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The rehabilitation of human bimanual coordination

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The Rehabilitation of Human Bimanual Coordination

by

Peter Stanley Lum

DISSERTATION

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San Francisco

[redacted]

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I would like to thank Lawrence Stark for first sparking my interest in motor control and initiating the long and arduous journey which has culminated in this dissertation. I am very fortunate to have worked closely with Dave Reinkensmeyer and Steve Lehman. Without Steve's guidance, this dissertation would not have been possible. Steve is always encouraging and knows how to laugh when something goes awry, because he has everything in the proper perspective. It is a running joke that Dave and I share frontal lobes. Dave and I complement each other perfectly, and challenge each other academically. Working with Dave has resulted in the most productive period in my life.

I would also like to thank Elise Burmeister and Margot Damaser for their encouraging words and cheerful company which have made the lab an ideal workplace. And special thanks to Andrea Nasser for donating her time and effort in recruiting stroke patients for our studies and for her limitless enthusiasm.

Abstract

The Rehabilitation of Human Bimanual Coordination

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Doctor of Philosophy in Bioengineering

University of California

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Many activities of daily living (ADLs) are bimanual contact tasks: the manipulation of an object with two hands. Our overall goal is the characterization of coordination in these tasks and the use of this information in the design of robotic devices to assist rehabilitation of these tasks in patients with disability in one hand. The target population for bimanual robotic therapy devices are stroke patients with hemiparesis. Our studies of coordination in normals show that in focal\postural tasks, feedforward neural structures stabilize the postural hand. In focal\focal tasks, feedforward neural structures partition the task between the hands in a preferred way. Also, special feedback neural structures allow the two hands to respond *together* to perturbations applied to only one hand. We used this knowledge of normal coordination to design two devices. First, we prototyped a simple device that could substitute for and assist one hand in a simple one degree of freedom task: two hands hold an object between the fingertips and move it

horizontally with flexion\extension of the wrists. We used what we learned from this device to design another prototype device that could assist in a useful ADL: two hands hold an object and move it vertically with elbow rotation. This prototype could substitute for one hand in this task and will be ready for clinical testing with a few improvements. The concept of this prototype can be adapted to other bimanual ADLs, developing a family of devices that could revolutionalize stroke rehabilitation therapy. Assuming clinical testing shows such devices to be useful for stroke patient rehabilitation, they can greatly improve the current state of physical\occupational therapy: 1) They can increase the availability of therapy by automating many of its repetitive aspects. 2) They can objectively measure patient progress, providing a platform for testing different therapies. 3) They can adapt to the patient more precisely than a human therapist, delivering only that amount of assistance that is needed.

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Chapter 1

Introduction

Many activities of daily living (ADLs) are bimanual contact tasks: the manipulation of an object with two hands (fig. 1). Our overall goal is the characterization of coordination in these tasks and the use of this information in the design of robotic devices to assist rehabilitation of these tasks in patients with disability in one hand. Devices operating under the same principles may also be useful as prostheses or powered orthoses [1].

The target population for bimanual robotic therapy devices is the class of patients with motor impairment of one upper extremity, but little or no impairment on the other side, such as stroke patients with hemiparesis. The potential impact of a physical therapy robot is huge even if it is applied only to stroke patients: the incidence of stroke is 0.8 per 1,000 person years, and the prevalence is 6.0 per 1,000 persons. Approximately 10% of the survivors of the acute phase return to work without disability, about 40% have mild disability, another 40% are severely disabled, and 10% require institutionalization [2]. Robotic devices could aid therapy of the 40% with relatively mild disability, thousands of people each year.

The devices we propose to design would assist rehabilitation of these patients by helping to restore bimanual coordination when one hand has been disabled. The <u>bimanual</u> nature of the therapy offers four distinct advantages. 1) Bimanual therapy would allow

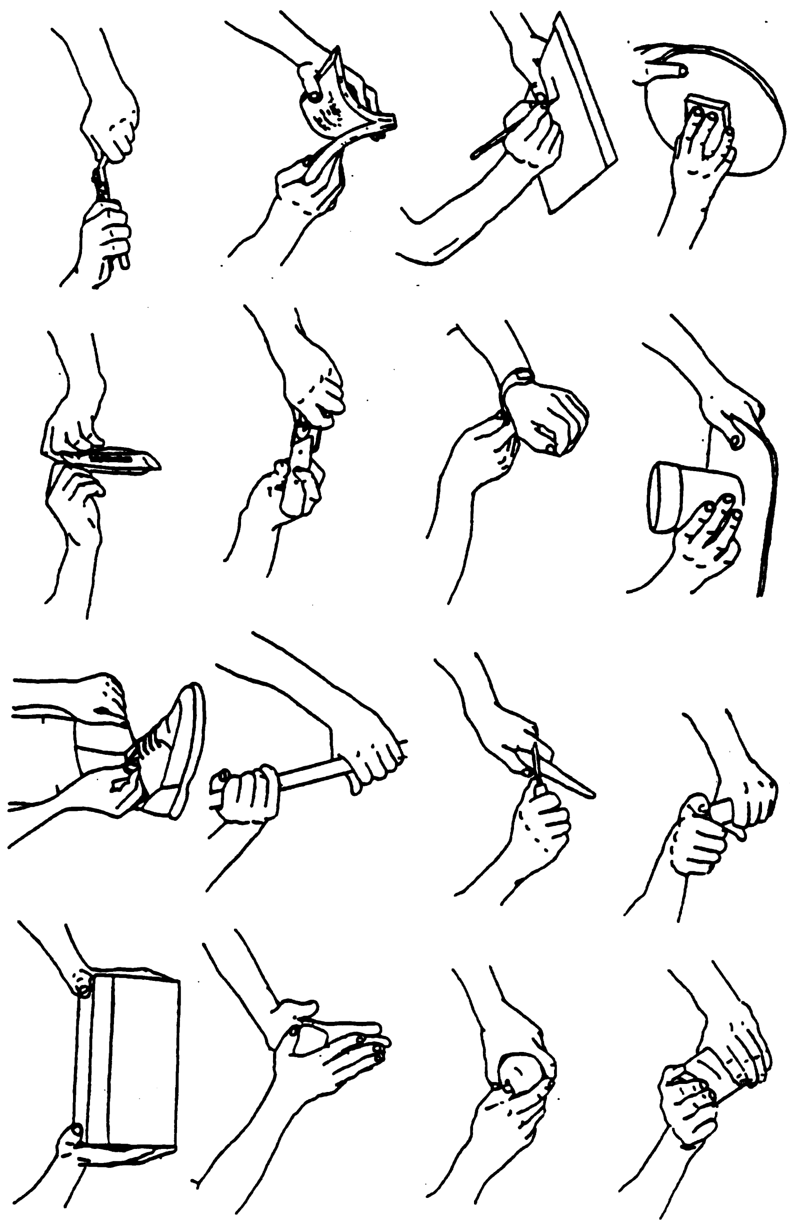

Fig. 1. Activities of daily living requiring bimanual coordination.

the able arm to drive therapy of the disabled arm, so that the patient could initiate and control therapy in a natural way. 2) Since our target patients already have one healthy hand, they need most to extend their range of performable tasks by relearning how to use the two hands cooperatively. 3) Restoring bimanual coordination may be easier than restoring single hand control. Stroke patients with disability in only one hand still retain the cortical areas controlling the undamaged hand. Recent results also suggest that neural circuits responsible for coordination in certain bimanual tasks are subcortical [3]. With therapy, it may be possible for hemiplegic patients to relearn how to access subcortical neural circuits used in postural stabilization [4]. Subcortical neural circuits used in bimanual tasks might also be accessible with proper therapy. 4) Trying to perform bimanual tasks can elicit activity from the disabled hand earlier after stroke than trying to move the paretic limb directly.

Given the complexity of the tasks in fig. 1, we first identified two broad classifications of bimanual tasks. One class of bimanual tasks is "jigging", where the two hands apply forces to an object without moving it. Examples of jigging tasks are peeling fruit, washing dishes, writing on a surface supported by the other hand and lifting a heavy object off a surface supported by the other hand. Another class of bimanual tasks is "transport", where the two hands grasp and move an object about in the workspace. Examples of transport tasks are moving a large box or pot, using a broom or mop and flossing. Jigging and transport are duals of each other: jigging is applying arbitrary internal forces to an object while maintaining constant position; transport is moving an object about with an arbitrary trajectory while maintaining approximately constant internal forces in the object. We view these two tasks to be the building blocks for all of the bimanual contact tasks in fig. 1.

Our strategy for rehabilitating these tasks can be broken down into three stages: 1) Study normal coordination in simple one or two degree of freedom (DOF) jigging and transport tasks. 2) Based on this information, design robotic devices to assist a disabled hand in performance of the task with an able hand. 3) Test these devices on hemiplegic subjects. The basic knowledge gained in studying these simple tasks can be used as a springboard for studying and rehabilitating more complex multi-DOF tasks.

Chapter 2 summarizes normal coordination in bimanual unloading, a simple jigging task. In this task one hand supports a load (object) that is removed with the other hand. The jigging phase of this task is when muscles from both hands are applying forces to the load. Once the load is removed from the supporting hand, this ceases to be a bimanual contact task. Squeezing an object with two hands is another jigging task that as been previously studied [5,6].

Chapter 3 summarizes the normal coordination in a simple transport task where the two hands hold an object between the fingertips of the two hands and move it horizontally with wrist flexion\extension of the wrists. Chapter 4 details the design and testing of a robotic device that assists one hand in the performance of this horizontal transport task. Holding is one part of this horizontal transport task. Chapter 5 summarizes the reflexes that exist when two hands hold an object by squeezing it.

In chapter 6 we apply the concepts developed in previous chapters to a useful bimanual ADL, vertical transport: two hands hold an object and move it vertically with elbow flexion. Coordination in normals is studied and a prototype device that assists one arm in this task is designed and tested.

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Chapter 2

Feedforward Stabilization in a Bimanual Unloading Task[1]

Abstract

When one hand removes a load from the other hand, feedforward motor commands stabilize the position of the postural (unloaded) hand. We studied the stabilization of the postural hand using a novel apparatus that allowed unloading at different rates, and unexpected uncoupling of the unloading force from the postural hand. Feedforward stabilization of hand position was observed in all subjects. This stabilization was achieved both by deactivation of postural agonist muscles and by activation of postural antagonist muscles. The neural feedforward command apparently increased with unloading rate. However, the command only partially canceled the interaction torque generated by removing the load, and stabilization became less effective as unloading rate increased.

Introduction

Since body segments are mechanically coupled to each other, voluntary

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[1] Portions of this chapter have been previously published under the same name in Experimental Brain Research, (1992) 89:172-180.

movements of one set of segments destabilize the posture of other segments. Stabilizing activity in postural muscles has been observed before, during and after voluntary movements. Postural muscle activity that precedes the voluntary movement and predicts the impending dynamical disturbance is called feedforward stabilization, and has been observed in several experiments [1,2,3,4,5,6].

A simple and tractable experiment for exploring feedforward stabilization has been developed by J. Massion and his group in Marseilles [7]. Subjects attempt to maintain posture in a preloaded arm as the load is removed with the opposite arm. Massion's group has demonstrated that stabilization is achieved by feedforward deactivation of the postural muscles that supported the load.

Many researchers have investigated the origin of the feedforward stabilization in this unloading task. Forget and Lamarre [8] found this stabilization present in a deafferented man, demonstrating the central origin of this stabilization. Furthermore, experiments by Dufosee et al. [9] suggest that the deactivation in postural arm muscles is not a voluntary descending signal but is generated from an efference copy of the voluntary descending signal sent to the other arm. Feedforward deactivation was not present when subjects triggered the unloading by pressing a button, or when the experimenter performed the unloading after a warning tone. Yet the deactivation was present when subjects triggered unloading by lifting a weight with the other arm. There is evidence that the supplementary motor area plays a role in the generation of this efference copy: Massion et al. [10] observed a lack of feedforward stabilization in patients with damaged supplementary motor areas.

The most detailed investigation of this unloading task has been reported by Paulignan et al. [7]. In their experiments, the load was suspended from the subject's right forearm and removed in a step-like fashion. The subject was instructed to maintain posture of the right arm as the unloading was triggered by activity in the left arm. Left arm trigger signals included threshold crossings in electromyogram (EMG), movement, and force. A brief period of decreased postural arm biceps EMG was observed when the subject triggered the unloading. This deactivation was coincident with EMG rise in the subject's left arm and must therefore be feedforward. The deactivation was absent when the experimenter triggered the unloading. Once the coordination had been established with practice, the deactivation was independent of the weight lifted by the left arm: the same deactivation occurred if the left arm lifted .5 kg, 1 kg, or 2 kg. This suggested that the deactivation might be a preprogrammed response, triggered by unloading arm activity but insensitive to its level. However, as Paulignan et al. [7] point out, the use of an unloading apparatus which always removed the load with the same rate could have obscured the dependence of the feedforward stabilization on unloading arm activity.

Our aim was to study the dependence of the feedforward stabilization on the voluntary unloading signal, which we parameterized by the maximum rate of unloading. The apparatus we used allowed unloading at various rates. By separating the torques due to the unloading arm from those due to the postural arm, our apparatus also allowed us to trick the subject occasionally, leaving the load on the postural arm despite an unloading movement by the other arm. By this means we could measure kinematic, rather than electromyographic, consequences of the feedforward command.

Methods

Overview of experiments

Subjects sat upright in a chair with the right forearm resting on a table, and the right elbow at approximately 90 degrees. The right hand held up a load attached by a cable and pulley to a rigid handle. Subjects were instructed to keep the right hand as still as possible as the load was relieved. Each experimental trial consisted of a single unloading of the right hand, either by the subject or by the experimenter. The apparatus shown in Fig. 1 and described below was designed to separate the torques due to the two hands, and thus allow the forces applied by the left hand to be separated from the right hand.

Five subjects were tested. Three were highly practiced, and two had only enough practice to perform the experiment. Each subject performed three sets of fifty trials. In the first two sets of trials, the subject removed a load from his right hand using his left hand. To guarantee a wide range of unloading rates, the experimenter viewed a display showing the peak unloading force rate for the trial just completed, and a histogram of peak unloading force rates for all the trials in the set. He then informed the subject of the peak unloading force rate of the trial just completed, and directed him to unload faster or slower, according to the contents of the histogram.

These "self"-unloadings were intermixed randomly with from six to ten trials in which the subject performed the unloading action with the left hand, but the load

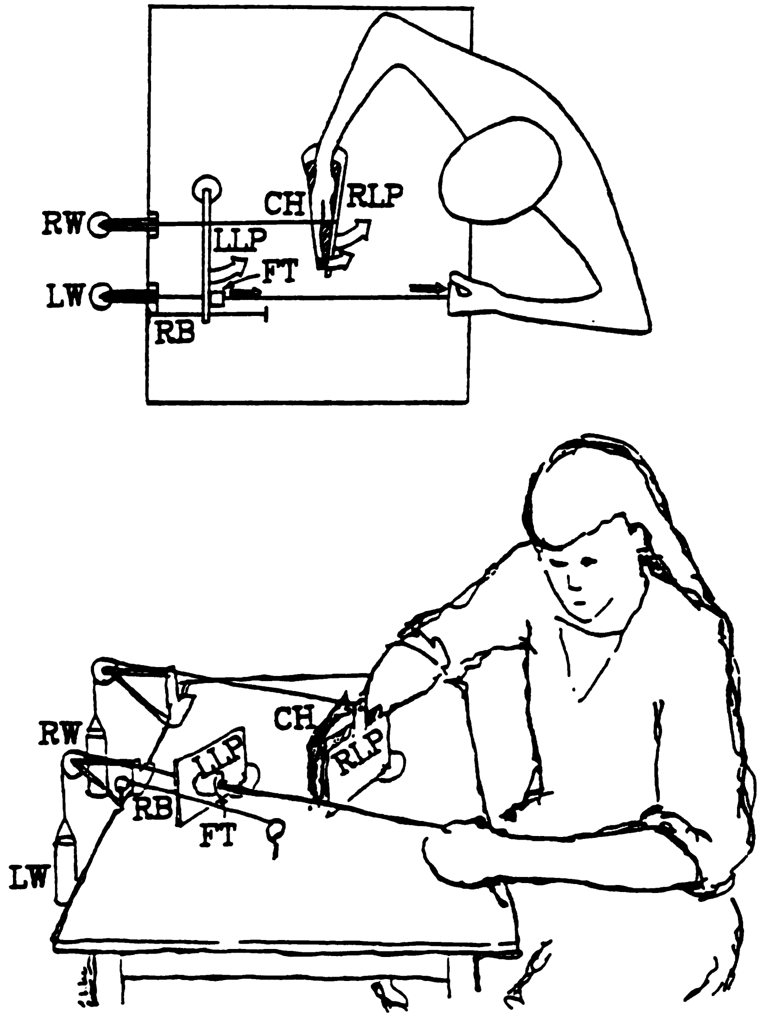

Fig 1. Experimental apparatus for simulating bimanual load lifting, shown from two views. A 1 kg weight (RW) suspended over the back of the table pulls on a load plate (RLP) which applies a torque about the subject's right wrist. The right wrist is held by a light weight cardboard handle (CH) which is free to extend away from the right load plate. The subject unloads the right hand with the left hand by lifting a matched weight (LW) supported by a second load plate (LLP). A force transducer (FT) measures the force produced by the left arm and feeds it back to a motor under the right hand, which produces a torque to lift the right hand weight. The cable connecting the left hand to the left weight passes through a hole in the left load plate. When the subject is not lifting the left weight, the force transducer rests against the left load plate, and rubber bands (RB) support the left load plate/force transducer/left weight system. When the subject lifts the left weight, the force transducer is pulled away from the left load plate, and the rubber bands simulate the compliance of the right hand.

remained on the right hand. A special apparatus, described below, allowed the experimenter to unexpectedly uncouple the torques applied by the unloading (left) arm from the postural (right) hand by blocking the input to the motor. We will call these unexpectedly uncoupled movements "blocked" trials. We limited the number of blocked trials in order to minimize disturbance of the subject's normal self-unloading strategy. Many of the subjects rested during the two sets of self-unloadings to assure fatigue was not a factor.

After performing the two sets of "self" unloading trials, the subjects were allowed to rest, and then performed a third set of fifty trials. In the third set of trials, the experimenter performed the unloading action. As before, the subject was instructed to keep the right hand as still as possible. We call these unloadings by the experimenter "imposed" trials. The experimenter was careful to duplicate the subject's unloading arm technique in an attempt to guarantee similar force traces. Fifty of these "imposed" trials were recorded with a wide range of unloading rates.

Apparatus

The apparatus, shown in Fig. 1, simulates natural unloading. A 1 kg weight suspended over the back of the table is attached to a load plate which applies a torque (1.3 Nm) about the right wrist. An equal weight applies a force to a second load plate via a cable passing through the plate and connected to a force transducer. Each load plate is free to rotate about an axis through the table (the shaft of an electric motor attached to the bottom of table). The subject lifts the left weight by pulling on a cord connected to the force transducer, displacing the transducer off of the left load plate. The force in the transducer is measured, scaled appropriately, and used as a command signal to the torque motor under the right hand. The torque motor acts on the right load plate to unload the postural arm, thus simulating natural unloading.

This simulated unloading allows the experimenter to unexpectedly uncouple the torques applied by the unloading arm from the postural wrist by blocking the input to the motor. For such "blocked" trials, any torques applied to the postural wrist arise solely from the feedforward stabilizing command. Thus the apparatus effectively isolates the effects of the feedforward stabilizing command.

The left load plate/weight/transducer system is made dynamically similar to the right hand/load plate/weight system by use of equal weights at equal moment arms and similar load plates. Rubber bands attached to the end of the left load plate simulate the stiffness of the right wrist (10-20 Nm/rad [11]). The impedance seen by the unloading arm is thus very close to the impedance seen by the torque motor.

To test the apparatus, one subject also performed "natural" unloading, for which the motor was turned off and the transducer was attached directly to the right load plate. Either the subject or the experimenter pulled on the transducer, generating "natural self" and "natural imposed" trials.

The right wrist is free to flex or extend, and is held only by a light weight cardboard handle against which the right load plate presses. The angle of the right wrist with respect to the table is measured by subtracting the output of a potentiometer which measures the angle of the cardboard handle with respect to the right load plate from the output of a second potentiometer which measures the angle of the right load plate with respect to the table.

Data collection and analysis

Force transducer and potentiometer traces were sampled at 500 HZ and stored on an IBM PC. Hand velocity and acceleration traces were obtained by digitally differentiating the position traces (Lancos differentiating filter [12]). The force transducer traces were digitally differentiated to generate force rate traces.

Electromyograms from the postural hand of one practiced and one unpracticed subject were also recorded. Eight mm Ag/AgCl surface electrodes recorded the electrical activity of the flexor carpi radialis and extensor carpi radialis. Three unloading rates (measured by maximum force rate) that spanned the subject's range of unloading were chosen and twenty "self" and twenty "imposed" trials were taken at each of these maximum unloading rates. EMGs were bandpass filtered (low frequency cutoff was 30 Hz, high frequency cutoff was 300 Hz), sampled at 500 Hz, synchronized at the time of peak force rate, rectified, and averaged.

Results

Subjects stabilize the postural hand more effectively when they perform the unloading than when the experimenter does. Typical "self", "imposed", and "blocked" trials, selected for similar unloading force traces, are contrasted in Fig. 2. The "self"

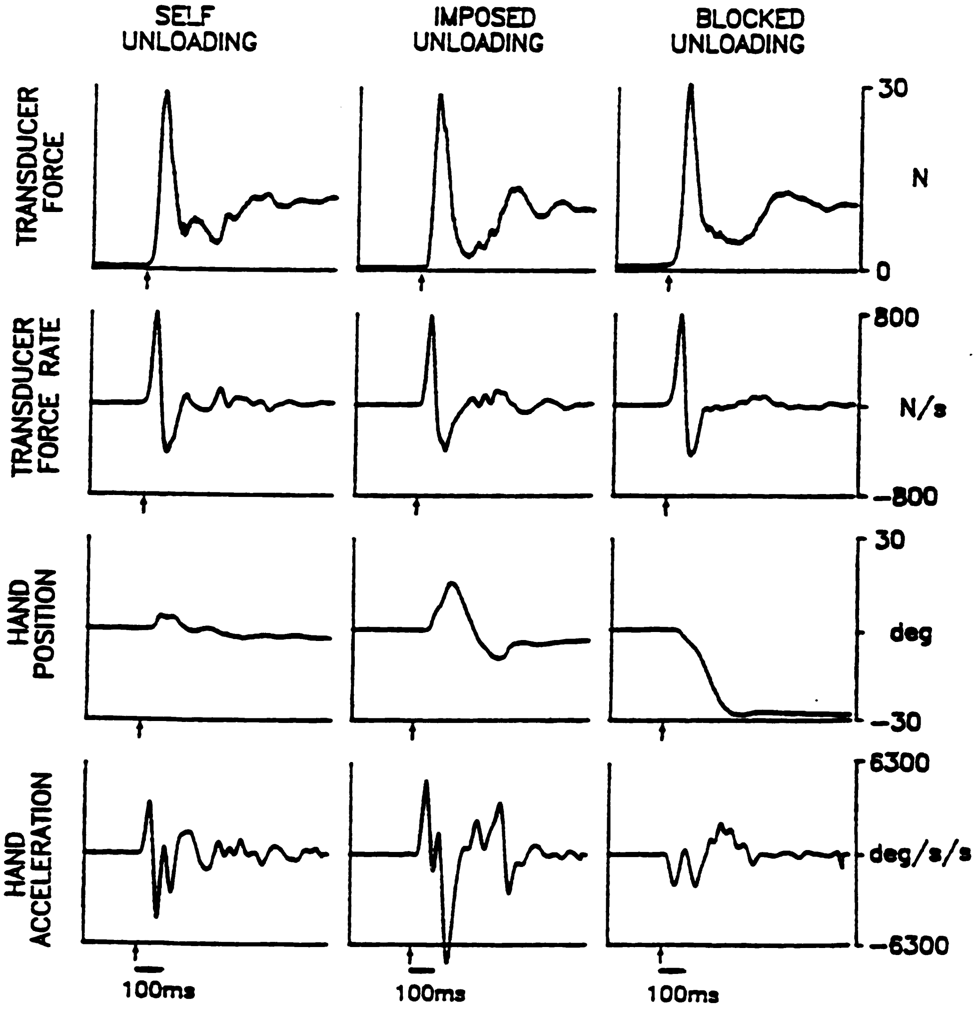

Fig 2. Force and rate of change of force for the left hand, and position and acceleration of the right hand for single "self", "imposed" and "blocked" trials. Arrows indicate first change in force.

hand displacement is much less than the "imposed" hand displacement. The mechanism for this stabilization is a feedforward motor command -- the postural arm compensates for the unloading command, rather than for the disturbance created by the unloading command. The compensation occurs even in the absence of an unloading torque: in the "blocked" trial, the hand moves backwards. Movement during the "blocked" trial also shows that the feedforward command does not simply stiffen the wrist.

We chose the first peak of postural wrist acceleration to parameterize the effects of the feedforward stabilization. The first peak of hand acceleration is well-defined for "self", "imposed" and "blocked" trials (e.g., Fig. 2). It occurs before feedback signals influence the acceleration: for all but the slowest unloading rates, the delay from unloading onset to peak acceleration was less than 100 ms (Fig. 3). The first changes in EMG in "imposed" trials occurred from 65 - 80 ms after initiation of unloading. At this latency we observed a stretch reflex in the postural antagonist muscles and an unloading reflex in the postural agonist muscles (the unloading reflex is the feedback deactivation of muscles supporting a load when that load is removed [13,14]). We also measured a 35 ms delay from EMG to force. The first peak in hand acceleration is therefore not corrupted by reflexes and can be used as an indicator of feedforward influences. Root mean square (RMS) hand position, averaged over the first 100 ms after unloading, was also used to quantify feedforward influences. RMS position can be interpreted as average position error.

We measured feedforward stabilization over a wide range of unloading rates for "natural" unloading by one practiced subject (Fig. 4). For "natural" unloading the force

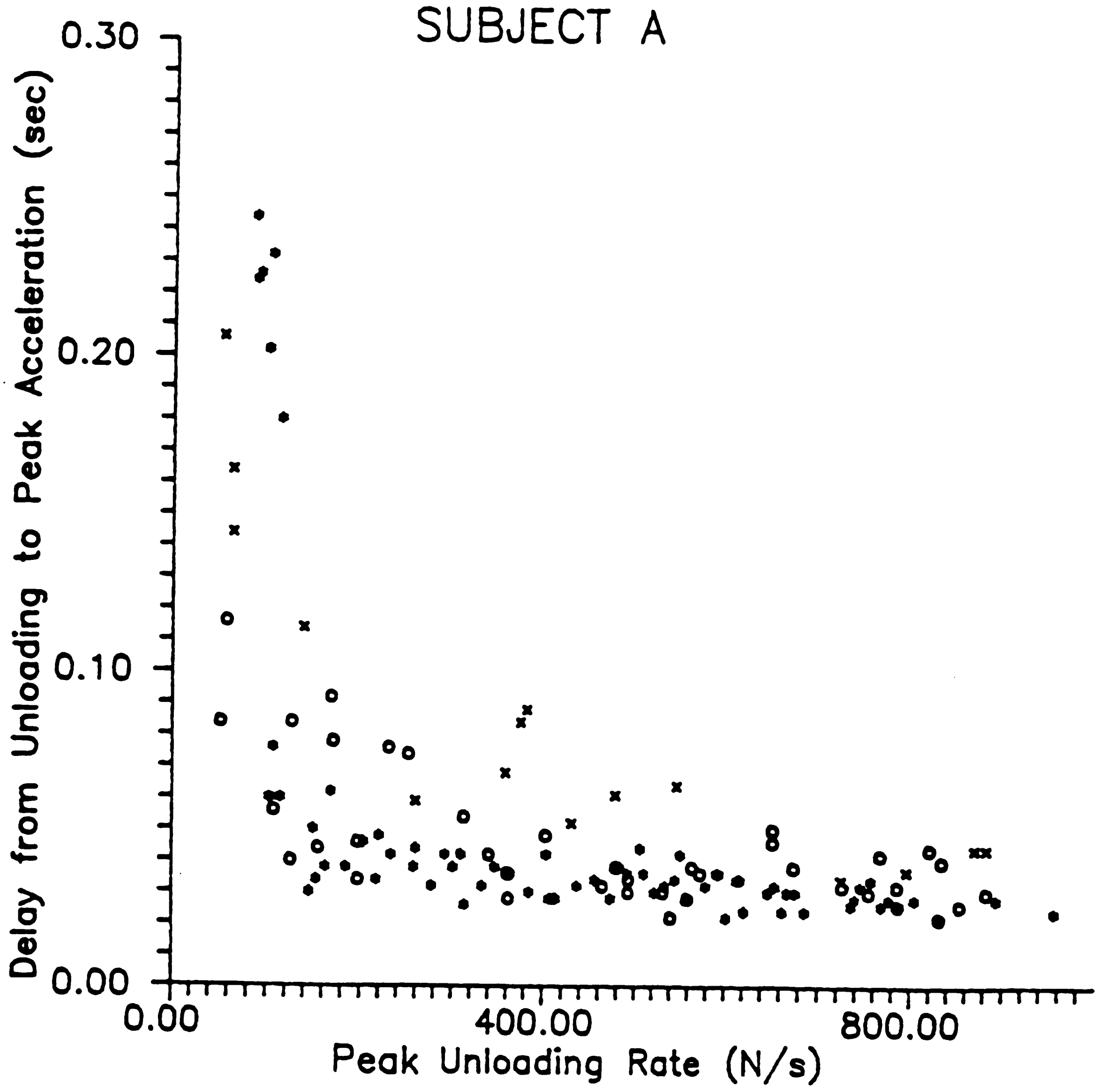

Fig 3. Delay from initiation of left hand unloading (first change in force) to peak angular acceleration of right hand as a function of peak unloading rate. * "self"; o "imposed"; x "blocked".

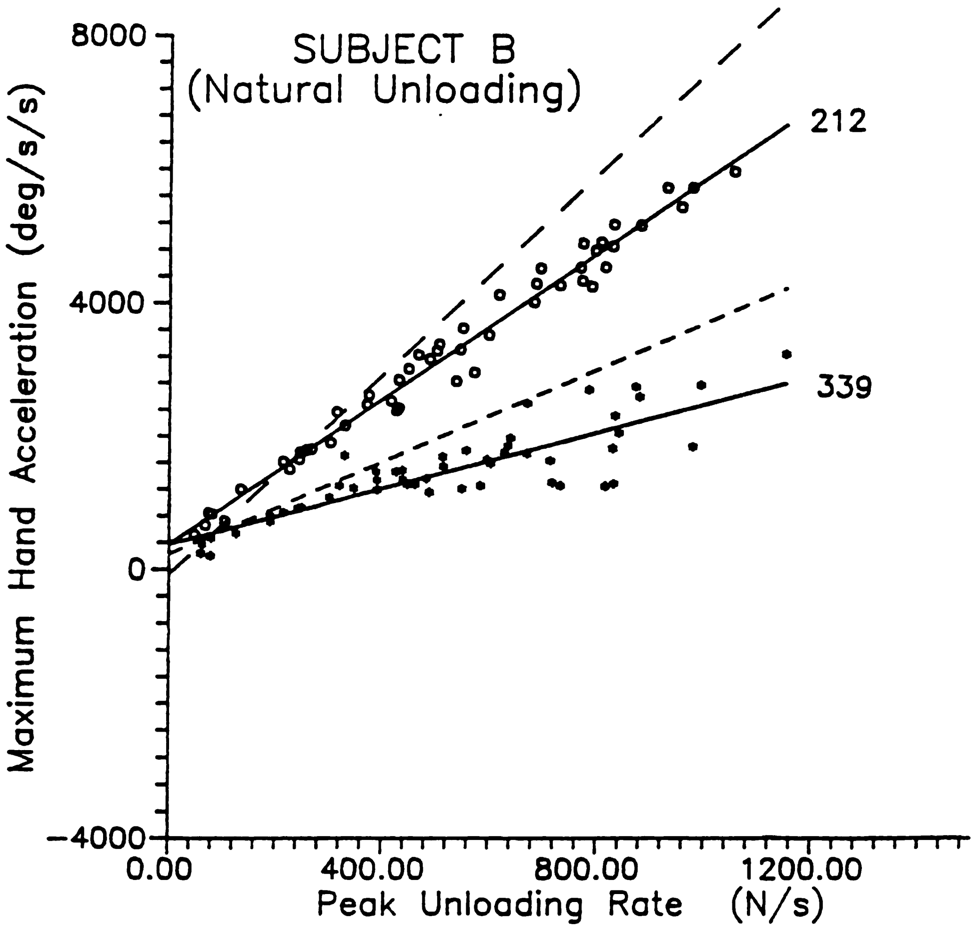

Fig 4a. Maximum angular acceleration of right hand versus peak unloading rate for "natural" unloading (left hand pulls on the right hand plate). Linear regressions are plotted for "natural imposed" (o) and "natural self" (*) trials. Root mean square error about the regression line is indicated next to each line. Linear regressions for subject B "imposed" (long dashes) and "self" (short dashes) are also plotted.

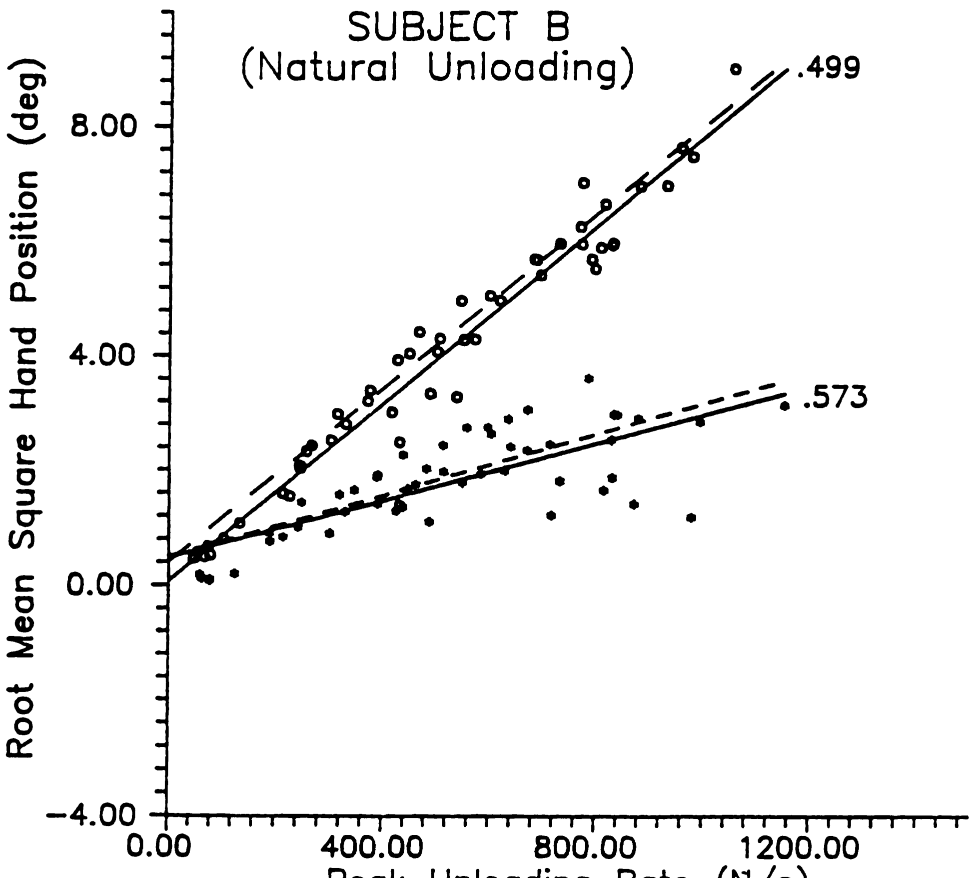

Fig 4b. Root mean square right hand angle (over first 100ms) as a function of peak unloading rate for the same subject.

transducer was attached directly to the load plate and the unloading force acted directly on the load plate. The subject was able to decrease peak hand acceleration and RMS position error in "self" unloading compared to "imposed" unloading, at any given peak unloading rate. The variability in "self" unloadings was greater than that in "imposed" unloadings. Even though the peak hand accelerations are smaller for the "self" trials than for the "imposed" trials, the hand still accelerates during the self trials. Thus the feedforward stabilization only partially cancels the unloading torque. The compensation, measured either by peak acceleration or RMS position, is less effective at faster unloading rates.

When the motor was used to simulate unloading, all subjects showed feedforward stabilization similar to that seen in the "natural" unloading for subject B (compare Figs. 5 and 6 with Fig. 4) -- peak hand accelerations (Fig. 5) and RMS positions (Fig. 6) were smaller for the "self" trials than for the "imposed" trials; the spread of the "self" trials was greater than the spread of the "imposed" trials; the compensation was imperfect, becoming less effective at faster unloading rates. The simulation of unloading gave us access to kinematic, rather than electromyographic, effects of the feedforward stabilization command -- postural hand kinematics during "blocked" trials. These showed that the feedforward command increases with unloading rate.

Closer examination of the kinematics of the "blocked" trials emphasizes this increase in the feedforward command with unloading rate (Fig. 7). Postural hand acceleration increases with force rate of the "unloading" arm from the very beginning of the postural hand movement.

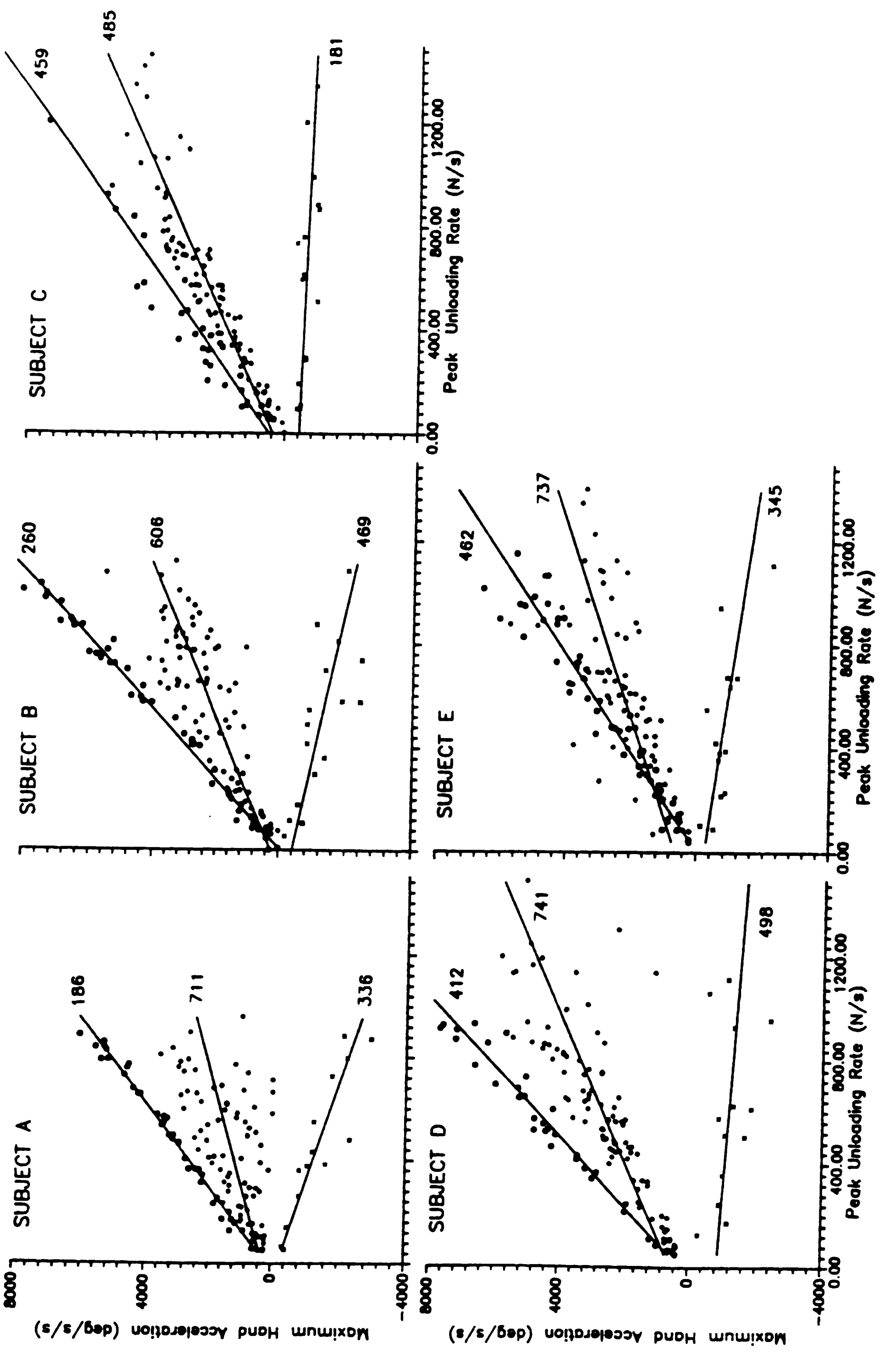

Fig 5. Maximum angular acceleration of right hand versus peak unloading rate for all subjects. Top row (A,B,C) experienced subjects; bottom row (D,E) unexperienced subjects. Linear regressions are plotted for "imposed" (o), "self" (*) and "blocked" (x) trials. Root mean square error about the regression line is indicated next to each line.

