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Fast corrective responses are evoked by perturbations approaching the natural variability of posture and movement tasks

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Crevecoeur F, Kurtzer I, Scott SH. Fast corrective responses are evoked by perturbations approaching the natural variability of posture and movement tasks. *J Neurophysiol* 107: 2821–2832, 2012. First published February 22, 2012; doi:10.1152/jn.00849.2011.—A wealth of studies highlight the importance of rapid corrective responses during voluntary motor tasks. These studies used relatively large perturbations to evoke robust muscle activity. Thus it remains unknown whether these corrective responses (latency 20–100 ms) are evoked at perturbation levels approaching the inherent variability of voluntary control. To fill this gap, we examined responses for large to small perturbations applied while participants either performed postural or reaching tasks. To address multijoint corrective responses, we induced various amounts of single-joint elbow motion with scaled amounts of combined elbow and shoulder torques. Indeed, such perturbations are known to elicit a response at the unstretched shoulder muscle, which reflects an internal model of arm intersegmental dynamics. Significant muscle responses were observed during both postural control and reaching, even when perturbation-related joint angle, velocity, and acceleration overlapped in distribution with deviations encountered in unperturbed trials. The response onsets were consistent across the explored range of perturbation loads, with short-latency onset for the muscles spanning the elbow joints (20–40 ms), and long-latency for shoulder muscles (onset > 45 ms). In addition, the evoked activity was strongly modulated by perturbation magnitude. These results suggest that multijoint responses are not specifically engaged to counter motor errors that exceed a certain threshold. Instead, we suggest that these corrective processes operate continuously during voluntary motor control.

feedback control; internal models; long-latency reflex; motor variability; upper limb

FEEDBACK CONTROL OF MOVEMENT is essential to guarantee movement success in the presence of sensory, central, and motor sources of movement variability (Churchland et al. 2006; Jones et al. 2002; Osborne et al. 2005; Todorov and Jordan 2002; van Beers et al. 2004). Besides the intrinsic neural variability, feedback control is also essential to compensate for perturbations arising from our interaction with the external world. For example, spinal reflexes of the arm can counter mechanical perturbations in as little as 20–40 ms. These purely spinal processes are followed by long-latency responses (50–105 ms) that can be modulated by explicit task goals (Crago et al. 1976; Marsden et al. 1981; Mutha et al. 2008; Pruszynski et al. 2008; Rothwell et al. 1980) and use internal models of limb and environmental dynamics (Gielen et al. 1988; Kimura and Gomi

2009; Krutky et al. 2010; Kurtzer et al. 2008, 2009; Soechting and Lacquaniti 1988). In this respect, the capabilities of long-latency responses mirror those shown in voluntary control. An important question in experimental and theoretical neuroscience is how feedback is used by the central nervous system (CNS) to correct for intrinsic as well as external sources of disturbances.

The functional similarities between rapid motor responses (<100 ms) and voluntary movements highlight the relevance of using perturbation paradigms to address feedback control mechanisms in general (Scott 2004). However, a current impediment to filling the gap between perturbation-evoked responses and voluntary control is that, to evoke strong and robust electromyographic (EMG) activity, many studies have used relatively large perturbations well beyond errors or variability usually encountered during voluntary control. For instance, typical upper limb perturbations found in the studies cited above induced a range of displacements that exceeded 1° in <100 ms for protocols using torque perturbations or 4 cm in the same time interval in studies using linear perturbation displacements. Such dramatic and abrupt shifts in limb position and velocity are clearly larger than the small deviations or variability inherent in natural motor behaviors and well beyond the movement perception thresholds from muscle afferent feedback (Hall and McCloskey 1983); note that none of the foregoing studies reports the unperturbed motor variability, which further compounds this ambiguity.

On the one hand, corrective responses could reflect the feedback processes inherently associated with voluntary control. Alternatively, corrective responses may be distinct from unperturbed behavior and only evoked when the system is significantly pushed beyond a certain threshold that is above natural variability. For example, when the stumble corrective response or flexion and crossed-extension reflexes are triggered, the walking process is interrupted to engage the corrective response (Duysens and Loeb 1980; Forssberg 1979; Lisin et al. 1973). Although often tacitly assumed, it remains unknown whether upper-limb rapid motor responses evoked by large perturbations are representative of a general feedback control policy for voluntary control or, alternatively, whether it is specific to situations requiring large corrective responses. We address this issue by reporting responses to a range of mechanical perturbations while participants performed postural control or reaching. We examined both reaching and posture as previous studies have highlighted distinct control processes for these behaviors (Brown et al. 2003; Kurtzer et al. 2005; Scheidt and Ghez 2007). In addition, the small pertur-

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bations were intended to induce joint motion that approached the natural variability observed during motor behavior. Thus the larger inherent variability of reaching movements can in this case be seen as a powerful asset because mechanical perturbations can more easily match the natural range of variability of this behavior.

We explore a known pattern of rapid motor responses observed with a combination of shoulder and elbow torques that produces pure elbow motion (Kurtzer et al. 2008, 2009). This mapping between multijoint torque and single-joint motion is due to complex intersegmental dynamics. In this condition, short- and long-latency responses are observed in muscles spanning the elbow, but long-latency responses are only observed in muscles spanning the shoulder. Long-latency activity at the shoulder in the absence of shoulder motion after the perturbation reflects internal knowledge of intersegmental dynamics. We tested this property across a range of perturbations, including very small ones, in an attempt to generalize the response pattern across different loads. Our results clearly demonstrate that the corrective response was engaged irrespective of the size of the perturbation-induced motion. These results suggest that feedback control processes observed for large perturbations are reflective of control processes for small perturbations or errors approaching those encountered during voluntary motor performance.

METHODS

Subjects

A total of 25 subjects (16 males and 9 females, between 18 and 37 yr of age) participated in 1 or more of the experiments. All subjects gave written informed consent and were paid for their time following procedures approved by the ethics committee at Queen's University.

Apparatus

As described in previous studies (Scott 1999; Singh and Scott 2003), subjects performed the task with a robotic exoskeleton (KINARM Exo-

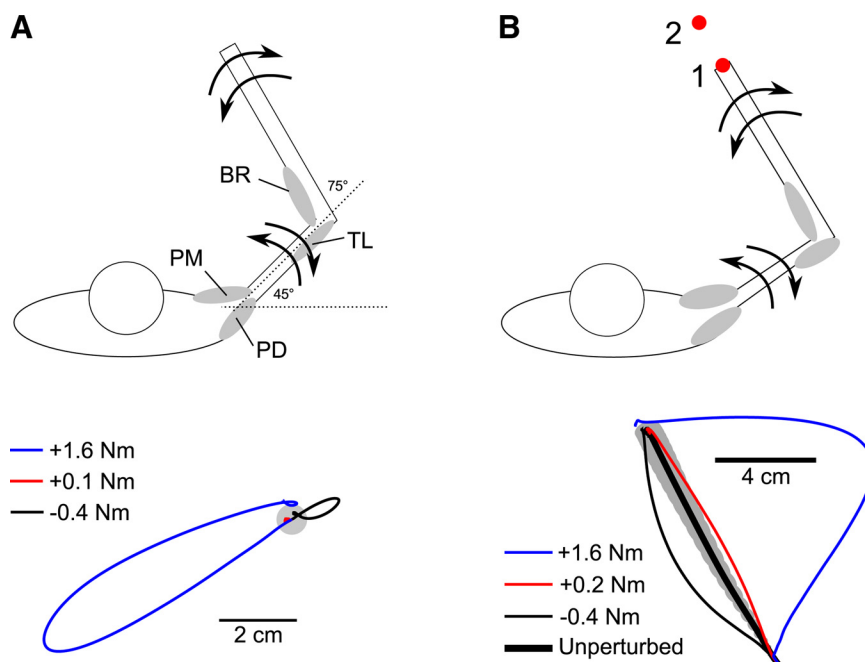
skeleton; BKIN Technologies, Kingston, Ontario, Canada) that allows planar movements of the limb and can selectively apply loads at the shoulder and/or elbow joints. A virtual reality display presented a hand-aligned cursor and visual targets in the same horizontal plane. Direct vision of the limb was blocked.

Tasks

The experiments used combined perturbations of shoulder and elbow joints to induce single-joint elbow motion (Kurtzer et al. 2008, 2009). Indeed, applying similar torques at each joint results in almost pure elbow motion up until ~ 100 ms postperturbation.

