IMPROVING FINGER FORCE CONTROL WITH VIBRATIONAL HAPTIC FEEDBACK FOR MULTIPLE SCLEROSIS

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ABSTRACT
Pathologies including stroke and multiple sclerosis can leave patients with impaired tactile and proprioceptive sensation, which contributes to their difficulty in performing everyday tasks. We present the results that indicate that a combination of fingertip force sensing and vibrational feedback can help patients with multiple sclerosis improve their performance in manipulation tasks. The feedback can take the form of an "event cue" in which patients are alerted when forces at the fingertips stray outside of a recommended range, or proportional feedback, in which trains of vibration pulses are correlated directly with the fingertip forces. While both types of feedback allow the patients to handle objects with more accurate force control, the former approach is more successful and preferred by subjects with mild impairment while the latter approach appears to be most effective for patients with severe impairment.

KEY WORDS
Multiple sclerosis, Vibration, Force sensing, Haptic feedback, Rehabilitation.

1 Introduction

Multiple sclerosis (MS) is an inflammatory disease affecting the human brain and spinal cord characterized by loss of myelin and axons in the nerve tracts. It is the most common cause of neurological disability affecting young adults in the United States and Northern and Central Europe. The symptoms and signs of the disease include motor and sensory dysfunction of the hand and arm. This dysfunction is often asymmetrically distributed between the left and right upper extremity. The clinical picture differs in severity from one patient to another, yielding individual combinations of reduced modes of sensation, reduced muscle power and increased muscle tone. Sensory disturbances of the upper extremities are usually related to lesions in the posterior columns of the cervical spinal cord (typically loss of proprioception) but they may also be due to cortical pathology of the brain. Disorder of the autonomic nervous system is relatively common but the peripheral nervous system is usually spared.

Realtime feedback has been established as an important consideration for people learning and performing new motor skills (e.g. as related to a task or job) [1, 2, 4]. In the case of MS patients who have reduced or distorted haptic sensation in one limb, there is a tendency to reduce the utilization of that limb and shift tasks to the opposite limb since many tasks are not easy to perform without sensation. The hypothesis behind the work in this paper is that patients' performance in manipulation tasks, and confidence in using the more impaired limb, could be increased by providing haptic feedback to the corresponding digits of the opposite limb so that utilization of the impaired limb will be increased.

Our approach draws upon several related investigations in which researchers have tried to improve subjects' performance in motor related tasks when the normal haptic information channel is either blocked or diminished, by providing sensory augmentation or sensory substitution through vibrotactile and/or force feedback. For example, Murray [10] designed a wearable vibrotactile glove for telemanipulation and evaluated the efficacy of different types of vibrotactile feedback to determine which ones helped subjects to achieve better control of force. Lieberman and Breazeal [9] developed a wearable vibrotactile feedback suit system that can detect errors in the motions of a subject's upper limbs and provide vibrotactile feedback to help them improve their performance. In the rehabilitation field, several investigations have addressed providing stroke patients with haptic feedback to the upper or lower limbs [5] [3]. However for multiple sclerosis, comparatively little has been done to explore the effectiveness of haptic feedback in response to measured manipulation forces.

2 Methods

2.1 Apparatus

The portable haptic rehabilitation apparatus consists of five parts: force sensors, signal conditioning circuits, microcontroller, amplification circuits and vibrotactile stimulators. (Figure. 1).

Initial experiments were conducted with commercial force sensitive resistors (FSRs InterLink Electronics Inc.). However, difficulties with nonlinearity, drift and hysteresis

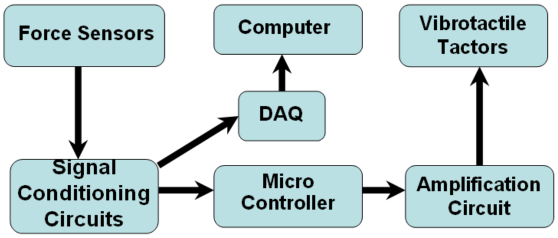

Figure 1. System hardware diagram

lead to their being abandoned in favor of a custom fabricated solution. The devices shown in Figure 2 utilize low-cost point contact force sensors (FSS1500NST, Honeywell Inc.) embedded in a cast urethane plate. To improve accuracy and make the fingertip force sensor less sensitive to the point of application of the contact force, the signals from three point-contact sensors are summed to obtain the resultant force on the plate.