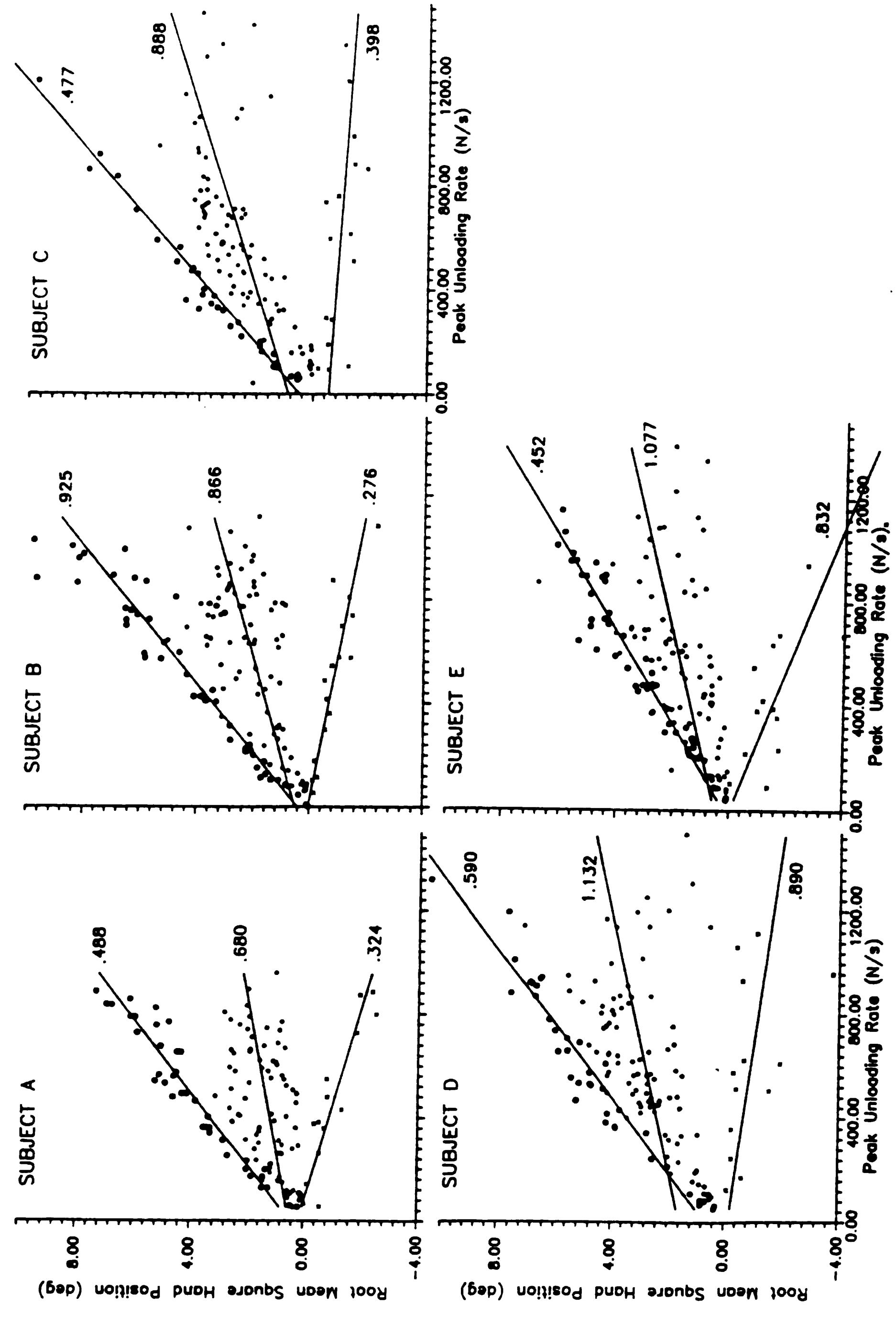

Fig 6. Root mean square right hand angle (over first 100ms) versus peak unloading rate for all subjects.

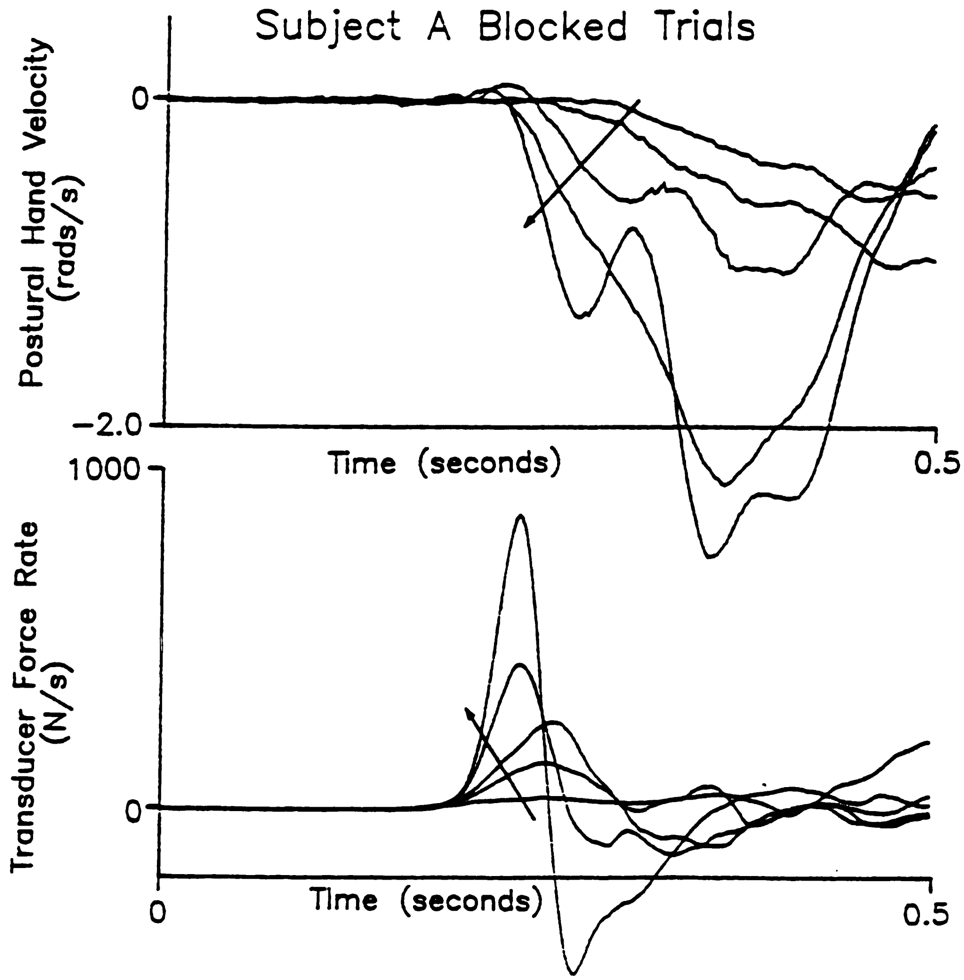

Fig 7. Postural hand angular velocities (top) for blocked trials "unloaded" at different rates (bottom). Trials are synchronized on rise in unloading force.

The apparatus adequately simulates the main features of the "natural" unloading process as evidenced by the qualitative similarity between Fig. 4 and Figs. 5 and 6. Quantitatively, however, the regression line slopes for peak accelerations in "self" and "natural self" trials are significantly different, as are the slopes for "imposed" and "natural imposed" trials (Fig. 4a). These differences in slope are probably due to differences between the dynamics of the "artificial wrist" and the subject's own wrist. The quality of simulation appears better when measured by RMS hand position. The slopes for subject B's RMS hand position in the "imposed" and "natural imposed" are not significantly different, nor are the slopes for RMS hand position in the "self" and "natural self" trials (Fig. 4b). Differences in acceleration peaks may be obscured by equal differences in deceleration valleys, integrating to the same RMS positions.

One other difference between the simulated unloadings and the natural unloadings is a smaller spread in the "natural self" trials versus the "self" trials (compare Fig. 4a and subject B in Fig. 5). The smaller spread might be attributed to the intermixing of "blocked" trials with "self" trials. "Blocked" trials are disconcerting to the subject because the subject's hand moves backwards unexpectedly. Subjects may have altered the feedforward command in anticipation of more blocked trials. Attempts were made to stabilize feedforward performance after each blocked trial with 5-10 practice unloadings. Nevertheless, it is possible that the disconcerting effects of "blocked" trials contributed to "self" trial spread.

Electromyograms from an experienced subject show that the feedforward stabilization occurs by deactivating postural agonists and activating postural antagonists

(Fig. 8). The first drop in flexor EMG in the "self" trials occurs more or less simultaneously with the rise in unloading force, thus showing the feedforward origin of this deactivation. Simultaneous with this deactivation is feedforward activation of the postural antagonist. Changes in the sizes and shapes of the feedforward commands are difficult to quantify from these averages. Notice that the EMGs from the "imposed" unloading show no change until about 70 ms after rise in force, and thus have no feedforward component. At 70 ms, the unloading reflex is seen in the postural agonist muscle and a stretch reflex in the postural antagonist muscle.

The feedforward commands to postural agonist and antagonist muscles are also apparent in the "self" trials and absent from the "imposed" trials in EMG's from an inexperienced subject (Fig. 9). However, the onset of feedforward commands relative to unloading onset varies from -17 ms for the slowest speed to about -40 ms at the fastest speed: i.e. the EMG changes precede unloading.

Discussion

In summary, we have the following results. Feedforward stabilization of postural muscles in "self" unloading is observed in all subjects, and at all unloading rates, during simulated unloading. This stabilization is achieved by a deactivation of postural agonist muscles and an activation of postural antagonist muscles. The feedforward command apparently increases with unloading rate. However, the command only partially cancels the interaction torque generated by removing the load. As unloading rate increases, the

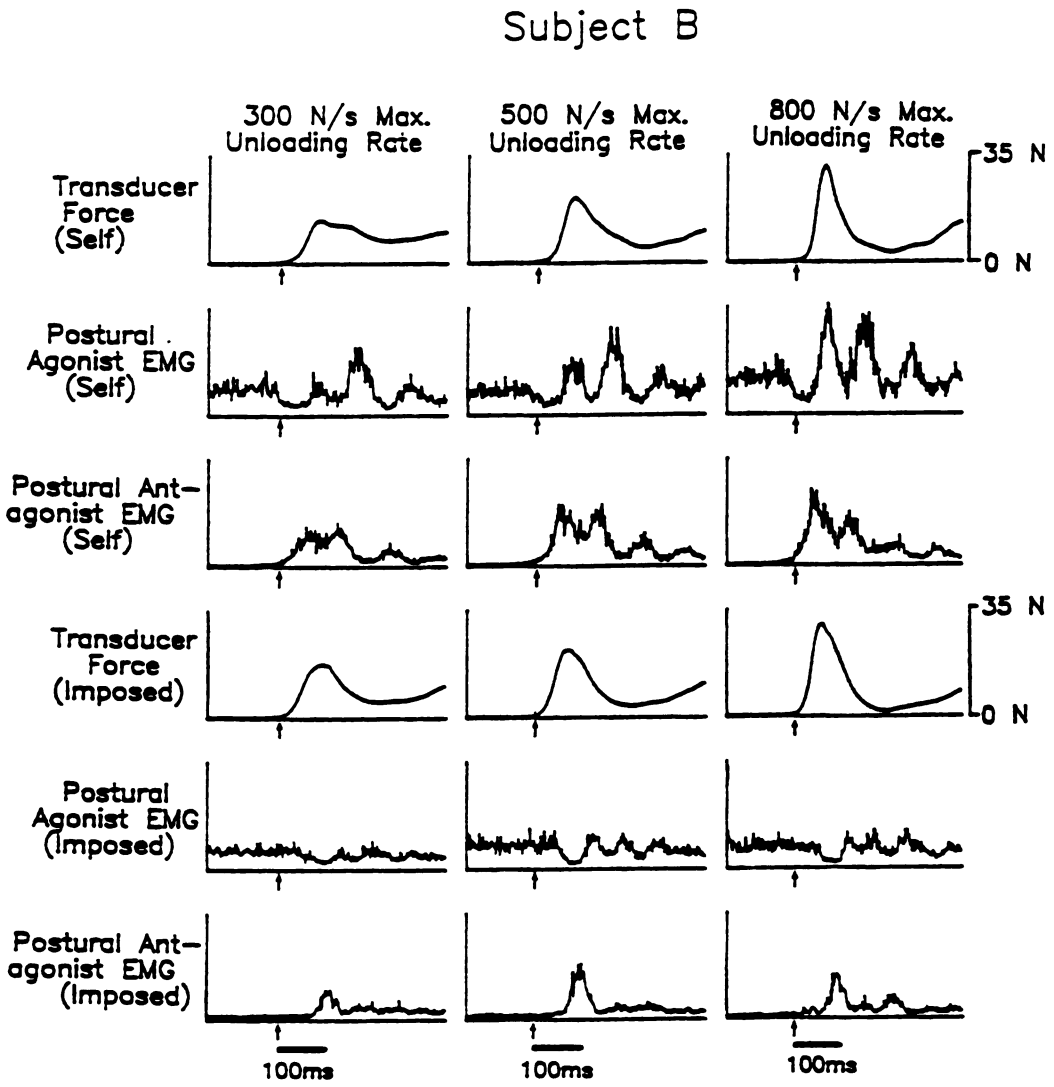

Fig 8. Force and postural arm EMG for "self" and "imposed" unloadings at three different rates (columns). Arrows indicate initiation of unloading. All traces are 20 trial averages taken from an experienced subject.

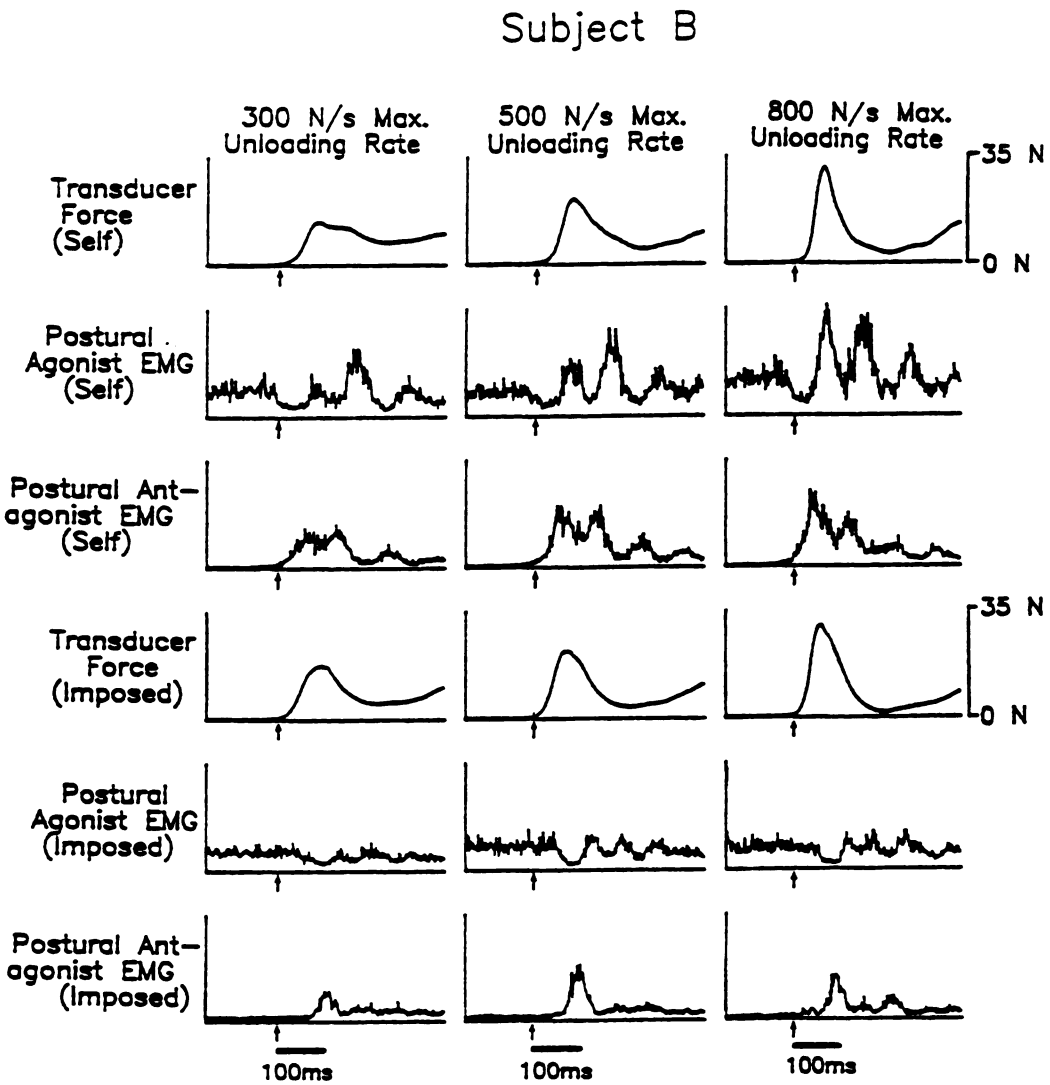

Fig 8. Force and postural arm EMG for "self" and "imposed" unloadings at three different rates (columns). Arrows indicate initiation of unloading. All traces are 20 trial averages taken from an experienced subject.

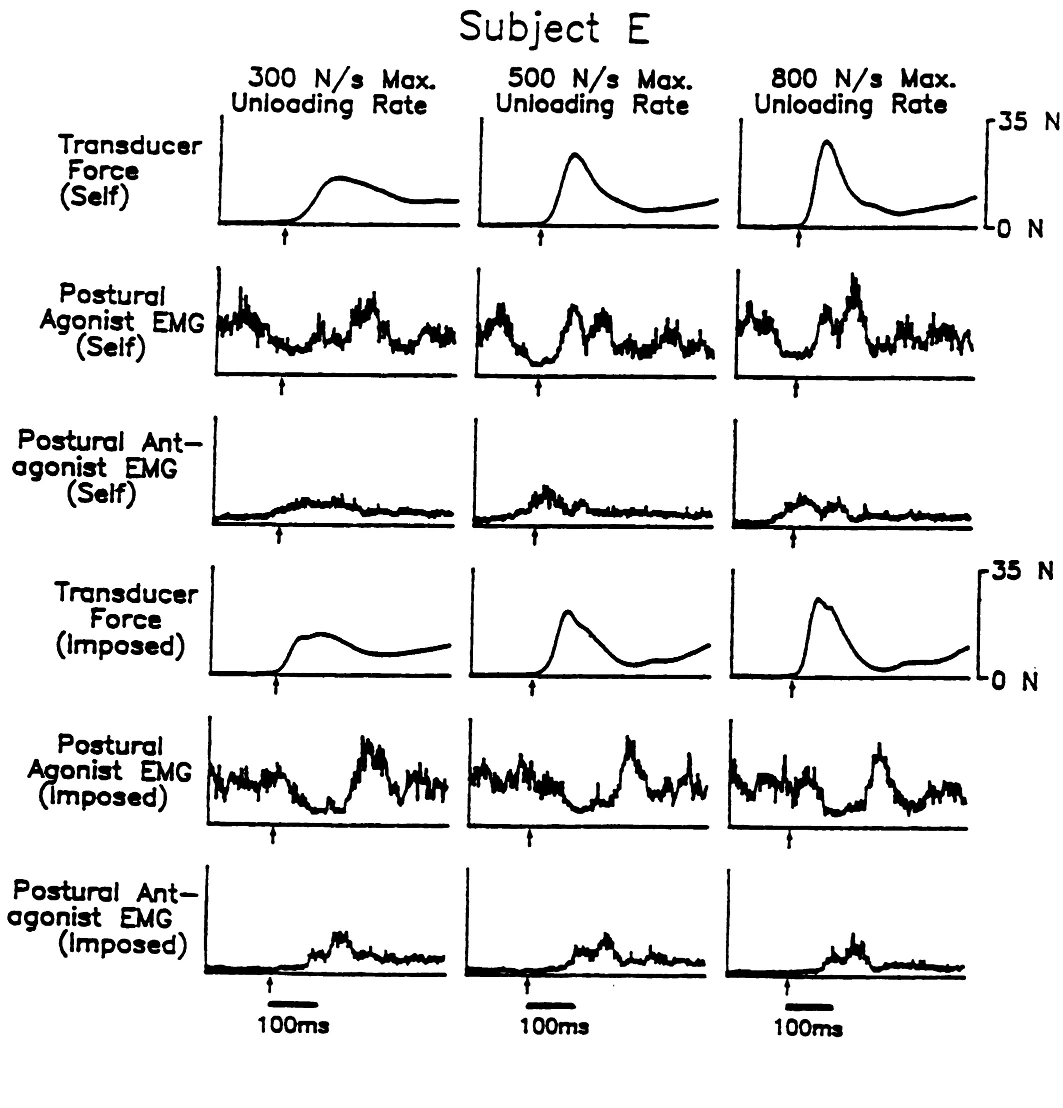

Fig 9. Force and postural arm EMG at different rates for an inexperienced subject.

stabilization becomes less effective.

One strategy for generating feedforward commands would be to use preprogrammed stabilizing commands, independent of the unloading rate and triggered by some threshold of activity in the unloading arm. The results of Paulignan et al. [7] suggest such a strategy. In our experiment, such a strategy would result in two populations for the "blocked" trials: one population for which unloading arm activity crossed the threshold, and one population for which it did not. The steady increase in "blocked" hand acceleration for faster unloading rates argue against this strategy and suggests the magnitude of the feedforward command increases with unloading rate. Grading of postural muscle activity with rate of voluntary movement similar to our findings was reported by Lee et al. [2], who studied postural muscle EMG responses to voluntary arm movements.

EMG recordings indicated that the feedforward stabilization is achieved by deactivating the postural agonist and activating the postural antagonist (Figs. 8 and 9). Thus stabilization is achieved by cancellation of interaction torques rather than by stiffening the wrist. Feedforward activation of the postural antagonist was also observed by Forget et al. [8] but not by Paulignan et al. [7].

The fact that the postural hand accelerates during "self" unloading implies that feedforward stabilization only partially cancels the interaction torque generated during unloading. Stabilization may be only approximate because of limitations in the neuromuscular hardware that implements the stabilization. On the other hand, more precise stabilization may not be necessary.

The stabilization is not only imperfect, but also noisy. The variability in the stabilization is probably not in the postural arm dynamics, muscles, or the tonic postural signal, because the standard deviation in "self" trials is always greater than that in "imposed" trials, for which the signal to the postural arm doesn't change (up to reflex time). It is also unlikely that the noise is in the unloading arm or the voluntary unloading signal, because the experimenter and the subject show about the same variability in their unloading trajectories (Fig. 10). The noise is therefore likely to be in the transformation from voluntary unloading signal to the feedforward postural adjustment signal.

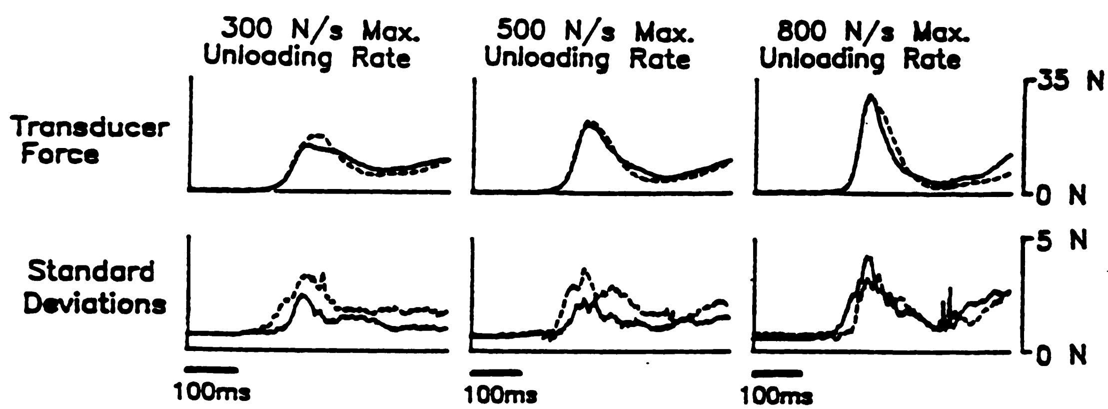

Fig 10. Average force traces for three peak unloading rates (columns) for "self" (solid) and "imposed" (dashed) trials. The second row plots standard deviations of the force averages.

Given the lack of an obvious cost of an error, it is perhaps surprising that the stabilization is as good as it is. We hypothesize two other possible uses for this coordination. First, the coordination may be used to simplify control of the unloading hand when removing an object supported by the other hand. Suppose an object is supported by the postural arm and the person wishes to remove this object with a desired trajectory from the support. Due to the spring-like nature of muscles and tendons, there will be a period early in the unloading when both the postural and unloading arms are exerting forces on the object. The central nervous system (CNS) must coordinate the voluntary unloading signal with the force the postural arm places on the object during this coupled phase. If however, the CNS can counteract the spring-like nature of the postural arm with feedforward stabilization, the coupling time is reduced and the postural arm mimics a grounded surface. Then the CNS can use the same unloading signal as it uses to lift the object off any grounded surface. This is analogous to the dynamic simplifications in moving the arm about in the workspace with the shoulder joint stabilized by feedforward signals.

The second possible use of the coordination is that it may permit a person who is manipulating an object with two hands to vary the internal force in the object while keeping the object stationary. Examples of such tasks are peeling a carrot, writing on a surface supported by the other hand, peeling an orange, washing dishes and lifting a heavy object off a tray supported by the other hand. In each case, the "focal" or voluntary hand operates on the object, applying very specific and complicated forces to the object. The other hand automatically compensates for these forces and stabilizes the object. The mechanism for this stabilization is similar to the feedforward compensation seen in this unloading task.

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Chapter 3

Horizontal Transport : Two Hands Controlled as One

I. Introduction

Many activities of daily living (ADLs) involve manipulation of an object with two hands (see fig.1 of Chapter 1), yet surprisingly little is known about control of these tasks. Our overall goal is the characterization of the control of these tasks in normals so that this knowledge can be used in the design of devices that can be used by physical or occupational therapists to rehabilitate these tasks in patients. This knowledge can also be useful in functional electrical stimulation, prosthetics and robotics.

We have identified a simple class of bimanual manipulation tasks which we call transport tasks. These tasks involve moving an object about with two hands. This chapter describes the research done on a specific transport task: moving an object held between the fingertips of the two hands in the horizontal plane with wrist flexion/extension (figs. 1 & 2). We call the system consisting of the two hands grasping an object and constrained to one degree of freedom (DOF), the hand-object-hand (H-O-H) system.

A more "natural" horizontal transport task might be holding an object with two hands and moving it with a combination of torso, shoulder, elbow and wrist rotation. The motivation for choosing the H-O-H system is threefold: 1) The "natural" horizontal transport task described above has 7 DOFs. Our H-O-H system has only one DOF and yet is still capable of horizontal transport. 2) Control of multi-joint movements is an active area of research [1,2,3]. Most of this work concentrates on control of serial linkages such as the shoulder and elbow. Transport with the H-O-H system also requires coordination between two joints but is simpler dynamically due to the absence of coriolis and centrifugal interaction torques present in serial linkages. In fact, previous results have shown that the dynamics of the H-O-H system are similar to single hand dynamics [4]. Therefore, we can study interjoint coordination in a dynamically simplified setting. 3) The control of free wrist movements are well understood [5,6,7]. This task allows us

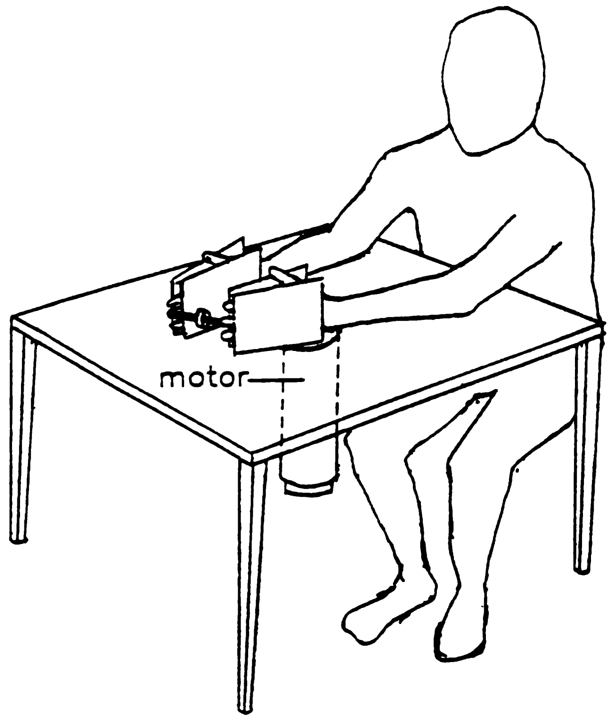

Fig. 1. Hand-Object-Hand system. Handles allow only wrist flexion/extension. The object is a force transducer extended with two machine screws. The object length is equal to the distance between the axes of rotation of the two handles.

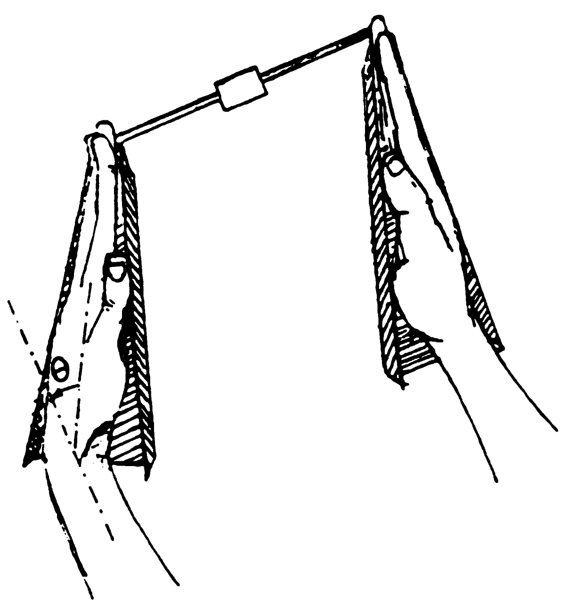

Fig. 2. Top view of H-O-H system. Positive torque and rotation are defined to be in the counterclockwise direction.

to study how single joint control can be adapted and used in multijoint tasks.

The motivation for studying this task is driven more by dynamic simplicity and basic research in motor control rather than its usefulness as a ADL. It is unlikely that this is a task that needs to be rehabilitated. However, we believe study of the H-O-H system to be a building block for study and rehabilitation of more complex bimanual ADLs. For example, further study in horizontal transport might start with a system consisting of the wrist and elbow of both arms, eventually building up to the "natural" horizontal transport system described above.

The goal of this chapter is the characterization of normal control of the H-O-H system in two types of movements: discrete and rhythmic. The general hypothesis which we wish to investigate is:

General Hypothesis : the two hands in the H-O-H system are controlled like one hand.

Single hand free movements are made with an agonist muscle that initiates the movement and an opposing antagonist muscle that breaks the movement. Our general hypothesis states that the H-O-H system is controlled like one hand; one hand takes on the role of agonist for the H-O-H system and other hand takes on the role of antagonist. An alternative hypothesis is that *each* hand is controlled as in single hand free movements; that the control signals sent to *each* hand in bimanual tasks are the same as the control signals used in single hand free movements. To test these hypotheses, subjects performed both single hand and bimanual movements (discrete and rhythmic). The kinematics and emgs of the bimanual movements were compared to the single hand movements. The implications of our general hypothesis are discussed in Summary.

II. Discrete movements

Introduction

We studied discrete movements of the H-O-H system and compared them to single hand discrete movements at the same speed. The movement speeds we tested ranged from "as fast as possible" to as "slow as possible while still making a single movement". Neural control of rapid single joint movements is characterized by a triphasic burst pattern of muscle activity (fig. 3). This triphasic burst pattern has been shown to mediate rapid elbow rotations [8], wrist rotations [6], as well as finger flexions [9]. The amplitude of the first agonist burst is dependent on both movement size and required movement force [10]. Therefore, there is little question as to the central origin of this burst. The antagonist burst is present, but of lessor amplitude when actual movement is unexpectedly prevented [11]. This demonstrates that the antagonist burst is centrally generated, but also suggests reflex modulation of the triphasic pattern after about 100 ms. Furthermore, the triphasic pattern is present in deafferented subjects, providing evidence for the central origin of the entire triphasic pattern [12,13]. Preliminary studies have shown triphasic patterns are used in H-O-H discrete movements [14]. We will compare the emg patterns used to move the H-O-H system to the triphasic patterns used in single hand movements.

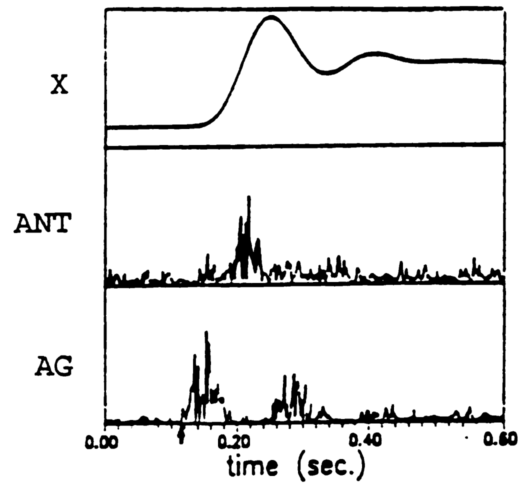

Fig. 3. Triphasic EMG pattern associated with rapid wrist movements. From top to bottom, hand position (x), antagonist muscle activity (ANT), agonist muscle activity (AG).

Many investigators have postulated that movements of different type (size,speed,load, etc.) are mediated by modulation of the pulse height, pulse width and pulse onset of the triphasic burst pattern. Gottlieb et al. [15] have postulated a "speed sensitive" (SS) strategy results from movement goals related to movement speed, movement time or target size. This strategy involves increases in agonist and antagonist pulse heights and decreasing antagonist onset latency with increasing movement speed (increasing speed instruction, decreased movement time instruction, or larger target size). Of particular importance is the fact that this strategy can be identified by examination of net inertial torques applied to the system. How and if these strategies may be generalized to other than single joint free movements is still an open question. We will test if the SS strategy generalizes to control of the H-O-H system.

Methods

Three subjects were tested. Subjects placed their hands in two rigid wooden handles and grasped a 17.5 cm long pencil-like object consisting of two machine screws attached to both sides of a force transducer (Sensotec model 31, 25 lbf range, 0.5 V/kgf sensitivity) (figs. 1 & 2). The handles constrained movement to wrist flexion/extension. Precision potentiometers measured the angular position of the hands (0.05 degree resolution), while a torque transducer (West Coast Research Corp. model 44A0-5.2, 500 in-lb range, 0.5 V/Nm sensitivity) measured the torque in the shaft of the left handle. In a typical trial, hand angular position, grasp force (transducer force) and emgs from flexors and extensors were sampled at 500 Hz, digitized and stored on a IBM PC/AT. Electrodes were placed over the right flexor carpi radialis (RF), right extensor carpi radialis (RE), left flexor carpi radialis (LF) and the left extensor carpi radialis (LE). Movement angular velocity and angular acceleration were obtained by digital differentiation (Lancos differentiating filter [16]) of angular position records.

H-O-H position was presented to the subject along with a two degree target on an oscilloscope. The initial target location corresponded to 10 degrees clockwise (cw) of the center position of the hands. The subject moved into the target, and after a random delay, the target moved to another location corresponding to 10 degrees counterclockwise (ccw) of the center position. Subjects were instructed to move into the new target window at one of three different speeds: fast as possible while still ending up in the target (fast), slow as possible while making a single movement (slow) and one intermediate speed (medium). Subjects were informed of the peak velocity of each movement. For each of these three speeds, an acceptable peak velocity range was determined for each subject and only movements that fell within that range were recorded. Recorded movements ranged from 25 to 50 movements for each speed. After performing these bimanual movements, each subject performed single hand free movements with first his right hand and then his left hand at the same three movement speeds as his bimanual movements. Again, recorded movements ranged from 25 to 50 trials. All three subjects performed these 20 degree ccw movements. One subject performed a similar array of 20 degree cw movements.

Movements of the same speed were synchronized on peak velocity and ensemble averaged. Emgs were synchronized on peak movement velocity, rectified, ensemble averaged and low pass filtered with a 14 ms moving average filter.

We have shown previously that a simple dynamics calculation determines the torques produced by each hand in moving the H-O-H system [4] :

$$\tau_r+\tau_l=(J_r+J_l+Mr^2)\ddot{\theta}$$

$$F=\frac{\tau_r-\tau_l}{2r\cos\theta}$$

where τ_r is right hand torque, τ_l is left hand torque, F is the grasp force, J_r is right side inertia, J_l is left side inertia, M is the mass of the object, r is the distance from the wrist axis of rotation to the fingertip-object contact point, and Θ is the angular position of the H-O-H system. Positive rotation and torque are in the ccw direction. Parameters for this calculation were all obtained experimentally. Handle inertias were determined by rotating the handles sinusoidally with a torque motor at a frequency of 3 Hz and measuring the amplitude of the resulting movements. Two sets of handles were used. Subject DR used handles with inertias of 0.0017 kg m^2. Subjects PL and MD used handles with inertias of 0.00036 kg m^2. Hand inertias were estimated by measuring the volume of water displaced by the hand, assuming a density of 1 gm/cm^3 to calculate mass of the hand, and modeling the hand as a rectangular slab [7]. The hand inertias for subjects ranged from 0.0031 to 0.0053 kg m^2.

Two subjects performed "stopped" trials. These were trials in which the hands were unexpectedly stopped during a sequence of bimanual movements under the "fast as possible" instruction, as described above. The H-O-H system was unexpectedly stopped and the isometric torque from each hand measured. Twenty two of 174 trials were randomly and unexpectedly stopped for subject ES. Forty six of 300 trials were stopped for subject DR. The apparatus used to stop the system was a bicycle caliper brake mounted so that it could clamp onto a metal bar extending from the shaft of the left handle. The brake was set up to stop movements either at the center line or just before peak overshoot. All of these movements were in the ccw direction. After the brake was applied, the whole H-O-H system was stopped. The right hand isometric torque was determined by the force transducer. The left hand isometric torque was determined by subtracting the right hand isometric torque from the torque transducer trace which measured the torque in the shaft of the left handle.

Three subjects performed "stopped" trials of right hand free movements also under the "fast as possible" instruction. From 39-60 of 300 trials were stopped for each subject. The brake was set up to stop movements at one of three locations: at the center line, near peak overshoot, or at a location intermediate between these two points. The torque transducer measured the hand isometric torque.

Results

We have 5 results that support our general hypotheses. 1) The kinematics of bimanual movements up to peak overshoot are identical to single hand kinematics. 2) The emg patterns used to move the H-O-H system are similar to emg patterns used in one hand free movements, if the velocity of movement is high enough. 3) The bimanual emgs are modified from single hand emgs so that one hand acts as "agonist" and the other hand acts as "antagonist" for the H-O-H system. 4) The "speed sensitive" single joint movement strategy used for dealing with a movement speed instruction task requirement is used in this bimanual task. 5) Braking strategies used in "fast as possible" single hand movements generalize to this bimanual task. All of the results presented are for ccw movements. One subject also performed movements in the cw direction. All of the results generalize to cw movements.

The kinematics of bimanual movements up to peak overshoot are identical to single hand kinematics. Fig. 4 overlays the bimanual and single hand trajectories at three speeds for three subjects. A pointwise t-test was done comparing the bimanual with the single hand movements. The bars above each plot in fig. 4 indicate where the bimanual trajectories are significantly different from the right (upper bars) and the left (lower bar) hand trajectories. Except for a few isolated points, the single hand and bimanual trajectories were not significantly different ($p > 0.05$) from the onset of movement up to

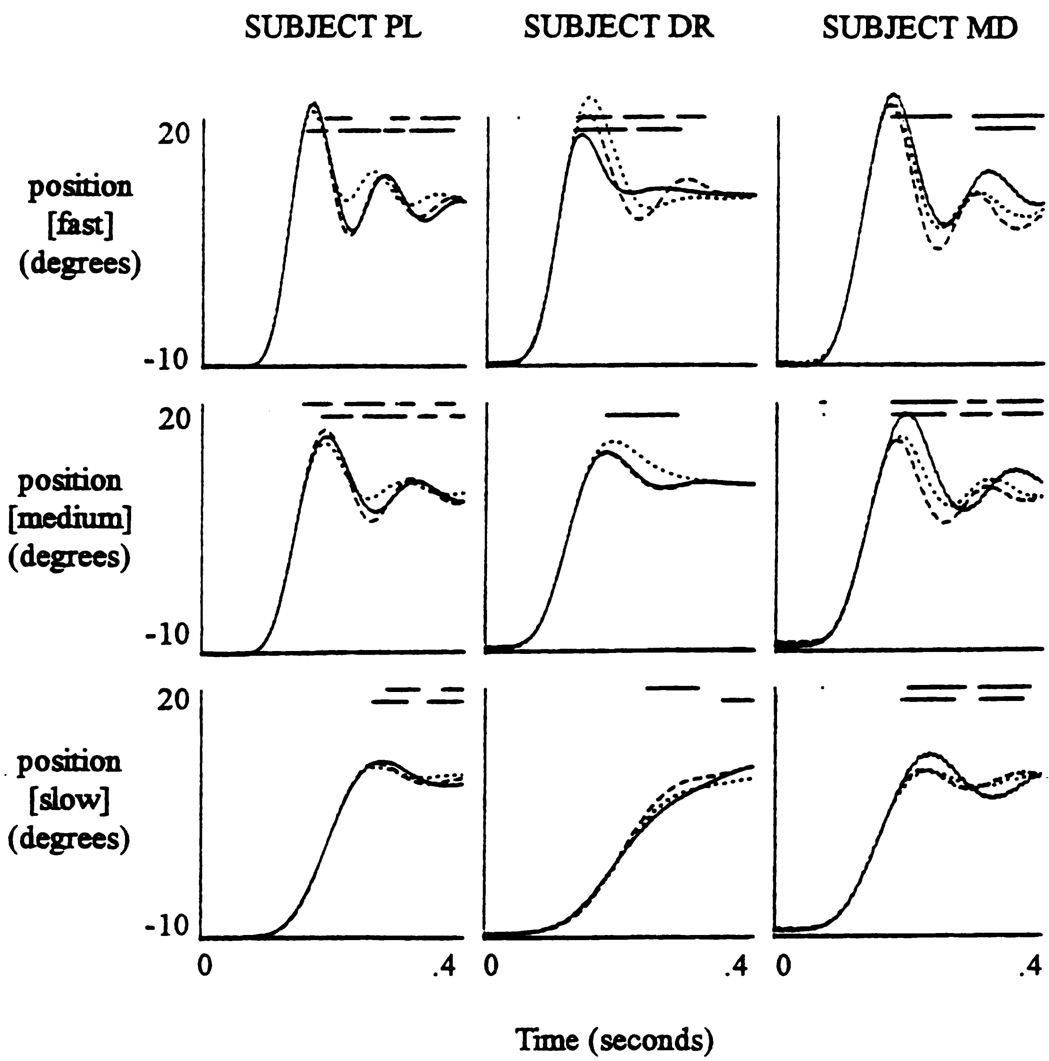

Fig. 4. Bimanual (solid), right hand (dashed) and left hand (dotted) discrete movement kinematics at three speeds for three subjects. All trajectories are ensemble averages of 25-50 trials.

near peak overshoot at all three speeds for all subjects. However, the trajectories did show a significant difference after peak overshoot.

