Force Compensation and Recreation Accuracy in Humans

by

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ABSTRACT

As industry becomes increasingly reliant on robotic assistance and human-computer interfaces, the demand to understand the human sensorimotor system's characteristics intensifies. Although this field of research has been going on for over a century, new technologies push the limits of the human motor system and our knowledge of it. With new technologies come new abilities, and, in the area of medical care and rehabilitation, the need to expand our knowledge of the sensorimotor system comes from both the patient and physician.

Two studies relating to human force interaction are presented in this thesis. The first study focuses on humans' ability to bimanually recreate forces. That is, to feel a force on one hand and reproduce that force on the other. This skill is applicable in everyday lives from tasks such as a gardener using shears to trim a bush to a surgeon tying a delicate suture. These two tasks illustrate the different factors in this study on force recreation, which are the effects of: (1) occupational force dexterity, (2) force magnitude, and (3) the number of fingers used in the recreation task. Results showed statistical significance for force magnitude and number of fingers as factors in bimanual force recreation but not for occupation.

The second study examines how humans compensate for force perturbations in different directions with respect to the line of action and the effects of restricting movement time. A dynamic tracking task was presented to participants in which they were told to follow a moving target as accurately as possible. During a fixed interval along the target's path, a force field would perturb them in an undisclosed direction. Nine force conditions and three speeds were tested on both the left and right hands. Statistical analyses and comparison of error data indicate an effect of force direction on compensation accuracy. Speed is demonstrated as a statistically significant factor on accuracy, and a linear relationship between speed and error is posited.

CHAPTER 1: INTRODUCTION

One look at the cortical sensory and motor homunculi will demonstrate the evolutionary importance of humans' motor capabilities and our ability to communicate them [1]. Our hands and mouth dominate the representation in the motor and sensory cortices. The ability to design and create tools, devise complex motor tasks to create and use them, and even teach all of these steps is unrivaled. Woodworth [2] was one of the first to study the accuracy of human movement when he noticed ostensibly unskilled manual labor involved quite a lot of skill. He observed road workers hammering spikes into the ground with sledgehammers and counted the number of successful hits to quantify their accuracy. His argument was that, although hitting a spike does not require a lot of traditional knowledge, hitting a spike 999 times out of a thousand demands a high level of manual skill: a different kind of knowledge. Robotics researchers are intimately aware of this proposition as many everyday tasks require feedback from numerous inputs in our sensorimotor system and complicated, adaptive motor programs, which must be replaced by manufactured sensors and man-made algorithms and controls. Members of the medical committee are not naive to this complexity either. Working with patients who have lost motor function will quickly demonstrate that these skills cannot be taken for granted and occupy a different but vitally important area of knowledge and ability [3].

Despite the amazing flexibility of the human sensorimotor system, it is not free from inefficiency and bias. These deficits range from neurological asymmetries to physical sensory limitations to incomplete environmental practice. One such inefficiency, and a topic of this thesis, is force perception. Several studies have shown that self-generated forces are attenuated, which leaves a gap between what we *think* we are doing and what we are *actually* doing [4]. The medical community provides one of the most compelling motivations for working toward a solution to this problem. Physical therapists, for example, rehabilitate many patients through direct manipulation. This hands-on interaction involves the communication of muscle tone, force, and motion to the therapist and the selective delivery of force and motion by the therapist [5]. Misapplication of the desired forces may slow or reduce patient outcomes. These issues can be compounded in more complicated motor tasks. Similarly, force application errors during surgery account for a significant percentage of negative consequential patient outcomes such as recovery time and post-operative pain [6]. These errors likely result from insufficient training [7, 8].

Dynamic force perception and generation are primarily responsible for the competency of compensating for a perturbation in online movement (i.e., an unexpected force during a task). Previous research has shown that force compensation accuracy has many factors including brain hemisphere [9], arm position [10], and force type [11]. However, many of these studies necessarily focus on very simplified tasks and thus leave an open question as to how these findings combine in a more real-world task: a question partially addressed in this thesis. More studies simulating real world tasks will inform the development of motor training devices that seek to meet the requirements of new human-computer interfaces.

CHAPTER 2: BACKGROUND

Humans routinely preform motor tasks that require years of experience. The competency these tasks demand differ between individuals: a surgeon will have different skills than a machinist, for example. However, the underlying sensorimotor systems are the same, and researchers have sought to understand the mechanisms controlling them and the relationships between them for well over a century [2].

2.1 Hemispheric Asymmetries in Upper Limb Control

When researchers began investigating the roles of our hemispheric brains, they initially thought hemispheric differences were due to one side being more specialized for language than the other. However, Goldberg et al. [12, 13] were some of the first to recognize the hemispheric distinction was more universal than this. They contended, essentially, that the left hemisphere was optimized for processing the known while the right hemisphere was concerned with the unknown. Similarly, several decades later, researchers have disproved the popular myth that the dominant arm is "better" than the nondominant: it is more nuanced than that. Each arm has hemispheric specializations, which are responsible for different task components. The fact that the arms are different has sparked questions into what the differences are and how the arms work together in bimanual tasks, a question addressed in Chapter 4. The asymmetries between the control schemes of the human arms have been well documented in many diverse studies. Sainburg [9] introduced the dynamic dominance hypothesis, which stated that the dominant hemisphere is specialized in controlling limb and task dynamics while the nondominant hemisphere specializes in controlling stability and limb posture. Numerous other studies support theory that the dominant and nondominant arm performance asymmetries arise from hemispheric specializations [14–18].

The dominant arm's adept handling of limb dynamics manifests itself in many places. In Sainburg's proposal study, he found better coordination between dominant arm muscle torques (shoulder and elbow joints) compared to nondominant arm joints [9]. Roy and Elliot found that force variability does vary between hands, with the nondominant hand having greater variability, but is not affected by the visual condition [19, 20]. The dynamic dominance hypothesis has been upheld in many different scenarios including studies investigating unsupported reaching [21, 22], load compensation [23], left and right-handers [24], and patients with hemispheric brain damage [25].

The ability of subjects to make corrections based on changes in the inertial resistance experienced during movement was examined by Elliot et al. [26]. The authors used magnetic forces at the home position to increase the force needed to initiate movement (inertia). Once the participant's hand left the home position, the magnetic force was no longer acting on them and would experience an inertial change as a result. Subjects then had to make corrective adjustments as they moved toward the target location. The dominant hand yielded lower peak velocities between visual and no visual conditions as well as faster movement times compared to the nondominant hand. Furthermore, removal of visual feedback had a greater impact on the peak velocity of the left (nondominant) hand than the right: a result which is explored further in the next section. These observations suggest that the dominant hand was more adept at compensating and adjusting for the inertial change than the nondominant hand was.

2.1.1 Visual and Proprioceptive Asymmetries

Many aspects of motor control have been shown to be affected by hemispheric asymmetries and sensory feedback is no exception. Asymmetries have been demonstrated for both visual and proprioceptive feedback separately. Furthermore, research has shown that neither arm relies on visual or haptic sensory feedback to such a degree that it will become effectively useless if one of these streams is lost [27].

As one might expect, when visual feedback is present in a task, performance is better for both arms relative to their performance when it is absent. Apker et al. [28] conducted 3D reaching tasks where the visual condition of the fingertip was varied, and found that the behavior of the dominant and nondominant arms was comparable. However, when vision of the fingertip's position was removed, the variability of the dominant arm increased significantly more than the nondominant's did. This implies that the dominant arm has a proclivity toward visual feedback, but the contralateral hemisphere is also more efficient at online processing during motion. Goble et al. [29, 30] also investigated the reliance of the dominant and nondominant arms on visual and proprioceptive feedback. Their results agree with those produced by the previous study. Additionally, Goble et al. showed an advantage for the nondominant arm when visual feedback is removed and only proprioceptive information is available during the task. Moreover, it has been shown for the case of movement accuracy that visual information takes precedence over proprioceptive feedback [31].

2.2 Directional Arm Control

In this section, several areas are explored in which literature has shown directional differences in arm stiffness, force compensation, and force discrimination. Barbagli et al. [32] tested humans' ability to discriminate between force directions. They used several haptic/visual conditions, but the most precise and the most relevant, was a condition where they showed the subjects congruent haptic and visual information about force direction. They concluded that the perception threshold for force direction was 18.4° for this condition. Another study yielded similar angular precision of approximately 15° in their results [33]. These experiments were performed with a Phantom haptic device and a Microsoft Sidewinder force feedback joystick, respectively. However, a more recent study found a significantly reduced precision compared to that seen in the aforementioned studies. By applying forces to the fingertip, the authors determined that subjects could differentiate between 3D dynamic forces on average approximately 10° apart and above but this number could be as low as 7° depending on the way the force was applied [34]. For the purposes of this research, it suffices to say that humans can discriminate between force directions above a 20° difference.

From a mechanical perspective, a system's response to a force will be highly dependent on its stiffness and viscous properties in the direction of the force. Since the human arm is a complex, multi-joint system actuated by many different muscle groups, these mechanical properties vary by subject, posture, and task. Experiments have yielded stiffness ellipses of the arm with the major elliptical axis often oriented vertically. However, for forces parallel to the forearm, the major axis of the stiffness ellipse is also parallel. For example, this occurs in quadrants 2 and 4 in a two dimensional Cartesian coordinate system. For other force directions, of which the cardinal axes are of special note, the major axis is vertical (or close to it) [10, 35].

Other studies have demonstrated that the direction of reaching (perpendicular or parallel to the forearm) has an effect on the accuracy of planar reaching. Gordon et al. [36, 37] found a relationship between dynamic factors, such as velocity and acceleration, and movement direction with respect to the axial direction of the forearm. Movements in the high inertia direction had lower endpoint accuracy (due to subjects overshooting the target) than movements in lower inertia directions. These dynamic differences in movement amounted to an accuracy difference for movement direction favoring movements parallel with the forearm's axial direction. It is important to emphasize these observations are given with respect to the forearm and not with respect to desired movement direction as in this thesis.

Smyrnis et al. [38] observed the same directional bias. Results yielded a direct relationship between accuracy and inertial characteristics of the arm. However, Smyrnis et al. also observed that the mean directional error decreased over the trajectory timeline (i.e., directional error near the end of the trajectory was less than it was near the beginning). This intra-trial time-dependent variation is due in part to the initial bias at movement onset toward the axis of minimum inertia.

Mugge et al. [39] performed a self-timed goal-directed reaching task with assistive, resistive, and perpendicular force fields measured with respect to the target (i.e., an assistive force pulled the subject toward the target regardless of the subject's position around it). The purpose was to test directional effects of a force field on guiding someone to a target. For the visual target condition, statistically significant effects were seen for reaching time and path length but not for accuracy or precision of the path (although there was a trend with assistive having the best results and perpedicular the worst). Futhermore, the resistive condition had significantly longer paths than any of the other conditions. A second experiment, where subjects were given no visual feedback, produced statistically significant effects of guidance condition on accuracy, precision, reaching time, and path length.

Specific force types, such as multi-directional forces, are key ingredients missing from the existing literature discussed thus far. Kurtzer et al. [11] made a foray into this space, testing both force type and direction. However, their work focused mostly on whether humans learn to separate individual force fields in a multi-force environment. Their results indicated forces parallel to the line of action had less deviation than perpendicular forces did. This is interesting because the errors were calculated only with respect to the direction of the perturbation. This seems to suggest that there is an inherent difference in the way force disturbances are handled by humans depending on the direction of the force. Additionally, their results show less variability in the deviation for parallel compared to perpendicular forces. The authors used an Omni with 4.9N used as the constant force. This has very similar characteristics to the study presented in this thesis (see Chapter 4) although the task requirements vary somewhat: with this work asking subjects to compensate for a force rather than to adapt to one (more info in Chapter 4).

2.3 Reaction Time

Another factor in the visual feedback loop is the time to process a visual information and begin a corrective motion to achieve a desired motor outcome. This is termed reaction time (RT): also known as visual or corrective reaction time. Reaction time has been shown to vary by sex, age [40], individual [41, 42], and even generation [43]. Asymmetries in visual and proprioceptive reliance and processing efficiency also play a role in determining RT. These are just a few characteristics that affect reaction time. For an exhaustive list of factors affecting reaction times, see Kosinski's compilation [44].

Some of the first studies to estimate RT were done by Keele and Posner [45] and by Beggs and Howarth [46] who obtained mean RTs of 260 ms and 290 ms, respectively. Many other studies have gone on to quantify RT. An extensive review of existing data on RTs was done by Silverman [43] with strict conditions for inclusion in the summary data. This synopsis of a dozen different studies and thousands of combined participants places the cross-experimental average RT at 250 ms for men and 278 ms for women. A large range of reaction times was observed among both sexes with males varying from 183 to 324 ms and females exhibiting a slightly tighter grouping of 224 to 318 ms. In recent, thorough studies, Lin et al. [41, 42] have demonstrated similar RTs to the previously mentioned results despite corrections for procedural and analytical deficits in those studies.

Inefficiencies and biases in processing feedback (formerly discussed in Section 2.1.1) are not the only deficiencies in the sensorimotor loop pertaining to visual feedback. A second serious bottleneck is the psychological refractory period. The psychological refractory period is the time after a visual correction is made where new visual information is received, but cannot be encoded by the central nervous system [47]. This natural phenomena was taken into account for experimental design in this thesis (more discussion in Section 4.1.2).