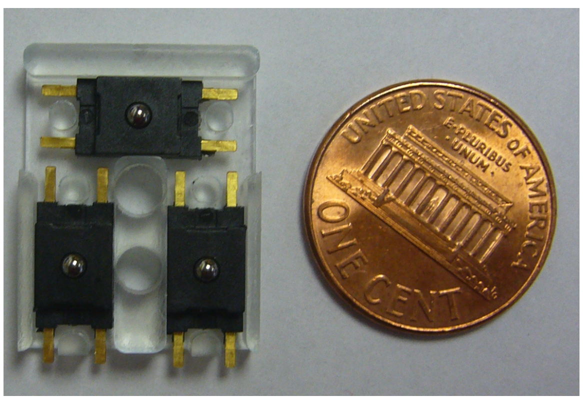

Figure 2. Sensor plate with one cent coin. Three sensors are held in the base plate, sensors are not wired in this figure to show a clear layout.

This sensor package including three sensors and base has dimensions of 22mm $\times$ 17mm $\times$ 4mm and can measure forces up to 44 N before saturating; however in this experiment forces never exceeded 15 N. The measured forces have a linearity within 0.7% and hysteresis of 0.5%.

Signal conditioning circuits are used to amplify and filter the force signal from each force sensor. A low pass filter is set at 40Hz, which is sufficient given the approximately 10Hz control frequency of human forces in manipulation. A microcontroller (PIC18F4431) monitors the forces with a 10bit A/D resolution for a resulting force accuracy of approximately 0.015N at each fingertip. The micro controller samples the data at 200Hz and sends corresponding drive signals to the vibrotactile stimulators. To obtain the data reported in this paper, a laptop computer with a 12 bit USB data acquisition board (National Instruments USB-6008) also monitored the force signals at 1000 Hz.

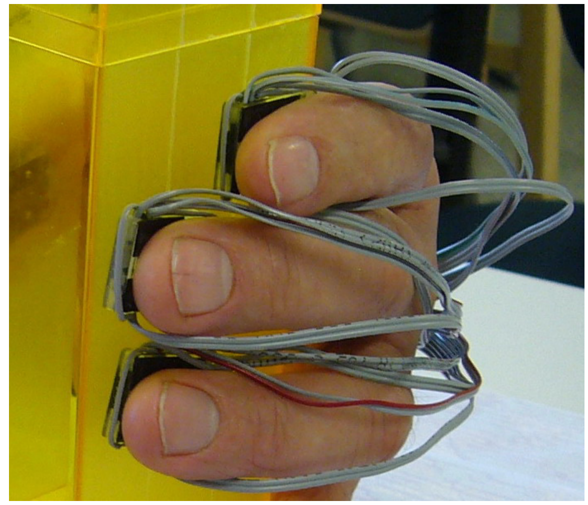

Figure 3. Fingers with sensors attached

Figure 4. Experiment setup: force sensors are attached on fingertips of subject's right hand. Vibrotactile tactors are attached on the fingernails of subject's left hand.

The stimulators are small cylindrical pager motors, 7 mm in diameter by 3mm high. The motors are powered by a 5V DC power supply(9V battery through a voltage regulator) through darlington transistors which are triggered by the parallel output channels from the microcontroller. The motors have a resonant frequency of approximately 200Hz when taped to a patient's hand, as shown in Figure 4.

2.2 Procedure

For patients who have sensation lost or reduced on one limb, there is an opportunity to provide haptic feedback to the healthier limb to improve performance in handling objects. For the experiments in this paper, force sensors were attached to the index, middle and ring fingers of a subject's impaired hand and vibrotactile stimulators were attached to the back of the fingernail on the corresponding fingers of the healthier hand (Figure. 4).