The EMG patterns used to move the H-O-H system are similar to emg patterns used in one hand free movements if the velocity of movement is high enough. Figs. 5,6,7 overlay the emgs from the bimanual movements with single hand movement emgs at three speeds and for three subjects. For the most part, each muscle showed a similar emg pattern in bimanual and single hand movements. For example, consider subject PL. He showed bursts in the RF, RE, and LF at all three speeds in both bimanual and single hand movements. In each case the number of bursts and the timing of the bursts were about the same. The LE for this subject showed an initial fast rise in emg and a slow decline in both bimanual and single hand movements. The other subjects' bimanual and one hand emgs were also similar, except in the case of subject DR at his slow speed. However, his slow speed was only 60% of the slow speed of subjects PL and MD and seemed to be close to the lowest speed possible while still making a single movement. In fact, many of his movements at this speed had two velocity peaks indicating two discrete movements. At this low speed, the extensors in the bimanual task were not active at all and movement was made with the flexors only.

Despite similarity in form and approximate timing, there are consistent differences between the bimanual and single hand emgs. The initial burst in the RF emg was larger in amplitude for all three subjects and at all three speeds. There were only minor differences in the initial RE emg burst at the medium and fast speed. However, at the lower speed, the RE was larger for subject MD, but smaller for subjects PL and DR.

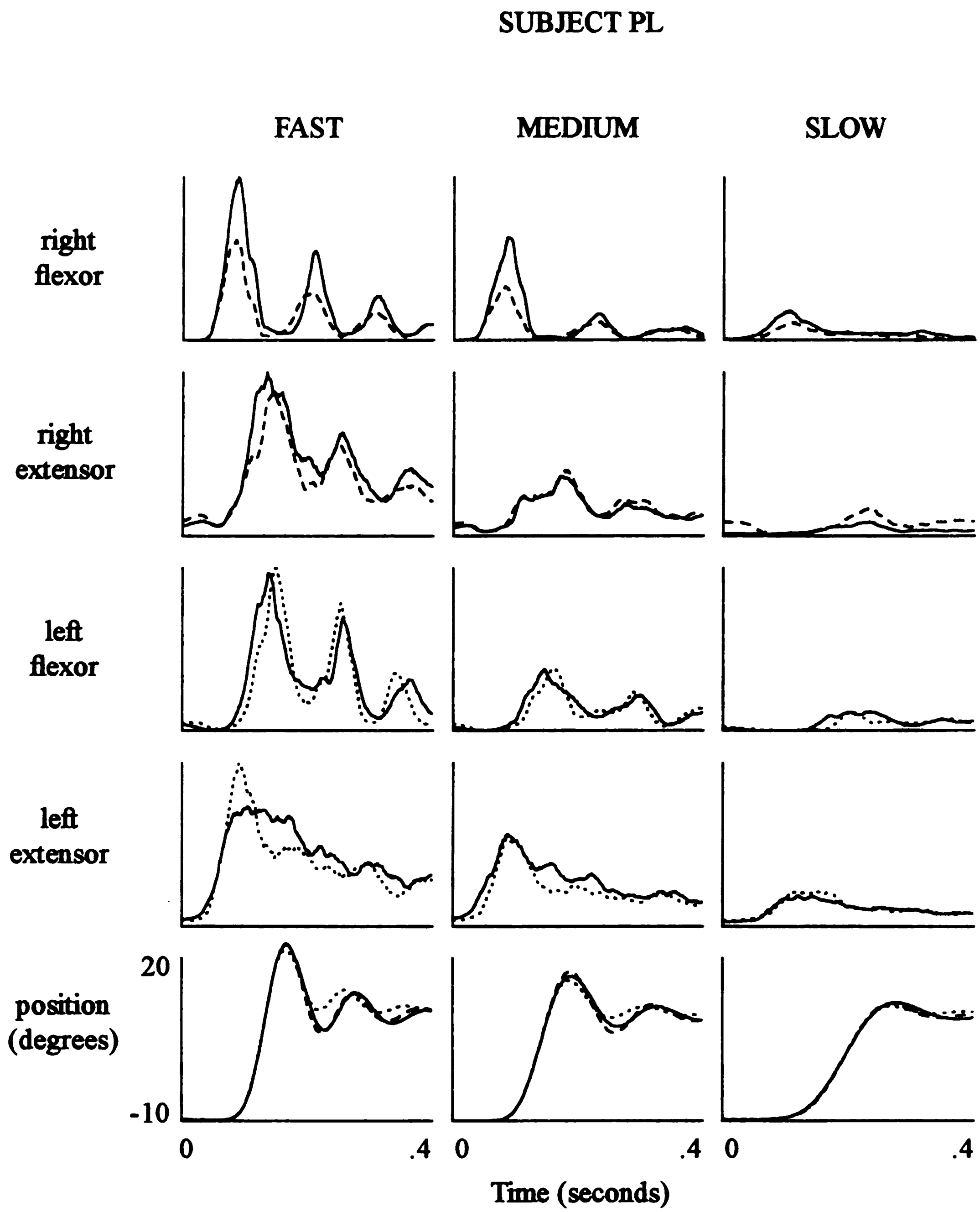

Fig. 5. Emgs for bimanual (solid), single hand right (dashed) and single hand left (dotted) movements for subject PL. All trajectories are ensemble averages of 50 trials.

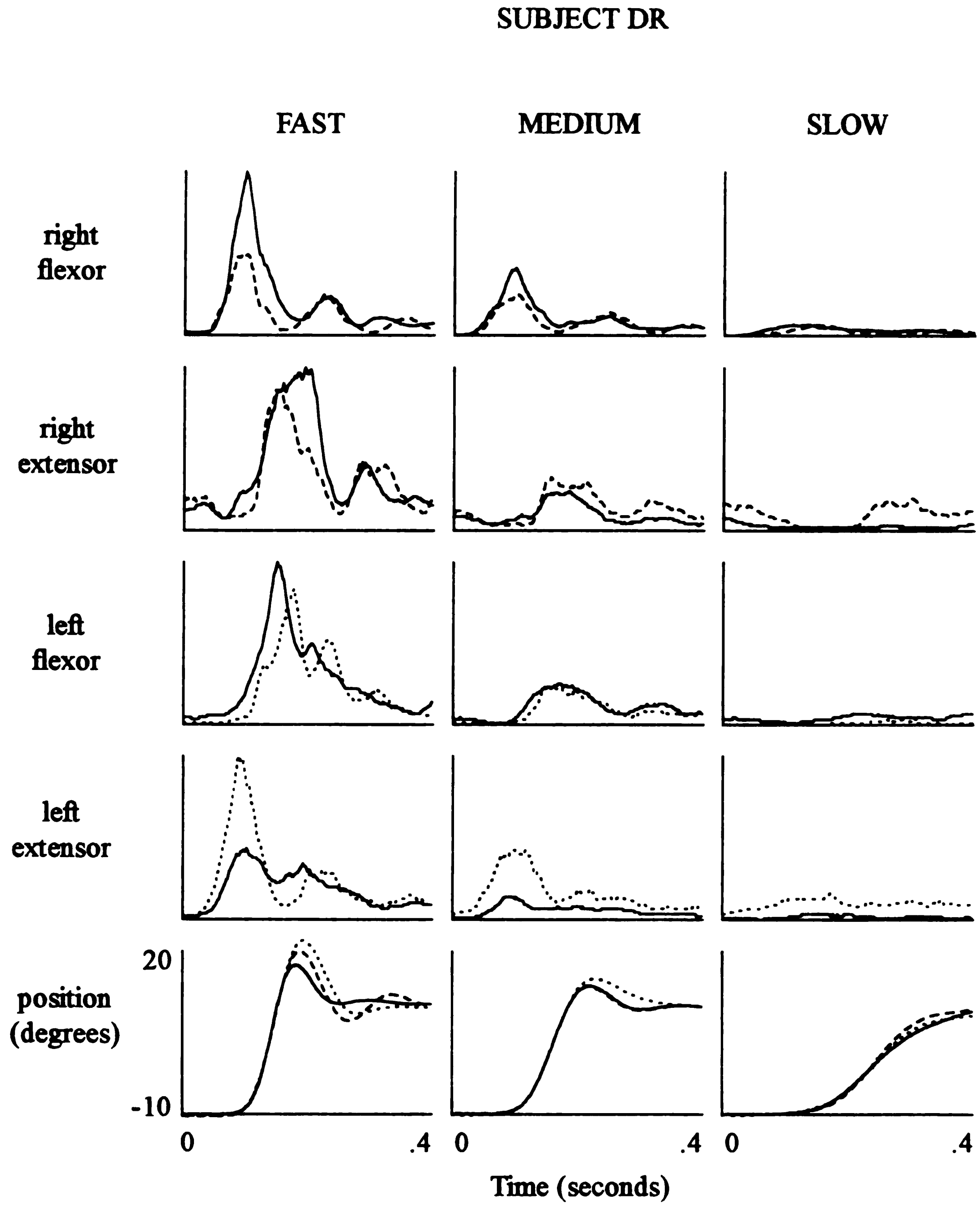

Fig. 6. Emgs for bimanual (solid), single hand right (dashed) and single hand left (dotted) movements for subject DR. All trajectories are ensemble averages of 50 trials.

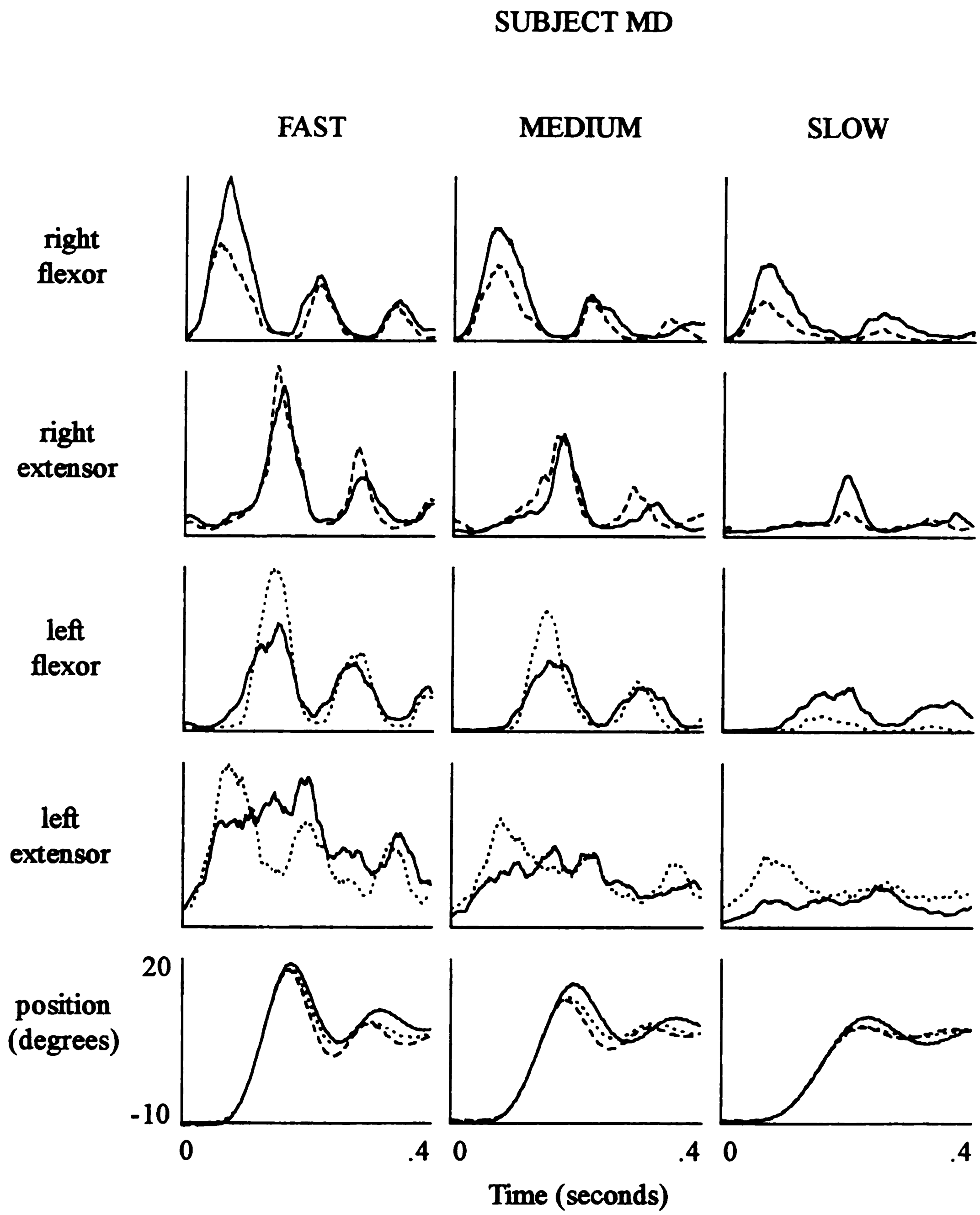

Fig. 7. Emgs for bimanual (solid), single hand right (dashed) and single hand left (dotted) movements for subject MD. All trajectories are ensemble averages of 25 trials.

The LF initial burst was advanced at all three speeds and for all three subjects, but the amplitude was variable. Subjects DR and PL showed no difference in the amplitude of LF, while subject MD showed a smaller LF amplitude burst at the middle and high speed but a larger burst at the slow speed. The LE emg was smaller or the same in amplitude vs. single hand movements in all cases. To summarize, emgs for bimanual movements had increased amplitude of the RF emg, advanced onset of the LF emg and decreased amplitude of the LE emg.

One hand acts as "agonist" and the other hand acts as "antagonist". Figs. 8,9,10 show the torques produced by each hand in the bimanual and single hand movements for three subjects. For subject DR, the right hand always produced all the positive torque needed to accelerate the system while the left hand produced all the negative torque needed to brake the system. Hence, the right hand acts as the agonist of single hand free movements and the left hand acts as the antagonist. For subjects PL and MD, this strategy was also present, but less pronounced. The right hand did produce some negative torque but only a small fraction of the braking torque from the left hand. Similarly, the left hand did produce some positive torque but only a small fraction of the positive torque from the right hand. As a result, subject DR had the highest grasp forces, while PL and MD had much smaller grasp forces.

The "speed sensitive" single joint movement strategy used for dealing with a movement speed instruction task requirement is used in this bimanual task. Fig. 11 shows how the emgs of single hand and bimanual movements were modulated with speed. The single hand movements showed the SS strategy: increase in burst amplitude in both

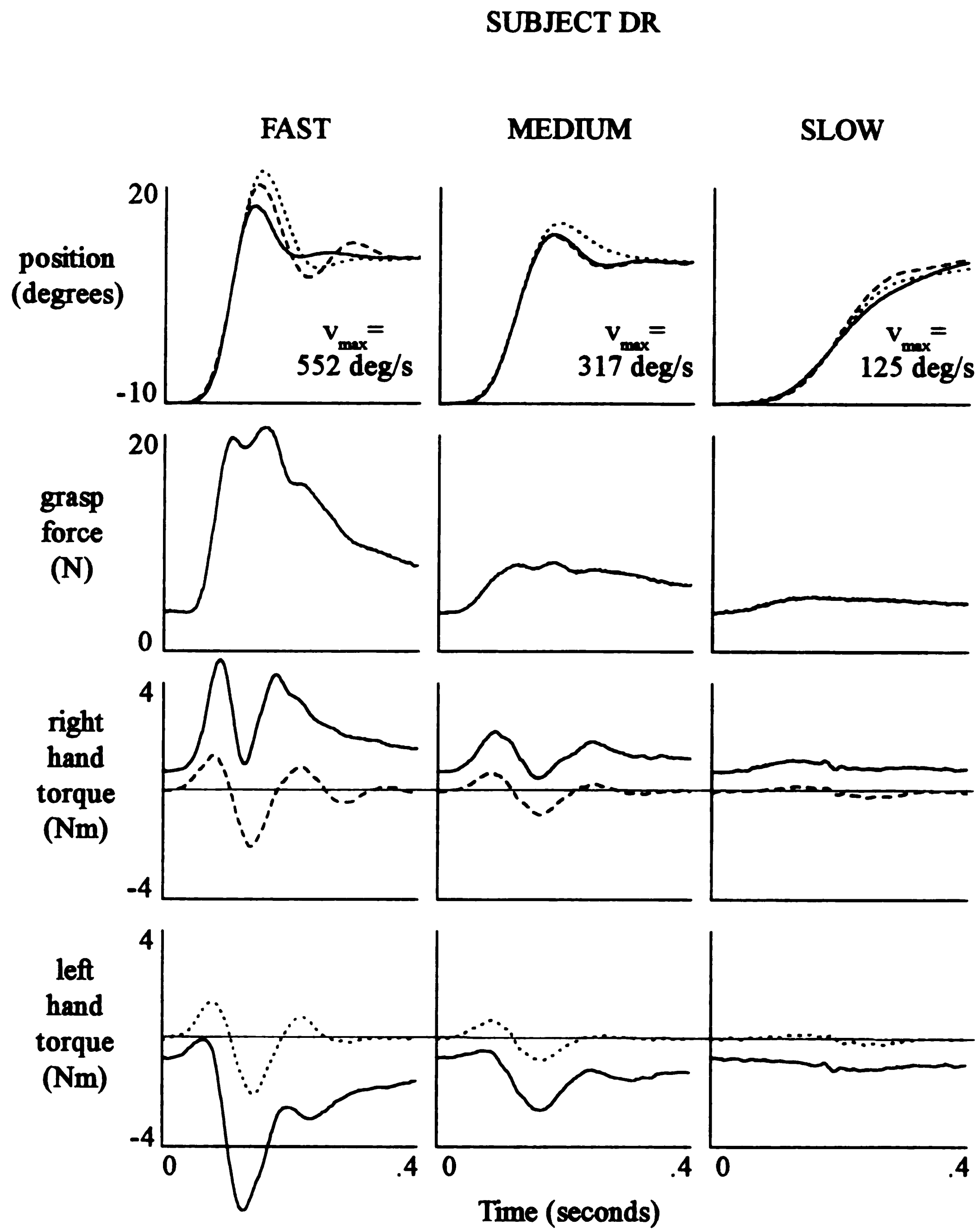

Fig. 8. Forces and torques for bimanual (solid), single hand right (dashed) and single hand left (dotted) movements for subject DR. All trajectories are ensemble averages of 50 trials.

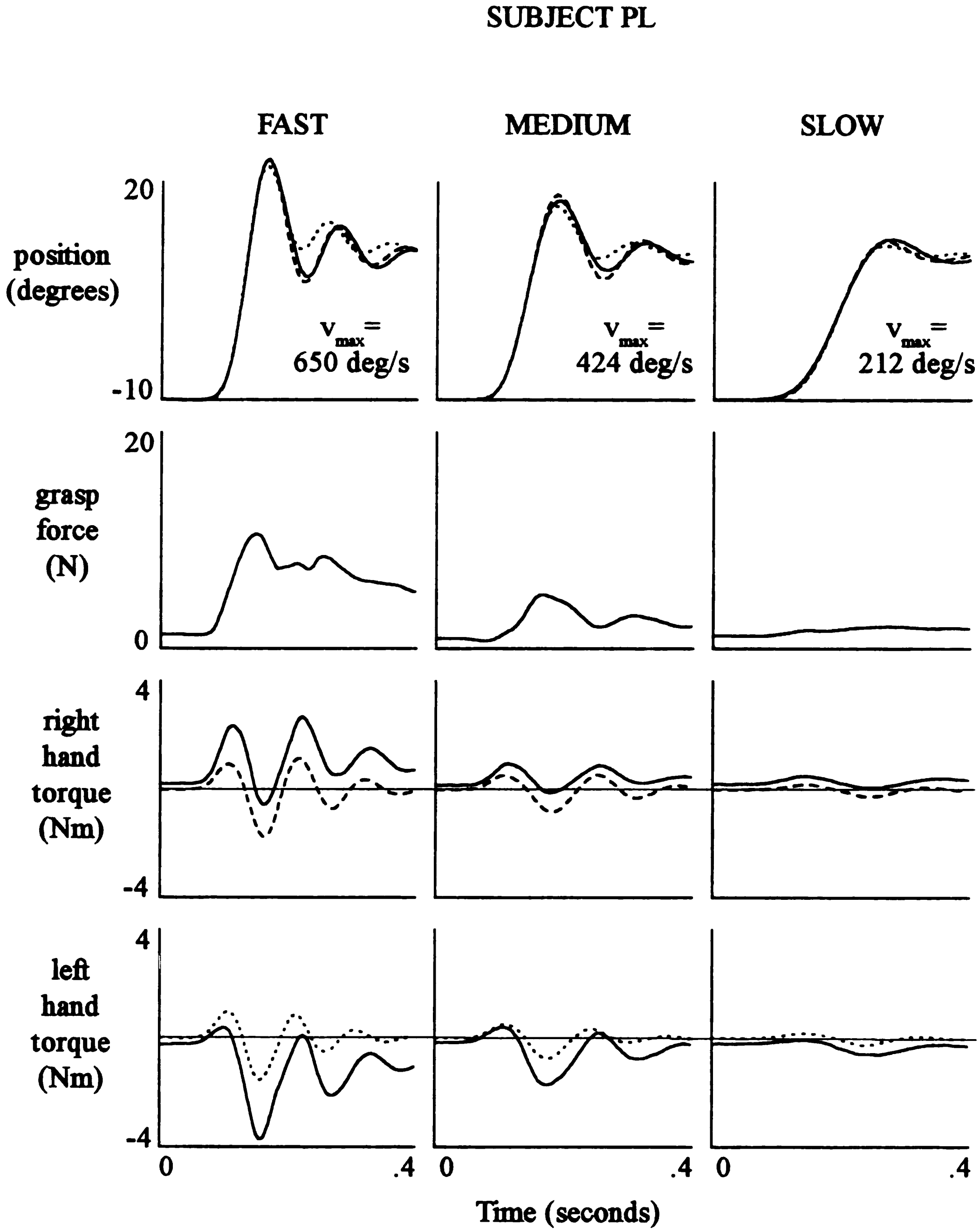

Fig. 9. Forces and torques for bimanual (solid), single hand right (dashed) and single hand left (dotted) movements for subject PL. All trajectories are ensemble averages of 50 trials.

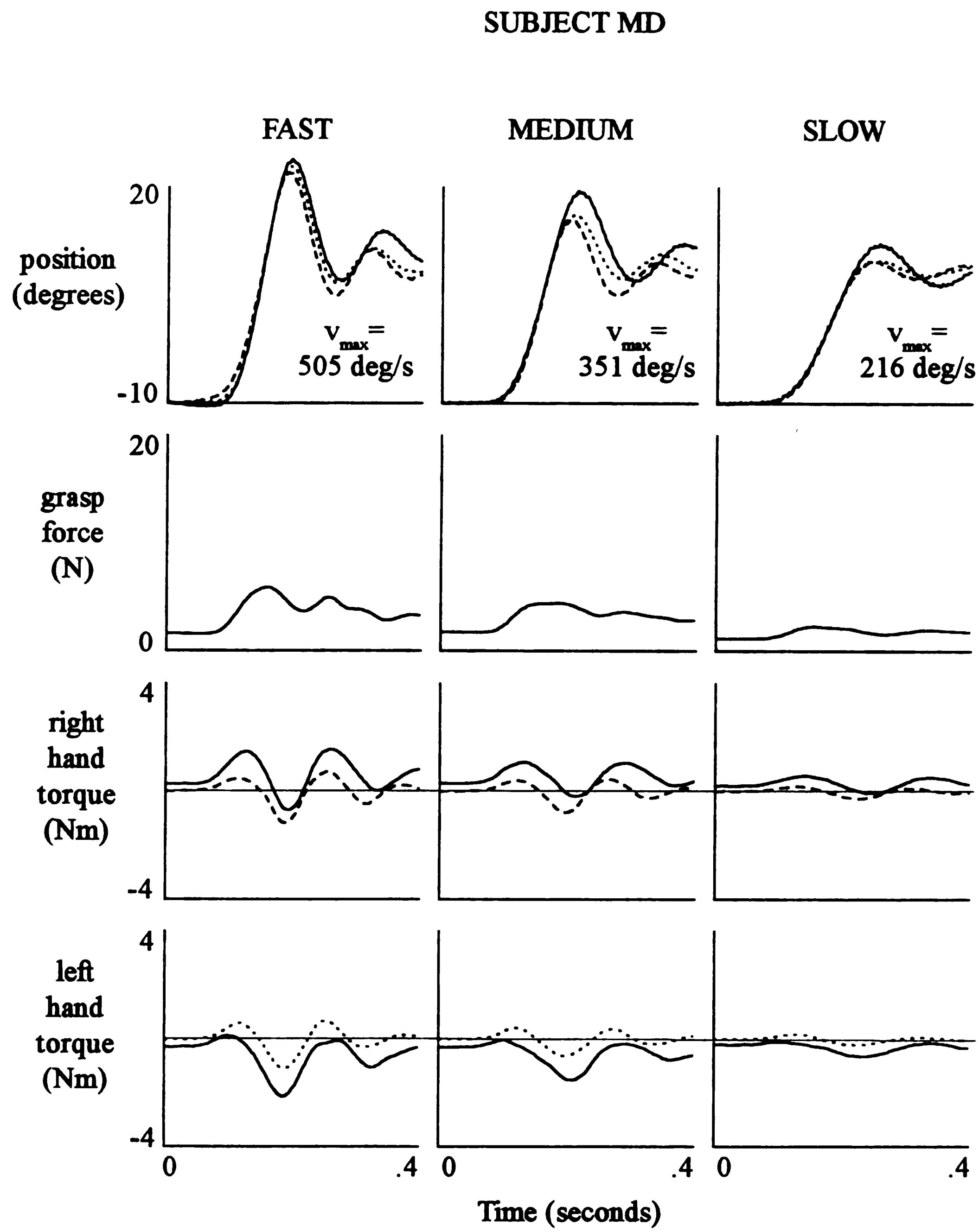

Fig. 10. Forces and torques for bimanual (solid), single hand right (dashed) and single hand left (dotted) movements for subject MD. All trajectories are ensemble averages of 25 trials.

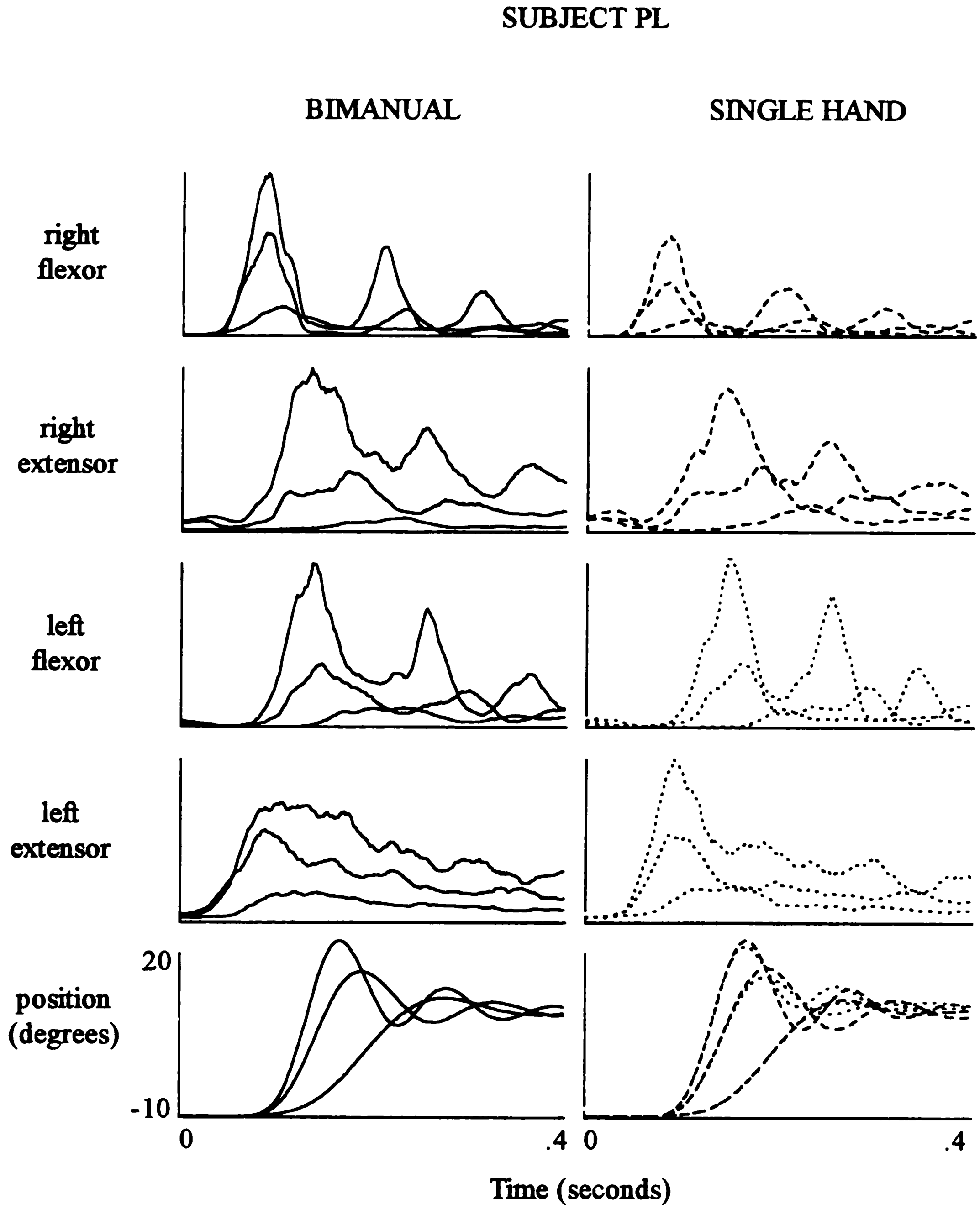

Fig. 11. *left column*: overlay of bimanual emgs at three speeds for subject PL. *right column*: overlay of single hand emgs at three speeds for subject PL. All trajectories are ensemble averages of 50 trials.

agonist and antagonist and advanced onset of the antagonist. The emgs in each hand in the bimanual task showed this same SS strategy. The SS strategy was also recognizable in the inertial torque trajectories of the single hand and bimanual movements (fig. 12). The trajectories diverged immediately and the initial zero crossing was advanced with increasing speed. This is characteristic of a SS movement strategy [15].

Braking strategies used in "fast as possible" single hand movements generalize to this bimanual task. Fig. 13 shows the isometric torques after single hand stopped movements at three places during the breaking phase of the movement. The emgs showed no change from the normal unperturbed pattern until 50 ms after the stop, so that the first 90 ms of the isometric torque pattern after the stop wasn't influenced by reflex activity triggered by the stopping action (fig. 13a). The stopped hand produced a braking torque proportional to stopped position. Stops later in the movement resulted in larger negative isometric breaking torques (fig. 13b). This has been interpreted as springlike behavior in the antagonist muscle [17]. This type of springlike braking was also present in bimanual movements (fig. 14). Stops later in the movement resulted in larger negative breaking torques from the left hand.

Discussion

All of the results confirm our general hypothesis that these bimanual movements are controlled as single hand movements. However, questions remain as to why and how this is true. In the following, we address three questions: 1) Why are the bimanual trajectories identical to single hand trajectories up to peak overshoot? 2) Why do the

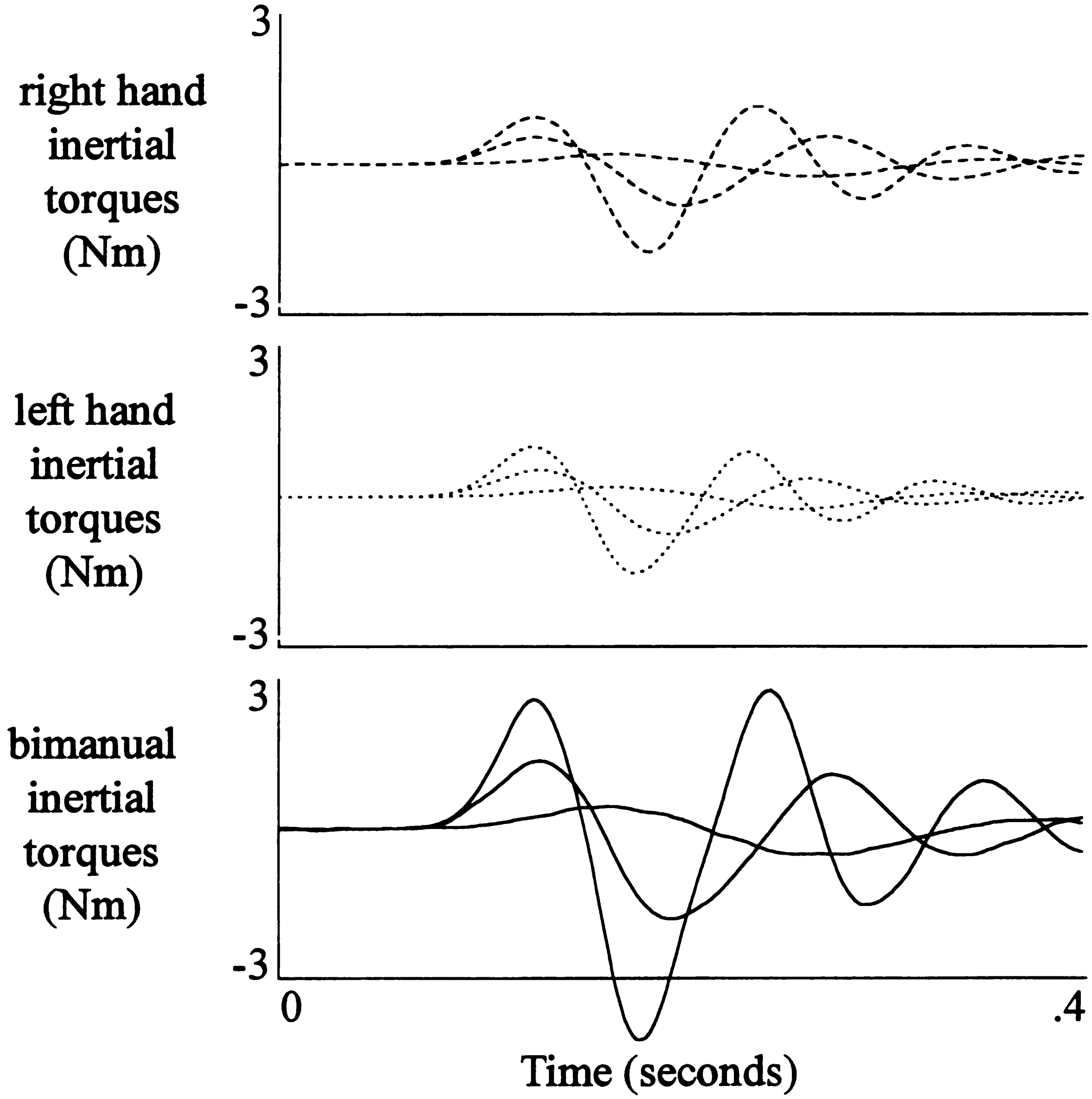

Fig. 12. Inertial torques at three speeds for right hand movements, left hand movements and bimanual movements for subject PL. All trajectories are ensemble averages of 50 trials.

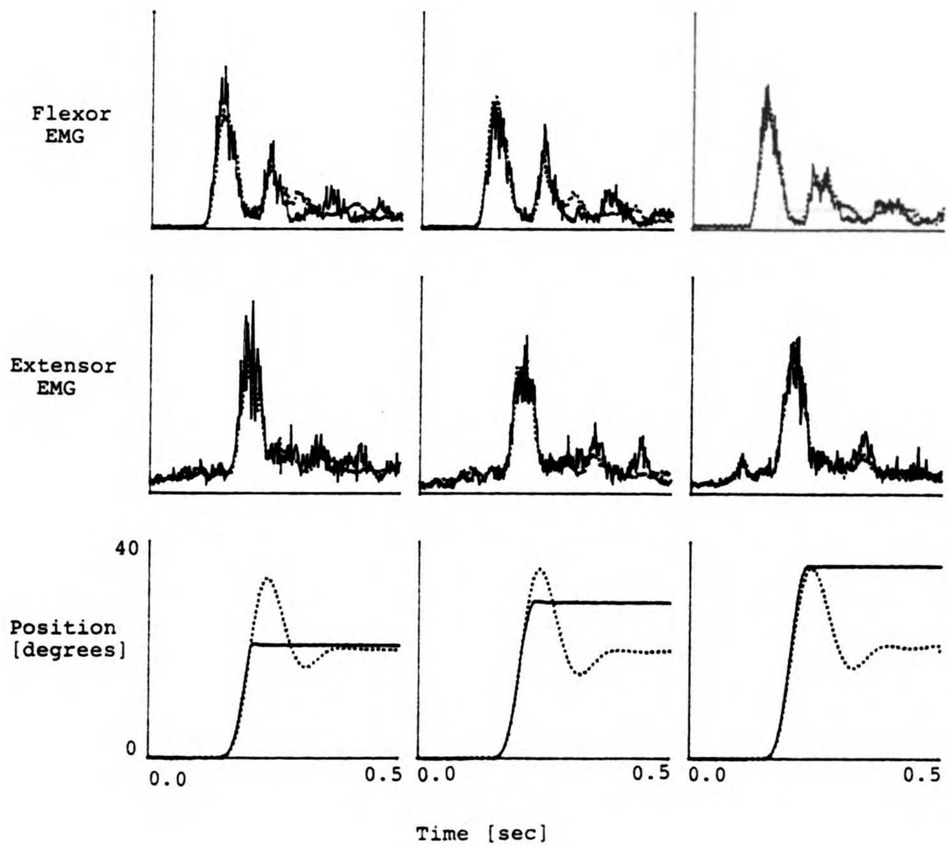

Fig. 13a. Emg and position for movements of one subject stopped near intended position (left column), near maximum overshoot (right column), and halfway between (middle column). For stopped trials, ensemble averages are of 13-20 trials. For unstopped trials, ensemble averages are of 82-85 trials.

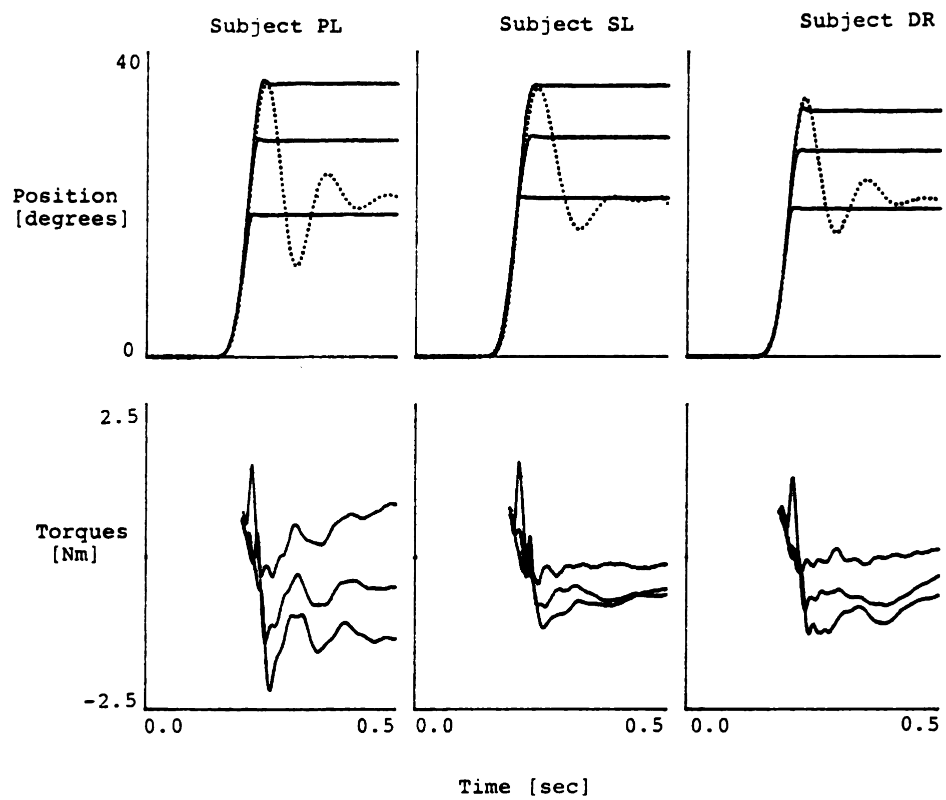

Fig. 13b. *1st row*: Unperturbed "fast as possible" movements (dotted) and movements stopped at three positions for three subjects. *2nd row*: Isometric torques after stopped movements. Lower torque traces correspond to later stops. Each trajectory is the average of 13-20 trials.

