Perceptual Limits for a Robotic Rehabilitation Environment Using Visual Feedback Distortion

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***Abstract*—Imperceptible visual distortion, in the form of a disguised progression of performance goals, may be a helpful addition to rehabilitation after stroke and other brain injuries. This paper describes work that has been done to lay the groundwork for testing this hypothesis. We have constructed and validated an experimental environment that provides controllable visual distortion and allows precise force and position measurements. To estimate the amount of visual distortion that should be imperceptible, we measured the limits for force and distance/position perception in our rehabilitation environment for young and elderly unimpaired subjects and for a single traumatic brain injury (TBI) patient. We found the Just Noticeable Difference (JND) for produced force to be 19.7% (0.296 N) and the JND for movement distance/finger position to be 13.0% (3.99 mm) for young subjects (ages 18–35). For elderly subjects (ages 61–80), the JND for force was measured to be 31.0% (0.619 N) and the JND for distance/position was 16.1% (5.01 mm). JNDs of 46.0% (0.920 N) and 45.0% (14.8 mm) were found for the motor-impaired individual. In addition, a subject's rating of effort was found to be profoundly influenced by visual feedback concerning the force magnitude. Even when this feedback was distorted, it accounted for 99% of the variance of the effort rating. These results indicate that substantial visual distortions should be imperceptible to the subject, and that visual feedback can be used to influence the subject's perceived experience in our robotic environment. This means that we should be able to use imperceptible visual distortion to alter a patient's perception of therapeutic exercise in a robotic environment.**

***Index Terms*—Feedback distortion, kinesthetic perception, perception of effort, rehabilitation robotics.**

I. Introduction

ROBOTIC therapy has been shown to increase strength, range of motion, and possibly functional performance in stroke patients [1], [2]. Such improvements are hypothesized to result from robotic therapy's focus on intensive, repetitive movements. Intensive practice increases use of an impaired limb in daily life [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, Burdea *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 redress psychological influences that may impede therapy.

A fundamental assumption of our work is that a patient's performance in the rehabilitation environment is influenced not only by overt goals established with a therapist, but by the patient's own perception of his or her performance. During therapy, patients receive kinesthetic input concerning joint positions and exerted forces. They may also perceive accumulated effort through signals of fatigue, or they may observe environmental outcomes such as success or failure at a target task. Importantly, at least some patients may use these signals to self-impose limits on their performance [6]–[9]. Our goal is to distort the perception of performance to prevent a patient from setting limits below his or her actual capabilities. Our method of distortion first sets a visual metric to measure performance, then gradually adjusts the level of objective performance that is necessary to achieve a given visual outcome. Such distortion might be called "disguised progression." It is different from distortions such as prism lenses that directly change the mapping between movement and perception. Though visual feedback distortion could be applied to any rehabilitation task, we plan to focus on tasks designed to improve fine motor control and function in the hand. Hand function has been shown to be important in predicting patient ability to carry out self-care activities [10], but robotic therapy has been infrequently applied to this domain.

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 [11], which could reduce the influence of visual distortion on a patient's perception of therapeutic exercise. Since the visual outcome signal that we use is task-specific and not an intrinsic result of performance, the gradual progression can only be detected if patients 1) map the outcome metric onto intrinsically perceptible results of their actions (e.g., exerted forces or joint angles) and 2) detect changes in the mapping as the performance period progresses. We seek to identify circumstances in which this detection will not occur.

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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 Difference (JND) 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 a motor-impaired individual.

While it is necessary to identify quantifiable factors such as the force and distance/position JNDs, a patient's qualitative perceptions of subjective factors may also influence our paradigm. A subject may be unable to immediately detect a distortion of 10% with regard to force, but over a period of time, may notice that they are exerting themselves more than expected. This expense of "effort" is defined in terms of metabolic expense or work output, including cardio-respiratory variables and other metabolic measurements [12]. We performed an experiment to investigate whether we could influence a subject's rating of effort using visual distortion. This experiment tests whether or not distortion can affect an individual's subjective experience of a task in a robotic environment. The results should indicate whether or not we can expect visual distortion to influence a patient's experience during rehabilitation in a robotic environment.