Postural control experiment. This experiment investigated the scaling of short- and long-latency activity of monoarticular shoulder and elbow muscles during postural control (Fig. 1A). Subjects had to stabilize a hand-aligned cursor on a visual target (radius 0.4 cm) against an extensor or flexor background load of 2 Nm applied at both shoulder and elbow joints. One group ($n = 8$) was exposed to a flexor background load (+2 Nm) that preexcited the shoulder and elbow extensors (posterior deltoid, PD, and triceps lateralis, TL; Fig. 1A), whereas a second group ($n = 8$) was exposed to an extensor background load (−2 Nm) that preexcited the shoulder and elbow flexors (pectoralis major, PM, and brachioradialis, BR). The target was located at shoulder and elbow angles of 45 and 75°, respectively. After a random delay ranging from 0.5 to 2 s, equal values of joint torque following a step profile were applied to both shoulder and elbow joints, producing almost pure elbow motion as described previously (Kurtzer et al. 2008). The same values of perturbation magnitudes were used for all subjects, which induced different amounts of elbow motion due to differences in inertial and mechanical properties of each individual's arms. Perturbations were randomly interleaved and included a range of 8 different values: −0.4, −0.2, −0.1, 0.1, 0.2, 0.4, 0.8, and 1.6 Nm, relative to the background load. Subjects were asked to stabilize on the target as accurately as possible and compensate for the perturbations applied by the robot. The cursor corresponding to the subject's hand was extinguished at the time of perturbation onset so that corrective responses were only driven by limb afferent feedback. For each subject, 48 trials were collected per perturbation magnitude. Hand paths are presented in Fig. 1A for 3 selected perturbations.

Fig. 1. Posture and reaching experiments. *A*: overhead view of the joint configuration used for the posture task. The 4 muscles of interest are represented: pectoralis major (PM), posterior deltoid (PD), triceps lateralis (TL), and brachioradialis (BR). The *bottom* diagram represents hand trajectories for the 3 selected perturbations. The gray disk represents the target (radius 0.4 cm). *B*: overhead view of the joint configuration and visual targets used in the reaching experiment. Data presented are for the condition in which there was an extensor load to preexcite the flexors. Mean hand trajectories for the unperturbed and perturbed reaches are presented below. The shaded area represents the 95% confidence ellipses on the 2-dimensional distribution of unperturbed reaches estimated every 20 ms.



An additional control experiment was performed to confirm that elbow motion evoked a response at the shoulder. At the same joint configuration used in the previous task, 5 subjects were 1st exposed to a flexor background load at the shoulder to evoke activity in the subjects' posterior deltoid (20 trials $\times$ 5 s). Subjects had visual feedback of the evoked activity displayed on the screen of an oscilloscope. Then, the shoulder joint of the KINARM robot was locked, and subjects were asked to push on the arm trough with shoulder extension until the EMG displayed on the oscilloscope matched the activity evoked by the previous 2-Nm background load. Perturbations of ± 1 Nm were then applied at the elbow, and subjects were again required to stabilize their hand back at the spatial target (same procedure as in the main experiment). Perturbation direction was randomly interleaved to avoid anticipation, and subjects performed a total of 40 trials in each direction.

Reaching control experiment. This experiment addressed how perturbation-evoked muscular activity scaled as a function of perturbation magnitude when subjects performed a reaching task. Subjects were presented with two visual targets: the start target was identical to the target used in the experiments investigating postural control, and the goal target was initially an open red circle of radius 1 cm. After a random time following stabilization in the start target (0.5–1 s), the goal target became a filled red circle, cueing the subject to reach for it. The reaction time was not constrained, but the movement time from leaving the start target to the time the subject remained in the goal target was constrained to <750 ms. This time constraint encouraged the subjects to keep consistent reaching velocities across trials, and all trials, even when unsuccessful, were included in the analysis. The perturbations were triggered at the beginning of the reach based on a position threshold of 0.5 cm toward the goal target. The same perturbation magnitudes as in the postural control experiment were used in this experiment. We collected 48 trials for each perturbation magnitude, 48 unperturbed trials without visual feedback, and 168 unperturbed trials with visual feedback of the hand-aligned cursor (600 trials in total, of which 36% were unperturbed). Having a fixed threshold for triggering the perturbations made their time and location highly predictable. However, this experimental choice is important to standardize joint angles and background muscle activities across trials, as these 2 factors can influence the magnitude of motor responses. The randomness of perturbation direction and magnitude was therefore critical to avoid anticipation.

The x - and y -coordinates of the two targets used in this experiment were (2.5, -4.5) cm and (-2.5 , 4.5) cm relative to shoulder and elbow angles of 45 and 75°, resulting in movement amplitude of ~ 10 cm between the two targets (Fig. 1B). A first group of eight subjects performed reaching movements requiring shoulder flexion and elbow extension (from *target 1* to *target 2*; Fig. 1B). An extension background load (-2 Nm) was applied at the two joints to preexcite the shoulder and elbow flexors. A second group of eight subjects performed the reaches in the opposite direction (from *target 2* to *target 1*) against a flexor background load to preexcite the extensors ($+2$ Nm). The target locations and threshold for perturbation onset were chosen such that the shoulder and elbow configuration at perturbation onset were roughly similar to the arm configuration used for the postural control experiment, and the perturbations were applied near the end of the shoulder agonist activity. In addition, the reach direction was aligned on the orientation of the forearm making the motion induced by the perturbation, at least locally, orthogonal to the movement. As in the posture experiment, the hand-aligned cursor was extinguished at perturbation onset.

Data Collection and Analysis

Angular positions of the shoulder and elbow joints were collected at 1 kHz and digitally low-pass filtered with a 4th order 0-lag Butterworth filter with a 20-Hz cutoff frequency. Muscle activity was collected with surface electrodes (DE-2.1; Delsys, Boston, MA)

attached after light abrasion of the skin. A reference electrode was attached on the ankle or on the knee. The signal was amplified by 10,000 and collected at 1 kHz. The signal was digitally band-pass filtered with a 4th order 0-lag Butterworth filter between 15 and 500 Hz. After filtering, the signal was rectified and normalized to the average muscle activity in a 500-ms time window during the initial stabilization period for each trial.

An initial concern with the small perturbations was that the frictional properties of the KINARM might oppose the motion induced by the very small changes in load. The static friction measured on the robot is ~ 0.06 Nm. To compare the actual perturbation magnitude with the commanded value, we used accelerometers attached to the structure supporting the forearm at a fixed location relative to the elbow rotation axis (EGAXT-5; Entran, Fairfield, NJ). The offset was removed, and the signal was integrated in a time window corresponding to the buildup of the perturbation. The integrated accelerometer signal was then calibrated with the velocity data based on 0.8-Nm perturbations, used as a calibration perturbation. As expected, the smallest perturbations in absolute value (± 0.1 Nm) were more affected by energy dissipation and opposition of frictional forces, but the overall loss of torque buildup was $<10\%$ of the commanded value. Correcting for this effect had no influence on the results. Therefore, unless specified, we use the commanded torque values for simplicity.

We followed classic definitions of the different epochs of time found in the literature (Crago et al. 1976; Lee and Tatton 1982; Nakazawa et al. 1997; Pruszynski et al. 2008). The first epoch (R1) occurs from 20 to 45 ms postperturbation and is referred to as the short-latency response. The long-latency response is divided into two epochs: R2 between 45 and 75 ms postperturbation and R3 between 75 and 105 ms postperturbation. The voluntary activity was defined between 120 and 180 ms postperturbation.

The overlap in distribution between perturbation-induced motion and the natural variability was addressed by receiver operating characteristic (ROC) analysis (Corneil et al. 2004; Pruszynski et al. 2008). This technique is used to compute a time series of estimates of the probability to discriminate two random variables based on their empirical distribution. If the two distributions are the same, the ROC area is 0.5, meaning that discriminating the two variables is equivalent to the toss of a coin. If the two distributions are strictly nonoverlapping, the ROC area is 1 or 0, meaning that there is a decision value achieving exact discrimination with probability 1 (Metz 1978). For the posture experiment, we used the data before the perturbation to estimate the joint displacement across a given time window corresponding to the natural variability of postural control. For the reaching experiment, the time series of ROC were simply obtained by comparing the perturbation-evoked motion with that of the unperturbed reaches. ROC areas were computed on the distribution of individual trials for each subject, and the resulting time series were averaged across them. ROCs for reaching movements were based on the comparison between perturbed and unperturbed distributions after subtracting the average unperturbed movement as well as the initial angle and velocity at 20 ms before perturbation onset for each individual trial. This approach is therefore not strictly related to perturbation-evoked motion since, unlike in the posture case, some variability inherent to reaching movement persists and the histograms do not reflect only perturbation-related motion. We verified that removing the position and velocity offsets at different points in time had qualitatively no impact on the results.