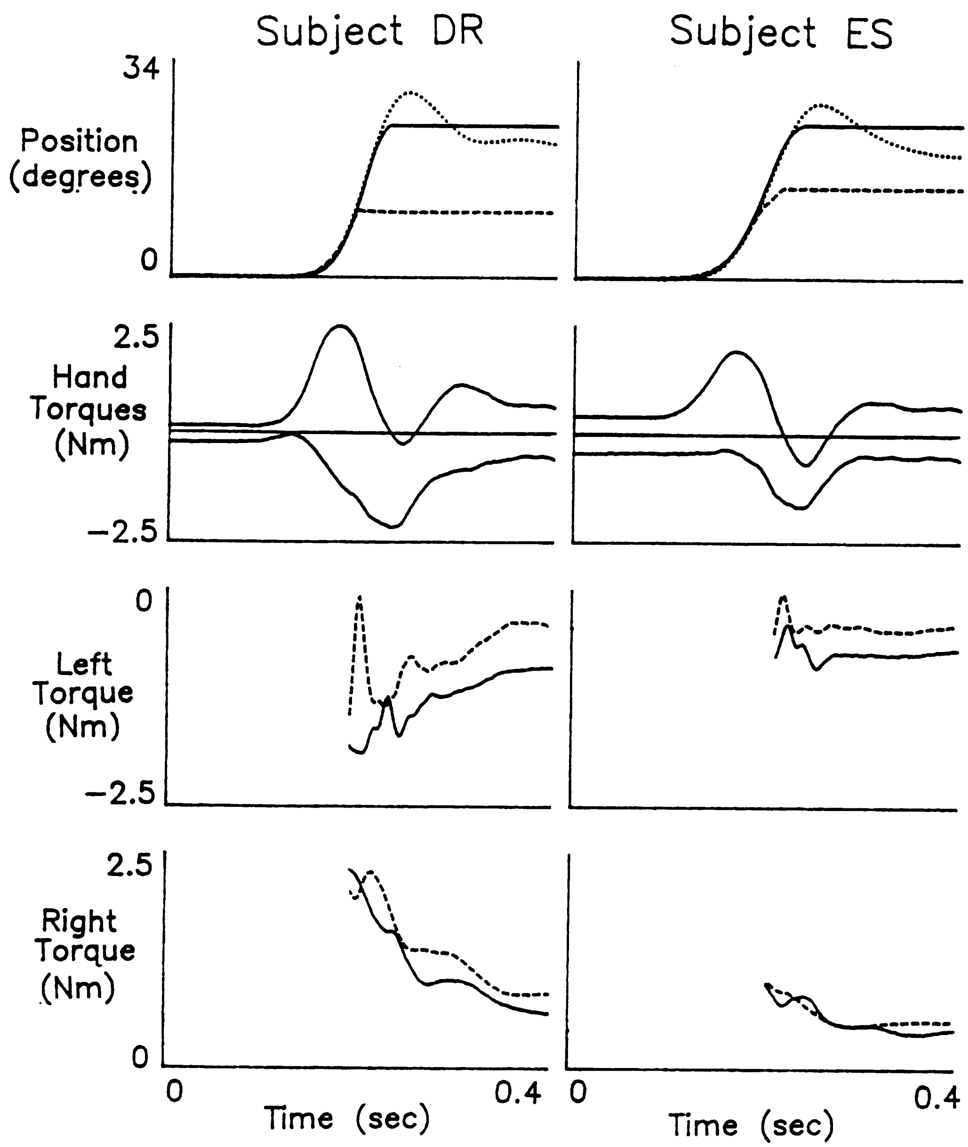

Fig. 14. Stopped bimanual movements for 2 subjects. All traces are ensemble averages. *1st panel*: position traces for unperturbed (dotted), early stopped (dashed) and late stopped (solid) movements. *2nd panel*: Right hand torque (top trace) and left hand torque (bottom trace) for unperturbed movements. *3rd panel*: Left hand isometric torque for stopped movements. *4th panel*: Right hand isometric torque for stopped movements.

trajectories diverge after peak overshoot? 3) How do the emg patterns used in bimanual movements produce single hand trajectories up to peak overshoot?

The movement instruction only specified the amplitude of the movement and the peak velocity; yet the bimanual and single hand trajectories were statistically identical up to peak overshoot. The fact that the trajectories were identical for such a large section of the movements indicates a choice made by the human control system. It has been shown that the dynamics of the H-O-H system are a scaled version of single hand dynamics [4]. The fact that bimanual movements were identical to single hand movements up to peak overshoot shows that the dynamics of the system dictate the control chosen by the CNS.

The question remains as to the why the bimanual kinematics diverged from single hand kinematics after peak overshoot. The right and left hand movements also diverged after peak overshoot. This indicates inherent differences in the control of the braking phase of right hand flexions vs. left hand extensions. This might be related to the fact that braking is a combination of feedforward and feedback control. Wadman et al. has shown that in stopped single hand movements, the first agonist burst is present unaltered while the second burst is also present but smaller in amplitude [11]. They conclude that the first agonist burst is totally feedforward; while the second burst is also feedforward, but facilitated by peripheral feedback. It may be that the CNS has less control of the braking phase due to this type of braking mechanism. This might explain why movement variability was largest in the braking phase for both bimanual and single hand movements. In contrast, the acceleration phase was totally under feedforward control and was kinematically identical for left hand extension, right hand flexion or the bimanual task.

How do the emg patterns used in bimanual movements produce single hand trajectories up to peak overshoot? We measured a 40 ms delay from a change in emg to a change in the kinematics. If one counts back 40 ms from where the trajectories diverge, one sees that at this point, the first RF and LE bursts were completed and the RE & LF emgs were rising. The emgs up to this point are responsible for the identical sections of the bimanual and single hand trajectories. The larger first RF burst produced a larger initial right hand torque peak while the smaller LE burst produced a smaller left hand initial torque peak. This distribution of torque between the left and right hands was exquisitely matched to accelerate the system as in single hand movements while maintaining the grasp force above a critical level. System breaking was not controlled by changing the amplitude of the antagonist emgs, but by advancing the onset time of the LF. As a result, the LF generated braking force before the RE, which guaranteed that the grasp force remained above the critical level, yet with the same kinematics as single hand braking up to peak overshoot.

How does the left hand generate so much braking torque with no larger EMG amplitude? Inspection of figs. 8,9,10 shows that the left hand generated more than twice the peak negative braking torque during bimanual movements vs. single hand movements. At the muscle level, it is unclear whether this results from less positive torque from the LE and/or more negative torque from the LF. Subject PL's middle speed movements are evidence that the LF is generating more braking torque than single hand movements. The

LE emg was actually slightly larger than that generated in left hand movements. The torque profile reflected this. The first positive peak in left hand torque in his bimanual medium speed movements approached the peak in the single hand torque profile. However, the negative peak in the bimanual movements was more than two times larger than the single hand movements. Kinematics were identical up to this point and the only obvious difference in the LF emgs was a 10 ms onset advancement. The explanation may be the spring-like braking hypothesis supported by the stopped movements. The 10 ms advancement of the LF emg resulted in earlier grounding of the spring and consequently at peak overshoot, even though kinematically the left hand was at the same location as single hand movements, the LF had been stretched further in the bimanual case and consequently generated more braking torque.

III. Rhythmic movements

Introduction

We studied rhythmic movements of the H-O-H system and compared them to single hand rhythmic movements at the same frequency. Coordinated rhythmic movements of unconnected hands have been studied in detail by Kelso [18,19]. In his studies, subjects were asked to move the hands rhythmically together in asymmetric mode, one hand flexing while the other extends. When the subject is asked to increase the cycling frequency, a phase transition occurs from the asymmetric mode to a symmetric mode with the two hands flexing at the same time, at a critical frequency. The transition occurs abruptly at around 2 Hz. This transition is analogous to gait transitions in locomotion when the speed of locomotion is increased. Locomotion is mediated by central pattern generators (CPGs) in the spinal cord. Kelso's data support the hypothesis that rhythmic movements of the hands are also mediated by CPGs.

In our bimanual task, the hands must be moved rhythmically in asymmetric mode at 1 to 4 Hz, with the added task requirement of holding an object between the two hands. We investigated whether the control of single hand rhythmic movements is adapted to perform this bimanual task.

Methods

The subject placed his hands in the H-O-H apparatus described above and moved the object back and forth rhythmically at a desired frequency. Subjects were presented hand position on an oscilloscope along with two targets 20 degrees apart. One target was located 10 degrees cw of the center line of the H-O-H system. The other target was located 10 degrees ccw of the center line. Subjects were asked to move rhythmically at a desired frequency while using the targets as a limit on the minimum size of the movements. The computer generated an audible click each time hand position passed the center line in the ccw direction to help subjects move consistently at the desired frequency. The experimenter corrected the subject's frequency until the subject was close enough to the desired frequency. Trials lasted 5 seconds. After each trial, a histogram of movement periods for that trial was presented to the experimenter. Only trials whose period histogram fell predominantly at the desired frequency were recorded. The desired frequencies were 1,2,3 and 4 Hz. Approximately 60 movement cycles were recorded at each frequency. One additional trial was recorded under the "fast as possible" instruction. Subjects then performed the same series of movements with just the right and then just the left hand. Two subjects were tested.

A single cycle was determined to occur between velocity zero crossings that mark the rightmost position of the system. Cycles were synchronized on the velocity zero crossing that marks the left most position of the system and ensemble averaged. Emgs were rectified, ensemble averaged and low pass filtered with a 14 ms moving average filter.

Results

We have three results that support our general hypothesis. 1) The statistical difference between bimanual movements and single hand movements is smaller than the difference between left and right hand movements. 2) In general, the emg patterns used to move the H-O-H system are alternating bursts in the flexor/extensor of each hand. These alternating bursts are similar to the emg patterns used in one hand free movements. 3) Bimanual emgs are modified from single hand emgs so that one hand acts as "agonist" and the other hand acts as "antagonist" for the H-O-H system.

The statistical difference between bimanual movements and single hand movements is smaller than the difference between left and right hand movements. The first row of figs. 15 & 16 overlay the bimanual and single hand movements at 4 frequencies for two subjects. A pointwise t-test was done between bimanual and single

SUBJECT PL

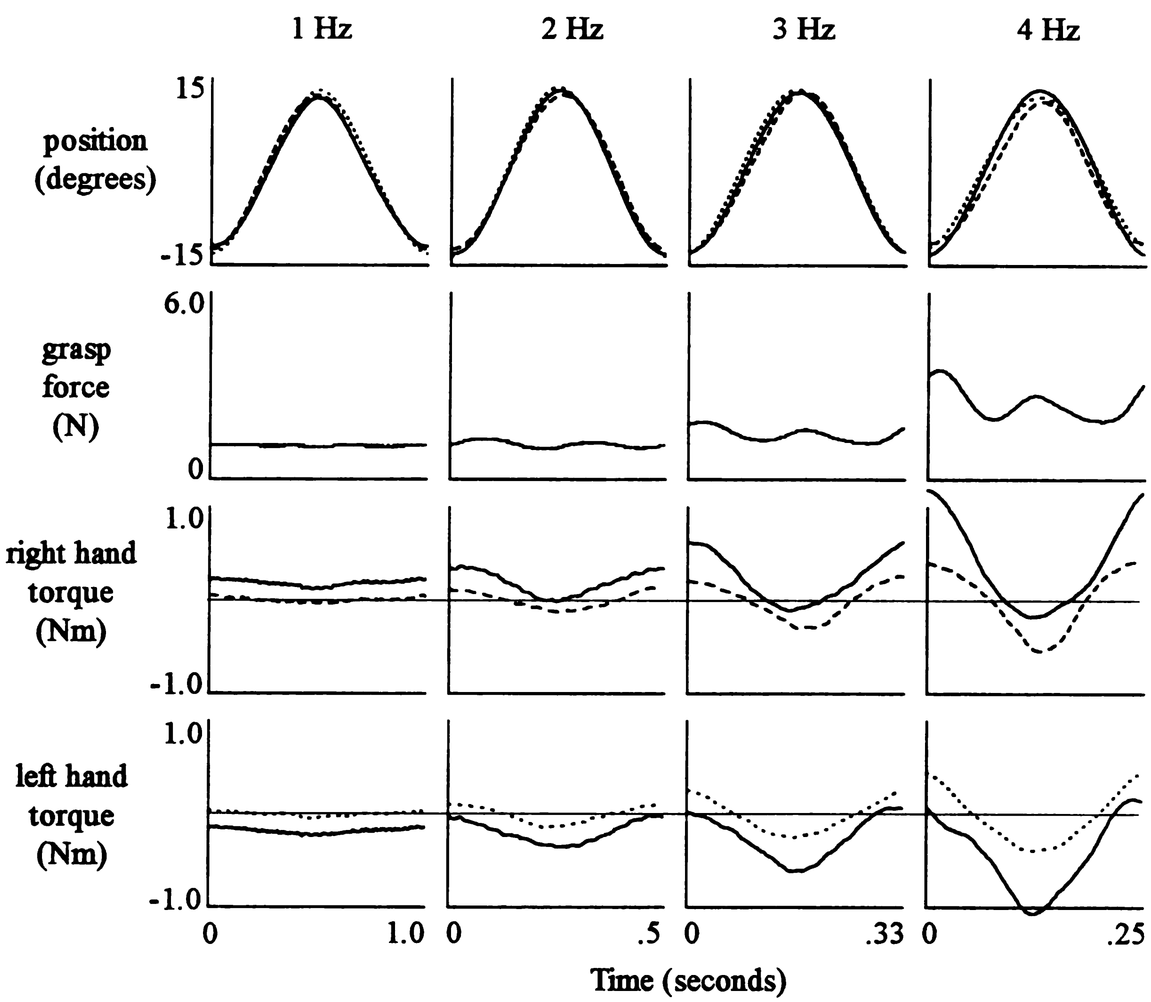

Fig. 15. Forces and torques for bimanual (solid), single hand right (dashed) and single hand left (dotted) rhythmic movements for subject PL. All trajectories are ensemble averages of 50-60 cycles.

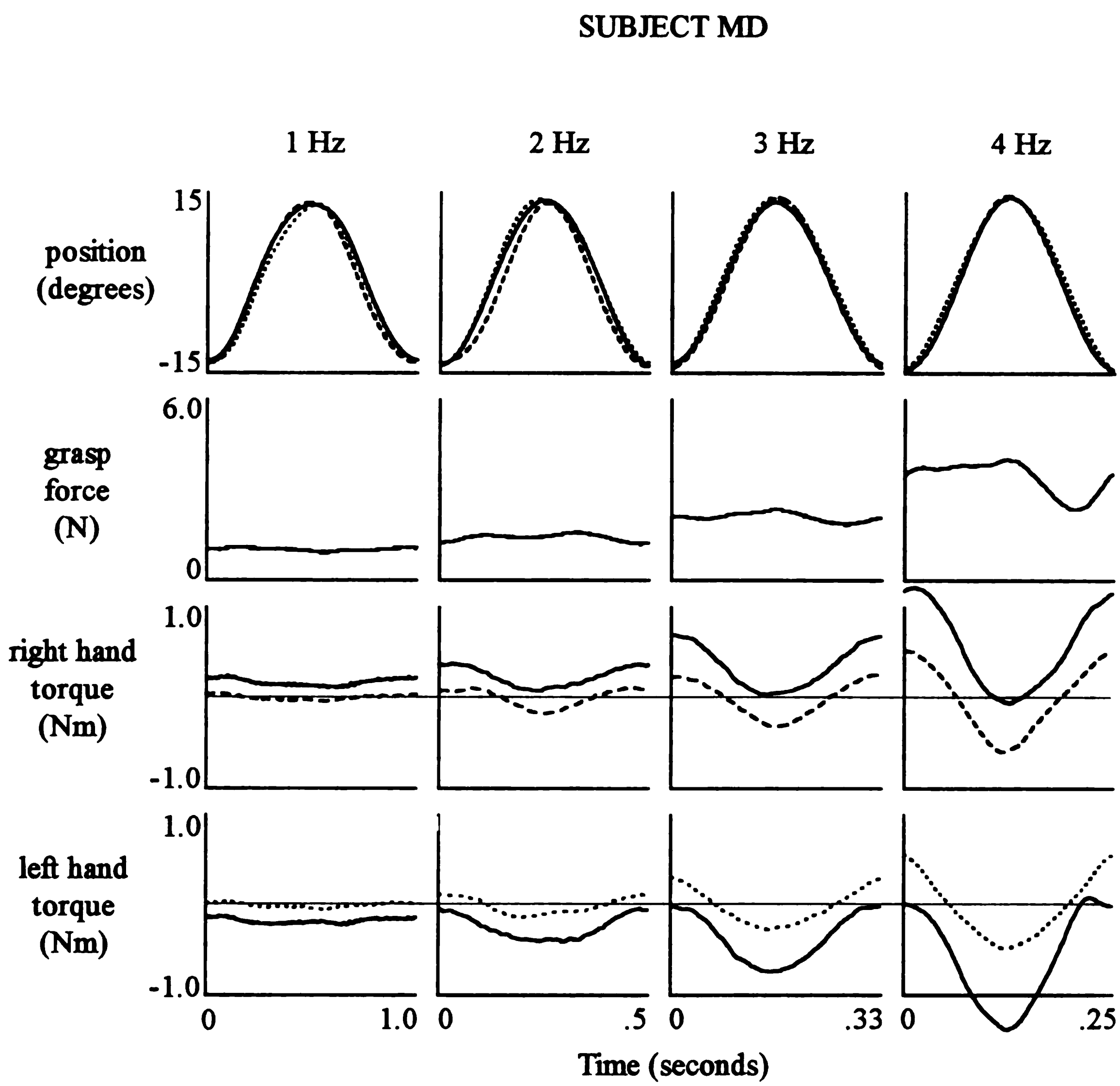

Fig. 16. Forces and torques for bimanual (solid), single hand right (dashed) and single hand left (dotted) rhythmic movements for subject MD. All trajectories are ensemble averages of 50-60 cycles.

hand movements, and between left and right hand movements at the same frequency. In many cases, the trajectories had points that were statistically different. We used the average t value over the cycle as a measure of the difference in two trajectories. Tables 1 and 2 summarize these t values. We averaged the first 2 rows of each table to get a measure of the statistical difference between bimanual and single hand movements. We averaged the third row of each table to get a statistical measure of the difference between left and right hand movements. In both subjects, the statistical difference between the bimanual and single hand movements were slightly smaller than the difference between right and left hand movements.

Table 1. subject PL t-test values

<table>
<tr><th></th><th>1 Hz</th><th>2 Hz</th><th>3 Hz</th><th>4 Hz</th><th>averages</th></tr>
<tr><td>bimanual vs. right</td><td>2.06</td><td>3.63</td><td>2.88</td><td>6.58</td><td rowspan="2">3.15</td></tr>
<tr><td>bimanual vs. left</td><td>3.14</td><td>1.66</td><td>2.34</td><td>2.89</td></tr>
<tr><td>right vs. left</td><td>2.28</td><td>3.30</td><td>3.42</td><td>5.32</td><td>3.58</td></tr>
</table>

Table 2. subject MD t-test values

<table>
<tr><th></th><th>1 Hz</th><th>2 Hz</th><th>3 Hz</th><th>4 Hz</th><th>averages</th></tr>
<tr><td>bimanual vs. right</td><td>2.70</td><td>4.01</td><td>1.40</td><td>2.89</td><td rowspan="2">2.31</td></tr>
<tr><td>bimanual vs. left</td><td>1.62</td><td>0.99</td><td>2.43</td><td>2.42</td></tr>
<tr><td>right vs. left</td><td>2.04</td><td>3.37</td><td>2.19</td><td>2.15</td><td>2.44</td></tr>
</table>

In general, the emg patterns used to move the H-O-H system are alternating bursts in the flexor/extensor of each hand. These alternating bursts are similar to the emg patterns used in one hand free movements. Figs. 17 & 18 overlay the bimanual movement emgs with single hand movement emgs at four frequencies and for two subjects. For the most part, each hand showed alternating bursts of activity in its flexor and extensor in both bimanual and single hand movements. In a few cases at the lower frequencies, some muscles showed tonic activity or no activity at all. However, the bursts were apparent in all the muscles at the higher frequencies.

There are consistent differences in bimanual emgs when compared with single hand emgs. Both subjects showed a larger and wider bursts in RF at all four frequencies. Both subjects showed smaller RE bursts, but as the frequency of the movements increases, the size of the bursts approached the size of RE bursts in single hand movements. Both subjects showed wider bursts in the LF, but subject MD also showed larger amplitude bursts in LF. Both subjects showed less LE activity, but as the frequency of the movements increases, the LE burst size approached that of left hand

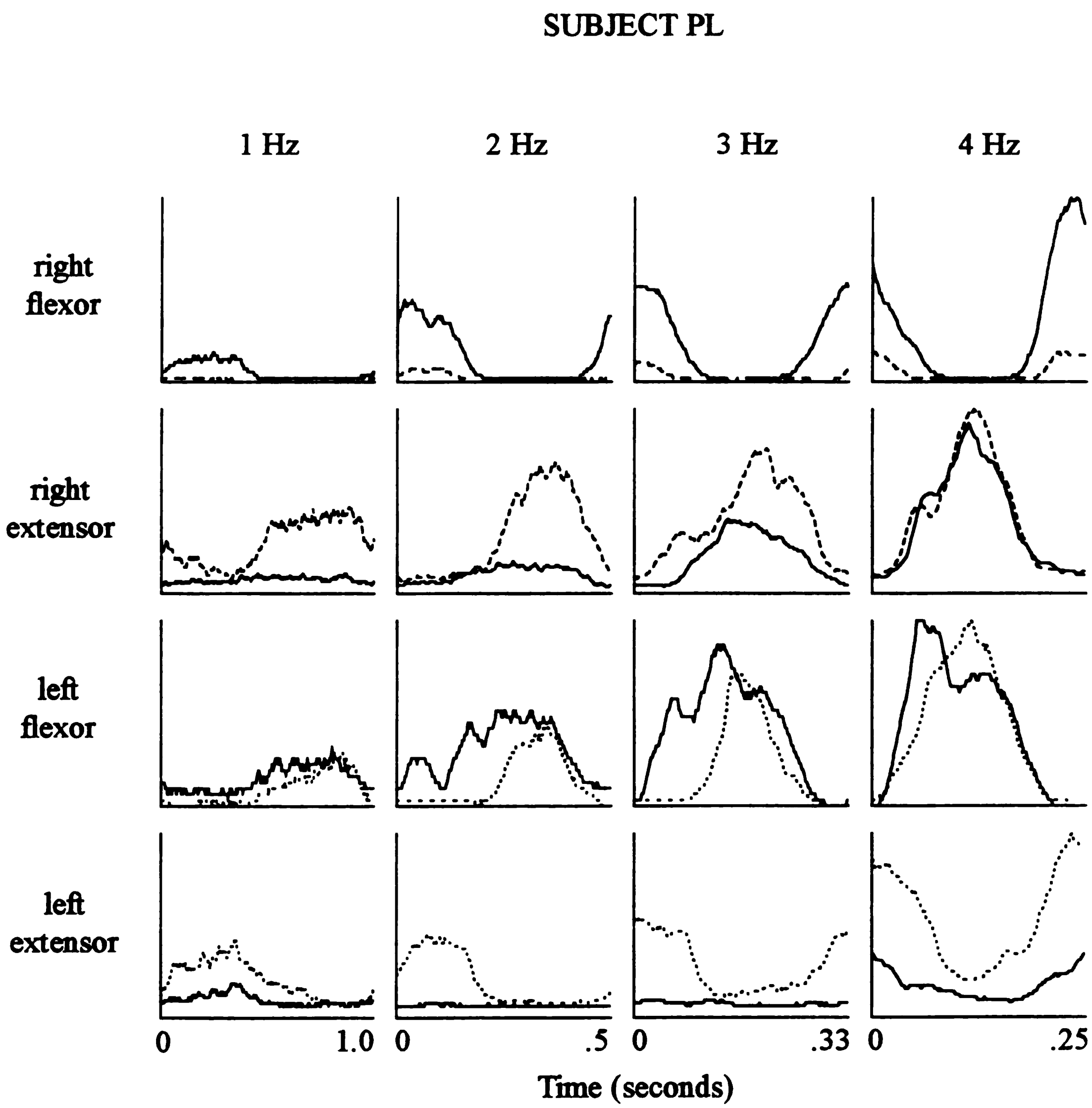

Fig. 17. Bimanual (solid) and single hand right (dashed) and single hand left (dotted) emgs at four frequencies for subject PL. All trajectories are ensemble averages of 50-60 cycles.

SUBJECT MD

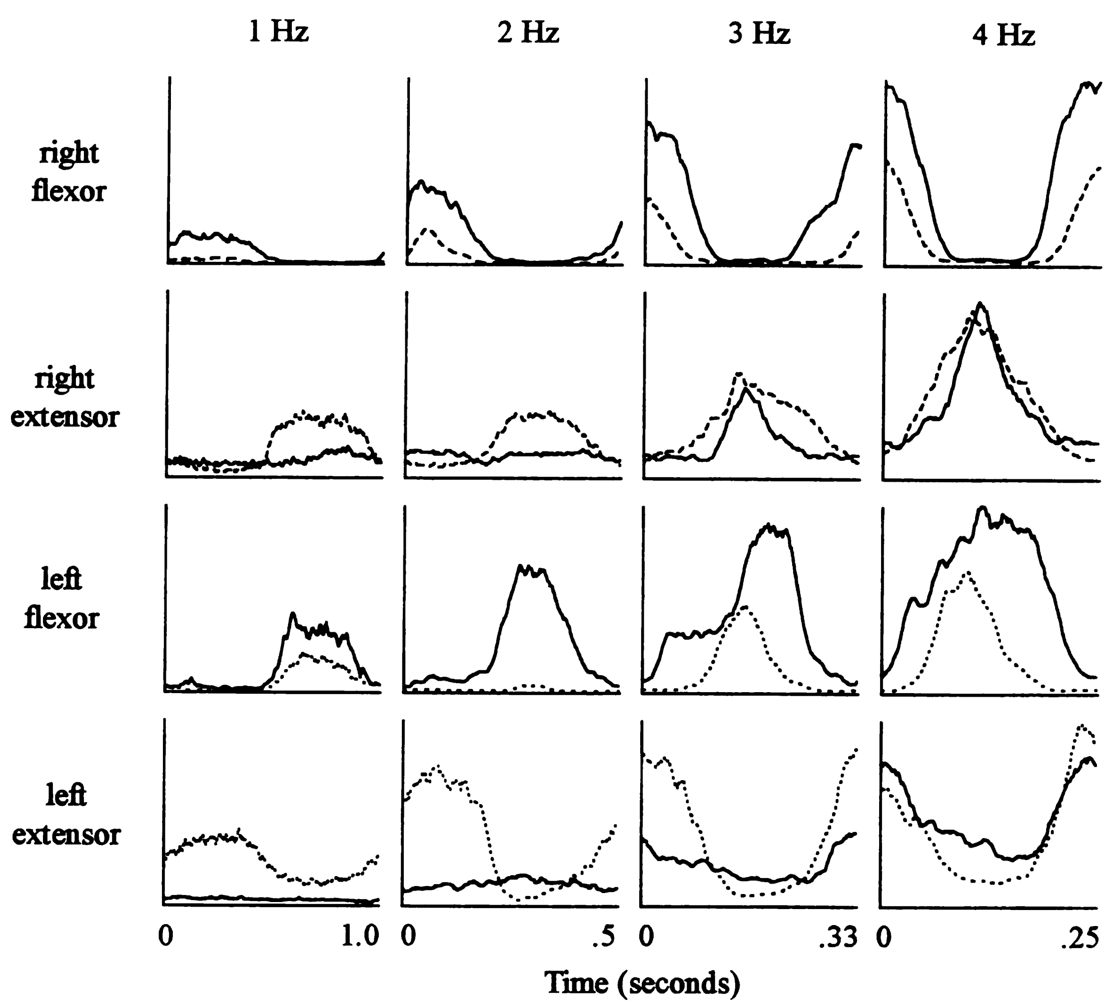

Fig. 18. Bimanual (solid) and single hand right (dashed) and single hand left (dotted) emgs at four frequencies for subject MD. All trajectories are ensemble averages of 50-60 cycles.

movements in subject MD.

One hand acts as "agonist" and the other hand acts as "antagonist". Figs. 15 & 16 show the torques produced from each hand in the bimanual and single hand movements for two subjects. The right hand produced the positive torque needed to move the H-O-H system while the left hand produced all the negative torque. This is in contrast to single hand movements where the hands must produce both positive and negative torque from its agonist/antagonist muscle pair. Thus the right and left hands act as an agonist/antagonist muscle pair for the H-O-H system.

Discussion

Our results support our general hypothesis that these bimanual movements are controlled as single hand movements. We address five questions: 1) Why are the bimanual trajectories on average significantly different from single hand trajectories. 2) How do the emg patterns seen in bimanual movements allow the hands to act as an agonist/antagonist pair? 3) Why do extensor emgs in many cases approach those of single hand movements at higher frequencies? 4) Why can subjects perform rhythmic movements at higher than 2 Hz, when a phase transition would normally occur from asymmetric to symmetric mode if the object were not being held between the hands? 5) Why are the single hand "as fast as possible" rates significantly higher than bimanual rates?

Why are the bimanual trajectories on average significantly different from single hand trajectories ($p<0.05$)? In an attempt to make the movements as "natural" as possible, we only specified the frequency and the minimum amplitude of the movements in the instructions to subjects. Therefore, due to handedness, it was not surprising that the right and left hand trajectories were on average significantly different. Since the bimanual movements use both hands, we would not expect the bimanual trajectories to be statistically identical to the single hand movements as was the case in the discrete movements. However, the fact that the difference in bimanual and single hand movements was on average smaller than the difference between right and left hand movements supports our general hypothesis that bimanual movements are controlled as single hand movements.

How do the emg patterns seen in bimanual movements allow the hands to act as an agonist/antagonist pair? The activity of the flexor/extensor pair of each hand was modified from single hand emgs to generate significant torque only in one direction. For the right hand, the torque profile was biased in the positive direction by making the RF emg burst larger and wider while making the RE emg burst smaller. The two subjects used their left hands slightly differently but with the same result of biasing the left hand torque in the negative direction. Subject MD did this by increasing the LF burst amplitude and width while decreasing LE activity. Subject PL seemed to only increase the LF burst width while decreasing LE activity.

Why do extensor emgs in many cases approach those of single hand movements at higher frequencies? Although the extensor emgs for both subjects were in general smaller than in single hand activity, the activity increased with increasing frequency, and at 4 Hz was about the same as for single hand activity for subject MD's RE & LE and subject PL's RE. Extensor activity may be needed to properly shape the torque profile from each hand at high frequencies. It is desirable to drop one hand torque to zero when the other hand is generating peak torque in the other direction, to avoid wasted effort and excessive squeezing of the object. At the higher frequencies, the hand torque must be brought down to zero much more quickly than the flexor can relax. Therefore, the activity of the extensor was increased at the higher frequencies to bring the hand torque to zero as the other hand torque was peaking. At the lower frequencies, the hand torque can be brought down to zero simply by relaxing the flexor. The only extensor emg that stayed much smaller than the single hand emg was subject PL's LE. This is explained when one considers that his LF activity was only slightly larger than in the single hand case. His left hand torque was biased in the negative direction mainly by keeping the LE activity small even at high frequencies.

Why can subjects perform rhythmic movements at higher than 2 Hz, when a phase transition would normally occur from asymmetric to symmetric mode if the object were not being held between the hands? If our general hypothesis that these bimanual movements are controlled as single hand movements is true, the problem is reduced from moving two hands together asymmetrically to moving one "hand" (the H-O-H system). Then it is logical to expect the maximum H-O-H movement speed to approach that of single hand movements. The "fast as possible" instruction yielded right hand, left hand and bimanual rates of 6.33,6.85 and 4.95 Hz for subject PL and 6.67,6.94 and 4.26 Hz for subject MD.

Why are the single hand "fast as possible" rates significantly higher than bimanual rates? This can also be attributed to the strategy of using the two hands together as one hand. This strategy requires the right hand to produce only positive torque and the left hand to produce only negative torque. Therefor, the right hand must generate enough torque to move the entire H-O-H system in the ccw direction, while the left has to generate enough negative torque to move the whole system cw. The right hand positive torque peak in bimanual movements is more than twice the positive peak in single hand movements. Similarly, the left hand negative torque peak in bimanual movements is more than twice the peak negative torque in single hand movements. This movement strategy prevents the H-O-H system from using all of the muscle power at its disposal; extensors are not used to their full potential. Consequently, it is not surprising that the "fast as possible" bimanual movements are significantly slower than single hand movements.

IV. Summary

All of our results support the general hypothesis that the two hands in the H-O-H system are controlled like one hand in discrete and rhythmic movements. Bimanual emgs are modified from single hand emgs so that one hand generates appreciable torque in only one direction, while the other hand generates the required torque in the other direction. Therefore, one hand acts as "agonist" and the other hand acts as "antagonist" for the H-O-H system. This type of bimanual control strategy results in bimanual kinematics that are similar to single hand kinematics.

All of the evidence supporting this general hypothesis may reflect a coordinative structure between the hands that exists to mediate two hand grasp. There is evidence that coordinative structures exist in bimanual movements; the two hands cannot be moved independently of each other if there is no mechanical connection between the hands. Fowler et al. [20] demonstrated synchronization of the hands in bimanual aiming movements when the targets for each hand differ in index of difficulty. Kelso has shown a transition from asymmetric rhythmic wrist movements to symmetric wrist movements at high frequencies [18,19]. It is tantalizing to conjecture that these coordinative structures exist to mediate bimanual tasks where the two hands manipulate an object. If so, these coordinative structures "interfere" in similar bimanual movements where the mechanical connection between the hands is removed. For example, subjects can move the hands rhythmically asymmetrically at over 4 Hz if an object is held between the fingertips, but can only move up to 2 Hz if the hands are not interconnected.

Our long term goal is to design robotic devices to assist a disabled limb in the performance of bimanual tasks in order to rehabilitate bimanual coordination. One of the primary difficulties of assisting a disabled limb is that the desired movement or force from the limb must be known. However, this problem is solved elegantly by the dynamic simplicity of the H-O-H system. Assume for a moment that we wish to design an assistive device for a disabled left hand in performing H-O-H tasks with an able right hand. Reexamination of the dynamic equations governing the H-O-H system shows that the required left hand torque (τ_l) is determined by the right hand torque (τ_r) and the grasp force (F).

$$\tau_l = \tau_r - 2rF\cos\theta$$

The cos(Θ) term can be approximated by unity for movements near the center position of the hands and r is a system constant. Assuming the right hand performs the task normally (normal τ_r trajectory), the device only needs to generate assisting torques for the left hand so that the grasp force trajectory resembles trajectories from normal subjects. An assistive device that works on this principle is guaranteed to produce normal left hand torque trajectories if the right hand trajectories are normal. Normal right and left hand torque trajectories guarantee normal kinematics.

$$\ddot{\theta} = \frac{\tau_r + \tau_l}{J_r + J_l + Mr^2}$$

J_l and J_r are the left and right hand inertias and M is the object mass.

The normal grasp force trajectory is determined by performance of the task by normal subjects. Grasp force trajectories in fast movements of the H-O-H system are complicated, but the force trajectories for slow movements may be approximated by a constant value (slow movements in figs. 8,9 & 10 ; 1 and 2 Hz movements in figs. 15,16). In the next chapter, we tested if an assistive device operating on such a principle could assist a disabled left hand in rhythmic movements of the H-O-H system at 1 and 2 Hz.

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Chapter 4

Robotic Assist Devices for Bimanual Physical Therapy : Preliminary Experiments[1]

Abstract

We present preliminary experiments in the design of robotic assist devices for bimanual physical therapy. Suitably designed devices could aid in the rehabilitation of people, such as hemiplegic stroke patients, with a disability affecting one hand but not the other. We performed preliminary experiments on a dynamically simplified system consisting of two outstretched hands constrained to flexion/extension while holding an object. A prototype device was designed to measure and assist in a horizontal transport task, in which the two hands moved a pencil-like object rhythmically back and forth. The device, operating under a simple proportional feedback control law, could substitute completely for one hand in the task. When the device assisted a hand partially disabled by ischemia, it produced torque that the disabled hand did not, allowing the task to be performed. These results demonstrate the capability of a device, operating under a simple control law, to assist a disabled hand, allowing performance of a coordinated bimanual task. Such a robot, extended to more useful bimanual activities of daily living,

[1] Portions of this chapter are currently in press under the same name in IEEE Transactions on Rehabilitation Engineering.

could afford cost effective, objectively measurable, self-driven therapy for a large patient population.

Introduction

We believe robotic devices could be used by therapists in conjunction with traditional physical therapy methods. These devices could improve the availability and cost-effectiveness of therapy by automating many of its repetitive aspects. Robotic aids also promise to enhance the capabilities of therapists by objectively measuring patient progress, thus providing a platform for evaluating specific therapy methods. We present in this chapter preliminary experiments in the design of these devices.

Examples of the use of robots in physical therapy are few. Continuous passive motion (CPM) machines can move limbs about repetitively, automating one aspect of physical therapy, but the present devices lack sophisticated programmability, adaptability to individual patients, and ability to measure patient performance or to generalize to more than one task. Khalili and Zomlefer [1] have suggested a more advanced CPM machine built around a two-joint robot that would allow repetitive motion to be programmed to suit the patient, and sensors on the robot which could be used for assessment. A prototype of such a device has not been reported. Two successful applications of robots in postural rehabilitation have been demonstrated for hemiplegic patients. Hocherman et al. [2] showed that the postural stability of hemiplegic patients could be improved with training on a moving platform. Godall et al. [3] used a single degree of freedom actuator with a force sensor to stabilize sway in hemiplegic patients, and proposed its use as a therapy technique. To our knowledge, no research has been done on a robotic aid for upper extremity rehabilitation.

There is a large population of patients whose disability is limited to one side of the body, such as hemiplegic stroke patients. The devices we propose to design would assist rehabilitation of these patients by helping to restore bimanual coordination when one hand has been disabled. There are several distinct advantages of bimanual therapy (see Chapter 1).

We studied a bimanual horizontal transport task which involved moving an object held between the two hands rhythmically back and forth. To limit the number of degrees-of-freedom, the hands were placed in two rigid handles that limited motion to wrist flexions/extensions (fig. 1). In previous work, we have characterized the control of this hand-object-hand (H-O-H) system during discrete and rhythmic movements in which both hands were able (see Chapter 3)[4,5].

We chose to study the H-O-H system because of its dynamic simplicity [4]. Control of this system requires bimanual coordination between the two wrist joints, but is constrained to a single degree-of-freedom. Also, some aspects of the control of free wrist movements are understood [6,7].

The transport task required control of the position trajectory of the H-O-H system while keeping grasp force approximately constant. Although it is unlikely this is a task that needs to be rehabilitated, we believe the experiments in this chapter to be the building blocks for the design of more complex robotic assist devices for rehabilitating

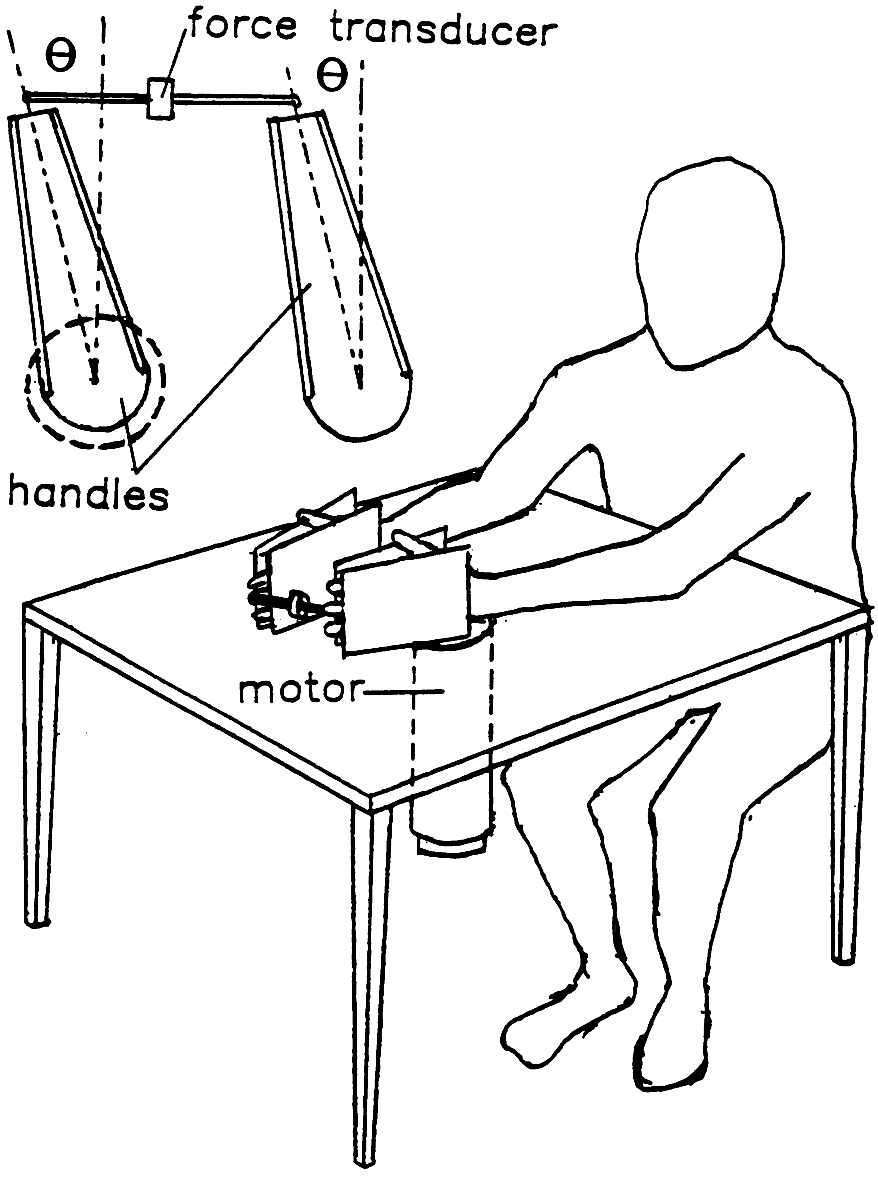

Fig. 1. Hand-object-hand system. Handles allow only wrist flexion/extension. The object is a force transducer extended with two machine screws. The object length is equal to the distance between the axes of rotation of the two handles. The robotic aid is a computer-controlled motor mounted under the left handle. Positive rotation and torque generation are defined to be in the counterclockwise direction.

more useful bimanual tasks such as opening a jar or medicine bottle or lifting a large box or pot with two hands.

In the present study, subjects first performed the tasks normally without assistance, then in conjunction with the robotic aid, which is a computer-controlled motor mounted under the handle of the left hand (fig. 1). To show that the robotic aid could substitute for the left hand, subjects removed their left hands from the left handle and performed the task with the right hand and robotic aid. To show the robotic aid could assist a partially disabled hand, the left forearm was progressively disabled by ischemia while subjects performed the task bimanually.

These experiments were designed to answer two basic questions. 1) Can the robot, operating under a simple feedback control law, substitute for one hand in this task? and 2) Can this simple control law allow the robot to aid a partially disabled hand, generating only that amount of torque that the disabled hand does not?

Methods

The H-O-H system has been previously described in detail in Chapter 3 (methods section). The only addition was a computer-controlled permanent magnet motor (Electrocraft model 0728-31-011) attached to the shaft of the left handle. This motor could apply up to 6 Nm torque to the left hand (4.1 ms mechanical time constant, 1.6 ms electrical time constant).

Three subjects performed the transport task. Subjects moved the pencil-like object back and forth rhythmically to audible ticks generated by the computer. No instructions were given concerning movement amplitudes to two subjects. One subject was fed back the position of the H-O-H system and minimum bounds for the movement on a CRT directly in front of him. Individual cycles of the rhythmic movements were defined to be between the zero crossings of velocity which marked the right-most position of the hands. Cycles were synchronized on the velocity zero crossing that marked the left most position of the hands and ensemble-averaged.

