

Coordinating Two Degrees of Freedom During Human Arm Movement: Load and Speed Invariance of Relative Joint Torques

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SUMMARY AND CONCLUSIONS

1. Eight subjects performed three series of pointing tasks with the unconstrained arm. *Series one* and *two* required subjects to move between two fixed targets as quickly as possible with different weights attached to the wrist. By specifying initial and final positions of the finger tip, the first series was performed by flexion of both shoulder and elbow and the second by shoulder flexion and elbow extension. The third series required flexion at both joints, and subjects were instructed to vary movement speed. We examined how variations in load or intended speed were associated with changes in the amount and timing of the electromyographic (EMG) activity and the net muscle torque production.

2. EMG and torque patterns at the individual joints varied with load and speed according to most of the same rules we have described for single-joint movements. 1) Movements were produced by biphasic torque pulses and biphasic or triphasic EMG bursts at both joints. 2) The accelerating impulse was proportional to the load when the subject moved "as fast and accurately as possible" or to speed if that was intentionally varied. 3) The area of the EMG bursts of agonist muscles varied with the impulse. 4) The rates of rise of the net muscle torques and of the EMG bursts were proportional to intended speed and insensitive to inertial load. 5) The areas of the antagonist muscle EMG bursts were proportional to intended movement speed but showed less dependence on load, which is unlike what is observed during single-joint movements.

3. Comparisons across joints showed that the impulse produced at the shoulder was proportional to that produced at the elbow as both varied together with load and speed. The torques at the two joints varied in close synchrony, achieving maxima and going through zero almost simultaneously.

4. We hypothesize that "coordination" of the elbow and shoulder is by the planning and generation of synchronized, biphasic muscle torque pulses that remain in near linear proportionality to each other throughout most of the movement. This linear synergy produces movements with the commonly observed kinematic properties and that are preserved over changes in speed and load.

ous work (Corcos et al. 1989; Gottlieb et al. 1989a,b) suggested that pulses of motoneuron excitation are programmed based on specific force requirements of the intended movement task (see also Hoffman and Strick 1989, 1993). From the task parameters, motoneuron excitation pulse patterns can be generated by specifying their heights, widths, and relative timing, and these lead to muscle contraction and force development (Gottlieb 1993). The movement trajectory is an emergent property of the muscle-load dynamics (Gottlieb et al. 1995b). Of course, muscles and their supporting reflexes are compliant so that force also depends on limb kinematics and cannot be specified by the muscle activation signal alone. We assume that when movements are made under predictable and well-known conditions, these properties are accounted for in the planning of the movement. Thus we can speak of specifying forces to move the limb/load system that operate in parallel with and relatively independently from compliant mechanisms.

With two or more joints, however, the muscles about each joint produce only one component of the torque. Motion of other limb segments produce interaction torques so that each muscle's contraction influences motion at every joint. As a consequence of the physics, the relationship between the muscle torque and joint rotation is complex, even with only two joints. For the same reasons, the relationships between the muscle activation patterns and motion are complex (Flanders et al. 1994). This complexity presents the CNS with an apparent surplus of degrees of freedom for solving any individual kinematic task. Bernstein expressed this in his well-known statement: "The coordination of a movement is the process of mastering redundant degrees of freedom of the moving organ, in other words its conversion to a controllable system" (Bernstein 1967, p. 127). Little is known, however, of just how the CNS does this.

In a sense, even rules for single-joint motion are an example of such a mastering. That is, a set of relatively simple rules are used for agonist/antagonist muscle activation and coordination that are neither unique nor optimal according to any obvious criterion. Neither are they merely expressions of biomechanical constraints on neuromuscular activity (Gottlieb 1996). Such rules reduce the problem of control from deciding which of a virtually infinite set of potential control strategies to use to one of finding a small set of parameters for a specific set of control algorithms. This we have termed an "adequate" control strategy (Gottlieb et al.

INTRODUCTION

For single-joint movement of the elbow, the only torques acting on the forearm are those produced by the elbow flexor and extensor muscles and gravity. Hence there is necessarily a close and simple relationship between net muscle torque and voluntary joint motion. These movements have been shown to be accomplished by stereotyped activation of the agonist and antagonist muscles in a biphasic or sometimes triphasic electromyographic (EMG) burst pattern (Angel 1974; Gottlieb et al. 1989b; Hallett et al. 1975). Our previ-

1995a). With movements involving more than one joint, however, the coordination of motion across joints must also be addressed. It is possible that the same kind of approach might apply to this problem as well. That is, there may be rules that relate the simultaneous activation of the muscles at different joints that again reduce the problem to one of finding task-specific parameters for those rules. What might those rules be? In what follows, we will simplify the discussion of this question to the problem of controlling movements of two joints, the elbow and shoulder, moving in a sagittal plane.

The simplest rule one could have across joints is to make their torques linearly proportional to each other. This was first suggested by Lissajous plots of elbow versus shoulder torque during arm movements performed with different inertial loads and at different speeds (Hong et al. 1994). Because the net muscle torque patterns for these movements are simple, biphasic pulses, these relatively straight lines imply that the peaks and zero crossings of the torques at the two joints must be closely coincident in time. A similar observation was made by Buneo et al. (1995) for planar arm movements in different directions. This has also been shown to be true for arm movements in which only a single joint (elbow or shoulder) was intentionally moved (Almeida et al. 1995; Gottlieb et al. 1996). The observation that a linear relationship might exist between joint torques is a surprising and provocative discovery if it is true for more than a small set of special movements.

To explore this, we analyzed a series of experiments involving pointing movements of the arm with different weights attached to the wrist or at different intended speeds. Targets were positioned to require shoulder flexion and either flexion or extension at the elbow. One issue we address is how the agonist and antagonist muscles at an individual joint are controlled to adapt to task-specific changes (load/speed) during such multijoint movements. The second issue these experiments were designed to explore was how the actions of the individual joints relate to each other. Preliminary results have been presented in Hong et al. (1994).

METHODS

Subjects stood at ease and faced a small target (a cotton ball, 2 cm diam) positioned so that movement of the right arm was performed in a sagittal plane. *Tasks 1* and *2* used four different weights on the wrist. The movements were performed as fast as possible between two stationary targets. The first task, illustrated in Fig. 1A, started with the right arm relaxed at the side and required a net flexion of ~30° at both shoulder and elbow. We will refer to this as the FF-Load task. The second task, illustrated by Fig. 1B, required ~40° of shoulder flexion and ~10° of elbow extension. We will refer to this as the FE-Load task. The reasoning behind selecting these two tasks is that simultaneous flexion at both joints might be comparable with two simultaneous, "single-joint" flexion movements. The second task, however, although requiring shoulder flexing torque to initiate the movement, does not necessarily require elbow extension torque from the elbow joint muscles. The flexing action at the shoulder by the shoulder muscles will simultaneously act to extend the elbow and could be exploited by the nervous system to minimize muscle contraction. Both of these tasks were performed with the unloaded arm and with three inertial loads (0.9, 2.2, and 3.12 kg, respectively) attached to the wrist with Velcro straps.

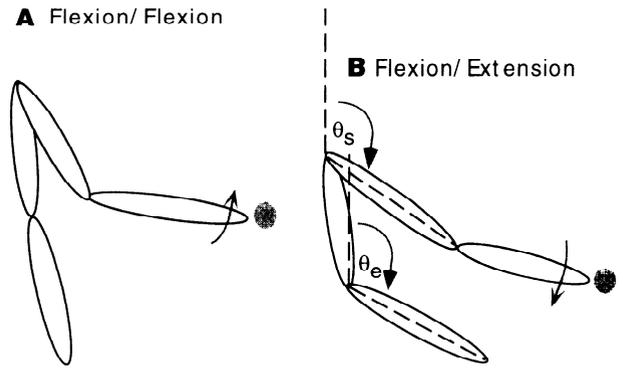


FIG. 1. This cartoon shows the initial and final limb positions of the subjects who performed the 3 movement tasks. The 1st task, performed with different weights strapped to the wrist, and 3rd task, performed at intentionally different speeds, began and ended from positions indicated by A. These movements required flexion of the shoulder and the elbow joints. The 2nd task, also performed with different weights strapped to the wrist and illustrated by B required flexion of the shoulder and extension of the elbow. Final positions are indicated by the gray target dot. The dotted arrow shows the relative elbow motion, which differed in the 2 halves of the figure. Dashed heavy lines indicate the coordinate system used to define rotation of each joint segment. In this coordinate system, forearm rotation was in the same, counterclockwise direction for all movements.