The onset of perturbation-evoked activity was estimated by the time when the averaged EMG signal exceeded 2 SD of the baseline for ≥ 50 ms. The baseline was defined as the activity before the perturbation for postural control and the activity in the unperturbed condition for reaching at the time the perturbations were triggered. Therefore, the time-varying activation profile corresponding to unperturbed reaches was subtracted from the activity of perturbation trials to extract the response. All remaining analyses used classic one-way ANOVAs, (paired) t -tests, or linear regressions. The t -tests were

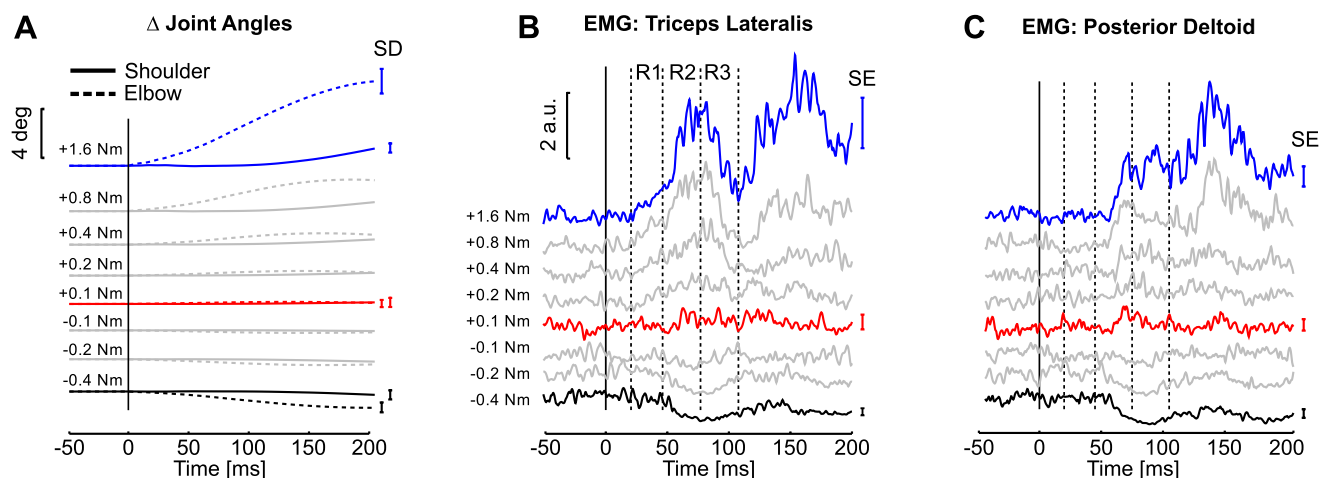


Fig. 2. Posture data. *A*: plot of the average elbow (dashed) and shoulder (solid) motions induced by the perturbation for all perturbation loads. Three perturbation magnitudes were emphasized with the same color code as in Fig. 1 for comparison: -0.4 Nm (black), +0.1 Nm (red), and +1.6 Nm (blue) relative to the 2-Nm background load. *B* and *C*: average electromyogram (EMG) of elbow (TL) and shoulder (PD) extensor muscle for the 3 selected perturbations (epochs 1–3, R1–3). The different epochs of the rapid motor responses are presented with dashed lines. deg, Degrees; a.u., arbitrary units.

based on the mean response for each subject, integrated across the different epochs of time used to characterize motor responses. The paired *t*-tests allowed us to examine perturbation-related changes in EMG relative to the unperturbed condition for each individual subject. For the posture experiment, the paired *t*-tests compared perturbation-evoked response with the pre-perturbation activity (-30 to 0 ms).

Linear regressions of perturbation-evoked activity as a function of loads were used to address whether the perturbation-evoked response scales as a function of the change in load. In addition, the offsets of the linear regressions allowed us to examine whether rapid motor responses scale down to 0 or whether offsets indicates possible limits or thresholds where no response is observed. These regressions were restricted to the lower perturbation magnitudes, including ± 0.4 , ± 0.2 , and ± 0.1 Nm, to avoid nonlinear effects, such as saturation present for higher loads, influencing the estimation of regression parameters. For the reaching experiment, the regressions were computed after subtracting the average activity of unperturbed reaches for each individual subject.

RESULTS

Postural Control

The hand trajectory and muscle activities of one representative subject are shown in Fig. 2. Three selected perturbations are emphasized with color traces, including the smallest perturbation used in this protocol (0.1 Nm; red traces). It can be observed that the small loads induced a very small amount of hand and joint motion compared with the larger ones (Figs. 1*A* and 2*A*). Also, the different perturbation magnitudes produced different amounts of elbow motion, whereas there was almost no perturbation-related motion at the shoulder for ≥ 100 ms after the perturbation (Fig. 2*A*). The activity of triceps lateralis in response to large perturbations (Fig. 2*B*; blue trace) presents a typical response pattern with a short-latency increase in activity after ~ 20 ms, followed by multiphasic bursts corresponding to long-latency and voluntary responses. The inhibitory response follows a similar overall pattern (black trace). In contrast, the activity of posterior deltoid presents a distinct signature where no short-latency activation can be observed (Fig. 2*C*).

In general, the perturbation-related motion followed the same pattern as shown in Fig. 2 across subjects and experiments. A more detailed analysis is presented in Fig. 3 where

average shoulder and elbow motion following the perturbation of +0.1 Nm were magnified (data from the 8 subjects involved in the experiment with extensors preexcited). It can be observed that the perturbation mostly displaced the elbow (black traces), whereas the shoulder (gray traces) displays only residual motion until ~ 100 ms post-perturbation. Within 50 ms of perturbation onset (+0.1 Nm), the maximum shoulder displacement averaged across subjects was $0.005 \pm 0.001^\circ$ (mean $\pm$ SD), the maximum shoulder velocity was $0.25 \pm 0.06^\circ/\text{s}$, and the maximum shoulder acceleration was $18 \pm$

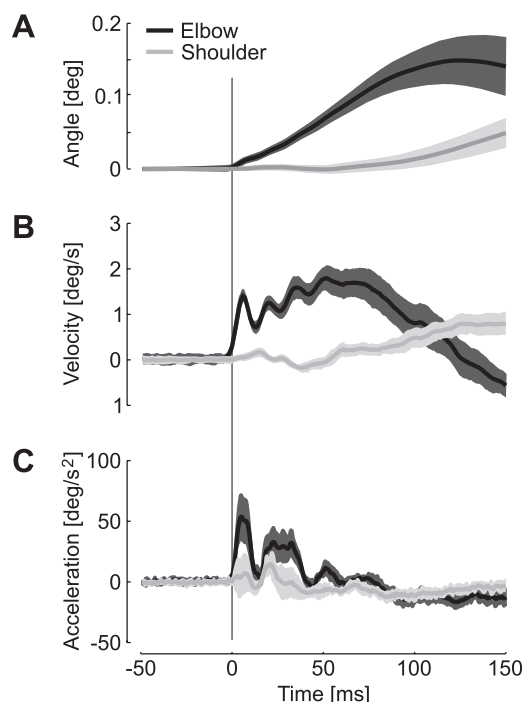


Fig. 3. Perturbation-induced motion. *A*: elbow (dark gray) and shoulder (light gray) displacement following the perturbation of +0.1 Nm. The shaded areas represent 1 SD across the 8 subjects involved in the postural control experiment with extensor muscles preexcited. *B*: same as *A* with joint velocity. *C*: same as *A* with the data collected from linear accelerometers attached to the exoskeleton.

