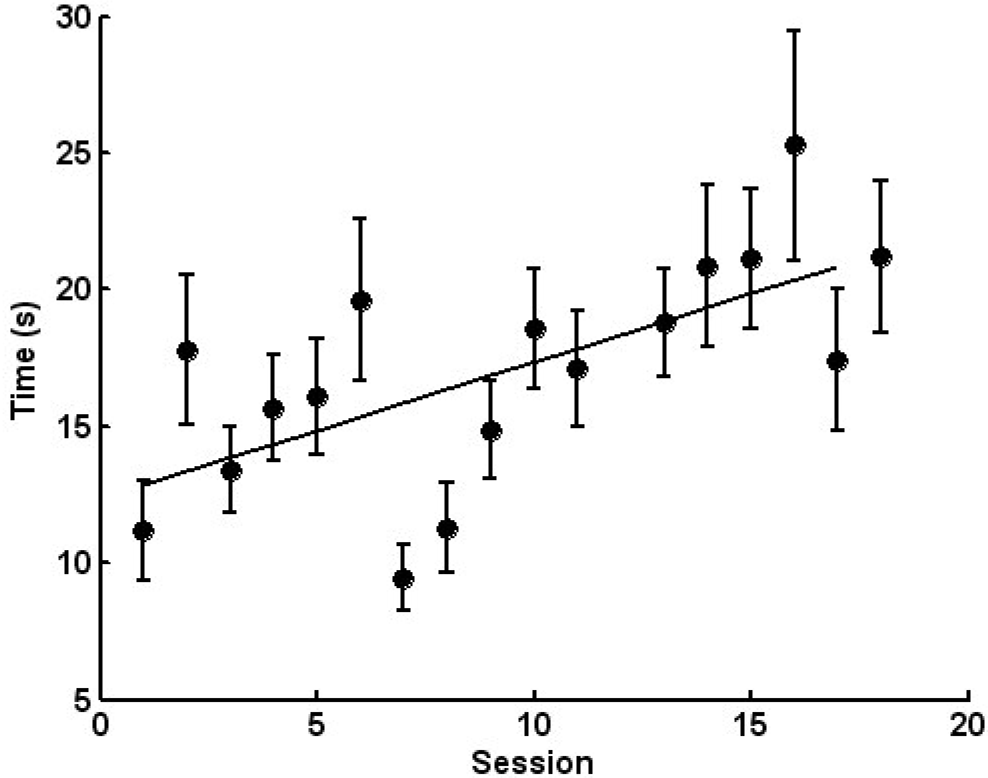
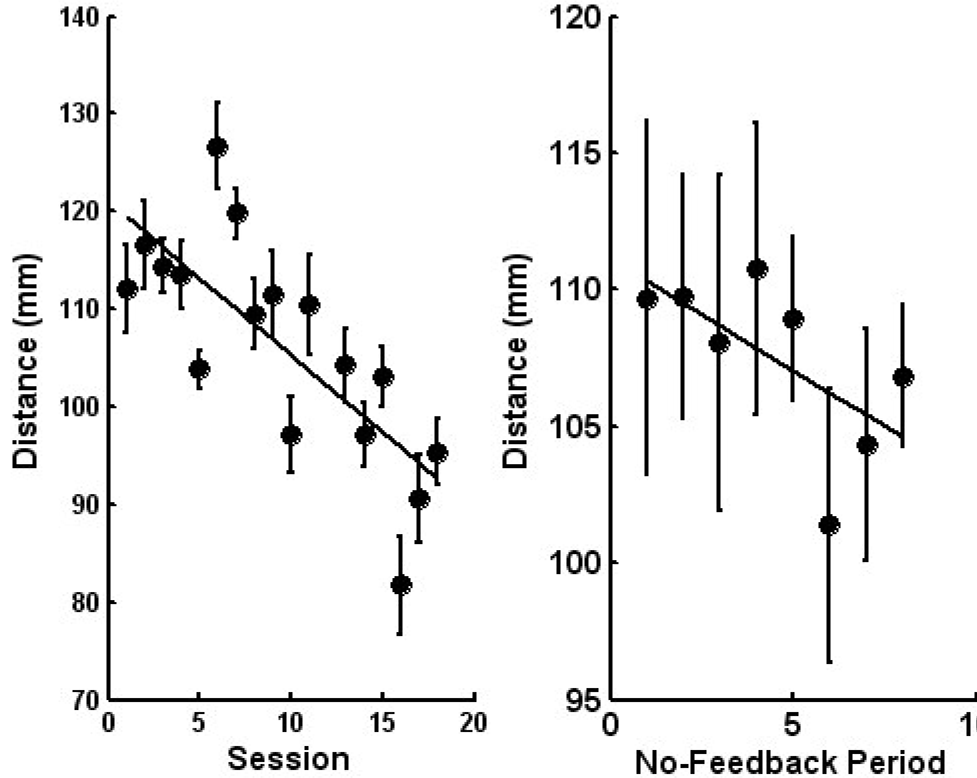


Figure 9.5. P3's mean time to select a letter increased as a function of session. Each point represents the mean time it took P3 to select a letter during that rehabilitation session. Bars represent standard error, and a regression line is shown. A significant upward trend was found for the mean search time with session.



(a) **(b)**

Figure 9.6. **(a)** P3's mean no-feedback maximum extension decreased as a function of session. P3's maximum extension was measured during a number of no-feedback periods spaced throughout the Hangman game. Each point represents the mean of these maxima for the given rehabilitation session. Error bars represent standard error. The downward trend indicated by the regression line was highly significant. **(b)** No significant trend was found for the no-feedback maximum extension as a function of time within a distortion session. For each no-feedback period, we averaged the maximum extension distance over distortion sessions. Standard error bars and a regression line for this data are shown.

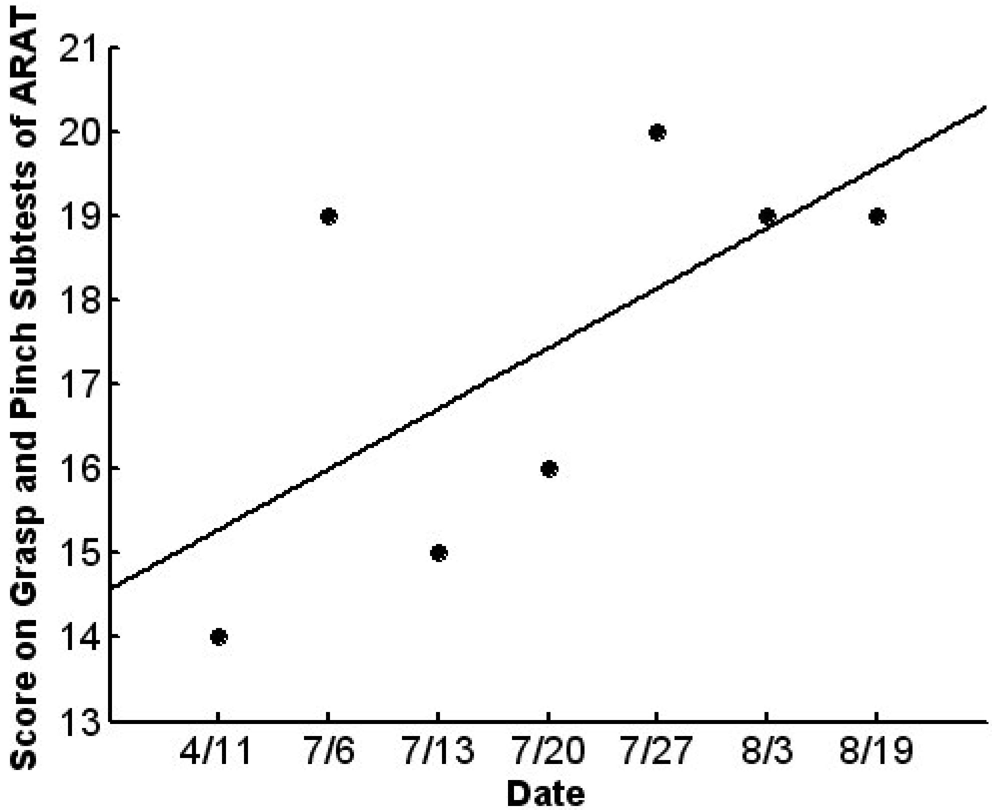


Figure 9.7. Throughout the six-week participation period, P3's practical pinching skills improved. A subtest of the ARAT was given to P3 once a week. The positive slope of the regression line shown was significant.

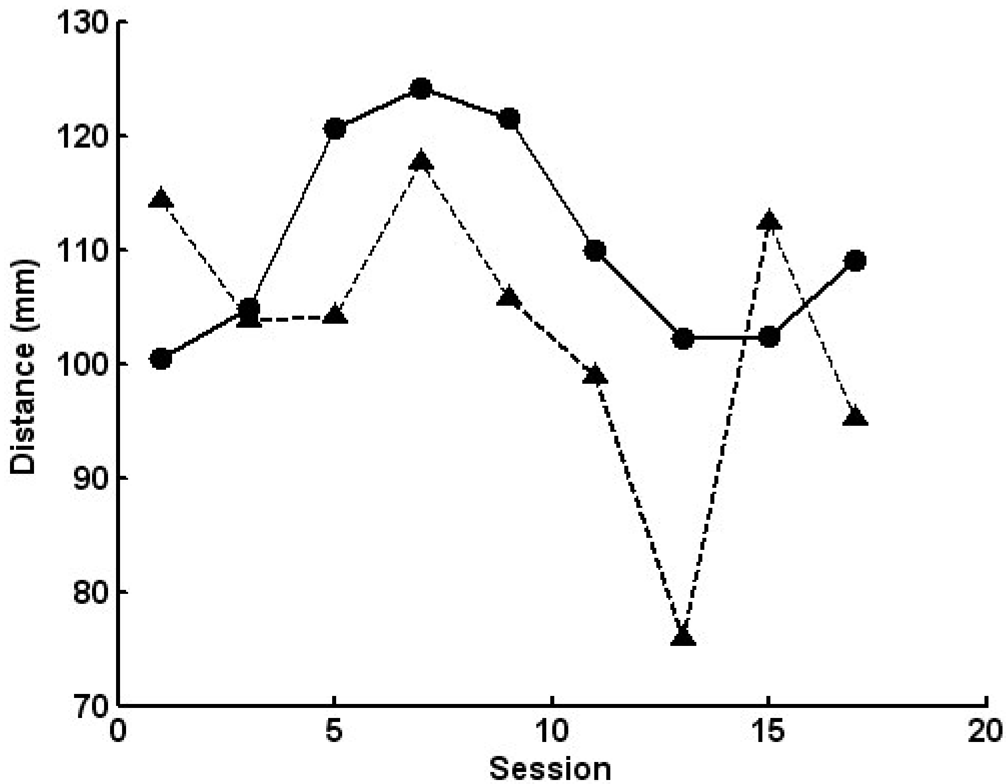


Figure 9.8. By the end of a distortion session, P3 usually produced more extension when shown the distorted visual feedback than with no feedback. The solid line indicates P3's last with-feedback maximum extension (last 'A' selected). The dotted line indicates her last no-feedback maximum extension.

Chapter 10. Case Study: Subject P4

10.1. Subject Description

Subject P4 was a 76-year-old female. Seven years before her participation in the study, she suffered a stroke that affected the right side of her body. Her left side was unimpaired. P4 met all the inclusion/exclusion criteria for stroke subjects given in Appendix A. Her right leg and foot were impaired, but she was able to walk using a cane. Her speech was slow but understandable, and she was cognitively unimpaired. She had limited use of her right arm and hand and used her right hand to participate in the experiment. She was independent in most activities of daily living. As was the case with P3, P4 could bring her index finger and thumb together such that the distance between them was less than the lower limit that could be consistently measured by the robots. Thus, we focus here on her results in extension. Again, data from session 12 was excluded from analysis because a programming error prevented sufficient data from being collected on this day.

P4 experienced distortion on alternate days during the rehabilitation protocol. In each distortion session, the upper limit of the range of distance mapped to the alphabet changed gradually from 80% of the maximum extension measured by the calibration program to 108% of the calibration maximum. Again, most sessions were conducted by an undergraduate assistant who did not know the details of the distortion. P4 was capable of asking the experimenter to contract the range of distance mapped to the visual display. After observing the experimenter use the 'x' key, P4 pressed this key herself when she wished to contract the distance range.