The third task was initiated from the same posture to the same target as the first task, but the movements were performed at four different speeds. A 0.9-kg weight was attached to the wrist. The instructions were "move as fast as you can," "move fast but not at your maximal speed," "move at a comfortable speed," and "move slowly." We will refer to this as the FF-Speed task.

No instructions were given about the hand path. On a verbal "get ready" signal, subjects positioned their arm at the starting position until the experimenter said "go," at which they reached out to the target, staying there until they heard a computer-generated tone. Movements were visually monitored during the experiments to make sure there was no significant out-of-plane motion. Eight adult male subjects gave informed consent according to medical center-approved protocols and then performed 10 trials for each load or speed.

Kinematic/dynamic analysis

A three-dimensional, electrooptical motion measurement system (OPTOTRAK-3010) recorded the locations of four markers attached to the shoulder, elbow, wrist, and index finger tip.

A simplified model of the kinematic linkage of the human arm was used that includes sagittal plane shoulder, elbow, and wrist joint rotations. Joint angles and their derivatives were calculated from the measured coordinate data of the distal and proximal segment endpoints. Muscle torques were computed by Newtonian equations of motion shown below in simplified form. The actual dynamic analysis of these movements was based on five degrees of freedom. These were horizontal and vertical, sagittal plane translation of the shoulder, and sagittal plane rotation about shoulder, elbow, and wrist. The data and analysis presented in this manuscript are of two of those degrees of freedom, shoulder and elbow rotation. The angles of the joint segments θ_s and θ_e are defined in Fig. 1B. The angle of the elbow joint is given by $\phi = \theta_e - \theta_s$

$$\text{Elbow Torque} = I_e \ddot{\theta}_e + r_e l_e m_e \cos \phi \ddot{\theta}_s + r_e l_e m_e \sin \phi \dot{\theta}_s^2 + r_e m_e \sin \theta_e g \quad (1)$$

$$\text{Shoulder Torque} = (I_s + l_s^2 m_s) \ddot{\theta}_s + (r_s l_s m_s \cos \phi) \ddot{\theta}_e - r_e l_e m_e \sin \phi \dot{\theta}_e^2 + (r_s m_s + l_s m_s) \sin \theta_s g + \text{Elbow Torque} \quad (2)$$

We have included gravitational terms appropriate to vertical plane movements and have explicitly used the absolutely referenced angles of the two limb segments with elbow joint angle (ϕ) shown only for notational simplicity. The lengths of the upper and lower limb segments are l_u and l_l , and their centers of mass are located r_u and r_l from their proximal ends. These equations represent the net torque produced by all the muscles about each joint. To perform these calculations, the inertial parameters of upper arm, forearm, and hand (mass, location of mass center, and principal moment of inertia) were estimated with the use of statistical data (Winter 1979) and measurements of whole body weight and limb lengths of each subject. Each additional weight attached to the wrist was assumed to be a point-mass located at the joint center.

The focus of this paper is on the transient pulses of torque that propel the limb toward and arrest it at its intended target. On these are superimposed the static torque requirements for resisting gravity. We assumed the separability of the two components, a static one proportional to gravity and a dynamic one independent of it. The gravitational component is a function of angle and load and is directly computed from *Eqs. 1* and *2* with all derivatives set to zero. Net muscle torques, including the gravitational component, were illustrated for one of our subjects performing these experiments in Hong et al. (1994). Here we show (in Fig. 2) the effects of removing the gravitational terms from the analysis for the same subject. This residual torque, computed by setting $g = 0$ in *Eqs. 1* and *2*, we will refer to as the dynamic muscle torque. We also computed the time integral of the dynamic muscle torque from movement onset to its first zero crossing and refer to this as the impulse.

The dynamic muscle torques analyzed here were always biphasic with distinct acceleration and deceleration phases. For all of the experiments described here, the first peak was always into flexion and the second into extension at both joints. We measured three temporal landmarks of the biphasic torque pulse; the time to the first extremum into flexion (t_f), the time of reversal when the torque crossed zero (t_z), and the time of the second extremum that was into extension (t_e). It made little difference if these times were measured from total or dynamic torque records if the movements were as fast as possible. For intentionally slow movements, however, the dynamic components became small in comparison with the gravitational terms, and t_z could sometimes be defined only for dynamic torque because the total torque did not go through zero.

To compare torque patterns across joints and task variables, we performed the following normalization on the dynamic torque terms. First we divided the dynamic torque for each joint and movement by its own first peak into flexion (t_f). Second we scaled the time axis for both joints by t_{zs} , the torque zero crossing time measured at the shoulder. Normalized torques are defined by *Eqs. 3* and *4*

$$\hat{\tau}_s(t) = \frac{\tau(t/t_{zs})}{\tau_{fs}} \quad (3)$$

$$\hat{\tau}_e(t) = \frac{\tau(t/t_{zs})}{\tau_{fe}} \quad (4)$$

EMG analysis

EMG surface electrodes (pediatric electrocardiographic electrodes with 2 cm between centers) were taped over the bellies of the biceps brachii, triceps (lateral head), and anterior and posterior deltoid muscles. The EMG signals were amplified, full-wave rectified, and low-pass filtered [100-Hz Paynter filter (Gottlieb and Agarwal 1970)]. All signals were sampled at 200/s.

We assumed that, like the muscle torque, the EMG can be partitioned into two additive components, a static component that depends on position and a dynamic component that is a function of the velocities and accelerations of the limb segments. At a movement's

endpoints the static component accounts for 100% of the EMG and at intermediate times is a proportional function of the instantaneous position of the limb. We subtracted this component from the measured EMG signal before performing any analyses, and these waveforms are shown in Fig. 2. Although the net static torque component depends only on gravity, the static EMG component in each muscle also depends on muscle elastic forces and the degree of cocontraction by its antagonist. Thus the amount of static EMG activity is probably in excess of what can be accounted for by the need to resist gravity. Our method is similar to subtracting the EMG recorded during a very slow movement from those of movements made at higher speeds (Buneo et al. 1994) and serves the same purpose of removing that component of the EMG signal that scales linearly with joint angle.

From these phasic EMG components, we computed the areas of the flexor bursts (Q_{ag}), and the area of the antagonist burst (Q_{ant}), integrated from movement onset to the time hand velocity fell to 5% of its peak. To obtain a measure of the slope of the rising phase of the EMG burst (Q_{rise}), we integrated a 40-ms window centered around the time the agonist EMG reached 50% of its peak, including only the roughly triangular area of increase in that interval.

To pool data across subjects for some analyses, we normalized EMG and impulse measures by dividing each subjects' values at each load (or speed) by its average for the four loads (or speeds).

RESULTS

Single-joint EMG and torque dependence on task variables

We illustrate our findings for the three tasks with representative data from one subject in Fig. 2. A statistical analysis [a single factor repeated measures analysis of variance (ANOVA)] of data from all eight subjects is given in Table 1. The figure is similar to Fig. 1 of Hong et al. (1994), but here we have removed the static components of both torque and EMG. The torque waveforms (2nd row), corrected by removal of the gravity-dependent components, are all biphasic pulses. For FF-Load and FE-Load tasks (Fig. 2, A and B), muscle torques at both joints initially rise into flexion at load-independent rates, in spite of the different intended direction of the elbow. For the FF-Speed task shown in Fig. 2C, the torques initially rise at speed-dependent rates. At both joints, there is a highly significant correlation between impulse and four load conditions in both Load experiments or with the four instructed speeds in the Speed task as shown in Table 1. Impulse at the shoulder was always greater than at the elbow as would be expected from *Eq. 2*.