The task in the experiments is inspired by the everyday task of lifting a glass of water. The proxy for the glass is a hollow plastic parallelepiped, 5.7mm $\times$ 5.7mm $\times$ 15.5mm, weighing 73 grams. We asked subjects to grasp this object and raise it several centimeters from a table top, hold it for several seconds and then replace it. We instructed subjects to apply equal amounts of force on their index, middle and ring fingers in holding the object. When subjects reported that they felt the forces were balanced, we recorded the forces for 5 seconds and then asked them to replace the object. This task was chosen to test our hypothesis for several reasons. First, to quantify the effect of adding haptic feedback to improve MS patients' ability to control their finger forces in an everyday task, measuring the patients' ability to balance their finger forces is a logical and straightforward approach. Also, the balanced force task inherently leads the test subjects to distribute their forces evenly and minimally among their fingers. Because patients' hands are normally weak and quick to fatigue, the use of minimal sums of forces is desirable to decrease the risk of fatigue during the task. Finally, the task of balancing forces was simple and easy to understand, which allowed the elderly patients to complete the experiment successfully.

Three haptic feedback modes were selected: no haptic feedback (NHF), amplitude based feedback (ABF), and event-cue feedback (ECF). The characteristics of each feedback mode are discussed in the next subsection. A within subject test was conducted. 24 multiple sclerosis patients were recruited as subjects at the Masku Neurological Rehabilitation Center in Finland. Eight of those subjects are males, sixteen are females. The range of ages is from 33 to 64 with a mean of 56.4. The recruited subjects all have reduced sensation in one hand and good sensation in the other hand. They were all able to fully understand the human consent form and able to follow the simple instructions required to complete the sessions. With three feedback modes, there were a total of 6 different possible orderings in which the feedback modes could be presented to the test subject (NHF-ABF-ECF, NHF-ECF-ABF, etc...). For each test subject, one of these orderings was randomly chosen to present the different stimuli. For each feedback mode, all 24 subjects completed the task 3 times. The process was then repeated so that each feedback mode was presented a total of 3 times, which resulted in 9 trials total for each feedback mode and an overall total of 27 trials.

Before the tests, subjects were given time to get familiar with conducting the task under the three different feedback modes. The pretest practice sessions took from 30 to 60 minutes depending on the individual. Also, whenever subjects switched modes, they were given several practice trials with the new mode.

2.2.1 Force to vibration mapping

The ABF and ECF modes provide vibration feedback during the task in two quite different ways. The amplitude based feedback (ABF) is based on mapping the intensity of vibration feedback to the magnitudes of forces in the handling task. After several pilot tests involving variations of continuous amplitude and frequency, and pulses of varying frequency and duty cycle (i.e., the fraction of each period that a motor is turned on), the following scheme was adopted: For each finger, if the measured force exceeds a threshold of 0.05N, the microprocessor commands a vibratory stimulus. The vibration takes the form of a train of 5V pulses applied to the pager motor such that the pulse frequency and the duty cycle both increase with increasing force. At 0.05N, the frequency of pulses is 1Hz and the duty cycle fraction is (0.01/1), corresponding to 10ms during each period when the vibrator is on. For larger forces, the period decreases and the duty cycle fraction increases as shown in Figure 5, ultimately saturating at a force of 7N. (Note that in some cases, the pulses are short enough that the motor never reaches its steady-state speed.)

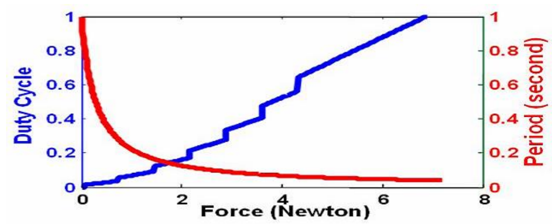

Figure 5. Period length and duty cycle versus force, in the amplitude based mapping method (ABF).

Event cue feedback (ECF) involves a quite different approach from amplitude based feedback. In this case, transient vibratory stimuli are applied only when there is a large range in the forces applied by the fingers:

$$if(max(f_i) - min(f_i) > f_r)\ then \text{ apply vibration}$$

where $i = 1, 2, 3$ for the index, middle and ring fingers, respectively and f_r is a threshold set empirically during the practice trials for each subject.