$11^\circ/\text{s}^2$. The maximum elbow displacement and velocity after the same perturbation were an order of magnitude greater with 0.06 ± 0.008 and $1.8 \pm 0.23^\circ/\text{s}$, respectively (Fig. 3, *A* and *B*). Elbow acceleration was $54 \pm 18^\circ/\text{s}^2$ (Fig. 3*C*). For the largest perturbation ($+1.6$ Nm), the maximum shoulder displacement across subjects within the same time window was $0.04 \pm 0.007^\circ$, the velocity was $5 \pm 0.65^\circ/\text{s}$, and the acceleration was $330 \pm 210^\circ/\text{s}^2$. The maximum elbow displacement and velocity at 50 ms following 1.6 Nm were consistently greater than for the shoulder motion by 1 order of magnitude: $1.3 \pm 0.1^\circ$ and $38 \pm 5^\circ/\text{s}$. The maximum elbow acceleration was $1,014 \pm 245^\circ/\text{s}^2$.

Figure 4 shows that the smallest perturbations induced motion at the elbow that approached the natural variability present during postural control. Indeed, the distributions of joint angle, velocity, and acceleration clearly overlapped with the range corresponding to unperturbed postural control (red and black histograms, Fig. 4*A*). Figure 4, *A–C*, illustrates this overlap by showing the empirical distributions of joint angle, velocity, and acceleration at different times corresponding to the times when the time series of ROC area is maximum (Fig. 4*D* for elbow angle, Fig. 4*E* for velocity, and Fig. 4*F* for acceleration). Therefore, the histograms presented in Fig. 4, *top*, show the individual distributions of each kinematic parameter at the moment when the perturbation-related shift in distribution is maximum. Subjects informally reported that they were not consistently aware that a perturbation had been

applied to their arm, suggesting that the smallest perturbation was close to or below perceptual thresholds. In contrast, perturbations as small as 0.4 Nm already produced shifts in position, velocity, and acceleration that resulted in no overlap with the unperturbed distribution of each variable (blue histograms). Figure 4, *D–F*, quantifies the shifts in distributions by showing time series of ROC areas. It can be observed that loads of 0.4 Nm generate joint velocity and acceleration that exceed the range of unperturbed postural control almost immediately after perturbation onset, whereas the shift in joint angle totally exceeds the natural range after ~ 50 ms. The same comments apply to negative perturbations relative to the background load. Subjects exposed to the opposite background and perturbation loads displayed similar tendencies. Perturbations of higher magnitude than 0.4 Nm obviously generate more drastic changes in kinematic parameters.

Notably, even for perturbation magnitude approaching natural variability, evidence for motor responses to the smallest perturbations could be reliably extracted and are presented in Fig. 5 (± 0.1 Nm, 5% of the background load). Figure 5 shows that the average muscle activity deviates from baseline for both elbow (*top*) and shoulder muscles (*bottom*). These perturbations failed to evoke significant R1 activity of the elbow flexor, although some deviation was clearly present in this time window (Fig. 5, *top left*). Significant R2/3 activity for both elbow and shoulder flexors was observed [paired *t*-test, $t_{(7)} > 2.6$, $P < 0.05$]. The evoked activity was more vigorous for the

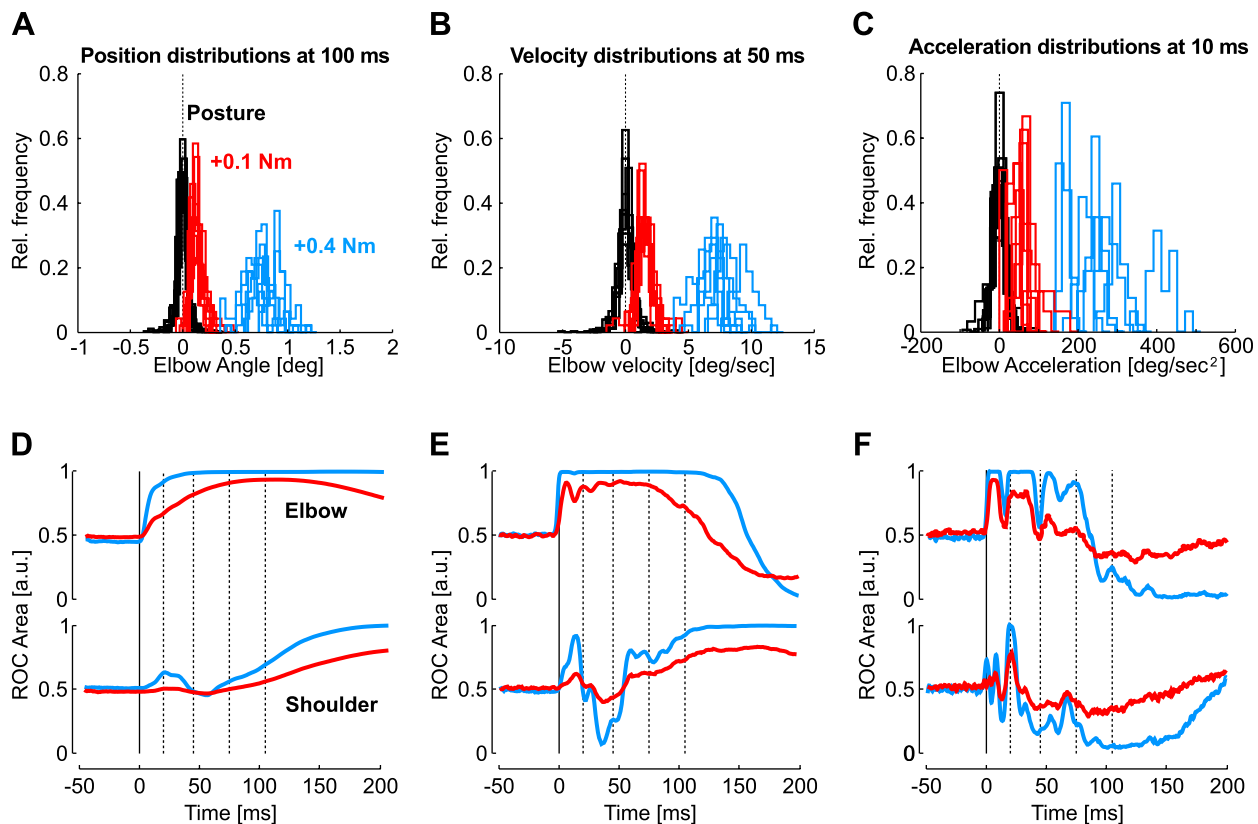
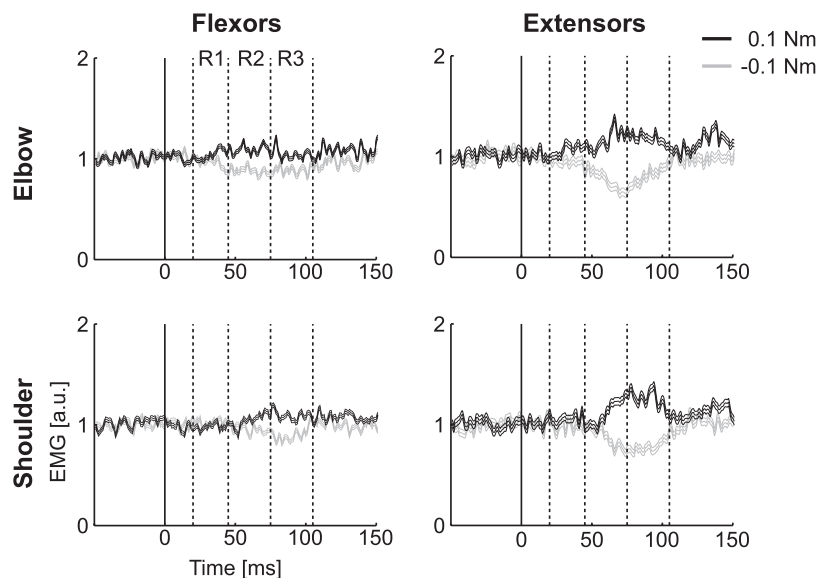


Fig. 4. Distributions of elbow angle, velocity, and acceleration during posture. *A*: distributions of elbow angle displacement and velocity for all subjects during postural control (black), after +0.1-Nm perturbation (red) and +0.4 Nm (blue). Joint angle distributions are shown 100 ms postperturbation, which corresponds to the time when the motion induced by the +0.1-Nm perturbation is maximum. The histograms of individual subjects were defined relative to the average motion during unperturbed postural control. *B*: same as *A* with velocity data. *C*: same as *A* with acceleration signals. *D*: time series of receiver operating characteristic (ROC) areas comparing shoulder and elbow angle for unperturbed vs. +0.1 Nm (red) and unperturbed vs. +0.4 Nm (blue). *E*: same as *D* for joint velocity. *F*: same as *D* for the accelerometer signals. Rel., relative.