The *bottom panels* in Fig. 2 show the amplitude and time-normalized torques for each joint. The accelerating peaks are all identically unity as is the time of zero crossing for the shoulder due to the normalization procedure. Note that the elbow zero crossings do not deviate far from unity (they are normalized on the shoulder's zero crossing time), and the deceleration peaks are also nearly coincident except for the slowest, FF-Speed movement.

The flexor muscle EMG bursts rise for longer times and have longer durations at both joints with increased loads. The burst durations in the FF-Speed task shown in Fig. 2C do not appear to be strongly sensitive to movement speed. Increases in inertial load and intentional increases in movement speed are both associated with increases in the areas of the agonist bursts (Q_{ag}) in the elbow and shoulder flexors.

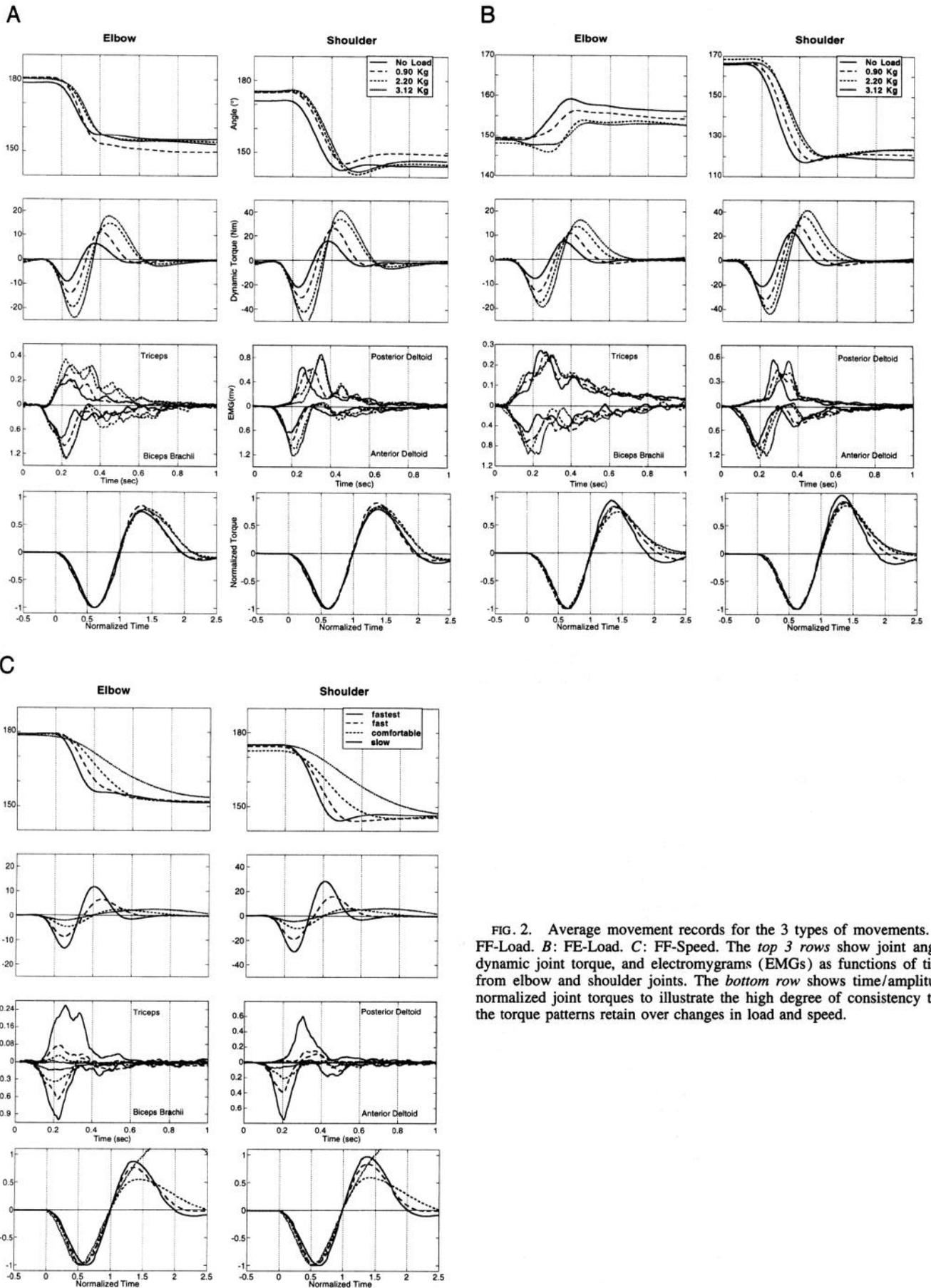


FIG. 2. Average movement records for the 3 types of movements. A: FF-Load. B: FE-Load. C: FF-Speed. The top 3 rows show joint angle, dynamic joint torque, and electromygrams (EMGs) as functions of time from elbow and shoulder joints. The bottom row shows time/amplitude normalized joint torques to illustrate the high degree of consistency that the torque patterns retain over changes in load and speed.

TABLE 1. Statistical analysis of data from all eight subjects

| Variable/Task | FF-Load | | FE-Load | | FF-Speed | |
|------------------|---------|---------|---------|---------|----------|---------|
| | F(3,21) | P | F(3,21) | P | F(3,21) | P |
| Elbow impulse | 70.67 | <0.0001 | 105.66 | <0.0001 | 43.143 | <0.0001 |
| Shoulder impulse | 37.35 | <0.0001 | 81.973 | <0.0001 | 46.485 | <0.0001 |
| Bi- Q_{ag} | 19.575 | <0.0001 | 12.542 | <0.0001 | 22.407 | <0.0001 |
| AD- Q_{ag} | 4.402 | 0.0149 | 4.644 | 0.0121 | 25.683 | <0.0001 |
| Bi- Q_{rise} | 1.14 | 0.354 | 4.19 | 0.108 | 14.5 | <0.0001 |
| AD- Q_{rise} | 0.531 | 0.666 | 0.814 | 0.50 | 52.6 | <0.0001 |
| Tri- Q_{ant} | 0.618 | 0.611 | 0.288 | 0.834 | 37.41 | <0.0001 |
| PD- Q_{ant} | 1.556 | 0.230 | 9.374 | 0.0004 | 54.8 | <0.0001 |

Impulse and electromyographic measures for 3 tasks are correlated with the 3 tasks and 4 conditions. The FF-Load and FE-Load columns show $F(3,21)$ and P values for a repeated measures analysis of variance with 4 loads. The FF-Speed column shows the values with 4 speeds. FF-Load, task with net flexion of $\sim 30^\circ$ at both shoulder and elbow; FE-Load, task with $\sim 40^\circ$ of shoulder flexion and $\sim 10^\circ$ of elbow extension.

The dependence of Q_{ag} on load is significant for all three tasks as shown in Table 1.

For movements of only a single joint, torque and the area of the agonist EMG burst are always highly correlated, regardless of the task (Gottlieb et al. 1989b), but the rates at which those variables rise are sensitive to the type of task performed. We analyzed the effects of load and speed on the rate of rise of the flexor muscle EMG bursts (Q_{rise}). For the FF-Load task, load has no significant effect (Table 1). These results are consistent with what is seen during movements of a single joint. For the FE-Load task, load increases the rate of EMG rise slightly in biceps, and the effect is significant at the 0.05 level. Post hoc, pair-wise analysis shows that the values of Q_{rise} are significantly different from each other ($P < 0.01$) only for the largest and smallest pair of loads.

The magnitude of the load has less effect on the area of the antagonist burst (Q_{ant}). The intended speed, however, has a strong influence. Both antagonists show positive trends for Q_{ant} as load or speed increases, but for the two load tasks, only the Posterior Deltoid FE-Load trend is statistically significant. For the FF-Speed task, the trends are highly significant in both muscles (Table 1).