10.2. Results

Our analysis of the data measured with P4 paralleled that described in Chapter 9 for P3, and our results were similar. Unlike P3, P4 did occasionally choose to contract the range of distance mapped to the visual display. She chose to do this during 3 out of 9 distortion sessions. These contractions of the distance range took place on the last word of the session, on word 9 of 15, and on word 9 of 11, respectively. However, P4 also chose to press the 'x' key (or have it pressed) during 3 out of 9 no-distortion sessions. During no-distortion sessions, pressing the 'x' key had no effect. Like P3, P4 produced greater amounts of extension (relative to the initial maximum measured by calibration program) during distortion sessions. The median proportion of time in each session for which d_{it} was greater than 75% of the calibration maximum was marginally greater for distortion sessions ($p = 0.059$; Figure 10.1). For each of the first 9 words presented in the Hangman game, we averaged over all distortion sessions the maximum extension distance used to select a letter. We used the first 9 words because P4 completed at least 9 words in every session except session 2, in which she completed only 6 (explained below). For this reason, session 2 was omitted from all analyses that investigated performance as a function of word number within a session. Linear trend analysis showed that the maximum extension distance increased as a function of word number within distortion sessions ($p = 0.0051$), but not within sessions without distortion ($p = 0.28$) (Figure 10.2).

Figure 10.3 shows P4's maximum extension distance measured by the calibration program at the beginning of each rehabilitation session. Surprisingly, there was a significant downward slope to this data ($p < 0.001$). There was also a significant downward trend in the standard deviation of the distance between the index finger and the thumb ($p = 0.049$; Figure 10.4). This corresponded to P4's ability to play the Hangman game with an increasing number of letters shown on the screen. Unlike P3, P4 was able to play the game with more letters without increasing the mean time needed to select a letter ($p = 0.23$; Figure 10.5).

To assess P4's performance during the no-feedback periods interspersed throughout the Hangman game, we found the maximum extension distance measured during each of the no-feedback periods. We took the mean of these distances for each session and found a highly significant downward slope over the 6-week rehabilitation protocol ($p < 0.001$; Figure 10.6(a)). There was no significant difference between sessions with distortion and sessions without distortion ($p = 0.89$).

We examined P4's maximum extension during no-feedback periods as a function of time within a session for the first 9 words of distortion sessions. We did not find a significant linear trend (p = 0.26; Figure 10.6(b)). However, as for P3, the first with-feedback maximum extension for each session (the first distance used to select 'A') was significantly less than the first no-feedback extension (p < 0.001), while the first no-feedback maximum extension was not significantly different from the calibration maximum (p = 0.33). At the end of the session, the last no-feedback maximum extension was not significantly different from the last with-feedback maximum extension for distortion sessions (p = 0.91) or sessions without distortion (p = 0.64).

Force-feedback was integrated into the Hangman game for P4. During every session, P4 experienced a resistive force in flexion that rose from 0 N at the neutral position to 2 N at the lower bound of the range of distance mapped to the alphabet. At the beginning of the 6-week period, P4 was unable to play the Hangman game while encountering a resistive force in extension. Our attempt to apply a resistive force in session 2 is the reason P4 completed only 6 words in that session; those 6 words were completed after we removed this force. By session 13, P4 was able to play the Hangman game with all 26 letters displayed on the screen and completed 24 words in a 90-minute session. At this point, we added force feedback in extension. During the last five rehabilitation sessions, P4 experienced a resistive force in extension that rose from 0 N at the neutral position to 1 N at the upper bound of the range of distance mapped to the alphabet. This resistive force likely accounts for the spike at sessions 14 and 15 in P4's mean time to select a letter. The direction of the resistive force applied to each finger was based on a least-squares fit to the path of that finger. When the robot was started at the beginning of the Hangman game or after a break, no force was applied to P4's finger. The robot recorded the position of P4's fingers until it collected sufficient data to determine a least-squares fit for each finger. After the program determined the least-squares fit, P4 began to encounter a resistive force tangent to the path of each finger. Due to the delay before the robots began to produce force and because force was present only in flexion, P4 encountered force only 22.9% of the time she spent playing the Hangman game in sessions 1 through 13. In sessions 14-18, P4 experienced force during 70.1% of the time she spent playing the game. P4 chose to contract the range of distance mapped to the visual display only once in the 5 sessions with force present in extension;

this occurred during a distortion session. No force was present during calibration or during the no-feedback periods.

In general, P4's passive ROM tended to decrease between the therapist evaluation at the beginning of the 6-week protocol and the ending evaluation (Table 10.1). Her active ROM for flexion was more constant, though after the 6-week paradigm, P4 displayed less active hyperextension in the index finger and more in the ring and little fingers (Table 10.2). The angle between the index finger and the thumb increased from 30° to 43° in the plane perpendicular to the palm and decreased from 48° to 40° in the plane of the palm. In terms of the Ashworth scale, P4 scored a 1 for her fingers and thumb both before and after the 6-week protocol. The Ashworth score for her elbow decreased from 2 to 1, while her wrist increased from Ashworth 1 to Ashworth 2.

Figure 10.7 shows P4's score over time on the Grasp and Pinch subtests of the ARAT. P4's subtest score had a significant upward trend over time ($p = 0.0013$). P4's score on the entire ARAT increased from 38/57 before her participation in the experiment to 40/57 after her participation, and this increase is not clinically important. However, on the AMAT, her mean Functional Ability score increased from 2.29/5 to 2.86/5, and her mean Quality of Movement score increased from 2.14/5 to 2.71/5. Both of these increases were clinically important. At the beginning of the experiment, P4 ranked herself as 2/10 in her desire to approach challenging activities and 0/10 in her desire to move onto a harder activity after accomplishing a difficult task.

At the 6-month follow-up, P4 demonstrated less flexion and more hyperextension than she showed at her first two evaluations (Tables 10.1 and 10.2). This was true for both active and passive ROM. P4 also showed increased hypertone at her 6-month follow-up. She scored an Ashworth 2 for the elbow wrist, and fingers, while her thumb maintained an Ashworth 1 score. P4 had not maintained her improvement on the ARAT; her ARAT score was 36/57. Her AMAT improvements were partially maintained with her Functional Ability score and her Quality of Movement score both being equal to 2.32/5.

10.3. Discussion

Many of our results with P4 were similar to those described in Chapter 9 for P3. Like P3, P4 was capable of moving farther than she did during the calibration program at the beginning of each session. She consistently followed the visual distortion and was able to play the Hangman game as the range of distance required increased from 80% of her calibration maximum to 108% of this maximum. P4 did occasionally request a reduction in the range of distance mapped to the alphabet. However, her desire to have the range of distance reduced was not correlated with the distortion, as she made this request equally often on days with no distortion. P4 chose to contract the distance range during the first distortion session in which she experienced resistive force in extension, but during the second distortion session featuring this force, she followed the distortion. On our informal questionnaire, P4 indicated that she was reluctant to attempt difficult goals and to progress to new goals after a success. These qualities may partially explain why P4 extended her fingers farther with the distortion than she did during the calibration program. Whatever the reason for it, P4's positive response to the distortion adds additional qualitative evidence indicating that stroke patients may be capable of performing better than they indicate when asked to execute a specific movement.

Over the 6-week protocol, we found a significant downward trend in the mean maximum extension of P4 during the no-feedback periods. This decrease was analogous to that observed with P3 and can be similarly attributed to P4 becoming bored with the no-feedback periods as time progressed. Again, this result indicates that distortion and visual feedback in general may be important to maintaining patient performance during rehabilitation.

When we examined P4's performance in the no-feedback periods as a function of time in distortion sessions, we found that her no-feedback maximum extension did not increase over the first 9 words. As was the case for P3, this is probably because her no-feedback maximum extension at the beginning of the session was greater than her with-feedback maximum extension. P4's performance during the first no-feedback period was similar to her calibration performance. At the end of the session, P4's no-feedback maximum extension was similar to her with-feedback extension (Figure 10.8), even though the distortion required her to produce up to 108% of her calibration maximum when feedback was present. These results emphasize P4's underperformance during the calibration program. As for P3, P4 was required to hold her maximum with-feedback extension for 2 s, while she was not required to hold her maximum extension during calibration or no-feedback periods.

Instead of the expected upward trend, we observed a significant downward trend over the 6-week period in the maximum extension measured for P4 during the calibration program. We suspect that over the 6 weeks, P4 may have become bored with the calibration program and exerted less effort during it. It is possible that providing the subject with visual feedback during calibration would avoid this problem. We also observed a significant downward trend in the standard deviation for distance measured by the calibration program. This trend corresponded to P4's ability to play the Hangman game with an increasing number of letters without an increase in the mean time required for her to select a letter.

P4 exhibited no improvements in range of motion as a result of her participation in the experiment, but like P3, P4 improved her performance on the ARAT over the six-week period. We measured a significant positive slope in her score on the ARAT subtests given throughout the 6 weeks. This is important, given that the ARAT is the occupational therapy test most closely related to the tasks practiced in our rehabilitation paradigm. P4's improvement on the entire ARAT did not reach the level of clinical importance, but the fact that we did observe an improvement suggests that transfer can occur between our rehabilitation program and functional tasks. P4 did demonstrate a clinically important improvement on the AMAT, which also suggests that practice in our environment can lead to functional improvement. Unfortunately, unlike P3, P4 had not maintained her gains on the ARAT and AMAT at the 6-month follow-up. We cannot know the reason for this, but it may have been caused by a failure to utilize the functional improvements in the home environment. Like P3, P4 had a large gap between her initial evaluation and the beginning of the 6-week protocol.

To summarize, our results for P4 were very similar to those we observed for P3. P4 was capable of a greater range of motion than she displayed during the calibration at the beginning of each session, and distortion was used to encourage P4 to improve her performance to 108% of her calibration maximum. Downward trends over the 6-week protocol in P4's calibration maximum and mean no-feedback extension indicate that patient performance may deteriorate over time in the absence of visual feedback.

Passive Range of Motion

Finger	Pretest Extension	Pretest Flexion	Posttest Extension	Posttest Flexion	Follow-up Extension	Follow-up Flexion
Index	H 52	290	H 20	260	H 70	260
Middle	H 38	300	H 10	265	H 60	260
Ring	H 30	295	H 65	275	H 60	240
Small	H 75	225	H 10	250	H 85	230

Table 10.1.

Active Range of Motion

Finger	Pretest Extension	Pretest Flexion	Posttest Extension	Posttest Flexion	Follow-up Extension	Follow-up Flexion
Index	H 32	240	20	230	H 40	195
Middle	H 10	240	H 10	245	H 30	230
Ring	H 15	230	H 65	225	H 35	220
Small	H 45	220	H 70	225	0	185

Table 10.2.

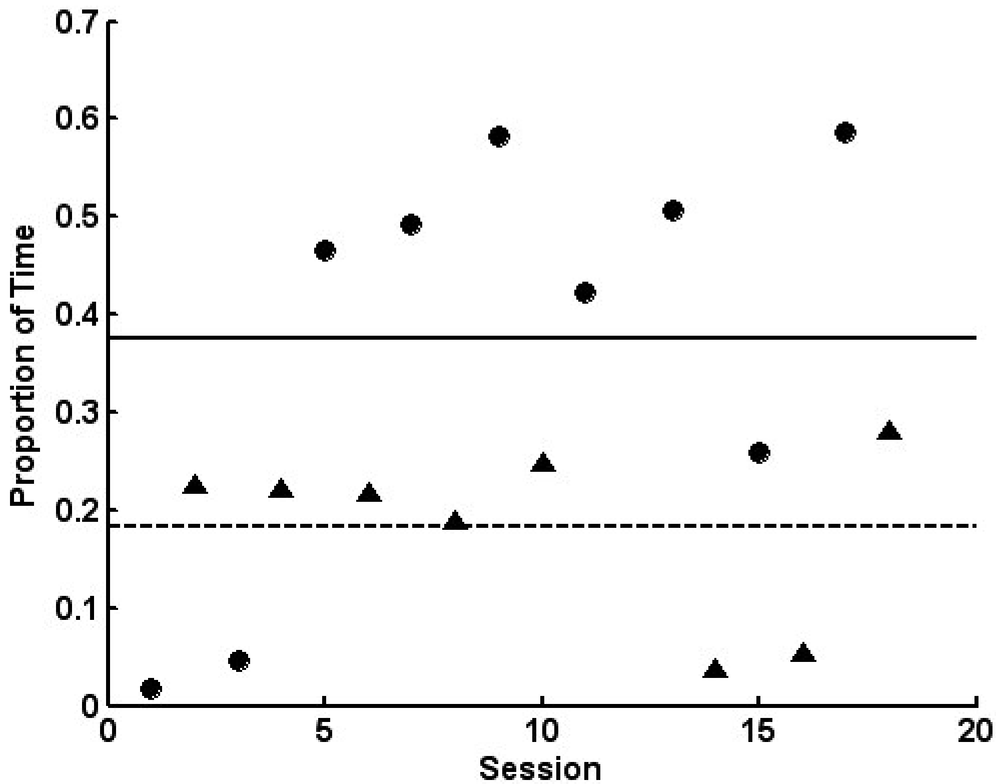


Figure 10.1. Relative to the maximum extension distance measured at the beginning of each rehabilitation session, P4 spent more time at greater levels of extension on days in which she experienced distortion. The *x*-axis represents session number, and the *y*-axis represents the proportion of each session during which P4's extension was greater than 75% of the maximum extension measured by the calibration program at the beginning of the relevant session. Circles indicate sessions with distortion and the solid line, the mean of these sessions. Triangles represent sessions without distortion, and the dashed line indicates the mean of the no-distortion sessions. The proportion of each session in which P4's extension was greater than 75% of the maximum calibration extension was marginally greater for distortion sessions.