This paper describes the hardware and software details of an experimental environment suitable for testing the effects of feedback distortion on performance of fine motor control tasks; experiments we performed to validate our system will also be discussed. The JNDs for produced force and distance/position for the index finger in this environment are reported. Finally, we relate how accurately and consistently a subject could judge his or her sense of effort and whether that judgment could be influenced using visual distortion.

II. Experimental Environment

A. Hardware

The experimental environment used in the experiments described here consists of haptic and visual displays, as shown in Fig. 1. The subject moves his or her index finger against a resisting force while receiving visual feedback on the computer screen. The force feedback is provided by a Premium 1.5 model PHANTOM robot.[1] This robot has three active degrees of freedom (DOF) and can measure position to within 0.03 mm. The maximum exertable force is 8.5 N, while the largest continuously exertable force is 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

[1]SensAble Technologies, Inc., Woburn, MA; http://www.sensable.com

Fig. 1. Haptic and visual displays that compose our experimental environment. The restraints that isolate the movements of the index finger and the restraint that prevents the hand from tilting are shown. The screen that concealed the hand during experimental trials is not shown.

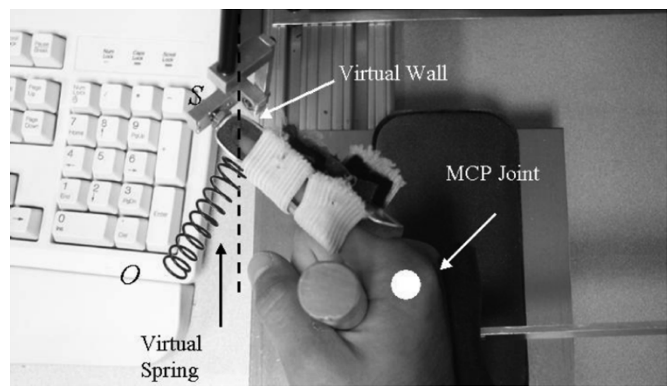

Fig. 2. Our software simulates a virtual compression spring between O and the index finger. A virtual wall is placed at a location 65 mm from O (the point S). The rest length of the spring is SO. The custom-made finger cuff is shown, and the MCP joint is indicated.

Fig. 2. Our finger cuff adjusts easily to fit a wide range of finger sizes and restrains the finger so that it moves only about the metacarpophalangeal (MCP) joint. Three pairs of ball bearings give this finger cuff three passive DOF. When combined with the three active DOF inherent to the robot, the six DOF of our system allow the finger to move comfortably in any direction within the robot's workspace. The total weight of the finger cuff is 32 g. The finger cuff has been tested with subjects ranging in age from 18 to 81, including one traumatic brain injury (TBI) patient, 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 instruct the subject to grasp a post with the remaining fingers and thumb throughout the experiment (Fig. 1). In addition, the subject's forearm is 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 is placed against the back of the hand (Fig. 1). A screen conceals the subject's hand throughout each experiment.

B. System Software

Our software consists of a graphics thread that runs at 30 Hz and a haptics thread that runs at 1 kHz. This program is written in Visual C++ and uses the GHOST tool kit provided by SensAble Technologies. This tool kit provides 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, provides no information about the actual forces generated by the robot; only the commanded forces are known. To calibrate the relative encoders, the tool kit assumes that the robot is initialized at a particular angle, and the software uses a Cartesian coordinate frame with the origin located at the endpoint position of the robot during initialization. In this frame, the y axis is vertical and the xz plane is horizontal. All position and force information is given in terms of this Cartesian coordinate frame.

Each subject begins a session in our environment by moving back and forth between the two positions marked O and S in Fig. 2. Point O is the position of the tip of the index finger when the finger is fully flexed, and point $S = (x_S, y_S, z_S)$ is the point at which the subject has extended the finger a Euclidean distance of 65 mm from O. The distance of 65 mm was chosen because it is a distance that subjects can comfortably extend the index finger without shifting the hand position. While the subjects moves between S and O, the path of the finger is recorded. The path is an arc between S and O that lies largely in the xz plane.

