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Scaling of joint torque during planar arm movements

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Abstract The present study scrutinized the “Motor Program” concept for aimed arm movements. Human subjects pointed at visual targets in a horizontal plane, with movements of varying starting positions, amplitudes and directions. We recorded movement kinematics and subsequently calculated the shoulder and elbow joint torque profiles. Our results indicate that the *shape* of torque profiles is rather uniform across movements and joints. We defined the *size* of those profiles by six “landmark variables”, which could be subsequently reduced to three factors using factor analysis: one factor represented torque magnitude and two represented different aspects of torque timing. Additional analyses indicated that total torque duration is an important controlled signal. Our findings conform with the view that movements are executed by playing back scaled versions of prototypical joint torque profiles.

Key words Pointing · Arm movements
Motor programs · Rescaling · Torque · Human

Introduction

It has been proposed in the past that aimed arm movements are executed by playing back prestructured force-time functions, or *motor programs* (Keele 1968; Schmidt et al. 1979). According to this view, movements with different trajectories are produced by scaling the stored functions differently in magnitude and duration; thus, the potentially complex task of controlling a multijoint limb (e.g. Hollerbach and Flash 1982) would be reduced to the calculation of only two scaling parameters.

Experimental support for the motor program concept has been derived from studies of *hand* trajectories during aimed arm movements. A number of invariant

characteristics were found, as expected if the movements were brought about by stereotyped force-time functions. Thus, the shape of hand velocity profiles was invariant across changes of movement speed or external load (Hollerbach and Flash 1982; Atkeson and Hollerbach 1985), and the magnitude and duration of those profiles scaled consistently with load size (Bock 1990a). Furthermore, the well-known speed-accuracy trade-off was attributed to a proportional increase in the variability of scaling parameters with their intended magnitude (Schmidt et al. 1979; Meyer et al. 1982). Other experiments have indicated that motor programs can be modified if required: hand trajectories vary with task constraints (Marteniuk et al. 1987) and can be corrected in mid-flight if there is a change in target location (Georgopoulos et al. 1981; Gielen et al. 1984; Goodale et al. 1986) or in external force (Wadman et al. 1979; Lee et al. 1986; Johansson and Westling 1988).

Unfortunately, the evidence provided by the above studies is only indirect, due to the complex relationship between recorded hand trajectories and underlying driving forces (e.g. Hollerbach and Flash 1982). It is therefore conceivable that the observed invariances are better related to mechanical factors such as limb inertia and dynamic joint interactions rather than to the characteristics of motor commands. It appears that studies of the electromyographic (EMG) activity during arm movements provide a more direct insight into neural control strategies.

In agreement with the motor program concept, EMG studies documented certain regularities in the pattern of muscle activation during movements. Thus, fast arm movements exhibit a consistent tri-phasic EMG pattern (Hallett et al. 1975), which is preserved in deafferented subjects and is therefore of central origin (Forget and Lamarre 1983; Sanes and Jennings 1984). Furthermore, EMG magnitude scales consistently with movement speed (Karst and Hasan 1987; Flanders and Herrmann 1992; see, however, Brown and Cooke 1990), the duration of the first EMG burst increases with movement amplitude (Wadman et al. 1979; Benecke et

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al. 1985), and the distribution of EMG onset times and magnitudes across muscles depends in an ordered way on movement direction (Hoffman and Strick 1986; Flanders 1991; Karst and Hasan 1991).

In spite of the above regularities, however, EMG studies don't provide conclusive support for the existence of motor programs. As stated by some authors (Wadman et al. 1979; Darling and Cooke 1987; Karst and Hasan 1987), EMG signals exhibit a substantial trial-to-trial variability in the *shape* of individual bursts, their relative *timing*, and even their *number* per movement. Accordingly, most studies analysed means rather than individual recordings, or global measures such as the integral over a burst. Clearly, the reported variability of EMG signals is hard to reconcile with the view that they represent stereotyped control signals.

The non-uniformity of EMG signals could indicate that the motor program concept is incorrect. Alternatively, prototypical force-time functions could be implemented in the motor system at a higher hierarchical level than that of individual muscle activations; EMG variability could then originate from lower-level sources. The latter view, that invariant high-level representations of movement may correspond to variable low-level signals, was first proposed by Bernstein (1967, e.g. page 49).