We have shown previously that a simple dynamics calculation determines the torques produced by each hand in these tasks [4] :

$$\tau_r+\tau_l=(J_r+J_l+Mr^2)\,\ddot{\theta}$$

$$F=\frac{\tau_r-\tau_l}{2r\cos\theta}$$

where τ_r is right hand torque, τ_l is left hand torque, F is the grasp force, J_r is right side inertia, J_l is left side inertia, M is the mass of the object, r is the distance from the wrist axis of rotation to the fingertip-object contact point, and Θ is the angular position of the H-O-H system. Parameters for this calculation were all obtained experimentally (for full description see methods section of Chapter 3). The left handle inertia was 0.0047 kg m^2, and the right handle inertia was 0.0049 kg m^2. The hand inertias for the three subjects ranged from 0.0031 to 0.0045 kg m^2. The friction in the motor shaft was identified by having the motor apply a sinusoidal torque to the left handle. The inertial forces were then calculated by twice differentiating the angular position trace of the left handle and multiplying this by the left handle inertia. The difference between the inertial forces and the input torque is equal to the frictional forces since no other significant forces are present in the system. A value of 0.1 Nm was obtained in each movement direction and found to be independent of the velocity of the handle, indicating that the friction can be classified as static friction.

To characterize normal coordination, three subjects performed 60 rhythmic transport movements to 2 Hz clicks. Subject SL also performed 50 transport movements to 1 Hz clicks. Position and grasp force were recorded. To obtain an average grasp force value, the grasp force was ensemble-averaged over the cycles recorded. This trace was then time averaged over the length of the cycle to obtain an average grasp force value for each subject.

To determine whether the robot could substitute for one hand in the performance of rhythmic transport, subjects removed their left hands from the left handle and the left handle torque was controlled by the motor operating under the feedback law:

$$\tau = G(F - F_c)$$

where τ is the torque applied by the motor, F is the measured grasp force, F_c is the command grasp force and was chosen to be the average grasp force value obtained by averaging normal transport as described above, and G is the feedback gain. We chose G to be as large as possible without causing instability due to sampling delay (0.55 Nm motor torque/N grasp force for subject SL, 0.42 Nm/N for subjects PL and ES). Position, grasp force and motor output were recorded for 60 2 Hz movements for three subjects and 50 1 Hz movements for subject SL.

Finally, we wanted to investigate whether the robot could assist a partially disabled hand in this task. A blood pressure cuff was placed around each subject's left arm, and the blood flow occluded (200 mm Hg). As ischemia progressed, the subject gradually lost sensory and motor function in the left forearm and hand. While uncomfortable, ischemia of 20-40 minutes is not dangerous and has been used in many investigations of the use of afference in movement control [8,9]. Although we have no evidence that ischemia simulates hemiplegia in stroke patients, we believe the device should be able to adapt to any type of disability. We used ischemia to simulate a *progressive* disability in order to test the ability of the device to adapt to changing patient performance. For two subjects, the following protocol was used: periodically after onset of occlusion, the subject performed a maximum isometric voluntary contraction (MVC) with the left flexors and 30 seconds of rhythmic transport movements at 2 Hz. Cycles were initiated every 3.5 minutes. The cycles were repeated until MVC was less than 5% of its initial value. Subject SL followed this same protocol but performed an additional 50 seconds of 1 Hz transport in each cycle. During all the ischemia experiments, the motor attached to the left handle operated under the feedback law described above in order to assist the left hand in the task. All protocols were approved by the Committee for the Protection of Human Subjects.

Results

Transport movements can be made with the robot substituting for one hand and operating under a grasp force regulation control law. The robot successfully substituted for the left hand for all three subjects (fig. 2). For all subjects, however, the movement amplitudes were smaller than when the subject performed the task bimanually and the grasp force was always lower than the command force (F_c). The torque patterns in fig. 2 show the decreased torque levels associated with the smaller movements after robot substitution. The general movement strategy was positive torque from the right hand pushing the whole H-O-H system in the counterclockwise (ccw) direction and negative torque from the left side (left hand or motor) pushing the whole system in the clockwise (cw) direction. A similar movement strategy was used by subjects in our previous studies of the H-O-H system (see Chapter 3)[4,5].

Rhythmic transport is also possible with the robot assisting a progressively disabled hand, with the robot operating under the same grasp force control law. The subjects' left hand was made ischemic as described in methods, causing a decrease in MVC for all subjects (fig.3). Despite this decrease, all three subjects were able to make rhythmic transport movements at the desired frequency throughout the ischemia experiments.

Since the normal function of the left hand is to generate negative torque, we wanted to determine how much negative torque the left hand applied compared to the robot and right hand during ischemia. We broke the movements into cw and ccw phases and integrated the negative torque from the left hand, robot and right hand over these phases. These integrated torque values were then normalized by the total integrated negative torque applied to the H-O-H system in that phase. Fig. 4 shows the percent contribution to negative torque from the left hand, robot and right hand in the cw and

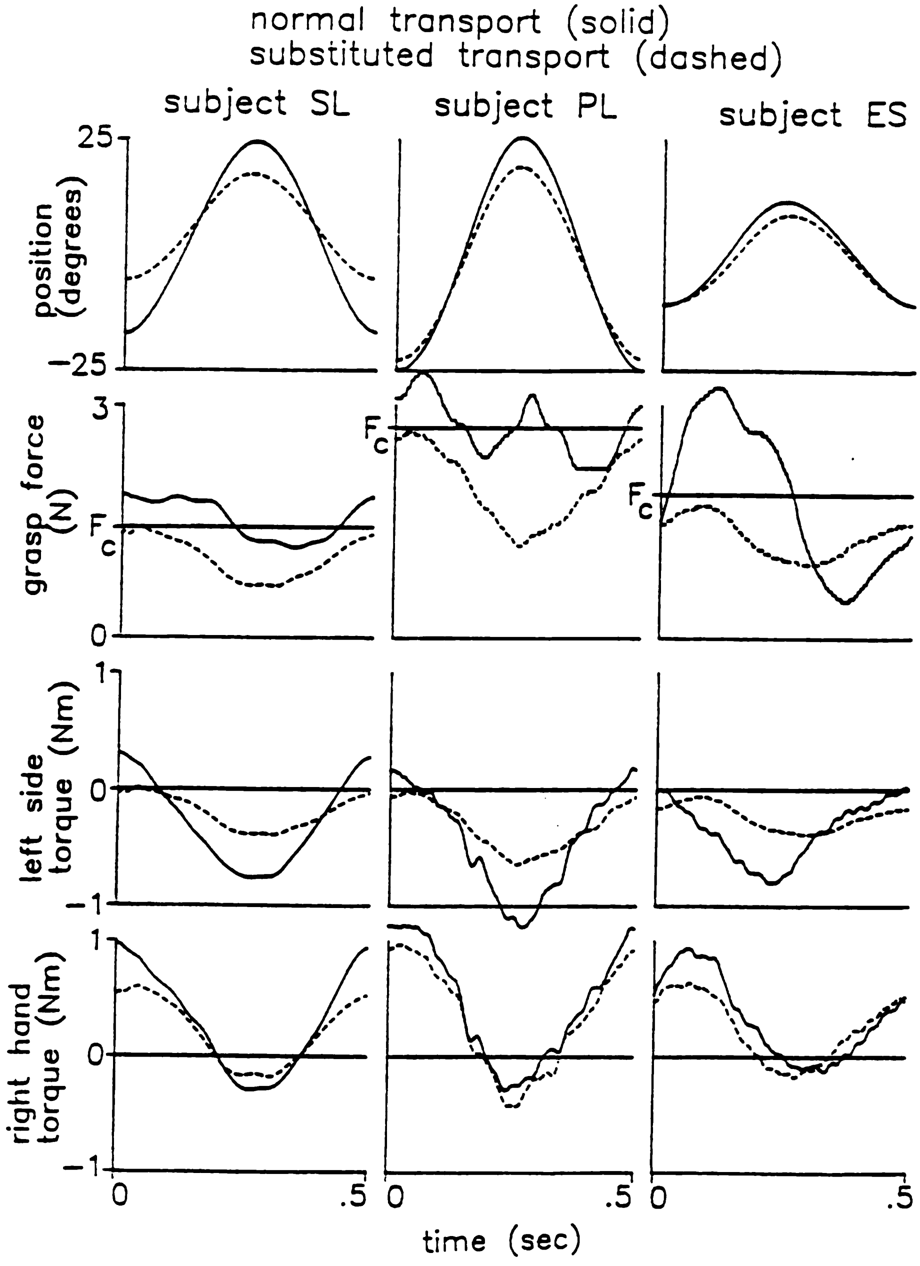

Fig. 2. Transport performed normally (solid) and with the robot substituting for the left hand (dashed) for three subjects. The horizontal lines in the 2nd row are the set points for the feedback control law. All traces are ensemble-averages of 60 movements.

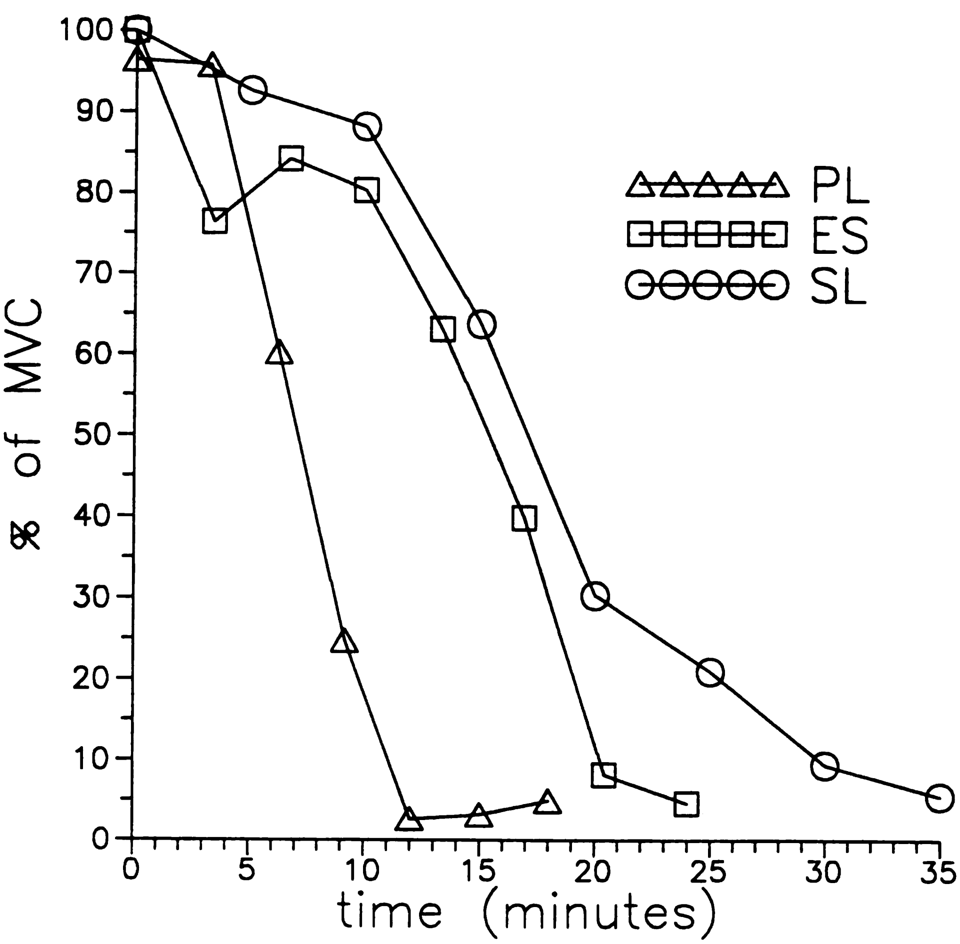

Fig. 3. The drop in maximum voluntary contraction (MVC) during the ischemia experiment for three subjects.

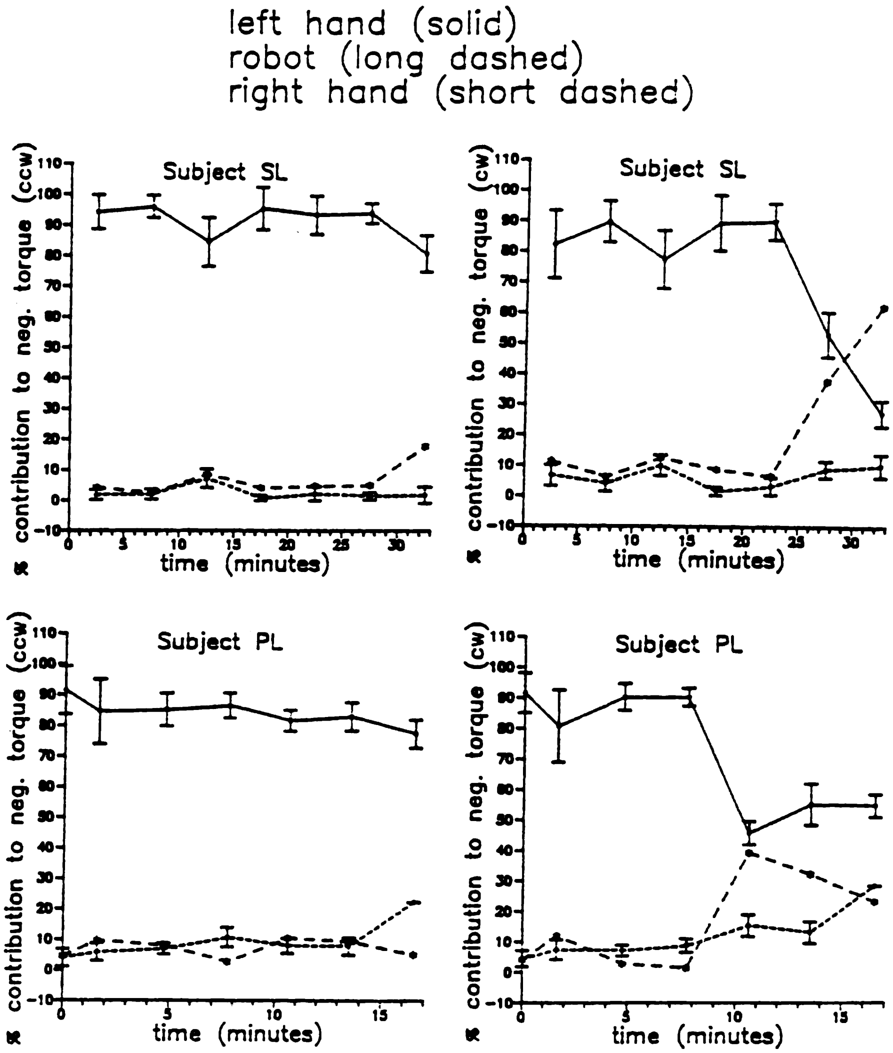

Fig. 4. Ischemic transport for two subjects. *Left column*: percent contribution to negative torque applied to the H-O-H system during the ccw phase of movement from the left hand(solid), robot(long dash) and right hand(short dash). *Right column*: same as the left column but in the cw phase of movement. All points are ensemble-averages of 60 movements. Error bars for the robot are omitted for clarity.

ccw phases of the movement during ischemia for two subjects. For both subjects, as ischemia progressed, the left hand continued to generate the torque required as the H-O-H system moved in the ccw direction, but stopped generating all of the torque required in the cw phase of movement late into ischemia. In subject SL, the robot generated all of the missing torque in the cw phase. In subject PL, the robot initially generated all of the missing torque in the cw phase. However, by the end of the experiment, this subject's right hand adapted and generated about half of the missing torque, with the robot generating the other half.

A third subject, whose data is not shown, allowed the robot to "take over" left hand function even before occlusion was begun. He relaxed the left hand and allowed the motor to move it throughout the ischemia experiment. Again, the motor adapted to the reduced performance of the subject, even though the reduced performance was caused by a lack of effort instead of ischemia.

The fact that the robot adapts to subject performance is also apparent when examining torque patterns. Fig. 5 shows the left hand torque and robot torque patterns for subject SL. This subject's left hand torque pattern changed dramatically in the final 10 minutes of occlusion. In the cw phase of the movement, his ischemic left hand produced less torque and the robot produced more torque to compensate. In fact, the robot so precisely compensated for the changing left hand torque pattern that the sum of the left hand torque and the robot torque retained the same pattern during the last 10 minutes of occlusion.

The robot only generates torque in phases of the movement when the disabled

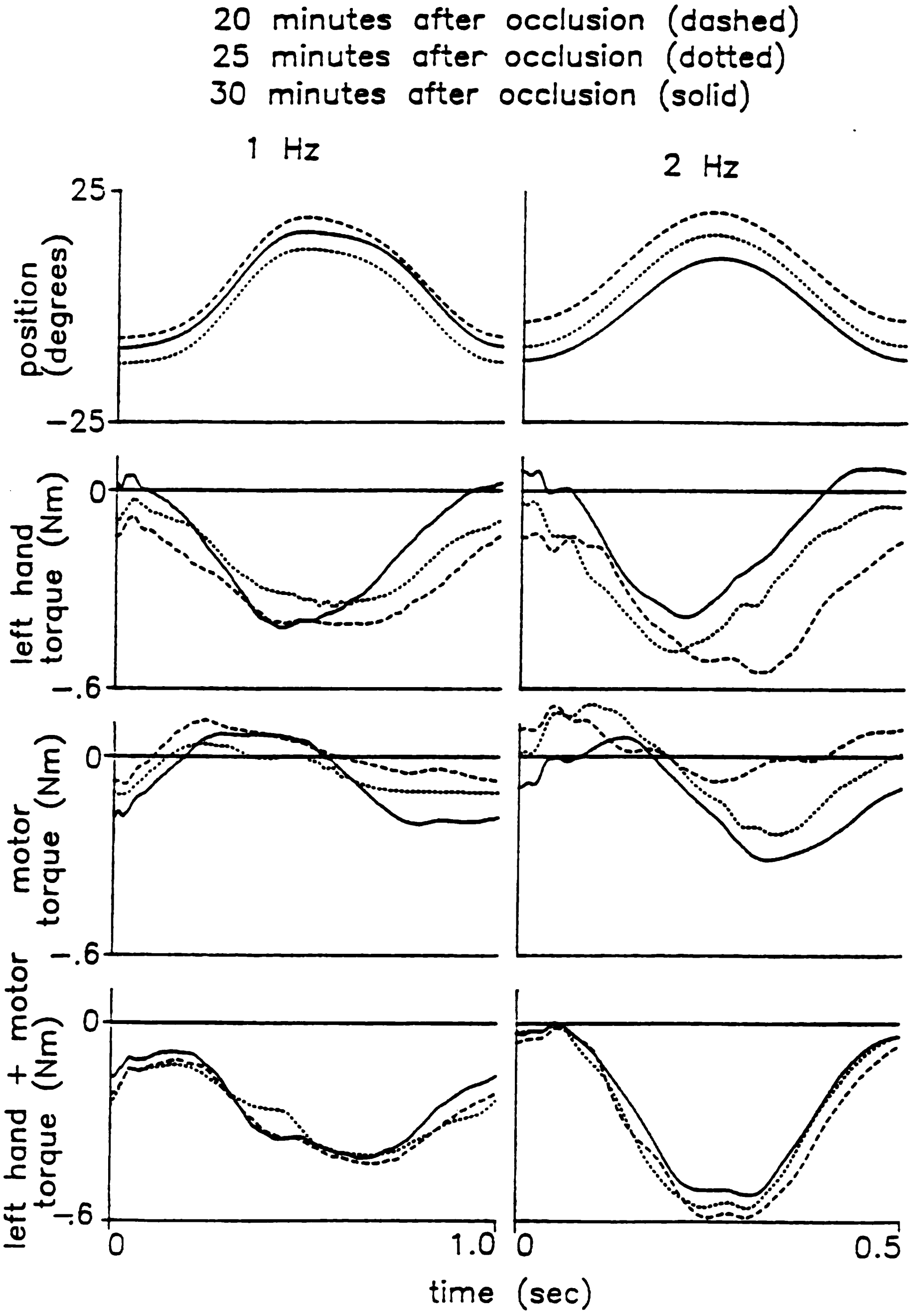

Fig. 5. Torque patterns during ischemic transport for subject SL. *Left Column*: angular position, left hand torque, robot torque and total left side torque 20 (dashed), 25 (dotted) and 30 (solid) minutes after occlusion at 1 Hz. *Right column*: same as left column but at 2 Hz. All traces are ensemble-averages of more than 50 movements.

hand torque is inadequate. Fig.6 overlays the left hand torque, robot torque and their sum for performance of the task 30 minutes after occlusion for subject SL. In the ccw phase of movement, the normal function of the left side is to generate negative torques to brake the system. At 1 Hz, the robot actually generated a small positive torque in the ccw phase and only began to generate negative torque at the end of the ccw phase. The left hand torques alone were adequate to brake the system in the ccw direction. At 2 Hz, the robot began to generate negative torque about 3/4 of the way through the ccw movement, indicating left hand braking torque was adequate for about the first half of the braking phase in the ccw direction. But in the cw phase where the normal function of the left side is to generate negative torques to accelerate the system, the left hand torque was clearly inadequate at both frequencies and the robot generated large negative torques to compensate.

We suspected the large torques from the left hand at the end of ischemia were produced by the passive mechanics of the hand. To test this hypothesis, we measured the passive stiffness of one subject's left hand. The subject placed his left hand in the motored handle and relaxed his muscles as the motor moved his hand in the same kinematic pattern as he produced during his ischemia experiment. EMGs were recorded from left hand flexors and extensors to guarantee no reflexes were being excited. The mechanical properties measured during relaxation have been shown to be identical to those measured in patients under general anesthesia and so can safely be called the passive properties of the hand [10]. Fig.7 overlays the position and left hand torques from this experiment with those generated from the left hand after 30 minutes of

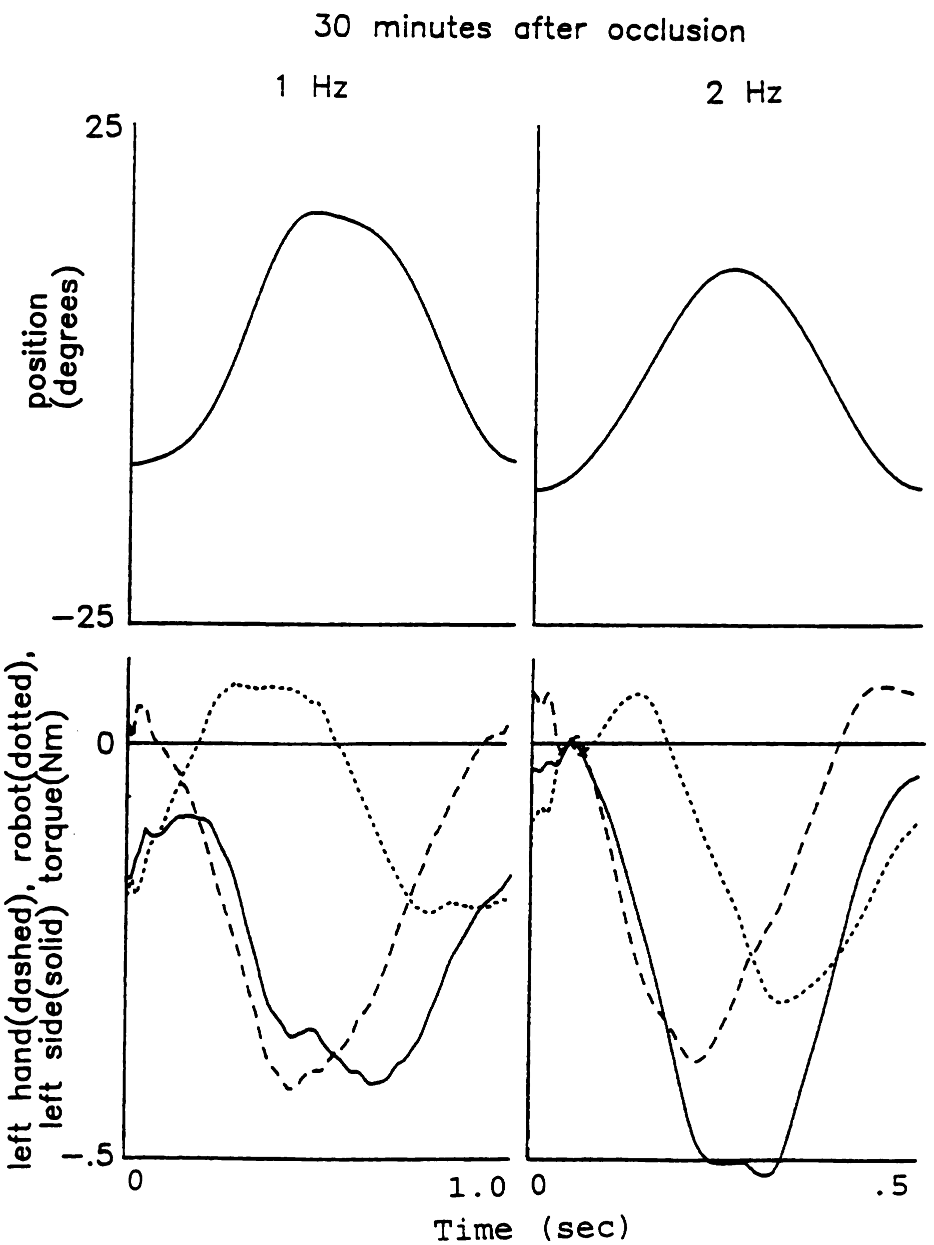

Fig. 6. Overlay of left hand torque (dashed), robot torque (dotted) and total left side torque (solid) 30 minutes after occlusion at 1 Hz (left column) and 2 Hz (right column). All traces are ensemble averages of 50 (1 Hz trials) or 60 (2 Hz trials) periods.

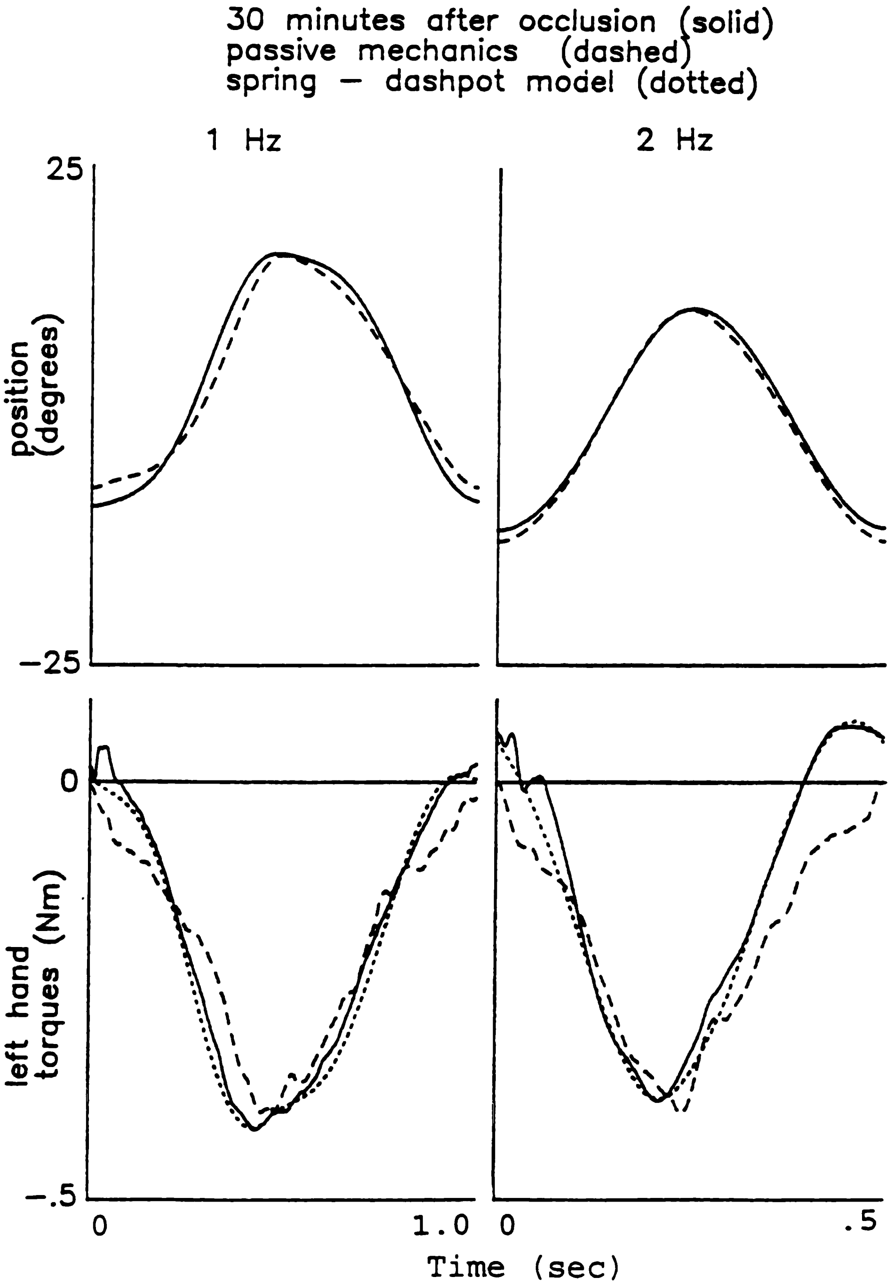

Fig. 7. Torques from the left hand 30 minutes after ischemia (solid), from passive mechanics (dashed), and from a spring-dashpot model (dotted). Model parameters at 1 Hz: stiffness = .912 N*m/rad, viscosity = .0452 N*m*s/rad. Model parameters at 2 Hz: stiffness = 1.026 N*m/rad , viscosity = .0452 N*m*s/rad.

ischemia. The position traces were almost identical and the resulting torques were not very different. We conclude the torques from the hand after 30 minutes of ischemia are the result of passive sources only. A spring dashpot model of the passive hand describes the subject's passive behavior quite well (dotted trace, in Fig.7).

Discussion

The two major results of this study are: (1) rhythmic transport can be performed with the robot substituting for one hand and operating under a simple feedback control law; and (2) rhythmic transport with the robot assisting a partially disabled hand are possible with the robot operating under this same control law.

When the robot substituted for the left hand, transport could be performed, but somewhat differently than when performed bimanually. The amplitude of transport movements were smaller than normal movements and the grasp force was always lower than the command force. We hypothesize that the low grasp force caused the subject to move more slowly (and hence make smaller movements); if he had moved quickly, the force would have dropped below the minimum grasp force required to hold the object and it would have been dropped. A higher feedback gain in the control law could prevent large drops in the grasp force, but stability problems have to be overcome to increase the gain. Computer simulations show that by doubling the A/D sampling rate, which should be possible with a faster computer, the force feedback gain can be doubled without causing instability.

When the robot assisted a progressively disabled hand, transport could still be performed and the robot generated torque that the hand did not. This is not surprising given the nature of a servo control: a servo device intervenes only when actual performance does not equal desired. Ischemia caused the subject's performance to become abnormal and the servo intervened. The robot operated in *assist* mode and didn't take over the task from the left hand. Only devices that can operate in *assist* mode can be useful in rehabilitation.

During ischemic transport for subjects SL and PL, left hand performance didn't change appreciably until MVC dropped to below 25% of normal. MVC at this point for subject SL was 4.0 Nm. The force velocity curve for this subject's wrist flexors has been previously identified [11] and a quick calculation showed that he could still generate 3.3 Nm at the peak velocity of his transport movements. Since the peak torque required from this subject's left hand during transport was only 0.6 Nm, lack of muscle force was not the cause of the sudden drop off of torque from the left hand. A more plausible explanation is related to motor unit recruitment. Given the low levels of torque required, it is likely that only the smaller motor units were recruited in performance of the task. Larger motor units were affected by the ischemia first, causing large drops in MVC. But the smaller units were still intact until late in ischemia, hence there was no change in left hand performance. Only after the smaller motor units were affected did the left hand performance decrease.

Subjects can adapt to the device. One subject allowed the robot to produce large assisting torques even before ischemia was begun. For assistive devices to be useful in therapy, the patient must do the part of the task that he or she can. Better instructions, and enough feedback to allow the instructions to be carried out may solve this problem. For the case of this subject, feedback of the torque he was producing with his left hand may have prevented his laziness. Another subject adapted during ischemic transport by using his right hand to take over some of the function that was lost from his left hand. For the case of this subject, feedback of the torque he was producing with his right hand may have prevented this adaptation.

Ideally, robotic therapy devices would be used in reverse order of the ischemia experiment. Initially, the patient would have little ability to properly activate muscles on the disabled side, and the robot would generate the missing torque. With practice, the subject would gradually relearn proper use of the disabled limb and the robot torque would correspondingly decrease. Relearning might be facilitated by visually presenting robot torque to the subject during the performance of the task. The subject could judge his own performance by attempting to minimize robot torque.

The results of this study generalize readily to other transport systems. This robotic aid prototype could be used to restore transport coordination between any pair of joints, such as the two elbow joints, the two shoulder joints, or a system consisting of a metacarpal phalangeal joint from each hand. Many transport tasks involve the coordinated control of the wrist, elbow and shoulder linkage. Consequently, future work should include transport systems with more degrees of freedom, such as a system consisting of the wrist and elbow joints on both sides.

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Chapter 5

Bimanual Reflexes During Two Hand Grasp

Abstract

We present evidence for a bimanual "set" when two hands grasp an object. This "set" uses afferent information from one hand to drive appropriate responses in both hands. Many have studied the feedforward aspects of bimanual coordination. Our results demonstrate the presence of feedback bimanual coordination when two hands manipulate an object.

Introduction

We examined the reflex responses to perturbations when two hands grasp an object between the outstretched hands (fig.1). A special apparatus allowed us to apply horizontal perturbations to one hand during this two hand grasp. We examined the EMG and force responses of both hands. We also applied downward perturbations directly to the grasped object. Again, the emg and force response of both hands were examined. Another apparatus allowed us to apply the downward perturbation to only one hand. By applying perturbations to only one hand during two hand grasp, we can examine how afference from one hand can drive reflex responses in both hands.

Grasping a Large Box With Two Hands

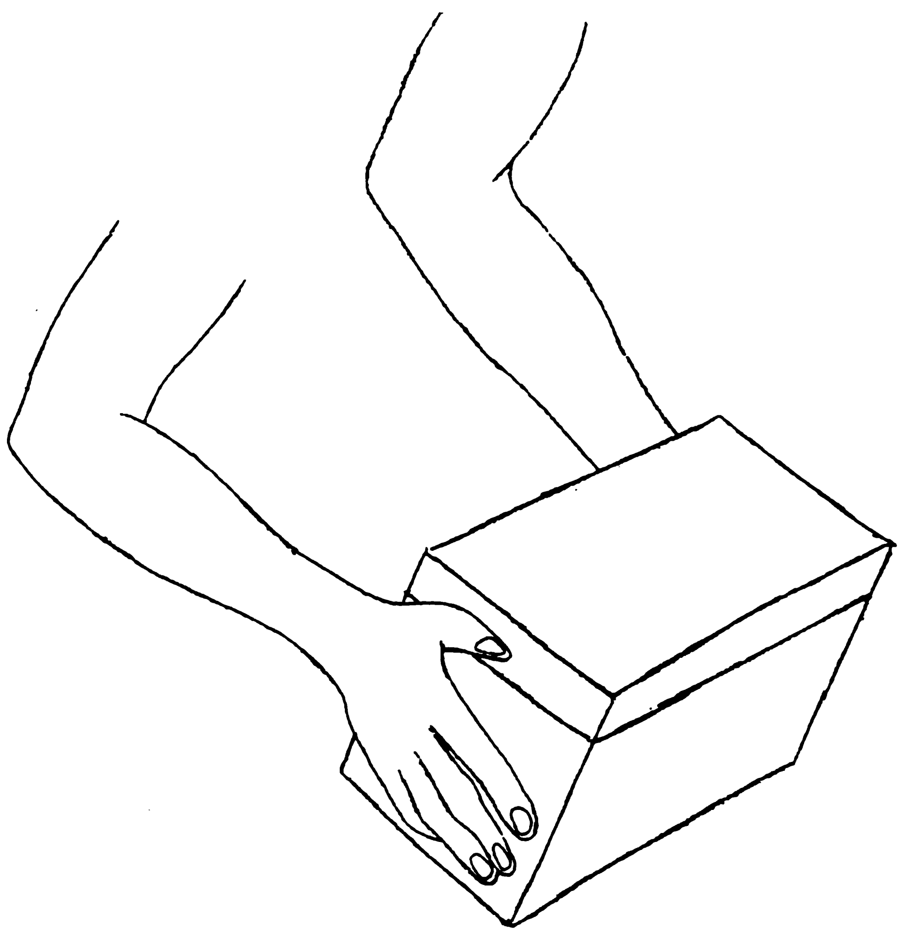

Fig. 1. Grasping a large box with two hands.

Review of current knowledge in reflexes

The stretch reflex response of muscle has been studied extensively. When an activated muscle is suddenly stretched, a short latency (M1) excitatory response is seen followed by a longer latency (M2) excitatory response. The short latency response is mediated by Ia muscle spindle afferents making direct monosynaptic (and possibly polysynaptic) connections to alpha motoneurons. The origin of the longer latency response is controversial. There are three possibilities: 1) M2 could result from phasic Ia spindle firing due to mechanical oscillations of the muscle-tendon system excited by the sudden stretch [1]. 2) M2 could be driven by the slower conducting type II muscle spindle afferents. 3) M2 could be driven by the same Ia spindles that drive M1, but through a longer supraspinal pathway. There is evidence to support each of these theories and each mechanism may contribute to the M2 response in varying degrees. Of particular interest for our study is the fact that the M2 response dominates the M1 response in muscles of the forearm [2] . The latency for the wrist flexors is 20-25 ms for M1 and about 40 ms for M2 [1].

The dual of the stretch reflex is the unloading reflex: the deactivation of an active muscle when the muscle is suddenly shortened by removing the load the muscle supports. Short latency and long latency deactivation responses have been observed at the same latencies as their stretch reflex counterparts. The mechanism for these responses may be the disfacilitation of ongoing short and long latency stretch reflex responses during tonic contractions [3]. Some forearm muscles do not show the short latency unloading response (flexor digitorum profundus and flexor pollicis longus [3]). This is not surprising considering the M1 stretch response is small when compared with the M2 response for these muscles.

Cutaneous reflexes have been less extensively studied. Electrically stimulating the cutaneous nerves of the fingers can elicit responses in active muscles of the forearm and hand. The response is a short latency excitation (E1), followed by inhibition (I1), followed by a longer latency excitation (E2). The E1 response has been observed at a 35 ms latency and E2 at a 60 ms latency in 1st dorsal interosseous [4]. However, Matthews et al. [3] observed no short latency responses in flexor digitorum profundus. They do see a long latency response at a 70 ms latency. Similarly, Johansson and Westling [5] report only the longer latency response in 1st dorsal interosseous 68 ms after stimulation. They also demonstrate how cutaneous reflexes can be used in thumb-forefinger grasp. Object slips are sensed by tactile afferents in the fingertips and triggered increases in grasp force to prevent further slips. They measure a 64 ms latency from slip to emg which is comparable in latency to the electrically stimulated cutaneous reflexes they observed. This latency is also comparable to the M2 stretch reflex response latency in these muscles. They postulate these long latency slip responses may be mediated by a supraspinal pathway.

Afference from near one muscle can excite reflex responses in other muscles. Cole and Abbs [6] have shown that during thumb-forefinger pinch movements, perturbation of the thumb results in a 50 ms latency M2 response in the thumb and a 65 ms latency response in the finger flexors. They were unable to determine if the response in the finger flexors was mediated by muscle or cutaneous afferents from the thumb.

Marsden et al. [7] have shown that perturbations on one side of the body can lead to reflex responses in muscles on the other side. These contralateral responses only exist if they assist posture or the maintenance of the position of the contralateral hand in space (holding a teacup). In such cases, they measure a M2 response at 45 ms latency when directly stretching the flexor pollicis longus and a response in the contralateral flexor pollicis longus at a 67 ms latency. They attribute the crossed responses to muscle afference routed across the body through supraspinal pathways to contralateral muscles.

The routing of muscle afference to contralateral muscles can also be spinal. Lagrasse [8] delivered tendon taps to knee extensors when the knee extensors in both legs were contracting tonically at MVC. Afference from the perturbed knee extensors crossed the spinal cord and inhibited contralateral knee extensors while facilitating contralateral knee flexors. Similarly, tendon taps to knee extensors have been shown to inhibit the tendon tap response of contralateral knee extensors at a 25 ms latency [9].

Methods

Horizontal perturbations

Subjects sat with their forearms resting on a table (fig. 2). The right forearm was immobilized and the right hand splinted. The left hand was placed in a handle that allowed only wrist flexion and extension. Grasping was simulated by having subjects push with their right hand on a force transducer. The torque from the right hand was measured and an equal torque was applied to the left handle by a motor mounted under

Apparatus for Unloading Perturbations

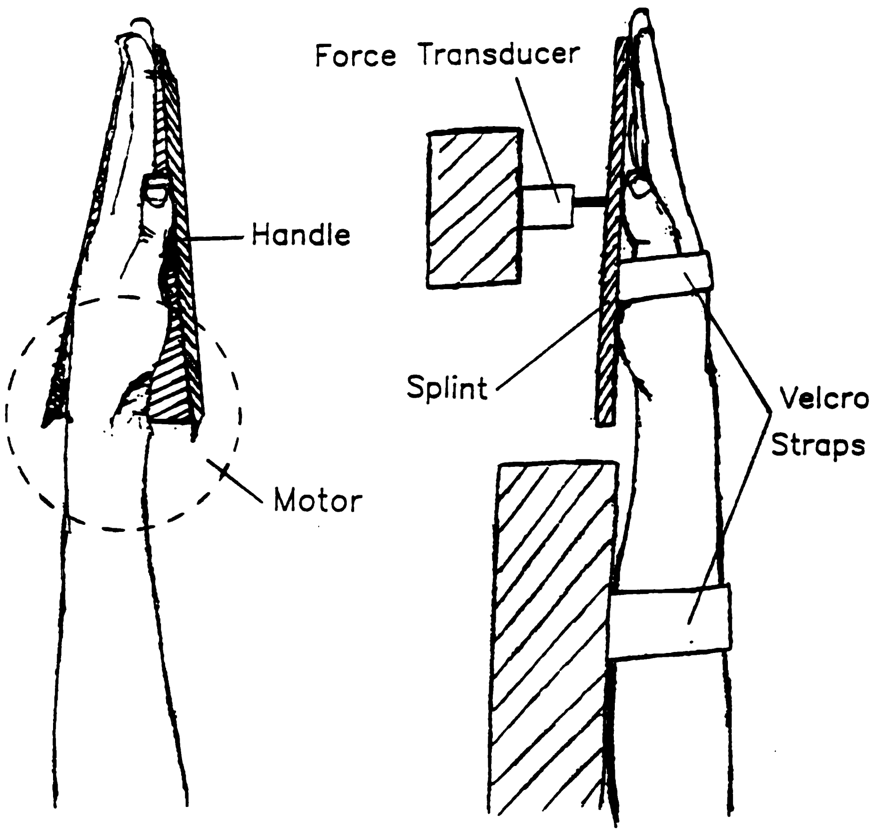

Fig. 2. Sketch of grasping apparatus for horizontal perturbations.

the table. The force transducer signal was presented to the subject on an oscilloscope. A target at 3 kg was presented on the scope. The subject was instructed to squeeze with both hands to this constant preload. At a random time after the preload was achieved, the motor was turned off, thus unexpectedly unloading the left hand. Subjects were instructed to relax after the perturbation occurred. EMG's were recorded from the flexor carpi radialis and the extensor carpi radialis of both hands. EMG's were rectified, smoothed and ensemble averaged (100 trials). The tonic emg level for supporting the preload was determined by time averaging 200 ms of the ensemble average trace prior to the perturbation. The onset of emg rise/fall was chosen to be the point where the emg crossed above/below the tonic level plus/minus two standard deviations. Left hand position and transducer force were also recorded. Three subjects were tested.