2.4 Speed-Accuracy Tradeoff

Given the corrective lag in the sensorimotor system, the relationship between speed and accuracy becomes apparent. With a fixed corrective resolution, the maximum accuracy the sensorimotor system can achieve will suffer. Fitts and Peterson [48] were the first to quantify this relationship to Fitts' earlier model (commonly know as Fitts' law) describing the difficulty of a movement. It shows a logarithmic relationship between the target size and the distance to the target [49].

Fitts' law is given in Equation 2.1, where ID is the Index of Difficulty, W is the width of the target, and D is the distance to the target.

$$ID = log_2\left(\frac{2D}{W}\right) \tag{2.1}$$

Fitts' law has been used in a wide variety of applications such as describing foot movements [50], pupil dilation during a task [51], and information transfer rate in brain-computer interfaces [52]. In this thesis, it will be discussed in its traditional form.

Fitts and Peterson [48] went on to expand Fitts' original paper to include temporal information as a factor in the equation (see Eq. 2.2). In this case, MT is the movement time and is linearly related to the difficulty of the movement, where a, b are case dependent constants. A table of different formulations of Fitts' law was assembled by Plamondon and Alimi [53].

$$MT = a + b * log_2\left(\frac{2D}{W}\right) \tag{2.2}$$

Given humans' physical limitations described by Fitts' law and RTs, a relationship between the speed of a given motion and the accuracy of that motion is inevitable. In fact, we know and account for it innately [54]. Several studies have examined the functional relationship between speed and accuracy or the "speed-accuracy tradeoff". Howarth et al. [55] designed an experiment in which subjects moved from a position near their shoulder and to a surface 50 cm in front of them. Their task was to tap different targets on the surface. A more recent study had participants interact with a 2D visual environment through a joystick. Participants were asked to move to different target locations on screen. Results showed an effect on maximum speed and acceleration with horizontal movements having the higher values compared to vertical movements. Finally, they showed that different parameters affected the response at different periods of motion. The direction and distance of the target had significant effects in the initial stage of execution while the target size affected the last part of the response [56].

Another study aimed to test the limits of Fitts' law by using moving targets at various velocities and distances. Subjects were asked to move from a starting position at the center of a screen to target falling vertically on one side of the screen. Thus, unlike previous studies, the distance to the target, D, was not constant. An inverse relationship between speed and accuracy was demonstrated, and the authors found Fitts' law accurately described the results [57].

2.5 Force Perception

Thus far, sensory feedback review has been focused on the literature surrounding the visual system. The next sections will center more on haptic sensory feedback.

Our sense of touch, including force perception, is gained through two main systems: cutaneous and kinesthetic. In general, the cutaneous system is responsible for exteroceptive sensing while the kinesthetic system governs the proprioceptive elements [58]. The cutaneous system gathers information from sensors in the skin on surface textures, temperature, pressure on the skin, etc. Proprioceptive details such as bodily positions and information from sensors at muscles, joints, and tendons pertain to the kinesthetic system [59].

One way to describe the perception of force at various intensities is Weber's law, which posits a linear relationship between stimulus intensity and just noticeable difference to a subject. Weber's law is also commonly rearranged to find the constant relating the stimulus intensity to the perceptual difference. These realtionships are known as Weber's law and Weber's fraction and given in Equations 2.3 and 2.4, respectively, where ΔI is the just noticeable difference in intensity, I is the stimulus intensity, and K is a constant relating the two variables.

$$\Delta I = IK \tag{2.3}$$

$$\frac{\Delta I}{I} = K \tag{2.4}$$

Jones and Hunter [60] applied Weber's fraction to demonstrate an exponential relationship in human stiffness perception with a bimanual stiffness matching task where a subject received a reference stiffness on one hand and adjusted the stiffness of a motor felt on their other hand until perceived matching was achieved. Results showed an exponential decrease in the Weber fraction with increasing stimulus intensity. The same relationship was demonstrated for force in an earlier work [61]. Similarly, Weber's fraction was seen to decrease with increasing mass up to 200g when it began to level off [62].

The central nervous system (CNS) builds internal models in part from experiential information gained through the cutaneous and kinesthetic systems. This is also where the formerly discussed biases emerge. The CNS builds internal models of the body, motor tasks, and the external environment. These internal models are thought to formulate motor commands with input from sensing systems as well as behavioral goals. Therefore, the discrepancy in force perception would arise within these models when the difference between the predicted response and the actual response is not equal to zero; where the predicted response is generated by the internal models and the actual response is given by the exteroceptive and proprioceptive information [63]. Thus, force attenuation is a purposeful, albeit unconscious, reaction to external events. One explanation for the purpose of attenuating self-generated forces is to counteract the noise produced within the different levels of hierarchy in the human motor system. By decreasing proprioceptive forces, our force perception becomes more focused on exteroceptive forces [64].

These naturally occurring mismatches manifest themselves in interesting ways. An example of sources of error in a force perception task is given by the size-weight illusion. The size-weight illusion occurs when the size and weight of an object are negatively correlated. That is, when one is given two objects of the same weight but different size, the larger object will be perceived as the lighter of the two. This illusion originates from the size-weight relationship often occurring in nature which is to have a positive correlation between size and weight [65]. Another motivational example is attenuation of self generated forces. One of the consequences of force attenuation is overestimating a force during a force recreation task. This phenomenon has been documented by several experiments including Shergill et al. [4] who found an average increase in self-recreated forces of over 30 percent. Another study subsequently showed that this phenomenon held true for a bimanual force perception case in which participants received an input force on one hand and tried to simultaneously recreate that force on their other hand [66].

2.6 Force Perception Training and Experience

Internal models stored in the brain can be improved through deliberate training. Previous work has shown that an individual can be trained to become more accurate with force discrimination. Of course, the sensors themselves cannot be changed, but our perception of them can and has been. Flanagan et al. showed that these internal models could be augmented through experience to mitigate the natural tendency toward the size-weight illusion. Participants gained experience lifting objects with negatively correlated size-weight relationships of different sizes and shapes. The participants adapted through the training to accommodate the inverse relationship between size and weight of the experimental objects [67]. Valles and Reed showed that participants could be trained to overcome nature force attenuation in a bimanual recreation task. Participants received visual feedback of how much force they were applying with each hand during force recreation trials. An improvement in perception without feedback was found after a one day rest period, which suggests that internal models had been augmented through training [66].

Despite the studies showing the efficacy of targeted training in reversing some perceptual biases, the question of whether on-the-job occupational training can do this indirectly is still an open question. If not, occupations relying on high skilled bimanual motor skills may benefit from such training. This subject has been of particular interest to the field of laparoscopic surgery where force feedback and perception are paramount. Analysis of common surgical errors has demonstrated excessive force application is a significant problem [68]. In fact, a recent study showed that excessive force application accounted for 55% of all consequential errors during surgery [6].

Real world studies have shown that experience plays a role in specific force perception tasks. Experienced surgeons were seen to have a higher force perception threshold, but faster reaction times than novices [69, 70]. Similarly, experienced surgeons performed significantly better in a study on bimanual force recreation using laparoscopic instruments (LI). Interestingly, out of the two interaction modalities (finger and LI), subjects were more accurate when they used an LI than when they used their own fingers. Finally, results showed accuracy and force intensity were positively correlated (i.e., accuracy increased as the force magnitude became larger) [71, 72].

2.7 Force Production and Finger Configurations

Much of the work relating to finger configurations has been in force production tasks. Although this represents the opposite side of the coin to perception in some sense, it is very relevant to a force recreation task. Also, this work demonstrates the potential of finger configurations as a factor, which could affect force perception accuracy. The ability to control each finger's force production independently was shown for the gripping task in which participants were able to control the amount of force each finger produced independently. In this case, anticipatory adjustments to the amount of force on each finger were made in response to an imminent slip [73]. This further showed that fingers could be recruited for a specific goal such as a gripping task.

Researchers have examined the maximal voluntary contraction (MVC) associated with individual finger and multi-finger configurations. A force deficit was found when comparing the MVC of a given individual finger with the force output by the same finger in a multi-finger configuration. One study looked at the force production task extensively with relation to how finger configurations affected it. The study found that the aforementioned force deficit increased with an increasing number of fingers with respect to each finger. In terms of force production dominance, the index and middle fingers were the most dominant out of the four digits followed by the ring and little fingers, respectively. Different synergy levels were proposed as determining how much force each finger will contribute to total applied force. This whole process is controlled by the CNS, which ultimately delegates the force of each individual finger [74].

The previously mentioned literature examined force production and its relationship to finger configuration. Ambike et al. [75] investigated the affects of removing visual force feedback in a bimanual force production task. Participants were initially given continuous force feedback in relation to a desired reference force as well as the ratio of force production between hands. Subjects' goal was to maintain steady state at a given production ratio between the arms for a given reference force magnitude, which was based on a percentage of MVC. Both forms of feedback were removed after a set time interval. Despite being asked to maintain specific force magnitudes and production ratios, participants tended to reduce force magnitude and converge on a production ratio of 0.4-0.5 between the hands. They further found that the force drop, which after visual feedback was removed, was significantly greater for the nondominant hand than the dominant.

CHAPTER 3: BIMANUAL FORCE RECREATION

Humans perform many tasks throughout the day that require bimanual coordination and skill. Depending on a person's occupation, there are varying skill levels required to perform their jobs in an accurate, efficient manner. An increasingly relevant problem for people who require high bimanual skill is the inaccuracy of self-recreated forces, or bimanual force perception. The purpose of this study is to expand our understanding of human force perception by examining the effect of occupational bimanual skill levels, interaction methods, force levels, and confidence on one's ability to recreate a force. This is applicable in surgery and physical therapy where the application of force is an important aspect of the treatment. Note this is a force recreation task where recreation and reproduction are used interchangeably to describe the task subjects are performing.

Portions of this chapter have been previously published by IEEE in *Assessing the Effect of Experience on Bimanual Force Recreation* [76] and *Effect of Weight and Number of Fingers on Bimanual Force Recreation* [77]. Copyright 2016 IEEE.

3.1 Experiment

3.1.1 Experimental Setup

The experimental setup consisted of two Omega force sensors (LCM703-10) interfacing with a Phidget Wheatstone Bridge (Phidget 1046) to collect the force data from participants pressing down on two levers. The Phidgets interfaced with a RaspberryPi 2 computer. This was done to allow for a portable setup that could easily be taken to the offices of people working in different professions (i.e., engineers, physical therapists, and surgeons). The computer was attached to the back of the setup where the experimenter could see the screen but the participants could not (see Figure 3.1). A mechanical system of weights was used to provide a force on the input hand while the output hand tried to recreate the force on the opposing side. One force sensor was attached to each of the levers to determine the applied force.

3.1.2 Procedure

Participants were seated in front of the device and asked to make themselves comfortable. A padded arm rest was used to elevate the subject's forearms to the height of the finger pads. Two levers were presented to the participant. One lever was static and one lever was attached to known weights. The participant was asked to press down on the lever with weights to lift up the weights (unseen by the participant) and then to recreate the force with the other hand on the unmoving lever (see Figure 3.2). The procedure was demonstrated to the participants prior to the experiment. Questions regarding the experimental procedure were answered, and the system was demonstrated to ensure each participant's full comprehension of their task during the experiment.

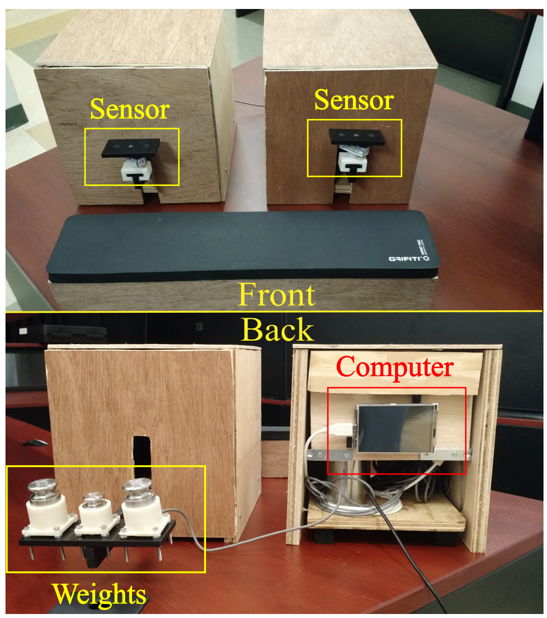

Figure 3.1: The setup for the bimanual force recreation experiment is shown with the input hand shown on the right. The front of the system, including the arm rest, is at the top of the figure. The back of the system is below with the Raspberry Pi and onboard display for the administrator. These forces were generated using a purely mechanical system with a lever and pre-measured weights. The weights were verified using a scale accurate within 0.01N.

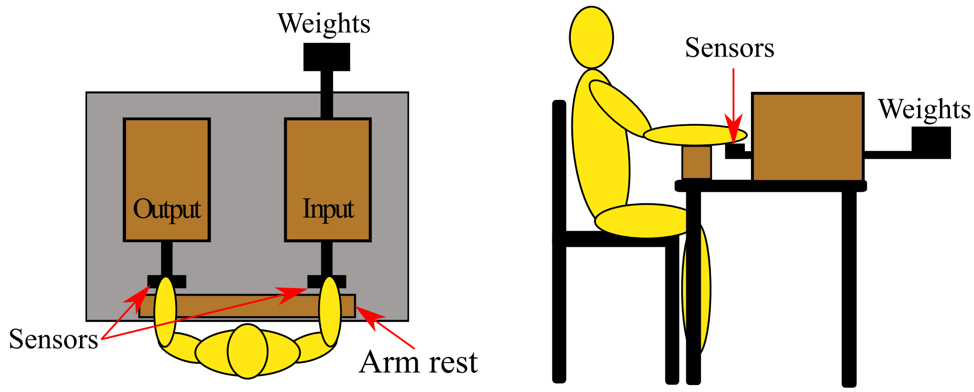

Figure 3.2: A schematic of the recreation device showing the force sensors, weights, and experimental setup.