Fig. 5. Response to small perturbations during posture. EMG activity as a function of time for the 4 muscles of interest in response to perturbations of 0.1 Nm (black) and -0.1 Nm (gray) relative to the background load (2 Nm). Displays are the means $\pm$ SE across the 8 subjects involved in each series of experiment.



extensors: significant R1 activity of the elbow extensor was observed in response to the $+0.1$ -Nm perturbation [$t_{(7)} > 3.3$, $P < 0.05$], and both shoulder and elbow extensors exhibited significant R2/3 activity [$t_{(7)} > 4.4$, $P < 0.01$].

Figure 6 presents the group activity as a function of load for short- and long-latency epochs (R1 and R2/3 grouped together). Separate analysis of each subject revealed a significant modulation of R1 activity in 3/8 subjects' elbow extensor and in 6/8 subjects' elbow flexor (linear regression of EMG activity

of individual trials activity vs. perturbation load, $P < 0.01$). Since the perturbation did not stretch the shoulder, no change in activity was observed in the R1 time window of these muscles (Fig. 6, Shoulder, R1). However, all subjects exhibited highly significant modulation of activity in R2, R3, and voluntary time windows in the 4 muscles of interest ($P < 0.05$, all but 1 $P < 10^{-5}$). Importantly, the linear regressions of subjects' averaged activity as a function of the perturbation loads suggest that the EMG-load relationship passes through 0. Indeed, a non-0 intercept would argue for threshold-based mechanisms. This analysis was restricted to $+0.1$ -, $+0.2$ -, and $+0.4$ -Nm perturbation to avoid that the presence of positive and negative values enforced the regression passing through 0. We also used the corrected load based on accelerometer signals for each subject. We found no estimated intercepts in the R1 and R2/3 time windows that significantly differed from 0 [4 muscles $\times$ 2 epochs of time, lower bound of 95% confidence interval < -0.03 arbitrary units (a.u.), upper bound > 0.04 a.u.].

One critical result is whether the response onsets vary as a function of the perturbation load, which would indicate that error accumulation and/or threshold-based mechanisms are necessary to launch the corrective response. The analysis of estimates of activity onsets presented in Fig. 7 allows us to reject this possibility. Indeed, although some estimates vary across loads, all data points show a short-latency elbow response (R1; dashed gray), whereas all estimates of shoulder activity onset occur in the early long-latency epoch of time (R2; solid gray). It is important to realize that the onsets for smaller perturbation loads are slightly overestimated because the change in activity is not as steep for these perturbations as those of the larger ones and therefore takes a little longer to exceed the 2-SD threshold. Thus the apparent trend observed for small loads in Fig. 6 is likely due to the determination of the onset rather than shifts in the actual timing of the response. Overall, the typical pattern of activation observed in response to larger perturbation magnitudes was reproduced across the explored range of loads.

Although there is little to no shoulder motion until 100 ms of perturbation onset, the analysis of joint kinematics revealed a

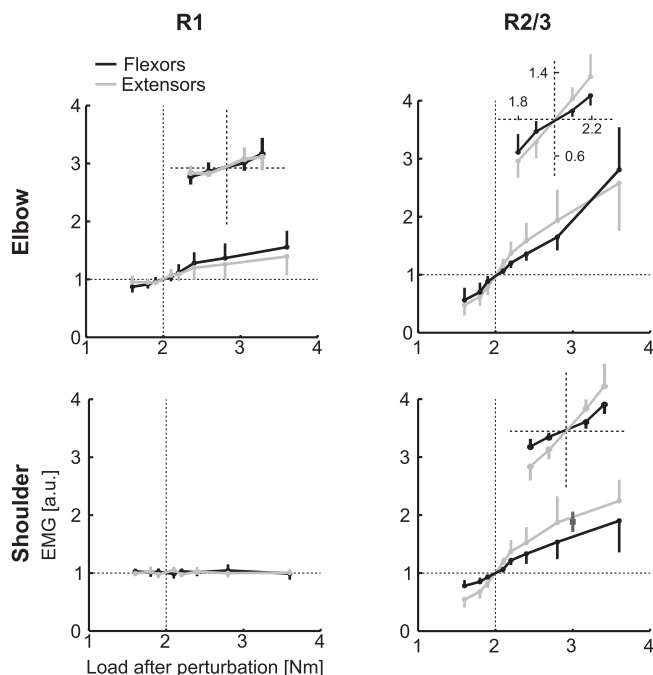


Fig. 6. Time course of modulation gain during posture. A: modulation of evoked activity as a function of the perturbation loads for flexors (black) and extensors (gray) in the short-latency (R1; left) and long-latency (R2/3; right) epochs of time. Means $\pm$ SD across subjects is represented for each perturbation magnitude. The modulation of muscle activity is presented across the whole range (-0.4 to 1.6 Nm relative to the background load), and smaller perturbations are magnified in the insets (-0.2 to 0.2 Nm). The data from the control experiment with the shoulder physically locked are plotted in dark gray (means $\pm$ SD across the 5 subjects).

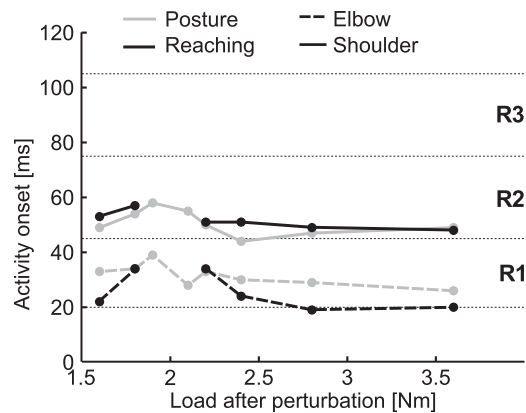


Fig. 7. Estimation of activity onsets. Plot of the estimated activity onset as a function of perturbation loads for elbow (dashed) and shoulder (continuous) muscles during posture (gray) and reaching (black). The discontinuous traces for the reaching enhance that the onset in response to ± 0.1 Nm could not be reliably extracted.

peak in shoulder acceleration at ~ 20 ms (Figs. 3C and 4F), which could potentially evoke a short-latency response at 40–45 ms. It was therefore necessary to verify that the long-latency shoulder response, analyzed above, was not generated by a short-latency response to the small shoulder acceleration. The control experiment with the shoulder joint locked in position confirmed that, in the absence of shoulder motion, elbow flexion and extension evoke a clear response in the shoulder extensor muscle (Fig. 8B). This shoulder response is qualitatively similar to that observed in the shoulder-free condition: no change in activity during R1 [paired t -test compared with preperturbation activity across subjects, $t_{(4)} = 2.21$, $P > 0.05$], whereas R2, R3, and early voluntary epochs of time exhibit significant responses [Fig. 8B; $t_{(4)} > 2.8$, $P < 0.05$] in all but one case (R3 following -1 -Nm perturbation; Fig. 8B). The average activity in the R2/3 window evoked by the flexion load is plotted in Fig. 6 (means $\pm$ SD; dark gray) to illustrate that the amplitude of the response is compatible with the response measured in the shoulder-free condition.

Reaching

Typical traces from one representative subject are shown in Fig. 9. The reaching movements involved simultaneous shoulder flexion and elbow extension (Fig. 9A). Typical movements were smooth and presented single-peak velocity profiles. The difference in joint angle between perturbed and unperturbed reaches following the perturbation is shown in Fig. 9A, right. It can be observed that the combined perturbation induced pure elbow motion relative to the unperturbed reaches until ~ 100 ms postperturbation. Figure 9B, left, presents average EMG activity of elbow and shoulder flexors across the whole reaching movement for one exemplar subject. Traces were aligned on the position threshold used for perturbation onset. The time window from -50 to 200 ms relative to perturbation onset is magnified in Fig. 9B, right. A clear increase in activity of the elbow flexor in response to large perturbations can be observed in the R1 time window (blue traces), whereas the shoulder flexor shows a later increase in activation.