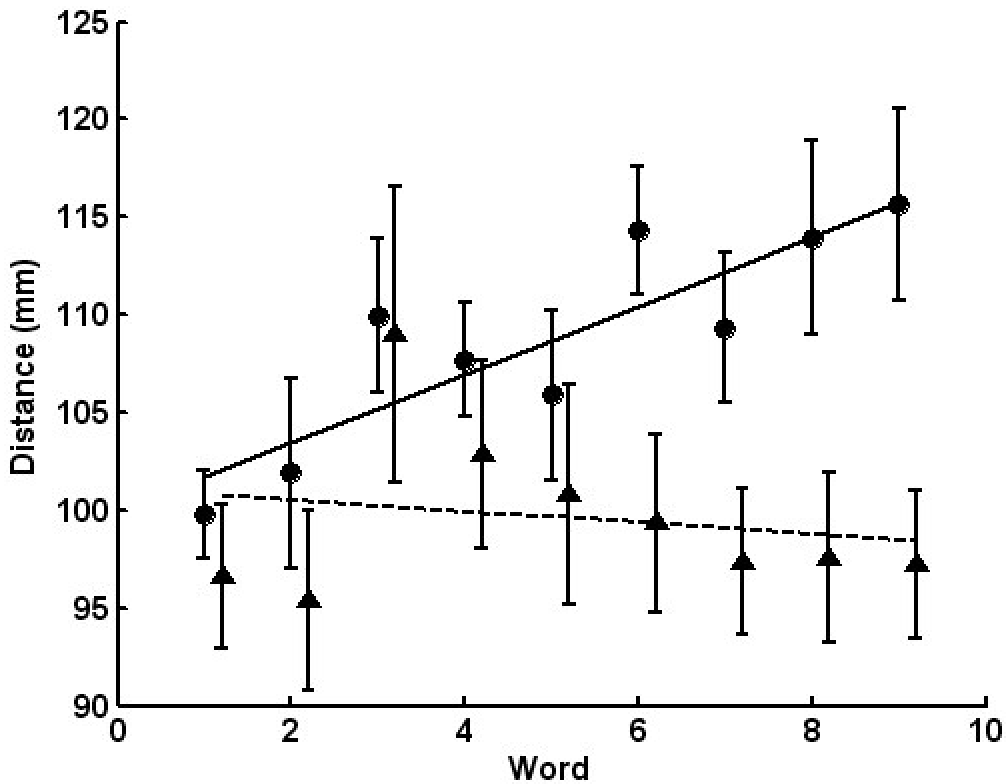


Figure 10.2. P4 consistently followed the distortion, causing an upward trend in her extension during distortion sessions. Circles represent the mean over distortion sessions of the maximum extension distance used to select a letter in each word. Triangles represent the same quantity for no-distortion sessions. No distortion data is slightly horizontally offset from distortion data for clarity. Regression lines for each data set are shown. All error bars represent standard error. A significant linear trend in extension distance with word number was observed for distortion sessions, but not for no-distortion sessions.

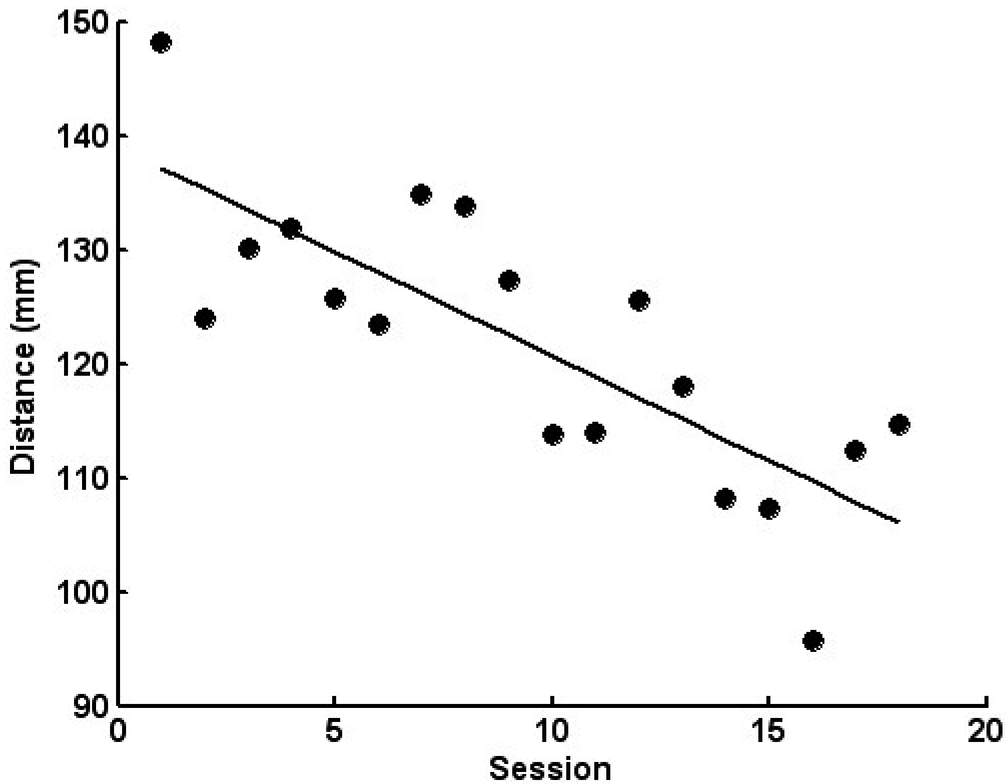


Figure 10.3. The maximum extension distance measured for P4 by the calibration program at the beginning of each rehabilitation session. A regression line is shown. The slope of this line was significantly less than zero.

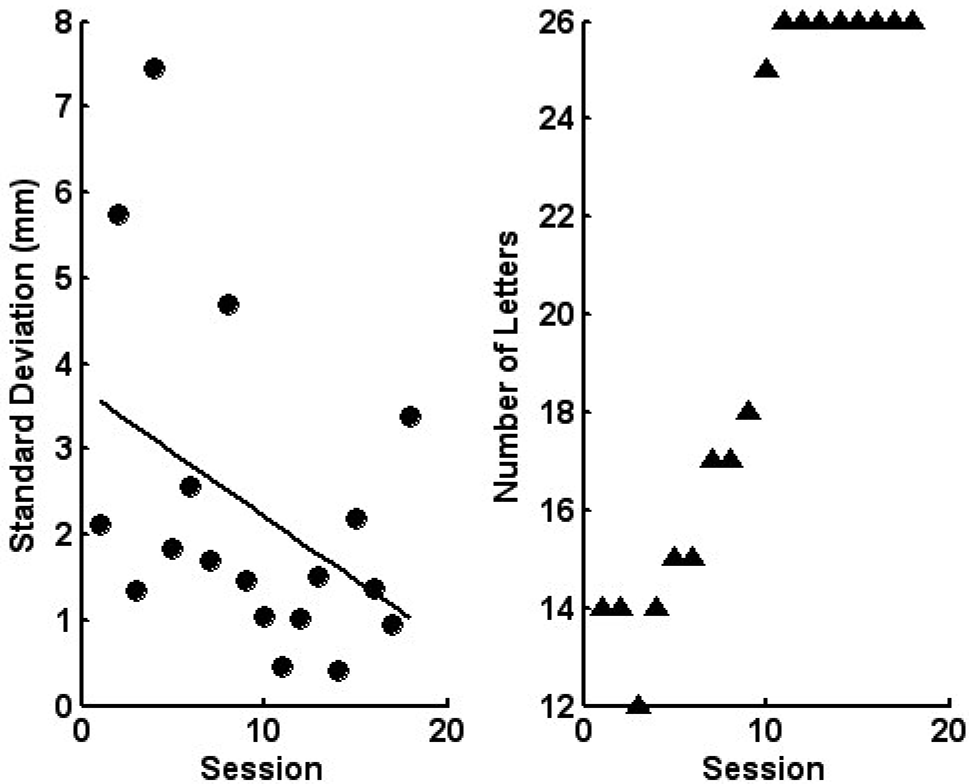


Figure 10.4. **(a)** For each session, the calibration program measured the standard deviation in the distance between the thumb and index finger as P4 tried to keep a visual cue representing this distance near a particular point on the screen. This data was used to determine the number of letters shown on the screen, with the caveat that the number of letters shown was not allowed to increase from session to session. A regression line is shown for this data, and this line had a significant negative slope. **(b)** The number of letters from which P4 selected for each session of Hangman. The downward trend shown in (a) corresponded to P4's ability to play the Hangman game with an increasing number of letters. The decrease in the number of letters from session 2 to session 3 was due to experimenter error.

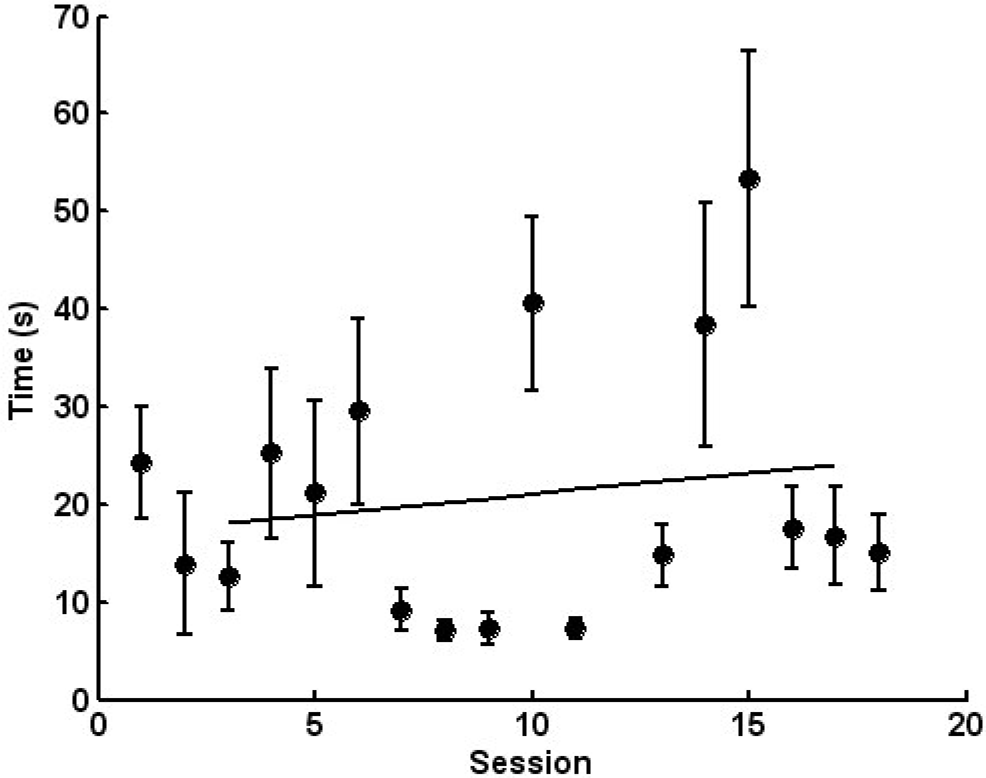


Figure 10.5. Each point represents the mean time it took P4 to select a letter during that rehabilitation session. Bars represent standard error, and a regression line is shown. No significant trend was found for the mean search time with session.