The shoulder extensor bursts are delayed when movement time increases, either because of added loads or intentionally reduced speeds. These same extensor EMG patterns are found for single-joint elbow flexion movements during similar tasks. There is greater variability between the tasks in the modulation of the elbow extensor. Triceps onset in Fig. 2A is at a constant latency following the agonist while burst area increases with load in this subject. Activation of the triceps in Fig. 2B is more like the single-joint pattern with a small, constant latency early component and a later component with a load-dependent latency, but there is little evident change in area with load. Across our eight subjects, we found no consistency in the latency of the extensors. Some subjects activated them shortly after the agonist and did not vary the onset time with load. Others increased the latency with load. Both patterns were seen at both elbow and shoulder. This was also true for the third task. This lack of consistency is a common observation in the movement antagonists (Almeida et al. 1995; Virji-Babul et al. 1994).

Because both Q_{ag} and impulse are affected by the tasks in the same way, we examined whether the two variables are also correlated with each other as they are for single-joint tasks. Figure 3 shows the correlation between Q_{ag} and

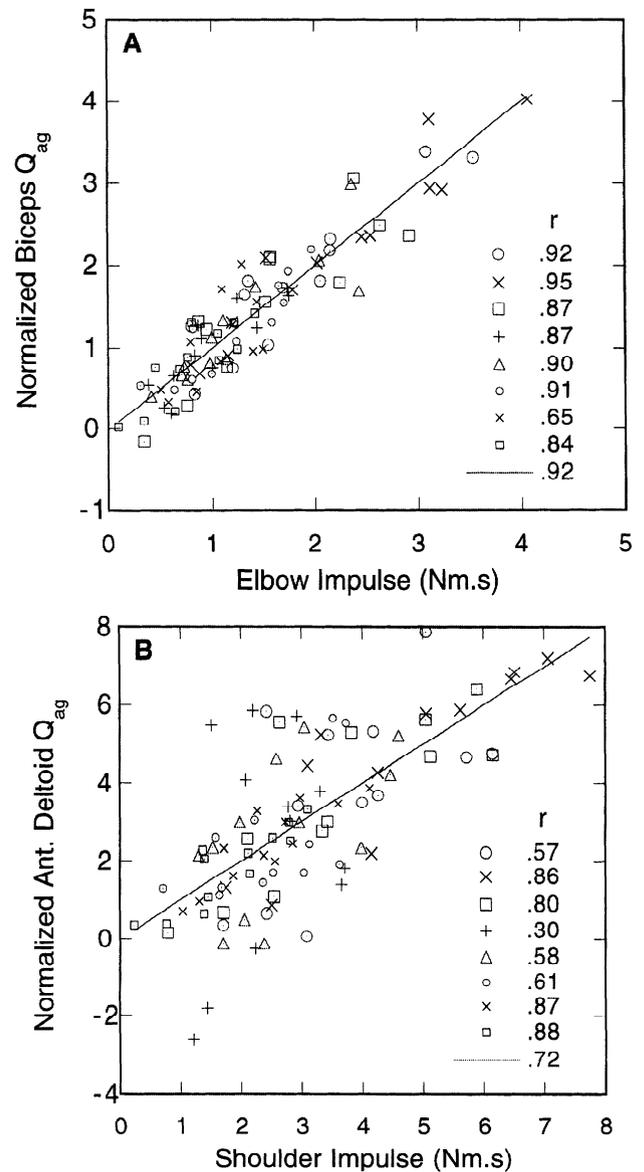


FIG. 3. Impulse produced at the elbow (A) and shoulder (B) joints is correlated with the area of joint's agonist muscle EMG burst. Data from each of the 8 subjects have been normalized to unity slope, and the correlation coefficients (r) are shown in the figures. The line has been drawn for the regression curve calculated from the pooled data.

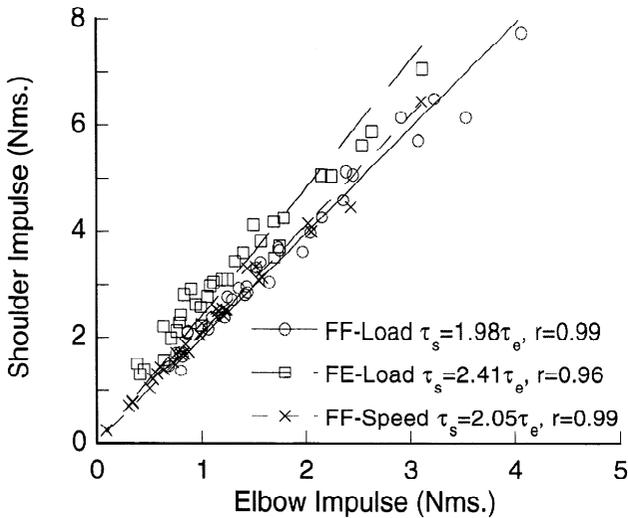


FIG. 4. Impulse produced at the shoulder remains proportional to the impulse at the elbow during movements with different inertial loads of intended movement speed. The lines are drawn from the regression equations shown in the figure.

impulse for all eight subjects. In this figure we have performed a linear transform on Q_{ag} to compute a normalized value (\hat{Q}) for each subject. It is defined by $\hat{Q} = (Q - b_i) / m_i$, where m_i is the slope and b_i is the y-intercept of the linear regression curve, computed for each subject (i). This places the data from all subjects on a line with unity slope running through the origin without affecting the within-subject variance. The column of figures in the graph is the value of the correlation coefficient for each subject, and the final value is for the pooled data.

Coordination between joints

All subjects demonstrated a very strong tendency to scale torques at the elbow and shoulder in parallel, increasing both with load or intended speed. Proximal joint torque is about twice that at the distal joint, and the peak values vary over about a fourfold range for the different subjects and tasks. Figure 4 plots the impulse at the shoulder versus that at the elbow for all eight subjects and all three tasks. The two FF series are indistinguishable, but there is a small separation of the FE movements. The shoulder is obligated to support the forearm and so the correlation between the two joints is in part due to their mechanical coupling (see Eq. 2). To determine how important this component is, we also computed the linear regressions between elbow impulse and the residual shoulder impulse after subtracting the elbow component. The slopes of those regression curves are equal to those of the original curves minus one. Their correlation coefficients fall slightly ($r = 0.93, 0.82, 0.97$) in comparison with the values shown in Fig. 4.

The quantitative correspondence of impulse between the two joints is accompanied by synchronization of the biphasic torque pulses. The three temporal landmarks on the torque waveforms, the peaks into flexion and extension, and the zero crossing between them are almost simultaneous at the two joints, and this relationship does not differ between subjects or tasks as illustrated by Fig. 5.

The interjoint correlations of the impulse and of the tem-

poral landmarks imply that the torque waveforms of the two joints should be highly correlated over time. To examine this, we superimposed the normalized torque waveforms of the two joints. Figure 6 shows a subject who demonstrated one of the strongest linear relationships between the torques at the two joints. The top row shows the two normalized torque waveforms for the four load/speed conditions. For each condition, elbow and shoulder torque are aligned on their initial values. The four different conditions are vertically offset for clarity. The middle row Lissajous figures show elbow torque plotted versus shoulder torque. The bottom row shows the path of the hand in the sagittal plane.