From the above considerations, it seems desirable to investigate signals which represent less peripheral events than the EMG, while at the same time being more closely related to neural commands than hand trajectories. In the present work we decided to study *joint torque*, since it reflects the combined activities of all muscles acting about the joint, and is not contaminated by the effects of limb mechanics. In support of the motor program concept, our results indicate that torque profiles have an invariant shape and, therefore, that shape invariance is not an artefact of limb dynamics. However, our data suggest that the original concept needs to be modified: torque profiles are more adequately described by three, in contrast to the original two, scaling factors.

Materials and methods

Two right-handed men (aged 30 and 33 years) executed sequences of arm movements in a horizontal plane. Both subjects were healthy and had previous experience in motor control studies. Figure 1 illustrates the experimental setup. Subjects sat, with their back supported, in front of a horizontal panel. Their right arm was elevated to shoulder level, thus moving just above the panel. Nine numbered targets of 1.5 cm diameter, interconnected by straight lines, were drawn within a 40 × 30 cm (lateral × sagittal) panel area.

The subjects were instructed to point from target to target at a comfortable speed, without touching the panel throughout the movement sequence. We emphasized that accuracy was of little concern in the present study, and that subjects should refrain from correcting their movements. Each sequence started with the index fingertip just above target 1 (square in Fig. 1). After a verbal command, subjects pointed at target 2 and kept the finger in its (accurate or inaccurate) final position above the panel. After the next command, they pointed at target 3, etc., until a sequence of eight movements was completed. Figure 1 shows that the movements

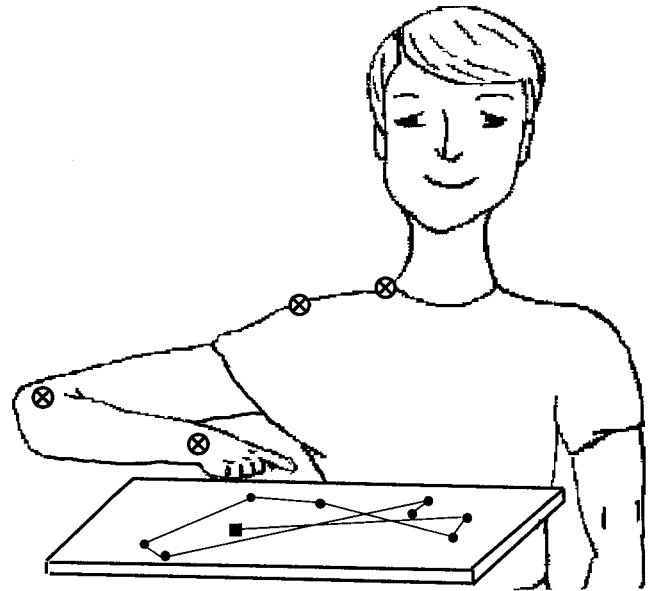


Fig. 1 Scheme of the experimental setup, showing a subject with infrared light-emitting diodes (*crossed circles*) and the target panel. For better clarity, the numbering of targets is omitted; instead, target 1 is plotted as a *filled square* and subsequent targets as *filled circles*. Note the wide range of prescribed movement amplitudes, directions, and starting points

varied considerably in their directions, amplitudes and locations in the workspace. Each subject repeated the movement sequence six times, thus executing a total of 48 movements.

Arm position was recorded contact-free in three dimensions (3-D) by the Watsmart motion analysis system, using two infrared-light-sensitive cameras and infrared light-emitting diodes (LEDs) on the subjects' wrist, elbow, shoulder joint and neck; the spatial accuracy of the Watsmart is about 1.5 mm. For each movement, data were recorded over an interval of 2 s at a frame rate of 100 Hz, and stored to disk for later analysis.

From the recordings, we calculated elbow and shoulder joint angles θ_e and θ_s ,¹ lowpass-filtered the results at 4.2 Hz, and calculated elbow and shoulder joint torque τ_e and τ_s using the two-link dynamics equations:

$$\tau_s = \left(m_1 \frac{l_1^2}{3} + m_2 \frac{l_2^2}{3} + m_2 l_1 l_2 \cos \theta_e + m_2 l_1^2 \right) \ddot{\theta}_s + \left(m_2 \frac{l_2^2}{3} + m_2 l_1 \frac{l_2}{2} \cos \theta_e \right) \ddot{\theta}_e - m_2 l_1 \frac{l_2}{2} (2\dot{\theta}_s + \dot{\theta}_e) \dot{\theta}_e \sin \theta_e \quad (1)$$