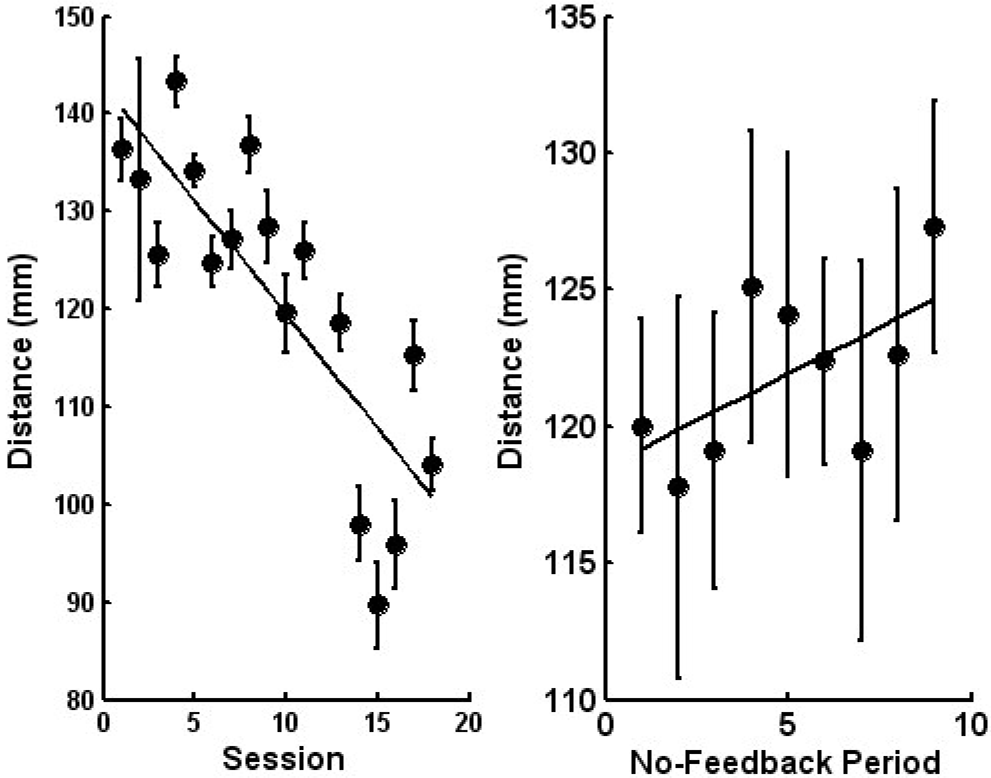


Figure 10.6. **(a)** P4's mean no-feedback maximum extension decreased as a function of session. P4's maximum extension was measured during a number of no-feedback periods spaced throughout the Hangman game. Each point represents the mean of these maxima for the given rehabilitation session. Error bars represent standard error. The downward trend indicated by the regression line was highly significant. **(b)** No significant trend was found for the no-feedback maximum extension as a function of time within a distortion session. For each no-feedback period, we averaged the maximum extension distance over distortion sessions. Standard error bars and a regression line for this data are shown.

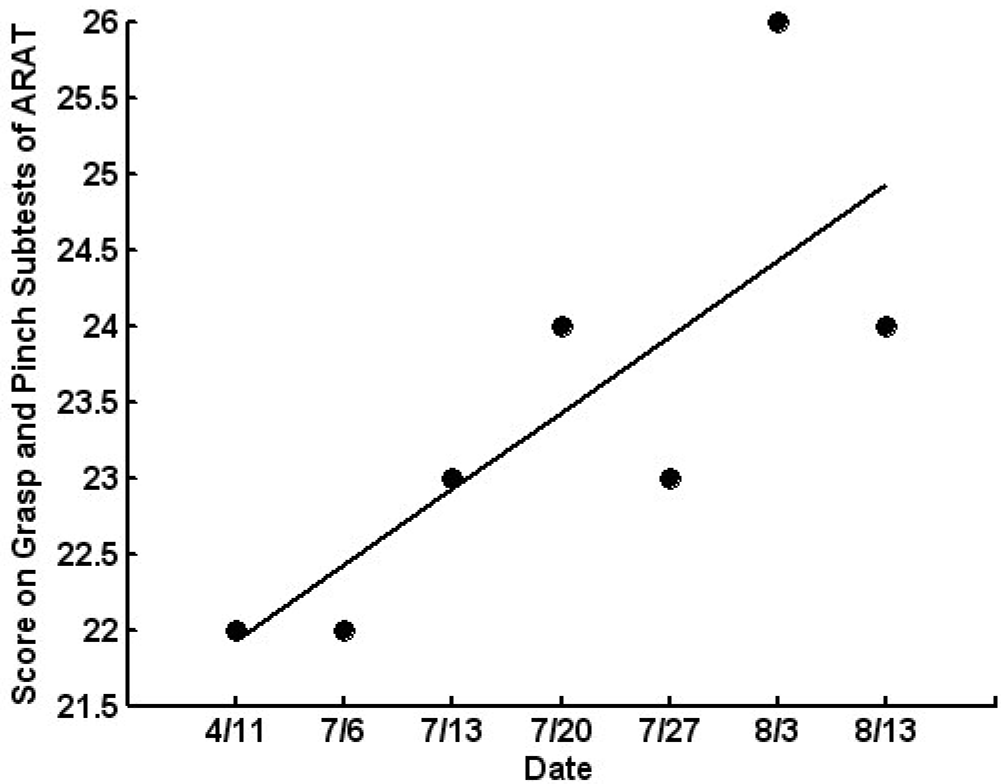


Figure 10.7. Throughout the 6-week participation period, P4's practical pinching skills improved. A subtest of the ARAT was given to P4 once a week. The positive slope of the regression line shown was significant.

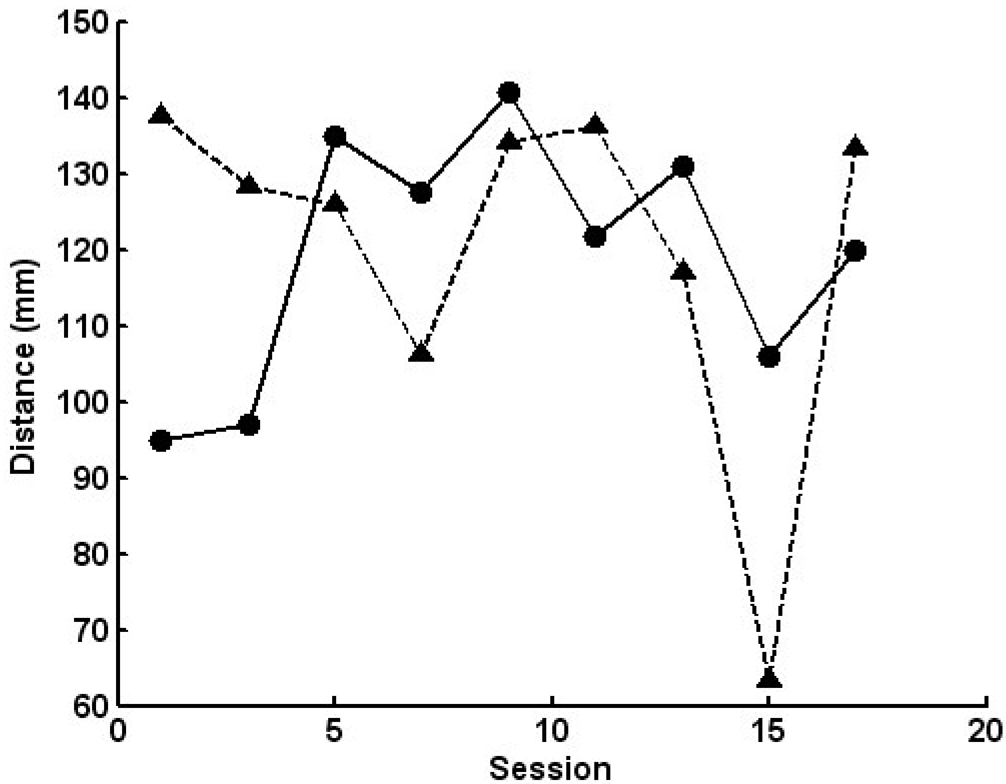


Figure 10.8. At the end of a distortion session, P4 often produced as much extension with no feedback as when shown the distorted visual feedback. The solid line indicates P4's last with-feedback maximum extension (last 'A' selected). The dotted line indicates her last no-feedback maximum extension.

Chapter 11. Case Study: Subject P7

11.1. Subject Description

Subject P7 was an 82-year-old male. He suffered three strokes before his participation in our study: one 12 years before participation, one 10 years before, and the last 9 years before participation. He participated with his impaired left hand, which had limited movement. The range of motion of his left arm was good, though he was unable to lift the arm above his head. His left leg wa s impaired, but he was able to walk using a cane. His speech was clear, and he was cognitively unimpaired. Prior to the therapeutic sessions, P7 could not bring his index finger and thumb together, and he was also limited in the amount of separation he could achieve between the fingers. However, at the beginning of the 6-week paradigm, P7 was capable of playing the Hangman game with the vowels presented on both the left and right sides of the screen and with all consonants shown (31 letters total). For this reason, we do not discuss the standard deviations in distance measured by the calibration program. His lack of tremor made it fairly easy for him to select letters, so we also do not address the mean time it took him to select a letter.

P7 experienced a 6-week progression protocol. The amount of the progression that occurred in odd-numbered sessions was computed as described in Chapter 8. During progression sessions, P7 was encouraged to decrease his minimum pinch from 120% of the calibration minimum to 87% of this minimum. He was simultaneously encouraged to increase his maximum extension distance from 80% of the calibration maximum to 105% of this maximum. During each progression session, P7 was told that it would become harder to reach the far right and far left of the list of letters shown on the screen. He was told to alert the experimenter if he needed her to make it easier to reach the outside letters.

11.2. Results

P7 predominantly followed the progression during the 6-week protocol. He chose twice to have the range of distance mapped to the screen reduced (sessions 15 and 17). In both cases, P7 struggled for some time to reach the far right of the display before allowing the experiment to reduce the range of distance required. The effect of the progression can be seen in Figures 11.1 and 11.2. Figure 11.1(a) shows the proportion of time in each session for which d_{it} was greater than 75% of the maximum measured by the calibration program. This proportion was marginally greater for sessions with progression (p = 0.063). Figure 11.1(b) shows the proportion of time that d_{it} was less than 125% of the calibration minimum. This proportion was not significantly different for sessions with progression (p = 0.14). To further investigate the effects of progression within a session, we considered the first 12 words and averaged across sessions the maximum extension distance used to select a letter in that word (Figure 11.2(a)). We observed a significant upward trend for progression sessions (p < 0.001), but not for sessions without progression (p = 0.56). We also performed the same analysis using the minimum distance used to select a letter (Figure 11.2(b)). The result was a significant linear trend for both progression sessions and sessions with no progression (p = 0.013; p = 0.015). These results show that progression affected P7's performance more in extension than in flexion. This is because P7 was given the choice to select vowels using flexion or extension, and he chose extension 94.7% of the time. There was no difference between progression and no-progression sessions in the proportion of vowels selected using extension (p = 0.79).

The calibration program measured P7's maximum extension and minimum pinch at the beginning and end of each rehabilitation session (Figure 11.3). He was required to hold each of these positions for 5 s. The slope of the maximum extension distance over the 6-week period was not significantly different from 0 for either group of measurements (p = 0.99 for measurements made at the beginning of sessions; p = 0.90 for measurements made at the end). The results were similar for the minimum pinch distance (p = 0.21 for beginning measurements; p = 0.33 for end measurements). The mean maximum extension measured at the beginning of a session was significantly smaller than that measured at the end (p < 0.001). There was no significant difference between the mean minimum pinch measured at the beginning of a session and that measured at the end (p = 0.12).

We computed P7's mean maximum extension and minimum pinch during the no-feedback periods for each session (Figure 11.4). Due to a programming error, no maximum no-feedback extensions were recorded during session 1. Thus, this session was omitted from the relevant analyses. Contrary to our results with P3 and P4, we found a significant upward trend over the 6-week period for the mean no-feedback maximum extension ($p < 0.001$). In addition, we found a significant downward trend with time in the mean no-feedback minimum pinch ($p < 0.001$).

In our analyses for P3 and P4, we computed, for each no-feedback period, the mean over sessions of the maximum extension. We then performed linear trend analysis using this data. This analysis could not be done for P7, because he occasionally failed to hold his maximum extension during a no-feedback period (this happened about once a day). The resulting gaps in the data prevented us from using linear trend analysis. The first no-feedback maximum extension was significantly greater than both the calibration maximum ($p < 0.001$) measured at the beginning of the session and the first with-feedback maximum extension ($p < 0.001$). The first no-feedback maximum was also greater than the maximum extension measured in the last no-feedback period ($p = 0.0086$). In addition, the mean last no-feedback maximum extension was significantly less that mean maximum extension measured by the calibration program at the end of a session ($p = 0.0023$). These relationships are summarized in Figure 11.5. The last no-feedback maximum extension was significantly larger than the last with-feedback maximum extension for sessions without progression ($p = 0.0078$) but not for progression sessions ($p = 0.74$).