At the beginning of each trial, participants were told to use either one finger (index finger) or three fingers (index, middle, and ring fingers) prior to each trial. There were 40 trials lasting five seconds each. Participants were instructed to indicate when they were applying the same force on both hands. Therefore, the length of the whole experiment depended on the individual participant, but usually lasted approximately 12-15 minutes. The forces tested during the experiment followed a linear distribution ranging from 4 N to 13.7 N. Each load was applied four times in a random order during the experiment.

3.1.3 Participants

The participants came from three occupational fields: engineering, surgery, and physical therapy. Demographics on occupation, time active in their field, and handedness were collected. Ten participants participated in the study, which were divided into two equal groups of high skilled (i.e., physical therapists and surgeons) and average skilled (i.e., engineers) occupations. Note that skill here refers to a generalized level of skill for each subject's occupation and does not represent individual abilities necessarily. Physical therapists and surgeons comprise medical disciplines which heavily rely on manual skill and were thus considered to belong to high manual skill occupations. Engineers, generally, do not require a higher than average level of manual skill for their jobs. They were therefore used as an average skill group representing the general public.

Of these participants, there were five engineers, two surgeons, and three physical therapists. All the participants were male. Five of which identified as right hand dominant: five left. This study was approved by the University of South Florida Institutional Review Board and all participants signed an approved consent form.

3.2 Results and Discussion

A three-way repeated measures ANOVA was performed with relative error between the hands as the dependent variable and independent variables were the number of fingers (one or three), weight applied (4, 6.4, 8.8, 11.3, 13.7)N, and learning effects (over four repetitions). As expected, learning was not statistically significant. Number of fingers was statistically significant ($F(1,9) = 14.89$, $p = .004$) with force recreation error means of 26% and 47% for one and three fingers, respectively. The amount of weight was statistically significant ($F(4,36) = 9.02$, $p < .01$).

A one-way ANOVA was performed with relative error between the hands as the dependent variable and the independent variable as experience level (average or high). The difference between skill groups was not statistically significant ($F(1,4.34) = 1.56$, $p = .21$).

3.2.1 Effect of Weight on Bimanual Force Recreation

Results reaffirmed the findings of previous research showing that humans have a tendency to overestimate self generated forces. However, the tendency shown in another study for subjects simultaneously producing self-generated forces on both hands to move toward an equal production ratio was not seen in this recreation task [75]. Also, force attenuation, as the sole explanation of the error, has been challenged by Onneweer et al. [78]. Their argument is that errors arising from feedback deficits should be eliminated if a force is generated and reproduced by the same hand. Since the same hand was used to create and reproduce a force in their study, they posit that force attenuation alone cannot describe the differences seen in their results.

For finger configuration, using one finger was more accurate than three fingers with 45% less error. Both of these results, in combination with a participant survey (see Section 3.2.4

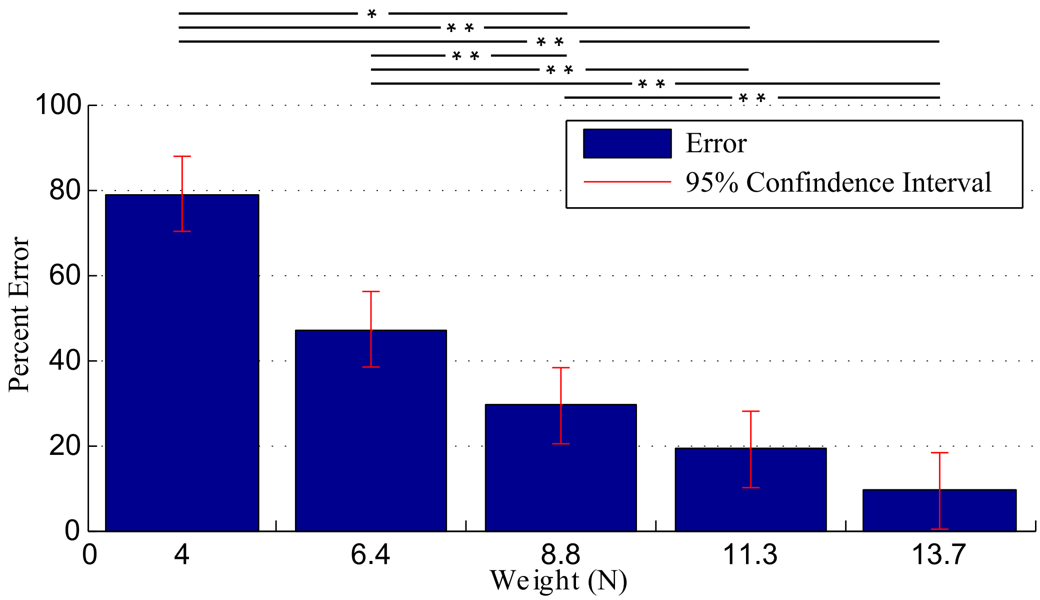

Figure 3.3: The post-hoc test with Bonferroni corrections is shown for accuracy vs. weight. Note: the bars at the top indicate statistical significance between pairs where * means $p < .05$ and ** means $p < .01$. Copyright 2016 IEEE [77].

for more details), confirmed the disconnect between intended and actual actions. Interestingly, although the weights followed a linear distribution, the weight-accuracy relationship seen in Figure 3.3 follows what appears to be an exponential function with weight inversely affecting error. Another study saw the same tread between weight and accuracy [71]. A correlation between force magnitude and reproduction error has been shown. Indeed the same trend of decreasing error with increasing force (up to 100N in their experiment) was seen [78].

One explanation for this relationship might be Weber's law, which shows an exponential relationship for haptic stimuli such as stiffness [60], force [61], and mass [62]. These relationships decrease exponentially in all cases. Although the results in Figure 3.3 are not directly comparable to the Weber fraction for force, they do represent subjects' force perception as a function of increasing intensity. Therefore, the results presented here may be analogous to the relationship of just noticeable difference to stimulus intensity: the definition of Weber's law.

3.2.2 Effect of Number of Fingers on Bimanual Force Recreation

A statistically significant difference was found between using a single finger and using three fingers during the bimanual force perception. The participants' force perception error with one finger was approximately half that of the three finger configuration with mean force recreation errors of 26% and 47% for one and three fingers, respectively. The force applied by each hand differed by only about 10% for trials using a single finger configuration; however, this increased significantly for the three finger configuration, which had a difference of almost 50%.

It is unclear why one finger was more accurate than three fingers. One explanation might come from subjects thinking that using one finger was less accurate (see Section 3.2.4 for more details), thus increasing their focus and effort. This might also be why Prasad et al. [71] observed a performance increase when subjects used a laparoscopic instrument instead of their own finger. Another explanation could be a compound error arising from a multi-finger combination. Finally, an experiment to further explore these potential explanations is outlined in the Future Works section.

3.2.3 Effect of Experience on Bimanual Force Recreation

Although my initial hypothesis and the existing research pointed to experience playing a role in force recreation, the results did not confirm this. However, the high skilled subjects were much more consistent as a group compared to the average skilled group with standard deviations in percent error between subjects within their groups of 13% and 68%, respectively. Both the most and least accurate subjects belong to the average skilled group.

Figure 3.4 shows the participants grouped together by motor skill level with engineers on top in the average category and surgeons and physical therapists in the high category. The post-hoc test revealed the most and least accurate participants were engineers. The high accuracy of two of the five engineers may mean that the assumption that engineers would be a good control group may be false; although this is unlikely. Participants' results came as somewhat of a surprise as it had been previously shown that direct training could make an individual more accurate at a force perception task [66]. In any case, those that were trained with delicate forces and precision of motion (i.e., surgeons) and applying whole body forces and motions (i.e., physical therapists) do not appear to be any better at recreating an applied force than those who have not been formally trained.

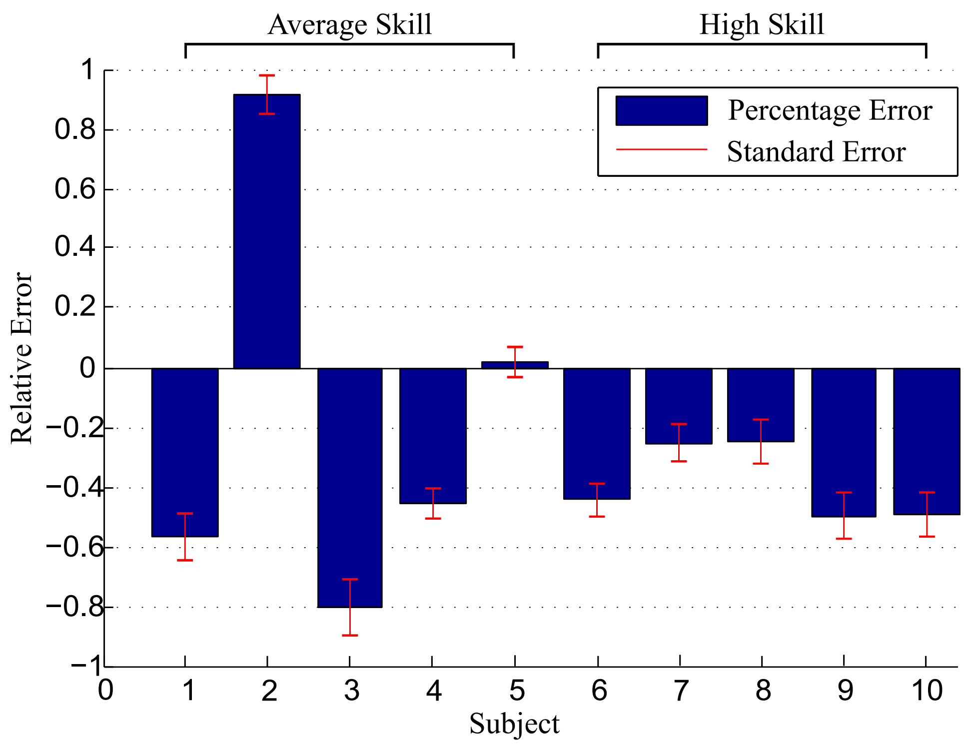

Figure 3.4: The average and standard errors are shown for relative error vs. subject. Relative error was calculated as the difference of the force applied by the input hand minus the recreation (output) hand all relative to the input (reference) force. Subjects are shown grouped according to occupational experience (see Participants section above for skill definitions). Copyright 2016 IEEE [76].

As in the case of the size-weight illusion (discussed in Section 2.5), the highly skilled participants may not have experienced situations that caused them to confront their natural tendency to attenuate self-produced forces. Individuals have been trained to overcome this natural discrepancy by contrasting their perception and reality [66]. Furthermore, several studies have shown experienced surgeons are more accurate at force perception and make fewer in task errors than residents do [71, 72]. Results presented in this thesis do not agree with previous studies likely due to the small sample size and the mix of PTs and surgeons in the high skill group.

3.2.4 Confidence

Occupational skill level was not a factor in judging which configuration was most accurate; participants from both high and average motor skill groups were equally likely to answer incorrectly. At the end of the experiment, participants were asked how accurate they thought they were on average during the experiment on a scale of 1 to 5. Most of the engineers thought they had performed at a 3; only one responded more confidently at a 4. However, the higher skilled participants responded with more confidence and had an average of confidence of 4. This further indicates that these medical professionals were not only unable to accuracy recreate forces, but they also thought themselves more accurate than they actually were. This poses an interesting question about whether surgeons and physical therapists are actually applying the forces they think they are.

When asked which finger configuration, if any, they thought made them more accurate, 60% of participants said they were more accurate using three fingers than one. Here again, participants were unable to correctly judge the most accurate interaction method available to them since they were actually much more accurate with one finger compared to three.

CHAPTER 4: FORCE COMPENSATION IN A TRACKING TASK

As discussed at the beginning of this thesis, the motivation for exploration of the human sensorimotor system is due in large part to the medical community. This second study seeks to determine whether asymmetries in arm stiffness and movement accuracy continue to a dynamic force compensation task. Such knowledge is useful when designing human-robot interfaces for delicate tasks where forces need to be simulated and perceived with a high degree of accuracy.

4.1 Experiment

4.1.1 Experimental Setup

Subjects used a Sensable Omni to track the target as it moved across the screen (see Figure 4.1). The Omni displayed the perturbation forces while a C++ program, which ran at 1000Hz, recorded the position of the on-screen cursor and other experimental data.

The experiments were performed in a well lit room. Subjects' vision was not altered, so they could see their arms, the Omni, the screen, etc. during the experiments. The screen used to display the tracking task was a standard 23" desktop monitor with a refresh rate of 60 Hz. The colors seen in Figure 4.2 were chosen based on the contrast of all the elements to each other. This ensured participants could easily see any element at any position. During the experiments, subjects had their arms supported by a pillow to ensure a consistent arm position between participants and trials. The support kept their arms approximately level with the plane they were moving in.