$$\tau_e = \left(m_1 \frac{l_2^2}{3} + m_2 l_1 \frac{l_2}{3} \cos \theta_e \right) \ddot{\theta}_s + m_2 \frac{l_2}{2} \ddot{\theta}_e + m_2 l_1 \frac{l_2}{2} \dot{\theta}_s^2 \sin \theta_e \quad (2)$$

where $m_1 = 1800$ g and $l_1 = 26$ cm are mass and length of the upper arm, while $m_2 = 1450$ g and $l_2 = 31$ cm are mass and length of the forearm. These values are means, yielded from six subjects in a previous study (Bock 1990a). The total number of torque profiles thus yielded was 48 (movements) × 2 (subjects) × 2 (joints) = 192.

Results

Figure 2 shows representative elbow and shoulder joint torque profiles from four movements of one subject. The movements differed widely in amplitude, direction and location in the workspace (see Fig. 1) and therefore re-

¹As conventional, positive values for angles or torques denote the counterclockwise direction

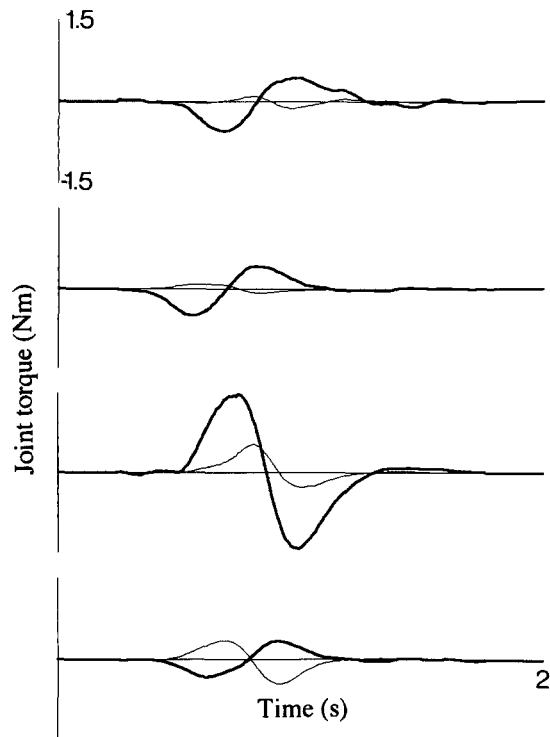


Fig. 2 Shoulder (*bold traces*) and elbow (*thin traces*) joint torque profiles of four aimed movements from one subject; *upward deflections* represent counterclockwise torque. The spatial characteristics of the movements were (*from top to bottom*): amplitude, 5.5, 7.5, 24 and 21 cm; workspace location, distal right, proximal left, centre and centre left; direction (as on a clock facing upwards), 4, 1, 9 and 11 h

quired substantially different torque. Indeed, the recorded torque profiles vary in their polarity, magnitude and timing; however, all profiles exhibit a similar, biphasic *shape*. It is not difficult to envisage that all profiles in Fig. 2 are scaled versions of the same prototype.

The shapes of all recorded torque profiles were comparable with those in Fig. 2. In some cases, a superposed oscillation of about 5 Hz was also observed (see top trace in Fig. 2). We have analysed such oscillations in a previous study and concluded that they represent contaminations of the motor control signal by intrinsic neural oscillators (Bock 1990b). In the present study, we therefore decided not to include the oscillations in further analyses and rather to evaluate only the biphasic component. In 42 recordings, however, oscillations encumbered the accurate assessment of the biphasic component. These recordings were therefore discarded and further analysis was limited to the remaining 150 torque profiles.