The vibration is applied to whichever finger deviates most from the average force, $f_a = (f_1 + f_2 + f_3)/3$, and is of Type I (high force) or Type II (low force) depending on whether the associated finger is above or below the average. The Type I stimulus consists of regular pulses at 35 Hz with a 50% duty cycle; the Type II stimulus consists of pulses at 1.33 Hz with a duty cycle of 10%. As with the amplitude based feedback, these parameters were determined empirically in pilot tests and found to be noticeable and easily distinguishable. The result of this approach is that when the fingertip forces are approximately equal, there is no stimulus. If one of the fingers drops substantially below the average value, a Type II (low force) vibratory cue is applied to that finger until it returns to within f_r of the mean. Conversely, if one of the fingers applies an excessive force, a Type I (high force) cue is applied until it returns to within f_r of the mean.

2.3 Data analysis and results

2.3.1 Metrics for the experiment

The primary metric chosen for the experiments was the degree to which subjects could achieve an even distribution of grasping forces among the fingers. The metric is computed as

$$f_s = \sum_{i=1}^{3} abs(f_i - f_a) \tag{1}$$

where f_a is the average force and, again, $i = 1, 2, 3$ for the index, middle and ring fingers, respectively.

Subjects were asked to maintain an even force balance for 5 seconds after lifting the object and the value of f_s was recorded continuously during this time. If the subject dropped the object or was unable to hold the object for 5 seconds, a failure event was noted.

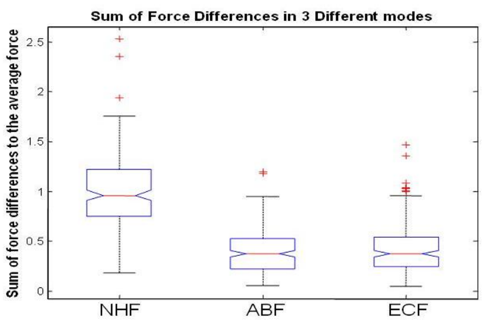

Figure 6. Box plot: Sum of force differences, f_s, under the different feedback modes (NHF: no haptic feedback, ABF: amplitude based feedback, ECF: event cue feedback). Both ABF and ECF result in significantly smaller variations in force compared to NHF; no significant difference is found between ABF and ECF. (Data for each subject are normalized by the subject's average value under no haptic feedback.)

To reduce the effects of subject to subject variability, all force data from each subject are normalized by the subject's average value with no haptic feedback (NHF). As seen in Figure 6, there is a significant difference in the force variations, f_s, when using ABF or ECF ($p < 1 \cdot 10^{-10}, p < 1 \cdot 10^{-10}$ respectively, using a Bonferroni corrected T test [6]). However no significant difference was found between the ABF and ECF modes ($p < 0.27$ in Bonferroni corrected T test). The standard deviations in f_s also show a significant reduction ($\sigma = 0.17$ for ABF and $\sigma = 0.16$ for ECF) as compared to the no-feedback case ($\sigma = 0.38$).

2.3.2 Correlating the effect of feedback with the degree of impairment

Although the overall data for 24 subjects do not show a significant difference between the ABF and ECF modes, we noticed during the experiments that subjects with greater impairment seemed to prefer the proportional feedback mode (ABF) while those with less impairment preferred event cue feedback (ECF). Accordingly, we divided the subjects' data into two groups based on their level of impairment, to look for a correlation between impairment and the improvement achieved using either ABF or ECF.

The subjects had all previously been evaluated using a standard clinical test. The 9-Hole Peg Test [7] is a quantitative measure of upper extremity function, widely used in MS clinical trials. It is one of the components of the Multiple Sclerosis Functional Composite that measures three important clinical dimensions of the disease, namely arm function, leg/walking function and cognition [12]. The test consists of moving nine pegs into one of nine holes on a peg board, then back into an open box. Subjects are scored in terms of the time required to complete the test, as compared to the normal range of times for subjects in the same age group. Published data indicate that results for unimpaired subjects are approximately normally distributed around the mean value for each age range.

