



2016-12-01

# The Feasibility of Using a Markerless Motion Capture Sensor (Leap Motion™ Controller) for Quantitative Motor Assessment Intended for a Clinical Setting

Clay Jordan Kincaid  
*Brigham Young University*

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The Feasibility of Using a Markerless Motion Capture Sensor (Leap Motion™ Controller) for  
Quantitative Motor Assessment Intended for a Clinical Setting

Clay Jordan Kincaid

A thesis submitted to the faculty of  
Brigham Young University  
in partial fulfillment of the requirements for the degree of

Master of Science

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Department of Mechanical Engineering  
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## ABSTRACT

### The Feasibility of Using a Markerless Motion Capture Sensor (Leap Motion™ Controller) for Quantitative Motor Assessment Intended for a Clinical Setting

Clay Jordan Kincaid  
Department of Mechanical Engineering, BYU  
Master of Science

Although upper limb motor impairments are common, the primary tools for assessing and tracking these impairments in a clinical setting are subjective, qualitative rating scales that lack resolution and repeatability. Markerless motion capture technology has the potential to greatly improve clinical assessment by providing quick, low-cost, and accurate tools to objectively quantify motor deficits. Here we lay some of the groundwork necessary to enable markerless motion capture systems to be used in clinical settings. First, we adapted five motor tests common in clinical assessments so they can be administered via markerless motion capture. We implemented these modified tests using a particular motion capture sensor (Leap Motion™ Controller, hereafter referred to as the Leap Motion sensor) and administered the tests to 100 healthy subjects to evaluate the feasibility of administering these tests via markerless motion capture. Second, to determine the ability of the Leap Motion sensor to accurately measure tremor, we characterized the frequency response of the Leap Motion sensor. During the administration of the five modified motor tests on 100 healthy subjects, the subjects had little trouble interfacing with the Leap Motion sensor and graphical user interface, performing the tasks with ease. The Leap Motion sensor maintained an average sampling rate above 106 Hz across all subjects during each of the five tests. The rate of adverse events caused by the Leap Motion sensor (mainly jumps in time or space) was generally below 1%. In characterizing the frequency response of the Leap Motion sensor, we found its bandwidth to vary between 1.7 and 5.5 Hz for actual tremor amplitudes above 1.5 mm, with larger bandwidth for larger amplitudes. To improve the accuracy of tremor measurements, we provide the magnitude ratios that can be used to estimate the actual amplitude of the oscillations from the measurements by the Leap Motion sensor. These results suggest that markerless motion capture systems are on the verge of becoming suitable for routine clinical use, but more work is necessary to further improve the motor tests before they can be administered via markerless motion capture with sufficient robustness for clinical settings.

Keywords: leap motion sensor, markerless motion capture, motor assessment, motor test, motor control, neurological exam, clinical rating scale, kinematics, hand, fingers, palm

## ACKNOWLEDGEMENTS

I would like to thank Dr. Steven Charles for his guidance, encouragement, hours of hard work and partial funding of the project. It was a pleasure working with him. I would also like to acknowledge and thank Paula Johnson, Michael McCain, Dan Carpenter and Kevin Cole for their technical support. Most significantly I would like to thank my lovely wife Stacie for her patience and constant support throughout our graduate school experience.

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## 1 INTRODUCTION

Despite the fact that millions of Americans are living with upper limb motor impairments [1-5], the current tools for assessing and tracking these types of impairments in a clinical setting are inadequate and need to be improved. Upper limb motor impairments are most commonly due to brain injuries caused by stroke, trauma, or neurodegenerative disorders. Clinicians typically assess these deficits using a number of standardized tests in which they observe a patient's movements and rate them on a rough numerical scale [6-9]. Although these methods are widely used, they carry certain challenges that prevent optimal, individualized care. First, as each injury is unique, there are subtleties in each patient's movements that observers simply cannot see. Second, the rough numerical scales are not fine enough to accurately quantify each individual's impairments. Third, there exists inherent subjectivity in each assessment as the clinician observes and rates patients. These same challenges are problematic when trying to track improvements in a patient's motor function due to treatment (rehabilitation, medication, or surgery); these challenges limit the opportunity to fine-tune treatment early and often. A more precise, objective, and automated method is needed to improve the diagnosis and treatment of the millions of people with upper limb motor impairments.

Methods have been developed to accurately and reliably track human movement, however these methods have their own challenges which make them impractical in a clinical setting and

prevent their widespread use. The advent of markerless motion capture technology (MMC), however, has introduced quick, low-cost, and relatively accurate motion capture sensors which have the potential to greatly enhance clinical assessments. Past methods of motion capture including optoelectronic systems, electromagnetic systems, inertial measurement units, and electrogoniometers require markers or sensors to be precisely affixed or calibrated to the patient's body by trained personnel, and the necessary setup takes too much time to be practical in a clinical setting—especially when frequent assessments are desired. The equipment necessary for such motion capture can also be prohibitively costly for many clinics and rehabilitation centers. Robotic methods of motion capture have also been developed [10-12], but these methods come with similar challenges, including lengthy setup time and high cost, which limit their practicality in a clinical setting. MMC systems eliminate many of these challenges. They require little to no setup, are very inexpensive, and are becoming increasingly accurate, making them ideal for widespread clinical use.

The purpose of this work was to adapt some of the most common clinical motor tests so they can be administered using MMC, and to evaluate the feasibility of administering these tests via MMC. We focused on a particular MMC sensor (Leap Motion sensor) and on five assessments that are common in neurological exams: reaction time, postural tremor, finger tapping, balance, and visually guided movement assessments. We developed computer programs that interfaced with the Leap Motion sensor and prompted the user to perform tasks similar to those in standard motor tests. To evaluate the feasibility of administering these tests via MMC, we implemented the software in the Leap Motion sensor and administered the tests to 100 healthy subjects. We present our observations of the subjects interacting with the modified assessments, the consistency of the sampling rate of the sensor, and discuss the type and

frequency of any problems that occurred during data collection. Additionally, we conducted a separate investigation of the frequency response of the Leap Motion sensor, which has not been attempted in previous studies. More specifically, we oscillated a plastic hand model at various amplitudes at frequencies between 1 and 15 Hz, recorded the hand position simultaneously with the Leap Motion sensor and a highly sensitive encoder, and compared the two signals. We present the magnitude ratios of the Leap Motion sensor for various frequencies and amplitudes and discuss a method to estimate the actual amplitude of tremor from the tremor amplitude recorded by the Leap Motion sensor.

## 2 QUANTITATIVE MOTOR ASSESSMENTS

### 2.1 Introduction

Although upper limb motor impairments are common, the primary tools for assessing and tracking these impairments in a clinical setting are generally subjective rating scales with limited resolution. Millions of Americans live with motor impairments caused by stroke, trauma, or neurodegenerative disorders [1-5]. It is likely that the prevalence of motor impairments will rise with increases in the aging population and further improvements in medical treatments and procedures. Presently, clinicians typically assess motor deficits by observing a patient's movements and rating them on a rough numeric scale. For example, a common task in a neurological exam is the "finger-to-nose" test in which a patient attempts to move their finger back and forth between the clinician's finger and their own nose. During the activity the clinician observes the patient's movement and rates them on a variety of characteristics such as tremor, dysmetria (undershoot and overshoot), movement speed, and so on. These types of scales generally have low resolution and are dependent on the observational acuity of the clinician. Furthermore, improvements in a patient's motor function due to treatment (rehabilitation, medication, or surgery) typically occur gradually, so observer-based assessments that rely on scales with coarse resolution inherently limit the opportunity to fine-tune treatment early and often. Despite clinicians' efforts to accurately assess motor deficits, a more precise, objective,

and automated method is needed to improve diagnosis and treatment of the millions of people afflicted with upper limb motor impairments.

The advent of MMC technology has introduced quick, low-cost, and relatively accurate sensors with the potential to enhance clinical assessments, but these assessments must first be adapted to be compatible with MMC. Past methods of motion capture, which include optoelectronic systems, electromagnetic systems, inertial measurement units, and electrogoniometers, typically require markers or sensors to be attached and calibrated to the subject's body. These processes are far too time- and labor-intensive to be practical for routine clinical use, especially when frequent assessments are desired. Additionally, the motion capture equipment typically costs between ten thousand and one hundred thousand dollars, which is prohibitively expensive for most clinics and rehabilitation centers. In contrast, MMC systems, such as the X-box Kinect, Organic Motion, and Leap Motion sensor, require little to no setup, are inexpensive, and are becoming increasingly accurate, making them ideal for widespread clinical use. In the future we envision MMC systems that record patients' movements and automatically assess motor deficits as the patient moves naturally in a clinical setting. As such MMC systems continue to increase in accuracy, they have potential to enhance clinical assessments, but current assessments must be adapted to be compatible with such systems.

The purpose of this work was to adapt some of the most common clinical assessments so they can be administered using MMC systems and evaluate the feasibility of their administration in a clinical setting. We focused on the five following assessments that are common in neurological exams: reaction time, postural tremor, finger tapping, balance, and visually guided movement assessments. This resulted in five modified tests called Quantitative Motor

Assessments (QMA). Our goal was to modify the traditional assessments as little as possible, but enough to enable their robust administration via MMC.

To determine the robustness of administering clinical assessments using MMC, we implemented these QMA using a particular MMC system—the Leap Motion sensor (Leap Motion Inc., San Francisco, CA) — and tested the QMA on 100 healthy subjects. Although we focused on the Leap Motion sensor in this study, our work also applies to other MMC sensors capable of tracking the position of a fingertip, palm, and tools.

In the following sections we describe the protocols of the QMA, how we implemented these QMA using the Leap Motion sensor, and the performance of the Leap Motion sensor during the administration of the QMA to 100 healthy subjects; however, the results of the QMA (i.e. the subjects' performance assessed using the measures described below) are not described here—they are presented in the work of collaborator Paula Johnson, PhD student in Neuroscience at BYU. As this was a feasibility study, we present the type and frequency of problems that occurred administering the QMA using the Leap Motion sensor, discuss our observations of subjects' experience with the QMA and interaction with the MMC system, and report the sampling rate of the Leap Motion sensor during the tests.

## **2.2 Methods**

### **2.2.1 Adaptation of Common Motor Tests for Markerless Motion Capture**

Clinicians conduct standard exams to assess the extent of impairment in patients with traumatic brain injury, cerebrovascular accidents (e.g., stroke), or neurodegenerative disorders. Such exams typically consist of batteries of individual tests, including a battery of tests to

determine motor deficits related to strength, muscle tone, reflexes, movement efficiency and speed, tremor, and postural control [6-9]. Although the details of the motor tests commonly administered differ for different disorders and clinician specialties (e.g., neurology vs. physical therapy), most batteries share a group of basic elements. We investigated a variety of motor batteries for different disorders and identified tests that are 1) most widely used and 2) most easily administered via MMC (Table 2-1). These tests necessarily exclude tests requiring the application or sensing of force (strength, muscle tone, reflexes).

**Table 2-1: Five Motor Tests Modified for Markerless Motion Capture**

Test	Behavior Attributes	Qualitative Measures of Conventional Tests	Quantitative Measures of Modified Tests
<b>Reaction Time</b>	Processing Time	N/A	Reaction Time
<b>Postural Tremor</b>	Upper Limb Postural Control	Hand Steadiness	Power Spectrum Area
<b>Finger Tapping</b>	Motor Speed, Movement Efficiency	Smoothness of Movement	Number of Taps, Regularity/ $\sigma$ of Amplitude and Tap Period
<b>Balance</b>	Postural Control	Estimate of the Amount of Sway	Normalized Mean Path of the Crown of the Head
<b>Visually Guided Movement</b>	Visuomotor Control, Intention and Kinetic Tremor	Smoothness of movement, Estimate of Over- or Undershoot to Target	Dysmetria, Path ratio, Power Spectrum Area

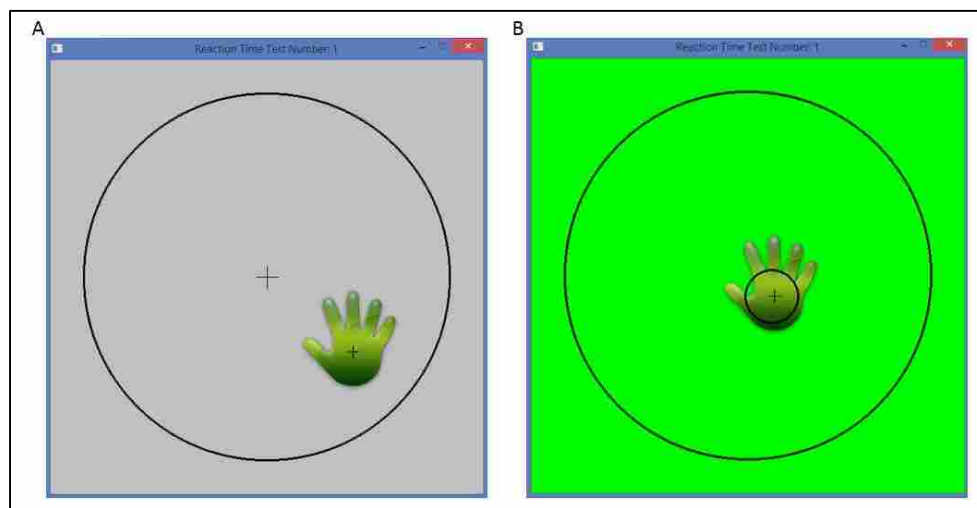
***Modified Reaction Time Test***

*Conventional Test:* The objective of a reaction time test is to measure the time it takes for a subject to react to a stimulus. There are many variations of the reaction time test, but among the most common tests is the simple reaction time test, which requires the subject to move his/her



hand or fingers as quickly as possible after a visual stimulus. The time between the visual stimulus and some measurable aspect of the movement (e.g., pressing a button) is recorded as the reaction time [13, 14].

*Modified Test:* For our test, we measured the time it took for the subject to move his/her hand a short distance after a visual stimulus. A blank grey screen was displayed with a crosshair in the center. Also displayed was a virtual hand that indicated the subject's hand position in the horizontal plane (Figure 2-1A). The subject was instructed to align the crosshair on the virtual hand with the crosshair in the center of the screen while maintaining the palm down and the fingers spread approximately 200 mm above the MMC system. As soon as the hand was in position—centered over the MMC system—the background turned white, which indicated to the user that they should hold still and anticipate a stimulus. After a period of time randomly chosen between 3 and 8 seconds, the screen turned green and the user moved his/her hand in any direction as quickly as possible. In the same instant the screen turned green a small circle (15 mm radius) appeared around the center of the palm in that instant (Figure 2-1B).