4.1.2 Procedure

Participants were presented with a tracking task. They were told they would be perturbed in different directions within the 2D plane they moved in, and they were to "try to stay as close to

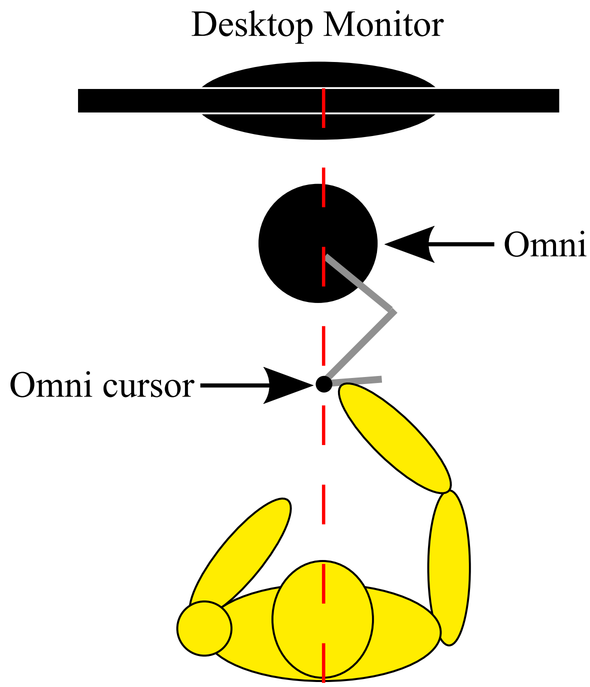

Figure 4.1: An overhead view of the setup showing a subject interacting with the visual-haptic environment. Also note the physical Omni cursor.

the center of the target as possible." No information about the force direction or speed of the target was provided to the subjects. The screen (seen in Figure 4.2) showed participants the moving target as well as the Omni position cursor during trials with full visual feedback. A tracking task was chosen for this experiment so that errors in two dimensions could be calculated (see Figure 4.3). During all trials the target moved in a horizontal line from left to right as depicted in Figure 4.2.

Participants experienced three different target velocities and nine different perturbations on each arm during the experiment (listed below). Each velocity condition was constant for a given set of perturbation conditions. For example, if 60 mm/s was set as the target velocity, all of the perturbations would be looped through before the target velocity was changed. They were given a one minute mandatory break on each arm after each set of perturbation conditions (9 trials). This was done in order to reduce fatigue effects and to ensure consistency between subjects.

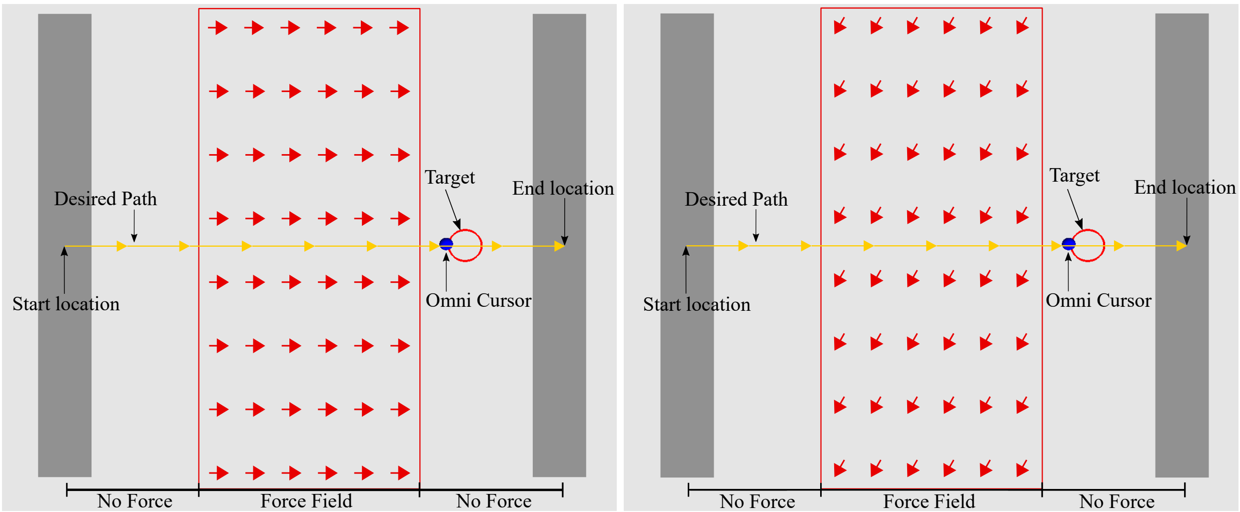

Figure 4.2: Two example screens of the experiment are shown. Representations of the force fields are shown for convenience, but were invisible to the participants. On the left side, a "right" force field is depicted while a "Quadrant 3" force field is on the right side. Salient portions of the screen are annotated: also not visible to participants.

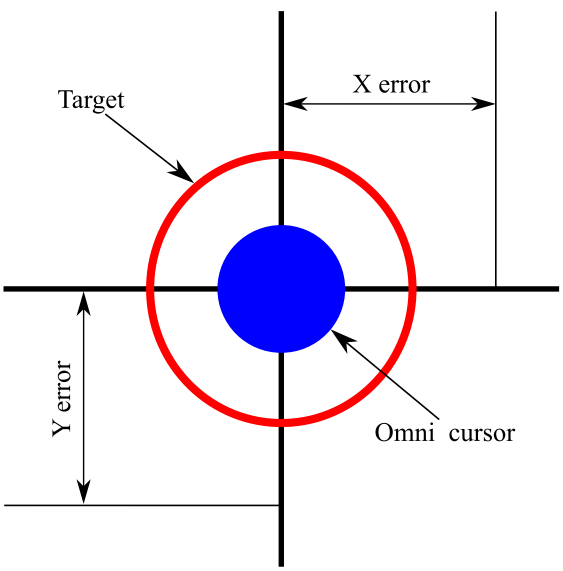

Figure 4.3: A depiction of how X and Y errors were defined. The Omni cursor shown at the zero error position. Note: only the Omni cursor (blue) and the target (red) were visible during the trial.

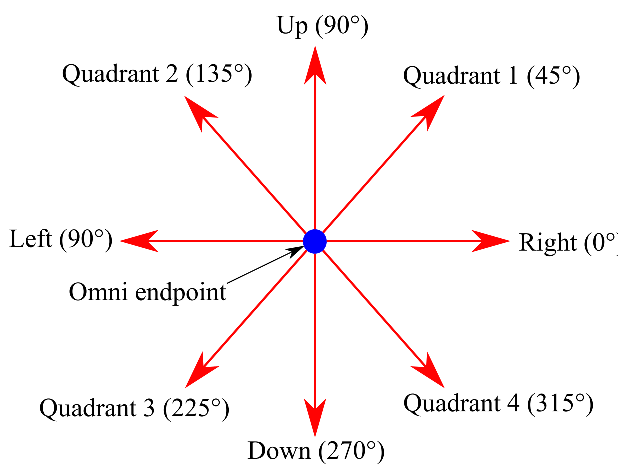

Figure 4.4: All force perturbation conditions are shown in a 2D plane acting on the Omni endpoint. There were nine total force conditions with the one condition not shown being the no force, baseline condition. The naming conventions seen in this figure will be used consistently throughout this thesis.

The following is the full structure of the experiment:

- Training trials (22 trials total)
  - 10 training trials at the beginning of each hand
  - 2 training trials at the beginning of each new velocity (3 times on each hand)
- Condition trials (54 trials total)
  - 3 target velocity conditions
    - 40 mm/s (Slow)
    - 60 mm/s (Medium)
    - 80 mm/s (Fast)
  - 9 perturbation conditions (see also Figure 4.4)
    - No force
    - Up
    - Down
    - Left
    - Right
    - Quadrant 1 (Q1)
    - Quadrant 2 (Q2)
    - Quadrant 3 (Q3)
    - Quadrant 4 (Q4)

4.1.3 Participants

Subjects who participated in the study were all healthy individuals who possessed no known physical or neurological disabilities that would corrupt their results. All six participants were right handed; half were male. This handedness restriction was implemented due to the variance found in several studies between left- and right-handers [24, 79–81]. Their ages ranged from 22 to 47 years old. All participants signed a consent form for this study approved by the University of South Florida's Institutional Review Board.

4.2 Results and Discussion

Statistical analysis was performed in SPSS with a multivariate, 3-way ANOVA on X, Y, and root mean square (RMS) error (see Table 4.1) and a 3-way ANOVA on peak acceleration. Post hoc tests using a Bonferroni correction were performed on statistically significant results for comparison. Independent factors were hand, speed, and force condition, and dependent variables were parallel (X), perpendicular (Y), RMS error (see Eq. 4.1), and peak acceleration. Error here means the difference between the position of the haptic interface point on screen and the center of the moving target (see Figure 4.3).

$$RMS = \sqrt{\frac{\sum_{i=1}^{n} E_i^2}{n}} \tag{4.1}$$

All position data was filtered using a first order Butterworth filter with a 20Hz cutoff frequency. This frequency was chosen because it is above the maximum frequency for human motion and below Omni sensor and motor operating frequencies. Time data was filtered using the average of the nearest five data points. This ensured an appropriate linear increase without time gaps from errors in encoding. Additionally, all data was interpolated to 7500 data points to account for different speeds (trial lengths). This number of data points was just above the number contained in the longest trial, which meant no data was lost in the interpolation process.

As shown in Table 4.1, target speed and force direction were statistically significant factors but hand was not. Post hoc tests are discussed in their respective sections below, and further analyses are also presented and discussed. A note on all plots depicting experiment paths, errors, etc. is the cause of the initial peak at approximately 50 mm (or 1000 data points) into the trial. This was due to the subjects catching up to the target after a small reaction delay when the target

Table 4.1: Statistical results obtained from a multivariate, 3-way ANOVA for hand, speed, and force condition factors.

Hand	RMS	F(1,5) = 0.04	p = 0.948
	Y	F(1,5) = 0.30	p = 0.585
	X	F(1,5) = 0.05	p = 0.819
Speed	RMS	F(2,10) = 39.43	p < 0.0001
	Y	F(2,10) = 24.36	p < 0.0001
	X	F(2,10) = 34.85	p < 0.0001
Force Condition	RMS	F(8,40) = 15.31	p < 0.0001
	Y	F(8,40) = 18.78	p < 0.0001
	X	F(8,40) = 21.79	p < 0.0001

Table 4.2: Average and standard deviation for groups of force perturbations reported for each component of error.

		RMS	**Y**	**X**
Average	All	5.19	2.58	6.30
	Parallel	6.77	2.10	8.97
	Perpendicular	4.42	3.40	4.378
	Diagonals	5.27	2.71	6.51
Standard Deviation	All	1.10	0.62	1.77
	Parallel	0.37	0.01	0.55
	Perpendicular	0.19	0.14	0.22
	Diagonals	0.23	0.20	0.34

began moving. As seen in the plots, enough time was given to allow subjects to reach steady state before they were perturbed.

4.2.1 Hemispheric Asymmetries

Hemispheric asymmetries, as discussed in Section 2.1, did not significantly affect the performance of either hand. Given that subjects could take advantage of full visual feedback during compensation, this result was expected. If the visual feedback condition was varied, it is likely that the dominant arm would suffer disproportionately [28, 29]. Removing visual feedback might also reveal some other differences as to whether there are hemispheric advantages based on force direction with respect to the line of action. This is because visual feedback, when available, takes precedence over proprioceptive feedback [31].

4.2.2 Force Direction

For force directions, post hoc tests unsurprisingly indicated the RMS error for the baseline force condition was statistically significantly different than all other force directions with the exception of a downward force. Furthermore, none of the four diagonal directions were shown to be statistically significantly different from each other in any of the error cases (X, Y, or RMS). However, a statistically significant difference between perpendicular and parallel groups, but not within groups, was found in each error case (see Figures 4.5, 4.6, & 4.7).

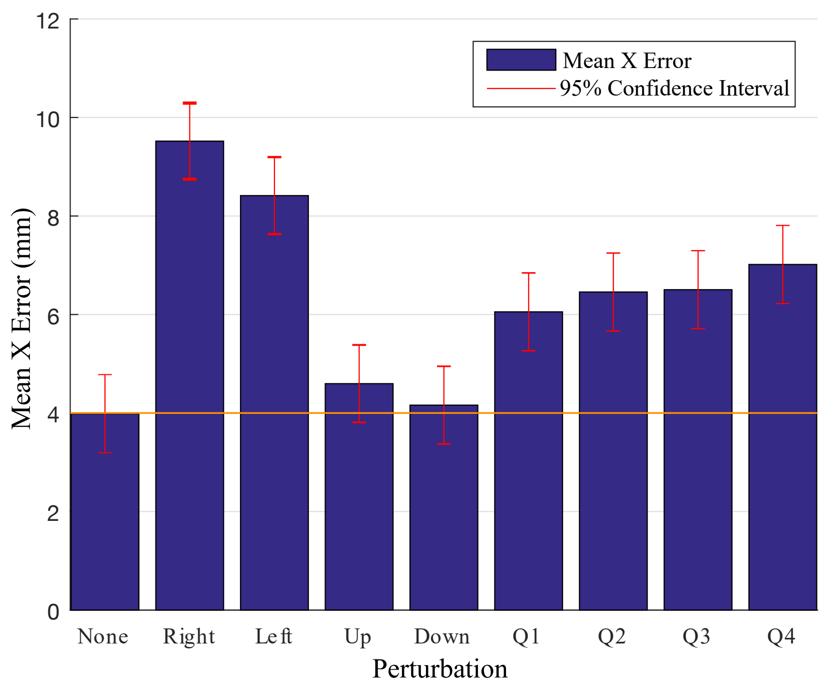

Figure 4.5: The average and standard errors are shown for the mean X axis error versus force condition.