In a separate series of experiments the left hand was perturbed in the opposite direction. Grasp was simulated with the apparatus described above. At a random time after subjects achieved a 2 kg preload, the motor torque was stepped to 4 kg. Subjects were instructed to relax after the perturbation. Three subjects were tested.

To look for reflexes between other muscles besides the flexors, the grasping apparatus was slightly modified. The gain between the force transducer and the torque motor was reversed so that the preload could be supported by the right flexors and the left extensors. The force transducer was moved to the other side of the right hand (see fig. 2) so that the preload could be supported by the right extensors and the left flexors. The gain was again reversed so that the preload was supported by the right and left extensors. In each condition, the preload was 3 kg and the left hand was unloaded by unexpectedly turning off the motor. Two subjects were tested.

We wanted to measure the mechanical disturbance to the right hand from the perturbations applied to the left hand. The right hand was strapped to the force transducer with the desired preload. The subjects were instructed to relax the right hand and support the preload with the left hand as before. The left hand was perturbed as before with 3 kg - 0 kg step unloading and 2 kg - 4 kg step loading. The EMGs from the right hand were silent throughout, so the force transducer trace was a measure of the perturbation as it traversed across the body to the right hand.

Downward perturbations

Subjects were seated and held an object (120 gm) which was a force transducer extended with two machine screws to be 17 cm long (fig.3). At each end of the object was a 2 cm x 9 cm wood surface which was where subjects grasped the object. Subjects extended their hands and held the object between the tips of the middle three fingers of each hand. The middle of the object was attached to a 41 cm long arm which extended from the axis of a permanent magnet motor. The motor could apply downward perturbations directly to the object. Grasp force was presented to the subject on an oscilloscope. A target was presented to the subject representing a grasp force slightly higher than the force which the subject naturally held the object with (0.4 kg for subjects PL and SL; 0.22 kg for DR). A random time after the target grasp force was achieved the motor applied a downward perturbation the object. The perturbation length was 40 ms. The perturbation magnitude was randomly varied between three values (0.18, 0.36

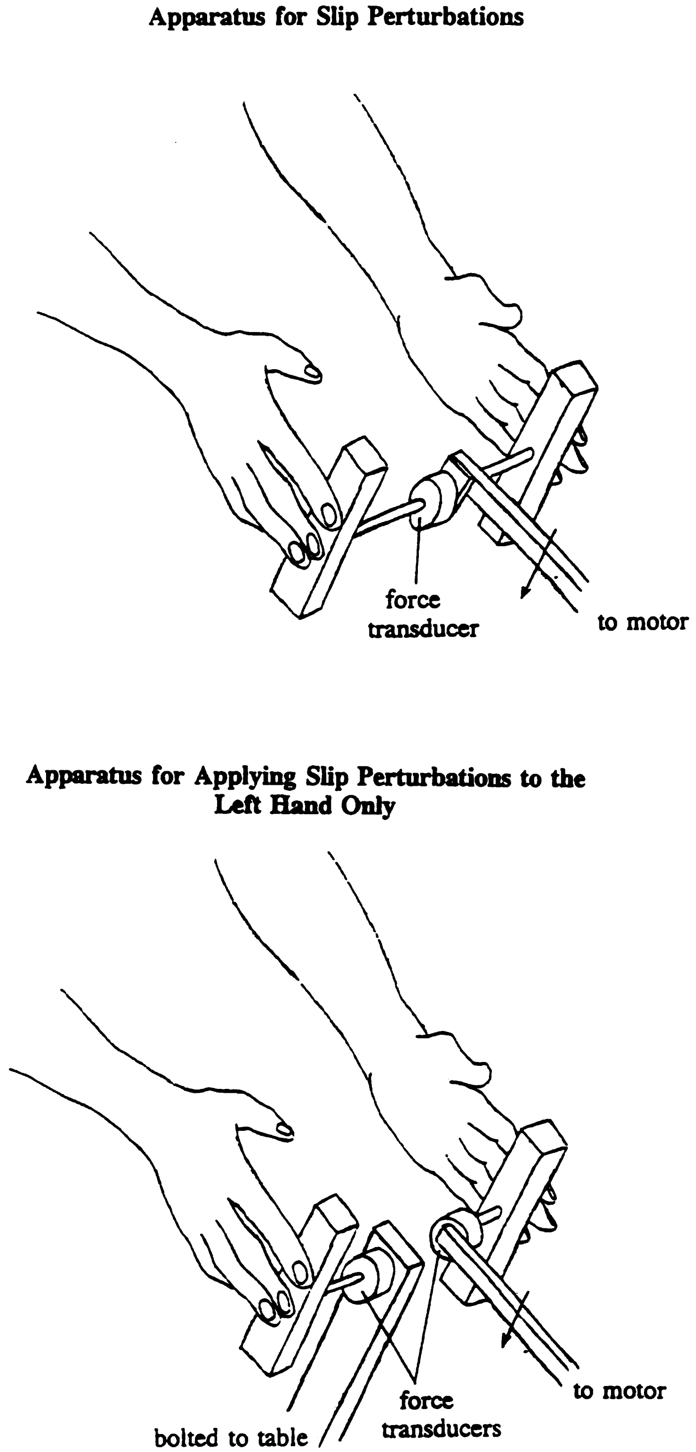

Fig. 3. *top*: Sketch of apparatus for downward perturbations applied to the object. *bottom*: Sketch of apparatus for isolating the two hands and applying downward perturbations to the left hand only.

or 0.54 kg for subjects PL and DR; 0.13, 0.26 or 0.39 kg for subject SL). A potentiometer attached to the shaft of the motor measured object displacement. Object displacement and grasp force were recorded for more than 20 trials at each of the perturbation magnitudes. Three subjects were tested.

EMGs were measured from one subject. Perturbations (0.36 kg) were applied to the grasped object as described above. EMGs were measured from the flexor carpi radialis or both hands. Sixty trials were recorded.

To perturb only one hand, a separate device was constructed (fig. 3). The object described above was detached from the arm which connected it to the motor. The left machine screw and contact surface was removed and the transducer base was rigidly attached to a block mounted to the surface of a table. The left contact surface was attached to a second force transducer whose base was attached to the 41 cm long arm connected to the motor. The "object" was held by pushing with the right hand on the grounded force transducer and pushing with the left hand on the left force transducer. The left force transducer was supported vertically by friction between the left fingertips and the left contact surface. The left force transducer was supported laterally by the arm which connected it to the motor. Both force transducer signals were displayed to the subject on an oscilloscope, along with a 0.05 kg wide target at 0.29 kg. The subject was instructed to grasp the "object" with this preload. A random time after both force transducer signals were within the preload target, the motor applied a 0.28 kg downward perturbation to the left transducer. Left transducer vertical position, both force transducer signals, and EMGs from the wrist flexors were recorded for 60 trials.

Results

Horizontal perturbations

Unloading the left hand when both wrist flexors supported the preload caused a reflex response in both flexors (fig.4). An unloading reflex of the left flexor occurred 34 ms after the perturbation. This is slightly faster than the latencies reported in literature (40 ms [1]). A similar deactivation was seen in the right flexor at a 47 ms latency. We call this response the crossed unloading reflex. This dropped muscle activation in the right flexor to about 60% of the tonic level before recovering. After a brief rise in activation, the emg again drops at around 108 ms. This response occurred well before voluntary response time (160 ms), and was not affected by instructing the subjects to contract maximally with the right flexor after the perturbation.

The crossed unloading reflex was present after unexpected unloading of the left hand when the preload was supported by the extensors of both hands (fig. 5), the left flexor and the right extensor (fig. 6) or the left extensor and the right flexor (fig. 7). In all cases the latency and form of the crossed responses were similar to the case where the load is supported by the flexors.

Loading the left hand when the flexors supported the preload did not result in a crossed loading reflex (fig. 8). The left hand flexors responded with stretch reflex at a 18 ms latency. This is slightly faster than the M1 response reported in literature for these muscles (20-25 ms [1]). Subjects PL and ES showed no excitatory response in the right hand at all, while subject DR showed only a modest rise in right hand emg at around 80

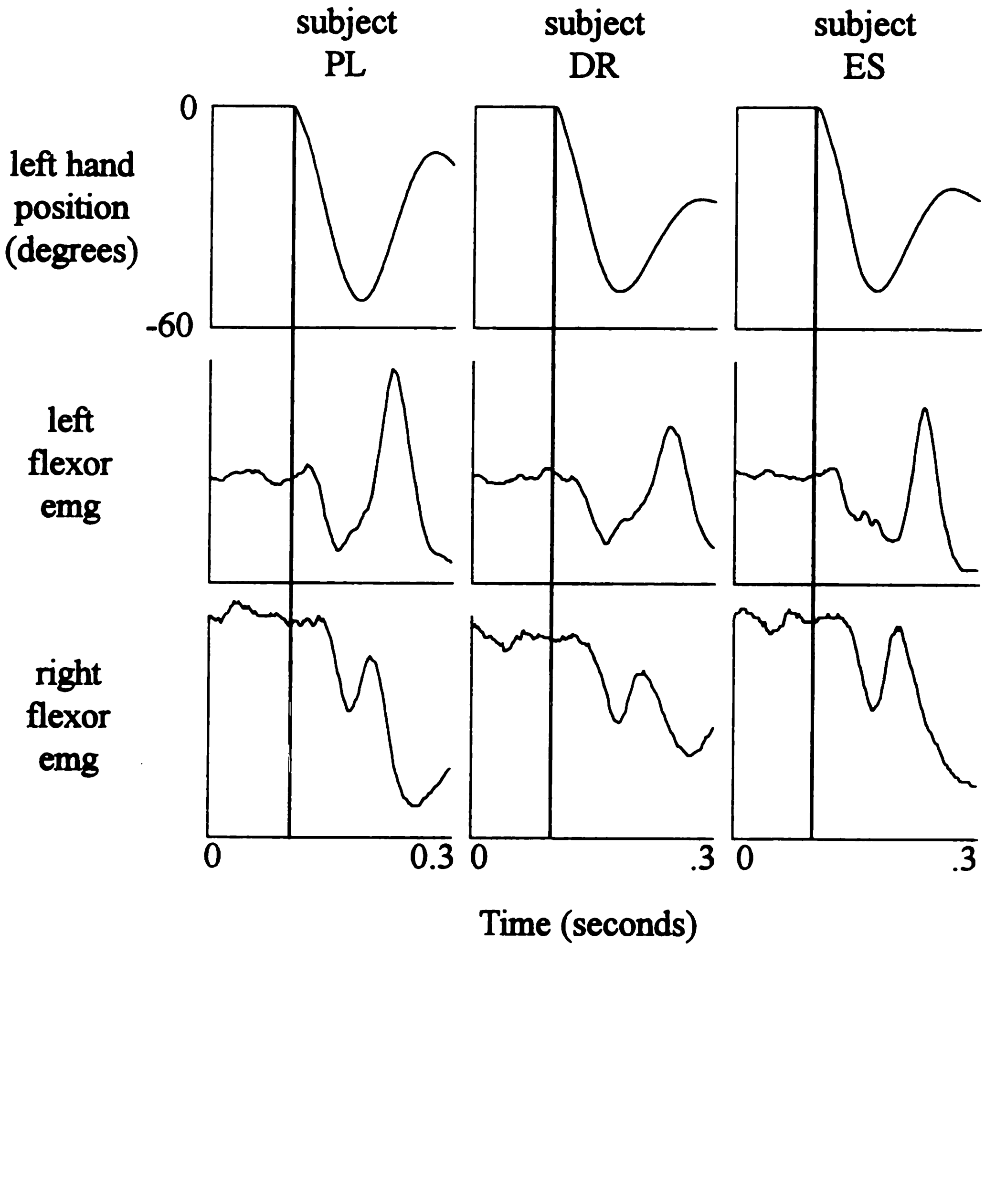

Fig. 4. The autogenic unloading reflex (left flexor) and the crossed unloading reflex (right flexor) when the flexors support the preload. Negative angular position in the left hand indicates flexion. The onset of the perturbation is indicated by movement in the left hand. All trajectories are ensemble averages of 100 trials.

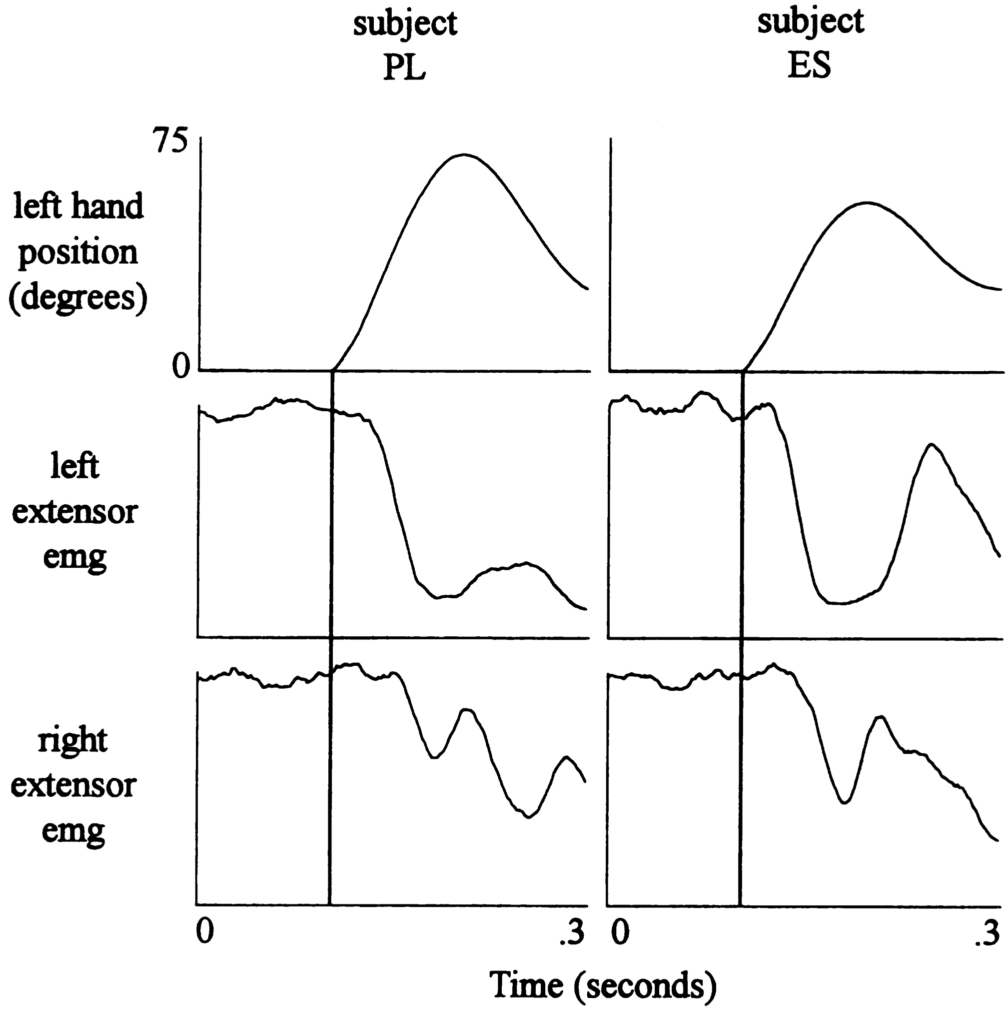

Fig. 5. The autogenic unloading reflex (left extensor) and the crossed unloading reflex (right extensor) when the extensors support the preload. All trajectories are ensemble averages of 100 trials.

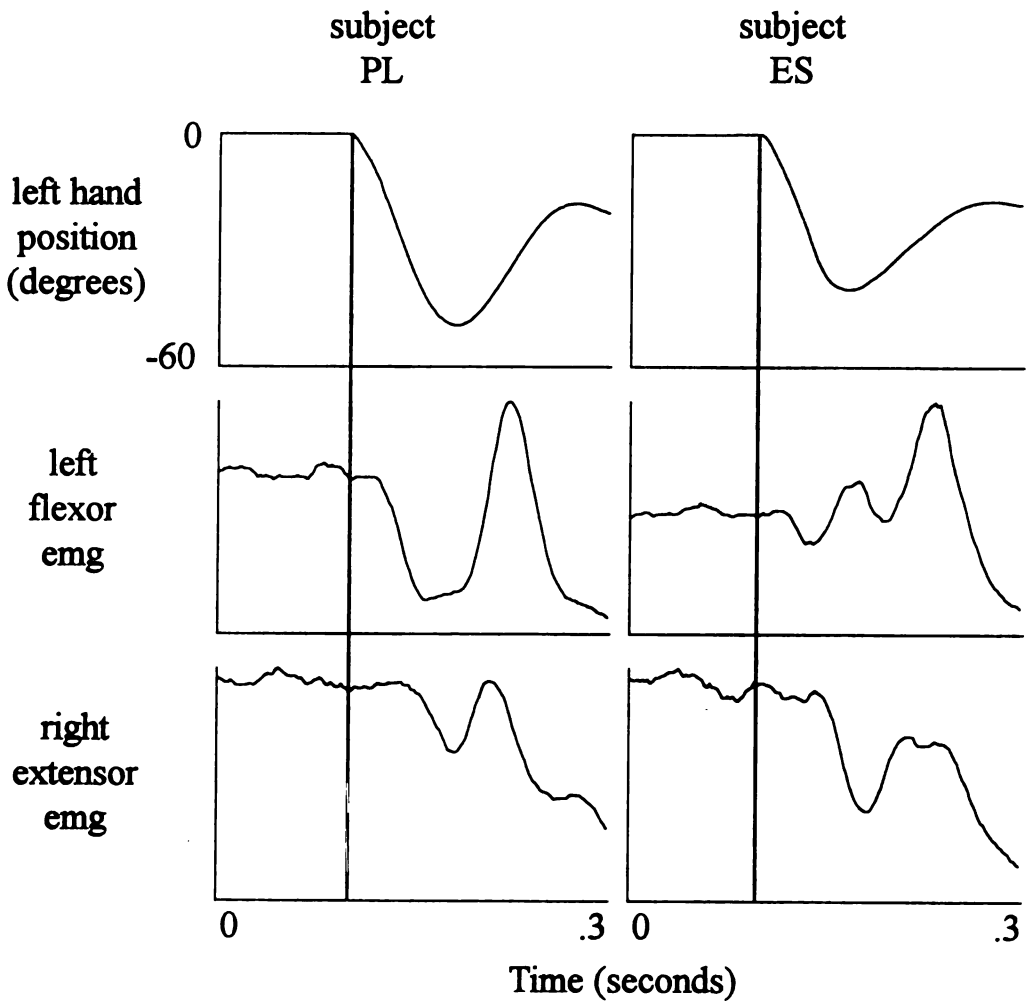

Fig. 6. The autogenic unloading reflex (left flexor) and the crossed unloading reflex (right extensor) when the preload is supported by these muscles. All trajectories are ensemble averages of 100 trials.

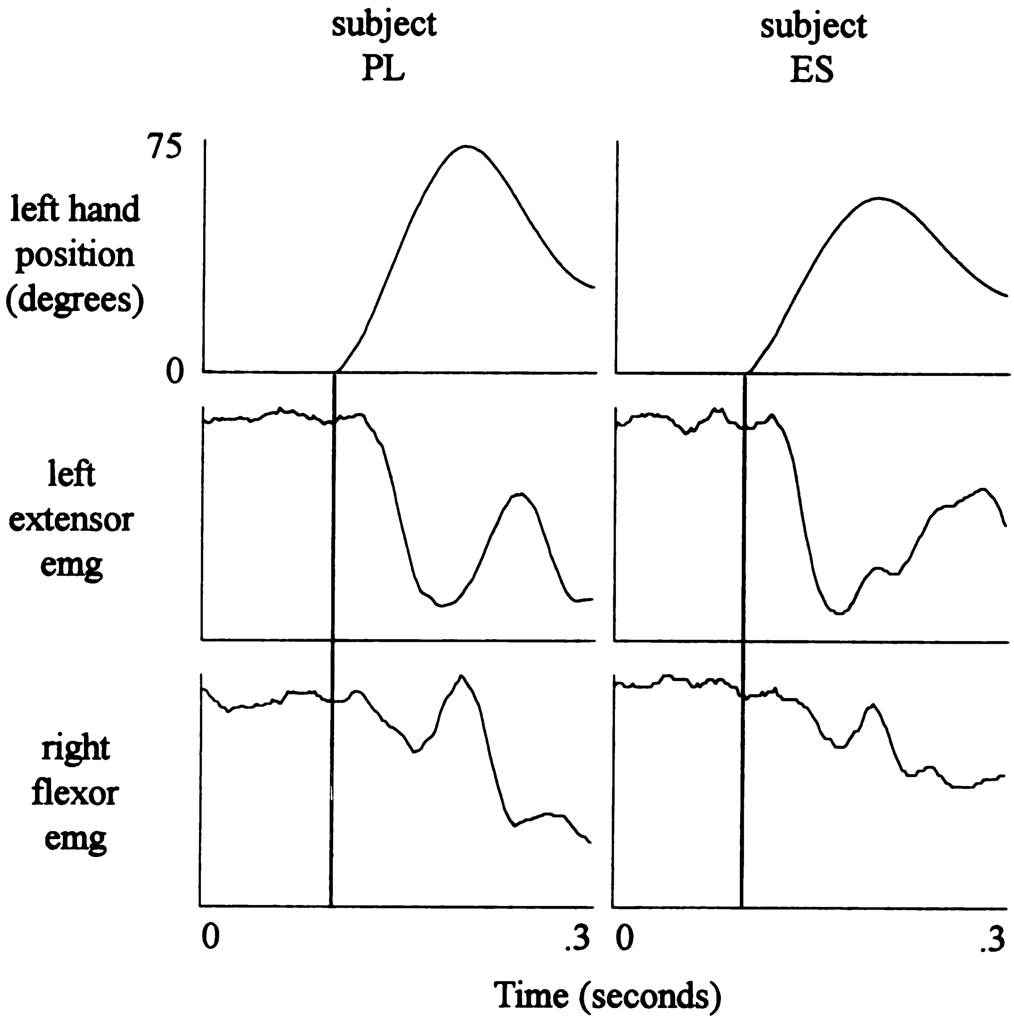

Fig. 7. The autogenic unloading reflex (left extensor) and the crossed unloading reflex (right flexor) when the preload is supported by these muscles. All trajectories are ensemble averages of 100 trials.

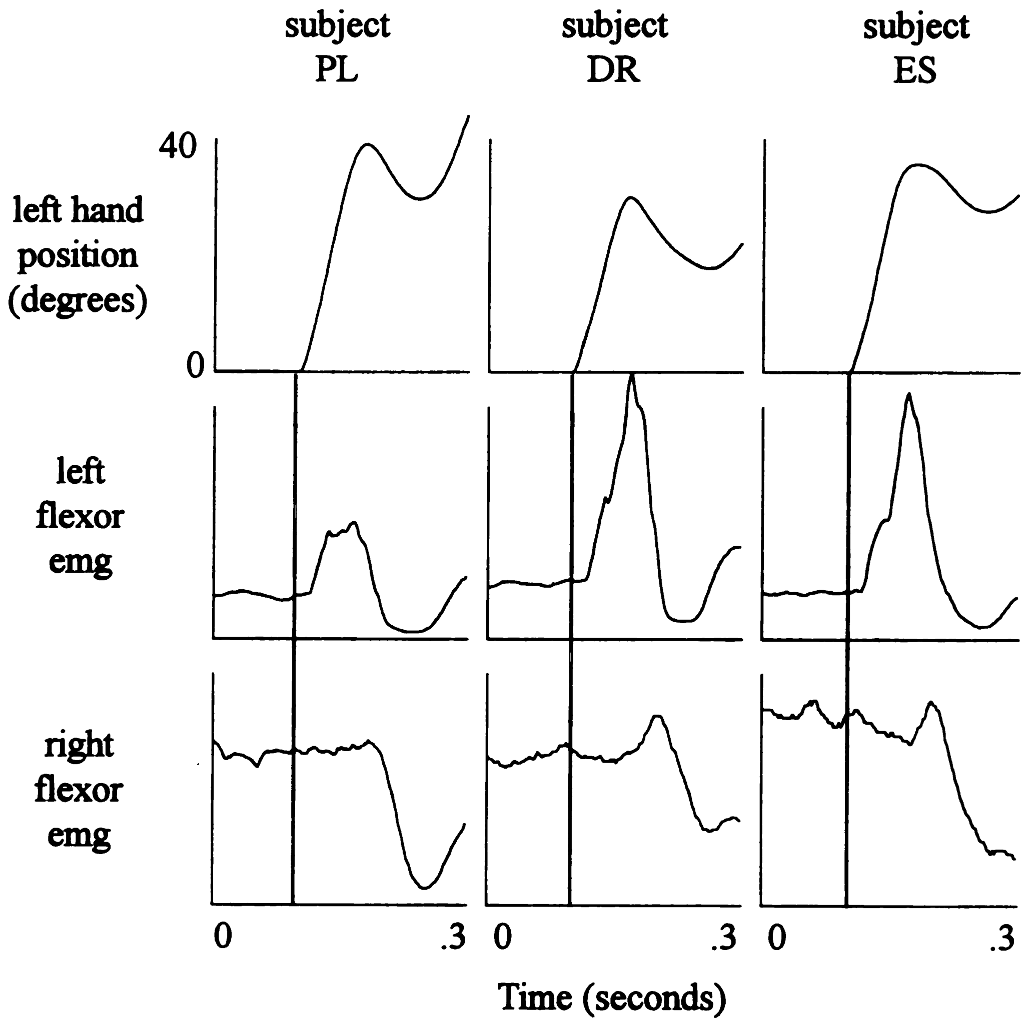

Fig. 8. The EMG responses to loading of the left hand when the flexors support the preload. All trajectories are ensemble averages of 100 trials.

ms. Since this comes 33 ms later than the crossed unloading reflex, this response is not the symmetric counterpart of the crossed unloading reflex and probably mediated by a different pathway. All subjects showed a deactivation response in the right hand at 108 ms.

The observed crossed reflexes were not a response to the mechanical disturbance transferred through the body, since EMG's began changing in the right hand about the same time as the mechanical disturbance reached it. We measured the mechanical effect at the right hand of the perturbation applied to the left hand (see methods). The mechanical wave arrived at the right hand after 42 ms and never exceeded .2 kg when measured by the force transducer strapped to the right hand. This wave never elicited any emg response in the relaxed right hand muscles. However, right hand muscles were contracting during the experiments. Even if this wave excited the M1 stretch reflex in the contracting right hand muscles (rather unlikely considering the size of the perturbation), the mechanical wave could not have affected the right hand emg until 60 ms after the perturbation (18 ms latency for M1 response).

Downward perturbations

Object slip triggers increases in grasp force (fig. 9). Downward perturbations resulted in downward movement of the object. This downward movement is a combination of object slip and downward arm movement. We applied the 0.36 kg perturbation directly to the hands to determine the grasp force response to downward arm movement alone. The downward movement was about 40% of object movement when

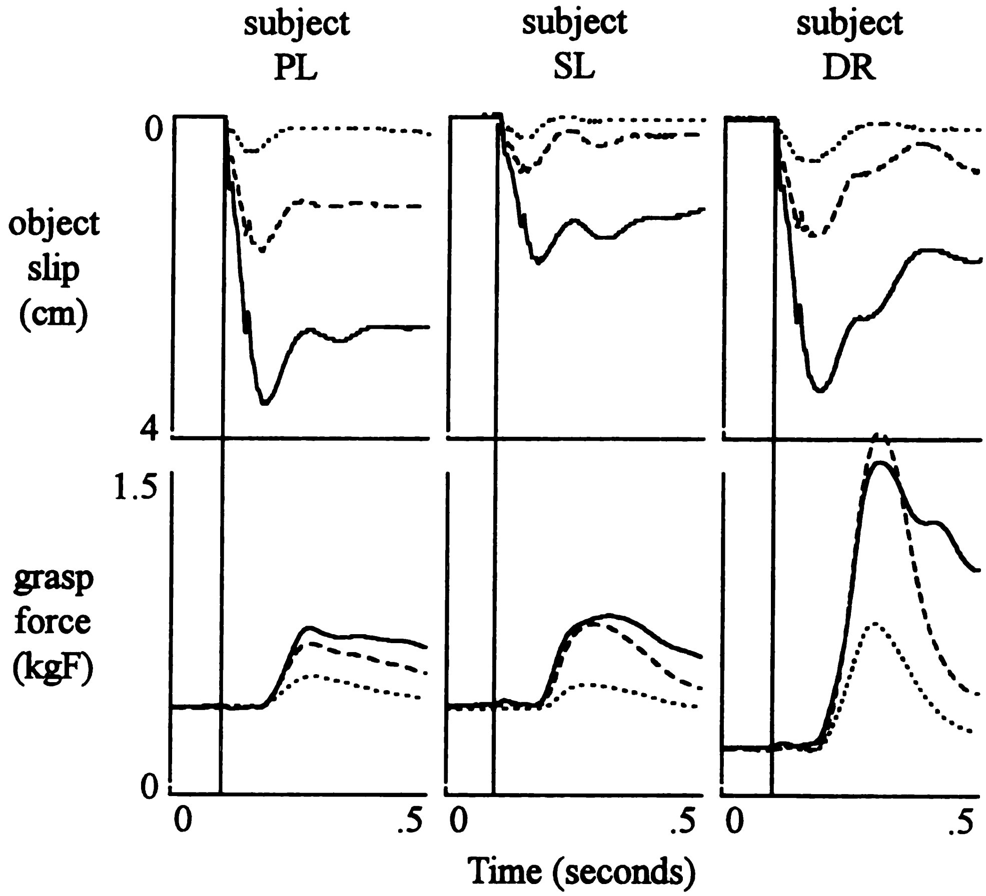

Fig. 9. Object downward movement and grasp force response for small (dotted), medium (dashed) and large(solid) downward perturbations applied to the object. All trajectories are ensemble averages of more than 20 trials.

the 0.36 kg perturbation was applied directly to the object. When the perturbation was applied directly to the hands, the grasp force showed a ripple that began at the onset of the perturbation, but the force never exceeded 0.025 kg above the preload value. This response was caused by the perturbation directly. In contrast, when the perturbation was applied to the object, the grasp force rose to 0.3 kg above the preload value. Furthermore, when the perturbation was applied to the object, the grasp force responses were unchanged when the hands and forearms of the subject were supported, preventing any arm movement. Therefor, increases in grasp force were driven by object slip and not movement of the arms. These slip responses were driven by the cutaneous sensors in the fingertips and not muscle afference.

The rate of grasp force increase was larger for higher velocity slips, but the maximum force rate saturates (fig. 9). The velocity and extent of downward slip was larger for larger sized perturbations. The grasp force began to rise 86 ms after the onset of object slip. Increasing the perturbation size from small to the medium clearly increased the rate of force increase. However further increase in the perturbation size from medium to large resulted in no further increase in force rate.

Emgs were recorded from one subject at the medium size perturbation (fig. 10). The emg in both hands appears biphasic and rises at a 46 ms latency. The grasp force rises at a 86 ms latency following a 40 ms delay from emg to force production.

Part of the slip response in each hand is driven by afference from the other hand (fig. 11). The two hands were isolated and the perturbation was delivered to the left hand only (see methods). As before, the emg rise in the left hand began at a 46 ms latency

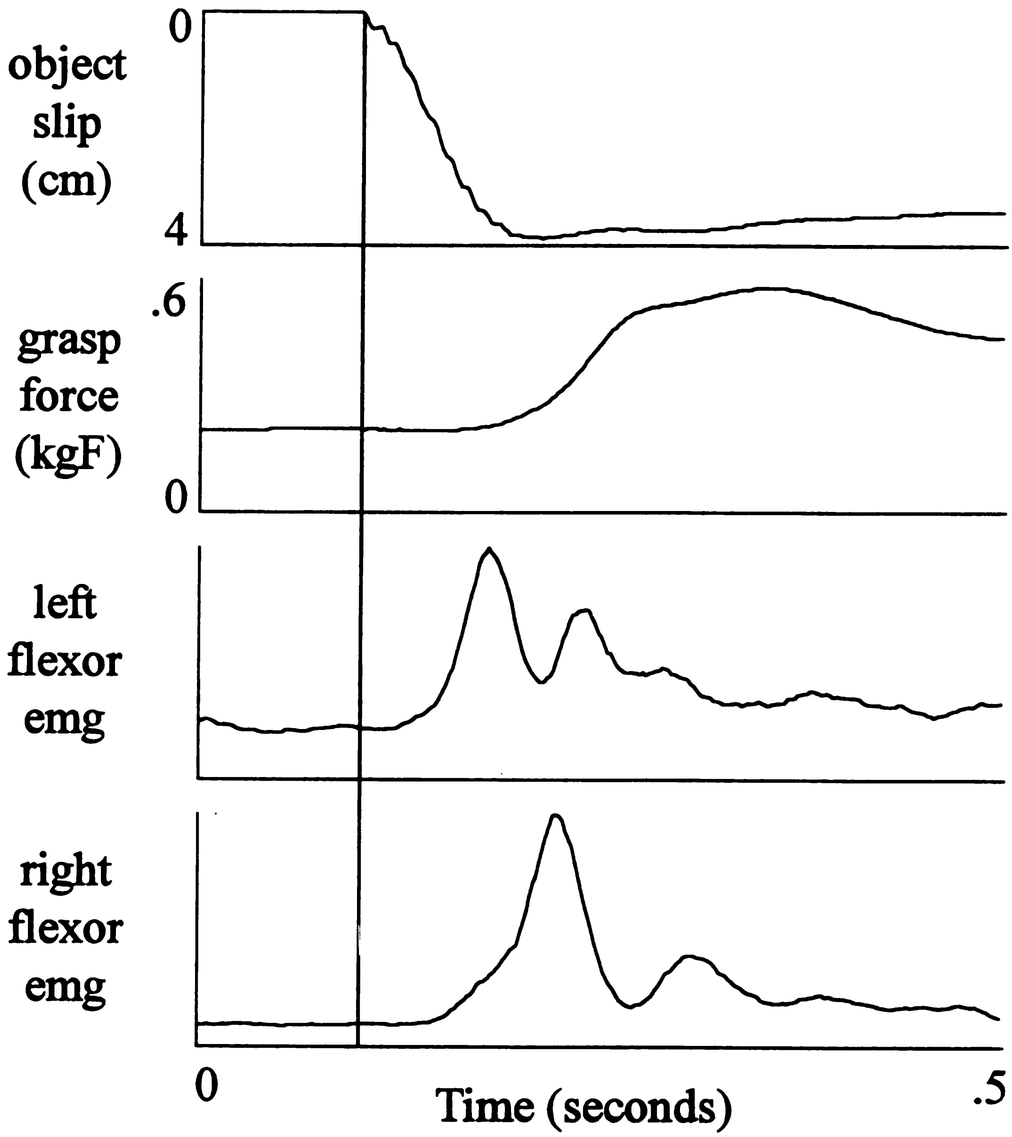

Fig. 10. Object downward movement, grasp force response and left and right flexor EMG for the medium perturbation. All trajectories are ensemble averages of 60 trials.

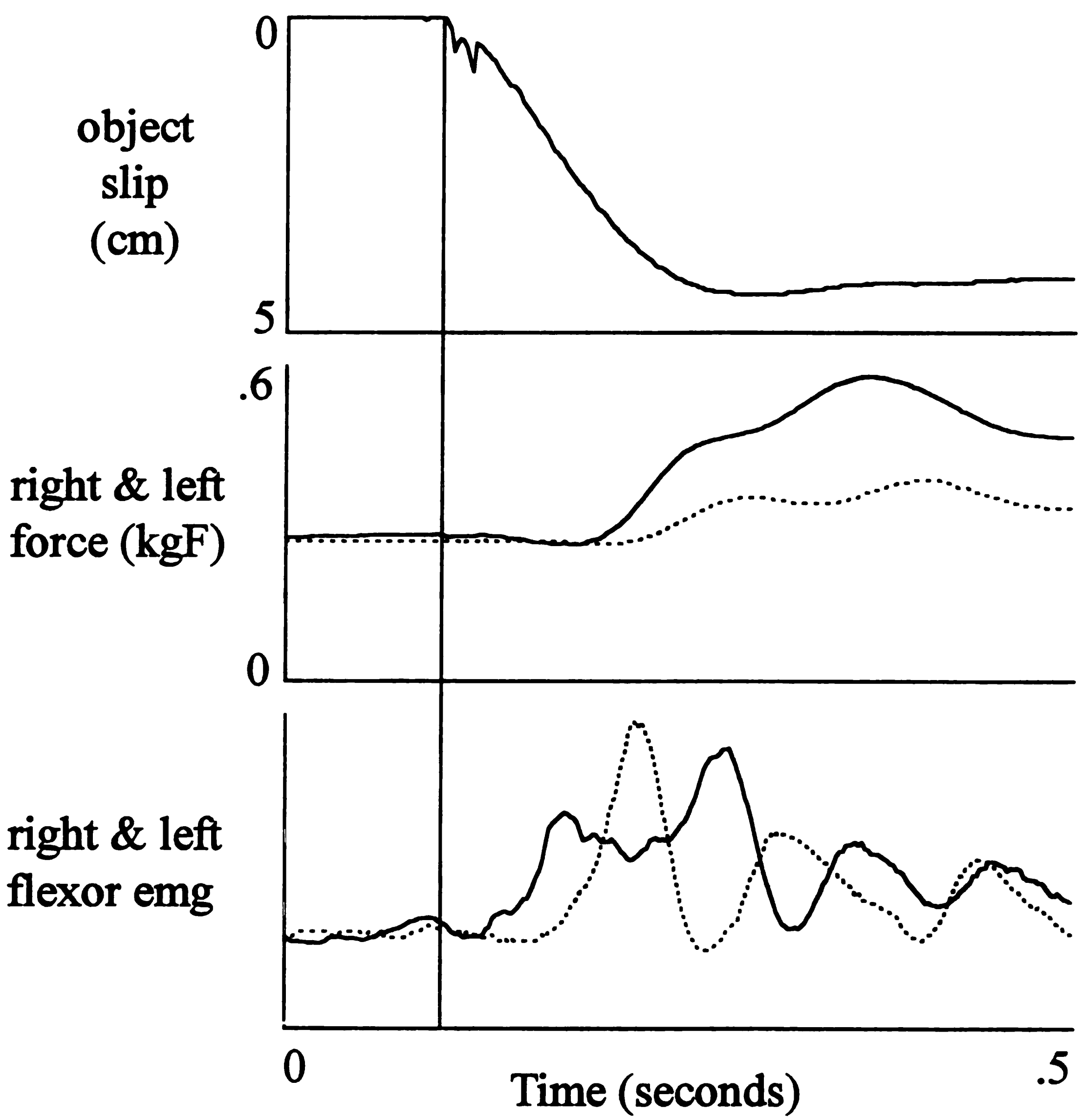

Fig. 11. Left transducer downward movement, left EMG and force responses (solid), and right EMG and force responses (dotted) when the downward perturbation is applied only to the left hand. All trajectories are ensemble averages of 60 trials.

followed by a left hand force increase at 86 ms. Surprisingly, the right hand emg also showed a sharp response at a 76 ms latency and a corresponding force rise at 116 ms.

Discussion

Horizontal perturbations

The initial deactivation phase of the crossed unloading response was driven by afference from the contralateral active muscle. This initial response was probably mediated by the unloaded spindles of the unloaded left hand muscle, the same stimulus that drives the autogenic unloading reflex. Cutaneous afference can be ruled out based on latencies. We measured a crossed cutaneous response with a 76 ms latency, compared to the 47 ms latency seen here. The Golgi tendon organs from the left hand active muscle and afference from the stretched left hand antagonist are unlikely candidates since they have been shown not to mediate the autogenic unloading reflex [10].

The second phase of deactivation at 108 ms is present in both loading and unloading perturbations and is still earlier than measured response time (160 ms). This later crossed response may have been mediated by the same afference that drove the initial crossed response but through a longer pathway. Alternatively, cutaneous afference may drive this longer latency response.

The crossed responses seen here are likely supraspinal. The non-symmetric responses to loading and unloading perturbations suggests considerable neural processing, which is indicative of a supraspinal pathway. The fact that the unloading response can be routed to any pair of contralateral muscles also suggests a supraspinal pathway. Therefore, it is unlikely these responses are related to the stereotyped crossed spinal responses to tendon taps in leg muscles [8,9].

The crossed unloading reflex does not cancel the mechanical disturbance to the contralateral hand. Consider the case where the flexors support the preload. Perturbation to the left hand initially causes a decrease in the force applied by the right hand on the force transducer. The reflex deactivation of the right flexor also decreases the force applied by the right hand. Therefore, the effect of the reflex is to add to the perturbation.