The overlap between perturbation-induced motion and natural variability is markedly stronger during reaching than postural control. This is partially due to the higher variability

inherent in reaching movements. Another factor that can influence the distribution of perturbation-related motion is that changes in joint stiffness during reaching induce smaller amounts of elbow motion following the perturbation. Indeed, the elbow displacement measured 20 ms after perturbation onset relative to unperturbed reaches was significantly smaller than the same quantity measured during posture (Wilcoxon rank sum test, $z > 4$, $P < 0.01$). The distributions are illustrated in Fig. 9, A and B, where the black histograms represent the distribution in elbow angle and angular velocity, respectively. The selected times correspond to the maximum perturbation-related shifts in distributions, as we performed for the posture data. The changes in position and velocity induced by the $+0.2$ -Nm perturbations are clearly in the same range as the variability of unperturbed reaches (Fig. 10, A and B, black and red histograms), whereas perturbation magnitude of 1.6 Nm produced dramatic changes in both position and velocity, resulting in little to no overlap with movement from the unperturbed condition (blue histograms). Figure 10C quantifies the overlap in distributions by showing the time series of ROC areas for shoulder and elbow angles and angular velocities. The ROC time series illustrate that shoulder and elbow motion following the smaller perturbations is almost indistinguishable from the unperturbed condition. Nonetheless, the typical response observed above could be reliably extracted as addressed below.

Responses to the smallest perturbations during reaching (± 0.1 Nm) were not significant, likely due to the higher

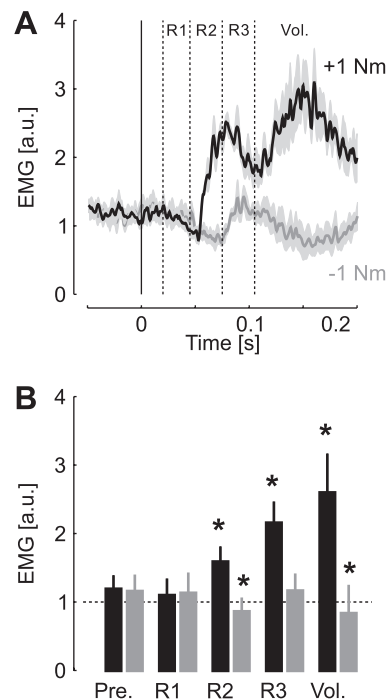


Fig. 8. Shoulder muscle response with shoulder joint locked in place. A: average perturbation evoked response by the shoulder extensor (PD). Responses to flexion and extension loads are shown in black and gray, respectively. The shaded area represents ± 1 SE across the 5 tested subjects. Responses were normalized to the 2-Nm evoked background activity for each individual subject. B: binned analysis of the evoked response across the different epochs of time with the same color code as in A. The dashed line represents the average activity evoked by 2-Nm background load. *Significant increase in EMG relative to the preperturbation activity at the level $P < 0.05$. Vol., voluntary; Pre., preperturbation.

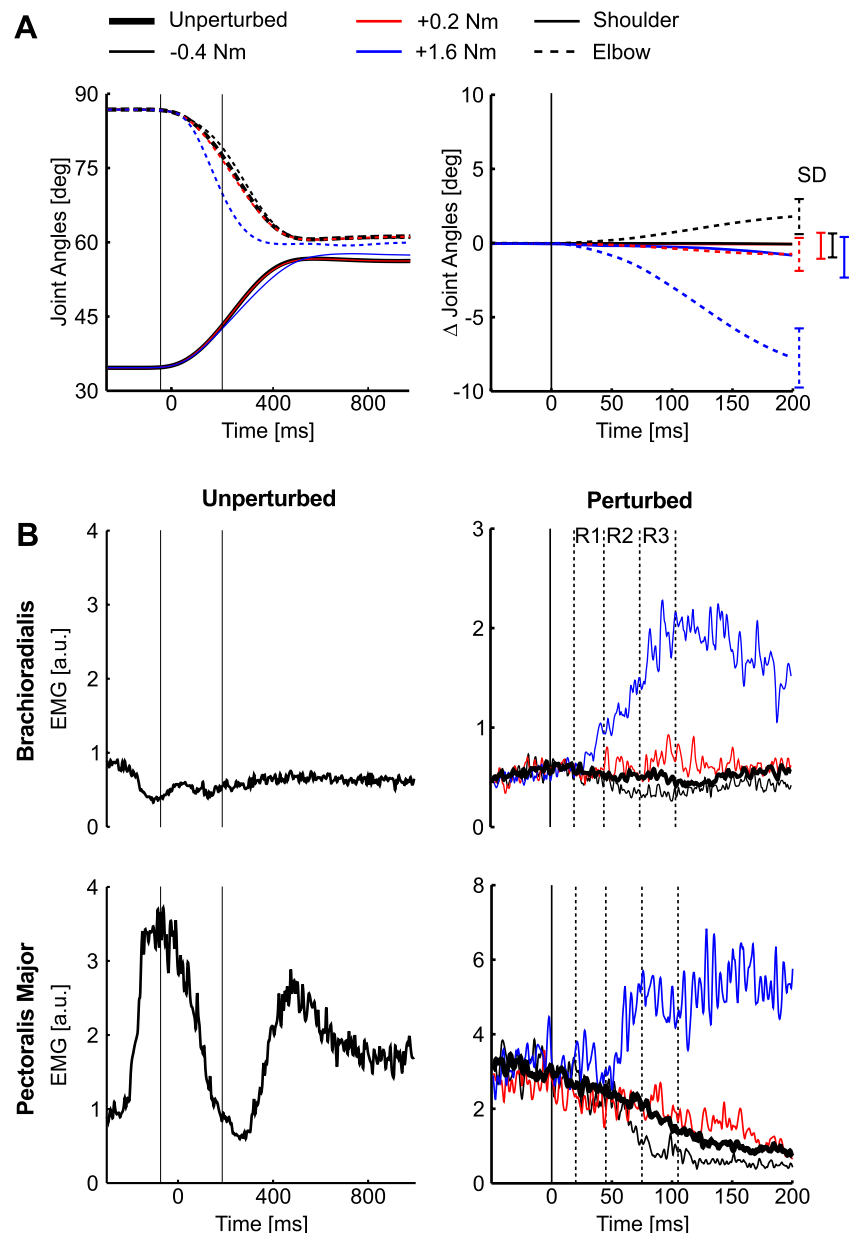


Fig. 9. Reaching data. *A*: average elbow (dashed) and shoulder (solid) angle as a function of time for 1 representative subject. The color traces correspond to 3 selected perturbation magnitudes including -0.4 Nm (black), $+0.2$ Nm (red), and $+1.6$ Nm (blue) relative to the 2-Nm background load. The dark gray trace is the average across unperturbed reaches. The 2 solid vertical lines illustrate the time window expanded on the *right* to magnify joint motion following the perturbation. The *right* presents the average change in joint angles relative to the unperturbed reaches. There is an overlap between the black trace (-0.4 Nm) and the red trace (0.2 Nm) due to the small amount of shoulder motion induced by each perturbation. SD measured at 200 ms of perturbation onset are shown at the end of each trace. *B*: *left* shows the average EMG of elbow and shoulder flexor muscles throughout the whole movement. The solid vertical lines on the *left* represent the time window expanded on the *right*. The dashed vertical lines on the *right* represent the different epochs of the rapid motor responses. These panels show the average EMG of an elbow flexor (BR; *top*) and shoulder flexor (PM; *bottom*) in the unperturbed (thick black) and perturbed conditions for a single subject.

variability of the unperturbed reaches used as baseline. However, responses could be clearly observed for perturbations corresponding to ± 0.2 Nm relative to the background load ($\pm 10\%$). Figure 11 shows the EMG as a function of time for the four muscles. Significant activity was observed during R1 for the elbow flexor [paired *t*-test, perturbed against unperturbed reaches, $t_{(7)} > 3$, $P < 0.05$] and during the R2/3 time window for the shoulder and elbow flexors [$t_{(7)} > 4.3$, $P < 0.01$]. The same perturbation magnitude failed to evoke significant R1 activity for the elbow extensor but evoked significant R2/3 response for both shoulder and elbow extensors [$t_{(7)} > 3.3$, $P < 0.05$].