Significance is also seen when contrasting error magnitudes in the two axial directions. Errors in the X direction are higher than in the Y even after subtracting baseline error in both cases. Comparing Figures 4.5 & 4.6, the differences continue when contrasting off axis errors (OAEs) where off axis error means the error in the direction normal to the direction of the pertur-

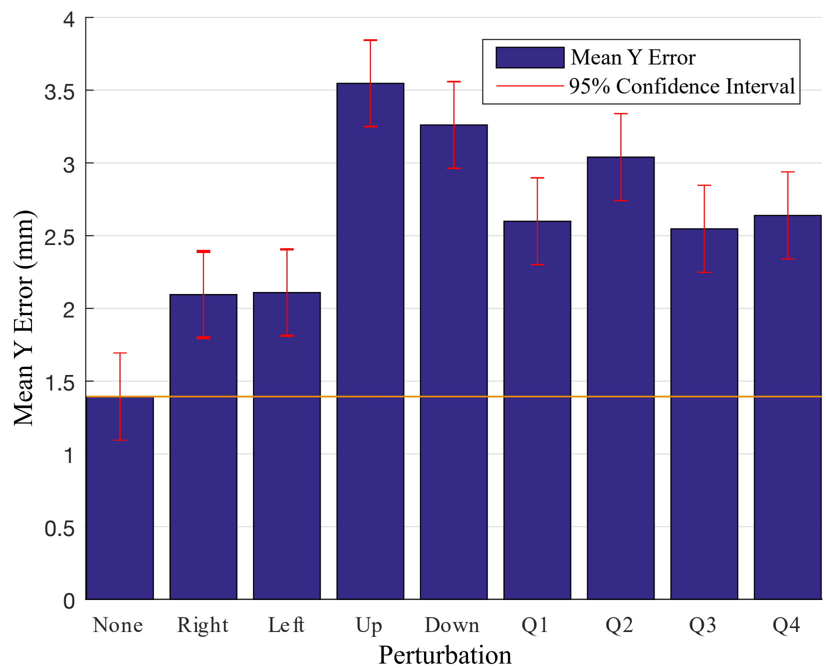

Figure 4.6: The average and standard errors are shown for the mean Y axis error versus force condition.

bation. OAEs for the right and left (parallel) perturbations are significantly higher than baseline while OAEs for up and down (perpendicular) perturbations are almost the same as the baseline component. Therefore, subjects actively or passively create a significant off axis error with respect to the baseline error only for perturbations that are along their line of action (LOA). Based on these observations, subjects are able to minimize OAE for perpendicular but not parallel forces.

In Figure 4.8, X error has a remarkably higher standard deviation than Y error does. Averages and standard deviations of parallel and perpendicular groups in Table 4.2 support the same conclusion. The average and standard deviation of the RMS error are significantly higher for perturbations parallel to the LOA than for perpendicular ones. This may link back to humans' reaction time and the refractory period associated with a corrective reaction [42, 44]. The desired position at the center of the target can be thought of as an intersection point of two lines (see Figure 4.3). In this scenario, the position of the horizontal line, which corresponds to zero Y error, never changes. This is not true for the vertical line corresponding to X error, which moves continuously with the speed of the target. Given the lag inherent in the human sensorimotor system,

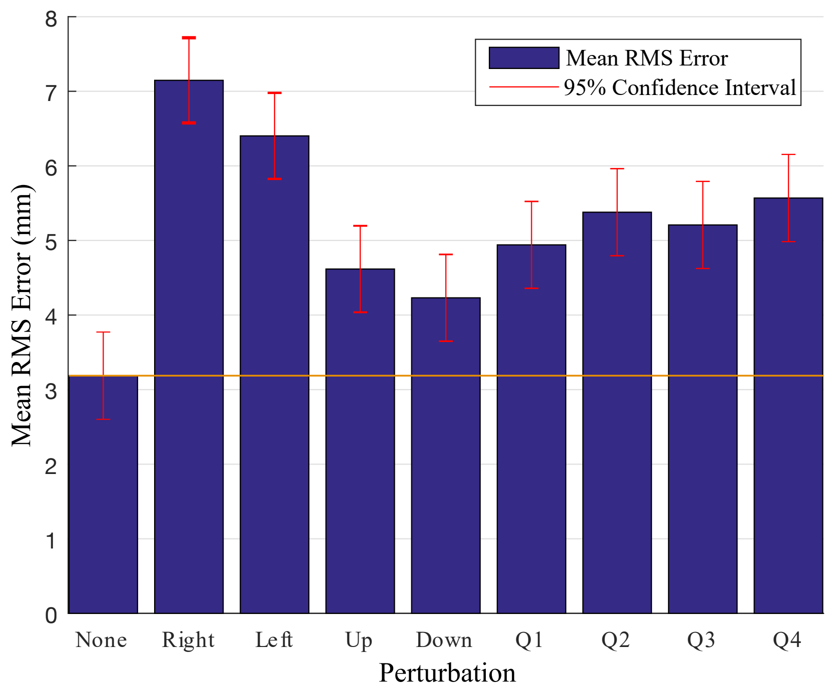

Figure 4.7: The average and standard errors are shown for the mean RMS error versus force condition.

converging on a continuously shifting reference point would be meaningfully more challenging. Comparing mean X and Y errors for baseline, this is clearly demonstrated. Likewise, the magnitude of X errors is much higher than those for Y on average for all other force conditions. Subtracting baseline error does not have an effect on the statistical results or on the salient portions of the estimate error plots. This is important because it shows that the higher magnitude of X error is independent of the tracking task to some extent.

Biomechanics also plays a role in determining directional dependence as previously discussed in Section 2.2. It has been well established that stiffness [10, 35] and force generation [82] vary with respect to the forearm's axis. However, the effect of task specifics has also been well studied and has demonstrated that factors, such as velocity and acceleration profiles [36], can be changed by a subject based on task requirements. Therefore, the question of whether a subject's ability to compensate for a perturbation with respect to the line of action does not directly follow from known biomechanical knowledge.

The results seen in planar reaching [36, 37] continue here when a subject is perturbed during a tracking task; with the highest accuracy corresponding to compensations parallel to the forearm (Note: this is approximately perpendicular to the LOA). One could conclude that either the differences seen are due to the orientation with respect to the forearm or with respect to the LOA. However, the two conclusions, while unique, are not contradictory.

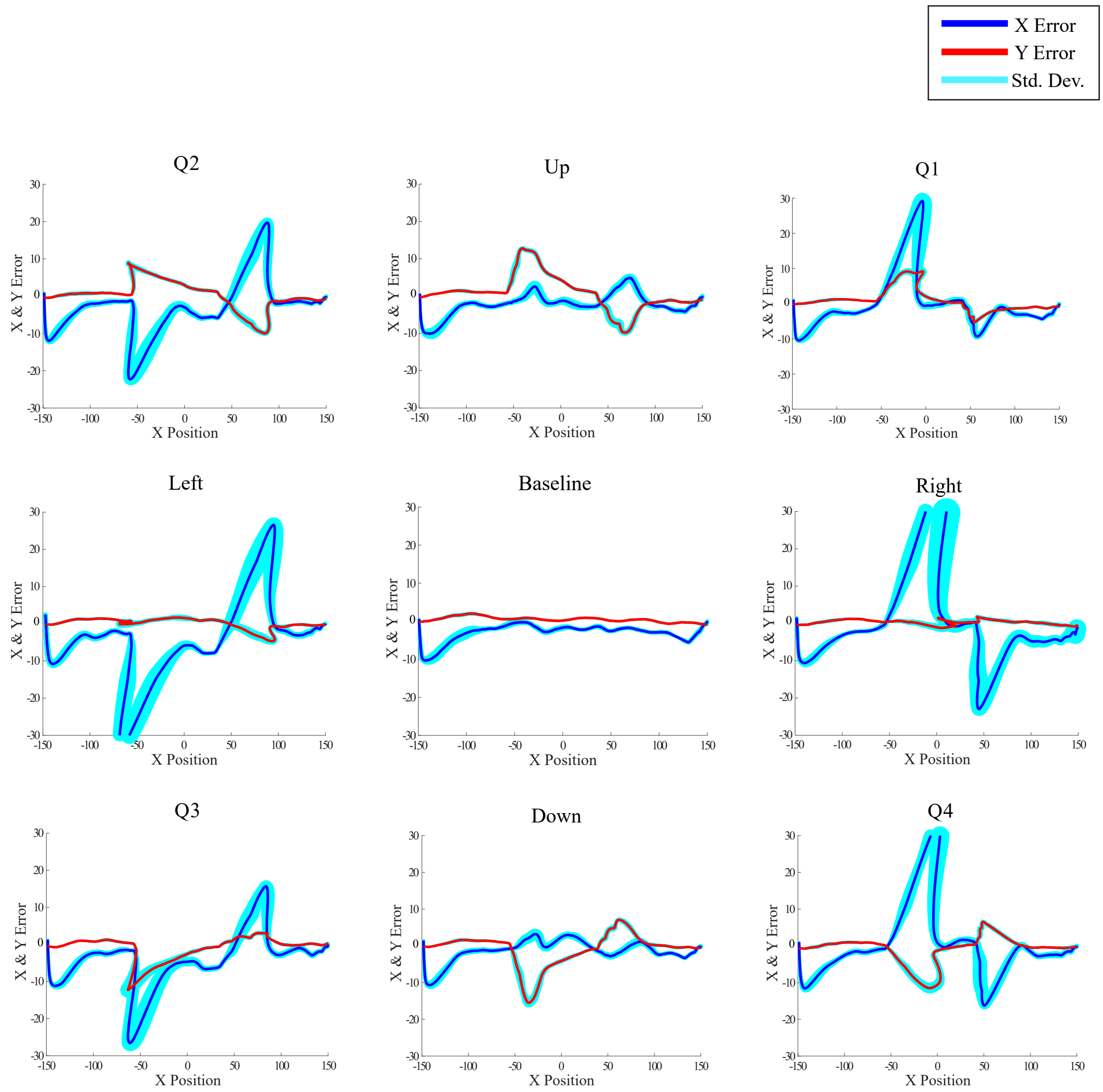

Figure 4.8: Average X & Y error of each force condition along the trial path with standard deviation at each point depicted with shading. All units are in mm.

4.2.2.1 Error Estimates

This section further assesses the potential differences in directional error and explores how errors might combine. This is an endeavor to shed light on the ambiguity discussed in the previous section between perturbations taken with respect to the forearm and perturbations with respect to the LOA. Error was estimated for a particular force condition by taking its two adjacent errors and combining them. For example, up and right condition data was used to simulate Q1 error. This simulation was then compared to the actual data for that condition. It is consequential to note that this method is unique from previously presented analyses. Results can be seen in Figures 4.10 & 4.11; slightly different methods were used for the diagonal and cardinal direction groups. The procedure for deriving these estimates is given below:

1. Appropriate component errors were taken from each component direction. For instance, in the case of estimating Q1, the X error of the right force condition and the Y error of the up were used. The absolute value was taken of all components, and the mean value at each data point was calculated.

2. The corresponding mean baseline error was subtracted from each component; e.g., if the Y error from the up condition was used, the baseline (no force condition) Y error was subtracted.

3. A scaling or correction factor was then applied.

   (a) For diagonal directions, a scaling factor of 80/20 was applied to get a good match between the estimate and actual errors. The component errors were combined with 20% of the X error and the remaining 80% from the Y error. Note this is not a 50/50 combination: the consequences of this are discussed below.

   (b) For cardinal directions, a correction factor of $\frac{1}{\sqrt{2}}$ was applied. This was to account for Omni force limitations and is discussed at length below.

4. Each component, estimate and actual errors were then plotted for comparison.

(a) For diagonal directions, actual error is the RMS error.

(b) For cardinal directions, actual error is the error only in that direction; e.g., only the actual Y error associated with up.

One assumption made in this process was that the relationship between force and error is linear ($E = a * F + b$); where E is the error (independent variable), F is the perturbing force (dependent variable), and a, b are constants. This means that, in the case of the diagonal direction estimates seen in Figure 4.9(a), some percentage combination of each error should be present. If the contribution is equal, then a simple average of the two errors should be a good approximation. If this is not true, then clearly one error component is contributing a majority of error.

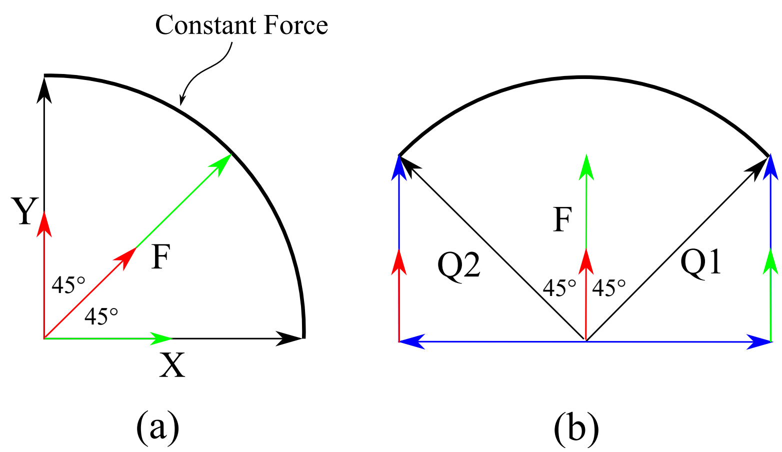

Figure 4.9: A representation of the force vectors and the vector addition required for simulation. (a) Perturbation forces from two axial direction conditions are used to recreate the diagonal force. No scaling is necessary in this case. (b) Component forces are taken from two diagonal directions and used to replicate the intermediate axial force. Scaling is necessary in this case since the average of the component forces is not equivalent to the magnitude of axial force.