Figure 7 shows the subject who, for the FE-Load series,

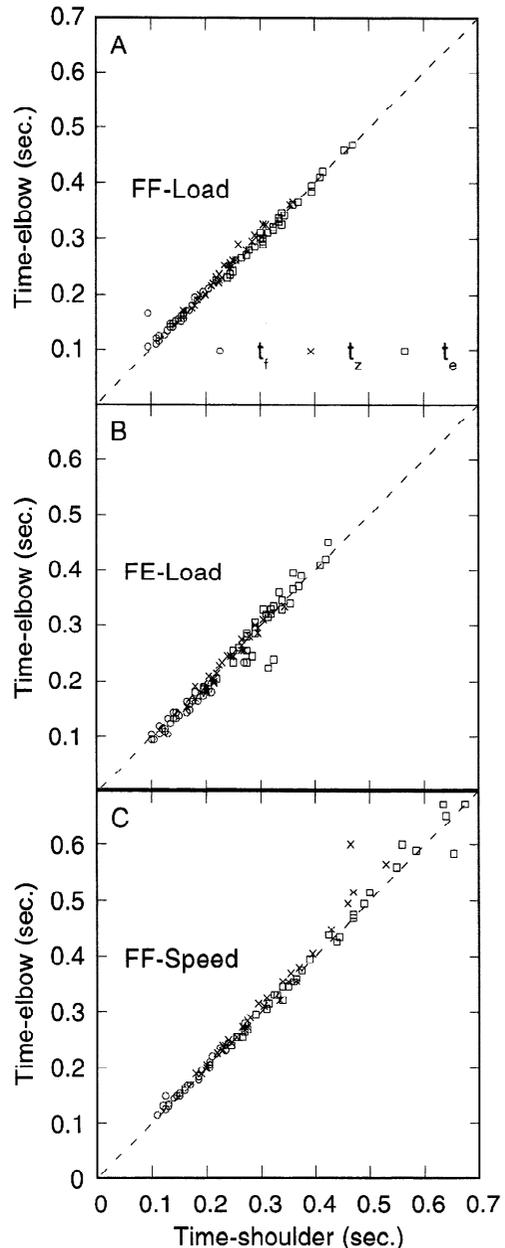


FIG. 5. Temporal landmarks of the shoulder and elbow torque waveforms occur almost simultaneously. Times to the 1st peak into flexion (\circ), to the peak into extension (\square), and to the zero crossing (\times) between them were measured after removing the gravity-dependent torque component. Dashed lines have unity slopes.

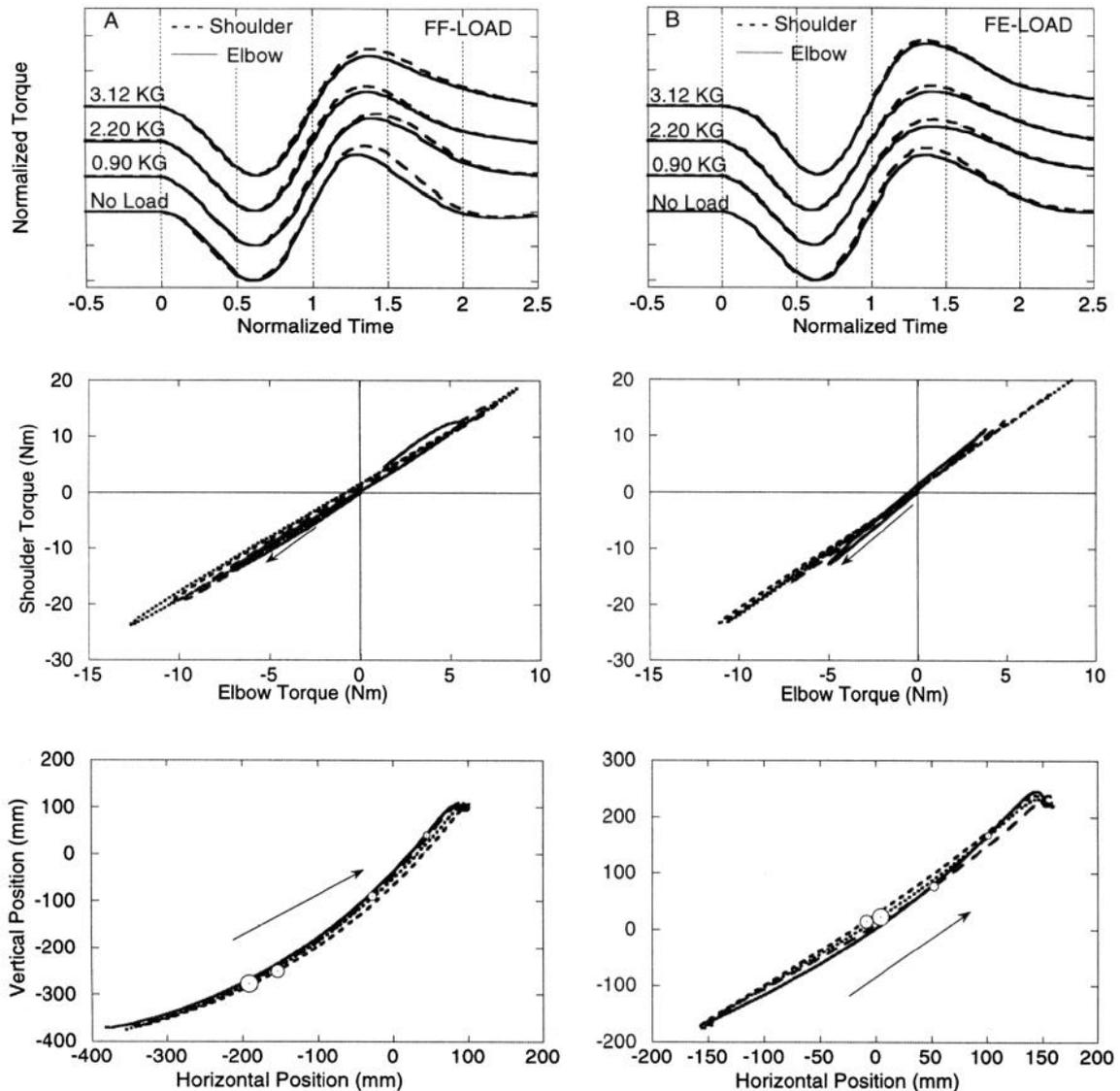


FIG. 6. Normalized elbow and shoulder torque (according to Eqs. 3 and 4) are plotted vs. time (top) and vs. each other (middle). A–C are for the 3 different tasks. At the bottom, the path of the hand is shown. This subject is one who showed a very linear relation between the torque waveforms at the 2 joints.

had the greatest deviation from elbow-shoulder torque linearity. His Lissajous figures are figure eights and the hand paths have a different curvature than the other two tasks. His FF-Load and Speed movements are the more typical narrow ellipses.

DISCUSSION

We recently proposed that single- and multijoint rules for movement are similar (Almeida et al. 1995) because the CNS retains rules of multijoint tasks when required to perform an unusual single-joint task. Multijoint pointing movements are natural elements of our movement repertoire, and, when asked to perform a novel and peculiar task, such as move a single degree of freedom manipulandum, their rules may be used because they are adequate and we have no motivation to change them. The results of the present study further demonstrate that the relationship between task vari-

ables, load and speed, and behavioral measures such as impulse and integrated EMG that we used to parameterize single-joint movements (Gottlieb et al. 1989b) applies to the individual joints of some multijoint movements. Torque increases with load or speed, and the area of the agonist burst is correlated with the impulse (Gottlieb et al. 1989a). For increases in load, the rates of rise of the torque and of the agonist EMG burst are usually load invariant, although this is imperfectly obeyed by the FE-Load movements. The variation of the rising phase of the agonist burst during FE-Load movements is, however, much smaller than it is during FF-Speed movements where the rate of muscle activation is proportional to the speed as it is during single-joint elbow movements (Corcos et al. 1989). The greatest differences from single-joint behavior are found in the antagonists. We found small but not statistically significant increases in the areas of the antagonist bursts and no consistent changes in latency with changes in load.