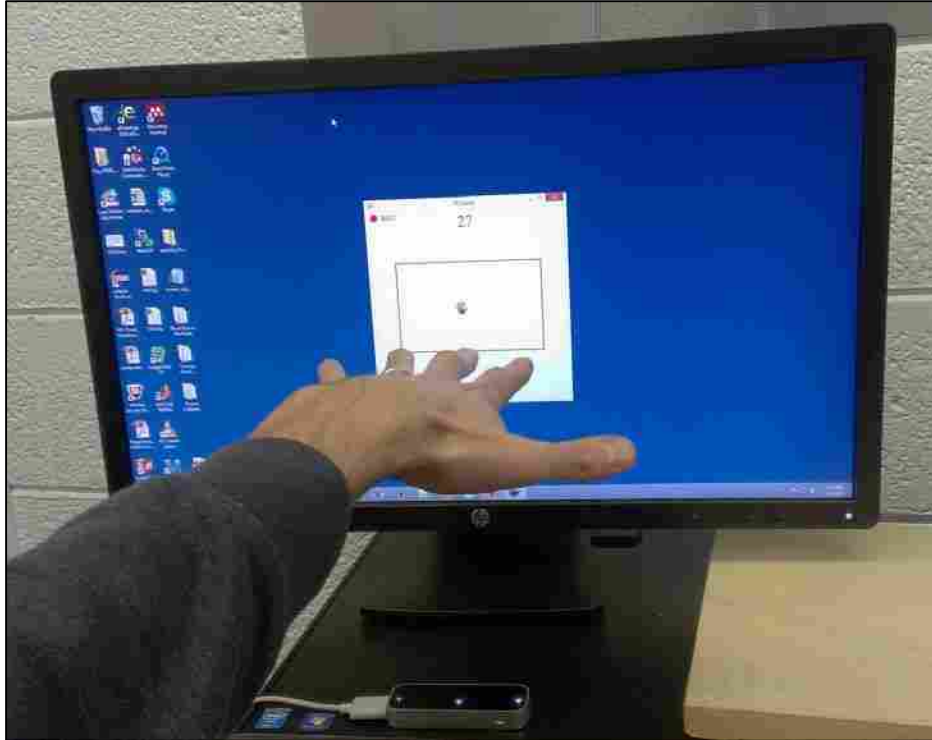
**Figure 2-1: Graphical User Interface of Modified Reaction Time Test.** (A) Grey screen is displayed while aligning the hand. (B) Screen turns green and a small circle appears at stimulus.

*Measures:* The reaction time was defined as the time it took for the subject's palm to exit the small circle after the stimulus. The small circle served two practical purposes: it allowed the subject's hand to drift before the stimulus without affecting the measured reaction time, and it allowed for a small amount of unintentional movement of the hand (e.g., tremor) after the stimulus without counting as intentional movement. The test was repeated 10 times on each hand and the average reaction time of 10 trials on each hand was calculated.

### ***Modified Postural Tremor Test***

*Conventional Test:* The postural tremor test measures the amount of tremor at the hand while the subject attempts to maintain his/her arm and hand in an unsupported posture. In a typical clinical test, the subject is often comfortably seated and the clinician instructs the subject to hold both hands straight out in front of his/her chest with arms parallel to the ground, palms down and fingers spread. The clinician assesses the tremor by observing and estimating the amplitude of oscillations using "the maximally displaced point of the hand of the greatest displacement along any single plane" [9].

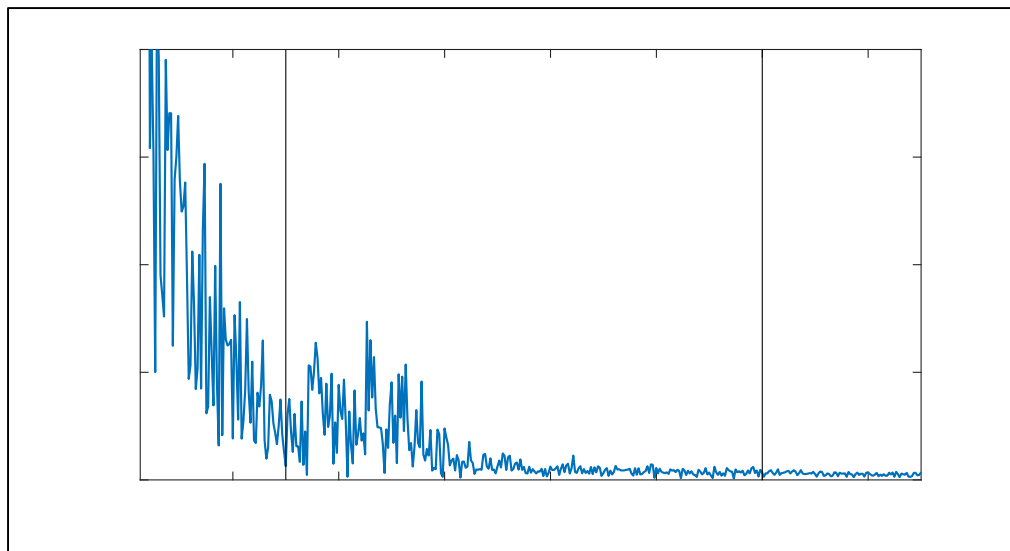
*Modified Test:* Little alteration was needed to adapt the conventional postural tremor test to MMC. Subjects were instructed to hold only one hand out at a time to ensure that the hand was centered over the MMC sensor. A window was displayed on the screen and a small hand was presented, which showed the position of the subject's hand in the horizontal plane (Figure 2-2). The subject was instructed to bring his/her hand near the center of the screen and hold it still approximately 200 mm above the MMC sensor. Once the subject's hand was in position and a 3-second countdown had elapsed, we began recording the position of the palm and finger tips for 30 seconds. The recording time was displayed on the screen. Two trials were performed for each hand in alternating order.



**Figure 2-2: Graphical User Interface of Modified Postural Tremor Test**

*Measures:* We calculated the area under the power spectrum curve between 3 and 12 Hz as a measure of overall tremor along each axis. To clarify, most types of tremor fall in the frequency range between 3 and 12 Hz [15], sometimes called the tremor band (Figure 2-3). Therefore, the area under the power spectrum curve over this frequency range represents the total amount of power in the tremor band. Because subjects with tremor do not always exhibit narrow peaks in the power spectrum, and because measures must be calculated for all subjects—even those without tremor—the area is a more robust measure of tremor than the amplitude of individual peaks. The area under the power spectrum curve was calculated individually for movement along each axis as follows. The positional data of a single 30-second trial were detrended by subtracting the mean from each data point. The data, which were collected at a variable sampling rate around 100 Hz, were then interpolated at a constant rate of 100

samples/sec with a piecewise cubic hermite interpolating polynomial (pchip) using Matlab's interp1 function. The power spectrum was calculated by Welch's method using a Hamming window, 50% overlap, and 4096 discrete Fourier transform points (implemented with Matlab's pwelch function). Finally, the area under the power spectrum curve was calculated by trapezoidal integration. This resulted in one measure value for each axis, hand, and repetition. Repetitions were averaged, reducing the measure values to six—one for each axis and hand.



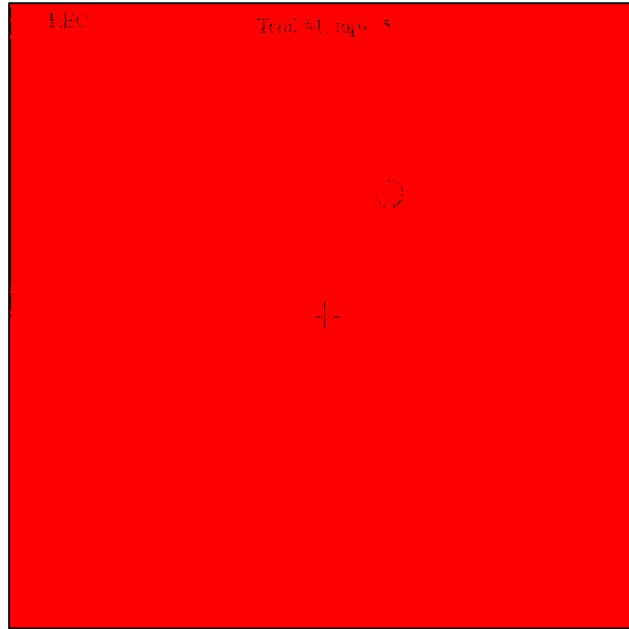
**Figure 2-3: Typical Power Spectrum Curve for Hand Tremor**

### ***Modified Finger Tapping Test***

*Conventional Test:* The traditional Finger Tapping Test (FTT) measures the number of times that a subject can tap his/her index finger in ten seconds. The subject rests his/her hand and fingertips on the table while resting his/her index finger on a small lever attached to a specialized counting device. The subject is instructed to press and release the lever repeatedly with his/her index finger while keeping the ball of the hand and other fingers stationary on the table. If the subject fails to completely press and completely release the lever, the tap is not counted.

Typically, the trial is repeated until five consecutive trials within five taps of each other have been recorded or until ten trials have been completed. The average tap count of those five consecutive trials is then computed. If, after ten trials, five trials within five taps of each other have not been recorded, the average of all ten trials is calculated [8].

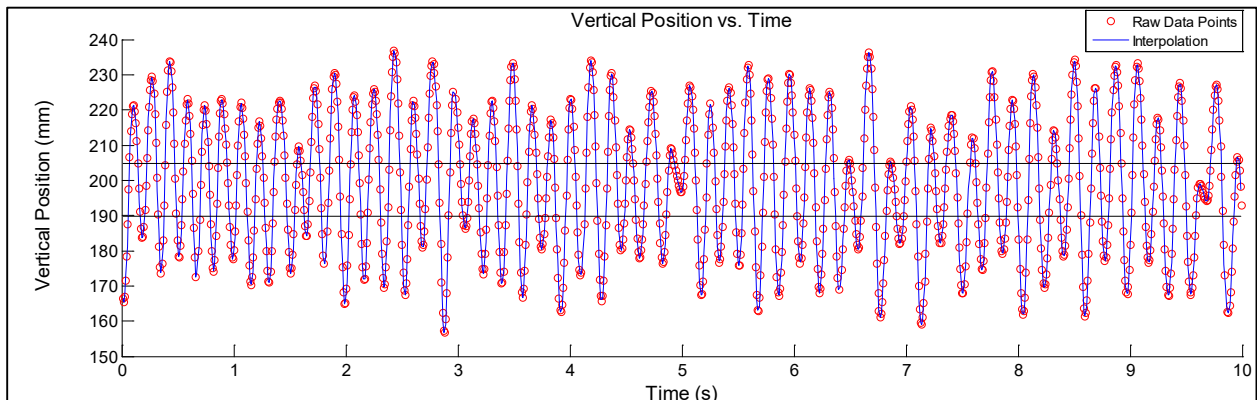
*Modified Test:* We modified the FTT so that it could be administered with MMC instead of the counting device. In the modified test, the subject held his/her hand steadily above the sensor instead of resting it on a board. A window was displayed on the computer screen containing two horizontal lines spaced 15 mm apart as shown in Figure 2-4. The distance between these two lines was equal to the lever stroke amplitude of the traditional counting device. The subject pointed directly into the screen with his/her index finger while keeping the rest of the fingers balled into a fist to keep them out of the way of the sensor. A solid black dot was also displayed that dynamically indicated the position of the subject's fingertip in the x-y plane. The subject was instructed to wag his/her index finger up and down as fast as possible for ten seconds such that, for each oscillation, the tip of the index finger started completely above the top line and went completely below the bottom line. Similar to the traditional test, if the fingertip did not go completely above the top line and completely below the bottom line the tap was not counted. A tap was counted the instant the fingertip descended below the bottom line after having started above the top line and the number of successful taps was displayed on the screen in real time (Figure 2-4). In a recent study, Fazio et al. showed that obtaining five *consecutive* trials within five taps of each other is not essential [16]. Therefore, unlike the traditional test, the trial was repeated only until any five trials within five taps of each other were recorded or until a total of ten trials were recorded.



**Figure 2-4: Graphical User Interface of Modified Finger Tapping Test**

*Measures:* Our measures included the tap count and the standard deviation of the tap period and amplitude, averaged across trials. As mentioned above, a tap was counted the instant the fingertip descended below the bottom line after having started above the top line. As in the conventional test, we calculated the average tap count across the five trials within five taps of each other, or across all ten trials if there was no set of five trials within five taps of each other. In the event that, after six or more trials, there were more than one set of five trials within five taps of each other, the set containing the lower tap counts were used for the average. In addition, the average standard deviations of the tap period [17] and amplitude were calculated as measures of regularity. The tap period and amplitude were calculated from the minima and maxima as follows. First, the y-position data were interpolated at 100 Hz (as described above) and the minima and maxima were identified as the points when the forward difference of the y-position changed sign. Due to the highly oscillatory nature of the test, the minima and maxima were easily identified (Figure 2-5). In particular, the data lacked high-frequency oscillations that often

interfere with peak detection, so there was no need to low-pass filter or use thresholds for peak detection. That said, occasional spatial jumps in the data created false peaks, so any peak that followed its preceding peak by less than 30 ms was deemed erroneous and ignored. This criteria was set after analyzing the data from the first 50 subjects and determining that it reliably removed most of the errors. Furthermore, such rapid oscillations are beyond the tapping rates generally observed in the traditional Halstead finger tapping test [18] as well as the frequency band for tremor. Therefore they were attributed to the sensor. The period was calculated as the time between adjacent minima, and the amplitude as half the distance from a minimum to the next maximum. The standard deviations of the period and amplitude were averaged across the same ten-second trials used to calculate the average number of taps. Even unsuccessful taps—tapping motions that failed to start above the top line and go below the bottom line—were included in the calculation of standard deviations of the tap period and amplitude.



**Figure 2-5: Typical Fingertip Position Data from Modified Finger Tapping Test**

### ***Modified Balance Test***

*Conventional Test:* The traditional balance test measures the amount a subject sways while standing in place. The subject is instructed to stand with arms crossed in front of the chest for 30 seconds under five different conditions: with eyes open and feet together on a hard surface; with eyes closed and feet together on a hard surface; with eyes open and feet together on an AirEx ® balance pad (pad); with eyes closed and feet together on a pad; with eyes open and feet in a tandem stance on a hard surface. In the standard clinical test, the clinician measures the amount of time the patient is able to maintain his/her balance [19], while in a newer test included in the NIH Toolbox, sway is measured with an accelerometer attached to the subject's waist [7].

*Modified Test:* Our test retained the same five conditions, but in order to measure sway with a MMC sensor developed to track hands or tools, subjects wore a modified bicycle helmet with two dowels protruding from the forehead. The dowels, which had an 8 mm diameter and rounded ends, were rigidly attached to the helmet. The tips of the dowels were spaced 122 mm apart and extended approximately 160 mm in front of the helmet. The modified helmet weighed 0.68 pounds and was adjusted to fit snugly on the subject's head. The MMC sensor was fastened to a tripod and positioned in front of the subject approximately 150 to 200 mm directly below the tips of the dowels (Figure 2-6). The subject performed all five tasks as the MMC sensor recorded the tip position and orientation of the dowels. The subject was instructed to look forward at the wall during the balancing tasks and was otherwise given no visual feedback.





**Figure 2-6: Modified Balance Test**

*Measures:* Similar to the NIH Toolbox, we calculated the normalized path length of the crown of the head. From the raw data we calculated the position of the crown of the subject's head as a function of time as follows. In a pre-experiment calibration, the position of the point on the helmet immediately above the crown of the head was expressed in terms of the position and orientation of the tips of the two dowels. Using this calibration allowed the position of the crown of the head to be calculated in the reference frame of the MMC sensor. The normalized path length (NPL) of the subject's head was defined as the 3-dimensional path length of the crown of the head divided by the duration over which the path was measured, calculated for each trial as

$$NPL = \frac{\sum_i \Delta p_i}{\sum_i \Delta t_i} = \frac{\sum_i \|\vec{p}_{i+1} - \vec{p}_i\|}{\sum_i (t_{i+1} - t_i)} \quad (2-1)$$

where  $\Delta p_i = \|\vec{p}_{i+1} - \vec{p}_i\|$  is the distance between the position of the crown of the head at one sample ( $\vec{p}_i$ ) and the position at the next sample ( $\vec{p}_{i+1}$ ), and  $\Delta t_i = t_{i+1} - t_i$  is the time between samples. Note that our measure includes the entire 3-dimensional path length, whereas the NIH Toolbox measure is limited to the anterior-posterior axis. During pilot testing we found that the MMC sensor would occasionally lose track of one or both dowels because the dowel(s) left the sensing volume or because the subject's face interfered with recognition of the dowels. This temporary loss of a dowel resulted in erroneous spatial jumps between successive samples, which would have artificially inflated the NPL. To eliminate these jumps, we removed any successive samples with a distance of more than 5 mm from each other. Likewise, the temporary loss of both dowels caused a larger-than-normal time gap between successive samples, which could have artificially lowered the NPL. To eliminate these gaps, we removed any successive samples spaced more than 50 ms in time. Because the measure depends on the sum of individual segments, removing some segments has little effect on the measure as long as the number of removed segments is small (see Results).