Given the Omni's maximum total force output, scaling was necessary when using the error from a perturbation with more than one force component (i.e. diagonal directions). Since only one component error was taken from each condition (either X or Y), taking the average of two of these constituents would not be equivalent to the perturbation force or the resulting error, assuming a linear relationship between perturbation force and compensation error. Figure 4.9 demonstrates this case. Mathematically, this is shown by considering a force of unit magnitude representing the maximum force an Omni can display ($F = 1$). For a diagonal force at 45°, either component of the diagonal force would be $\frac{F}{\sqrt{2}}$. Thus, when using the average of diagonal component forces (or their corresponding errors), a correction factor of $\sqrt{2}$ must be used.

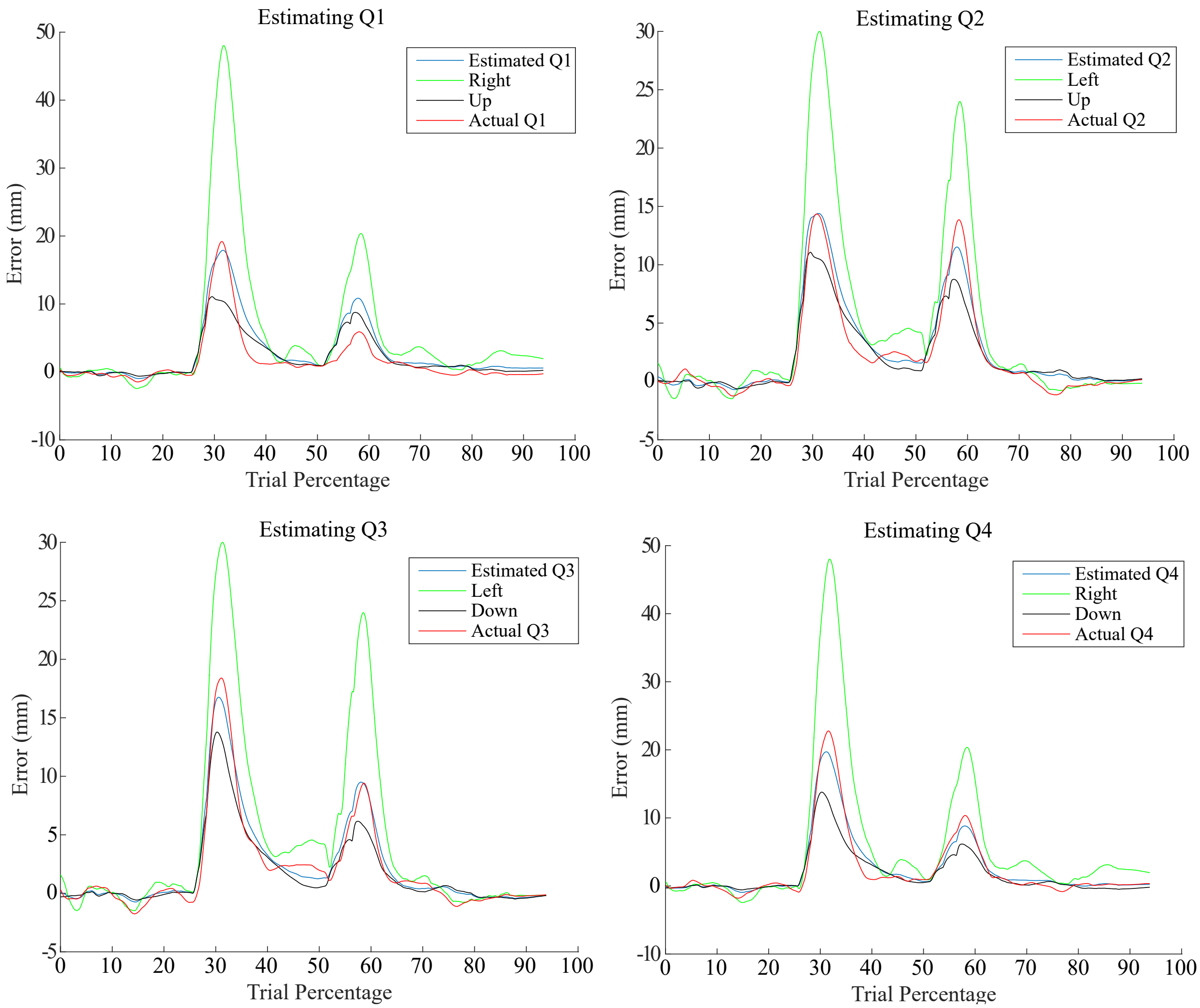

Figure 4.10: Comparing diagonal data with an estimate of the data generated from the two components' data.

As mentioned previously, one would expect an equal ratio when combining errors for a diagonal perturbation at 45°. However, using an equal ratio between the X and Y components did not provide a good match, and is, in fact, likely not the case. A ratio of 4:1 (Y:X error) provided a reasonably good estimate of the actual errors seen in Figure 4.10. The purpose of this ratio was not to determine the actual contributions from the X and Y components, but rather to show that their contributions are not equal. Furthermore, the 4:1 ratio use is different than ratio of Y error to X error for the baseline condition. This suggests that the error combination is not caused exclusively by the tracking task itself.

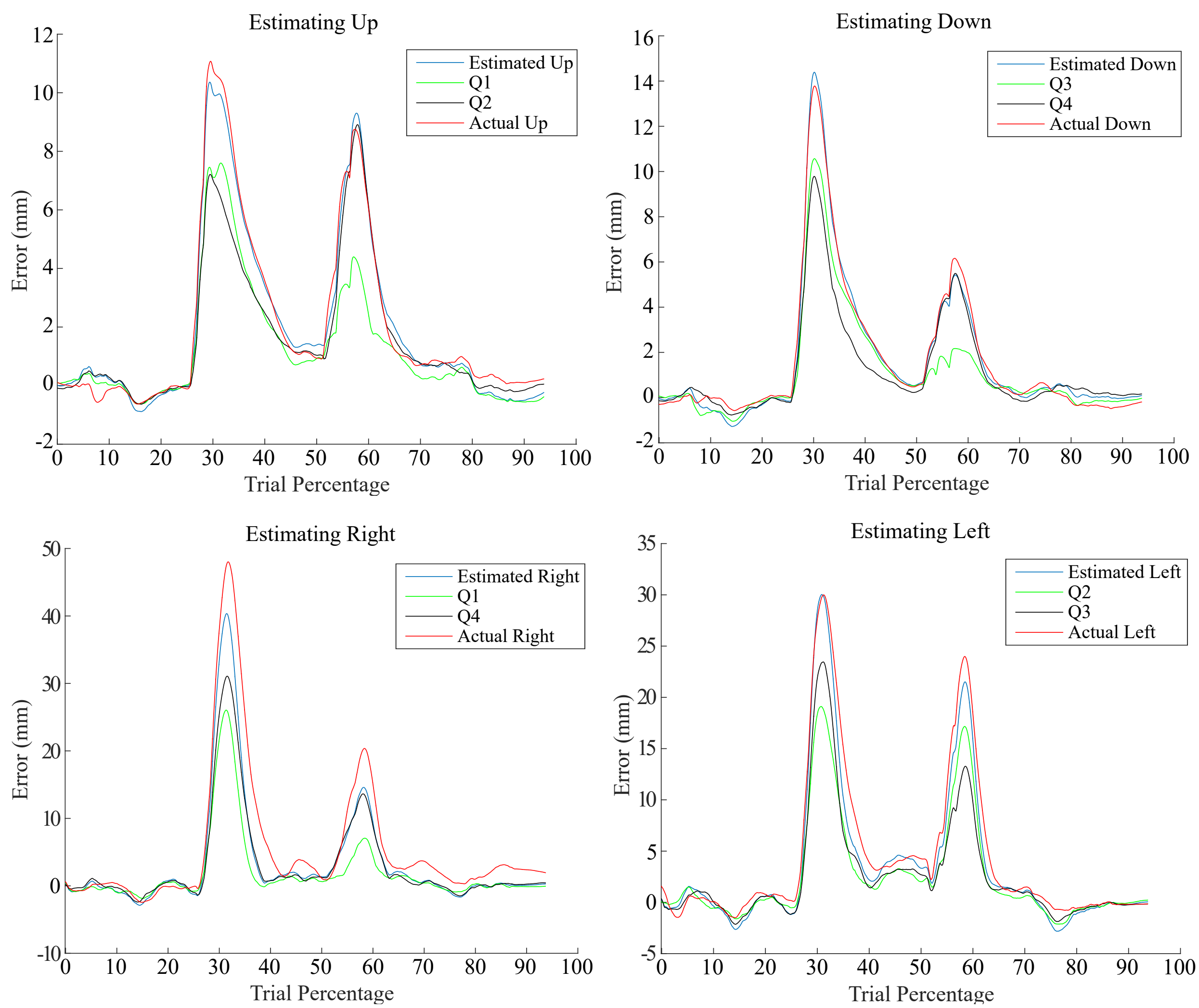

Figure 4.11: Comparing cardinal direction data with an estimate of the data generated from the two components' data.

Upon examining the cardinal direction estimates (see Figure 4.11), the results become more perplexing. Here a simple average of two component errors was used. For example, the Y error from Q1 and Q2 directions were used to estimate the up error, which was compared to the actual Y error of the up condition. This did not require a ratio adjustment presumably because X and Y errors were not combined (only one component was used per estimate).

Several phenomena seen here were also seen in a previous study [83]. In most of the conditions seen in Figures 4.10 & 4.11, the second error peak is significantly smaller than the first one. In both the previous study and this one, the are seemingly two plausible explanations for this. One is that participants were able to predict the end of the force field better than the beginning, and thus reduce their error upon exiting the field. The second explanation is that subjects are able to react more quickly to self-generated perturbations (exiting) than to externally generated ones (entering). Another phenomenon seen here and in previous work was a remarkably larger error peak in the right perturbation error versus the left (see Figure 4.11). The most obvious explanation of this is a biomechanical cause. However, there was no statistically significant difference shown between the hands, which one would expect if biomechanical differences between the hands had significant effects. One other cause might be how the motor system handles perturbations in goal directed tasks. It has been shown that forces with an assistive effect increase accuracy [39]. Although the right force was not pushing directly toward the target, it was pushing in the direction the target was moving.

4.2.3 Acceleration

Acceleration is a useful addition because it provides a window into intent. Once deceleration begins, one can assume the subject has started compensating for the perturbation even though direction might not be reversed yet. Thus, acceleration provides a different picture of the underlying system at work. Another example of the contrast between the position and acceleration plots is seen in Figure 4.12. The oscillatory acceleration responses can be seen upon the subject entering the force field, but position responses are much smoother (refer to Figures A.1, A.2, & A.3 in Appendix A). This is, at least in part, due to heavy reliance on visual feedback: a subject would find perceivable oscillations to be unnatural.

The acceleration was found numerically from position data using the central divided difference (CDD) method given in Equation 4.2 for the second derivative:

$$CDD = \frac{f(t_{i+1}) - 2f(t_i) + f(t_{i-1})}{(\Delta t)^2} \tag{4.2}$$

where the time step, Δt, is given by:

$$\Delta t = t(i+1) - t(i) \tag{4.3}$$

Time steps were filtered and interpolated and therefore assumed to be equal such that this numerical method is valid. Note: real world time was recorded and used here so that interpolation would not invalidate the acceleration numbers obtained.

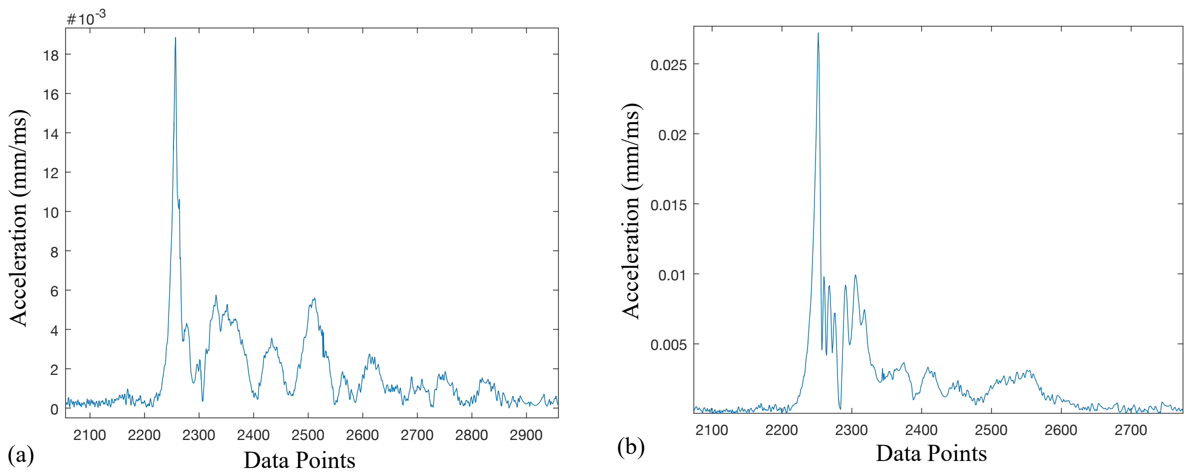

Figure 4.12: Examples of acceleration response when subjects entered force field during two perturbations. (a) is a perturbation toward the right. (b) is an upward perturbation.