In order to identify their common shape, we normalized all 150 profiles with respect to peak amplitude and total duration,² and then calculated the *mean* normal-

²Amplitude was normalized with respect to the *larger* of the two lobes. If the peak amplitude of the first lobe was larger, it was set to +1; if the peak amplitude of the second lobe was larger, it was set to -1. In consequence, all normalized profiles had a positive first lobe, and their peak amplitudes averaged less than 1.00

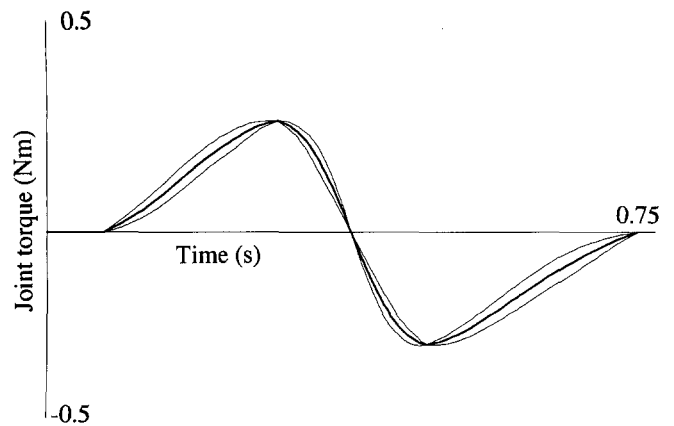


Fig. 3 Mean normalized joint torque profile, based on all 150 analysed recordings. *Error bars* indicate the standard deviations about the mean for the three “landmarks”: peak of first lobe, zero-crossover and peak of second lobe

ized profile as shown in Fig. 3. Furthermore, we documented the *variability* of individual shapes about the mean as standard deviations (SDs) of three profile “landmarks”: peak of the first lobe, zero-crossover between lobes and peak of the second lobe. As shown in Fig. 3, SDs were smaller than the respective means. For example, the normalized time of the first peak was 0.317 ± 0.053 (mean \pm SD), which implies that, for 68% of profiles, the first peak occurred in the relatively narrow time interval (0.264...0.370). We therefore conclude that the mean curve in Fig. 3 is a fair representation of individual profile shapes.

Our finding, that torque profiles vary in size but not in shape, supports the view that they are scaled versions of the same prototype. In analogy to the motor program concept, such scaling could be based on the two scaling factors “duration” and “peak amplitude” (see Schmidt et al. 1979; Meyer et al. 1982); however, other scaling factors could be envisaged as well. For example, one factor could determine both the peak amplitude and the time-to-peak for the first lobe, another factor the corresponding values for the second lobe, and a third factor could specify the remaining aspects of profile timing. We took the following approach to find out which factors were involved in torque scaling for the movements described above.

In a first step, we quantified all 150 non-normalized profiles by calculating the values of the six landmark variables:

1. T_1 , time from movement onset to peak of first lobe
2. T_2 , time from peak of first lobe to zero-crossover
3. T_3 , time from zero-crossover to peak of second lobe
4. T_4 : time from peak of second lobe to movement end
5. K_1 , absolute amplitude of first peak
6. K_2 , absolute amplitude of second peak

We deliberately selected a large number of landmark variables, to ensure that the actual scaling factors could be expressed as their combinations. In a second step, we

Table 1 Summary of factor analysis results, based on all 150 analysed profiles. T_1 , T_2 , T_3 , T_4 , K_1 and K_2 represent the “landmark variables” as defined in the Results section. $F1$, $F2$ and $F3$ represent the three factors extracted by factor analysis. The left part of the table indicates factor loadings (analogous to the correlations

between factors and variables), the middle part lists factor weights (analogous to the first-order regression coefficients between factors and variables), and the righthand column provides the final communality estimate (FCE ; describes how well a variable is explained by the factor solution)

Variable	Factor loading			Factor weight			FCE
	F1	F2	F3	F1	F2	F3	
T_1	0.083	0.076	0.943	-0.207	-0.139	-0.884	0.902
T_2	0.916	0.084	0.094	0.544	-0.009	-0.172	0.854
T_3	0.891	-0.002	0.149	0.517	-0.067	-0.096	0.814
T_4	0.477	0.288	0.572	0.131	0.035	0.375	0.639
K_1	0.056	0.965	0.143	-0.044	0.518	-0.058	0.955
K_2	0.063	0.957	0.091	-0.025	0.523	-0.109	0.929