The size of the perturbation load had a clear impact on the size of the EMG response. Group data are presented in Fig. 12. Although the response to ± 0.1 Nm was not significant, a trend can be observed in the *insets* of Fig. 12 where responses to ± 0.1 - and ± 0.2 -Nm perturbation were magnified. R1 activity of brachioradialis was significantly modulated for all subjects

(linear regressions, all $P < 0.01$). R1 activity of triceps lateralis was significant for only 2/8 subjects ($P < 0.05$). In contrast, a large majority of muscle samples were significantly modulated by the load magnitude during the R2 (29 of 32), R3 (31 of 32), and voluntary (all regressions) epochs of time (4 muscles $\times$ 8 subjects; $P < 0.05$).

Like in the posture experiment, the regressions of subjects' average activities as a function of perturbation loads revealed that the intercepts were generally not significantly different from 0, except in 1 case that is likely a false-positive result as it concerns the R1 time window of the posterior deltoid that is not stretched by the perturbation. In the 7 remaining cases (4 muscles $\times$ 2 time windows), the 95% confidence interval on the regressions intercepts included 0 (lower bound < -0.07 a.u., upper bound > 0.006 a.u.).

The response onset is again a critical parameter that can be used to determine whether similar response patterns are engaged across the explored range of perturbation loads. A slight

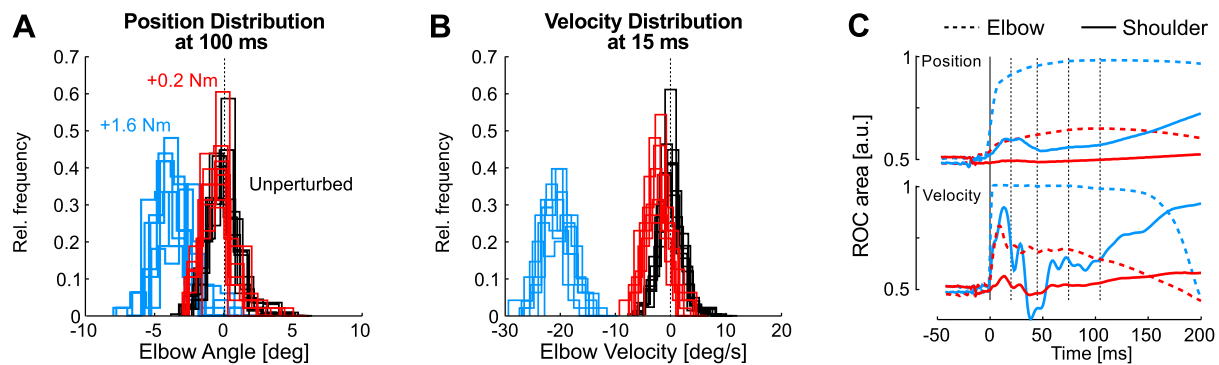


Fig. 10. Joint angle and velocity distributions during reaching. *A*: distributions of elbow angles for all subjects during unperturbed reaches (black) and after +0.2-Nm perturbation (red) and +1.6 Nm (blue). *B*: same as *A* with elbow velocity. Distributions are shown 100 and 15 ms postperturbation, respectively. *C*: time series of ROC areas comparing elbow (dashed) and shoulder (solid) angle and velocity for unperturbed vs. +0.2 Nm (red) and unperturbed vs. +1.6 Nm (blue). The dashed vertical lines represent the different epochs of rapid motor responses.

increase in response time for the smaller perturbation is expected as for the posture data, which is likely due to the method used to determine the onset. We found that the response times were similar across all loads and across the two experiments with short-latency responses for the elbow muscles (Fig. 7, dashed black; onset < 40 ms) and long-latency responses for the shoulder muscles (Fig. 7, solid black; onset > 45 ms).

DISCUSSION

The aim of this study was to observe whether rapid motor responses could be evoked by very small mechanical perturbations inducing position and velocity errors approaching those encountered during unperturbed postural control or reaching tasks. We reported clear responses to ± 0.1 -Nm loads during postural control, and a trend could be observed for the same perturbation loads during reaching (see Fig. 12, *insets*). More robust responses to ± 0.2 Nm could be observed during reaching, whereas the distributions of perturbation-related deviations almost entirely overlapped with those of unperturbed reaches. EMG responses scaled down with load magnitude and exhibited a linear trend passing through zero. Critically, response timing remained consistent across all perturbation magnitudes, with short- and long-latency activity for the muscles

spanning the elbow joint (onset < 40 ms) and long-latency for shoulder muscles (onset $\approx$ 50 ms).

There are several challenges to quantify rapid motor responses associated with very small perturbations. To begin, the variability of EMG measurements is a clear limitation to our ability to extract responses to very small changes in loads, in particular when the response must be extracted from a time-varying signal as during reaching. Second, the frictional forces can impact the transmission of motion from the applied loads when applying very small perturbations. This limitation was greater for the posture experiment, where the natural position and velocity distributions of the unperturbed condition were extremely narrow and therefore very hard to match with perturbations that reliably overcome frictional constraints. A final challenge is that although position and velocities between perturbed and unperturbed trials overlapped, there likely remain differences for higher-order derivatives, including joint acceleration, to which muscle spindles are possibly sensitive (Lennerstrand and Thoden 1968; Matthews and Stein 1969; Schäfer 1967). Indeed, recent studies on grasping and key-pressing tasks have emphasized that spindle discharge of wrist muscles during natural hand movements were best accounted for by velocity and acceleration signals rather than muscle

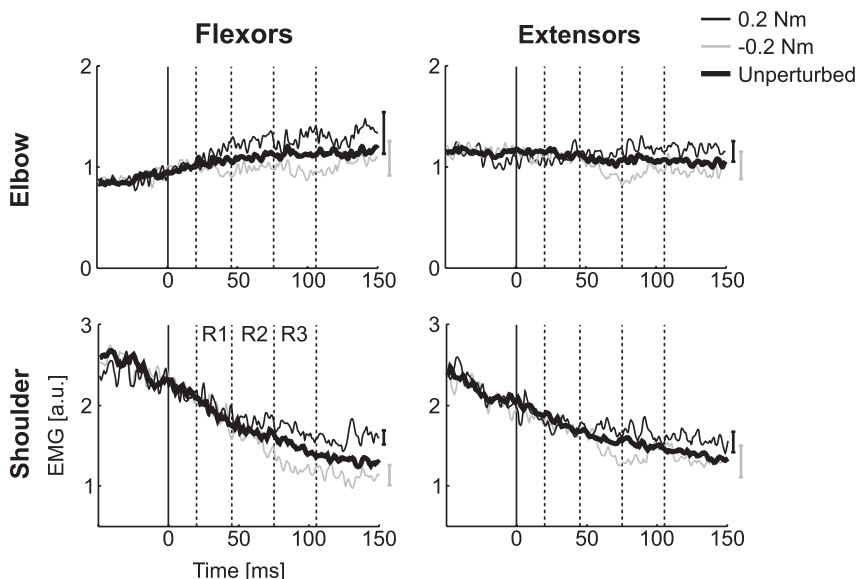


Fig. 11. Response to small perturbations during reaching. EMG as a function of time for the 4 muscles of interest. The thick black trace corresponds to the activity measured during unperturbed reaching averaged across trials and subjects. The thin traces represent the mean across subjects corresponding to perturbation trials using ± 0.2 -Nm perturbations ($\pm 10\%$ of the background load). Error bars represent the means $\pm$ SE across subjects at 150 ms of perturbation onset.