In our experimental environment, we created a virtual compression spring and a virtual wall, as shown in Fig. 2. One end of the virtual spring is fixed at the point marked O, and one end is attached to the subject's index finger. We chose to make our force simulation a virtual spring because as the subject moves the index finger, the virtual spring provides a continuously varying force that is intuitive for the subject. With the virtual spring, we can examine both force production and range of motion for a subject while varying the relationship between them. We have implemented a virtual wall at $x = x_S$ that discourages the subject from shifting the hand to extend the finger past this vertical plane. The subject can move only between O and the virtual wall at $x = x_S$; the values of y_S and z_S are reset each time the subject reaches 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 is based on the distance that the fingertip has moved from S. This arc length is 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 is the rest length of the virtual spring, and the force experienced by the subject is

$$F = k\sqrt{(x - x_S)^2 + (y - y_S)^2 + (z - z_S)^2} \tag{1}$$

where k represents the spring constant and (x, y, z) is the current 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 requires a model of the finger path. The component of the path that lies in the xz plane (the horizontal plane) and the component that lies in the y direction are considered separately. The horizontal path component is found by approximating z as a quadratic function of x, $z = ax^2 + bx + c$. The coefficients a, b, and c are computed using the method of least-squares. This least-squares fit is initialized with an array of 333 position data points at the beginning of the experiment. The use of a least-squares fit adjusts the force assignment for each subject's individual finger path, and the least-squares fit is updated throughout the experiment to account for any slight changes in the finger path that may occur. Every time the subject's index finger moves 2 mm in the xz plane, a new data point is added to the position array and the oldest data point in the array is deleted. Then the least-squares fit is recalculated, yielding a new set of coefficients.

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

Thus, the path of the finger can be modeled by

$$\vec{P} = x\hat{\imath} + m\sqrt{(x - x_S)^2 + (z - z_S)^2}\hat{\jmath} + (ax^2 + bx + c)\hat{k} \tag{2}$$

where $\hat{\imath}$ is the unit vector in the x direction, $\hat{\jmath}$ is in the y direction, and $\hat{k}$ is in the z direction.

Substituting for z in the y component and taking the derivative with respect to x, we find that the tangent to the finger path lies along the vector

$$\vec{t} = \hat{\imath} + \frac{m\left((x - x_S) + (2ax + b)(z - z_S)\right)}{\sqrt{(x - x_S)^2 + (z - z_S)^2}}\hat{\jmath} + (2ax + b)\hat{k}. \tag{3}$$

By normalizing this vector, we can find the direction in which the force should be applied. We combine this with the previously calculated force magnitude in (1), and the result is a virtual spring with the force applied tangent to the path of the finger.

C. Validation

To test the hypotheses in our experimental environment, we must know precisely the forces produced by the subject. Rather than add a force sensor to the finger cuff, which would add unnecessary inertia, we chose to investigate how accurately the commanded force predicts 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 $\pm$ 0.0069 N (mean $\pm$ standard error). None of our experiments use forces or force differences under 0.4 N. Therefore, we determined that the commanded force of the PHANTOM estimates 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 is chosen to be tangent to the path of the finger. We assessed the performance of this algorithm using position and force data recorded from ten young subjects (ages 18–35) and ten elderly subjects (ages 61–81) selected at random from participants in the JND experiments described later. 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 between the 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 ten young subjects, the mean angular error between the true path tangent and the applied force vector was 7.85° $\pm$ 1.33° (mean $\pm$ standard error). The mean angular error for the ten elderly subjects was 7.96° $\pm$ 1.24°. These errors were not significantly different ($p = 0.951$). The percentage of the applied force that lies along the true path tangent is given by the cosine of the angular error. Combining the young and elderly data, we obtain an overall mean angular error of 7.91°. The cosine of this angle is 0.991, which tells us that only 0.9% of the force is applied in a direction other than the true path tangent. Our algorithm approximates the path tangent extremely well, and it performs equally well for young and elderly subjects.

III. Methods

We conducted three experiments in this environment to test subjects' perception of force, distance/position, and effort. The first experiment measured the JND for force. The second measured the JND for finger position or movement distance. The third experiment measured the effects of visual distortion on subjects' perceived effort. 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 a single TBI patient are also reported. This patient, denoted SKL, was a 34-year-old female who was eight years post-injury. She performed the experiments with her left hand, which she had limited ability to move. SKL had no voluntary movement in her right hand. She was cognitively intact. All experiments were approved by the Internal Review Board of the university, and all subjects gave informed consent.