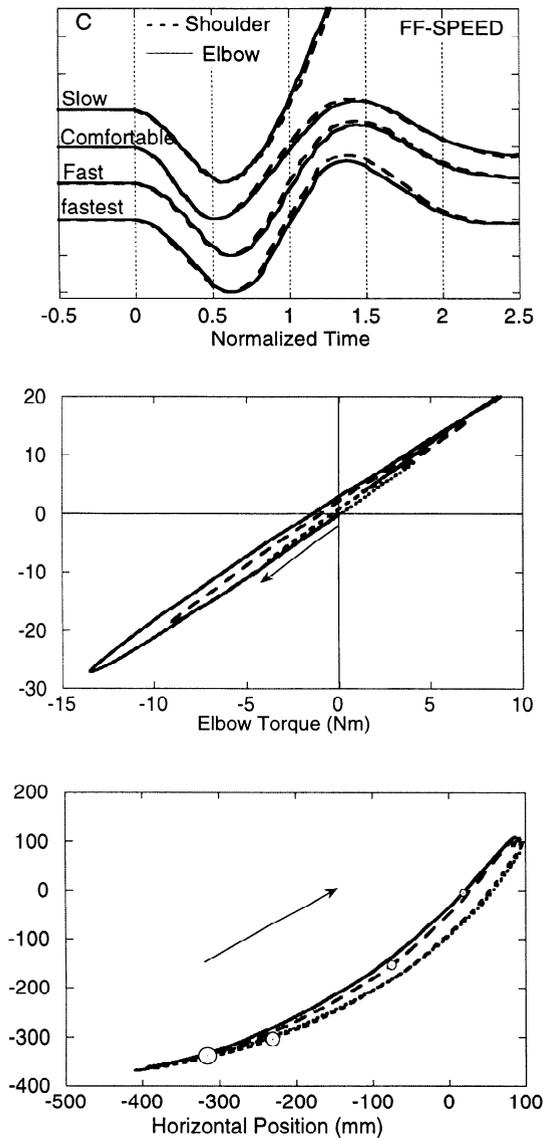


FIG. 6. (continued)

This weak relationship between movement task and antagonist activation is in contrast to the findings of single degree of freedom movements in which the antagonist burst scales with the level of inertial loading (Gottlieb et al. 1989a; Karst and Hasan 1987; Lestienne 1979; Sherwood et al. 1988). If the data for the two antagonistic muscles in the two loading tasks are considered, there is a small monotonic increase in Q_{ant} in all four instances. However, whatever degree of dependence exists, it is clearly less for multijoint than for single-joint movements. How can we account for these differences? It is possible that the nonlinear, viscous properties of the muscles play a role in this. Had we performed our analyses of variance on Q_{ant} without adjustments for gravity, Q_{ant} would have been larger, and the correlation between Q_{ant} and load would have been higher. Removing the static component from the EMG removes more than just a gravitational component from Q_{ant} . We must also remember that the assumption that static and dynamic components are independent and additive is one of convenience for which we have little empirical evidence. According to Karst and Hasan

(1987), the amount of antagonist activity does not depend exclusively on the torque needed to arrest the movement. Therefore our static correction may be excessive.

Even in the analysis of single-joint movement, interpretation of the antagonist has proven difficult. In multijoint movement, it is further complicated by gravity and the increased number of muscles involved in controlling the movement. We speculate that another cause may be that our measurements of EMG patterns were too narrowly focused on a single muscle from each joint. Even if a single-joint movement is regarded as just a simplified multijoint movement, the problems of stabilizing a multijoint limb in space has many more degrees of freedom and involves many muscles (Flanders 1991; Happee 1992) that act both in the plane of motion and out of it. Although we know that there was little out-of-plane motion by our subjects, it is possible that this was achieved by significant out-of-plane muscle torques that were balanced among synergists and antagonists to prevent such motion. Trying to associate the net muscle torque for a single axis of rotation at a joint (especially one like the shoulder, which has multiple degrees of freedom) with the activation patterns of a single muscle is probably too great a simplification to succeed in general. We conclude that the way torque is modulated at the individual joints during a free movement of the arm in the sagittal plane is only approximately like the way it is modulated during single-joint movement in the horizontal plane (Virji-Babul and Cooke 1995). Because movement is a consequence of the combined actions of many muscles, we cannot consistently correlate the task-dependent activity patterns of every individual muscle with the net muscle torque patterns, although such correlation can be found for some muscles (especially the agonists).

It appears that at the individual joints, the dynamic torque patterns are biphasic pulses that are scaled in amplitude and timing to perform the desired kinematic tasks. This is clear from single-joint experiments and is consistent both with the results here, our previous reports (Almeida et al. 1995; Hong et al. 1994), and as shown in Buneo et al. (1995), for many but not every direction around the work space. Single-joint rules cannot be a complete basis for the control of multijoint movements, because, in addition to rules for force production at each joint, we must also have rules for coordinating those forces among the joints. The evidence presented here and in Gottlieb et al. (1996) suggests that the coordinating rule may be

$$\text{torque}(t)_{\text{shoulder}} = K_d \text{torque}(t)_{\text{elbow}} \quad (5)$$

where these are the dynamic torque components and K_d is a constant that, for these movements, is approximately two. From this very strong rule, the relations between interjoint impulse and landmark times that are shown in Figs. 4 and 5 are a simple and necessary consequence.

Planar movements have distinct kinematic properties, usually being gently curved or occasionally straight and having almost symmetrical, bell-shaped velocity profiles. These patterns will not be produced by all interjoint coordination rules (Atkeson and Hollerbach 1985; Hollerbach and Atkeson 1987). Atkeson and Hollerbach (1985) showed analytically that scaling joint torque amplitudes can preserve the trajectory over changes in speed and load. From this it follows that if elbow and shoulder

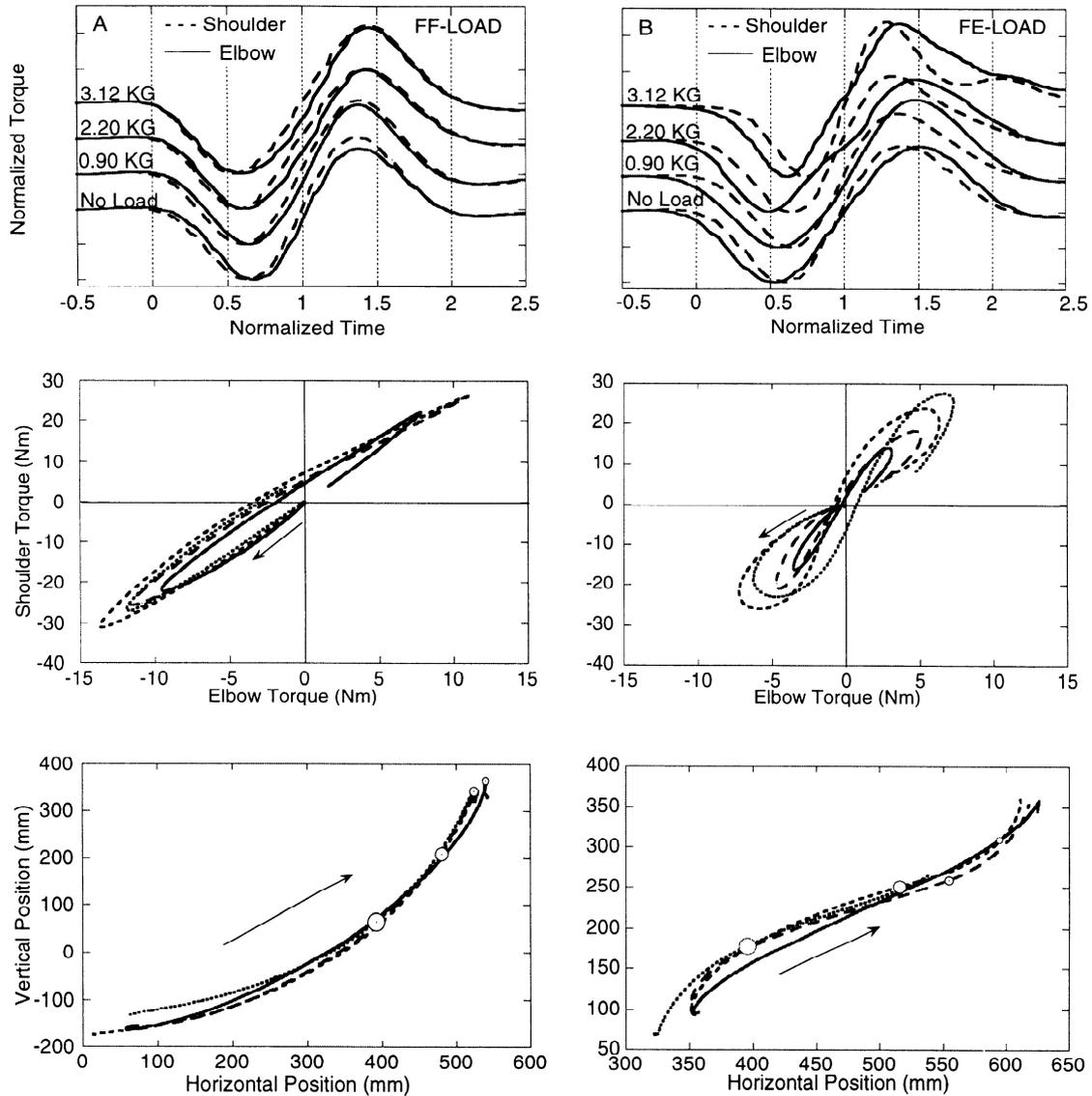


FIG. 7. Normalized elbow and shoulder torque (according to Eqs. 3 and 4) are plotted vs. time (top) and vs. each other (middle). A–C are for the 3 different tasks. At the bottom, the path of the hand is shown. This subject is one who showed the least linear relations between the torque waveforms at the 2 joints.

torque are linearly related for one load and speed, they can remain so for different loads and speeds to preserve the trajectory. We have demonstrated experimentally here that this appears to be the strategy used by the motor system.