The MATLAB algorithm *findpeaks* was used to find peak accelerations from the data with the maximum taken from the resulting peaks for each trial. The 3-way ANOVA showed statistical significance only for force condition ($F(8,40) = 72.33$, $p < 0.0001$). Hand ($F(1,5) = 1.68$, $p = 0.20$) and target speed ($F(2,10) = 0.87$, $p = 0.42$) were not statistically significant effects nor were any interactions between factors. Post-hoc tests using a Bonferroni correction were conducted on the statistically significant effect, force condition. Baseline was statistically significantly different than all other force conditions. See Figure 4.13 for all post-hoc comparisons. Left and right accelerations are not statistically significantly different, but up and down are. The same result was

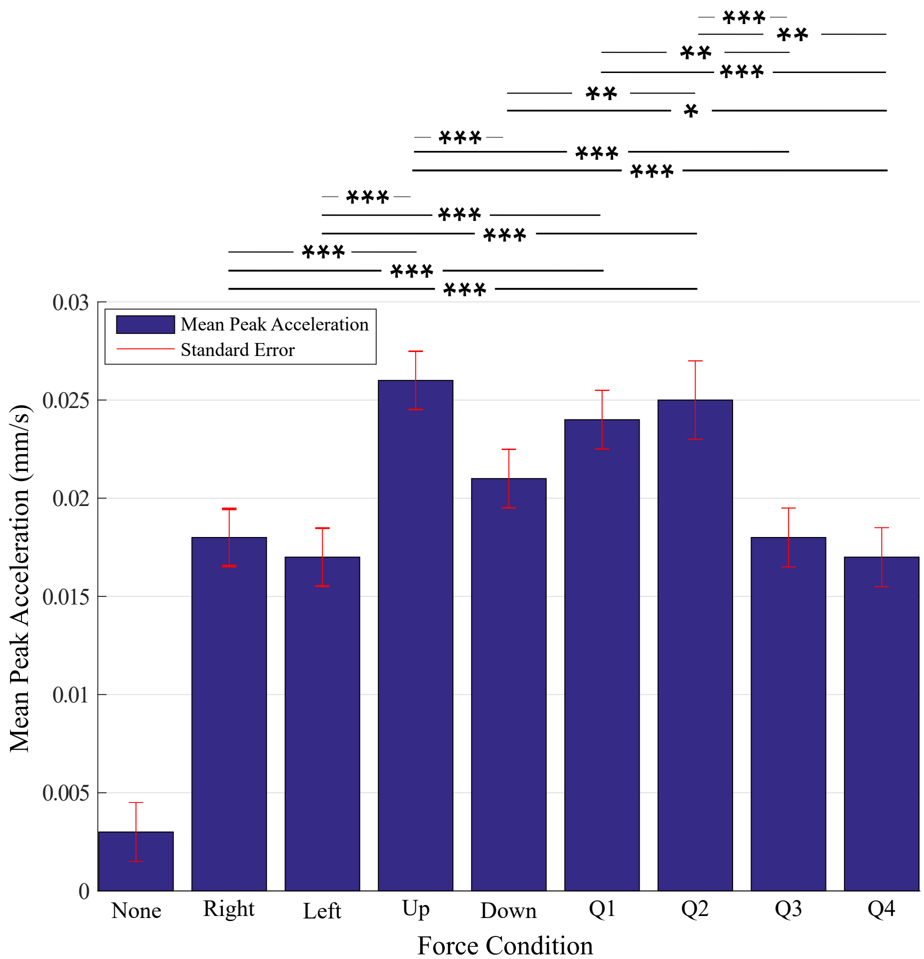

Figure 4.13: The post-hoc test with Bonferroni corrections is shown for the mean peak acceleration versus force condition. Notes: (1) baseline was statistically significantly different than all other conditions and is not explicit shown at top, (2) the bars at the top indicate statistical significance between pairs where * means $p < 0.05$, ** means $p < 0.01$, and *** means $p < 0.0001$.

seen in a study by Gentili et al. [84]. The up/down trend continues with the diagonal perturbations where Q1 and Q2 are not different from each other, but they are different from Q3 and Q4, which are not different from each other. This within group difference is likely due to the down condition being directed toward the body: a biomechanical difference.

4.2.4 Speed-Accuracy Effects

Results showed a clear increase in error with increasing speed, which agrees with previous research [56, 57]. Post hoc tests revealed each of the three speeds were statistically significantly

Table 4.3: Mean errors from each speed corresponding to each error type.

	RMS	**Y**	**X**
Slow	4.15	2.15	4.98
Medium	5.12	2.57	6.20
Fast	6.30	3.02	7.72

different from each other in each error case (RMS error shown in Figure 4.14. See also Tables 4.1 & 4.3).

As argued in Section 4.2.2, from a controls perspective, there is a difference in the input or reference goal to the human system between the X and Y directions (refer to Figure 4.3 for error definitions). In a tracking task, the Y reference is a step input while X reference can be thought of as a ramp function. This theory can further the explanation in this case of the decreasing accuracy with increasing speed. Given the relatively constant reaction times (RTs) for a subject seen in previous studies [41], an increase in speed between otherwise identical trials would yield an increase in the distance the target traveled between RTs.

Fitts' law, discussed in Section 2.4, can be rewritten in a linear form with the index of difficulty, ID, as the independent variable (Equation 4.4 shows this linear form). Similarly, there appears to be a linear relationship between mean error and speed in Figure 4.14. If one considers error as the distance between the subject's position and the target's center, then the distance in Fitts' law (Equation 2.2) is defined. Similarly, movement time, MT, can be thought of as the time it takes the participant to move across this distance. Finally, W, the width of the target, remains constant across all trials in this experiment.

$$MT = a + b * ID \tag{4.4}$$

However, this experimental design and Fitts' original definition may not be directly compatible. The distance to the target, as defined above, does not remain constant since the target's velocity relative to the subject's may not be zero or even a constant (the subject accelerates or decelerates at times). Additionally, the subject must counter a force field, which affects movement time. The question of moving targets (i.e., changing D) has been addressed previously with a different experimental design (see Section 2.4 for more details). Chiu et al. [57] found that using a constant ID, as defined by Fitts, but varying velocity resulted in a negative correlation between velocity and error (the same is seen in Figure 4.14).

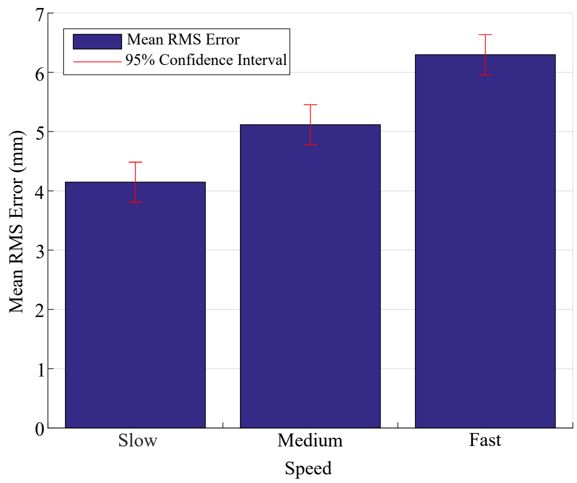

Figure 4.14: The average and standard errors are shown here for the mean RMS error vs. speed condition.

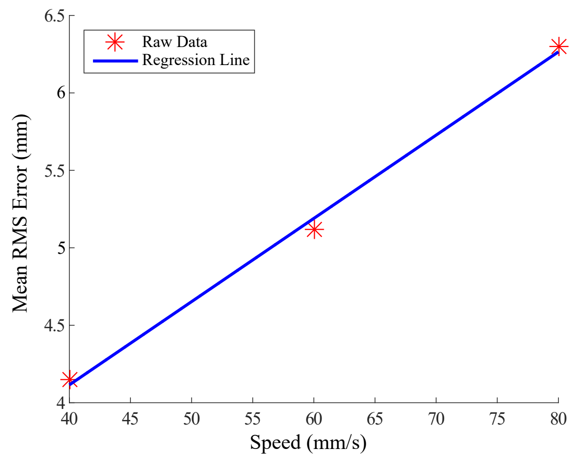

Figure 4.15: Linear regression equation for mean RMS error versus speed plotted with original data points.

$$E = a + b * v \tag{4.5}$$

Thus, the relationship between Equations 4.4 & 4.5 cannot be confirmed from this experiment alone. The linear regression equation is given in Equation 4.5; where a and b are constants proportional to a subject's reaction time and task specifics (target width/distance, accuracy requirements, etc.), E is error, and v is target velocity. Using the data from Figure 4.14, the linear regression model for Equation 4.5 is $E = 0.538 + 1.97v$ with $R^2 = 0.99$ (see Figure 4.15). It is unclear whether this linear relationship extends beyond these results based on this data only. More speeds are necessary to verify the relationship seen here between speed and accuracy for a tracking task where both movement time and movement amplitude are constrained.

CHAPTER 5: CONCLUSIONS

The work presented in this thesis expands the field of knowledge of human perception. Understanding human perception is essential to designing human-computer interfaces as new technology may cause unforeseen biases. In fact, nuances in interaction with the same tool or system may change our perception [85]. New tools and systems designed to advance human motor ability must also be implemented in a way that humans can benefit from [39]. Thus, investigating novel paradigms such as the tracking task presented in this thesis or the factors affecting force reproduction bimanually are important in a broader context.

Two studies have been presented in this thesis relating to human interactions with forces. In the first study, force magnitude and the number of fingers were shown to have significant effects in a bimanual force recreation task regardless of skill level. Skill level was not a main effect on the task; however, this is likely due to the small sample size. Subjects in the high skill group were not able to judge their performance better than the average skill group according to responses given in a post-experiment survey.

The second study examined the effects of perturbation direction on force compensation accuracy in a novel tracking task. Statistical significance was seen between parallel and perpendicular directions, and evidence was presented that the X and Y errors do not provide equal contributions in a diagonal direction. Furthermore, an analysis of peak acceleration was presented, which strengthened the conclusions obtained from the other metrics for directional effects. Additionally, the speed-accuracy tradeoff was investigated by varying the speed of the target subjects were tracking. The negative correlation between speed and accuracy persisted in the tracking task presented here.

5.1 Main Contributions

The main contributions of this thesis are:

1. A relationship between force magnitude and recreation accuracy was demonstrated.

2. A relationship between finger configuration (the number of fingers used) and recreation accuracy was demonstrated.

3. A directional dependent effect of perturbation on compensation accuracy was examined.

4. Speed-accuracy tradeoff effects were shown to continue in a tracking task (continuous error correction) scenario.

5. In depth results are presented for a novel tracking task, which allows the study of error unmeasured in previous work.

5.2 Future Work

There are several possible extensions of this research. For the force compensation study, an experiment to test static force compensation of each arm for different force directions. This would involve a target at a static location, which subjects would be perturbed out of by a force and then have to return to. To clarify the effects of biomechanics, the tracking task described in this thesis should be implemented for at least one other line of action. Additionally, the effects of Fitts' law should be investigated for a tracking task by extending the experiment done by Chiu et al. [57] for moving targets for two cases where the cases have perpendicular lines of action.

For the bimanual force recreation study, I have already begun work on a follow up experiment. The major changes in the new experiment will be a redesigned device with a force sensor for each finger allowing for individual monitoring of force output. Further inquiry should be conducted into the effect the number of fingers has on the accuracy of force perception and whether a deficit in perception accuracy can be found similar to the MVC deficit found in previous work [74]. Additionally, more forces and finger combinations should be tested to investigate the haptic capabilities of individual and groups of fingers and how Weber's law might relate to bimanual force recreation.