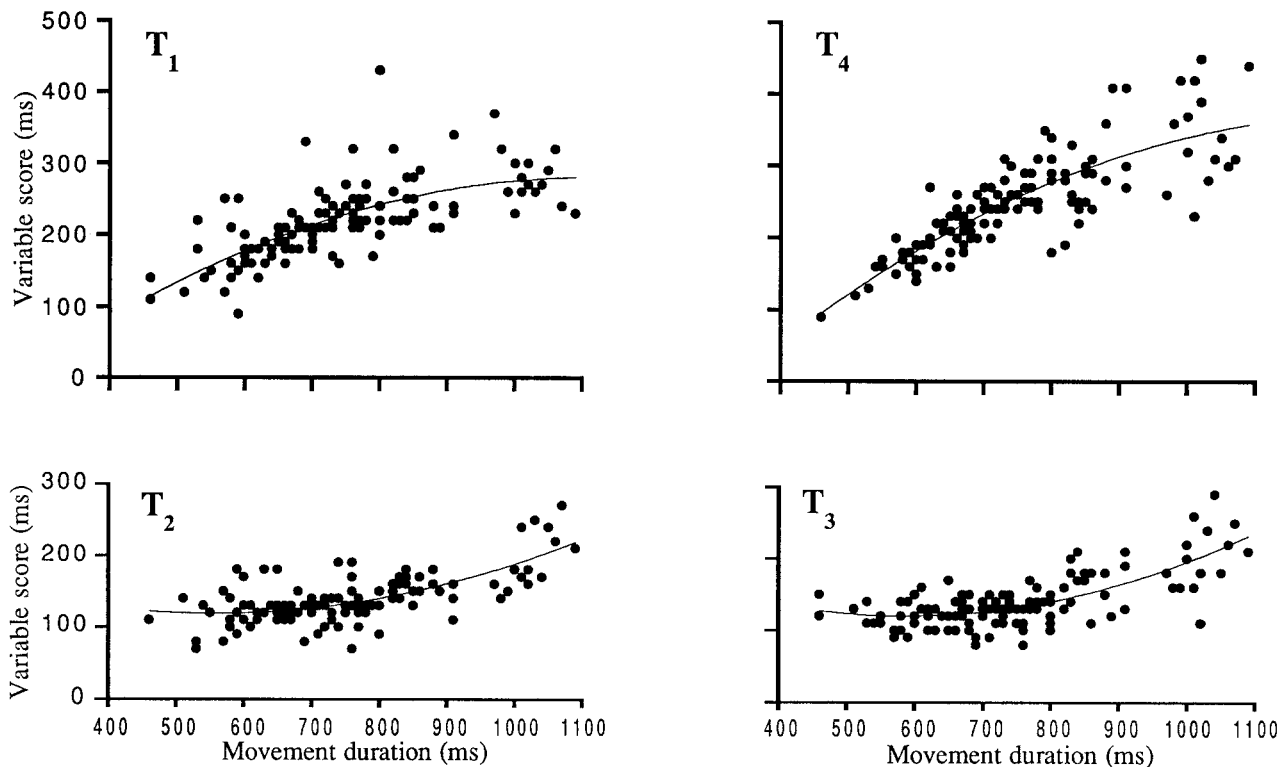


Fig. 4 Dependence of time-related variables T_1 , T_2 , T_3 and T_4 on total movement duration for all 150 analysed profiles. The curves represent second-order polynomial regression fits

then reduced the landmark variables to a smaller number of scaling factors using factor analysis.

Factor extraction was based on the “principal component” method and the 75% rule;³ we will report only the orthogonal solution, since oblique transformations yielded no notable improvements.⁴ As summarized in Table 1, the six landmark variables were reduced to three factors ($F1$, $F2$, $F3$), which explained 37%, 31%

and 16% of total variance, respectively. Further from Table 1, factor $F1$ was tightly coupled (near-unity loading) with landmark variables T_2 and T_3 , and loosely with T_4 . A change in $F1$ led to comparable changes (similar weights) in T_2 and T_3 , but only to minimal changes in T_4 . Factor $F2$ was very tightly related to landmark variables K_1 and K_2 . When $F2$ changed, both variables changed by similar amounts. Factor $F3$ was tightly coupled with T_1 and loosely with T_4 . A change in $F3$ modified T_1 more than T_4 .

Table 1 also indicates that the three-factor solution explains virtually all the variance in the variables T_1 , K_1 and K_2 (final communality estimate near unity), most of the variance in T_2 and T_3 , and less variance in T_4 .