A. 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 (six females and six males), ten elderly subjects (five females and five males), and SKL 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 ($p = 0.268$). This experiment consisted of 100 trials. A break lasting at least 4 min was given every 25 trials. On each trial, the subject sampled two forces. The subject began at the point S and flexed the finger against the virtual spring until he or she reached the base force F_0. The subject exerted this force for 2 s 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 SKL, and her ΔF was 0.8 N. 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 SKL because we anticipated that their JNDs would be larger, and the method of analysis requires that they be able to discriminate F_0 from $F_0 + \Delta F$ 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 SKL 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. 3(a)] 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.

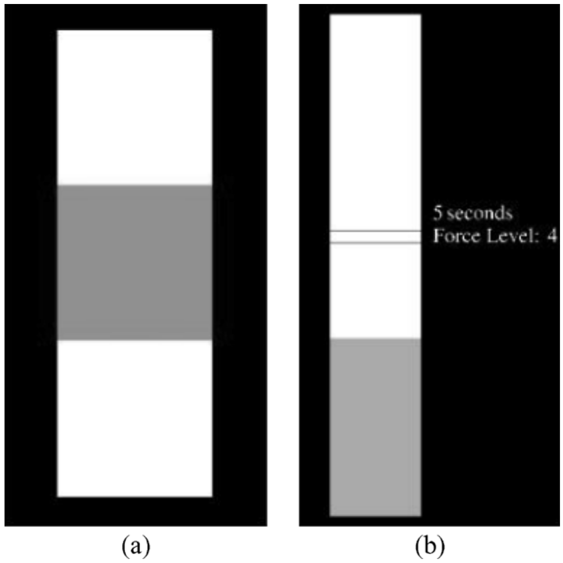

Fig. 3. (a) Visual display used for the force and distance/position JND experiments. The middle box was shaded when the subject was at the target value for force or distance, respectively. (b) The visual display used for the effort experiment. The subject sampled the target force by keeping the edge of the shaded box between the two lines for a specified amount of time. Information concerning the time period and the force level was not shown for control subjects.

For young subjects, the window was 1% of the target force. For elderly subjects, 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 JND for force using the method described by Berliner and Durlach [14] 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 following:

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

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

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

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.

B. *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 (two females and nine males), 11 elderly subjects (five females and six males), and SKL 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 ($p = 0.379$). 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 $D_0 = 30$ mm. The subject stayed at this displacement for 2 s 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 2 s, 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 SKL. 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 SKL 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 earlier.

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 $\pm 1\%$ of the target displacement, and elderly subjects 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.

C. Perceived Effort Experiment

This experiment was performed to investigate the effect of visual feedback on subjects' ratings of effort. Seventeen young subjects participated in this experiment: eight control subjects (three females and five males) and nine experimental subjects (five females and four males). Control subjects completed 70 trials, while experimental subjects completed 140 trials. On each trial, the subject started at S and flexed the finger against the virtual spring until he or she reached one of five possible target forces (within a force window of target force ±1%). The subject stayed at the target force for a predetermined time period ranging from 2 to 10 s. The subject was then asked to rate how much "effort" he or she expended while in the target zone, using a ten-point scale (1 = minimum effort, 10 = maximum effort). "Effort" was defined in the instructions as "sense of energy usage, exertion, or fatigue during the time that you were holding your finger against the haptic device." Neither force nor time was mentioned to the subject as part of the definition of effort.

A sample of the visual feedback used is shown in Fig. 3(b). The shaded area of the bar increased as the subject exerted more force, and the target window was depicted by the two lines on the feedback bar. The subject was asked to exert enough force so that the edge of the shaded box remained between the two lines for the specified time period. When the specified time period ended, the bar changed color to indicate this to the subject. For the experimental subjects, the position of the target force window on the visual feedback bar corresponded to the value of the target force, and the time duration and force level (1–5) for the trial were printed on the screen.