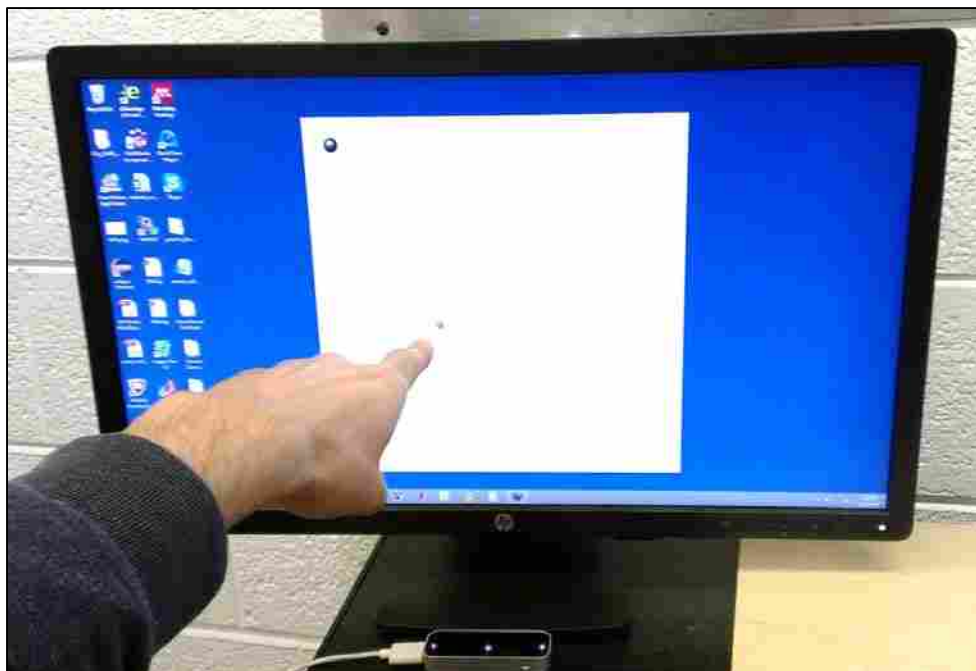
### ***Modified Visually Guided Movement Test***

*Conventional Test:* The goal of the visually guided movement test is to detect abnormalities in a subject's voluntary movements. In the conventional test (often call the "finger-to-nose" test), the clinician stands in front of the subject and holds his/her finger out as a target. The subject is instructed to reach out and touch the clinician's fingertip and then their own nose. This is repeated multiple times as the clinician moves his/her finger to different positions in front of the subject. The clinician looks for abnormalities such as tremor during movement (kinetic

tremor), undershoot or overshoot (dysmetria), slowness of movement (bradykinesia), movement decomposition, and jerkiness [9, 20].

*Modified Test:* We developed a goal-directed reaching task similar to the conventional visually guided movement test that was compatible with MMC. Instead of reaching between their nose and the clinician's finger, subjects reached to targets displayed on the screen in front of them. During initial tests we attempted to distribute targets in the depth dimension of the screen (using smaller icons for more distant targets) in addition to the two dimensions in the plane of the screen, but it became apparent that perceiving depth in this manner was not sufficiently intuitive. To reduce the sensitivity of the test to differences in subjects' ability to perceive depth on a computer screen, we limited the targets to the two-dimensional plane of the computer screen. Within this plane we presented four targets, located according to the following two constraints. 1) The targets were spaced as far apart as possible to approximate the amplitude of the movements in the conventional test and require involvement of the shoulder and elbow, and not just distal joints, as in the conventional test. 2) Subjects pointed directly at the targets on the screen, so the targets had to fit on a common computer screen and remain within the sensing volume of the MMC sensor. These constraints resulted in four targets at the corners of a 200 mm invisible square. The targets had a 10 mm diameter, sized to approximate a clinician's fingertip. The subject pointed directly into the screen with the index finger while keeping the rest of the fingers in a fist and the tip of the index finger approximately 20 to 100 mm from the screen. A small red circle dynamically indicated the position of the fingertip (Figure 2-7). The subject was instructed to move his/her fingertip as quickly and as accurately as possible from one target to the next as it appeared. Each target was presented fifteen times so as to elicit each of the twelve possible movements (between the four targets) five times for a total of sixty movements. The

order of the 60 movements was pseudo-random, but the same order was used for every trial. As soon as the center of the cursor was within the target and moving slower than 50 mm/s, the target was considered reached and immediately disappeared. The subject was instructed to hold his/her finger in place until the next target appeared 500 ms later and then move immediately to the new target. Two trials were performed on each hand in alternating order. We recorded the vertical and horizontal position of the index fingertip as a function of time.

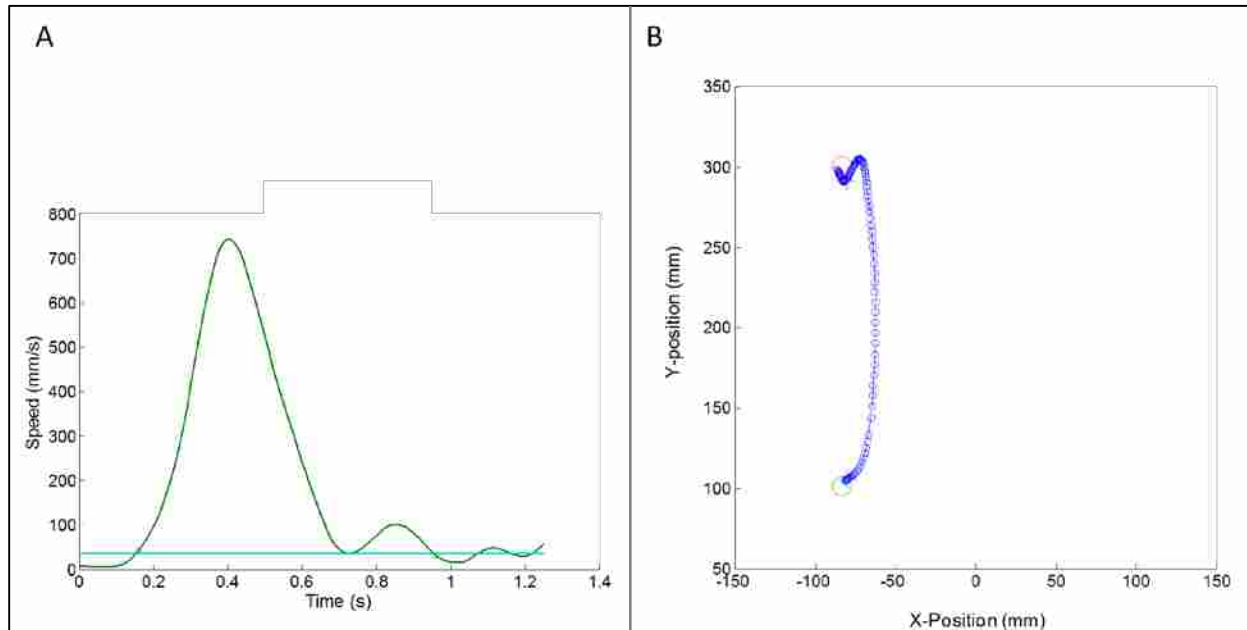


**Figure 2-7: Graphical User Interface of the Modified Visually Guided Movement Test**

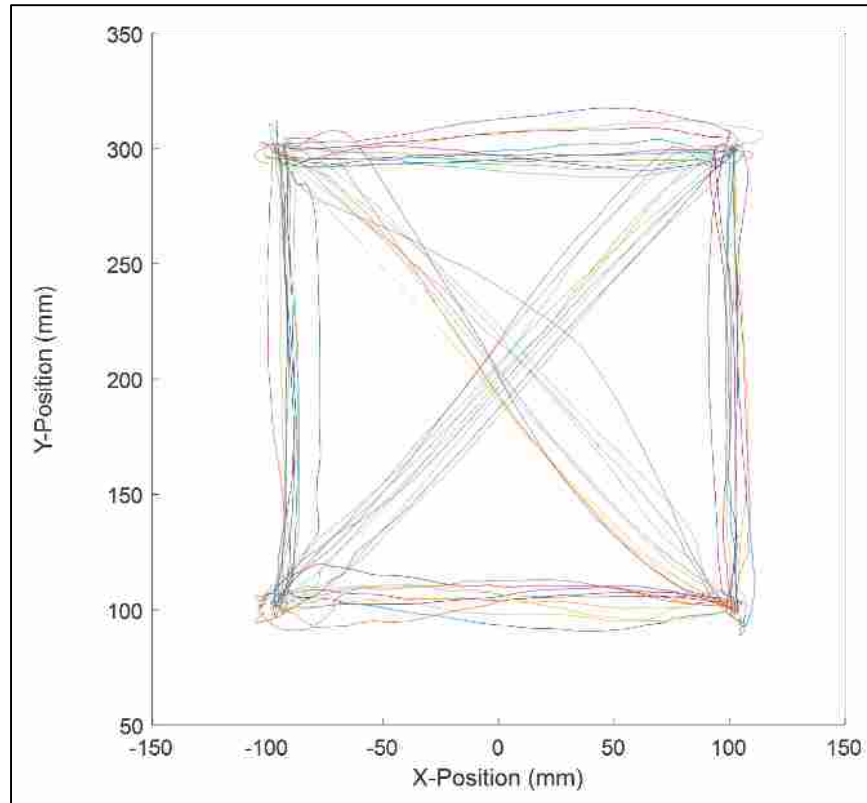
*Measures:* We calculated measures of dysmetria, path length, and kinetic tremor as follows. First, we removed errors that occurred occasionally during data collection. These errors generally took on one of three forms, any one of which rendered a single movement unusable: the sensor mistook the tip of the thumb for the tip of the index finger, the sensor momentarily lost track of the index finger altogether, or the subject did not wait the 500 ms before leaving the

target. These instances were easily identified. If there was a gap in time greater than 50 ms, a jump in position of more than 30 mm between samples, or the movement was less than 20 samples long, the movement was excluded from further analysis (except for calculating kinetic tremor—see below). Second, we interpolated the data at 100 Hz as described above. Third, we defined the beginning and end of each movement as 5% of maximum speed of that movement [21, 22]. More specifically, we calculated velocity in each direction using the forward-difference method and filtered with a 6th-order low-pass Butterworth filter with cutoff frequency at 6 Hz [23]. The speed in the vertical plane was calculated as the magnitude of the filtered velocities in the x- and y-directions. The start and end of a movement were defined as the final instance the speed crossed the 5% threshold before the maximum speed and the first instance the speed crossed the 5% threshold after the maximum speed, respectively (Figure 2-8A). Movements that never dipped below the 5% threshold before or after the maximum speed were defined to start at the first sample after the target appeared or at the moment the cursor entered the target, respectively. Fourth, we calculated dysmetria as the shortest distance between the center of the cursor at the end of the movement (defined using the threshold criterion above) and the edge of the target, normalized by the shortest distance between the center of the cursor at the start of the movement and the edge of the target (Figure 2-8B). If the movement ended inside the target, the dysmetria was considered zero for that movement. Fifth, we calculated the normalized path length as the length of the path from the start to the end of the movement, divided by the linear distance between those positions [24]. Sixth, we calculated the area under the speed power spectrum between 3 and 12 Hz as a measure of kinetic tremor. We used speed to combine all three movement directions into a single measure. More specifically, the raw x-, y-, and z-positional data for an entire trial (all 60 movements combined) were interpolated, differentiated,

and filtered (all as described above). The speed was calculated as the magnitude of the velocity vector in all three dimensions. We calculated the power spectrum of speed and determined the area between 3 and 12 Hz as described above (see modified postural tremor test). All 60 movements of a typical trial can be seen below in Figure 2-9.



**Figure 2-8: Typical Kinematics from the Modified Visually Guided Movement Test.** (A) Speed profile of a single movement. The speed threshold is indicated by the horizontal line, and the movement start and movement end by the circles. (B) The path of the same movement. Each circle represents a sample. The green and red circles represent the start and end targets, respectively. The green and red plus symbols represent the movement start and movement end respectively.



**Figure 2-9: Paths of All Sixty Movements of a Typical Modified Visually Guided Movement Test**

### **2.2.2 Implementation and Testing of Quantitative Motor Assessments**

To evaluate the feasibility of administering these QMA via MMC, we implemented the tests using a particular MMC sensor and administered them to healthy subjects.

#### ***Subjects***

One hundred healthy, right-handed subjects participated in this study (49 women, 51 men). Subjects were  $30.6 \pm 9.7$  (mean  $\pm$ SD) years old (range 18-50),  $1.74 \pm 0.10$  m tall (1.42-1.98 m), and weighed  $77 \pm 17$ kg (51-145kg), with BMI of  $25.2 \pm 4.6$  (16.1 – 47.3). All subjects reported that they had no history of movement disorders. Following procedures approved by

Brigham Young University's Institutional Review Board, informed consent was obtained from all subjects.

### ***Markerless Motion Capture System***

We implemented the QMA with the Leap Motion sensor using version 2.3.1 + 31549 of the accompanying software. The Leap Motion sensor is a small, infrared optical sensor that costs approximately \$70. It is designed as a plug-in controller to any Mac or personal computer and allows the user to interface with a computer using hand motions. Its onboard, proprietary software is designed to recognize and track the positions and orientations of features of the human hand such as fingertips, knuckles, and the center of the palm. Additionally, it is programmed to recognize and track the tip and orientation of pointer-like tools like a pencil. It has a field of view of about 150 degrees that extends approximately 600 mm from the surface of the sensor. Because the Leap Motion software exclusively tracks hands and pointers, it is better suited for the adapted upper limb assessments than other MMC systems like the X-Box Kinect (Microsoft Inc., Redmond, WA) or the Organic Motion (Organic Motion, New York, NY).

Prior studies have reported on the accuracy of this sensor. Weichert et al. used a highly accurate industrial robot with a pointer-like end effector to test the accuracy of the sensor in static and dynamic situations. They reported that the accuracy was <0.2 mm in static situations and <1.2 mm in dynamic situations [25]. Similarly, by tracking a stationary mannequin hand simultaneously with a highly accurate optical system and the sensor, Guna et al. showed the its accuracy to be <0.5 mm in static situations [26]. Most recently Tung et al. performed a study, which included the tracking of actual subjects' hands as they performed a pointing task. This study reported significantly worse accuracy ( $17.30 \pm 9.56$  mm) in quasi-static situations than the other two studies as well as poor sampling frequency [27], however we suspect that these were



the result of errors caused by placing an infrared light emitting diode at the end of the fingertip. We therefore proceeded under the assumption that the accuracy as reported by Weichert and Guna was correct. Brown et al. also reported the Leap Motion sensor to have an end-to-end latency of 85 ms [28]. During preliminary studies, we found the resolution of fingertip position to be consistently less than 0.05 mm and the resolution of palm position consistently less than 0.3 mm along each axis in quasi static situations. Note that the QMA presented here could be administered using any MMC sensor as long as it can accurately track the fingertip of the index finger, the palm, and multiple tools (the two dowels in the modified balance test). Additionally the sensor should have a minimum sensing volume of approximately 2ft by 2ft by 2ft, a minimum sampling rate of 24 Hz (to allow calculation of the power spectrum area up to 12 Hz), and an adequate bandwidth to measure hand oscillations (around 5-6 Hz for the modified finger tapping test and up to 12 Hz for the modified postural tremor test).