Similar analyses of the minimum no-feedback pinch data were performed (Figure 11.6). The mean minimum pinch measured during the first no-feedback period was not significantly different from the calibration minimum measured at the beginning of the session ($p = 0.19$) or the minimum pinch measured during the last no-feedback period ($p = 0.11$). The minimum pinch measured during the last no-feedback period was not significantly different from the minimum measured with the calibration program at the end of the session ($p = 0.33$).

P7 received force feedback from the robots throughout the 6-week rehabilitation protocol. The resistive force experienced in flexion ranged from 0 N at the neutral distance to 2 N at the lower bound of the range of distance mapped to the alphabet. The resistive force experienced in extension ranged form 0 N at the neutral distance to 1 N at the upper bound of the range of distance mapped to the alphabet. For the first five rehabilitation sessions, the direction of the force feedback was computed as described in Chapter 10 for P4. However, P7 moved quickly through the words in the Hangman game, and he was frequently almost finished with a set of three words before the robots had recorded enough data for the least-squares fits to the finger paths. Thus, by the time the robots began to exert force against his fingers, P7 had almost reached the point at which the program shut down to allow him to take a break. Because of this, we decided to determine the direction of the force based only on the positions of the two fingers at a given point in time. The force on each finger was exerted along the line connecting the index finger position with the position of the thumb (Figure 11.7). The direction of each force along this line was chosen such that the user experienced force pushing the fingers together if the distance between them was greater than the neutral distance. If the distance between the fingers was less than the neutral distance, the forces acted to push the two fingers apart. P7 experienced an average force of 0.060 N during flexion and an average force of 0.60 N during extension. No force was exerted by the robots during the calibration program or during the no-feedback periods.

P7's active and passive ranges of motion were measured by an occupational therapist before and after his participation in the experiment. The maximum flexion and minimum extension angles were summed over the MCP, PIP, and DIP joints of each finger, as in Chapter 9 and 10 (Tables 11.1 and 11.2). P7 exhibited greater passive flexion after the 6-week protocol. He also exhibited greater active flexion for all fingers except the index finger. The angle between the index finger and the thumb in the plane of the palm increased from 35° to 80°. The angle between the index finger and the thumb in the plane perpendicular to the palm increased from 40° to 55°. The angles between the index finger and thumb were measured passively. P7 measured an Ashworth 1 for his elbow, wrist, fingers, and thumb both before and after the 6-week protocol.

In terms of the functional tests administered by the occupational therapist, P7 scored a 56/57 on the ARAT before the 6-week rehabilitation protocol and a 57/57 after it. This improvement was clearly limited by a ceiling effect and was not clinically important. P7's score on the Grasp and Pinch subtests of the ARAT is plotted in Figure 11.8 as a function of time. However, P7 scored 36/36 on this subtest the first time he was given it. Due to this ceiling effect, we did not test whether the slope of his data was statistically significant. On the AMAT, P7's mean Functional Ability score was 3.61/5 before participation in the experiment and 4.11/5 after participation. This difference was clinically important. P7's mean Quality of Movement scores were the same as his Functional Ability scores. Before participation in the experiment, P7 rated himself as a 5/10 in both his desire to approach challenging activities and his desire to move onto a harder activity after accomplishing a difficult task.

11.3. Discussion

When we began these experiments, we expected that subjects who experienced the progression protocol might choose more frequently than distortion subjects to contract the range of distance mapped to the screen. Alternately, we thought that they might be highly motivated by observing their progress. Neither of these seemed the case for P7. He usually followed the progression. He chose to contract the distance range only twice and only when he had repeatedly tried without success to reach the vowels on the far right of the screen. As described above, the progression caused a significant upward trend in the maximum distance used to select a letter and also caused P7 to spend a greater proportion of the session time working at more than 75% of the maximum distance measured by the calibration program at the beginning of the session. P7 did not seem to notice when the range of distance required to play the game was increased. When discussing the progression, P7 said he never felt like he was being asked to move farther. This is most likely due to his large JND for distance/position (Chapter 4). Interestingly, P7 also did not visually notice any change in the distance scale shown along the bottom of the alphabet unless his attention was specifically drawn to this change by the experimenter. P7 did not seem very interested in watching his improvement on the distance scale. Thus, P7 was informed that the progression was occurring, but in practice, his perception of the experiment seemed to be little different than that of P3 and P4 in the distortion protocol.

While effects of the progression were seen clearly in extension, P7's performance in flexion was similar on days with and without progression. P7 was shown all 5 vowels on both the far left and the far right of the screen and could choose to select them by moving toward his maximum extension distance or his minimum pinch distance. Effects of the progression were not seen in flexion because P7 almost always chose to select vowels in extension. Discussions with P7 confirmed that he did this because he found the vowels easier to reach in extension. It is possible that reducing the force experienced in flexion might have made him more likely to select vowels in flexion. In the future, for subjects with deficits in both flexion and extension, alternating whether flexion or extension is required to select the vowels might be better than allowing the subject to use either.

Unlike P3 and P4, P7 had to hold his maximum extension and minimum pinch distances for 5 s each during the calibration program. We had hoped that this modification would allow us to observe a trend in these quantities over the 6-week paradigm. Nevertheless, we found no significant slope in the maximum extension or minimum pinch distance over the six-week period. As conjectured for P3 and P4, it is possible that P7 found the calibration program boring and that providing visual feedback during calibration would lead to observable trends over the course of our protocol.

On the other hand, we found a significant upward trend over the 6-week period in the mean maximum extension measured for P7 during no-feedback periods. We also found a significant downward trend in the mean minimum no-feedback pinch over the 6 weeks. These results are encouraging, in that they indicate an improvement in P7's range of motion over the course of the experiment, but they are very different from the significant downward trends in maximum no-feedback extension that we observed for P3 and P4. With our limited data, we have no way of accounting for this contrast, but two differences in the experimental protocols should be noted. P7 had to hold his no-feedback maxima and minima for 2 s each, while P3 and P4 did not. The 2 s hold reduced variation in the no-feedback maxima and minima, making it easier to observe any present trends. In addition, P7 encountered a resistive force in extension, while P3 and P4 were unable to play the Hangman game while experiencing such a force (P4 did experience resisted extension during her last 5 sessions). Resistance in extension has been shown to improve active range of motion in patients more than resistance in flexion

[119]. It is possible that countering resistive force, particularly in extension, helped P7 to strengthen his hand enough that we observed an improvement in his range of motion rather than the decrease in no-feedback performance we saw for P3 and P4, which we attribute to boredom during the no-feedback periods.

Investigation revealed several unintuitive changes in P7's performance over the course of a rehabilitation session. P7's maximum extension increased significantly between the initial measurement made by the calibration program and the first no-feedback period. These measurements were separated only by P7 spelling the word 'DAISY,' which took just 1 or 2 minutes. The reason for this dramatic increase is unclear, though perhaps P7 simply needed a brief "warming-up" period for his paretic hand. P7's maximum extension significantly decreased between the first and last no-feedback periods, which might be expected due to fatigue. However, the maximum extension measured for P7 at the end of the session by the calibration program was significantly greater than that of the last no-feedback period, despite the fact that these measurements were separated in time by only about 30 s. We think that the ending of the Hangman program and the transition to the calibration program may have refocused P7's attention on the extension task, but we can make no definite conclusions. Similar trends were not seen for P7's minimum pinch data, maybe because selecting the vowels in extension focused his attention on that end of his range of motion.

The large increase between the maximum extension measured during the initial calibration and the first no-feedback maximum reflects that P7's performance in calibration was not indicative of his true capabilities. At the end of sessions without progression, P7's maximum with-feedback extension was significantly less than his maximum no-feedback extension. There was no significant difference for sessions with progression. P7's maximum with-feedback extension never became greater than his no-feedback performance because the progression was based on the low maximum extension measured in the initial calibration.

Due to a ceiling effect, the ARAT was not very useful for tracking changes in P7's performance during our experiment. The variation seen in his ARAT subtest score in Figure 11.8 is due mostly to two features of the test. To score a 3 on a given item, P7 had to complete the movement within a given time limit while keeping his back against the back of the chair. Sometimes P7's score was lower because a couple of items were completed, but not within the given time limit. In addition, a high-backed chair was used in the experiment, and so his back left the chair more frequently than if a lower chair had been used.

While we could not observe any functional improvements with the ARAT, we saw a clinically important improvement on our other functional test, the AMAT. In addition the maximum angle between the index finger and the thumb increased both in the plane of the palm and in the plane perpendicular to the palm. These improvements, as well as the increased range of motion noted in the no-feedback measurements made over the 6-week period, imply that practice in our robotic environment increased the physical and functional capacity of P7's hand.

Passive Range of Motion

Finger	Pretest Extension	Pretest Flexion	Posttest Extension	Posttest Flexion
Index	H 10	255	0	265
Middle	H 5	250	0	255
Ring	H 5	240	0	250
Small	H 5	245	0	285

Table 11.1.

Active Range of Motion

Finger	Pretest Extension	Pretest Flexion	Posttest Extension	Posttest Flexion
Index	0	235	0	225
Middle	0	230	0	245
Ring	10	225	0	245
Small	0	235	0	250

Table 11.2.

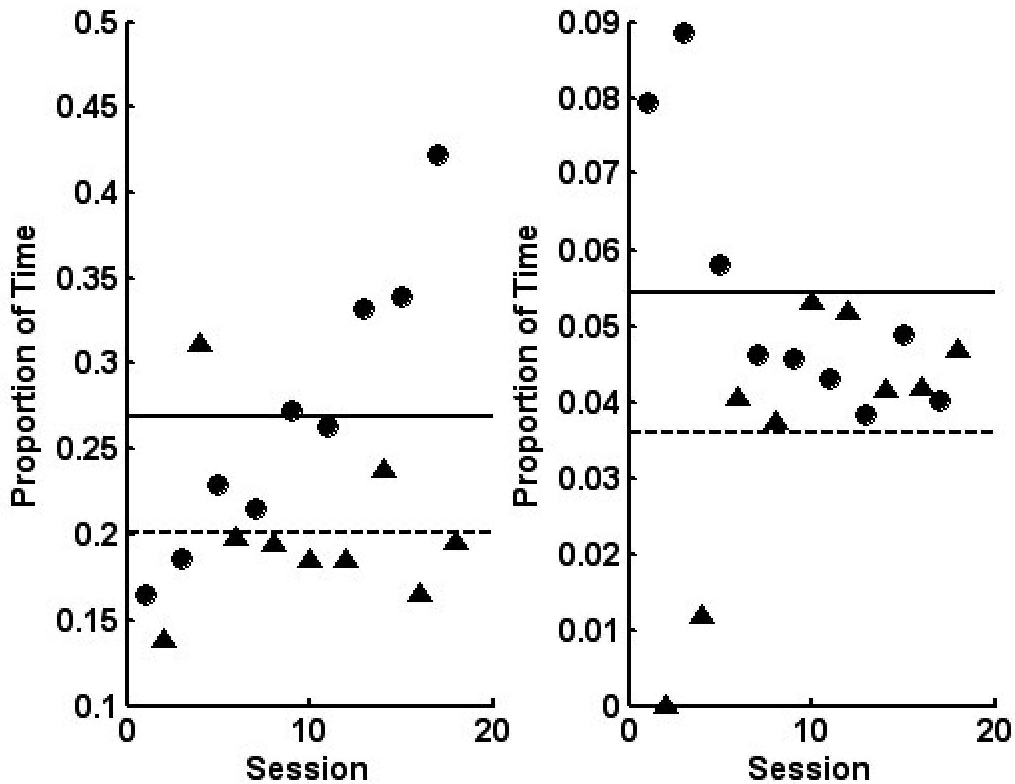


Figure 11.1. **(a)** Relative to the maximum extension distance measured at the beginning of each rehabilitation session, P7 spent more time at greater levels of extension on days in which he experienced progression. The *x*-axis represents session number, and the *y*-axis represents the proportion of each session during which P7's extension was greater than 75% of the maximum extension measured by the calibration program at the beginning of the relevant session. Circles indicate sessions with progression and the solid line, the mean of these sessions. Triangles represent sessions without progression, and the dashed line indicates the mean of the no- progression sessions. The proportion of each session in which P7's extension was greater than 75% of the maximum calibration extension was marginally greater for progression sessions. **(b)** We also examined the proportion of time during which the distance between P7's fingers was less than 125% of the calibration minimum. Symbols are analogous to part (a). Progression did not significantly increase the amount of time that P7 spent at small pinch distances.