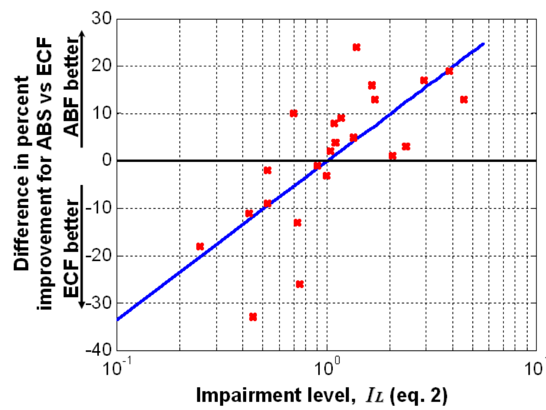

Figure 7. Difference in percent improvement for ABS vs ECF with respect to NHF mode. Each point corresponds to one subject. The x coordinate in log scale shows subjects' impairment level, I_L. A line shows the best log fit to the data.

For each subject, we computed an impairment level, I_L as follows:

$$I_L = \frac{T_S}{T_N} - 1 \tag{2}$$

where T_S is the time taken by the subject to complete the 9-Hole Peg Test and T_N is the average time for unimpaired individuals in the subject's age group. Thus, a value of $I_L = 0$ indicates no impairment and larger values indicate increasing impairment. For 23 of the 24 subjects, the value of I_L ranges from 0.25 to 4.53. (One subject was not able to finish the 9-Hole Peg Test test, so his value of I_L would be infinite.)

We then computed the difference in improvement obtained with ABF versus ECF over the baseline no-feedback (NHF) case. Figure 7 shows the results for the 23 subjects. The horizontal axis measures the impairment, I_L, plotted on a log scale. The vertical axis measures the difference in the percentage of improvement (in f_s) for the ABF versus the ECF feedback case. Subjects who showed the most improvement over the no-feedback case when using amplitude based feedback (ABF) are in the upper half of the plot and subjects who showed greater improvement with event-cue feedback (ECF) are in the lower half. A glance at the figure shows that the majority of less-impaired subjects ($I_L < 10^0$) are in the lower half of the plot while the more impaired subjects are mainly in the upper half. A best fit line is also plotted to the data. While the data show considerable scatter with respect to the line, the basic trend of greater improvement obtained with ABF *vs* ECF, with increasing impairment, is evident. Moreover, the relative improvement with ABF *vs* ECF appears to increase approximately logarithmically with the impairment level.

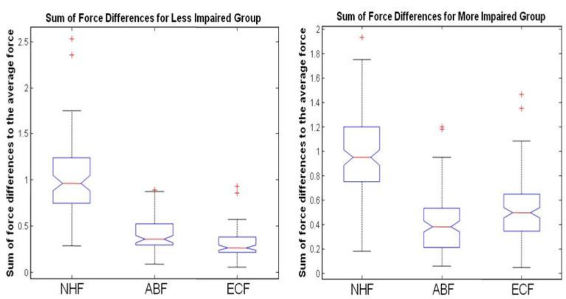

Figure 8. Force variation, f_s, for different feedback modes, with subjects divided into slightly and highly impaired groups. In both groups, the ABD and ECF feedback modes show improvement over the no-feedback (NHF) case. In the less-impaired group, the ECF mode was significantly better than ABF ($p < 1 \cdot 10^{-5}$); in the more impaired group the reverse was true($p < 1 \cdot 10^{-5}$).

The results in Figure 7 suggest that if we divide the subjects' data into two pools of less-impaired and more-impaired subjects, we should find significantly different levels of improvement with ABF versus ECF. Figure 8 shows the results of this division. We divided the subjects' data into two groups, depending on whether their I_L value was greater or less than 1. This division is somewhat arbitrary, as the subjects' I_L values are relatively evenly distributed on a logarithmic scale, as seen from Figure 7. However, the results are not much affected by whether the cutoff is at $I_L = 0.8$ or $I_L = 1.3$ instead of 1.

The results of dividing the subjects into two groups are shown in Figure 8. As in Figure 6, the measure of performance is the force imbalance, f_s from eq. 1. The less-impaired group contains 10 subjects and the more-impaired group contains the remaining 14. For the less-impaired group, the ECF mode provides significantly more improvement ($p < 1 \cdot 10^{-5}$ using a Bonferroni corrected T test) than the ABF mode. For the more-impaired group, ABF provides significantly better performance than ECF ($p < 1 \cdot 10^{-5}$).