If we start with the hypothesis that Eq. 5 is a default rule for interjoint coordination that is used by the CNS, we can consider how general it is and how and why it might be violated. The top rows of Figs. 6 and 7 show that overplotting the normalized torque of the two joints reveals very similar patterns across joints. The linear relationship is not perfect, however, as revealed by the Lissajous plots of elbow versus shoulder torque. The Lissajous patterns range from almost straight lines, to narrow ellipses, to figure eights. These torque patterns are produced, however, by only modest deviations from strict linearity.

If a biphasic torque pulse is produced at the elbow to accelerate and decelerate the limb, it can be approximated by a single cycle of a sine wave of period $T = 2$ (Fig. 8A,

solid line). If the shoulder's torque pulse were identical to the elbow's, a Lissajous plot of the two joint torques would be a straight line. If we distort the shoulder's torque and shift the occurrence of its zero crossing to lead the elbow's by 5% of the period T (e.g., 15 ms for a 300-ms movement time), we get the patterns shown in Fig. 8 by the dotted curves. The ellipsoidal shape looks much like those in Figs. 6 and 7. If instead we symmetrically shift the shoulder's peaks to first lag and then lead the elbow's peaks by 4% of the period (e.g., 12 ms for a 300-ms movement time) with the intervening zero crossing unchanged, we get the patterns shown by the dashed curves in Fig. 8. This resembles the FE pattern of Fig. 7. The two modeled torques in Fig. 8 have correlation coefficients of 0.99 and would qualify as "nearly linearly" related.

The data in this manuscript apply to movements with different inertial loads and at different intended speeds but to only two initial/final pairs of hand endpoints. We have

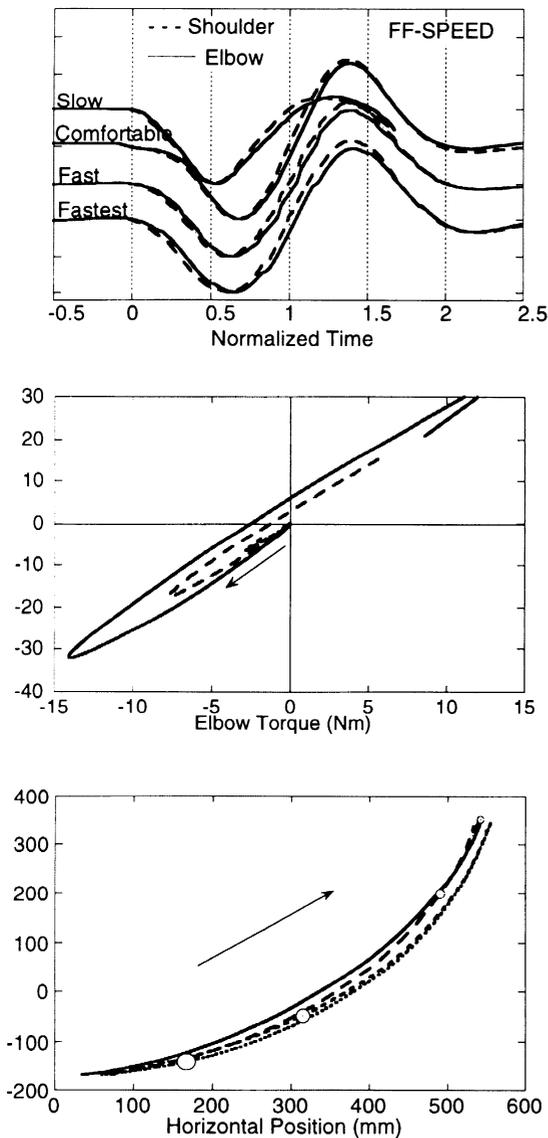


FIG. 7. (continued)

shown similar near linearity during movements in which only one joint is intentionally moved (elbow or shoulder) over different angular distances from 25 to 75° (Almeida et al. 1995). All of those movements and the movements that we have analyzed here were made with either an upward or upward and outward hand motion. They required initial flexion torques at both joints to either produce flexion motion or to prevent extension motion by the interaction torques from the other joint. We find that K_d is positive and approximately 2. Movements in which only the elbow or shoulder flex had different values of K_d (1.44 and 2.25) (Gottlieb et al. 1996). An overhand throw would have a positive K_d but with extension torques initially produced at both joints. A basketball set shot would have a negative K_d to produce shoulder flexion and elbow extension. In all of these cases, the joint torques would presumably take on biphasic patterns similar to those shown here and Eq. 5 might apply. Such joint torque patterns can be produced by biphasic EMG patterns. Note that K_d is identical for the two tasks that have identical endpoints (FF-Load, and FF-Speed) and slightly

different for the FE-Load task, which had a different initial position. Thus, according to this model, K_d should depend on the direction of movement and the location of the endpoint. In fact, the relative sizes of the joint torques may be the variables that specify movement direction.

If a subject performs a series of ‘center-out’ pointing tasks and the arm is directed to point to successive locations around the work space, we can reason that K_d must gradually change from positive to negative (unpublished data). For some directions of movement then, either the left or right hand side of Eq. 5 would vanish and the equation becomes ill defined or indeterminate when $K_d = 0$ or $1/K_d \rightarrow 0$. Buneo (Buneo et al. 1995) has shown that for some directions of planar arm movement, the torque pattern deviates from the almost sinusoidal, biphasic pulse shown here, especially during the deceleration phase. Flanders (Flanders et al. 1994) noted that EMG patterns around the work space are not uniformly bi/triphasic. This suggests that Eq. 5 may not be accurate for every point-to-point movement in the work space and similarly, that bi/triphasic muscle activation patterns may also be an incomplete rule. Movements where this rule may fail are likely to be those where K_d or its reciprocal is going through zero. These movements will be discussed in a subsequent paper (unpublished data).

Although we have concentrated exclusively on the production of muscle torque as the controlling principal for fast, phasic movements, we do not overlook the fact that such a

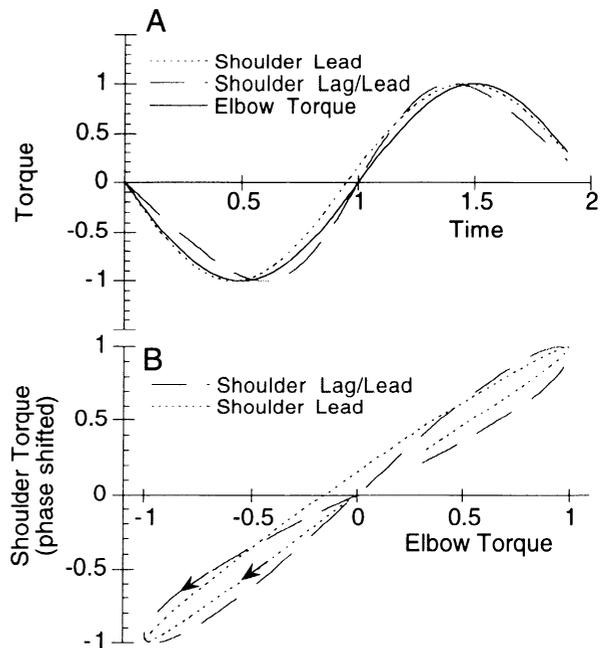


FIG. 8. A: hypothetical, sinusoidal torque pulse $\{ \sin [2\pi f(t)/T], 0 \leq t < 2 \}$ are plotted. The solid line represents the elbow torque and $f(t) = t$. Distorting the time axis for the shoulder torque using $f(t) = t + 0.1 \sin(\pi t/T)$ produces a biphasic torque pulse (dotted curve) with a unimodal phase shift that leads the elbow torque throughout the period (0-2) that is zero at the endpoints and ~ 0.1 at the midpoint $t = 1$. Distorting the time axis more rapidly using $f(t) = t + 0.1 \sin(2\pi t/T)$ produces a biphasic torque pulse (dashed line) with a bimodal phase shift that is zero at the endpoints and at the middle, leads the elbow during the 1st half of the period, and lags during the 2nd. B: plotting the Lissajous figure of the 2 temporally shifted torques on the abscissa shows that the unimodal shift produces an elliptical distortion of the curve while the bimodal shift produces a figure-eight distortion.