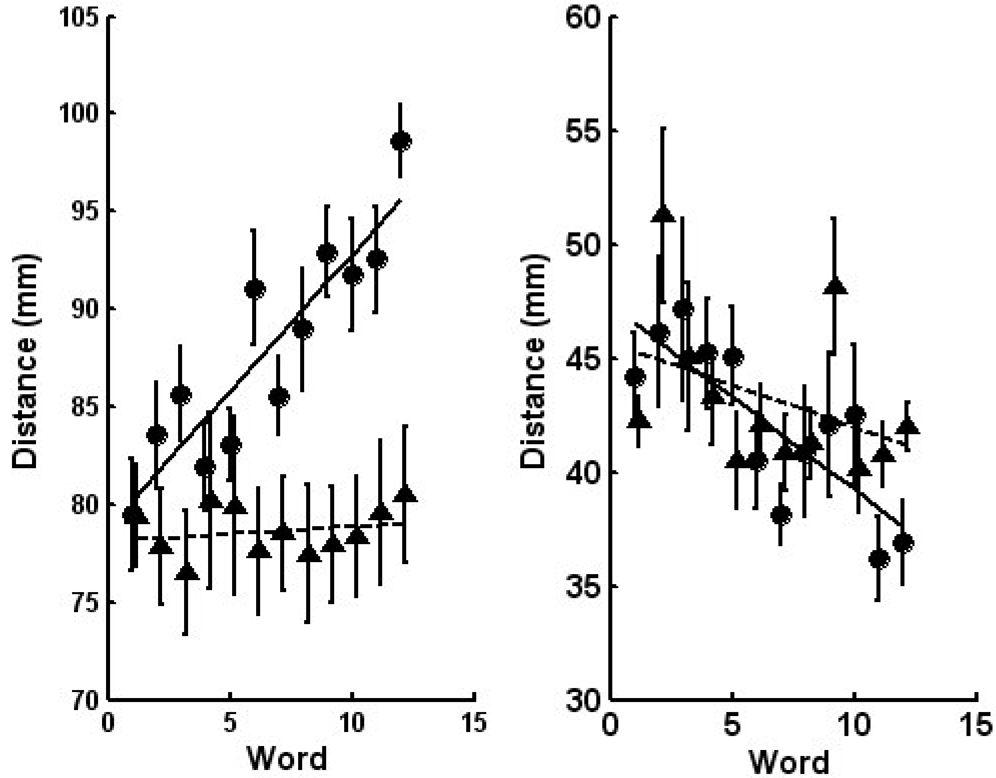


Figure 11.2. **(a)** P7 consistently followed the progression, causing an upward trend in his extension during progression sessions. Circles represent the mean over progression sessions of the maximum extension distance used to select a letter in each word. Triangles represent the same quantity for sessions without progression. No-progression data is slightly horizontally offset from distortion data for clarity. Regression lines for each data set are shown. All error bars represent standard error. A significant linear trend in extension distance with word number was observed for progression sessions, but not for sessions without progression. **(b)** A significant downward trend in the minimum pinch distance used to select a letter was observed for both progression sessions and sessions without progression.

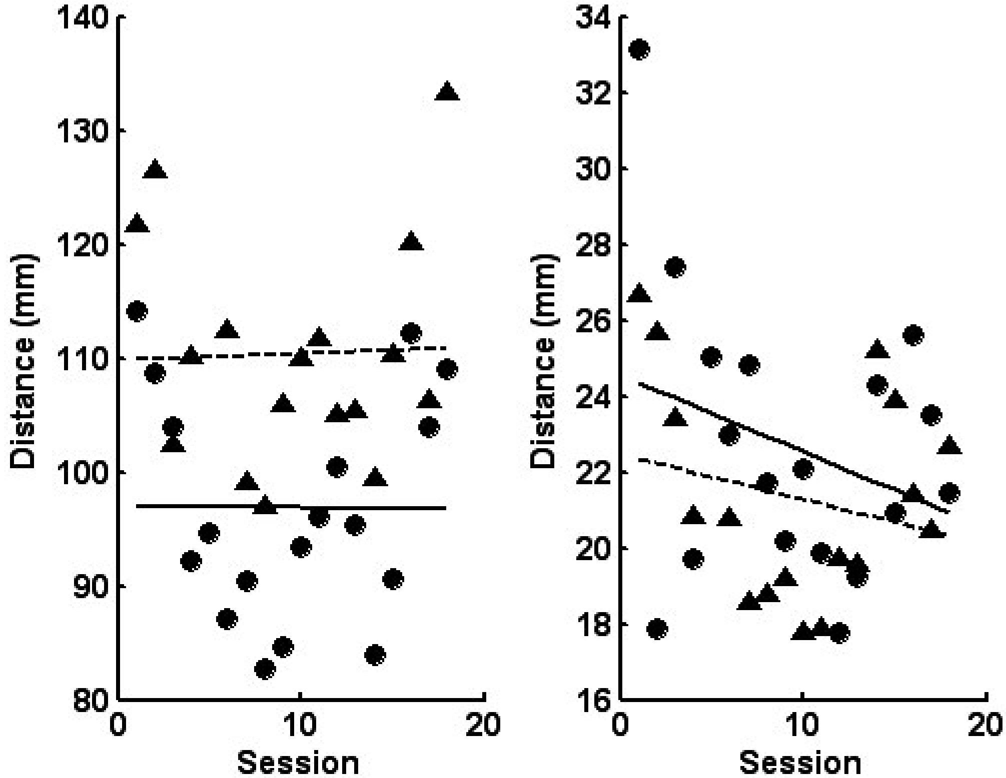


Figure 11.3. **(a)** The maximum extension distance measured for P7 by the calibration program at the beginning (circles) and end (triangles) of each rehabilitation session. A regression line is shown for each set of data (solid for beginning data; dashed for end data). Neither line's slope was significantly different from zero. **(b)** The minimum pinch distance measured for P7 by the calibration program at the beginning and end of each rehabilitation session. Neither line's slope was significantly different from zero.

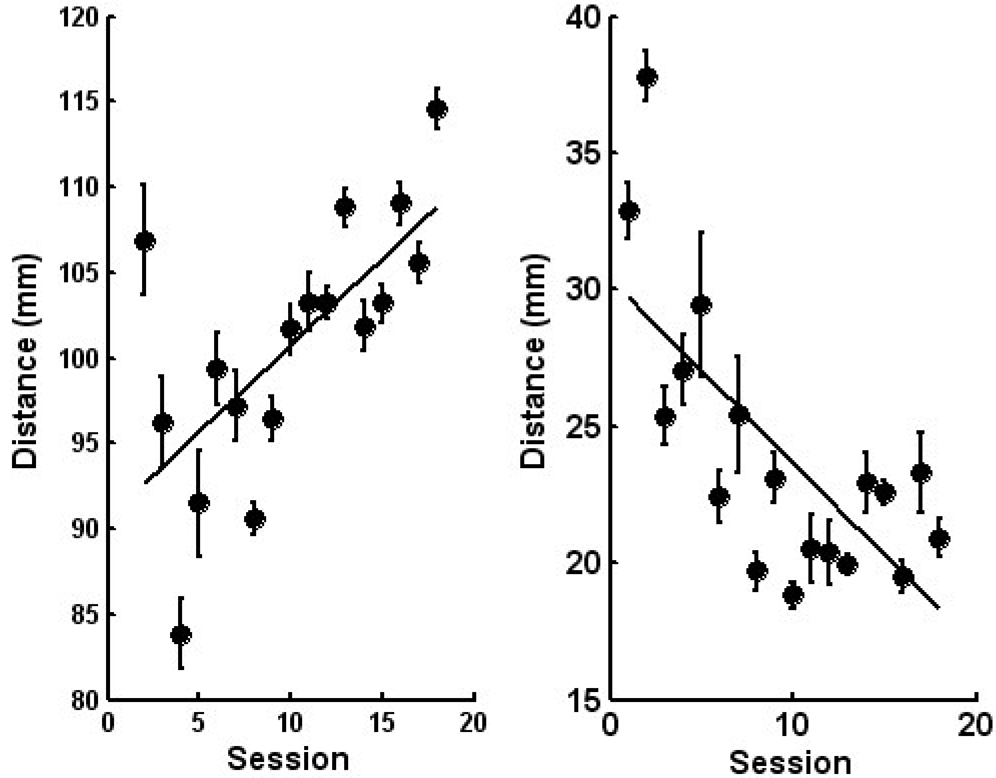


Figure 11.4. **(a)** P7's mean no-feedback maximum extension increased as a function of session. P7's maximum extension was measured during a number of no-feedback periods spaced throughout the Hangman game. Each point represents the mean of these maxima for the given rehabilitation session. Error bars represent standard error. The upward trend indicated by the regression line was significant. **(b)** P7's mean no-feedback minimum pinch decreased as a function of session. Again, the downward trend indicated by the regression line was significant.

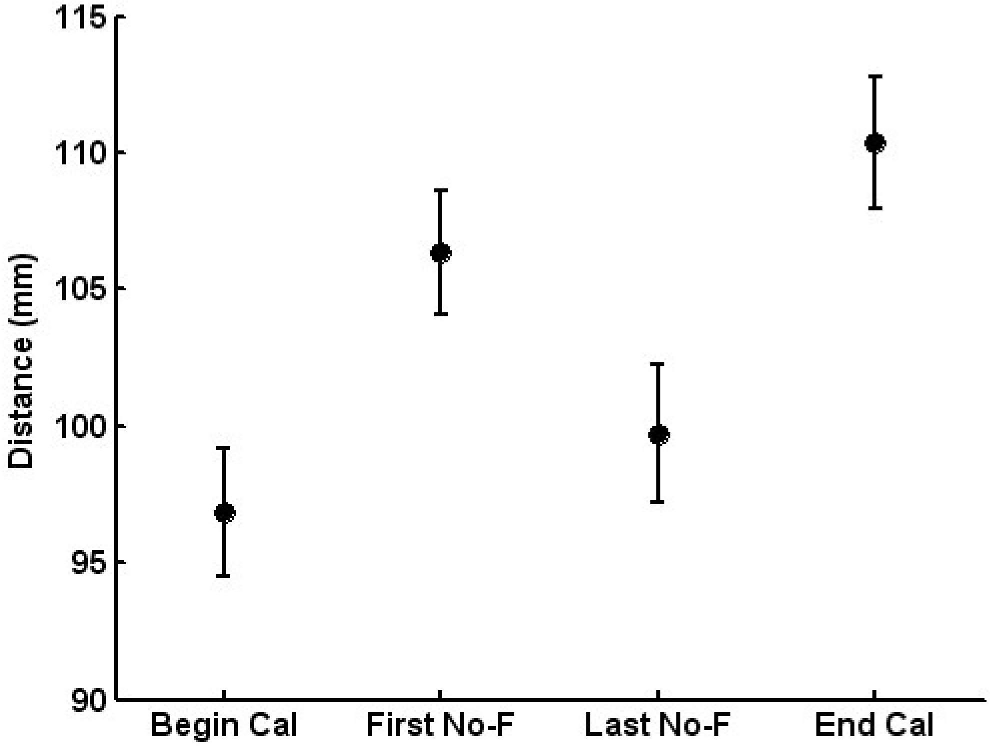


Figure 11.5. The maximum extension distance measured for P7 in the absence of feedback varied dramatically over the course of a rehabilitation session. Circles represent the mean maximum extension at a particular point during the session, while the bars represent standard error. P7's maximum extension during the first no-feedback period was significantly greater than the maximum measured by the calibration program at the beginning of the session and the maximum measured during the last no-feedback period. The maximum extension measured during the last no-feedback period was significantly less than that measured by the calibration program at the end of the rehabilitation session.