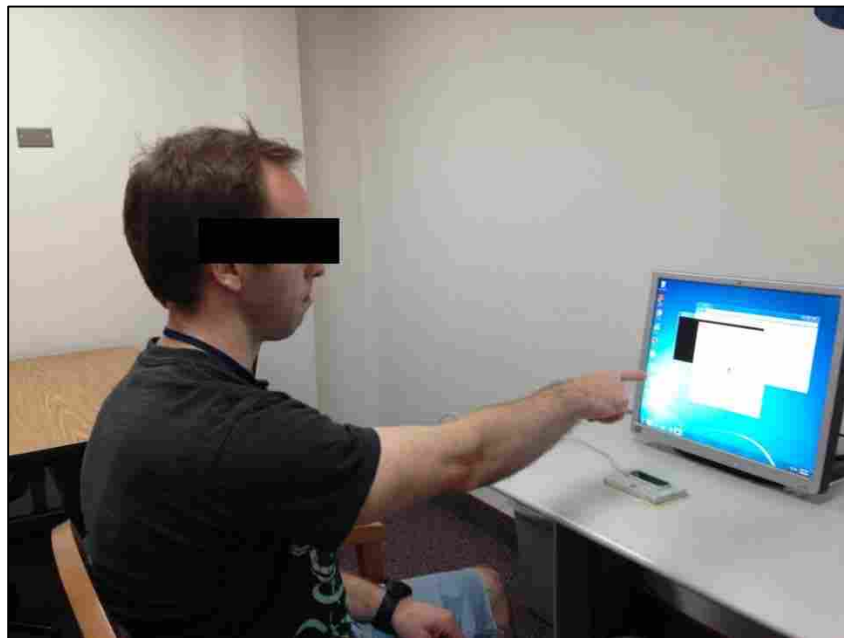
### ***Programming the Quantitative Motor Assessments***

We programmed the QMA in Python 2.7 using the Leap Motion application programming interface (API). These programs prompted the proctor to enter patient's name, instructed the user how to perform the QMA, provided visual feedback to the user with a graphical user interface (GUI), recorded the three-dimensional position of the user's finger(s) and/or palm, or tools and automatically calculated the measures described above upon completion of the assessment.

### ***Experimental Setup***

For all but the balance test, subjects were seated at a table approximately 750 mm in front of a computer screen centered in the sagittal plane. The sensor was positioned on the table in the sagittal plane 200 mm in front of the computer screen and the screen was raised such that the

center of the screen was 180 mm above the surface of the sensor. The subject held or moved his/her hands in front of the chest with the elbow slightly bent so as to stay within the most accurate portion of the sensors sensing volume (Figure 2-10). All tests were performed indoors under fluorescent lighting with the curtains drawn to limit the amount of infrared interference from outside as well as to mimic a typical clinical environment. A desktop computer was used to run the software and collect the data with a Windows 7 operating system, an Intel® Core™ 2 Quad Processor Q9550, 8 GB of RAM, and an Nvidia Quadro NVS 290 graphics card.



**Figure 2-10: Typical Setup for Administering the QMA**

The proctor used a two-step process to calibrate the system immediately before starting each battery of tests. First, the Leap Motion sensor was calibrated using the calibration program in the Leap Motion sensor's control panel. The Leap Motion software requires a minimum score

of 80/100 to ensure satisfactory tracking performance, but we required a minimum score of 90/100 before administering the test battery. Second, custom software was used to align the graphics display with the Leap Motion sensor so the cursor representing a subject's finger matched where the subject was pointing on the computer screen. The software displayed a series of 8 dots—the four corners of a square each displayed twice in sequence—located vertically and horizontally at a prescribed, constant number of pixels from the center of the screen. Keeping his/her index finger 5 to 10 mm from the screen, the proctor pointed directly at the center of each dot and recorded the position of their fingertip with the Leap Motion sensor. The recorded horizontal and vertical positions of all eight dots were averaged to estimate the center of the screen in Leap Motion sensor space as well as a pixel-to-mm ratio relating movement of the fingertip recorded by the Leap Motion sensor to pixels on the screen.

### ***Experimental Protocol***

In addition to the five modified tests described above (Reaction Time, Postural Tremor, Finger Tapping, Balance, and Visually Guided Movement), each subject also participated in three other tests for a companion study: the traditional Halstead Finger tapping Test, The Beery Visual-Motor Integration Test, and a standard grip strength test (not presented here). For practical purposes the grip strength test was performed first and the modified postural sway test was performed last. Otherwise, the order of the other six tests was randomized. Subjects were given short breaks between tests (1-3 minutes), and the entire battery of tests lasted approximately 90 minutes.

### ***Data Analysis***

The recorded variables (fingertip and/or palm position or tool tip position/orientation) were used to calculate the measures for each test as described above. Since the goal of this study was

to evaluate the feasibility of administering clinical tests via MMC, we present here 1) the ease-of-use of each QMA and its graphical user interface (GUI) according to our observations, and 2) the reliability of the Leap Motion sensor, quantified as average sampling rate and the frequency of adverse events (e.g., losing a finger, large time gaps between samples, etc). Unless otherwise stated, the average sampling rate and time step for each modified test was calculated from the total number data points across all trials and subjects. In other words, for each test the sum of all the time steps from every trial of every subject was found and divided by the total number of time steps.

### **2.3 Results**

With few exceptions, subjects found the QMA easy to use, and the Leap Motion sensor performed well. When adverse events did occur, they generally involved either an unusually large jump in time, which usually occurred when the Leap Motion sensor failed to recognize any hand feature or tool, or a sudden jump in space between two samples caused when the Leap Motion sensor briefly mistook one finger for another. Jumps in time affected all tests but occurred relatively rarely and were easily identified. As described earlier, after analyzing the data from the balance and visually guided movement tests, we defined time step thresholds, above which time steps would cause errors in the measures. We compared the difference between each sequential time sample in every trial to the defined threshold and removed the data as described earlier. No data were removed from the other tests due to a large time step. Jumps in space affected only those tests that depended on the position of fingers or tools (finger tapping, balance, and visually guided movement) but were more difficult to identify because of the non-constant sampling rate and varying velocities required by the tasks. To identify and remove

jumps in space, we used a number of task-specific criteria chosen after observing a large number of trials and adverse events. Below we describe the sampling rate, variability in time steps, and frequency of adverse events for each test (Table 2-2).

### **2.3.1 Modified Reaction Time Test**

Subjects understood the instructions and had no trouble reacting to the stimulus. The average sampling rate across all trials and subjects was  $108.9 \pm 6.4$  Hz (2.9-114.3 Hz). The average time step across all trials and subjects was  $9.3 \pm 38.6$  ms (0.84-22732.6 ms,  $N=1,275,785$ ), with 0.0019% beyond two standard deviations from the mean. No data were removed on account of large temporal or spatial gaps.

### **2.3.2 Modified Postural Tremor Test**

The (healthy) subjects had no trouble using the GUI or holding their hand over the sensor for 30 s. The average sampling rate across all trials and subjects was  $109.3 \pm 1.3$  Hz (108.8-113.6 Hz). The average time step across all trials and subjects was  $9.2 \pm 1.7$  ms (0.2-73.8 ms,  $N=1,311,027$ ), with 0.81% beyond two standard deviations from the mean. We recorded the three-dimensional position of the finger tips and palm but found that the palm data were more reliable; the Leap Motion sensor occasionally lost track of one or more finger tips. Averaged over all subjects, this occurred on  $5.5\% \pm 10.6$  of samples, ranging from 0% (never lost a finger) to 100% (never included all fingers) during an entire test consisting of four 30-second trials. In contrast, the Leap Motion sensor tracked the palm consistently. Therefore, we calculated tremor using only the palm position data and no data were removed on account of large temporal or spatial gaps.

### **2.3.3 Modified Finger Tapping Test**

The GUI was intuitive to use and subjects had no trouble holding their hands up for the duration of the test. Their hands easily stayed centered on the screen and they had no apparent trouble wagging their finger above and below the lines. However, the cursor often lagged the movement of the index finger, making it difficult for some subjects to use visual feedback effectively. Whether the lag was caused by the sensor or computer components involved in the graphical display is unknown. The average sampling frequency across all trials and subjects was  $108.8 \pm 3.8$  Hz (55.4-113.6 Hz). The average time step was  $9.2 \pm 3.1$  ms (0.1-850.5 ms,  $N=1,129,909$ ), with 0.94% beyond two standard deviations from the mean. Generally there were few sensor errors during data collection. The average number of false minima or maxima caused by sensor error (see Methods section) that were ignored per trial per subject was  $1.5 \pm 1.896$  (0-8.13) out of approximately 90 per trial. No data were removed on account of large temporal or spatial gaps.

### **2.3.4 Modified Balance Test**

Introducing the helmet into the traditional balance test did not seem to affect subjects' ability to perform the test. The average sampling frequency across all trials and subjects was  $106.6 \pm 6.0$  Hz (63.6-113.6 Hz). The average time step was  $9.4 \pm 0.1$  ms (0.1-3649.7 ms,  $N=1,596,822$ ), with 0.14% beyond two standard deviations from the mean. Occasionally, the tip of one or both dowels exited the sensing volume, and facial features sometimes interfered with the measurement. These adverse events created jumps in time or space. As described above, we removed any differences between successive samples larger than 50 ms in time and 5 mm in space. The average number of time steps larger than 50 ms per trial across all subjects was 3.8

$\pm 4.8$  (0-25.4) and the average number of jumps in space larger than 5 mm was  $25.2 \pm 44.4$  (0-197.6). Consequently, an average of  $1060.0 \pm 1220.0$  ms (20.0-7350.0 ms) of data was removed from every 30-second trial across all subjects.

### 2.3.5 Modified Visually Guided Movement Test

Generally the modified visually guided movement test worked well. Subjects easily interacted with the GUI and the Leap tracked the fingertip fairly consistently. The average sampling frequency across all trials and subjects was  $107.5 \pm 3.4$  Hz (88.0-113.5) and the average time step was  $9.3 \pm 17.6$  ms (0.054-9604.0,  $N=4,713,255$ ) with 0.03% beyond two standard deviations from the mean. Even though subjects were instructed to keep their index finger extended and all other fingers balled, the Leap occasionally failed to recognize the index finger or mistook it for another finger, creating jumps in time or space. We removed any differences between successive samples greater than 50ms in time or 30mm in space (this task required rapid movements, so the space threshold was larger than in the balance test). We also removed any movements with less than 20 samples, which indicated in part that the subject left the target before the new target appeared. An average of  $7.0 \pm 7.0$  (0-36) of 240 movements were removed per subject.

**Table 2-2: Results for Each Modified Test**

Modified Test	Sampling Rate (Hz)	Outliers in dt (%)	Time steps (N)	Removed
Reaction Time	$108.9 \pm 6.4$	0.0019	1,275,785	0
Postural Tremor	$109.3 \pm 1.3$	0.81	1,311,027	0
Finger Tapping	$108.8 \pm 3.8$	0.94	1,129,909	0
Balance	$106.6 \pm 6.0$	0.14	1,596,822	3.5% (1.06/30s)
Visually Guided Movement	$107.5 \pm 3.4$	0.03	4,713,255	7/240 moves

## **2.4 Discussion**

Tens of millions of people live with some form of upper limb motor impairment [1-5]. The current methods of assessing a patient's impairments and tracking his/her progress during treatment are qualitative and lack the ability to detect subtle changes. As a result, it is difficult to assess movement deficits accurately, track improvements over time, and fine-tune treatment. The development of MMC sensors, however, has provided the opportunity to develop motor assessments that are accurate, quick, and inexpensive. The first step, however, is to adapt conventional tests to the capabilities of MMC sensors so the tests can be administered in a robust manner. The purpose of this study was to adapt five common conventional motor tests so they can be administered via MMC, and to evaluate the feasibility of implementing the modified tests using a particular MMC sensor.

### **2.4.1 Summary of Results**

By administering our modified tests to one hundred healthy subjects, we evaluated the ease-of-use of the tests and the performance of the Leap Motion sensor. Overall the results were encouraging, suggesting that the Leap Motion sensor is on the verge of being sufficiently robust for clinical use. The average sampling rate across all subjects for each modified test was greater than 106 Hz. This is contrary to the findings of both Guna and Tung, who reported sampling rates of  $39.0 \pm 12.8$  Hz and  $65.47 \pm 21.53$  Hz respectively [26, 27]. The high average sampling rates we experienced are high enough to avoid aliasing of kinematic measures. The vast majority of the power in human position data lies below 12 Hz: voluntary movement typically resides below 5 Hz, finger tapping rarely exceeds 7 Hz, and most tremors are below 12 Hz. Therefore, according to the Nyquist Theorem, a minimum sampling rate of only 24 samples/sec is required



to accurately measure such signals. According to the Leap Motion API documentation, the sampling frequency depends on a number of factors, including available computing resources [29], so a minimum hardware specification could be made to ensure sufficiently high sampling frequencies.

## **2.4.2 Limitations**

### ***Modified Reaction Time Test***

The Leap Motion sensor performed well during this task; with the sampling rate consistently above 100 Hz, the reaction time can be expressed with a resolution of  $\pm 5$  ms. However, subjects with debilitating movement deficits may find this task challenging for two reasons. First, subjects with significant arm weakness may find it difficult to keep the arm extended against gravity while they anticipate the stimulus, which appeared randomly after 3-8 s. Second, subjects are required to keep their hand in a 130mm-diameter circle while they anticipate the stimulus. Subjects with severe tremor may have trouble keeping their hand in this circle. Also, the reaction time is defined as the time it takes for the subject to exit a 30mm-diameter circle centered on the palm at the time of the stimulus. Significant tremor may involuntarily push the palm outside of this circle, resulting in an artificially fast reaction time. Other reaction time tests measure the time it takes subjects to move their finger. As our test measures the time to move the greater mass of the hand, the reaction times measured by our test may be slightly longer than those measured by other tests. Recently released software updates have improved the Leap Motion sensor's ability to track fingers, so a future version of this test could track the finger instead of the palm.

### ***Modified Postural Tremor Test***

The performance of the Leap Motion sensor during the postural tremor test shows promise, however the tests need to be developed further before they can be implemented in a clinical setting. In theory, the high average sampling frequency we experienced during the administration of the test is adequate to measure tremor within the tremor band without any aliasing. However, the Leap Motion sensor does not have the bandwidth to accurately measure tremor amplitude, as will be shown in Chapter 3—at least not without additional post-processing. In addition, the dynamic accuracy of the Leap Motion sensor limits the range of tremor amplitudes that can be measured. Tremors with amplitude below 1.2mm are below the sensor's dynamic accuracy [25]. Although our test protocol does not differ greatly from that of the traditional assessment, the measures that we used differ from the measures of the traditional test, which may limit the adoption of the test in a clinical setting. In the traditional test the clinician typically observes the maximum displacement of the outstretched fingers, whereas our test focused only on the movement of the palm because of the limited bandwidth of the Leap Motion sensor, which was substantially smaller for the fingers than the palm (Chapter 3). We expect that future improvements in MMC technology (in the Leap Motion sensor or another sensor) will result in larger bandwidths and enable accurate tracking of tremor in the fingers as well. The conventional test results in a single amplitude of tremor even though tremor may include multiple frequencies. Instead of specifying the amplitude at a single frequency, we presented the area under the power spectrum over the tremor band. Although this measure is not as intuitive as tremor amplitude, it provides a more robust estimate of tremor activity.