movement strategy is based on the existence of a stable posture from which movements can be launched and into which movements resolve. This stability is a consequence, not of force-based mechanisms but of the compliant mechanics of the muscles and their supporting reflexes. We take it as self evident that such postural stability exists, at least at the endpoints of every movement. The present work addresses how we go between endpoints. Our data suggest that it may be straight forward to describe the control of sagittal plane movement between stable endpoints in terms of planned muscle torques and the activation patterns that will produce them. Such an approach provides a unified explanation for the control of movement speed and load and the control of distance and direction (unpublished data) that is simpler and more complete than approaches that rely primarily on the system's equilibrium properties to provide propulsion (cf. Gomi and Kawato 1996). Direct planning of the torques at each joint and of the coordination among joints (at least for 2) by central commands requires only a small number of parameters that characterize the excitation pulses that produce the biphasic muscle activation patterns. The linear relation between joint torques is an example of a basic coordination rule that masters "redundant degrees of freedom" (Bernstein 1967). Small deviations from this rule by changes in the relative phasing of the torques can change the hand's path (Gottlieb et al. 1996). These rules apply with appropriate parametric specification whether a joint flexes, extends, or is not supposed to move at all. A more complete description of the control of distance and direction will require further study.

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REFERENCES

- ALMEIDA, G. L., HONG, D. H., CORCOS, D. M., AND GOTTLIEB, G. L. Organizing principles for voluntary movement: extending single joint rules. *J. Neurophysiol.* 74: 1374–1381, 1995.
- ANGEL, R. W. Electromyography during voluntary movement: the two burst pattern. *Electroencephalogr. Clin. Neurophysiol.* 36: 493–498, 1974.
- ATKESON, C. G. AND HOLLERBACH, J. M. Kinematic features of unrestrained vertical arm movements. *J. Neurosci.* 5: 2318–2330, 1985.
- BERNSTEIN, N. A. *The Coordination and Regulation of Movement*. Oxford, UK: Pergamon, 1967.
- BUNEO, C. A., BOLINE, J., SOECHTING, J. F., AND POPPEL, R. E. On the form of the internal model for reaching. *Exp. Brain Res.* 104: 467–479, 1995.
- BUNEO, C. A., SOECHTING, J. F., AND FLANDERS, M. Muscle activation patterns for reaching: the representation of distance and time. *J. Neurophysiol.* 71: 1546–1558, 1994.
- CORCOS, D. M., GOTTLIEB, G. L., AND AGARWAL, G. C. Organizing principles for single joint movements. II. A speed-sensitive strategy. *J. Neurophysiol.* 62: 358–368, 1989.
- FLANDERS, M. Temporal patterns of muscle activation for arm movements in three-dimensional space. *J. Neurosci.* 11: 2680–2693, 1991.
- FLANDERS, M., PELLEGRINI, J. J., AND SOECHTING, J. F. Spatial/temporal characteristics of a motor pattern for reaching. *J. Neurophysiol.* 71: 811–813, 1994.
- GOMI, H. AND KAWATO, M. Equilibrium-point control hypothesis examined by measured arm stiffness during multijoint movement. *Science Wash. DC* 272: 117–120, 1996.
- GOTTLIEB, G. L. Voluntary movement of compliant (inertial-viscoelastic) loads by parcellated control mechanisms. *J. Neurophysiol.* 76: 3207–3229, 1996.
- GOTTLIEB, G. L. A computational model of the simplest motor program. *J. Mot. Behav.* 25: 153–161, 1993.
- GOTTLIEB, G. L. AND AGARWAL, G. C. Filtering of electromyographic signals. *Am. J. Physical Med.* 49: 142–146, 1970.
- GOTTLIEB, G. L., CHEN, C.-H., AND CORCOS, D. M. An "adequate" control theory governing single-joint elbow flexion in humans. *Ann. Biomed. Eng.* 23: 388–398, 1995a.
- GOTTLIEB, G. L., CHEN, C.-H., AND CORCOS, D. M. Relations between joint torque, motion and EMG patterns at the human elbow. *Exp. Brain Res.* 103: 164–167, 1995b.
- GOTTLIEB, G. L., CORCOS, D. M., AND AGARWAL, G. C. Organizing principles for single joint movements. I. A speed-insensitive strategy. *J. Neurophysiol.* 62: 342–357, 1989a.
- GOTTLIEB, G. L., CORCOS, D. M., AND AGARWAL, G. C. Strategies for the control of single mechanical degree of freedom voluntary movements. *Behav. Brain Sci.* 12: 189–210, 1989b.
- GOTTLIEB, G. L., SONG, Q., HONG, D., ALMEIDA, G. L., AND CORCOS, D. M. Coordinating movement at two joints: a principal of linear covariance. *J. Neurophysiol.* 75: 1760–1764, 1996.
- HALLETT, M., SHAHANI, B. T., AND YOUNG, R. R. EMG analysis of stereotyped voluntary movements in man. *J. Neurol. Neurosurg. Psychiatry* 38: 1154–1162, 1975.
- HAPPEE, R. Time optimality in the control of human movements. *Biol. Cybern.* 66: 357–366, 1992.
- HOFFMAN, D. S. AND STRICK, P. L. Force requirements and patterns of muscle activity. *Behav. Brain Sci.* 12: 221–224, 1989.
- HOFFMAN, D. S. AND STRICK, P. L. Step-tracking movements of the wrist. III. Influence of changes in load on patterns of muscle activity. *J. Neurosci.* 13: 5212–5227, 1993.
- HOLLERBACH, J. M. AND ATKESON, C. G. Deducing planning variables from experimental arm trajectories: pitfalls and possibilities. *Biol. Cybern.* 56: 279–292, 1987.
- HONG, D., CORCOS, D. M., AND GOTTLIEB, G. L. Task dependent patterns of muscle activation at the shoulder and elbow for unconstrained arm movements. *J. Neurophysiol.* 71: 1261–1265, 1994.
- KARST, G. M. AND HASAN, Z. Antagonist muscle activity during human forearm movements under varying kinematic and loading conditions. *Exp. Brain Res.* 67: 391–401, 1987.
- LESTIENNE, F. Effects of inertial load and velocity on the braking process of voluntary limb movements. *Exp. Brain Res.* 35: 407–418, 1979.
- SHERWOOD, D. E., SCHMIDT, R. A., AND WALTER, C. B. Rapid movements with reversals in direction. II. Control of movement amplitude and inertial load. *Exp. Brain Res.* 69: 355–367, 1988.
- VIRJI-BABUL, N. AND COOKE, J. D. Influence of joint interactional effects on the coordination of planar two-joint arm movements. *Exp. Brain Res.* 103: 451–459, 1995.
- VIRJI-BABUL, N., COOKE, J. D., AND BROWN, S. H. Effects of gravitational forces on single joint arm movements in humans. *Exp. Brain Res.* 99: 338–346, 1994.
- WINTER, D. A. *Biomechanics of Human Movement*. New York: Wiley, 1979.