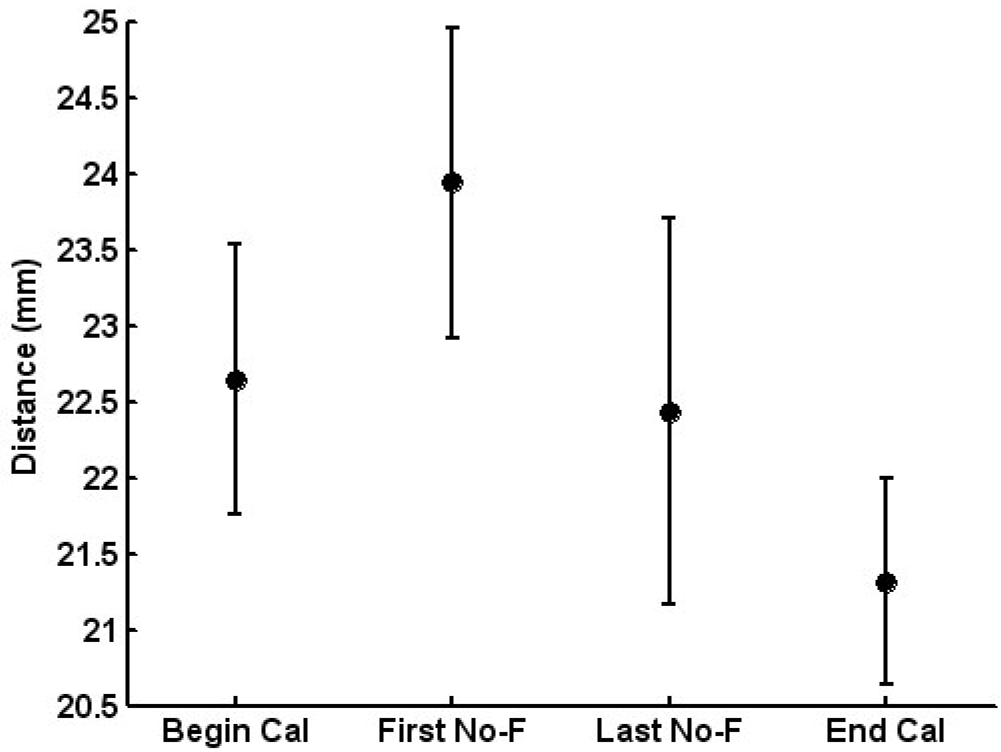


Figure 11.6. P7's minimum pinch performance in the absence of visual feedback was fairly constant over the course of a rehabilitation session. This graph is analogous to Figure 11.5. P7's performance in the first no-feedback period was not significantly different from either his minimum pinch during calibration at the beginning of the session or his minimum during the last no-feedback period. P7's minimum pinch measured by the calibration program at the end of the session was not significantly different from that measured during the last no-feedback period.

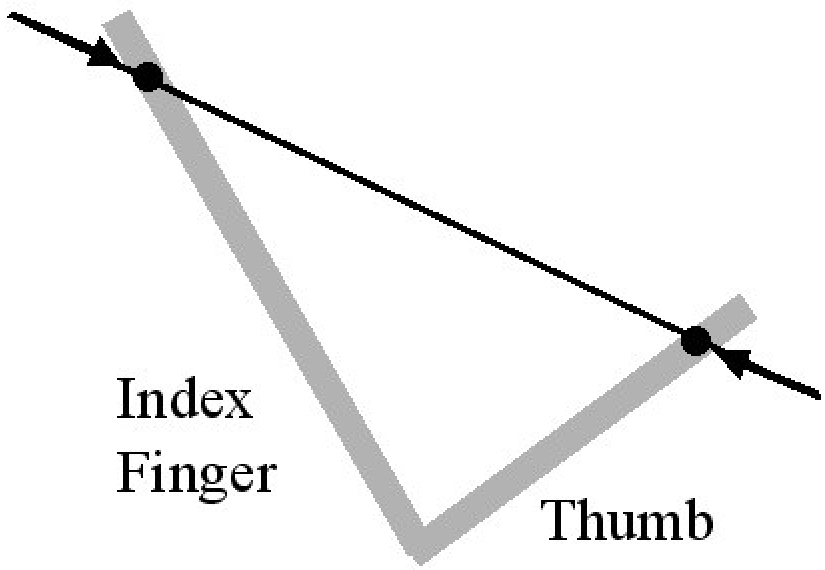


Figure 11.7. The force feedback experienced by P7 during the rehabilitation protocol was such that the force on each finger was exerted along the line connecting the index finger position with the position of the thumb. The direction of each force along this line was chosen so that P7 experienced resistive force both when extending the fingers (shown here) and when pinching.

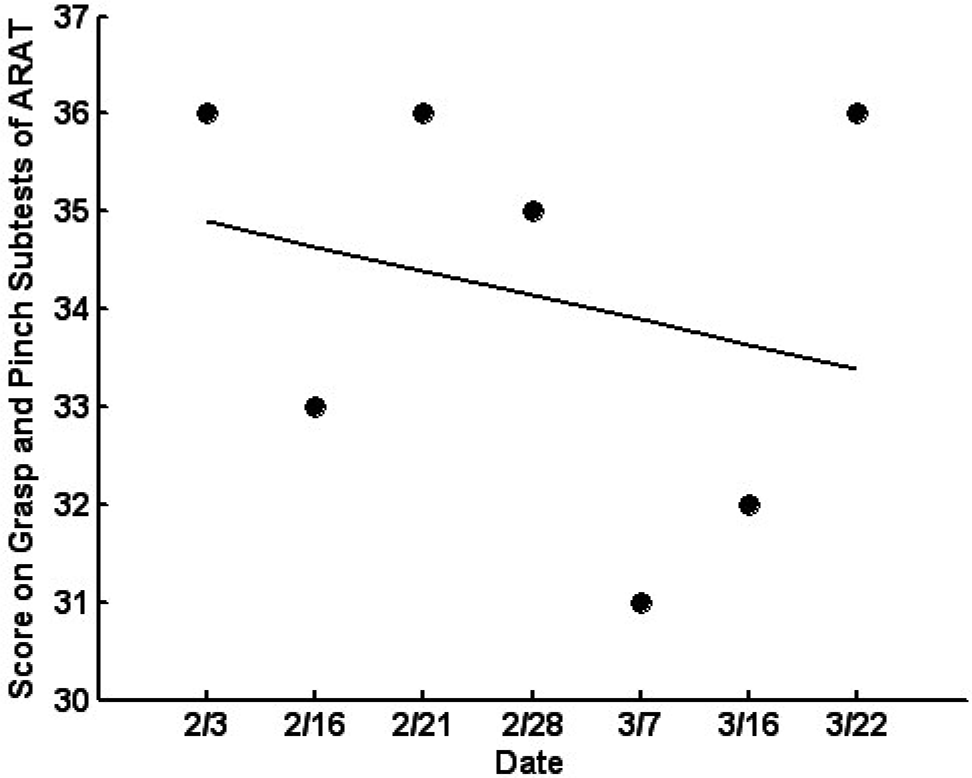


Figure 11.8. Due to a ceiling effect, P7's score on the ARAT subtest did not improve during the six-week participation period.

Chapter 12. Summary of Case Studies

In this chapter, results that were observed for all three chronic motor-impaired subjects are highlighted. At the beginning of each session, each subject was required to extend to 80% of his or her calibration maximum to reach the far right of the list of letters used in the Hangman game. We chose a starting point of 80% of the calibration maximum because we anticipated that a subject would expend considerable effort reaching the calibration maximum and would find it difficult to reach this distance repeatedly. We felt 80% of the calibration maximum would approximate a comfortable maximum distance for the subject. However, all subjects readily followed distortion or progression to levels of extension well above this starting point. A significant upward trend in the maximum distance used to select a letter was observed for all subjects during manipulation sessions (Figure 12.1). By the end of each manipulation session, the subject was reaching or exceeding his or her initial maximum. This is particularly notable for P3 and P4, because they were not required to hold their maximum extensions during calibration, while a given level of extension had to be held 2 seconds to select a letter in the Hangman program.

As mentioned above, if subjects performed maximally during calibration, they would find it difficult to repeatedly produce that range of motion. However, at the beginning of each rehabilitation session, the extension of each subject in the absence of feedback was significantly greater than his or her extension with feedback (Figure 12.2). Even at maximal distortion or progression, the mean no-feedback maximum extension was still similar to the mean with-feedback maximum extension for our subjects. The results of these experiments indicate that a subject's calibration performance did not represent that subject's actual abilities. This is the reason that they were able to follow the visual feedback manipulation to increasing levels of extension. This result recalls Taub's work with Constraint-Induced Movement Therapy (e.g., [120]). This technique addresses learned nonuse by restraining the unaffected limb of hemiplegic patients. Patients are often surprised to find that they can use the impaired arm for many activities that they usually perform with the unimpaired arm. The fact that no-feedback performance is comparable to with-feedback performance at maximal manipulation suggests that our assumption about the starting point for manipulation underestimated the extent to which subjects underperform in calibration. Increasing the starting point for the distortion or progression would allow us to obtain a clearer picture of a subject's actual abilities and the effects of visual feedback manipulation. In general, the underperformance of subjects during calibration has important implications for rehabilitation in general, and for robotic rehabilitation in particular. Robotic rehabilitation systems must be calibrated for each individual, and the activities a patient is asked to perform are based on this calibration. These preliminary data suggest that a patient's performance during calibration may not connote his or her actual abilities and may be a poor guide for determining what a patient should be asked to do in rehabilitation.

Previous applications involving visual distortion in rehabilitation (see Chapter 2) relied on the effects of distortion being maintained after the conclusion of the rehabilitation session. In part, this is true of our paradigm as well. Ideally, a subject who follows the visual manipulation to obtain a larger range of motion will continue to utilize the larger range after the conclusion of the rehabilitation session. This is likely, given that the effects of distortion have been found to wash out more slowly in stroke patients [84, 86]. However, the main goal of the manipulation is to encourage the subject to perform at a higher level during the rehabilitation session. We expect that even if the effect of the manipulation washes out, practicing at a higher level during each rehabilitation session will improve the outcome of rehabilitation.

Exercise in our environment led to clinically significant functional improvements in all three of our subjects, despite problems in standardizing the scoring of the occupational therapy tests. Our data cannot show how these functional improvements were related to the visual feedback manipulation, but practice in our robotic environment does transfer to real-life grasping tasks. This is important, because, as noted in Chapter 2, few functional gains have been measured after rehabilitation in a robotic environment. Future experiments are needed to investigate whether visual feedback manipulation can lead to greater functional gains than practice without feedback manipulation.

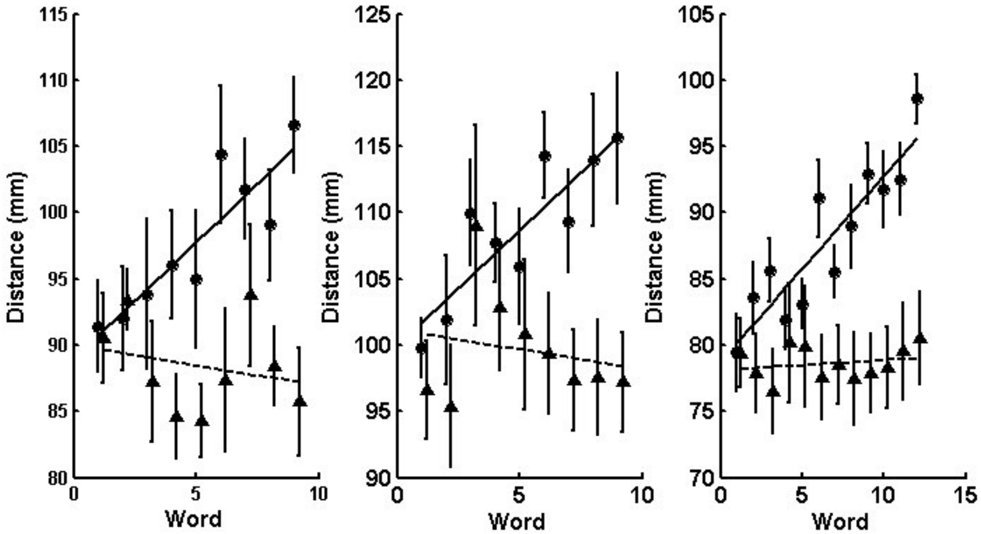


Figure 12.1. All subjects consistently followed the visual feedback manipulation, causing an upward trend in extension during manipulation sessions. From left to right, the graphs correspond to subjects P3, P4, and P7. Circles represent the mean over manipulation sessions of the maximum extension distance used to select a letter in each word. Triangles represent the same quantity for sessions without visual feedback manipulation. No-manipulation data is slightly horizontally offset for clarity. Regression lines for each data set are shown. All error bars represent standard error. For each subject, a significant linear trend in extension distance with word number was observed for manipulation sessions, but not for sessions without manipulation.