Experimental subjects completed 70 trials in which the force was 1.0, 1.5, 2.0, 2.5, or 3.0 N and the time period was 2, 4, 6, 8, or 10 s. After completing the first 70 trials, the experimental subjects were given an additional 70 trials. These trials were identical to the first except that the target forces were distorted by 0% (23 trials), 7% (15 trials), 15% (15 trials), or 22% (11 trials). Subjects were given the same visual feedback as in the undistorted trials, but they had to exert more force to reach the visual target window.

Control subjects experienced only 70 trials, and they experienced no distorted trials. Control subjects were given no information about the magnitude of the target force or the predetermined length of time to maintain the force. The feedback bar was scaled in such a way that the target force window was always depicted at the center of the bar on the visual display.

TABLE I
FORCE JND EXPERIMENT

Young Subjects JND	Bias	Elderly Subjects 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

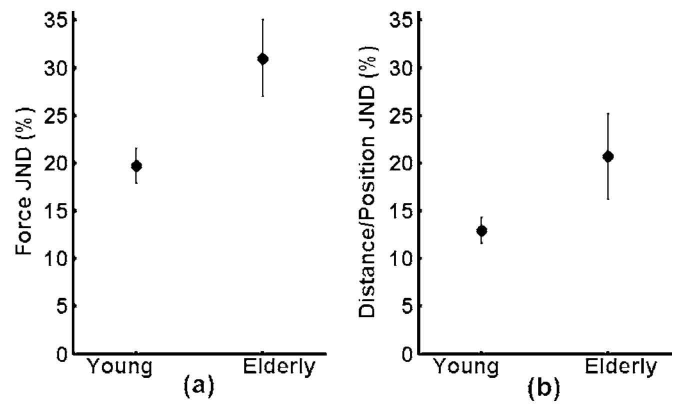

Fig. 4. (a) Results of the force JND experiment for young and elderly subjects. Mean and standard error for each group are shown. (b) Results of the distance/position JND experiment for young and elderly subjects.

IV. RESULTS

A. Force JND Experiment

Results for the young and elderly unimpaired subjects can be found in Table I and Fig. 4(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 I shows, the force JND varied widely with the individual. Individual JNDs ranged from 12.6% to 31.7% for young subjects and 12.3% to 50.5% for elderly subjects. Subject SKL's JND was 46.0% (0.920 N) with a bias of 1.14.

For elderly subjects, the percentage of correct responses on trials in which the two forces were experienced with the same

TABLE II
DISTANCE/POSITION JND EXPERIMENT

Young Subjects JND	Bias	Elderly Subjects 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

spring constant was 79.0% $\pm$ 4.40%. The percentage of correct responses for trials with the chosen combinations of different spring constants was 64.6% $\pm$ 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% $\pm$ 2.83%, while the percentage accuracy on different spring constant trials was 68.4% $\pm$ 2.26%. These values were not significantly different ($p = 0.474$), though 8 out of 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.

B. 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 II and Fig. 4(b). We refer to this JND as the distance/position JND because the distance moved and the terminal position of the finger are correlated in our 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 S, 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% $\pm$ 1.38% (3.99 $\pm$ 0.434 mm), and the mean bias was 1.10 $\pm$ 0.0601. The mean JND for elderly subjects was 20.7% $\pm$ 4.88% (6.32 $\pm$ 1.38 mm), and the mean bias was 1.01 $\pm$ 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. SKL's JND was 45.0% (14.8 mm), and her bias was 1.40.

The mean percentage accuracy for young subjects on the same spring constant trials was 85.0% $\pm$ 3.06%, and the mean percentage accuracy on the different spring constant trials was 72.3% $\pm$ 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% $\pm$ 3.02% and the mean percentage accuracy for the different spring constant trials was 73.3% $\pm$ 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.

C. Perceived Effort Experiment

Multiple regression analyses were done for the control subject data with force and time as the independent variables and mean effort rating as the dependent variable. Both independent variables caused statistically significant effects, with $p < 0.0001$ for force and $p < 0.05$ for time. The standardized coefficient was 0.92 for force and 0.18 for time, indicating that effort was predicted primarily by force; this can be seen in Fig. 5, which shows effort versus time for various forces. The multiple R^2 for the regression was 0.85. Similar results were found for the first 70 trials experienced by the experimental subjects. The standardized coefficient was 0.99 for force and 0.18 for time, and the multiple R^2 was 0.98.