The crossed unloading reflex is not a postural reflex. The subjects were seated with feet firmly on the ground. Their left hands were held in a fixed handle, and their right forearms were clamped to a solid surface, preventing torques applied about the wrist joint from rotating the torso. Thus, their posture was not disturbed by the perturbations. Furthermore, crossed postural reflexes have been reported to have a 22 ms delay between unilateral and contralateral responses in the forearm muscles [7], which is significantly longer than the 13 ms delay seen here. This suggests different pathways for crossed postural reflexes and the crossed unloading reflexes we observed. Since it is likely these crossed responses are supraspinal, it is not surprising that these crossed responses have different pathways given that different supraspinal structures are know to mediate posture vs. fine control of distal muscles.

We postulate that the crossed unloading reflex is the result of a bimanual "set" which is in effect when two hands manipulate an object. Often during bimanual tasks, one task requirement is to keep the object stationary. To accomplish this, it is important for the two hands to operate symmetrically. The bimanual "set" takes afferent information which drives the autogenic unloading reflex from the left side and uses it to quickly deactivate the active muscles on the right side. This enables the right hand force trajectory to mirror the left hand force trajectory with only a small delay. This bimanual "set" can also use afferent information for event detection. This might explain the lack of a crossed loading response. Consider the case where the flexors support the preload. When the left hand is loaded, this indicates the held object has been perturbed to the left. A crossed excitatory response in the right hand flexor would only add to the perturbation. This bimanual "set" is analogous to a unimanual "set" present during thumb-forefinger grasping movements. Perturbations to the thumb result in a 50 ms latency response in the thumb as well as responses in the finger muscles only 15 ms later [6].

Downward perturbations

The observed slip responses are likely to be related to the reflex activation of active muscles in the forearm and hand when the cutaneous nerves of the fingers are electrically stimulated. The latencies reported in the literature are variable. Sources report a latency of about 70 ms for flexor digitorum profundus [3] and 68 ms for 1st dorsal interosseous [5]. However responses as early as 35 ms have been reported in 1st dorsal interosseous [4]. The 46 ms latency response observed here could be mediated by a similar pathway.

Similar slip responses have been reported during thumb-forefinger grasp [5]. They report a latency of 64 ms from object slip to onset of emg responses in the intrinsic hand muscles. They isolate several types of tactile sensors in the fingertips that are responsive to object slips and could reliably trigger these responses.

The question remains as to why the rate of force increase saturated with large slip velocities. Increasing the perturbation from the small to medium increased the force rate. However, further increases in perturbation size resulted in no more increase in force rate. This saturation in response might be a result of saturation of the tactile cutaneous sensors in the fingertips.

There also exists the question as to the functional significance of the crossed component of these slip responses. The answer may again be a bimanual "set" that has a priority for keeping a held object stationary in space. If the object were to slip only on the left side, a unilateral slip response would activate the left hand flexors pushing the object to the right. The crossed component of these slip responses quickly activates the right flexors as well in order to increase grasp force while keeping the object stationary.

Conclusions

Feedback driven responses subserving posture and one hand grasp have been reported by other investigators. The CNS prepares for these tasks by engaging special coordinative structures or "sets" which use the feedback appropriately. However the existence of a bimanual "set" during two hand manipulations has not been investigated. We have shown that muscle and cutaneous afferents from one hand are used by this bimanual "set" to drive appropriate responses in both hands. Many have studied the feedforward aspects of bimanual coordination. It is apparent from these results that *feedback* bimanual coordination is also present during bimanual tasks.

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Chapter 6

Vertical Transport :

Normal Performance and a Prototype Assistive Device

I. Introduction

II. Device design

III. Vertical transport coordination in normals

IV. Substitution test of the device

V. Summary & future work

I. Introduction

We now apply the concepts developed in previous chapters to a useful bimanual ADL, vertical transport: lifting an object off a table with two hands, supporting it statically, then replacing it on the table. Our motivation for studying bimanual tasks and our philosophy for designing assistive devices has been expressed in previous chapters. In this chapter, we describe the design of a prototype device (section II). In section III, we use the device to study vertical transport coordination in normal subjects. In section IV, we test the ability of the device to substitute for a disabled limb in the performance of the task.

II. Device Design

Design goals

The device must satisfy three criteria to be considered a successful design. 1) It must simulate lifting a large box or object with two hands. Since our objective is coordination training and not strength training, the object should not be excessively heavy, but should be unwieldy enough so that subjects must use both hands to lift it. 2) The device must allow us to measure the vertical displacement of the object and the force applied to the object by each hand. 3) The device must be able to substitute for and assist a disabled limb during the lifting process.

There are two classes of devices to consider. The first class uses an exoskeleton that is strapped to the disabled limb and assists it in bimanually lifting a free unconstrained object. In these devices, the assistance is applied directly to the joints of the disabled limb. A disadvantage of this type of device is that the patient's limb must be strapped into the exoskeleton, which may be disconcerting to him. Also, assisting multi-joint movements would require many motors and a complicated, bulky device. The second class of devices uses a constrained object, with the active assistance applied to the object itself and not the disabled limb. The advantage of such devices is that patients grab the device just as they would grab an object in ADLs. Assistance to a disabled limb would be through the object. The disabled hand may have to be strapped to the object, but this is less disconcerting than strapping the entire limb into an exoskeleton. Also, these devices allow patients to use multi-joint movements to perform tasks which may only be 1 or 2 DOF in object coordinates. Indeed this is generally how objects are lifted in everyday life: normally, lifting is accomplished with coordinated movement of the wrist, elbow and shoulder of each limb, while the object itself is only moved in one DOF (vertically). For these reasons, we opted to design devices of this type.

1 DOF design

Our initial design was a simple 1 DOF U shaped rigid bar (fig. 1). Handles were attached to the bar which subjects grasped. The axis of rotation of the device was close to the axis of rotation of the elbow, so subjects lifted the device with elbow flexion. A motor was mounted on the axis of rotation of the bar on the left side (it is assumed the left side is disabled). A force transducer in the center of the linkage reads zero force only if subjects apply the same force at the two handles. Any asymmetry in the forces applied to the bar by the two hands would be indicated in the force transducer. The motor is controlled via a feedback law attempting to regulate the force transducer signal to zero. In this way, the motor would assist lifting of the left half of the bar if the force from the left hand is less than the force from the right hand.

This design was quickly abandoned for two reasons. First, the dynamics are too simple to capture the actual problems involved in lifting an object with two hands. In actual lifting, the two arms must behave reasonably symmetrically or the object tilts. In this device there is no penalty for large asymmetries in the two lifting arms. In fact the object could be lifted quite easily with one arm. Preliminary results showed that subjects produced rather large asymmetries in the forces applied to the object with this device.

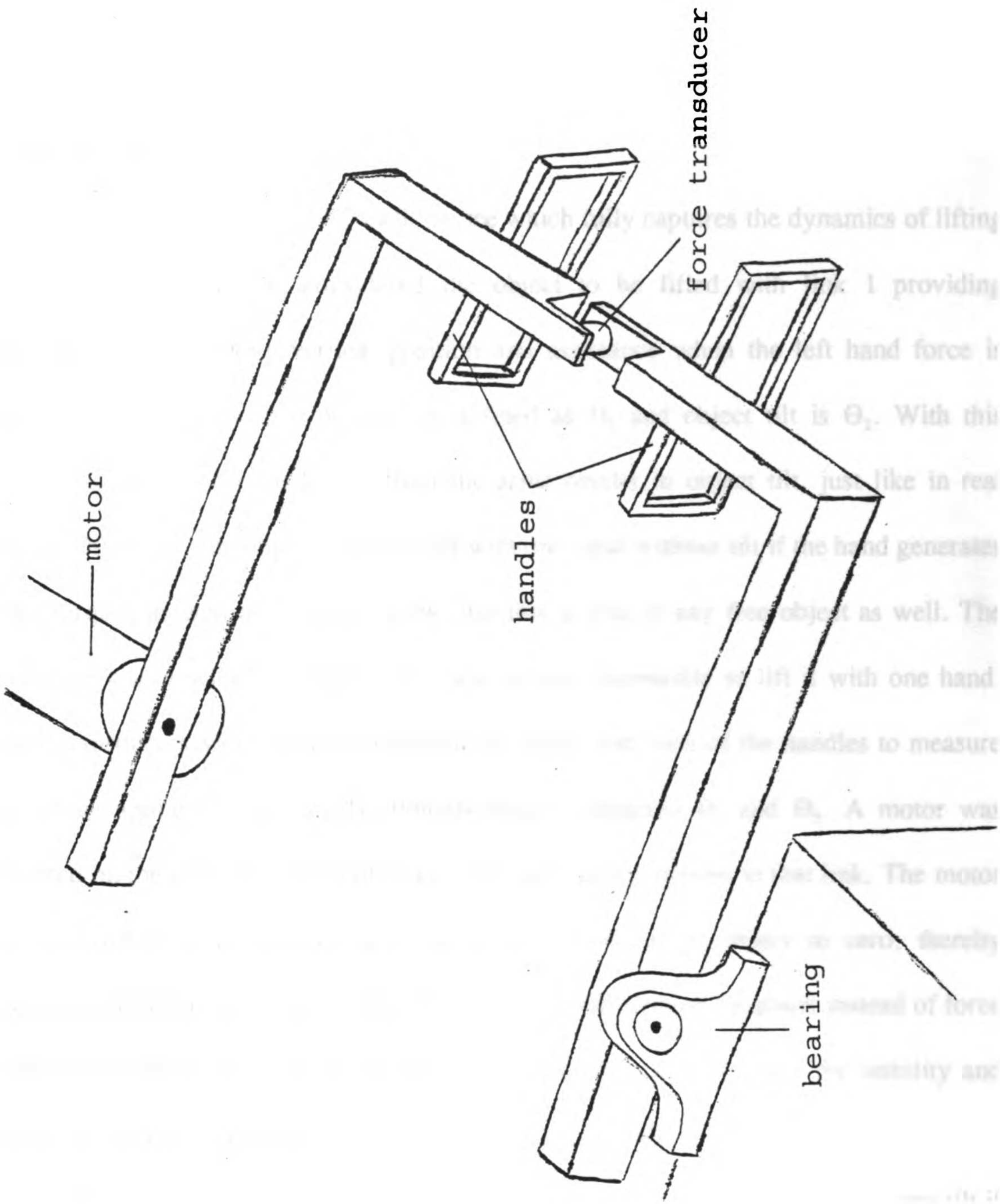

Fig. 1. 1 DOF design. The subject sits between the motor and the right bearing, places his elbows near the axis of rotation of the device, grabs the two handles and lifts the device with elbow flexion.

Secondly, force feedback control of the motor in this rigid configuration was found to be difficult due to instability and chatter.

2 DOF design

Our second design is a 2 DOF device which fully captures the dynamics of lifting (fig. 2). Link 2 can be considered the object to be lifted with link 1 providing measurement of object's vertical position and assistance when the left hand force is inadequate. Object vertical position is defined as Θ_1 and object tilt is Θ_2. With this device, asymmetry in the forces from the arms results in object tilt, just like in real lifting. Of course, the object can be lifted with one hand without tilt if the hand generates large torques as well as vertical forces. But this is true of any free object as well. The prototype was unwieldy enough that it was almost impossible to lift it with one hand. Force transducers were mounted between the object and each of the handles to measure the forces applied by the hands. Potentiometers measured Θ_1 and Θ_2. A motor was mounted on the axis of rotation of link 1 and could apply torques to that link. The motor was controlled by a feedback law regulating the tilt of the object to zero, thereby assisting a disabled left arm in lifting the object. Using position feedback instead of force feedback decreases the cost of the device substantially. Furthermore, the stability and chatter problems encountered in force feedback are avoided.

In order to properly simulate lifting free objects, the object must have zero tilt if the forces delivered at the two handles are the same. In general, this is not the case for this device given the extra inertia and weight on the left side due to link 1. However, it

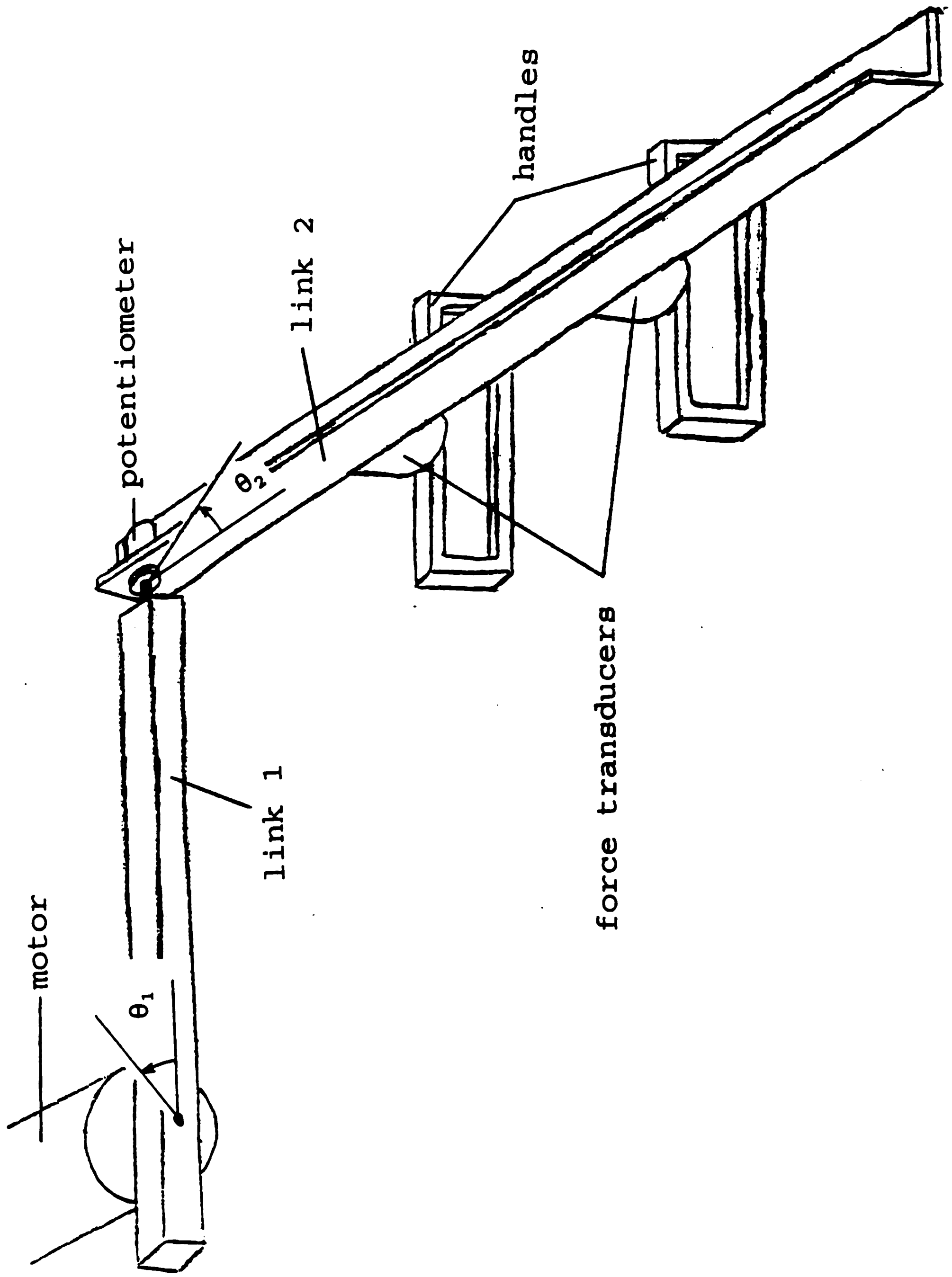

Fig. 2. 2 DOF design. The subject sits, places his elbows near the axis of rotation of link 1, grabs the two handles and lifts the device by rotating link one while keeping link two horizontal.

is possible to *balance* the device statically and dynamically so that equal left and right hand forces always results in no object tilt. The full dynamics of the device are derived in appendix A. The equations of motion are (see fig.3 for parameter definitions):

$$[M]\begin{bmatrix}\ddot{\theta}_1\\ \ddot{\theta}_2\end{bmatrix} = [V]_{centrifugal}+[V]_{coriolis}+[V]_{gravity}+[F]_{external} \tag{1}$$

$$M = \begin{bmatrix} I^1_{zz}+m_2l_3^2+m_1l_1^2+\sin^2\theta_2(I^2_{yy}+m_2l_2^2)+\cos^2\theta_2 I^2_{xx} & m_2l_2l_3\cos\theta_2 \\ m_2l_2l_3\cos\theta_2 & I^2_{zz}+m_2l_2^2 \end{bmatrix} \tag{2}$$

$$V_{centrifugal} = \begin{bmatrix} \dot{\theta}_1^2(m_2l_2l_3\sin^3\theta_2+m_2l_2l_3\cos^2\theta_2\sin\theta_2-m_2l_2l_3\sin\theta_2) \\ +\dot{\theta}_2^2(m_2l_2l_3\sin\theta_2) \\ \hline \dot{\theta}_1^2\cos\theta_2\sin\theta_2(I^2_{yy}-I^2_{xx}+m_2l_2^2) \end{bmatrix} \tag{3}$$

$$V_{coriolis} = \begin{bmatrix} \dot{\theta}_1\dot{\theta}_2\sin\theta_2\cos\theta_2(2I^2_{xx}-2I^2_{yy}-2m_2l_2^2) \\ 0 \end{bmatrix} \tag{4}$$

$$V_{gravity} = \begin{bmatrix} m_2g(l_2\sin\theta_2\sin\theta_1-l_3\sin^2\theta_2\cos\theta_1-l_3\cos^2\theta_2\cos\theta_1) \\ -m_1g(l_1\cos\theta_1) \\ \hline -m_2g(l_2\cos\theta_2\cos\theta_1) \end{bmatrix} \tag{5}$$

$$[F]_{external} = \begin{bmatrix} l_3\cos\theta_2(F_l+F_r) \\ l_2(F_l+F_r)-r_lF_l+r_rF_r \end{bmatrix} \tag{6}$$

The reference configuration of $\Theta_1=\Theta_2=0$ corresponds to the two links in the horizontal position. In order to *balance* the object, we require Θ_2 and all its derivatives be zero when $F_l=F_r\equiv F$. Applying these conditions, we get:

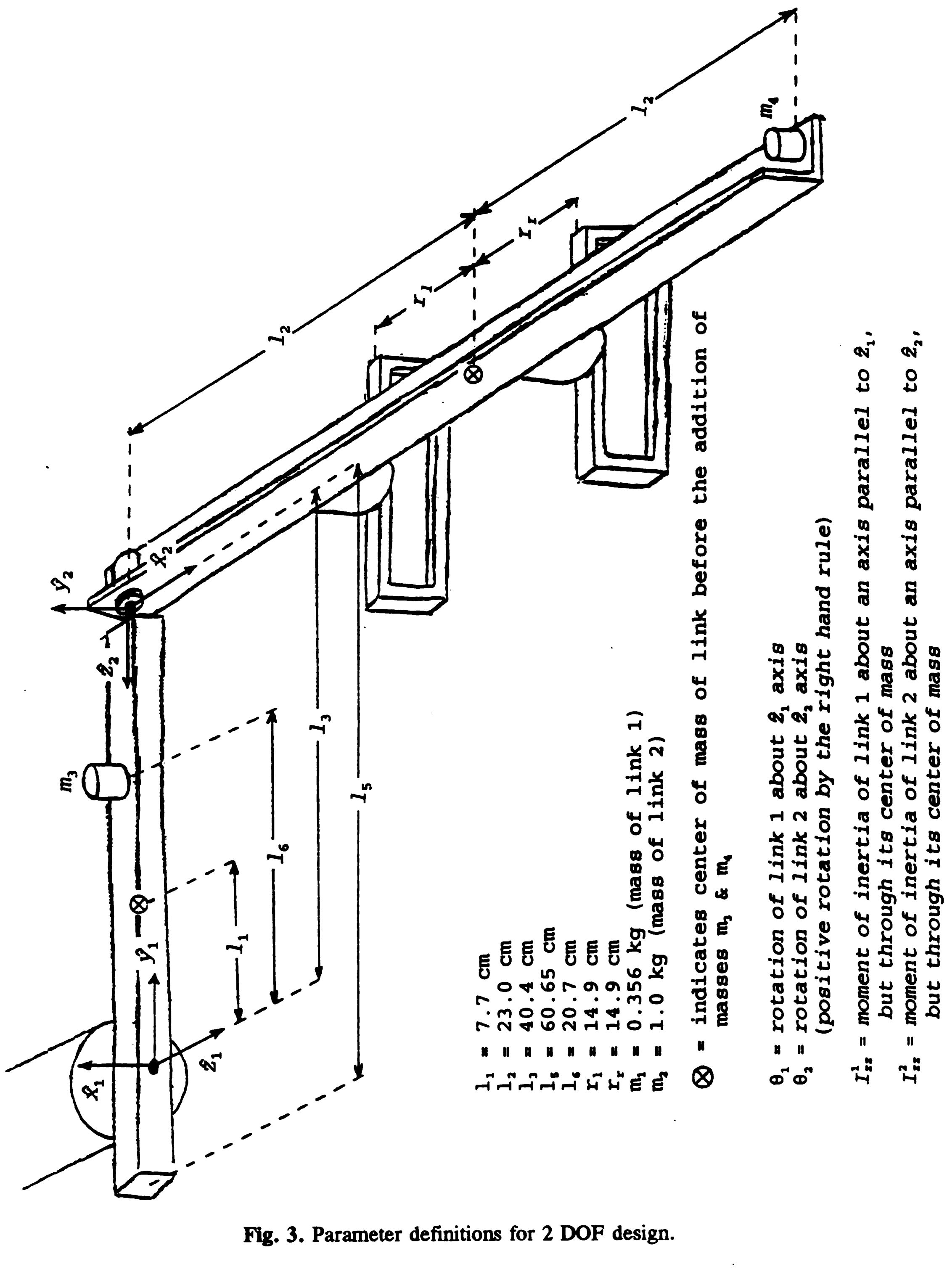

Fig. 3. Parameter definitions for 2 DOF design.

$$(I_{zz}^1+I_{xx}^2+m_1 l_1^2+m_2 l_3^2)\,\ddot{\theta}_1=l_3(2F-m_2 g\cos\theta_1)-l_1 m_1 g\cos\theta_1 \tag{7}$$

$$(m_2 l_2 l_3)\,\ddot{\theta}_1=F(r_r-r_l)+l_2(2F-m_2 g\cos\theta_1) \tag{8}$$

After combining these two equations and performing a few algebraic manipulations, we arrive at:

$$F[(r_r-r_l)(I+m_2 l_3^2)+2Il_2]-[Im_2 g\cos\theta_1 l_2+m_1 m_2 g\cos\theta_1 l_1 l_2 l_3]=0 \tag{9}$$

where

$$I \equiv I_{zz}^1+I_{xx}^2+m_1 l_1^2 \tag{10}$$

For the device to be *balanced*, we require equation (9) to be satisfied for all values of F. Thus we have two conditions which must be met :

$$I_{zz}^1+m_1 l_1^2 = m_1 l_1 l_3-I_{xx}^2 \tag{11}$$

$$r_l-r_r = \frac{2l_2 I}{I+m_2 l_3^2} \tag{12}$$

Equation (11) specifies the moment of inertia of the link 1 about its axis of rotation in terms of other device parameters. Equation (12) specifies the location of the center of mass of link 2 in terms of other device parameters.

We wish to *balance* the system by placing weights m_3 and m_4 at the locations specified in fig. 3. The value of m_3 can be determined from equation (11). By adding m_3 to link 1, we change its mass (m_1), center of mass (l_1) and inertia (I_{zz}^1). We'll denote these new parameters as $\mathbf{m_1}$, $\mathbf{l_1}$ and $\mathbf{I_{zz}^1}$.

$$\mathbf{m}_1 = m_1 + m_3 \tag{13}$$

$$\mathbf{l}_1 = l_1 + \frac{m_3(l_6 - l_1)}{m_1 + m_3} \tag{14}$$

We can now use equation (11) to derive an expression for m_3. Keep in mind that all the parameters in equation (11) should now reflect the addition of weights m_3 and m_4 (bolded values). Equation (11) can be used more easily if we recognize the left side as the moment of inertia of link 1 about its axis of rotation (with the added weight m_3). Using this fact, assuming link 1 is a slender rod, plugging in equations (13) and (14), and assuming I_{xx}^2 doesn't change with the addition of m_4, equation (11) yields:

$$\left(\frac{l_5^2}{12} + l_1^2\right) m_1 + m_3 l_6^2 = (m_1 + m_3)\left(l_1 + \frac{m_3(l_6 - l_1)}{m_1 + m_3}\right) l_3 - I_{xx}^2 \tag{15}$$

Solving for m_3 yields :

$$m_3 = \frac{m_1\left(\frac{l_5^2}{12} + l_1^2 - l_1 l_3\right) + I_{xx}^2}{l_3 l_6 - l_6^2} \tag{16}$$

A similar procedure can be followed to calculate m_4 using equation (12). Again, all the parameters must reflect the addition of weights m_3 and m_4. We denote new values as bolded. We also assume $r_l = r_r$ initially.

$$\boldsymbol{m}_2 = m_2+m_4 \tag{17}$$

$$\boldsymbol{l}_2 = l_2+\frac{m_4 l_2}{m_2+m_4} \tag{18}$$

$$\boldsymbol{r}_l-\boldsymbol{r}_r = \frac{2m_4 l_2}{m_2+m_4} \tag{19}$$

$$\boldsymbol{I}=m_1\left(\frac{l_5^2}{12}+l_1^2\right)+m_3 l_6^2+I_{xx}^2 \tag{20}$$

Plugging in equations (17-20) into equation (12), we get a quadratic in m_4:

$$m_4^2(l_3^2)+m_4(m_2 l_3^2-\boldsymbol{I})-m_2(\boldsymbol{I})=0 \tag{21}$$

Based on the numerical values in fig. 3, equations (16) and (21) can be used to calculate m_3 and m_4:

$m_3 = 0.089$ kg

$m_4 = 0.114$ kg

The addition of these two masses *balances* the device.

We simulated the device to verify the balancing calculations (Simnon nonlinear simulation program). Fig. 4a shows the position of the device in response to symmetrical inputs from the right and left hands. Notice there is considerable object tilt despite parallel inputs from the left and right hands. However after balancing weights m_3 and m_4 are added, the object has no tilt in response to symmetric inputs from the left and right

Fig. 4a Fig. 4b

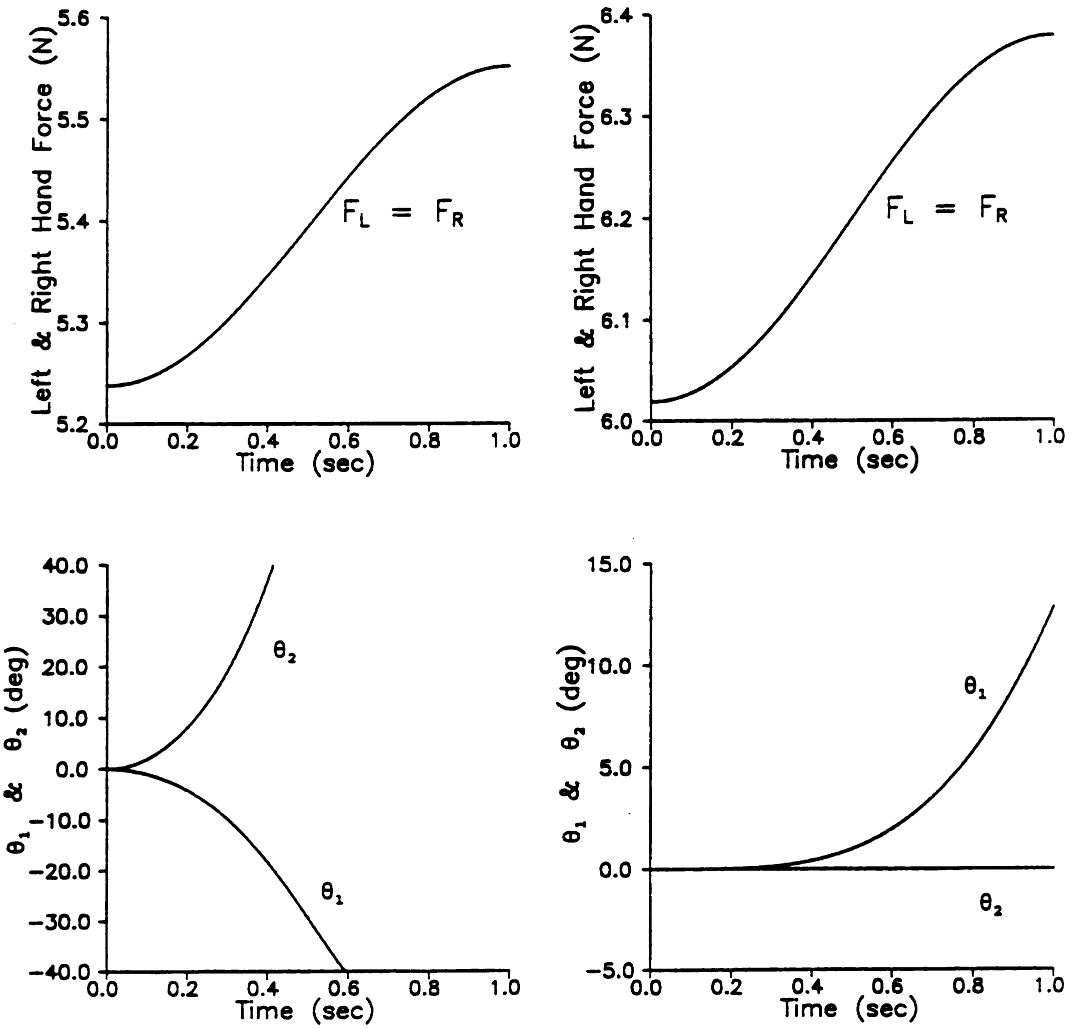

Fig. 4. Simnon simulation of the response of the 2 DOF device to symmetrical hand inputs before (a) and after (b) balancing.

hands and so behaves like a free unconstrained object (fig. 4b).

We also simulated the motor shown in fig. 3 controlled by a proportional feedback control law which regulated object tilt to zero. The effect of the feedback law was to change the $[F]_{external}$ matrix in equation (1) :

$$[F]_{external} = \begin{bmatrix} l_3 \cos\theta_2 (F_l + F_r) + K\theta_2 \\ l_2 (F_l + F_r) - r_l F_l + r_r F_r \end{bmatrix} \tag{22}$$

The feedback loop was simulated discretely at a sampling rate of 500 Hz. Fig. 5a shows the device position in response to asymmetric hand inputs. Notice the large object tilt in response to inadequate left hand force. Fig. 5b shows the device response to this same asymmetric input, but with the position feedback law implemented. The motor responds to differences in left and right hand force and generates torque to keep the object from tilting. By varying the feedback gain, the admissible steady state error can be varied.

III. Vertical transport coordination in normals

Methods

The 2 DOF device was modified slightly to test normal coordination. Since the motor was not to be used in these experiments, it was replaced with a low-friction bearing. Since link 1 would carry very little load, the original aluminum version was replaced with a lightweight wood version. After these modifications, the device was balanced as described in section II.

Subjects sat in front of the device which rested on a table. Initial values of Θ_1 and

Fig. 5a

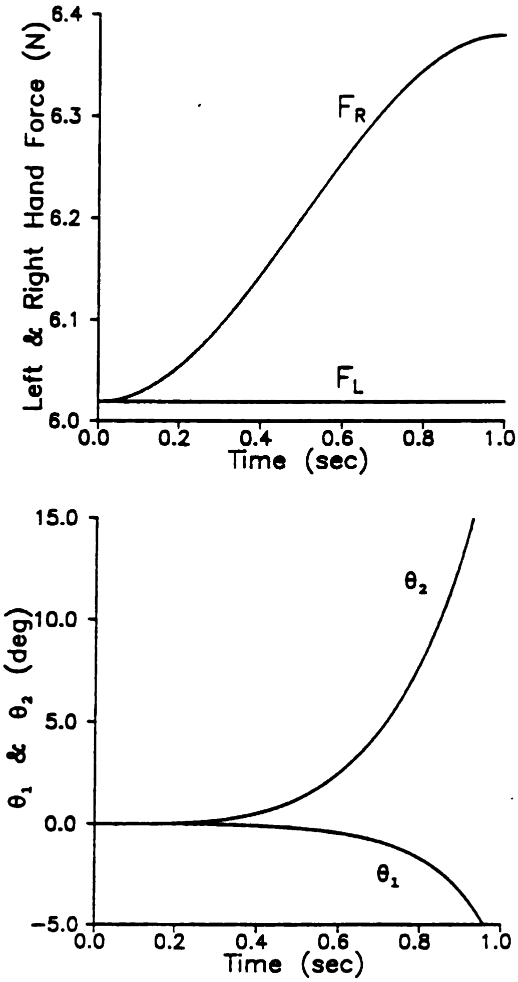

Fig. 5b

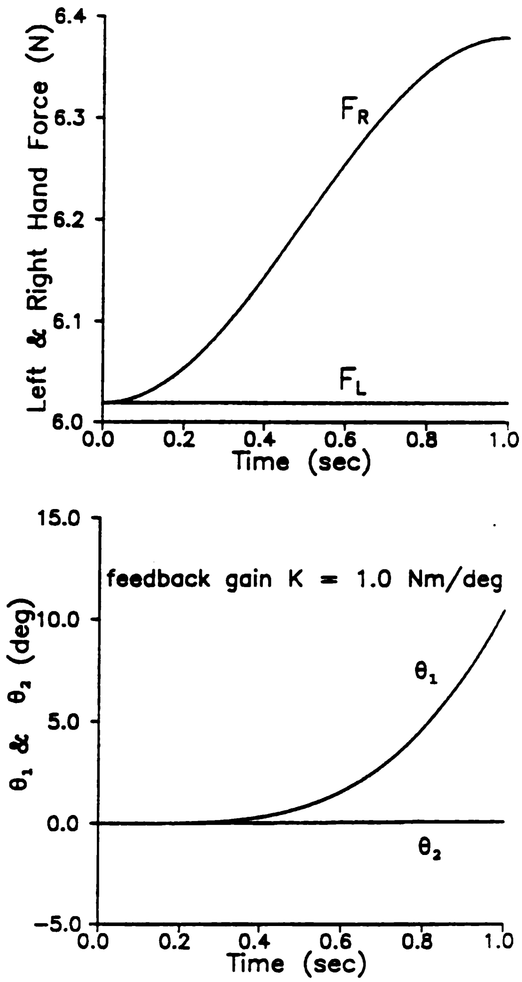

Fig. 5. Simnon simulation of the response of the 2 DOF device to asymmetrical hand inputs before (a) and after (b) the feedback law was implemented.

Θ_2 were -10 and 0 degrees. Subjects were presented Θ_1 and a 2 degree wide target on an oscilloscope. The initial target position was -10 degrees, so Θ_1 was already in the initial target as the device rested on the table. After the subjects grasped the two handles, the target was moved to a location corresponding to 10 degrees. Subjects lifted the object into the new target (lifting phase), paused for a second (holding phase) and then replaced the object on the table (placing phase). Since the axis of rotation of the elbow was close to the axis of rotation of link 1, subjects lifted the device predominantly with elbow flexion. One instruction was to move as fast as possible into the target. A second instruction was move at a natural pace. Both force transducer traces, Θ_1 and Θ_2 were recorded for thirty trials under each instruction. Trials were synchronized on peak Θ_1 velocity and ensemble averaged. Three subjects were tested.

To test asymmetric coordination, a 9 pound weight was added to one of the forearms. Subjects were asked to perform the movements as before. The left forearms of two subjects were weighted and the right forearms of two other subjects were weighted.

Results

The timing and shape of the force trajectories from the hands were similar for all subjects. Figs. 6 & 7 summarize performance of the lifting and holding phase under the two speed instructions and for three subjects. In the lifting phase, all subjects delivered similar left and right hand force trajectories, but subject AN biased her left hand force up. In the holding phase, subjects PL and AN preferred to hold with more force in their

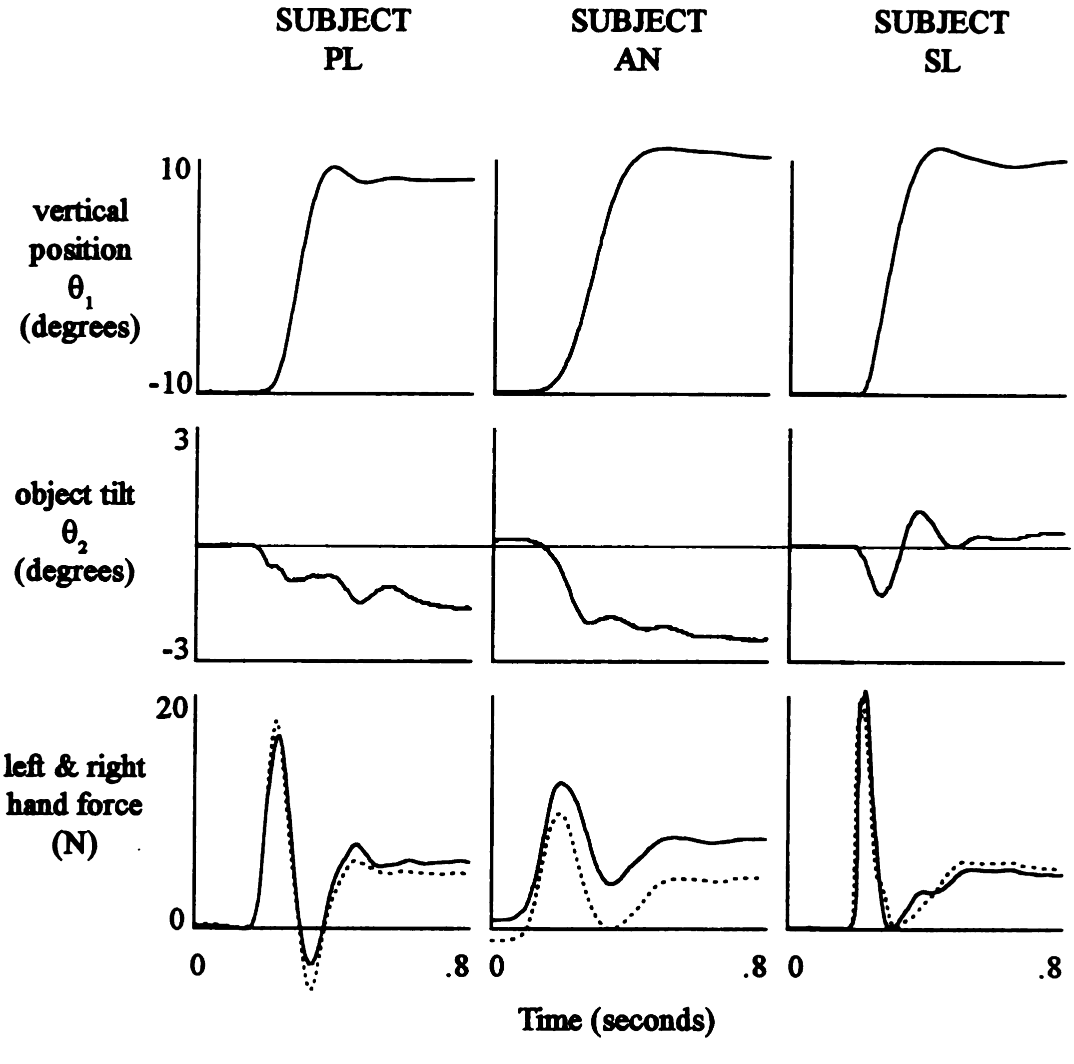

Fig. 6. The lifting and holding phases under the "fast as possible" instruction for three subjects. In the third row, solid lines are the left hand forces, dotted lines are the right hand forces. All trajectories are ensemble averages of 30 trials.

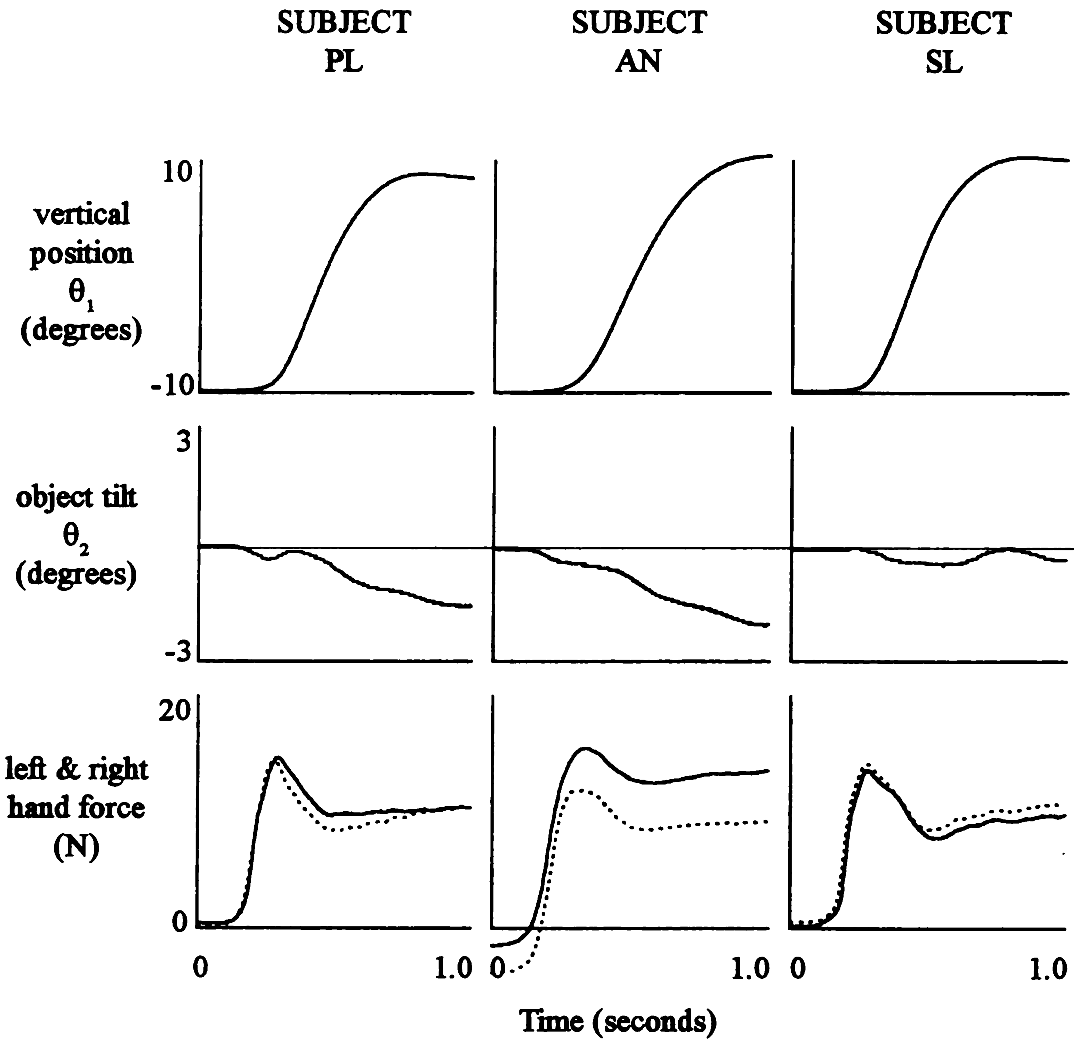

Fig. 7. The lifting and holding phases under the "natural paced" instruction for three subjects. In the third row, solid lines are the left hand forces, dotted lines are the right hand forces. All trajectories are ensemble averages of 30 trials.

left hands vs. their right hands while subject SL held the object with more force in his right hand. This characteristic partitioning of force between the hands was also reflected in the placing phase (fig. 8). In general, subjects PL and SL preferred to divide the required force equally between the hands, while subject AN preferred to use her left hand consistently more.