### ***Modified Finger Tapping Test***

The Leap Motion sensor shows potential to be used to administer the modified finger tapping test. Tapping rates can reach as high as 60 or 70 taps in 10 s [18], so the Nyquist frequency (10-12 taps/s) is well below the observed sampling rate (106 samples/s). However, the Leap Motion sensor does not have the bandwidth to accurately measure such oscillations, which is a significant limitation to this test, though there may be ways to reduce the effects this limited bandwidth (Chapter 3). Furthermore, there are aspects of the conventional test that are missing from the modified test, and their absence may have a significant effect on subjects' performance. For instance, in the conventional test, the tapping lever produced haptic and auditory feedback, which may help users tap more regularly or quickly. Additionally, the physical characteristics of the lever, which included resistance and a hard stop may affect the number of taps.

There are also some limitations with the modified finger tapping test which could cause significant challenges for individuals with upper limb motor impairments. First, the subject is required to hold his/her hand against gravity for 10 s at a time, repeated up to ten times for each arm. This could prove difficult for individuals with significant arm weakness. For patients with severe tremor, finger tremor may interfere with the assessment of their ability to voluntarily oscillate their finger. Although the same challenge may afflict the conventional test, its effect is likely smaller since the hand is resting on the table and pressing the mechanical lever takes additional force. Second, our test requires subjects to ball their fingers into a fist, which would prove difficult for some subjects with motor impairments. The newest API provides a more robust way of identifying specific fingers, so the test could be rewritten to reduce or even eliminate the requirement to make a tight fist. Lastly, in its current state, the modified test requires the proctor to watch for excessive movement of the hand, wrist, or arm, as such

movement is not possible in the conventional test. A more robust programming algorithm could be implemented to subtract non-finger movement before specifying the cursor position, reducing or eliminating the need for a proctor to watch for excessive non-finger movement.

### ***Modified Balance Test***

Although the sampling rate observed during the modified balance test was sufficiently high, there remain other issues that need to be resolved before this test is sufficiently robust to be administered in a clinical setting. First, positioning the tripod immediately in front of subjects may present a safety hazard to individuals who are unable to keep their balance. A stand with a boom arm could easily resolve this issue. Second, healthy subjects were generally able to maintain their balance enough to keep the dowels within the sensing volume of the Leap Motion sensor, but this would likely prove difficult for impaired patients. It is possible that this challenge could render the modified test unusable for some movement disorders. A potential solution could combine multiple, time-synchronized Leap Motion sensors to increase the effective sensing volume—the producers of the Leap Motion sensors have been said to be working on this. Third, it became apparent that the Leap Motion sensor would not recognize and track the dowels if the subjects leaned too far forward, presumably because of interference from features of the face and/or helmet. The dowels could be lengthened to reduce the likelihood of this error, but it may still be difficult for patients with significant balance impairment. Lastly, since it is impractical to place the dowels in the helmet with high accuracy, the position of the dowel tips relative to the location on the helmet immediately above the crown of the head needs to be measured for each modified helmet. These limitations suggest that this modified test is more suited for a MMC system that tracks the entire body (like the Kinect), but such full-body MMC systems would need to first become sufficiently accurate.

### ***Modified Visually Guided Movement Test***

The modified visually guided movement test has multiple limitations that need to be addressed before it is suitable for routine clinical testing. First, movements were limited to the 2D vertical plane of a computer screen, so motor deficits associated with depth perception may not be measurable. Furthermore, the amplitude of movements were limited both by the sensing volume of the Leap Motion sensor and by the size of the computer monitor used for visual feedback. The movements in our test were significantly smaller than those in the conventional test. Second, it became apparent during testing that the subject's forearm would occasionally block their view of the next target on the screen—specifically the bottom, right target was blocked by the right forearm when it appeared after a top left target, and the bottom, left target was blocked by the left forearm when it appeared after a top, right target. It is likely that this affected the kinematics of some movements because some subjects couldn't see the target on the screen until they moved their arm slightly from the previous target. Lastly, patients with motor impairments may have difficulty completing the modified test, which required subjects to extend their arm against gravity during the execution of 60 movements. This took healthy subjects approximately 110 s and is significantly more movements than included in the conventional test. Extending the arm against gravity for this duration may be difficult for patients with arm weakness. The number of movements was chosen to allow the assessment of movement variability. Variability is not usually measured in the conventional test and could be sacrificed in favor of a shorter test. In addition, the test required the speed of the fingertip to fall below 50mm/s within the target. This was intended to force the subject to come to an almost complete stop on the target while also allowing for some slight, unintentional movement of the

outstretched hand. However, patients with severe tremor may not be able to slow their hand below the 50mm/s threshold inside the target.

### **2.4.3 Conclusion**

The results of this study suggest that MMC sensors in general, and the Leap Motion sensor in particular, hold promise for clinical settings. They may allow the current rating scales, which are subjective and qualitative, to be replaced with objective, quantitative motor assessments (QMA). The sampling rate of the Leap Motion sensor observed during our tests was sufficiently high and regular to capture typical measures of human movement. Likewise, the dynamic accuracy measured previously [25] is sufficient for most measures (though certainly not all measures). Although the bandwidth of the Leap Motion sensor is too low to accurately measure the amplitude of tremor, additional post-processing can be used to estimate the actual tremor amplitude (Chapter 3). While the results of this study are encouraging, we have listed a number of limitations that must be overcome before the tests are sufficiently robust for clinical use.

### **3 FREQUENCY RESPONSE OF THE LEAP MOTION SENSOR**

#### **3.1 Introduction**

Although tremor is the most common movement impairment, the current methods of assessing tremor in clinical settings rely on subjective, qualitative rating scales. Tremor is the involuntary oscillatory movement of a body part; most tremors occur in the hands, which can disrupt many activities of daily life and cause social anxiety. Although there are a variety of disorders that result in tremor, the frequency of most tremors lies between 3 and 12 Hz [15, 30], while tremor amplitude varies greatly between and even within subjects. The current method of assessing tremor in a clinical setting involves visually observing a patient's movements as he/she completes different tasks and rating the observed tremor on a rough numerical scale. Because this method relies on visual observation and is inherently subjective, it lacks the resolution, accuracy, and repeatability needed to track subtle changes over time and make early corrective adjustments in treatment. Although quantitative tools, such as optoelectronic or electromagnetic motion capture systems, inertial measurement units, and robotic systems, have been used to accurately measure tremor in research settings, they are far too time-intensive, labor-intensive, and/or expensive for routine clinical use. A better method of assessing tremor needs to be explored.

Markerless motion capture (MMC) technology introduces an opportunity to improve assessment of tremor in a clinical setting. These types of sensors are inexpensive, require no setup, and are becoming more accurate and reliable. The Leap Motion sensor is one such device; it is a small, infrared optical sensor that costs approximately \$70. It is designed as a plug-in controller to any Mac or personal computer and allows the user to interface with a computer using hand motions. Its onboard, proprietary software is designed to recognize and track the positions and orientations of features of the human hand such as fingertips, finger segments, joints, and the center of the palm. Additionally, it is programmed to recognize and track the tip and orientation of pointer-like tools such as a pencil. Although actually cone-shaped, its sensing volume is roughly 2ft by 2ft by 2ft, with greatest accuracy approximately 25-250 mm above the sensor [26]. Previous studies have reported on the static and dynamic accuracy of the Leap Motion sensor. Weichert et al. reported that the accuracy was  $<0.2$  mm in static situations and  $<1.2$  mm in dynamic situations [25]. Similarly, Guna et al. showed the accuracy to be  $<0.5$  mm in static situations [26]. More recently, Tung et al. measured the accuracy in tracking human subjects' hands as they performed a pointing task. They reported significantly poorer accuracy ( $17.30 \pm 9.56$  mm) in quasi-static situations than the other two studies as well as lower sampling rates [27]. However, they placed an infrared light emitting diode at the end of the fingertip, which could have lowered the sensor's performance.

Whereas the accuracy of the Leap Motion sensor in static and dynamic situations has been characterized, its frequency response—critically important in measuring hand tremor—has not been measured. The purpose of this study was to characterize the frequency response of the Leap Motion sensor over the frequencies and amplitudes relevant to human tremor. To do this, we drove a mannequin hand at various frequencies and amplitudes and recorded finger and palm



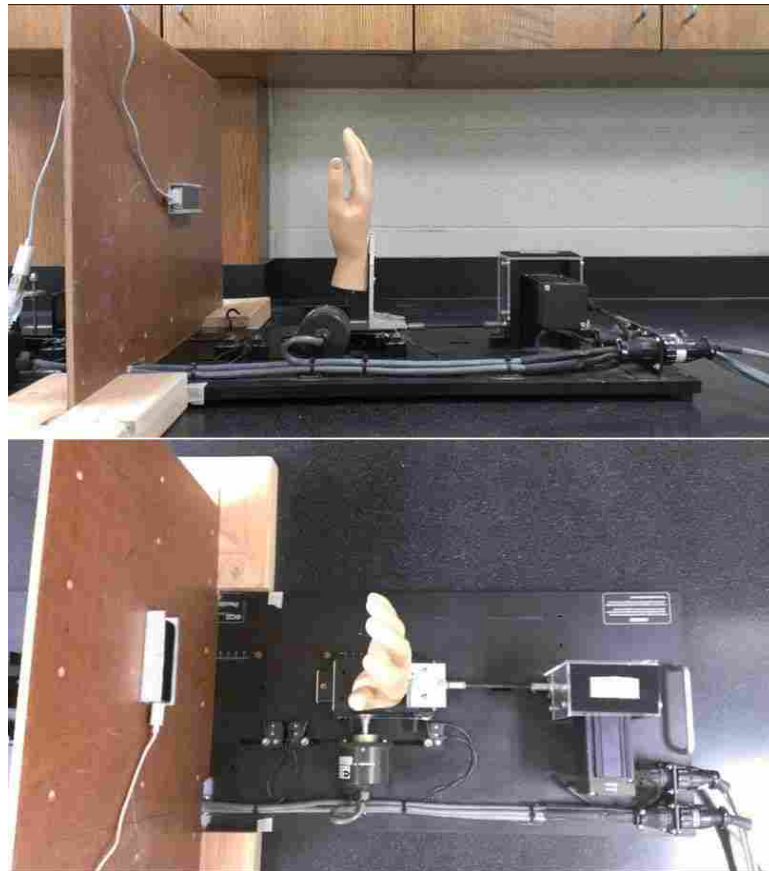
position simultaneously with the Leap Motion sensor and a highly accurate encoder. We present the sensor's magnitude ratio relating the actual amplitude of the tremor measured by the encoder (input) to the amplitude recorded by the Leap Motion sensor (output). We found the bandwidth of the Leap Motion sensor to be too small to accurately measure the amplitude of human tremor. To overcome this limitation, we also present a method to estimate the actual tremor amplitude from the attenuated tremor amplitude measured by the Leap Motion sensor.

## **3.2 Methods**

### **3.2.1 Experimental Setup**

A desktop computer running a Windows 7 operating system with an Intel® Core™ 2 Duo E8500 processor, 4 GB of RAM, and an NVIDIA Quadro NVS 290 graphics card was used to run a custom LabVIEW program. This program controlled a Model 210 Rectilinear Plant System (Educational Control Products, Bell Canyon, CA) consisting of a DC brushless motor that drove a sled along a linear bearing. The system also included a CP-850-HHC high-resolution encoder (Computer Optical Products, Chatsworth, CA) that recorded the position of the sled at 160 counts per mm. A stiff mannequin hand was rigidly attached to the sled and was oscillated back and forth by the motor along the linear bearing. A Leap Motion sensor was mounted to a vertical surface so that the cameras were oriented horizontally and the hand was centered in its field of view (Figure 3-1). At the near and far limits of the linear bearing, the distance from the Leap Motion sensor to the hand was 130mm and 200mm, respectively. The custom program recorded the time and the position as reported by the two sensors (the encoder and the Leap Motion sensor) at approximately 100 Hz. A single value was reported by the encoder, while the

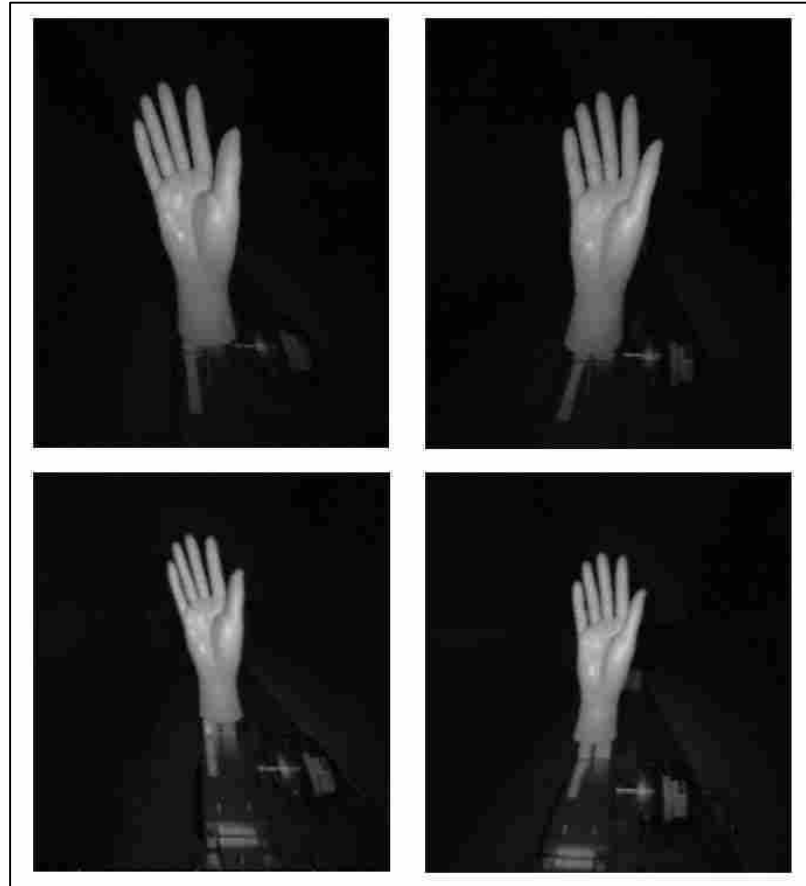
three-dimensional positions of the palm and each fingertip were reported by the Leap Motion sensor. Version 2.3.1 + 31549 of the Leap Motion software was used during the test.



**Figure 3-1: Experimental Setup**

We took a number of steps to maximize the sensor’s ability to recognize and track the hand. Immediately before the test was conducted, the Leap Motion sensor was calibrated using the Leap Motion sensor calibration software until a score of 95 out of 100 was achieved. We reduced the presence of objects in the environment surrounding the hand to limit potential interference. Additionally, a trial run was conducted to ensure that the sensor’s “hand confidence value” (HCV) was equal to unity. The HCV ( $0 \leq \text{HCV} \leq 1$ ) is a measure of “how well the

internal hand model fits the observed data” [29]. These steps resulted in a clearly discernable hand (Figure 3-2).