The results in Table 1 indicate that: the amplitude of both peaks is controlled jointly, since K_1 and K_2 are associated with a single factor; peak amplitude is con-

³The search for further factors was truncated when 75% of total variance was explained

⁴In orthogonal solutions, all extracted factors are independent of each other

Table 2 Regression coefficients between movement duration and time-related variables, calculated separately for durations below 800 ms (left column) and above 800 ms (middle column); results are based on all 150 analysed profiles. Also shown are the Pearson correlations between shoulder and elbow profiles for each variable (right column)

Variable	Slope		<i>r</i> between joints
	short duration	long duration	
T_1	0.335**	0.090 ^{n.s.}	0.518**
T_2	0.063 ^{n.s.}	0.271**	0.521**
T_3	0.037 ^{n.s.}	0.282**	0.653**
T_4	0.584**	0.377**	0.531**

The results were tested against zero by *t*-tests for slopes or correlations (both tests are numerically equivalent): ** $P < 0.001$, and ^{n.s.} $P > 0.05$

trolled independently of timing, since amplitude- and time-related variables are associated with different factors; total duration is not controlled as a uniform whole, since time-related variables are associated with more than one factor; and T_4 is not associated with a single factor.

The seemingly intricate grouping of time-related variables can be readily interpreted when the relationships between those variables and total duration (i.e. $T_1 + T_2 + T_3 + T_4$) are considered. Figure 4 illustrates that each variable depends on total duration in a non-linear way, which is well fitted by second-order polynomials (curves in Fig. 4). In support of this observation, we found that polynomial regression analyses yielded statistically significant *second-order* regression coefficients (*t*-tests for the significance of second-order coefficients, $P < 0.01$ for each variable). Further from Fig. 4, it appears that T_2 and T_3 change little with duration up to about 800 ms but consistently increase thereafter, while the opposite trend is seen for T_1 . In the case of T_4 , an increase is seen throughout the duration-range tested, with increased variability and reduced slope above 800 ms. We confirmed these observations by applying linear regression analyses separately for duration below and above 800 ms. As summarized in Table 2, short durations yielded non-zero slopes for T_1 and T_4 , and long durations for T_2 , T_3 and T_4 . It is therefore not surprising that factor analysis grouped T_4 with T_1 (similarity for short durations) as well as with T_2 and T_3 (similarity for long durations).

As a final step of data analysis, we compared the timing of shoulder and elbow torque profiles by pairing, movement by movement, the elbow and shoulder scores of each time-related variable. As summarized in Table 2, the Pearson correlations between all pairs were highly significant. A significant correlation of 0.787 ($P < 0.001$) was also found between the *total durations* of shoulder and elbow torque; the latter correlation is significantly higher than those for T_1 , T_2 and T_4 in Table 2, and higher without statistical significance than the correlation for T_3 (Fisher's *z*-test of two correlations: $P < 0.05$ for T_1 , T_2 and T_4 ; $P > 0.05$ for T_3). This outcome indicates that

total duration is better coordinated between joints than individual time-related variables.

Discussion

The present experiments evaluated the motor program concept, which states that aimed arm movements are controlled by prototypical force-time functions, scaled in magnitude and duration to execute different movements (Keele 1968; Schmidt et al. 1979). As detailed in the Introduction, previous studies derived their support for this concept from the observation of invariant features in the trajectories of arm movements. However, such invariances don't necessarily reflect central commands; they could also ensue from the constraints imposed on movements by limb mechanics. This critical view is seemingly supported by the finding that EMG recordings show *little* invariance. As an alternative interpretation, we have argued that EMG signals are variable because they are too "peripheral", while motor programs are implemented at a hierarchically higher level in the nervous system.⁵

To substantiate our assumption, and thus to re-confirm the plausibility of the motor program concept, the present experiments analysed the invariant properties of *joint torque*, a signal which doesn't reflect limb mechanics⁶ and which is probably higher-level than EMG (it reflects the composite activities of *all* muscles acting about a given joint).

In accordance with our assumption, the present study documented that joint torque profiles have a consistent, biphasic *shape*. Similar shapes can also be observed in published original recordings by others (e.g. Soechting and Lacquaniti 1981; Lacquaniti et al. 1982; Brooks 1983). We further found that the variable *size* of torque profiles can be described by three scaling factors. One of those factors represents overall torque magnitude: it is tightly coupled with the peak amplitudes of the first *and* second torque lobe, but is not associated with the timing of torque profiles. The other two factors represent complementary aspects of profile timing independent of peak amplitude.