Multiple regression analyses were then performed on the distorted trials of the experimental subject data, using effort rating as the dependent variable and *stated* force level, time, and distortion level as the independent variables. There were significant effects of all three variables ($p < 0.0001$ for stated force level and time and $p < 0.05$ for distortion). The stated force level was the most important contributor. The standardized coefficient was 0.99 for stated force level, 0.17 for time, and 0.08 for distortion. Fig. 6 shows effort versus distortion for different stated force

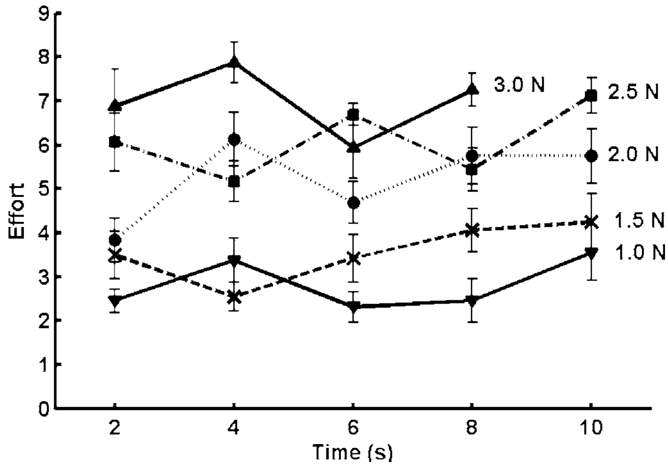

Fig. 5. Results of the control condition of the effort experiment. Effort as a function of time is shown for five different forces. Force was a better predictor of effort than time. Error bars represent the standard error.

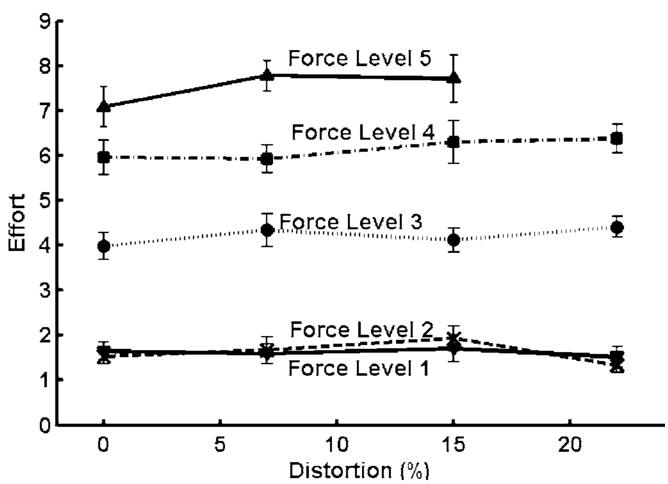

Fig. 6. Results of the experimental condition of the effort experiment. Effort versus distortion is shown for five stated force levels. Force level 1 corresponded to a nominal force of 1.0 N, force level 2 corresponded to 1.5 N, etc. When the level of distortion was greater than 0%, the subject had to exert more force to reach the stated force level; however, the visual feedback indicated no change. Stated force level had a greater effect on the effort rating than did distortion. Error bars represent the standard error.

levels and demonstrates the greater effect of stated force level. The multiple R^2 was 0.94. The actual force was higher for stated force level 4 (2.5 N) with 22% distortion than for stated force level 5 (3.0 N) with no distortion, but the mean effort rating for stated force level 4 with 22% distortion was 0.71 less than that for stated force level 5. This trend is interesting, though the difference was not statistically significant ($p = 0.219$).

V. Discussion and Conclusion

We have constructed an experimental environment consisting of a force-feedback robot and a visual display that is accurate enough to provide controllable feedback distortion during fine motor control tasks. This environment will allow us to measure small changes in the forces exerted by stoke and TBI patients, which could be important in tracking gradual recovery. Our system is safe because we use a robot designed to interact with humans, and our system tracks movements and exerts forces in three dimensions, giving us an advantage over systems that can only operate along a single line or plane [15]. Our system also adjusts automatically for movement differences between individuals and for an individual's changes in movement during the experiment. These advantages make our environment appropriate for testing our hypothesis that feedback distortion can improve the outcome of rehabilitation by influencing patients' perceptions of therapeutic exercise.