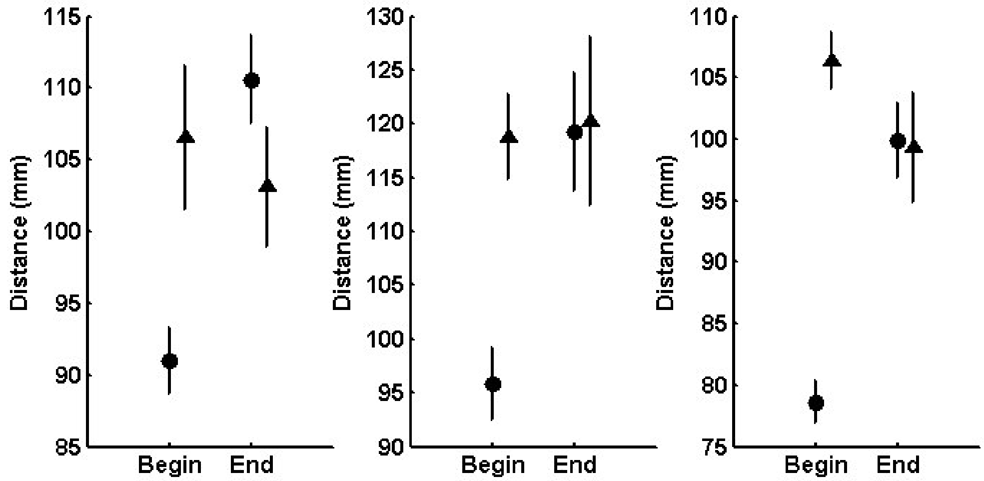


Figure 12.2. From left to right, the graphs correspond to subjects P3, P4, and P7. The first set of points in each plot represents the subject's with-feedback (circle) and no-feedback (triangle) maximum extensions at the beginning of a rehabilitation session. For all subjects, the no-feedback extension was significantly greater. The second set in each plot compares the with-feedback and no-feedback maximum extensions at the end of manipulation sessions (maximal manipulation). There was no significant difference for any subject, though P3's with-feedback extension was marginally greater than her no-feedback extension.

Chapter 13: Conclusions

13.1. Summary

The main contribution of this dissertation was to develop a body of work that provides a foundation for the use of visual feedback distortion in rehabilitation (Figure 1.1). First, we measured the Just Noticeable Differences for force and distance/position to find the lower bound for the amount of visual distortion that is imperceptible to subjects. We measured the JNDs for force and distance/position to be 19.7% and 3.99 mm, respectively, for young subjects. We found that age significantly increased the force JND to a value of 31.0% while marginally increasing the distance/position JND to 6.32 mm. Furthermore, we measured the JNDs for 4 motor-impaired individuals and found that these JNDs were often dramatically larger than those of the appropriate age-matched control group.

Next, we measured the effects of goal distortion on force production. Subjects attempted to produce consistent levels of force while the visual display was distorted in such a way that a given force value appeared progressively smaller with time. We found that subjects based their responses primarily on the visual display, even when told to attend to the force experienced with the hand. This result was observed for both young and elderly unimpaired subjects, as well as 4 subjects with chronic stroke or TBI. Similar results were also observed for distortion that encouraged increased movement distance. A total distortion of 2 JNDs was reached via a series of steps, and subjects did not notice the distortion. In fact, even when the subject was informed that the display might be distorted, his or her performance was still determined by the visual display. This point is particularly important when considering the use of visual distortion in a multi-week rehabilitation paradigm since over time, stray comments or mechanical glitches might make the patient suspect distortion.

Finally, we measured the effects of error enhancement on a coordination task involving the index finger and thumb. The subject was asked to produce a sinusoidal pattern of movement with both fingers while the error of either finger or both was visually enhanced. We found subjects were able to learn to coordinate the index finger and thumb in this way. When the error of only one finger was distorted, the subject's attention was drawn to that finger. This type of distortion did not improve subject performance when the errors of both fingers were distorted.

We used this body of work to design a paradigm to test the effects of visual feedback manipulation on performance in rehabilitation. This paradigm consisted of 18 rehabilitation sessions over 6 weeks; during each session, the subject used the index finger and thumb to play a game of Hangman using the robots. During odd-numbered sessions, the visual display was gradually changed so that a greater range of motion was required to play the game.

Three subjects participated in an initial test of our rehabilitation paradigm. Two subjects participated in a visual distortion protocol, and one participated in a visual progression protocol. These protocols were identical, except that the visual progression subject was informed that the display was changing. All three subjects showed an upward trend in maximum extension distance with manipulation. This improvement within sessions including visual manipulation was possible because a subject's performance during calibration was not indicative of his or her actual abilities. All three subjects showed functional improvements after participation in the study.

13.2. Future Work

The results of our initial tests were positive, but clinical trials with more patients are needed to confirm that visual feedback manipulation has a positive effect on performance in rehabilitation. Patients receiving manipulation at every rehabilitation session should be compared to patients who receive no manipulation. This subject design was not possible here due to the limited number of available subjects. Comparing the control group receiving robotic rehabilitation without visual feedback manipulation to a control group of subjects who practice a similar range of motion task without the robots would allow conclusions to be drawn about the importance of the game-like robotic environment. In addition, a study of how performance in calibration relates to actual capacity is needed to investigate how much manipulation should be used in therapy. Future patient work will also be needed to examine whether differences exist between visual distortion and visual progression. Assuming that clinical trials show visual feedback manipulation to have a positive effect on the outcome of rehabilitation, the next step would be to disseminate the technique to rehabilitation clinics on a large scale. A simplified, less expensive system utilizing visual feedback manipulation would be the best choice for large-scale use.

Appendix A. Inclusion/Exclusion Criteria for Study Participants

Traumatic Brain Injury Subjects:

Inclusion Criteria

- Age between 18 and 65
- Muscle strength at least Grade 3/5 but no greater than 4/5 muscle strength in the majority of the following muscle groups: wrist extension, elbow flexion and extension, and finger flexion.
- Minimal 20 degrees of isolated wrist extension and finger extension [47, 121].
- Greater than 12 months out from TBI.
- Subjects must have difficulty with object manipulation- as defined by an 1) inability to pick up a glass drink from it and replace it with the affected upper extremity, 2) removing and replacing a peg from a dowel, 3) use affected hand to open and close clothes peg (adapted from prior studies) [47, 121].
- Not actively engaged in Physical or Occupational therapy.
- Subjects must have some spasticity as measured by Ashworth score (1-2) and must be stable (defined by no change in Ashworth score).
- Stable antispasticity and psychoactive medication regimens for at least 6 weeks prior to study start.

Exclusion Criteria

- Ashworth scale tone of 3 or greater
- Blindness or visual field not sufficient for testing.
- Inability to follow commands as specified in the task.
- Seizure occurring in the previous 12 months.

Stroke Subjects:

Inclusion Criteria

- Minimal 20 degrees of isolated wrist extension and finger extension [47, 121].
- Greater than 12 months from stroke incident.
- Subjects must have difficulty with object manipulation- as defined by an 1) inability to pick up a glass drink from it and replace it with the affected upper extremity, 2) removing and replacing a peg from a dowel, 3) use affected hand to open and close clothes peg (adapted from prior studies) [47, 121].
- Not actively engaged in physical or occupational therapy.
- Subjects must not have excessive spasticity (resistance to movement), as measured by an Ashworth scale tone of 3 or greater
- Stable antispasticity and psychoactive medication regimens for at least 6 weeks prior to study start.
- Visual field sufficient for testing.
- Ability to follow commands as specified in the task and give informed consent.
- No seizures occurring in the previous 12 months.

Appendix B. Tests of Hand Function

Arm Motor Ability Test

Instructions

The test consists of 13 tasks that are subdivided into a total of 27 items. For each task, the subject is given very specific instructions. These instructions reference a template that is placed on the table so that all objects used in the test can be placed in a repeatable position. For example, for task 1, a subject with paresis in the right (formerly dominant) hand is instructed as follows:

"Begin with both forearms resting on the outer lines on either side of the plate before you. When I say 'go,' pick up the knife in your right hand and the fork in your left hand. Then cut two pieces of the 'meat' [really Play-doh] on the plate along the indicated lines. You can work on the cutting part of the task for up to two minutes. When you have completed cutting the 'meat,' put the knife down, switch the fork into your right hand, and use it to raise one piece of 'meat' to your mouth without touching your lips. Ready, set, go [114]."

Each part of a task is timed and has a time limit of one or two minutes. Each part of a task is given two scores from 0 to 5. One rates the functional ability of the subject for that subtask, how well the subject was able to accomplish the goal for that task. The other score rates the quality of the subject's movement, how smooth and precise his or her movements are.

The form that follows lists the tasks of the AMAT. This form was obtained from the University of Pittsburgh.

Activity	Time (s)	Functional Ability	Quality of Movement
I. Cut "meat"			
1. Pick up fork and knife			
2. Cut "meat"			
3. Fork to mouth			
II. Foam "sandwich"			
4. Pick up foam "sandwich"			
5. "Sandwich" to mouth			
III. Eat with spoon			
6. Pick up spoon			
7. Pick up dried kidney bean with spoon			
8. Spoon to mouth			
IV. Drink from mug			
9. Grasp mug handle			
10. Mug to mouth			
V. Comb hair			
11. Pick up comb			
12. Comb hair			
VI. Open jar on table			
13. Grasp jar top			
14. Screw jar top open			
VII. Tie shoelace			
15. Tie shoelace			
VIII. Use telephone			
16. Phone receiver to ear			
17. Press phone number			
IX. Wipe up spilled water			
18. Wipe up water (6 mov'ts)			
19. Discard towel in wastebasket			
X. Put on cardigan sweater			
20. Affected arm in sleeve, sweater over affected shoulder			
Finish donning sweater			
21. Button two lower buttons			

Activity	Time (s)	Functional Ability	Quality of Movement
XI. Put on t-shirt			
22. Arms in t-shirt sleeves			
23. Head through neckhole			
24. Pull down and straighten shirt			
XII. Prop on extended arm			
25. Prop on extended affected arm, reach across body with unaffected arm, pick up small object			
XIII. Light switch/door			
26. Pincer grasp of light switch and flip down			
27. Grasp door handle, rotate handle, open door 6 inches			
28. Close door			
SCORE			

Action Research Arm Test

Instructions:

There are four subtests: Grasp, Grip, Pinch, Gross Movement. Items in each are ordered so that:

- if the subject passes the first, no more need to be administered and he scores top marks for that subtest;
- if the subject fails the first and fails the second, he scores zero, and again no more tests need to be performed in that subtest;
- otherwise he needs to complete all tasks within the subtest

Each item is scored from 0-3 as follows:

- 0 = no movement
- 1 = movement partially performed
- 2 = movement abnormally performed (movement time greater than time limit or patient loses contact with the back of the chair)
- 3 = movement normally performed (movement time less than or equal to time limit)

The first part of the above information and the ARAT form that follows are taken from [122]. The scoring information was obtained from [117] and training provided by Dr. Skidmore of UPMC.

Activity	Time (s)	Time Limit (s)	Score (0-3)
Grasp			
1. Block, wood, 10 cm cube		4.2	
(If score = 3, total = 18 and go to Grip)			
Pick up a 10 cm block			
2. Block, wood, 2.5 cm cube		3.6	
(If score = 0, total = 0 and go to Grip)			
Pick up 2.5 cm block			
3. Block, wood, 5 cm cube		3.5	
4. Block, wood, 7.5 cm cube		3.9	
5. Ball (Cricket), 7.5 cm diameter		3.8	
6. Stone 10 x 2.5 x 1 cm		3.6	
Grip			
1. Pour water from glass to glass		7.9	
(If score = 3, total = 12, and go to Pinch)			
2. Tube 2.25 cm		4.2	
(If score = 0, total = 0, and go to Pinch)			
3. Tube 1 x 16 cm		4.3	
4. Washer (3.5 cm diameter) over bolt		4.0	
Pinch			
1. Ball bearing, 6 mm, ring finger and thumb		4.4	
(If score =3, total = 18 and go to Gross			
Movement)			
2. Marble, 1.5 cm, index finger and thumb		3.8	
(If score = 0, total = 0 and go to Gross			
Movement)			
3. Ball bearing, middle finger and thumb		4.1	
4. Ball bearing, index finger and thumb		4.0	
5. Marble, middle finger and thumb		3.8	
6. Marble, ring finger and thumb		4.1	
Gross Movement			
1. Place hand behind head		2.7	
(If score = 3, total = 9 and finish)			
2. (If score = 0, total = 0 and finish)			
3. Place hand on top of head		2.7	
4. Hand to mouth		2.4	

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