2.3.3 Failure rate

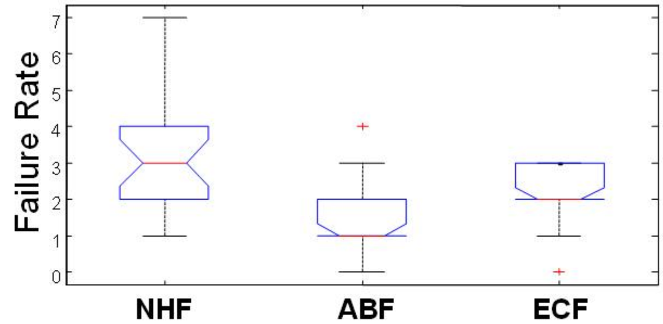

Figure 9. Failure rates for all 24 subjects under the three different feedback modes are statistically different. ABF provides the lowest failure rate.

The failure rate (i.e., the number of times that an object was dropped or could not be held for 5 seconds in each set of three trials with a given feedback mode) is another measure of the subjects' performance. As seen in Figure 9, the baseline NHF mode has the highest failure rate and the amplitude-based ABF mode has the lowest. A one-way ANOVA test was conducted and showed significantly different results ($p < 1.5 \cdot 10^{-5}$) among the modes. Bonferroni corrected T tests indicate that each pair of modes is statistically different: NHF *vs* ECF, $p < 8.08 \cdot 10^{-5}$; NHF *vs* ABF $p < 1.5 \cdot 10^{-9}$; ABF *vs* ECF, $p < 0.003$.

3 Discussion

The results in the preceding section indicate that either amplitude-based feedback (ABF) or event-cue feedback (ECF) in response to excessive variations in forces can help MS patients to better regulate their grasping forces in handling objects. On average, subjects' performance in balancing the forces among the index, middle and ring fingers was improved by more than 60% using haptic feedback.

Normalizing each subject's results obtained with haptic feedback by their no-feedback performance reduces the subject to subject variability sufficiently to reveal further significant trends. In particular, we found that subjects with mild impairment performed best with event-cue feedback while subjects with severe impairment performed better with proportional feedback. This finding also generally matches the preferences voiced by the subjects. A couple of reasons may explain this effect. First, the proportional feedback is always on and can become annoying when it is rarely needed by patients with mild impairment. In contrast, the event-cue feedback is only triggered when a subject is in danger of dropping or bobbling the object, perhaps due to fatigue or a momentary distraction. However, for subjects with severe impairment, the event-cue feedback is triggered frequently and can become distracting. These patients have relatively little sense of how much force their fingers are providing, and proportional haptic feedback provides a straightforward and continuous measure of their activity.

We observed also that the the event-cue feedback produced more failures than the proportional feedback (but fewer failures than no feedback). This result may be a reflection of the delay involved in detecting a dangerous situation (unbalanced forces), alerting the user and allowing the user to respond in time to prevent dropping the object. Providing an earlier warning of impending failures may reduce failure rate with ECF.

These findings are consistent with observations in other applications where a natural haptic feedback channel may be absent. For example, in experiments involving dexterous teleoperation of a slave robot [8] it was found that proportional feedback to the operators generally gave better performance than event-cues (e.g. grasp force too low or too high). The conclusion may be that when a natural feedback channel is absent, it is best to replace it with a proportional artificial one, but when an existing channel is active (perhaps with diminished effectiveness) it is better not to add a duplicate channel and instead to alert the user with cues of impending events.

The next step in this work is to determine whether lasting rehabilitation takes place as a result of utilizing the feedback. Our intention is to utilize the plasticity of brain to compensate for central nervous system damage that is a primary mechanism of functional recovery in MS [11]. Since the same phenomenon occurs in other important diseases of the central nervous system such as brain injury and stroke, the results may be of wide importance in neurological rehabilitation of hand and arm function. If rehabilitation can be shown, it would not be difficult to develop a miniaturized and more robust version of the apparatus.

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