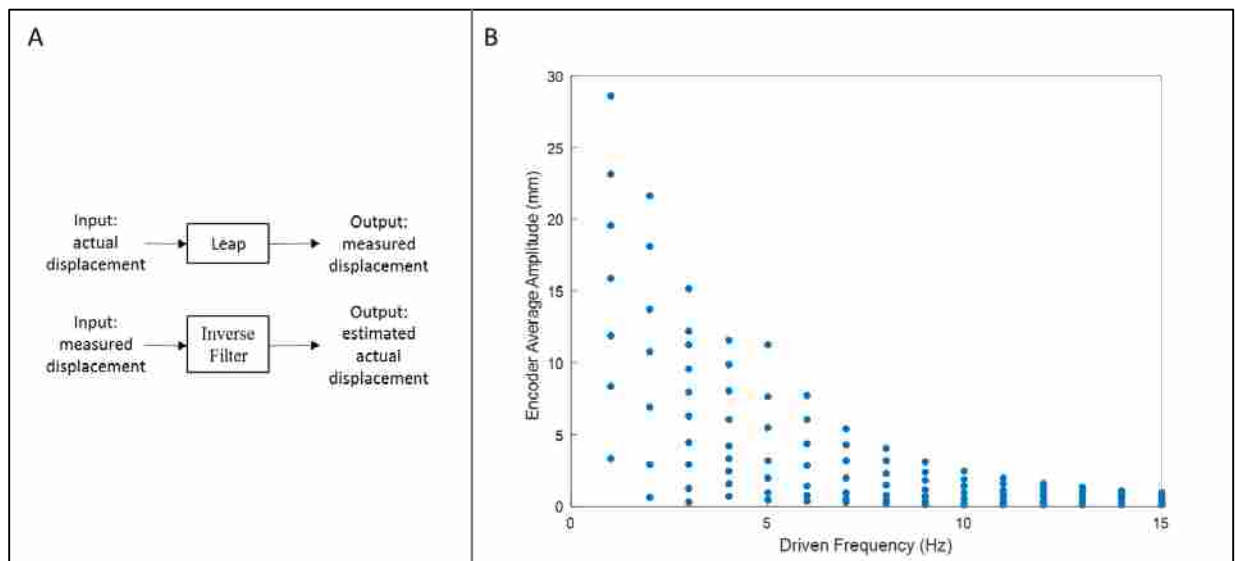


**Figure 3-2: Leap Motion Sensor Images.** Top Row: Hand at the near limit. Bottom Row: Hand at the far limit

### 3.2.2 Experimental Protocol

We characterized the Leap Motion sensor as an input-output system: the input was the actual displacement of the mannequin hand as measured by the encoder, and the output was the displacement of the hand recorded by the Leap Motion sensor (Figure 3-3A). To determine this input-output relationship, we drove the mannequin hand sinusoidally at a variety of frequencies

and amplitudes, resulting in 107 trials of distinct combinations of frequency and amplitude. The driving frequency and amplitude were kept constant during each 40-second trial. We drove the hand at integer frequencies between 1 and 15 Hz, with 6-10 different amplitudes per frequency. Due to the physical limitations of the motor, the range of available amplitudes decreased as the driving frequency increased (Figure 3-3B). Nevertheless, within this envelope we selected a wide distribution of amplitudes, ranging from 0.08 mm to 28.3 mm (mean  $\pm$  standard deviation:  $3.48 \pm 5.07$  mm).



**Figure 3-3: Sensor Input-Output Relationship and Inputs Included in Experiment.** (A) Top: Like any sensor, the Leap Motion sensor is a filter that can be characterized using the magnitude ratio, defined as the ratio of the amplitude of the output (measured by the Leap Motion sensor) over the amplitude of the input (actual amplitude as measured by the encoder). Bottom: The actual amplitude (as measured by the encoder) can be estimated from the amplitude measured by inverting the filter of the Leap Motion sensor. (B) Driving frequencies and input amplitudes tested in our experiment.

### 3.2.3 Data Analysis

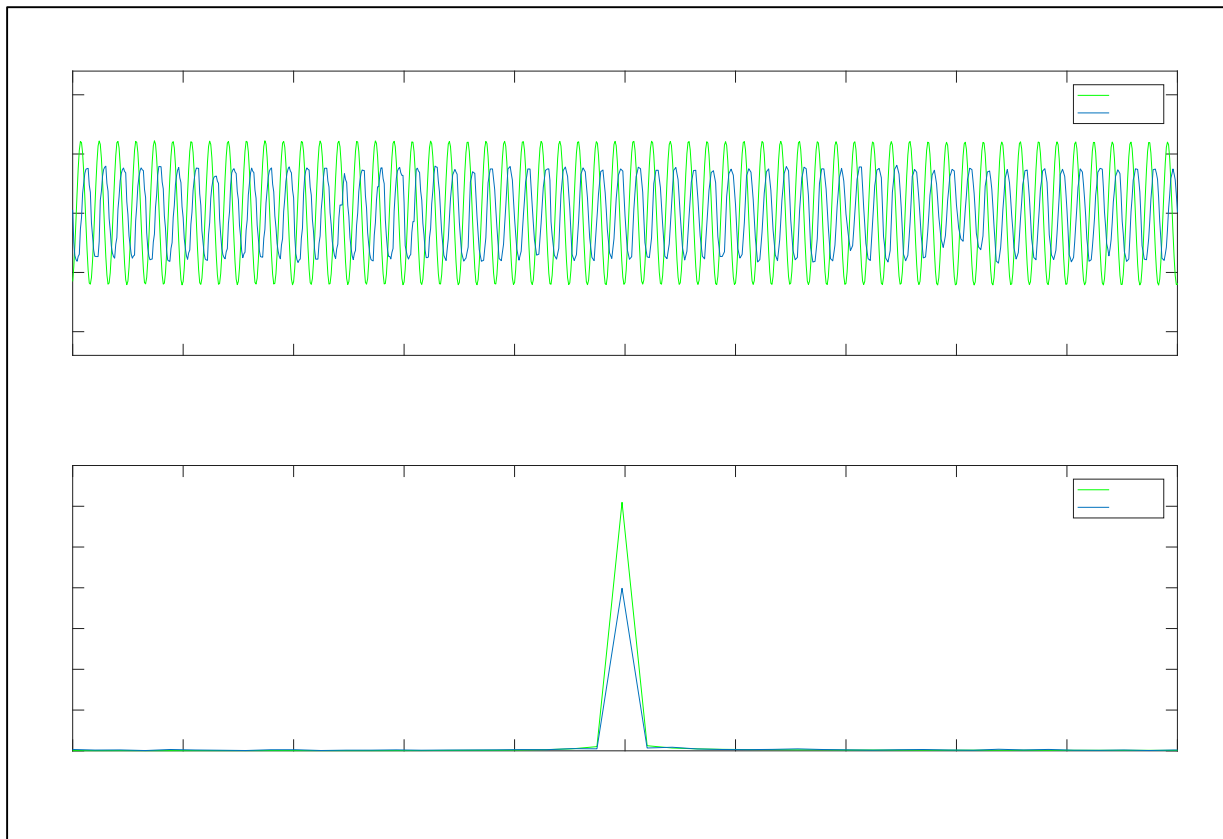
The input-output relationship of the Leap Motion sensor was characterized as the magnitude ratio, defined as the ratio of the amplitude of the output (as recorded by the Leap

Motion sensor) over the amplitude of the input (recorded by the encoder). The amplitudes of the input and output were calculated as follows. First, the data from the Leap Motion sensor and encoder were interpolated at a constant rate of 100 samples/sec with a piecewise cubic hermite interpolating polynomial (pchip) using Matlab's interp1 function. Second, using the three-dimensional position data reported by the Leap Motion sensor, we calculated the distance from the Leap Motion sensor to the palm and fingertips. The two time-synchronized signals are shown in Figure 3-4A. Third, we transformed both signals into the frequency domain using the Fast Fourier Transform (Matlab's fft function) as seen in Figure 3-4B. Since each trial lasted 40s, the transformed signals had a frequency resolution of  $1/(40s) = 0.025\text{Hz}$ . Fourth, the amplitude of the encoder signal was identified as the maximum of the transformed encoder signal, and the amplitude of the Leap Motion signal was determined as the magnitude of the transformed Leap Motion signal at the frequency of the maximum of the transformed encoder signal. These amplitudes were used to calculate the magnitude ratio at a given combination of frequency and amplitude, repeated for all 107 combinations. This resulted in 107 magnitude ratio values at different frequencies and input amplitudes.

For a linear system, the magnitude ratio is independent of input amplitude. If the Leap Motion sensor is linear, the magnitude ratios measured for different input amplitudes should be the same (at the same frequency). However, for general (non-linear) systems, the magnitude ratio can depend on input amplitude, so we present the magnitude ratios measured at each frequency as a function of input amplitude (the actual amplitude as measured by the encoder). We also present the magnitude ratios as a function of frequency (standard format) for different amplitudes. However, the amplitudes differed for every test, so we sorted the magnitude ratios at a given frequency into bins of similar inputs amplitudes and averaged the magnitude ratios

within each bin. The range of input amplitudes in the various bins were 0-0.13mm, 0.13-0.2mm, 0.2-0.45mm, 0.45-0.8mm, 0.8-1.5mm, 1.5-3mm, 3-5mm, 5-11mm, 11-30mm.

We also determined the bandwidth of the Leap Motion sensor. The Leap Motion sensor acts as a low-pass filter, resulting in decreasing magnitude ratios with increasing driving frequencies. Since the input and output signals involved the same variable (position), we defined the bandwidth as the range of frequencies over which the magnitude ratio was above  $1/\sqrt{2}=0.707$ .



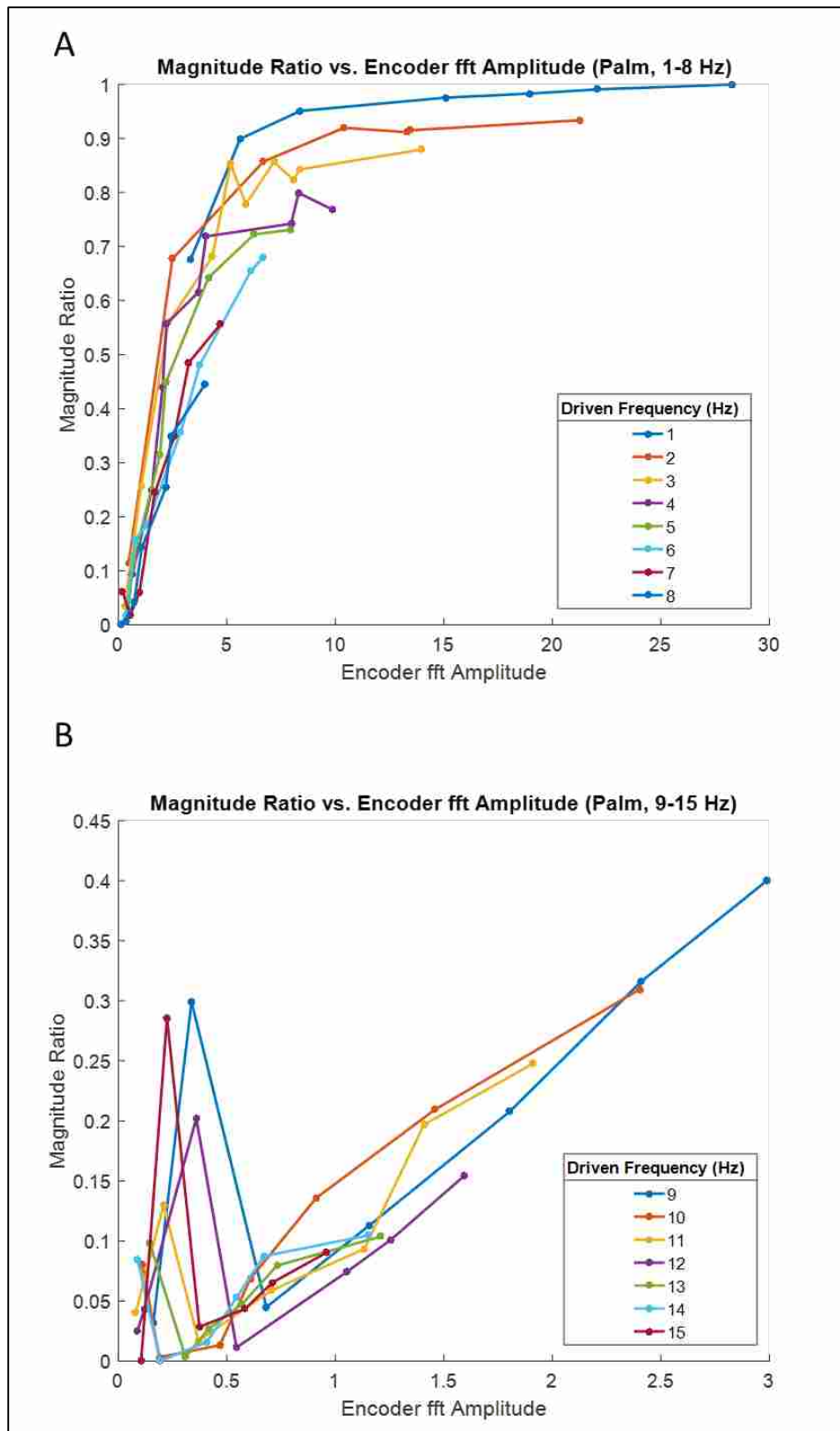
**Figure 3-4: Palm Position Measured by Encoder and Leap Motion Sensor.** (A) Position signals from the Leap Motion sensor and encoder in the time domain at 6 Hz. (B) Same signals in the frequency domain.

### 3.3 Results

Over the range of frequencies and amplitudes tested (1-15 Hz and 0.08-28.3 mm), the magnitude ratios for the palm position varied between 0.999 and 0.0001. Although the magnitude ratio was roughly constant for input amplitudes above approximately 10mm, the magnitude ratio at smaller input amplitudes depended strongly on input amplitude (Figure 3-5A), indicating that the Leap Motion sensor has a non-linear input-output relationship. The magnitude ratio generally decreased as the input amplitude decreased, though not for very small input amplitudes (<1 mm) and driving frequencies above 8 Hz, where decreasing input amplitude sometimes increased the magnitude ratio (Figure 3-5B).