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APPENDIX A: ADDITIONAL FIGURES

A.1 Position Paths

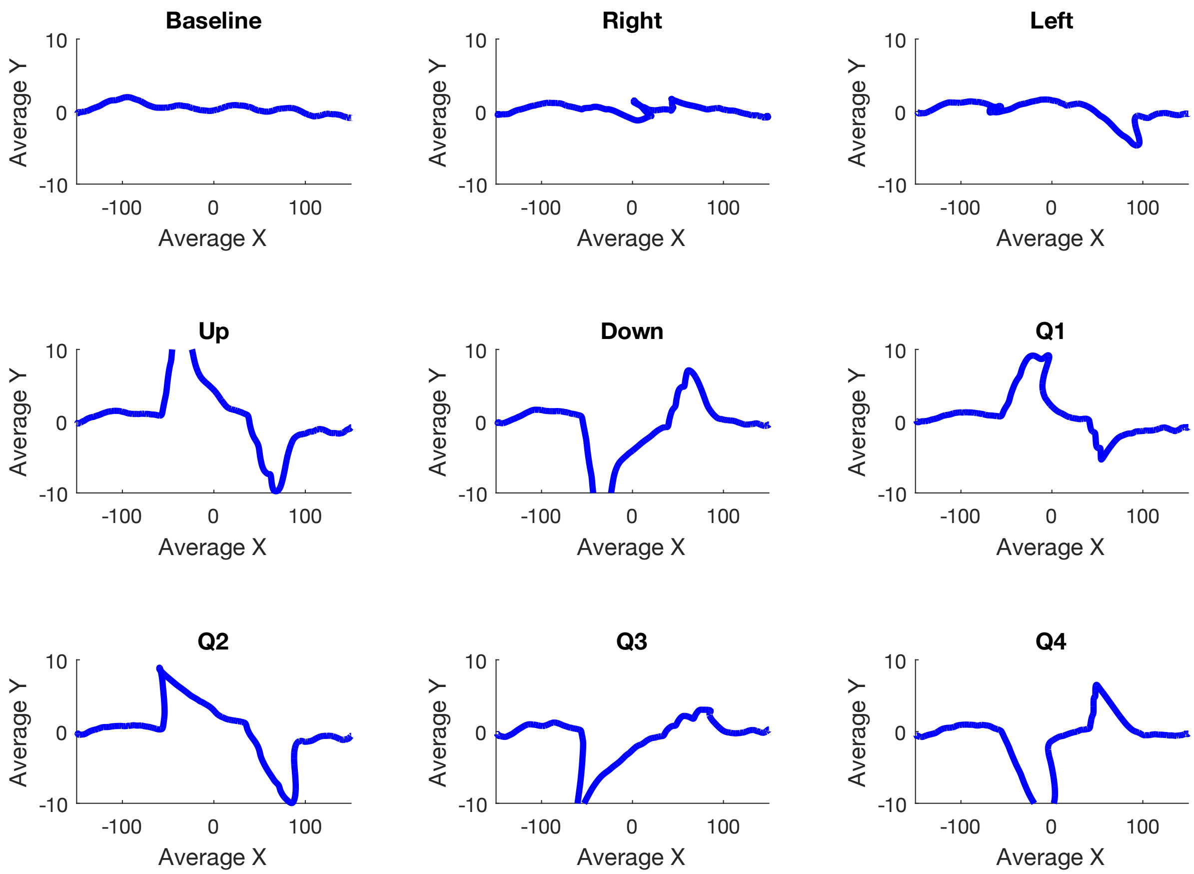

Figure A.1: The average Y vs. X position across all subjects for the duration of the trial.

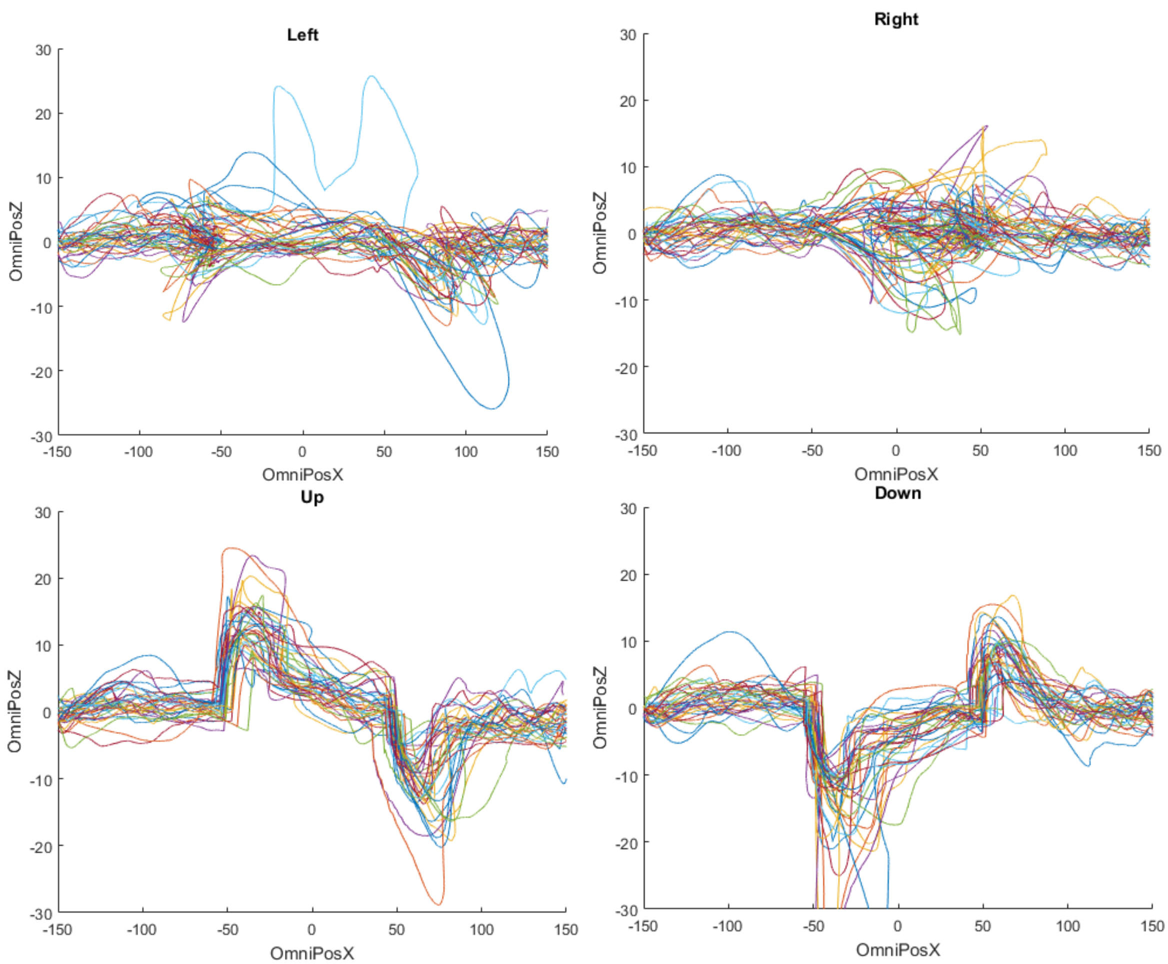

Figure A.2: Raw 2D paths of the four cardinal direction force perturbations are given across all the subjects.

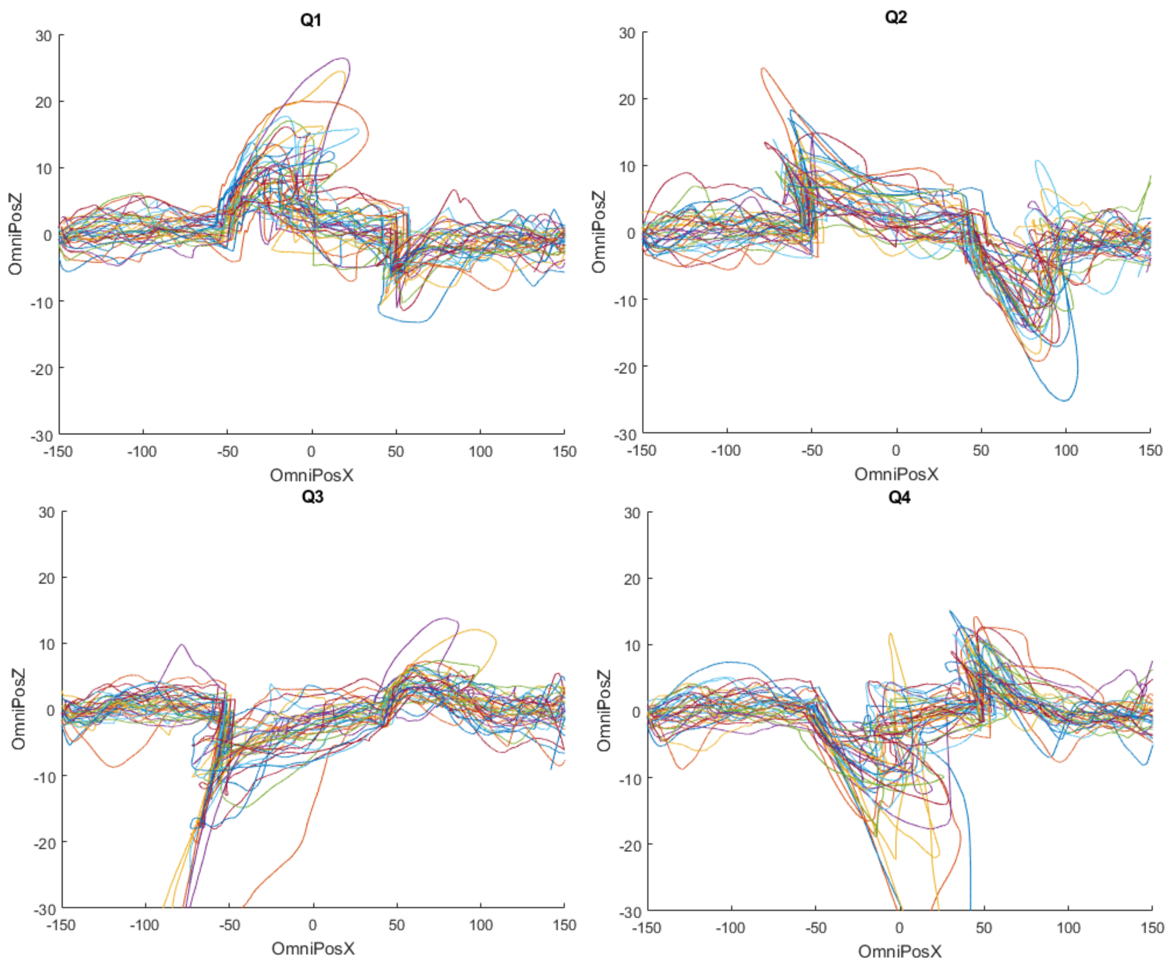

Figure A.3: Raw 2D paths of the four diagonal direction force perturbations are given across all the subjects.

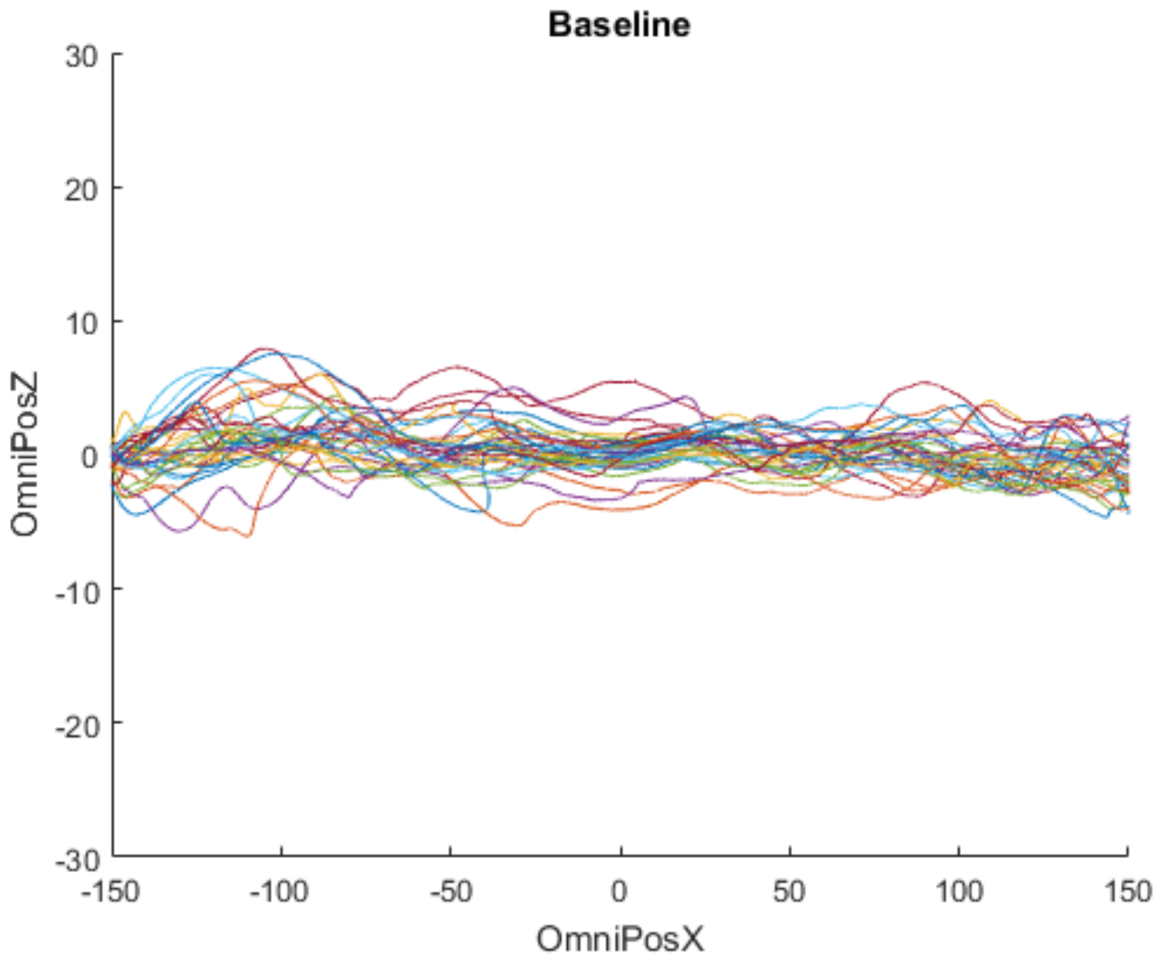

Figure A.4: The 2D path for the duration of the trial for all the subjects' baseline data (no force perturbation).

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