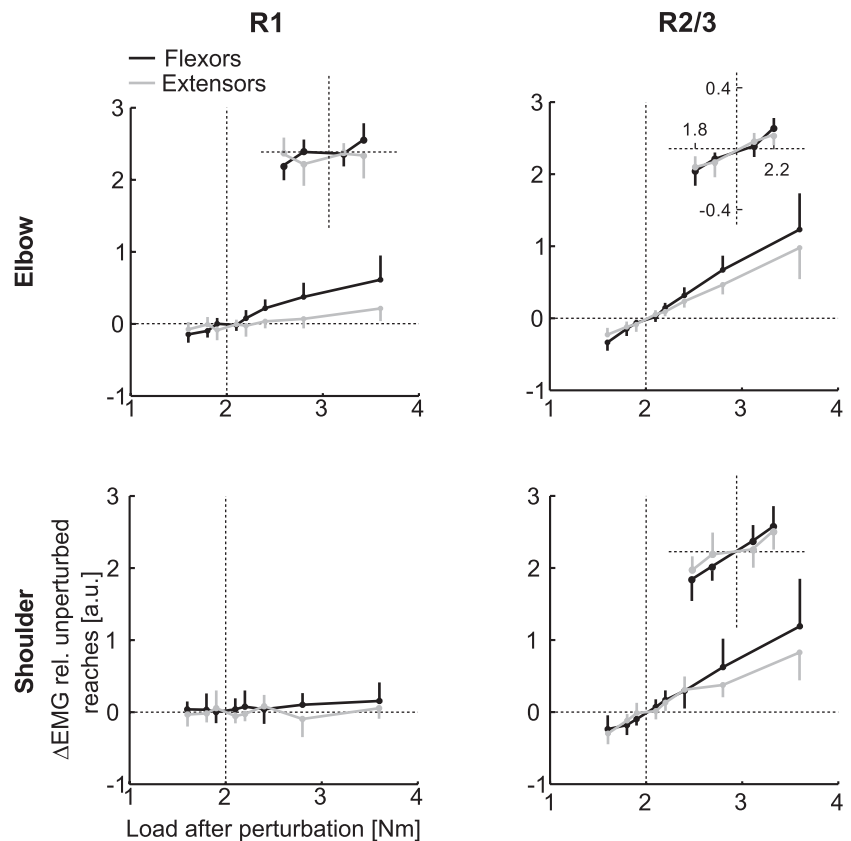


Fig. 12. Time course of modulation gain during reaching. Modulation of muscle activity as a function of the perturbation load in short- and long-latency time windows (R1 and R2/3) is shown. Each *inset* magnifies the results for smaller perturbations, including ± 0.2 and ± 0.1 Nm relative to the background load.

stretch (Dimitriou and Edin 2008a,b). The joint acceleration data reported here still presented some overlap in distribution with unperturbed joint acceleration, although combining velocity and acceleration signals may constitute sufficient information for the CNS to trigger a response, in particular during posture. Despite these technical difficulties, the present study highlights that rapid feedback responses likely exist across all perturbation loads. Specifically, we found that the scaling between perturbation magnitude and EMG responses scales down to 0 and, importantly, passes through the origin.

Although the influence of perturbation magnitude on short- and long-latency responses has been addressed in the past (Bedingham and Tatton 1979; Gottlieb and Agarwal 1980; Jaeger et al. 1982; Lewis et al. 2005; Marsden et al. 1981), our study provides several important extensions on the properties of perturbation-evoked responses. It is the first to characterize that these responses are present down to levels approaching the natural variability of unperturbed motor behavior to the extent possible given physical constraints of the technology. This key contribution clearly supports the hypothesis that rapid feedback corrections reflect a general process, thereby potentially engaged during voluntary control as well as feedback responses to perturbations. The consistent onsets found across all load magnitudes are incompatible with the hypothesis that position and/or velocity error must be accumulated before launching a corrective motor response. In addition, the rapid feedback responses observed for shoulder muscles even though they were not stretched by the perturbation is incompatible with theories assuming only stretch-related corrections during voluntary control (Feldman 2008; Feldman and Levin 1995; Foisy and Feldman 2006). Although there exist activation thresholds

at the level of individual spindles and receptors, our data suggest that these physiological limits are below the scale of behavior-related changes in muscle state. Interestingly, subjects' informal reports suggest that perceptual thresholds are higher than those engaged in corrective motor behavior. These observations are compatible with previous characterization of perceptual thresholds in perspective with the known sensitivity of muscle spindles to small stretches (Hall and McCloskey 1983; Hasan 1983).

As well, the use of multijoint perturbations designed to generate motion at only one joint, the elbow, allows us to look at several different neural processes. Direct scaling of length changes to EMG could be observed for muscles spanning the elbow joint compatible with a close relationship between load magnitude and perturbation-evoked motion following the perturbation. In contrast, the multijoint loads generated only residual motion at the shoulder, which in the case of very small changes in load cannot be disambiguated from the natural variability. Even in this case, we observed stereotyped responses at the shoulder. This supports the hypothesis that the CNS is able to take the elbow motion into account to extract the underlying shoulder torque via an internal model and initiate a response with a latency of ~ 50 ms (Kurtzer et al. 2008, 2009). Similar results regarding multijoint integration were reported with different perturbation paradigms (Gielen et al. 1988; Soechting and Lacquaniti 1988). The invariant long-latency onset for shoulder responses found in the present study demonstrates that the use of internal models of limb dynamics is engaged at very small perturbation levels. Therefore, it is quite possible that corrections for variability during voluntary

control and corrections for external perturbations can be mediated using similar neural pathways.

A qualitative comparison between responses evoked by small or large perturbations (Figs. 2 and 8) suggests possible differences in the phasic nature of the response across different loads. It is known that multiple components and/or pathways are involved in corrective responses to mechanical perturbations (Kimura et al. 2006; Pruszynski et al. 2011a; Shemmell et al. 2009). It is therefore possible that changes in phasic profiles reflect greater or reduced contribution of some components across the different load magnitudes. The present data cannot differentiate whether one or all of these are engaged for very small perturbations.

A common technique to address voluntary muscle onset is to apply a light perturbation at one joint and ask participants to produce a response at the other joint as soon as they feel the perturbation (Hammond 1956; Pruszynski et al. 2008; Yang et al. 2011). The light perturbation was meant to produce a perceptually robust sensation without eliciting a strong response at the joint of interest. Although long-latency activity was reduced, the present results suggest that, even in this situation, feedback corrections are still engaged and could be extracted provided a sufficient number of trials are considered.

Establishing that multijoint corrective responses are engaged regardless of the perturbation magnitude renders these responses directly relevant to address feedback control of voluntary movements. Indeed, similar properties of voluntary movement control and long-latency responses likely originate from shared neural pathways. The primary motor cortex (M1) is known to be a key region contributing to voluntary movements and long-latency responses (Evarts and Tanji 1975; Matthews 1991; Porter and Lemon 1993; Scott 2004). In particular, it has been shown recently that corticospinal excitability correlates with interaction torques at the elbow during voluntary reaching movements (Gritsenko et al. 2011). As well, studies using single-unit recordings in the monkey and transcranial magnetic stimulation in humans have recently demonstrated that M1 resolves the ambiguous torque-motion relationship induced by multijoint perturbations (Pruszynski et al. 2011b). As we showed that such mechanisms correct for large and small perturbation indistinctly, we suggest that they also reflect direct model-based control engaged during voluntary movements.

Similar results have reported in the context of reaching movements toward visual targets that occasionally jumped during participants' initial saccade (Gritsenko et al. 2009). Gritsenko and colleagues (2009) emphasized stereotyped correction times regardless of the amplitude of the target displacement, even for very small jumps that subjects did not systematically perceive. This result clearly supports the idea that corrections for small and large motor errors engage the same visuomotor pathway, which is known to modulate visuomotor corrective responses according to whether they interfere with the task (Franklin and Wolpert 2008). The present study establishes a similar bridge between perturbation-evoked responses and compensation for sensorimotor variability during voluntary control.

A direct parallel can be drawn with recent theories of sensorimotor coordination, which postulate that the CNS continuously estimates the state of the body from motor prediction and sensory feedback to correct for deviations that interfere

with the task success at minimum cost (Liu and Todorov 2007; Scott 2004; Todorov and Jordan 2002; Valero-Cuevas et al. 2009). A verified prediction of the optimal feedback control theory is that feedback corrections for motor errors integrate internal models of arm and environment dynamics (Izawa et al. 2008). The present study demonstrates that multijoint responses comply with this prediction at perturbation levels encountered during voluntary control and further emphasize the relevance of studying rapid motor responses to probe the processes underlying sensorimotor coordination.

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DISCLOSURES

S. H. Scott is associated with BKIN Technologies, which commercializes the KINARM robot used in this study.

AUTHOR CONTRIBUTIONS

F.C., I.K., and S.H.S. conception and design of research; F.C. performed experiments; F.C. and I.K. analyzed data; F.C., I.K., and S.H.S. interpreted results of experiments; F.C. and S.H.S. prepared figures; F.C. and I.K. drafted manuscript; F.C., I.K., and S.H.S. edited and revised manuscript; F.C., I.K., and S.H.S. approved final version of manuscript.

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