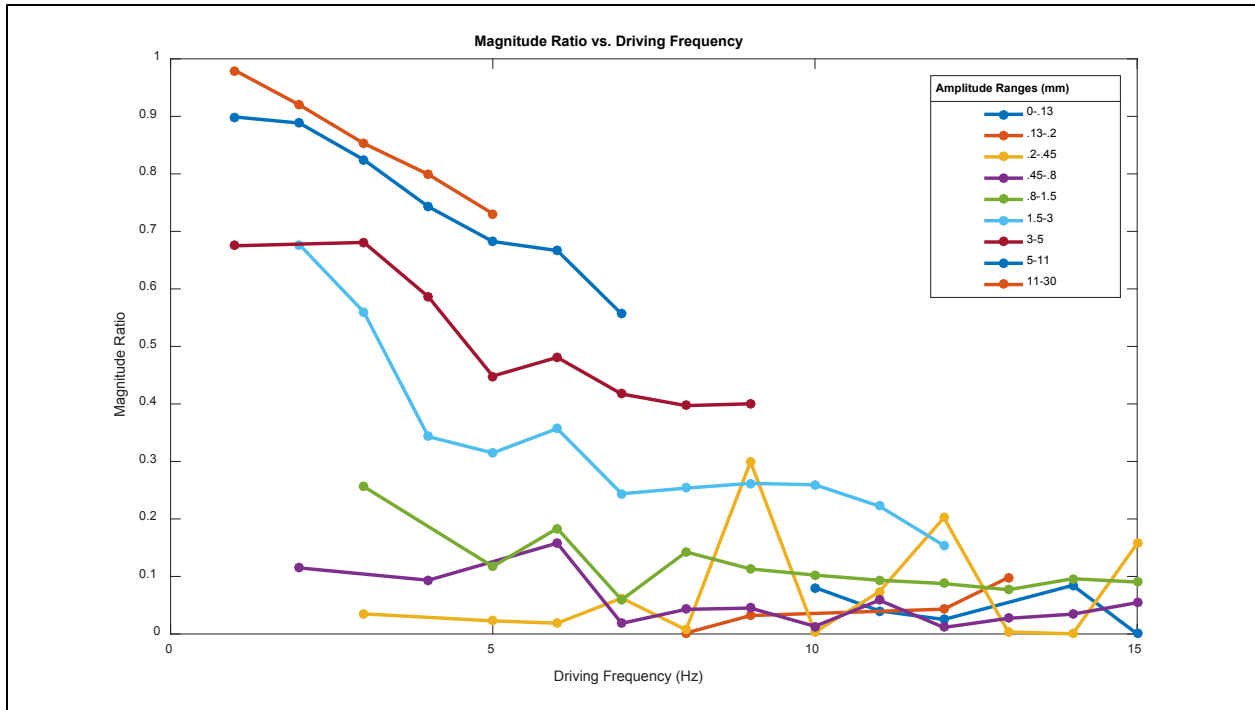
For input amplitudes above 1.5mm, the magnitude ratio for the palm position generally decreased with increasing driving frequency, indicating that the Leap Motion sensor acts as a low-pass filter (Figure 3-6). The bandwidth was approximately 1.7 Hz for input amplitudes of 1.5-3 mm, 2.7 Hz for 3-5 mm, 4.7 Hz for 5-11 mm, and 5.5 Hz for 11-30 mm. For input amplitudes below 1.5 mm, the magnitude ratios were below 0.3 for all driving frequencies tested and did not show clear trends with frequency.

The magnitude ratios calculated from the position of the finger tips followed similar trends but were substantially lower than the magnitude ratios calculated from the position of the palm (Appendix). For example, whereas the palm position magnitude ratio for a driving frequency of 1 Hz and input amplitude of 10 mm was 0.95, the fingertip position magnitude ratios at the same driving frequency and input amplitude were only 0.63, 0.50, 0.50, 0.50, and 0.47 for the thumb, pointer, middle, ring, and pinky finger, respectively. At frequencies above 8 Hz, the magnitude ratios calculated from the positions of the fingertips were more than an order of magnitude smaller than the corresponding magnitude ratios calculated from the palm position.



**Figure 3-5: Palm Position Magnitude Ratio vs. Input Amplitude for Driving Frequencies.**  
 (A) Driving frequencies of 1-8Hz. (B) Driving frequencies of 9-15Hz





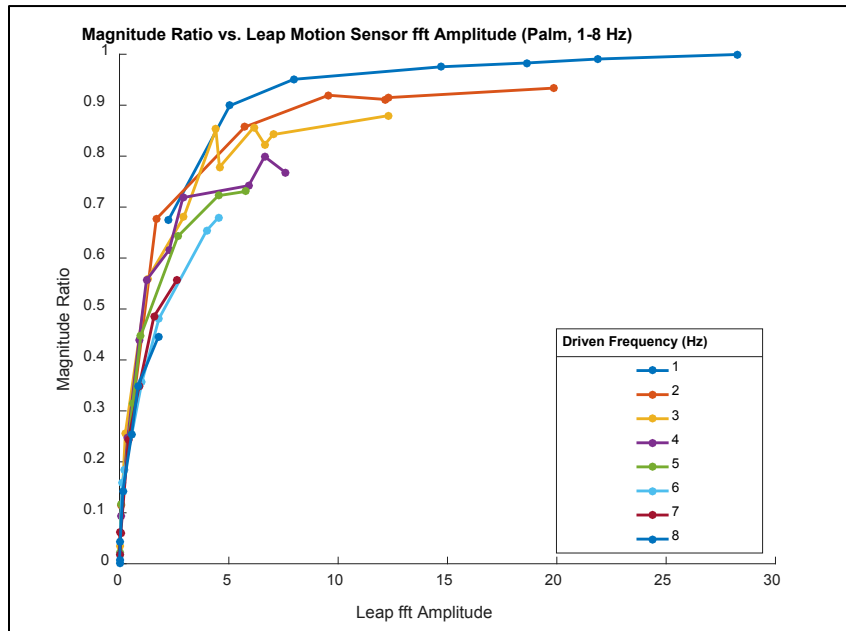
**Figure 3-6: Palm Position Magnitude Ratio vs. Driving Frequency**

### 3.4 Discussion

Although various types of tremor affect tens of millions of people, the current method for assessing tremor in clinical settings rely on subjective, qualitative rating scales, limiting health care providers' ability to accurately assess tremor and provide optimized care. MMC sensors introduce an opportunity to improve the clinical assessment of tremor. The purpose of this study was to characterize the frequency response of a particular MMC sensor (Leap Motion) and evaluate its ability to accurately measure hand tremor.

The Leap Motion sensor acts as a low-pass filter: the tremor amplitude measured by the Leap Motion sensor (output) is generally less than the actual amplitude of the tremor (input), and the magnitude ratio (output amplitude over input amplitude) decreases with increasing frequency. In addition, the magnitude ratio depends strongly on input amplitude; it increases with increasing input amplitude, indicating a non-linear input-output relationship. The bandwidth

for amplitudes above 1.5 mm varied between 1.7 and 5.5 Hz (larger bandwidth for larger input amplitude). Thus the bandwidth of the Leap Motion sensor is below or in the tremor band (3-12 Hz), indicating that the Leap Motion sensor is not suitable for measuring the amplitude of human tremor. This limited bandwidth could be due to hardware or software, but considering the bandwidth of other modern cameras (>5 Hz), we assume the bandwidth is due primarily to the Leap Motion software. Despite the limited bandwidth, the frequency response presented here can be used to estimate the actual tremor amplitude. Given the frequency and amplitude of tremor measured by the Leap Motion sensor, the actual tremor amplitude can be estimated by dividing the measured tremor amplitude by the magnitude ratio at that frequency and output amplitude Figure 3-7. The use of such a method is only recommended for input amplitudes greater than 1.5 mm, below which magnitude ratios become irregular (Figure 3-5B). This result is in line with the study by Weichert et al., who reported a dynamic accuracy of 1.2 mm [25].



**Figure 3-7: Palm Position Magnitude Ratio vs. Output Amplitude for Driving Frequencies.**

### 3.4.1 Limitations

The motor of the Rectilinear Plant System was unable to generate large-amplitude oscillations at high frequencies, so our characterization of the frequency response of the Leap Motion sensor was limited to the range shown in Figure 3-3B. That said, most patients' tremor is not both high-frequency and high-amplitude, so the range tested presumably covers the most common types of tremor.

We used a mannequin hand to allow the input to have a constant frequency and amplitude, which would not have been possible in vivo. The mannequin hand was similar to a real hand in terms of size, shape, color, texture, and posture. Consequently, the Leap Motion sensor algorithms, which are specialized for recognizing real hands, had no trouble sensing the mannequin hand. Nevertheless, the properties of the mannequin hand were not identical to those of real human hands, and it is possible that the frequency response of the Leap Motion sensor would be different for a human hand.

During the roughly 40 seconds of each test, the mannequin hand oscillated with nearly constant frequency and amplitude. Real human tremor is not constant in frequency or amplitude over such durations. In addition, measurements of tremor in clinical settings may be limited to shorter durations. Variable tremor frequency and decreased sampling duration would result in peaks in the frequency domain that are less well defined, which would likely reduce the accuracy in estimating the actual amplitude of the tremor using the algorithm presented above.

Because we took great care to maximize the accuracy of the Leap Motion sensor, the results presented here can be considered a best-case scenario. The Leap Motion sensor is most sensitive when the hand is centered above the sensor at a distance of less than 250mm [26]. In our experiment, the hand was positioned directly in front of the Leap Motion sensor and moved

only in the z-direction, with distances from the sensor between 130mm and 200mm. Interference from other infrared sources and visible light can affect the input. We took care to create an environment that limited such interference by performing our tests under fluorescent lighting with no natural light. We also minimized the presence of objects surrounding the hand. Finally, we calibrated the Leap Motion sensor immediately before the experiment. It is unlikely that all these precautions would be followed in clinical settings, which would like have an adverse effect on the accuracy of the measurements or the bandwidth of the sensor.

### **3.4.2 Conclusion**

Without additional processing, the Leap Motion sensor is unable to accurately measure the amplitude of hand tremor. However, using the magnitude ratios presented here, it is possible to estimate the actual tremor amplitude from the measured amplitude. We recommend this procedure for actual tremor amplitudes above 1.2 mm, which is the dynamic accuracy of the Leap Motion sensor [25]. We conclude that, with this additional processing, this MMC sensor is capable of measuring sufficiently large tremors with reasonable accuracy. As MMC systems continue to improve, we expect their ability to accurately measure tremor to increase as well.

## 4 CONCLUSION

The purpose of this study was to lay some of the foundation necessary for widespread use of markerless motion capture (MMC) sensors to evaluate motor deficits in a clinical setting. More specifically, we adapted five of the most common clinical motor tests so they can be administered via MMC and evaluated the feasibility of administering these tests using a particular MMC sensor (Leap Motion sensor).

We administered the five modified tests, termed quantitative motor assessments (QMA), to 100 healthy subjects. After an initial calibration of the Leap Motion sensor to the computer screen, there was little additional setup required, and the battery of tests flowed rather quickly. The subjects understood and followed the instructions easily. Some subjects initially had trouble mapping their movements to the feedback on the computer screen, but this was quickly learned, with few problems thereafter.

Generally, the Leap Motion sensor performed well during the administration of the QMA. It maintained an average sampling frequency above 106 Hz across all 100 healthy subjects and all five tests. However, the Leap Motion sensor occasionally recorded data artifacts or no data at all. These artifacts were usually due to occlusion of one finger by another, failure to differentiate neighboring fingers that were too close together, confusion of one finger for another, or the hand being too close to, or outside, the limits of the sensing volume of the Leap

Motion sensor. The majority of these artifacts can be eliminated with refined protocols and more robust algorithms, including programs to warn the user when errors occur.

The modified tests included a test to measure tremor, so evaluating the feasibility of using the Leap Motion sensor to administer these tests included a characterization of the sensor's ability to measure tremor. More specifically, we oscillated a mannequin hand at various amplitudes and frequencies and compared the measurements recorded by the Leap Motion sensor to those recorded by a highly accurate encoder. We found that the bandwidth of Leap Motion sensor is relatively small, the cut-off frequency being below or in tremor band, resulting in measured tremor amplitudes that are smaller than the actual tremor amplitudes. However, it is possible to estimate the actual tremor amplitude using the measured amplitude and the magnitude ratio characterized in this study.

Based on these observations, we conclude that MMC sensors are on the verge of becoming suitable for routine clinical use, but more work is necessary to further improve the motor tests before they can be administered via MMC with sufficient robustness for clinical settings. Our studies indicate that administering QMA via MMC shows potential, warranting additional work to improve both the MMC sensors and the QMA.

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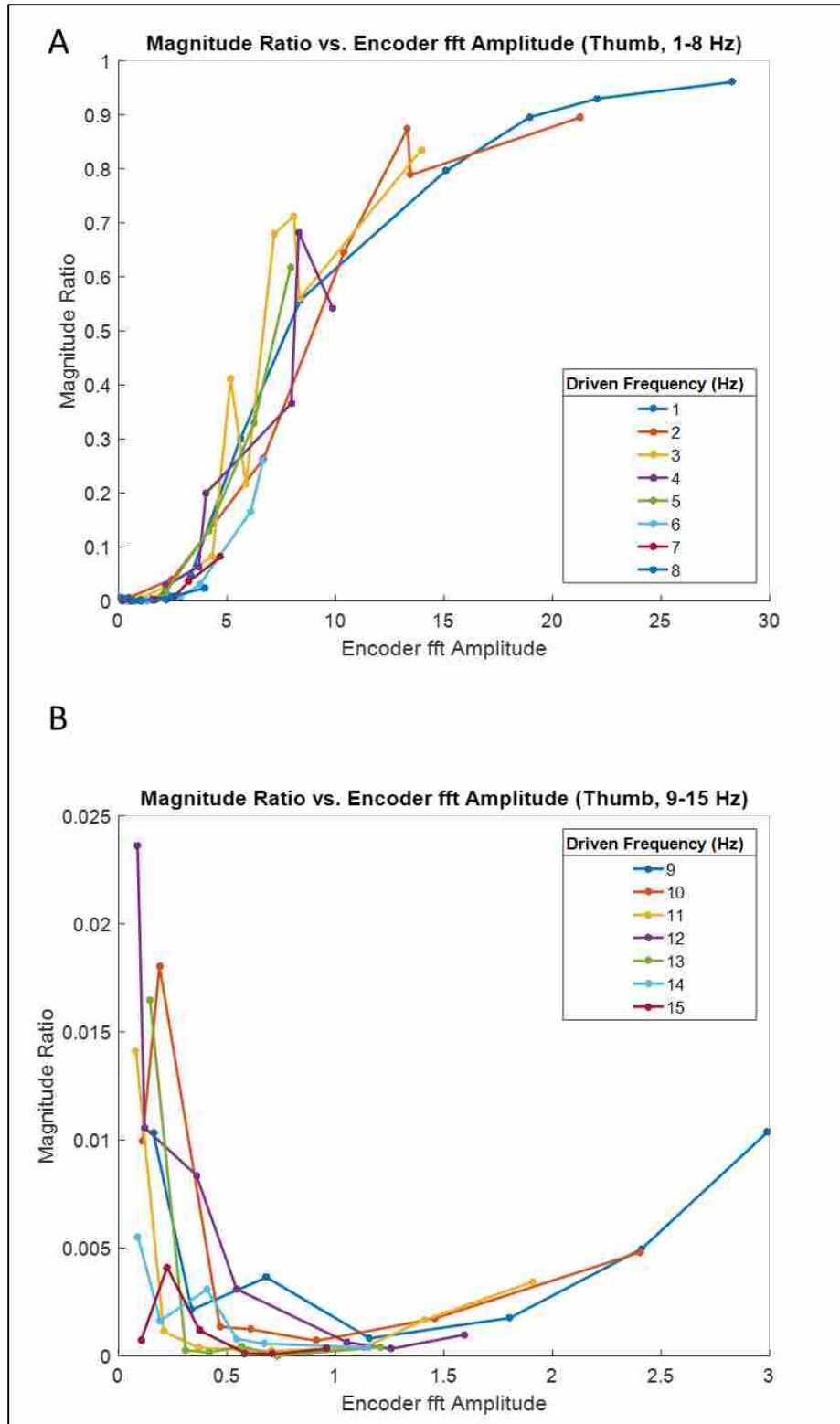