Despite some asymmetries in the force delivered by the hands (especially in the case of subject AN), the object was moved by all three subjects with less than 3 degrees of tilt. Subjects must have prevented large object tilts by generating torques at the handles in order to balance any asymmetry in vertical hand force. It would have been possible to prevent subjects from applying torques to the object with specially designed handles, but since it appears torque generation at the handles is a strategy used by normals, we might not want to eliminate that possibility when retraining stroke patients.

Subjects can produce precise asymmetric torques at the elbows in vertical transport. Fig. 9 overlays lifting performance with and without a 9 pound weight attached the right forearm. Rather surprisingly, subject PL's movement velocity was not slowed by the added weight and the peak force delivered to the object by the loaded right hand was only marginally smaller. In contrast, subject SL's movement velocity was reduced 16% and the force trajectories were markedly different after the weight was added. Weighing the right arm of subject SL caused the force pulses from *both* hands to be smaller in amplitude and wider. Whether or not the added weight affected the right hand force trajectory, the subjects' left hand force trajectories always mirrored the right hand force trajectories. In order to accomplish this, the torques about the left and right elbows

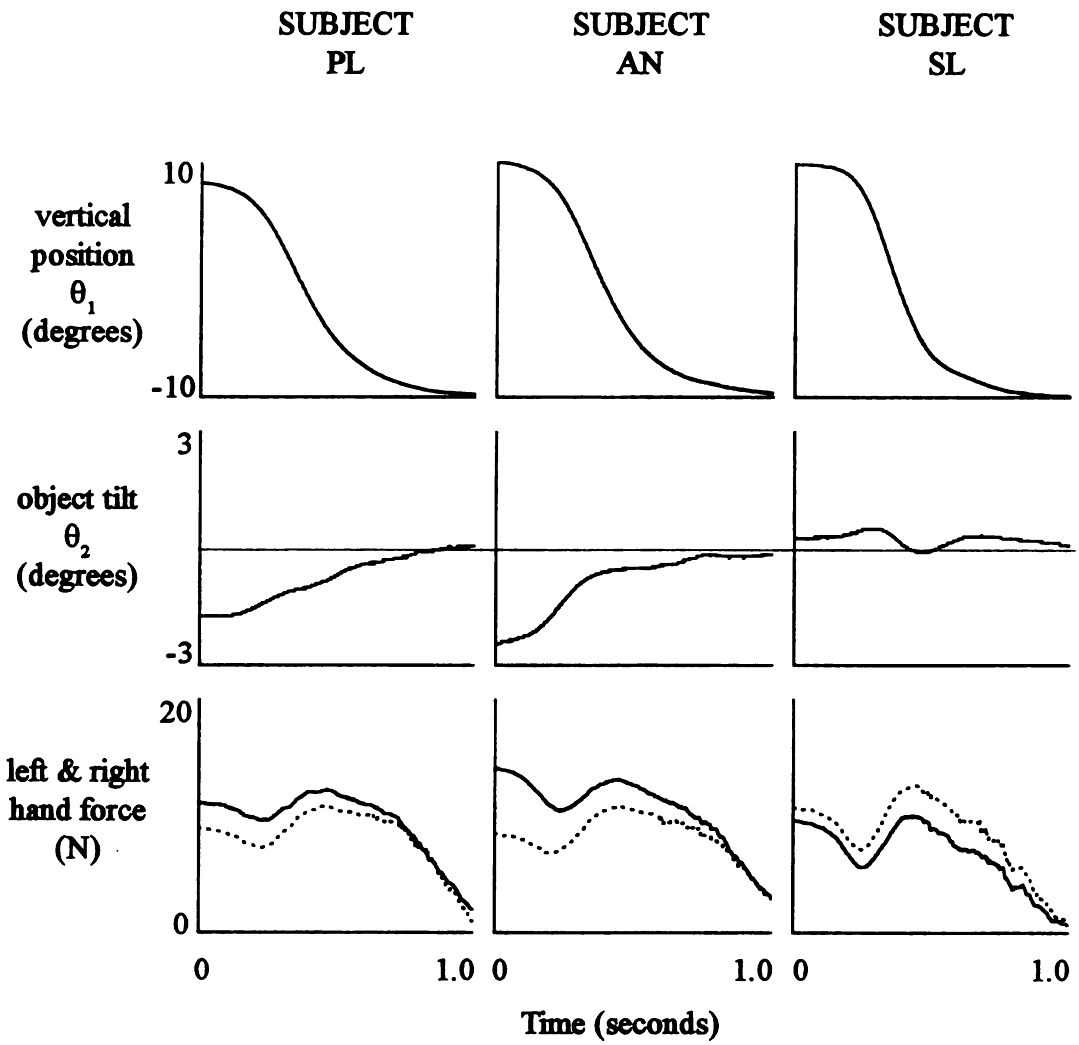

Fig. 8. The placing phase for three subjects. In the third row, solid lines are the left hand forces, dotted lines are the right hand forces. All trajectories are ensemble averages of 30 trials.

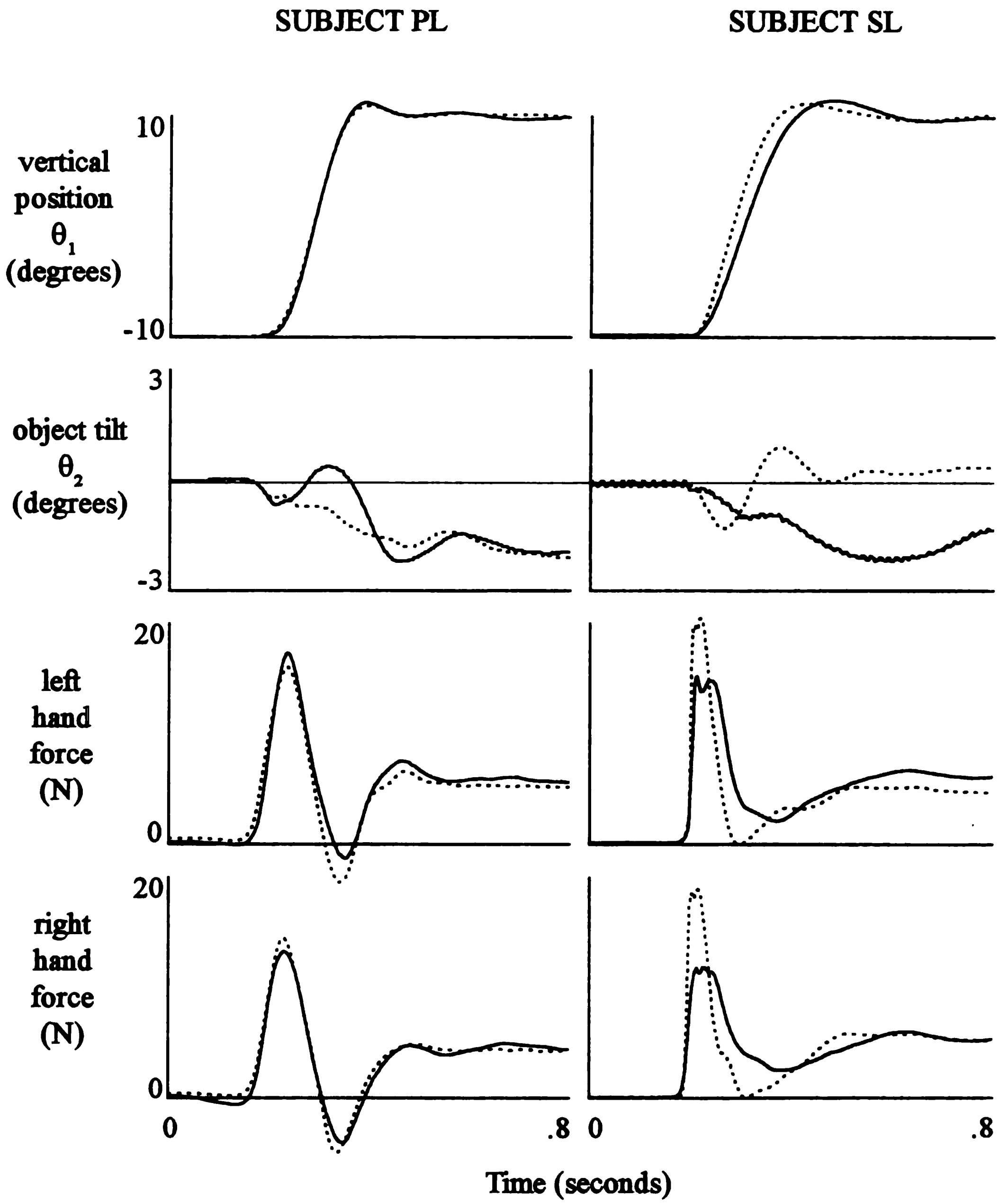

Fig. 9. The lifting and holding phase under the "fast as possible" instruction with (solid) and without (dotted) a 9 pound weight attached to the right forearm. All trajectories are ensemble averages of 30 trials.

must have been very different since the dynamics of the right forearm had been changed dramatically by the added weight. This asymmetrical coordination was precise enough to keep the object tilt less than 3 degrees throughout the movements.

Similar results occur if a 9 pound weight is added to the left forearm. Fig. 10 overlays lifting performance with and without a 9 pound weight attached the left forearm. Again, subject PL's movement velocity was only slightly decreased by the added weight. However subject AN's peak velocity was reduced 23%. The added weight caused her left hand to be biased downward before the start of the lift (she is resting her hand on the handle). Interestingly, the right hand is also resting on the handle. As a result of this bias, the force levels up to peak force in *both* hands lag behind trajectories delivered in the unweighted condition.

Discussion

Each subject has a coordinative structure which specified how the vertical force applied to the object was to be divided between the hands. Subjects PL and SL had coordinative structures which divided the task almost evenly between the hands, while subject AN preferred to use her left hand consistently more. These coordinative structures became evident when weights were added to only one hand. For subject PL, the added weights didn't greatly affect the force the weighted arm applied to the object. Consequently, the force trajectory from the unweighted arm also did not change dramatically. Weighing the right arm of subject SL caused the right force pulse to be smaller in amplitude and wider. His coordinative structure forced similar changes in his

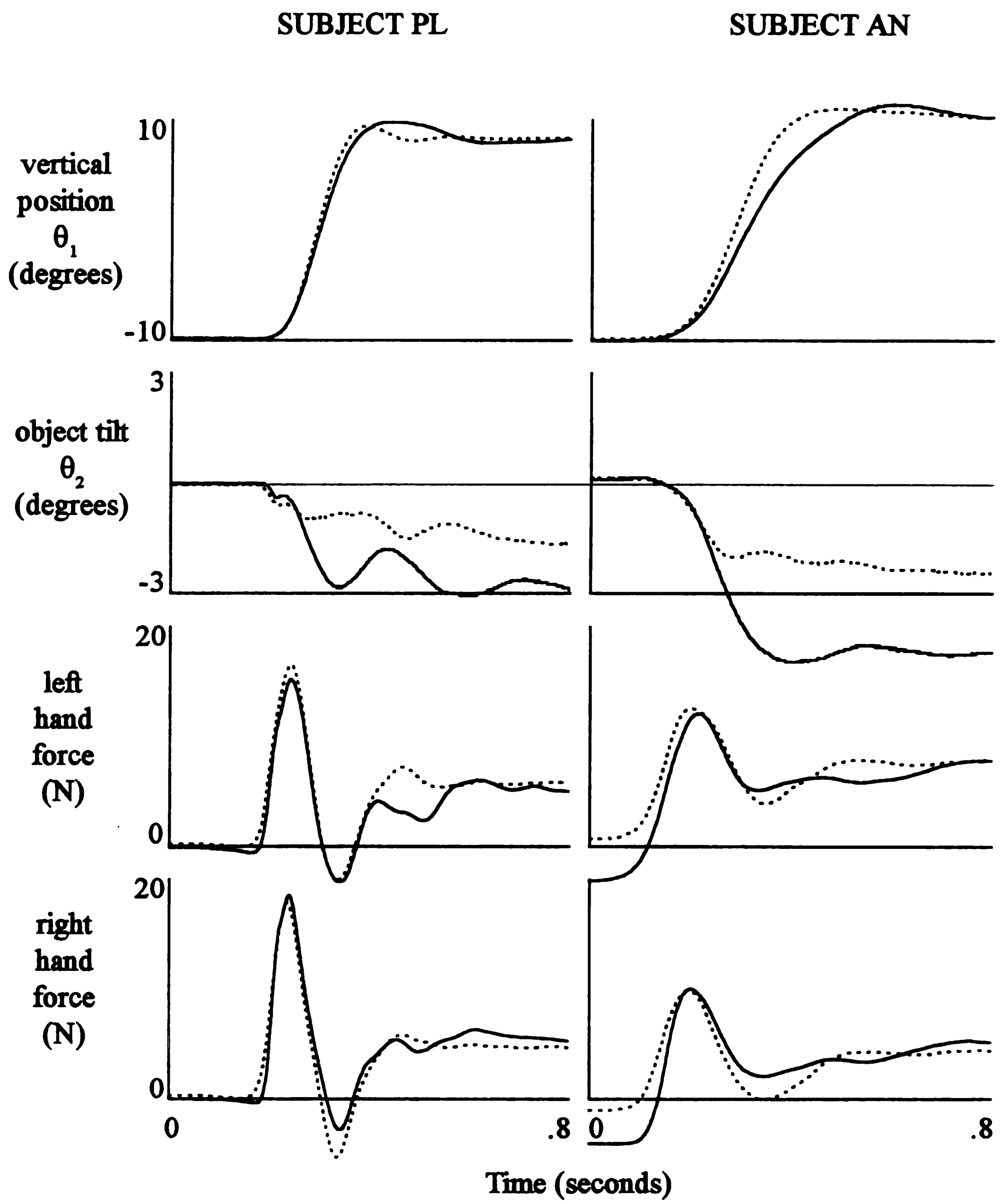

Fig. 10. The lifting and holding phase under the "fast as possible" instruction with (solid) and without (dotted) a 9 pound weight attached to the left forearm. All trajectories are ensemble averages of 30 trials.

left hand trajectory. Weighing the left arm of subject AN caused her left hand force to be biased in the negative direction and the force pulse to lag behind the force pulse delivered in the unweighted condition. Her coordinative structure forced similar changes in her right hand force trajectory, maintaining the right/left partitioning of the task preferred by this subject (more force from her left hand). For all subjects, the forces produced by the muscles of the two arms were necessarily *very* different after the addition of weights to one arm. The coordinative structure that specifies this asymmetrical muscle coordination was all the more impressive when one considers that subjects practiced with the additional weight less than 5 times before starting data collection.

We believe this coordinative structure can explain many observations concerning the fact that the two hands cannot be moved simultaneously independent of each other. It has been well documented that the hands have a strong preference for phase locking when being moved simultaneously. When moving the two hands to targets of different size, the hand moving to the easier target is slowed so that both hands start and stop moving at the same time [1,2]. This phase locking also exists in stroke patients. When stroke patients are asked to squeeze a ball in each hand as fast as possible, the able hand slows its frequency to match that of the disabled hand [3,4]. In fast bimanual elbow flexions, the able arm slows its velocity to match the velocity of the disabled arm [5]. This strong tendency to start and stop bimanual movements together is useful in tasks where the hands cooperate to manipulate an object. In the loaded movements we studied, the torque trajectories about each elbow must be very different due to the added weight and inertia added to one forearm. But the trajectories must also be phase locked (start and stop at the same time) in order to move the object with little tilt. We hypothesize that phase locking is a coordinative structure that has evolved to assist bimanual tasks where the two hands manipulate an object. An alternative hypothesis is that phase locking is a coordinative structure left over from quadrupedal locomotion, where the main function of the forelimbs was simultaneous phase locked movements. We may have to wait another 100 million years to determine which of these two hypotheses is true.

IV. Substitution test of the device

Methods

The device was returned to its original configuration as described in section II. The lightweight wood version of link 1 was replaced by the original aluminum version and the motor was re-mounted on the axis of rotation of link 1. The device was statically and dynamically balanced as described in section II. The effective weight of the object was adjusted to 1 kg. By effective weight we mean the net vertical force at the handles required to support the device with $\Theta_1=\Theta_2=0$.

Subjects performed lifting, holding and placing movements with the device as described in section III. The movement instruction was to move at a natural pace. To help subjects move consistently, the peak Θ_1 velocity was verbally fed back to the subject after each trial. A peak velocity range was determined for each subject. Only those trials that had a peak velocity within this range were recorded. After 30 trials were recorded, the motor was turned on and the subject performed the movements with just his right hand, with the motor substituting for the left hand. Again, the peak velocity was verbally fed back to the subject after each trial and only those trials that had a peak velocity in the desired peak velocity range were recorded. Thirty trials were recorded. Two subjects were tested.

The motor operated under a proportional-derivative feedback law regulating object tilt (Θ_2) to zero. The control law was :

$$T_m = K(\theta_2) + B(\dot{\theta}_2) \tag{23}$$

T_m is the motor command torque. The feedback law was implemented digitally at a sampling rate of 10,000 Hz. The object tilt was measured by a precision potentiometer, sampled and digitally low passed filtered. The low pass filter used was a 3 pole Butterworth filter [6] implemented digitally with a bilinear approximation [7]. The cutoff frequency for this filter was set at 1,000 Hz for these experiments, but a lower cutoff frequency might be used in the future to eliminate a 60 Hz chatter that originated in the power supply driving the potentiometer. Tilt velocity at sampling time n was calculated by running $(\Theta_2)_n$-$(\Theta_2)_{n-1}$ through another 3 pole Butterworth low pass filter. It was necessary to set this cutoff frequency at 30 Hz to eliminate 60 Hz noise that was amplified by the difference calculation. The proportional gain (K) was increased until the 60 Hz chatter became disconcerting and then decreased slightly (K=1.21 Nm/degree). With this high proportional gain, there were noticeable damped oscillations in the tilt variable during movements. This gave the movements an uncontrolled, unstable "feel". The derivative gain was increased until the oscillations were small enough to give the device a solid "feel" when lifting with one hand (B=0.0144 Nm*s/degree).

Results

With the motor substituting for the left hand, subjects can perform the lift, hold and placing movements with just their right hands. Fig. 11 overlays bimanual performance with performance after motor substitution. The Θ_1 kinematics were similar, but with the motor substituting for the left hand, the vertical movement was delayed; the tilt error must get large enough before the motor can generate enough torque to lift the left side of the object. Fig. 12 overlays bimanual performance of placing movements with performance after motor substitution. Again, the Θ_1 kinematics were similar for both subjects. Although the Θ_2 kinematics were very different, object tilt was maintained less than 4 degrees throughout the movements. This corresponds to a difference of only about 2 cm in the vertical position of the handles.

Subjects used their right hands differently when the motor substituted for the left hand. Force trajectories rose to a higher peak value during the lifting phase and stayed at a higher level throughout the holding and placing phases. In the holding phase, subject PL's right hand force increased from about 5.3 N to about 7.3 N after motor substitution. Subject SL's right hand force increased from about 5.6 N to 7.4 N in the holding phase. This 2 N increase in right hand force was apparent throughout the movement for both subjects.

The motor responds differently to the two subjects. The motor force is defined as the vertical force applied to the object (link 2) by the motor and calculated by dividing

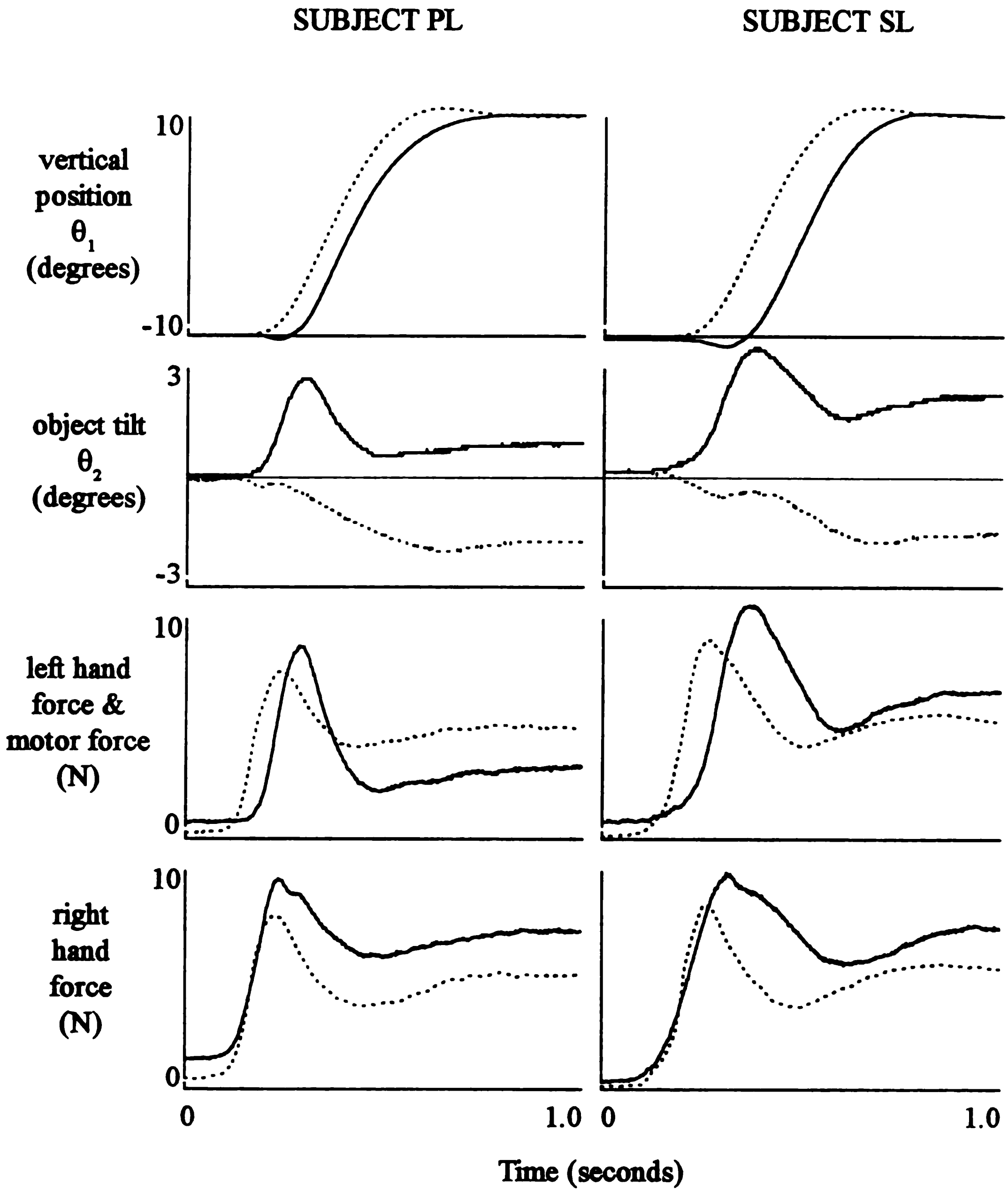

Fig. 11. The lifting and holding phase under the "natural paced" instruction performed bimanually (dotted) and with the motor substituting for the left hand (solid). Motor force is defined as the motor output torque divided by the distance from the axis of rotation of link one to link two. All trajectories are ensemble averages of 30 trials.

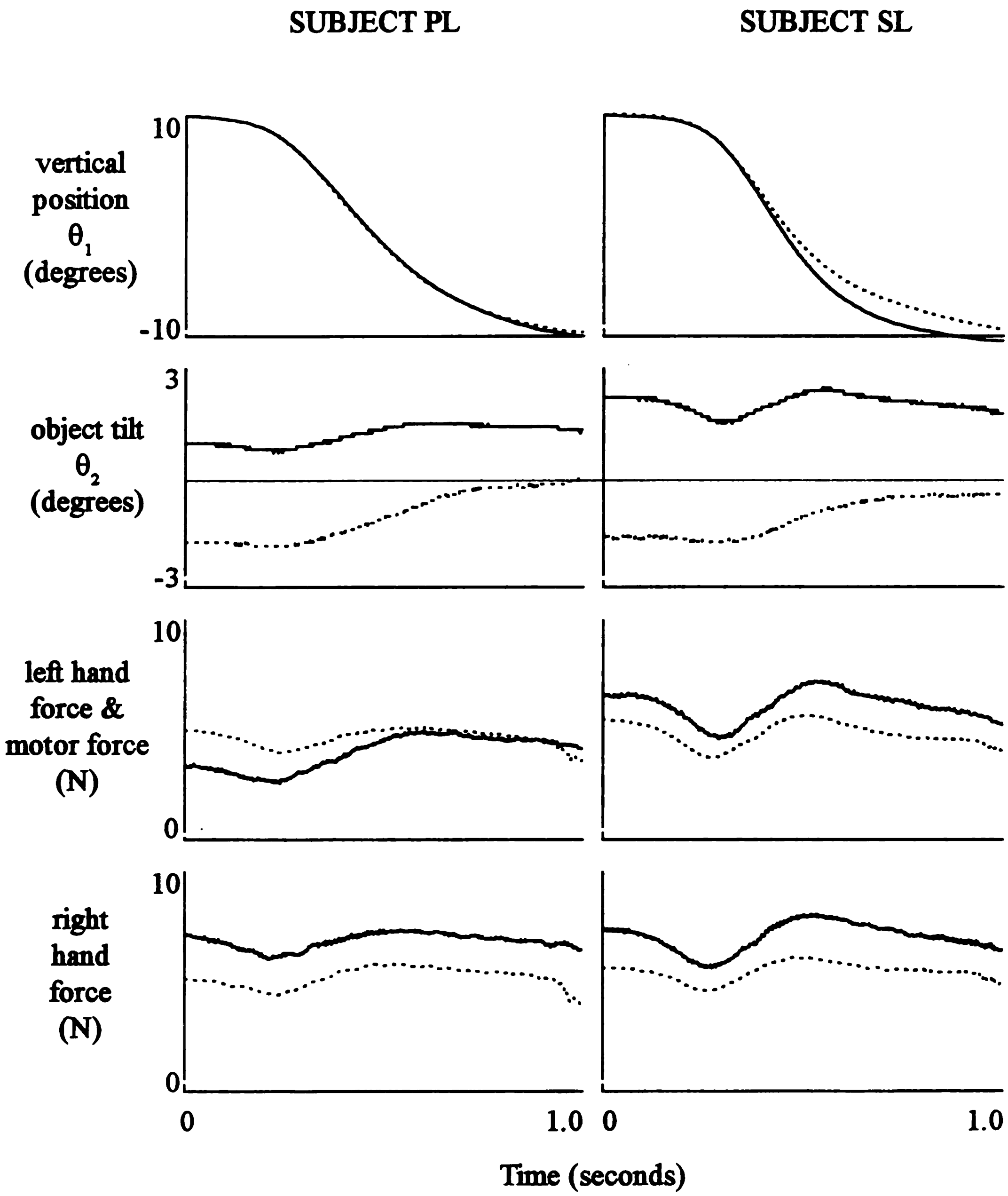

Fig. 12. The placing phase under the "natural paced" instruction performed bimanually (dotted) and with the motor substituting for the left hand (solid). All trajectories are ensemble averages of 30 trials.

the motor torque by the distance from the axis of rotation of link 1 to the object (l_3). For both subjects, the rise in motor force was delayed but then rose to a peak value that was larger than left hand force. In subject PL's holding phase, the motor force was about 3.3 N compared to a left hand force of about 5.1 N. In contrast, for subject SL, the motor force was 6.8 N compared with a left hand force of 5.4 N. This pattern of less motor force vs. left hand force in PL and greater motor force vs. left hand force in subject SL was also apparent in the placing phase of the movement.

Discussion

The object appears "heavier" to the right hand when lifting with the motor vs. lifting with the left hand. The right hand force trajectories rose to a higher peak after motor substitution and stayed higher throughout the movements. The peak in force corresponded to the onset of positive vertical object displacement. These force trajectories are what might be expected when lifting an object that was slightly heavier than before. The explanation for this "heavier" feel can be found in the device configuration. The device has been balanced so that to move the object perfectly with zero tilt, the force at the handles must be nearly identical. Normal performance indicated that about 10.7 N of force divided equally between the two hands was required to support the object statically. Consider the object (link 2) as a free body with a rotational degree of freedom about its center of mass. The motor substituted for the left hand by applying a force to the object that is farther away from its center of mass than the left handle (fig. 3). Assume the right hand produces 5.35 N of force as in the bimanual case. The torque applied to the object by this right hand force can be balanced by a smaller force when applied by the motor to the end of the object. But a net force of about 10.7 N is required to support the entire device. Consequently, both the right hand force and the motor force must be scaled up until their sum is equal to about 10.7 N. This explains subject PL's holding phase. His right hand force was about 7.3 N, and the motor force was about 3.3 N, which gives a sum of 10.6 N. Thus, the right hand must support more than half the weight of the device and feels "heavier".

The device substitutes for the left hand by adapting to right hand performance, allowing performance of the movements. In the holding phase, both subjects produce about 7.3 N of right hand force. What is confounding is how the object can be supported with a motor force of 3.3 N for subject PL and 6.8 N for subject SL. What is missing and what is not measured by the device is the torque that can be applied by the right hand. The sum of the right hand and motor force for subject SL in the holding phase was 14.2 N, which was significantly larger than the net force of 10.7 N needed to hold the object bimanualy. He must have generated a torque about the X axis of link 2 with his right hand after motor substitution. The motor adapted to this additional torque by increasing its output, allowing the movements to be performed. The reason the two subjects used their right hands differently might be related to rather vague instructions as to how to grab the handles. A subject that grasps the handles tightly is more likely to generate torques than a subject that allows the handles to rest in the palms of his hands. Where the handles are grabbed can also have an influence. If the handle is grabbed further back and not directly under the force transducer, a torque is generated about the object's X axis just by generating vertical force. More precise instructions in the future would eliminate this variation in subject performance.

IV. Summary & Future Work

We have designed and tested a prototype device which can measure and assist in bimanual vertical transport. When testing normal coordination on the device, each subject had a preferred way of dividing the required vertical force applied to the object between the hands. This partitioning was preserved even when the dynamics of one arm was changed dramatically with a 9 pound weight, forcing vastly different muscle forces in the two arms. We hypothesize the existence of a coordinative structure which performs this precise bimanual asymmetrical muscle coordination. When testing the ability of the device to substitute for the left hand, the device adapted to the performance of the right hand, allowing the task to be performed. However, the device felt "heavier" in substitution mode and subjects had a tendency to generate torques with their right hands as well as vertical forces.

Future work should concentrate on improvements to the prototype. 1) The left handle should be placed at the same location on the object as the point of application of assisting force. This would eliminate the "heavier" feel the device has in substitution mode. 2) Do we want to retrain patients to lift perfectly symmetrically with equal force from each hand or do we want to allow the able hand to do as much of the task as possible by generating torques as well as forces? This is an open question and perhaps future prototypes could be fitted with two types of handles: one set would allow the hands to apply torques to the object, the other set would not. The early stages of retraining would use the set that allows both torques and forces. After the patients can lift the device without assistance, the handles would be changed to the "force only" handles to further fine tune his coordination. 3) The tilt error can be decreased with a better control law. The proportional gain can be increased if the 60 Hz noise in the power supply driving the potentiometer can be reduced. Also, integral control can eliminate the steady state error. But do we really want a control law that decreases the tilt error as much as possible? For therapy, we may want patients to "feel" their errors in the form of object tilt so they can "try harder". Only if the patient has tried as hard as he can should the device provide assistance so that the task can be completed. In this case, we would want a "weak" control law that has perceptible tilt errors.

Appendix A

The following derivation is based on a general method which can be applied to any linkage [8]. First we define a base coordinate frame, coordinate frames attached to each link, and the two degrees of freedom (see fig. 3).

$\{\hat{x}_2, \hat{y}_2, \hat{z}_2\}^T$=*frame attached to link* 2 *(see fig.*3)
$\{\hat{x}_1, \hat{y}_1, \hat{z}_1\}^T$=*frame attached to link* 1 *(see fig.*3)
$\{\hat{x}_0, \hat{y}_0, \hat{z}_0\}^T$=*base frame that is coincident to frame* 1
in the reference configuration ($\theta_1=\theta_2=0$)
θ_1 = *rotation of link* 1 *about* $\hat{z}_1$ *relative to base frame*
θ_2 = *rotation of link* 2 *about* $\hat{z}_2$ *relative to frame* 1

In the following vector and matrix definitions, a leading superscript refers to the coordinate frame the vector or matrix is to be expressed in. The trailing subscript specifies the link. If the leading superscript is omitted, it is assumed the vector or matrix is to be expressed in the frame attached to the specified link.

P_n = vector from frame n to frame n+1
P_{c_n} = vector from frame n to link n center of mass
ω_n = angular velocity of link n
$\dot{\omega}_n$ = angular acceleration of link n
$\dot{V}_n$ = linear acceleration of frame n origin
$\dot{V}_{c_n}$ = linear acceleration of link n center of mass
F_n = sum of forces acting on link n
N_n = sum of moments acting on link n about its center of mass
f_n = force applied to link n by link n-1
η_n = moment applied to link n by link n-1
$[I]_n$ = inertia tensor of link n about its center of mass in principle coordinates

$$= \begin{bmatrix} I^n_{xx} & 0 & 0 \\ 0 & I^n_{yy} & 0 \\ 0 & 0 & I^n_{zz} \end{bmatrix}$$

The following rotation matrices can be easily derived:

$$R^0_1 = \begin{bmatrix} \cos\theta_1 & \sin\theta_1 & 0 \\ -\sin\theta_1 & \cos\theta_1 & 0 \\ 0 & 0 & 1 \end{bmatrix}$$

$$R^1_2 = \begin{bmatrix} \sin\theta_2 & \cos\theta_2 & 0 \\ 0 & 0 & -1 \\ \cos\theta_2 & -\sin\theta_2 & 0 \end{bmatrix}$$

These rotation matrices convert any vector expressed in the frame specified by the subscript into that same vector expressed in the frame specified by the superscript. The inverse of any of these rotation matrices is simply the transpose.

For each link, we calculate its angular velocity, angular acceleration, acceleration of the origin of its coordinate frame, and acceleration of its center of mass. Once the kinematics of the link have been determined, we use the Newton-Euler equations to relate the link kinematics to forces and moments applied to the link. By definition, the base link has no angular motion, but the effect of gravity can be taken into account if we assign the base coordinate frame an upward acceleration equal to gravity.

$$\dot{V}_0 = g\hat{X}_0$$

Now we can move on to link 1:

$$\begin{aligned}
\omega_1 &= \dot{\theta}_1 \hat{Z}_1 \\
\dot{\omega}_1 &= \ddot{\theta}_1 \hat{Z}_1 \\
\dot{V}_1 &= R_0^1 \dot{V}_0 = g\cos\theta_1 \hat{X}_1 + g\sin\theta_1 \hat{Y}_1 \\
\dot{V}_{C_1} &= \dot{\omega}_1 \times P_{C_1} + \omega_1 \times (\omega_1 \times P_{C_1}) + \dot{V}_1 \\
&= (g\cos\theta_1 + \ddot{\theta}_1 l_1)\hat{X}_1 + (g\sin\theta_1 - \dot{\theta}_1^2 l_1)\hat{Y}_1 \\
F_1 &= m_1 \dot{V}_{C_1} \\
N_1 &= [I]_1 \dot{\omega}_1 + \omega_1 \times [I]_1 \omega_1 \\
&= \ddot{\theta}_1 I_{zz}^1 \hat{Z}_1
\end{aligned}$$

Now we can move on to link 2:

$$\omega_2 = R_1^2\omega_1 + \dot\theta_2\hat z_2$$
$$= \dot\theta_1\cos\theta_2\hat x_2 - \dot\theta_1\sin\theta_2\hat y_2 + \dot\theta_2\hat z_2$$
$$\dot\omega_2 = R_1^2\dot\omega_1 + R_1^2\omega_1\times\dot\theta_2\hat z_2 + \ddot\theta_2\hat z_2$$
$$= \begin{bmatrix} \ddot\theta_1\cos\theta_2 - \dot\theta_1\dot\theta_2\sin\theta_2 \\ -\ddot\theta_1\sin\theta_2 - \dot\theta_1\dot\theta_2\cos\theta_2 \\ \ddot\theta_2 \end{bmatrix}$$
$$\dot V_2 = R_1^2\{\dot\omega_1\times{}^1P_2 + \omega_1\times(\omega_1\times{}^1P_2) + \dot V_1\}$$
$$= \begin{bmatrix} \sin\theta_2(\ddot\theta_1 l_3 + g\cos\theta_1) \\ \cos\theta_2(\ddot\theta_1 l_3 + g\cos\theta_1) \\ \dot\theta_1^2 l_3 - g\sin\theta_1 \end{bmatrix}$$
$$\dot V_{C_2} = \dot\omega_2\times P_{C_2} + \omega_2\times(\omega_2\times P_{C_2}) + \dot V_2$$
$$= \begin{bmatrix} -l_2\dot\theta_1^2\sin^2\theta_2 - l_2\dot\theta_2^2 \\ l_2\ddot\theta_2 - l_2\dot\theta_1^2\cos\theta_2\sin\theta_2 \\ l_2\ddot\theta_1\sin\theta_2 + 2l_2\dot\theta_1\dot\theta_2\cos\theta_2 \end{bmatrix} + \dot V_2$$
$$F_2 = m_2\dot V_{C_2}$$
$$N_2 = [I]_2\dot\omega_2 + \omega_2\times[I]_2\omega_2$$
$$= \begin{bmatrix} \ddot\theta_1\cos\theta_2 - \dot\theta_1\dot\theta_2\sin\theta_2 & \dot\theta_1\dot\theta_2\sin\theta_2 & -\dot\theta_1\dot\theta_2\sin\theta_2 \\ \dot\theta_1\dot\theta_2\cos\theta_2 & -\ddot\theta_1\sin\theta_2 - \dot\theta_1\dot\theta_2\cos\theta_2 & -\dot\theta_1\dot\theta_2\cos\theta_2 \\ \dot\theta_1^2\cos\theta_2\sin\theta_2 & -\dot\theta_1^2\sin\theta_2\cos\theta_2 & \ddot\theta_2 \end{bmatrix} \begin{bmatrix} I_{xx}^2 \\ I_{yy}^2 \\ I_{zz}^2 \end{bmatrix}$$

Now we can use these equations to calculate the internal forces and moments between links. We start with link 2:

$$f_2 = F_2 - (F_l + F_r)\hat y_2$$
$$\eta_2 + (-P_{C_2}\times f_2) + (r_r F_r - r_l F_l)\hat z_2 = N_2$$

The z component of this last equation is the equation of motion for link 2. We can now use these results to calculate the equation of motion of link 1.

$$f_1 = F_1 + R_2^1 f_2$$
$$\eta_1 - R_2^1 \eta_2 - (P_{C_1} \times f_1) - (P_2 - P_{C_1}) \times R_2^1 f_2 = N_1$$

The z component of this last equation is the equation of motion for link 1.

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Chapter 7

Summary

We have studied four bimanual contact tasks. 1) When one hand removes a load from the other hand, feedforward motor commands stabilize the position of the unloaded hand. 2) When an object is held between the fingertips of the hands and moved in the horizontal plane, the two hands are controlled as one hand. 3) When the two hands are holding an object by squeezing it, perturbations to one hand drive appropriate reflex responses in both hands. 4) When the two hands lift an object vertically, subjects have a preferred way of dividing the vertical force applied to the object between the hands. This partitioning is preserved even if it requires vastly different muscle forces from the two arms.

A general picture of bimanual coordination begins to emerge from these results. In focal\postural tasks (e.g. bimanual unloading), feedforward structures stabilize the postural hand. In focal\focal tasks (e.g. transport), feedforward structures partition the task between the hands in a preferred way. Also, special bimanual feedback structures allow the two hands to respond *together* to perturbations applied to only one hand. We hypothesize that these bimanual coordinative structures for contact tasks interfere with similar non-contact bimanual tasks.

In addition to scientific aims, this research has a clinical goal: to develop devices to be used to rehabilitate bimanual capability in patients, such as hemiplegic stroke

victims, who have lost motor ability in one side. To this end, we have prototyped two devices. The basic premise behind the devices is : motor learning is facilitated by being shown the proper way to perform a task, and then practicing the task while receiving appropriate assistance when you make errors. First, we prototyped a simple device that could substitute for and assist one hand in a 1 DOF horizontal transport task (chapter 4). We learned that the device, driven by a simple feedback law, could substitute for one hand, although somewhat imperfectly. We also learned that as one hand was progressively disabled by ischemia, the device performed only that part of the task that the disabled hand did not, allowing the movements to be made. Instead of perfecting this device, we used what we learned to design another prototype device that could assist in a useful ADL: vertical transport (chapter 6). This device could substitute for one hand in vertical transport, but the device felt "heavier" than when lifted bimanually. Despite its flaws, we believe this device will be ready for clinical testing with a few improvements.

The concept of this device can be adapted to other bimanual ADLs, developing a family of devices that could revolutionalize stroke rehabilitation therapy. Assuming clinical testing shows such devices to be useful for stroke patient rehabilitation, they can greatly improve the current state of physical\occupational therapy: 1) They can increase the availability of therapy by automating many of its repetitive aspects. 2) They can objectively measure patient progress, providing a platform for testing different therapies. 3) They can adapt to the patient more precisely than a human therapist, delivering only that amount of assistance that is needed.