Our finding of consistent profile shapes and a small number of scaling factors generally supports the motor program concept. However, two qualifications are in place. Firstly, the "force-time function" specified by central commands appears more closely related to joint

⁵In addition, EMG variability could reflect technical limitations, e.g. a variable relationship between recorded signal and actual muscle contraction

⁶Note that torque was calculated from the recordings of hand trajectories. It is therefore conceivable that components of the neural command which don't manifest themselves in the trajectories are missing in the reconstructed torque profiles (e.g. if limb mechanics act as a lowpass filter, high-frequency components of the central signal could be missing). However, the absence of components which have *no relevance for movement trajectories* should have little impact on our conclusions about the existence of invariances in motor control signals

torque than to individual muscle forces. Secondly, while "overall magnitude" is controlled by a scaling factor, "overall duration" seems to be controlled by *two separate* factors.

The latter finding can be reconciled with the original motor program concept when our analyses of time-related landmark variables are considered. We found that those variables depend in a complex but consistent way on total duration, and that the durations of shoulder and elbow torque are more closely correlated than the corresponding landmark variables. Both findings underline the important role of duration as a controlled variable, and suggest that torque profiles could be scaled by a two-stage process: a first stage could specify torque magnitude and duration, and a second stage could transform the duration command into detailed instructions about torque timing according to the pattern in Fig. 4.

One line of support for the motor program concept came previously from findings that the timing of kinematic landmarks changes in proportion to total movement duration (e.g. Armstrong 1970; Hollerbach and Flash 1982; Carter and Shapiro 1984; Schmidt et al. 1988), as expected if total duration is a scaling parameter. Conversely, violations of such time rescalability were considered as a serious challenge to the motor program concept: thus, two studies found that the first acceleration peak of hand movement occurred at a fixed time, irrespective of total duration (Gielen et al. 1985; Zelaznik et al. 1986). The proposed two-stage scheme would allow us to reconcile the latter studies with the motor program concept; in fact, constant timing of the first acceleration peak closely parallels our finding that T_1 remained constant for a range of movement durations.⁷

No attempt was made in the present work to search for simple relationships between task variables, such as amplitude and direction of hand movement, and scaling parameters. In fact, the relationships are likely to be complex, given the complexity of limb dynamics (see Eqs. 1, 2) and kinematics. Instead, a simulation study was carried out (Bock et al. 1993): a two-link planar robot was controlled by mapping task variables into torque scaling factors with an artificial neural network, and applying the scaled torque reference signals to the robot's motors. After training of the neural network, we recorded a mean areal error of only 0.8% in a 100 cm × 50 cm workspace, which confirms that movement control by prototypical, scalable torque profiles is a viable concept.

⁷The cited studies investigated a motor task that was considerably different from the present work (well-practised movements with a prescribed total duration); it is therefore not surprising that time-invariance of the first peak occurred at shorter durations (150–250 ms) than in the present study. See also the last paragraph, discussing task-dependence of motor programs. Note further that the cited studies investigated hand acceleration rather than joint torque, and that these signals are related in a complex way in multijoint movements

The motor program concept does *not* imply that all arm movements must invariably exhibit one and the same profile shape. Rather, different shapes can be envisaged and have indeed been documented for single- and multijoint acceleration profiles (acceleration and torque are proportional in single-joint, but not in multijoint movements):

1. Under strict spatiotemporal constraints, profiles can be distinctly multi-phasic (e.g. Stein et al. 1988), probably due to corrective signals based on sensory feedback. The existence of corrections would refine rather than refute the motor program concept.
2. Movements not confined to the same plane as in the present study and/or executed with different objectives (e.g. accuracy or time constraints) may require other profile shapes. Similarly, more complex tasks such as sequential, reciprocal or cyclic movements are characterized by different, yet scalable, shapes (e.g. Viviani and Terzuolo 1980; Carter and Shapiro 1984; Schmidt et al. 1988; Sherwood et al. 1988). The availability of different shapes seems beneficial for the objective to meet different task demands and/or strategic considerations.
3. Humans can be trained to produce various unusual profile shapes (Brown and Cooke 1990; Cooke and Brown 1990). This finding could reflect the ability of the motor system to learn new programs, or its ability to use alternative control approaches.

In conclusion, the present study provided renewed support for the motor program concept, albeit in a modified form: our results suggest that the prototypical function is more closely related to joint torque than to individual muscle forces, and that scaling in time is achieved by two separate scaling factors.

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