The JND for force measured for young subjects in our rehabilitation environment is more than twice as large as the JND measured for this dimension by other researchers. 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%. For instance, Brodie and Ross [16], [17] 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 [18] 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.* [19]. 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. As in our experiment, subjects in these experiments were given no visual feedback about the magnitude of the forces.

There are several reasons that our value for the force JND of young subjects is larger than that measured by other researchers. First, our environment and the joint tested are unique, which may affect the value of the force JND. In addition, our subjects received substantially less training in the discrimination task than those of [13] and [17]. Most importantly, all of the JNDs cited earlier were measured with all physical dimensions other than the target dimension fixed. These other physical dimensions are called background dimensions [20], and the most common way to measure kinesthetic JNDs is to keep the background dimensions constant. Our force JND experiment, however, has 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.* [21] 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.* [22] 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 that 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 vary the background dimensions in order to separate force and position cues. In general, we choose to vary the background dimensions in our experiments to increase the amount of visual distortion that is imperceptible.

The 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 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 impaired discriminatory mechanisms.

Researchers interested in other dimensions have also found increased JNDs in elderly subjects. For instance, Fitzgibbons and Gordon-Salant [24] measured the JNDs 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.* [25] 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.

As mentioned in Section III-B, 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. Since 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 S (percentages) and in terms of absolute position change (millimeters). Jaric *et al.* [26] 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. Since 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 based 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°.

No other researchers have measured a JND for distance or position in the same way that we have, but a few comparisons can still be made. Durlach *et al.* [27] and Ernst and Banks [28] measured the JND for the distance between the index finger and the thumb for young subjects. This judgment was based on the positions of the thumb and index finger and their results are thus comparable to ours. Ernst and Banks obtained a value of 6.0% (3.3 mm) for this finger span JND. Durlach *et al.* obtained a value of 1–2 mm and found that the JND did not obey Weber's law. At 3.99 mm, our distance/position JND agrees roughly with these measurements. It is slightly larger, probably because we vary the background dimensions of force and spring constant, as mentioned earlier.

Our 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 [29]. We believe that our force JND is 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.

Our reason for measuring the JNDs for force and distance/position was to discover the lower bound on the amount of distortion that will be undetectable by a subject, so that we can 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 [18] shows that the force JND does not decrease, even when subjects were exerting forces that were 85% of their maximum voluntary contraction.

While we cannot draw strong conclusions from a single patient, it seems, based on the JNDs measured for subject SKL, that we can reasonably expect the JNDs for stroke and TBI patients to be much larger than those of the appropriate control group. Since SKL is 34, her data are comparable to our young subject control group. Yet, her distance/position JND is more than twice as large as the largest young subject JND, and her force JND is almost 1.5 times the largest young subject JND. This 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.

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 [28].

The force and distance/position JNDs indicate the minimum amount of distortion that will be undetectable by a subject, but because vision is the dominant human sense in most situations [30], we expect that visual distortions well above the limit of the JND may be imperceptible. This is supported by the results of the perceived effort experiment. In general, people are capable of estimating their own metabolic expense with correlations between 0.70 and 0.80 [12], [31]; Williams and Purewal [32] confirmed this range for aerobic and anaerobic situations, finding a 77.5% correlation between user ratings on an "Effort Sense Rating Scale" and a measured metabolic variable. In the absence of visual information, the effort rating in our experiment was determined almost entirely by the force. However, when the subject was given visual feedback about the magnitude of the force, the effort rating was primarily determined by that information, even when the actual force produced by the subject was as much as 22% greater. This difference in force is larger than the JND, yet the subject's response depends on the distorted visual feedback. This shows that we may be able to use visual distortions beyond a subject's force or distance/position JND without the subject detecting the distortion.

Preliminary results for the experiments and robotic environment described here can be found in [33]–[36]. We have used this information about perceptual limitations and the effects of visual distortion to demonstrate that visual feedback can be used to increase the amount of force or movement distance produced by subjects [37]. We are currently planning work with patients using visual distortion in a rehabilitation environment to examine whether distortion can improve the outcome of rehabilitation by subtly encouraging patients to improve their performance in therapy.

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