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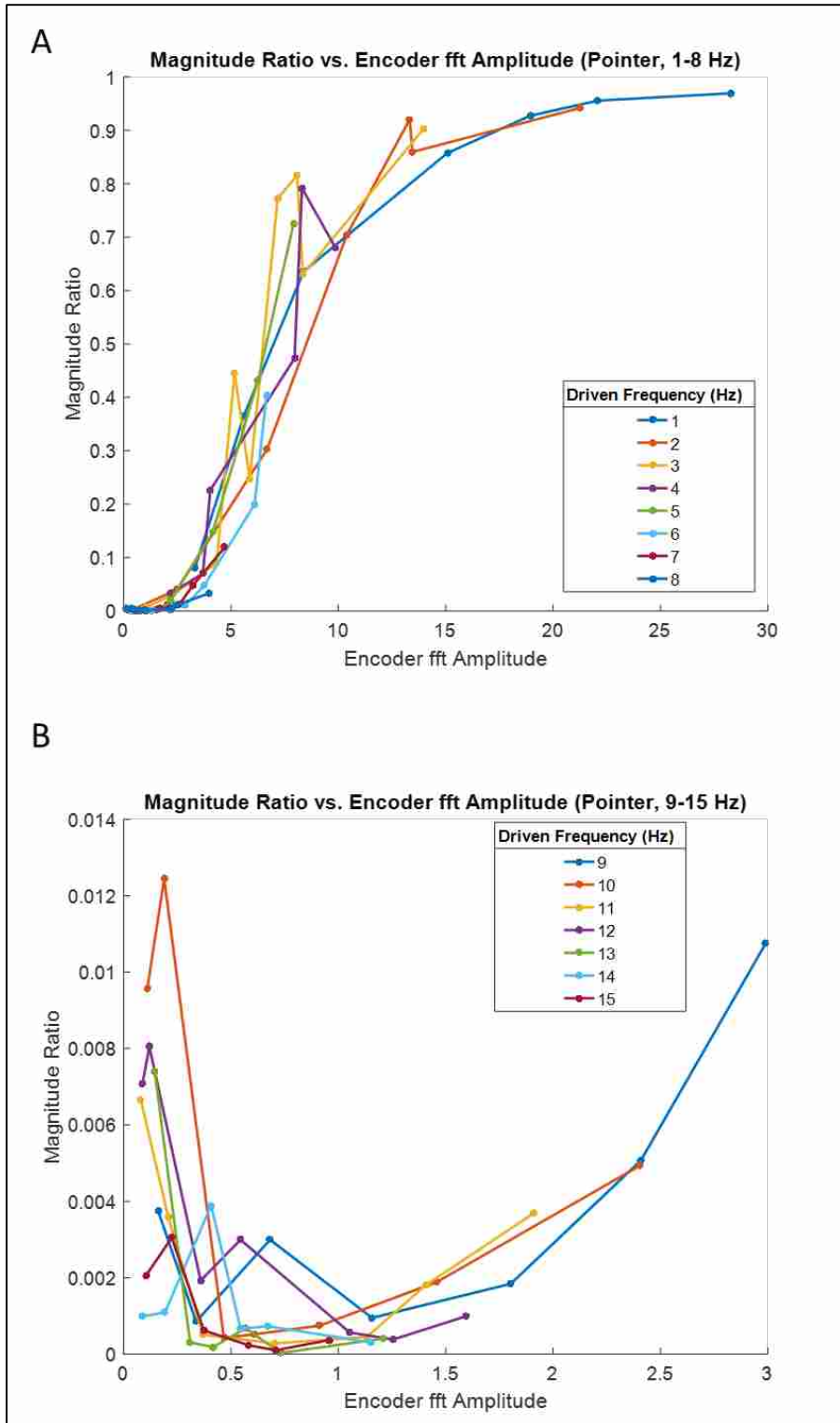


## **APPENDIX A.        MAGNITUDE RATIO VS. INPUT AMPLITUDE FOR FINGERS**

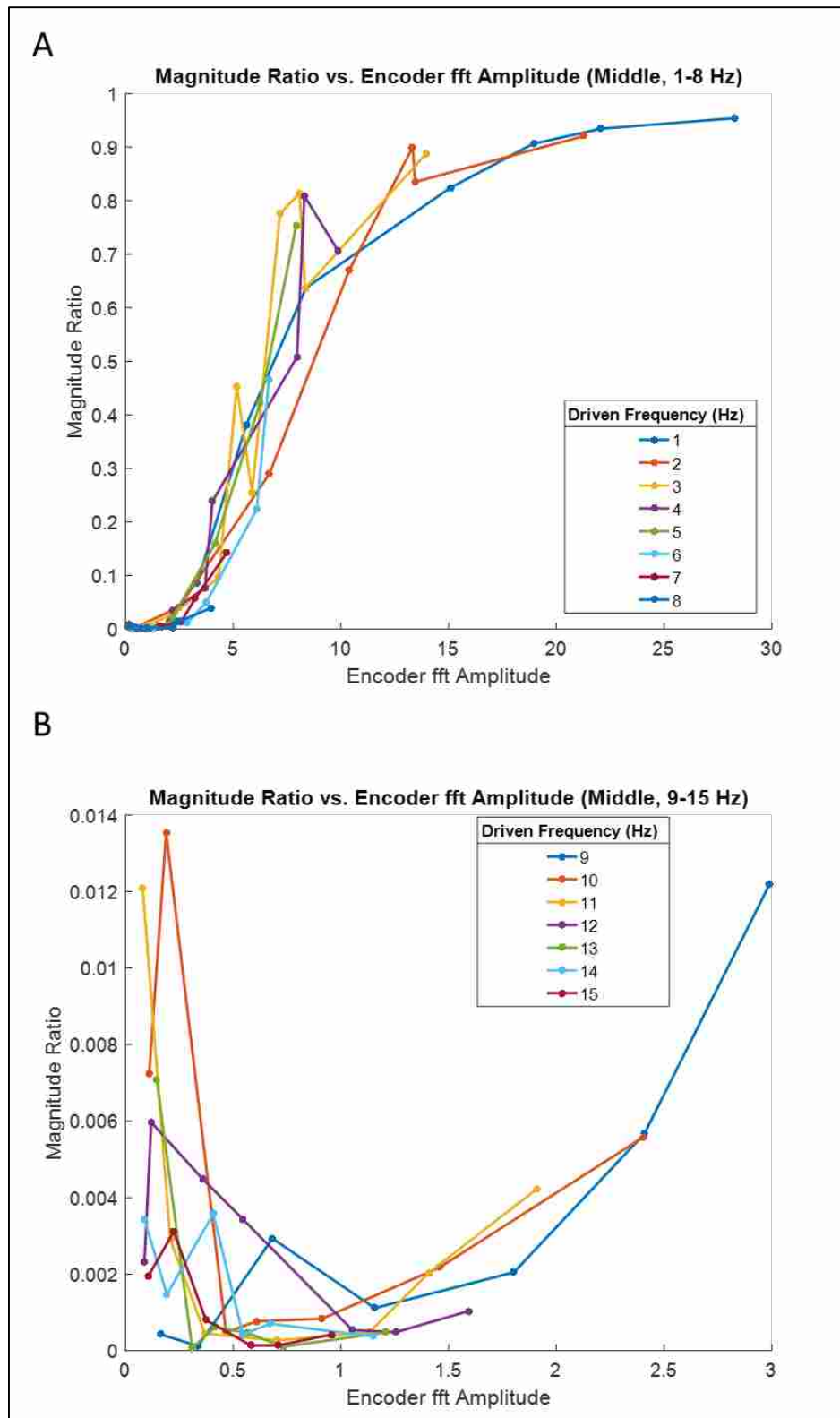
As mentioned in section 3.3, the magnitude ratios calculated from the position of the finger tips were lower than those for the palm. Here we present the magnitude ratios for all the fingers.



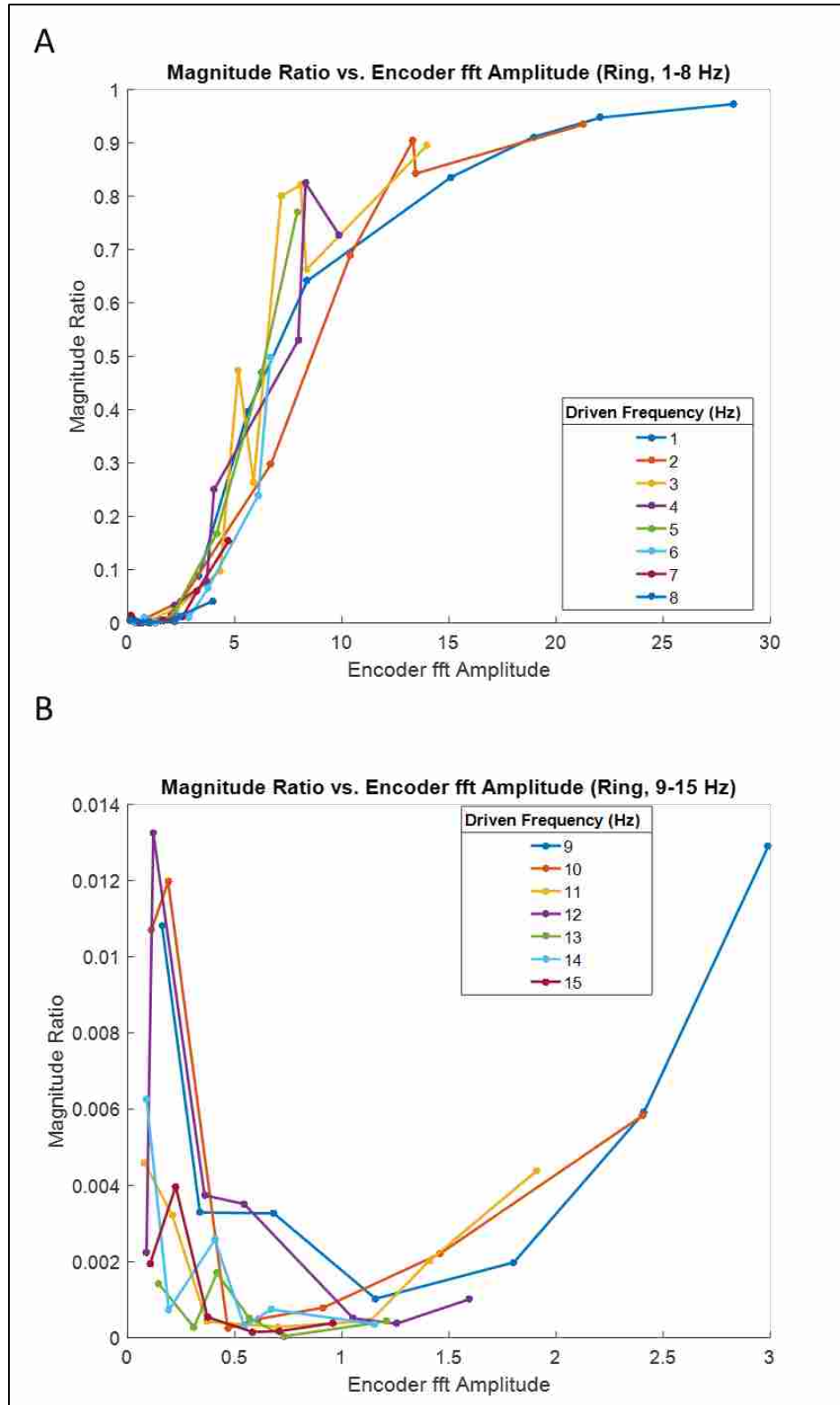
**Figure A.1: Magnitude Ratio vs. Encoder fft Amplitude for the Thumb. (A) Driving Frequencies 1-8 Hz. (B) Driving Frequencies 9-15 Hz**



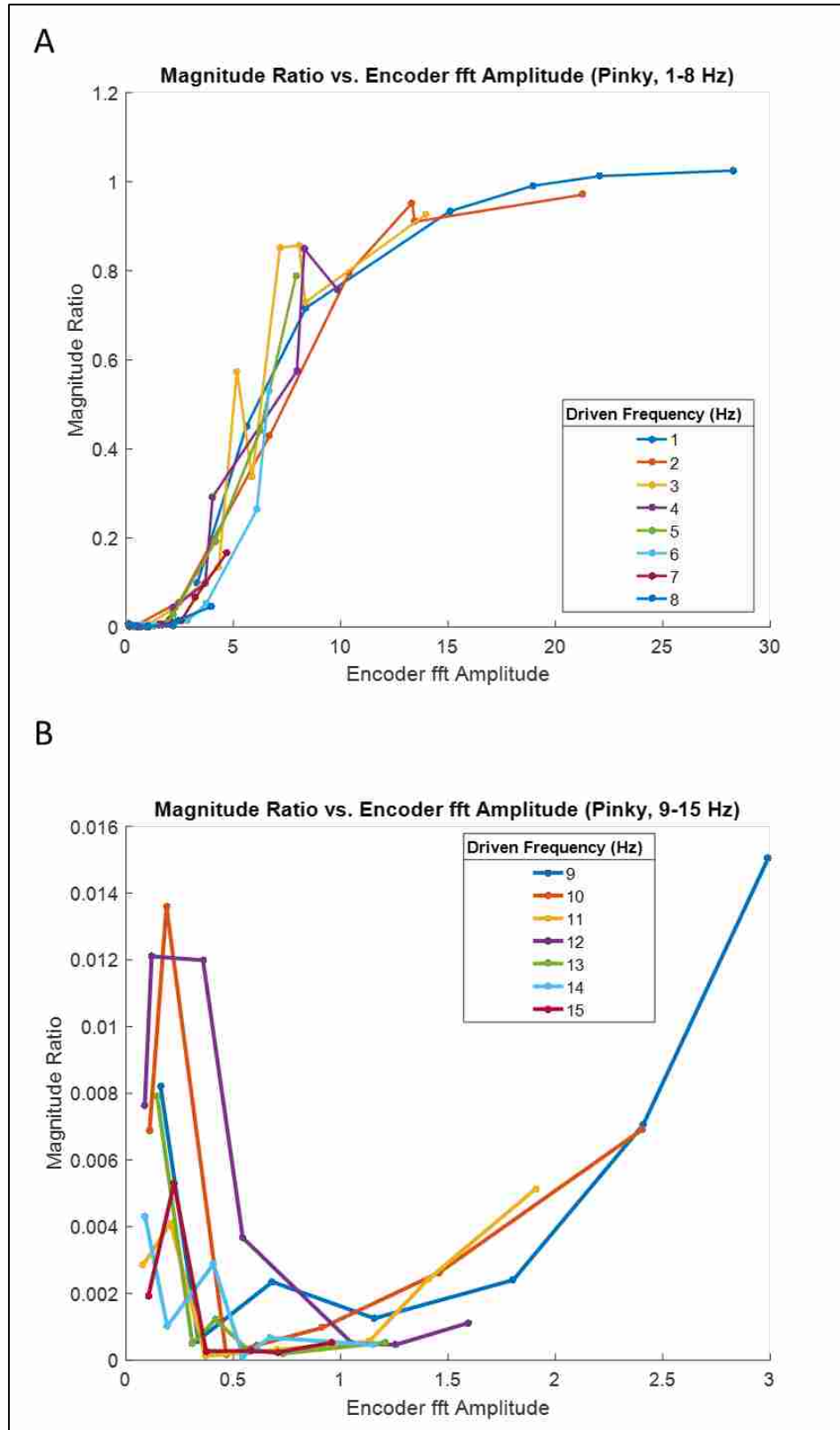
**Figure A.2: Magnitude Ratio vs. Encoder fft Amplitude for the Pointer. (A) Driving Frequencies 1-8 Hz. (B) Driving Frequencies 9-15 Hz**



**Figure A.3: Magnitude Ratio vs. Encoder fft Amplitude for the Middle. (A) Driving Frequencies 1-8 Hz. (B) Driving Frequencies 9-15 Hz**



**Figure A.4: Magnitude Ratio vs. Encoder fft Amplitude for the Ring. (A) Driving Frequencies 1-8 Hz. (B) Driving Frequencies 9-15 Hz**



**Figure A.5: Magnitude Ratio vs. Encoder fft Amplitude for the Pinky.** (A) Driving Frequencies 1-8 Hz. (B) Driving Frequencies 9-15 Hz