Control of 3D Limb Dynamics in Unconstrained Overarm Throws of Different Speeds Performed by Skilled Baseball Players

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Hirashima M, Kudo K, Watarai K, Ohtsuki T. Control of 3D limb dynamics in unconstrained overarm throws of different speeds performed by skilled baseball players. J Neurophysiol 97: 680–691, 2007. First published November 1, 2006; doi:10.1152/jn.00348.2006. This study investigated how the human CNS organizes complex three-dimensional (3D) ball-throwing movements that require both speed and accuracy. Skilled baseball players threw a baseball to a target at three different speeds. Kinematic analysis revealed that the fingertip speed at ball release was mainly produced by trunk leftward rotation, shoulder internal rotation, elbow extension, and wrist flexion in all speed conditions. The study participants adjusted the angular velocities of these four motions to throw the balls at three different speeds. We also analyzed the dynamics of the 3D multijoint movements using a recently developed method called “nonorthogonal torque decomposition” that can clarify how angular acceleration about a joint coordinate axis (e.g., shoulder internal rotation) is generated by the muscle, gravity, and interaction torques. We found that the study participants utilized the interaction torque to generate larger angular velocities of the shoulder internal rotation, elbow extension, and wrist flexion. To increase the interaction torque acting at these joints, the ball throwers increased muscle torque at the shoulder and trunk but not at the elbow and wrist. These results indicate that skilled ball throwers adopted a hierarchical control in which the proximal muscle torques created a dynamic foundation for the entire limb motion and beneficial interaction torques for distal joint rotations.

INTRODUCTION

In multi-joint movements, a joint is rotated not only by the muscle and gravity torques but also by the interaction torque that arises from motions of the linked limb segments. A key question in the study of neural control of limb movement is how the CNS coordinates the large number of degrees of freedom (DOF) in the presence of the interaction torque. Early studies considered the interaction torque as the “load” that is “passively” produced by the intersegmental interaction and must be compensated for by the nervous system (Hollerbach and Flash 1982). Studies on patients with nervous system injury have shown that a failure to compensate for the interaction torque results in an abnormally curved hand path (Bastian et al. 1996; Sainburg et al. 1995). In addition, several studies found that the CNS generates a motor command that can compensate for the interaction torque in a feed-forward manner (Gribble and Ostry 1999; Latash et al. 1995; Sainburg et al. 1999).

Recently an increasing number of studies have focused on the CNS’s strategy of using or exploiting the interaction torque—a strategy proposed by Bernstein (1967) long ago. Several studies have found that humans often utilize the interaction torque to accelerate elbow joint rotation during reaching and drawing tasks (Doumskaia et al. 2002a,b; Galloway and Koshland 2002). Sainburg et al. found that the skilled dominant arm can utilize the interaction torque more effectively than the nondominant arm (Bagesteiro and Sainburg 2002; Sainburg 2002; Sainburg and Kalakanis 2000). From this perspective, the interaction torque is considered as the “actuator” that the nervous system “positively” produces. What is important for the utilization of the interaction torque is that it enables larger joint angular velocity than the muscle torque can produce on its own. Therefore we would expect this strategy to be prominently used during ball-throwing or racket-striking movements in which the largest possible speed at the hand or racket is required. Supporting data showing that the interaction torque was effectively utilized were presented in our previous study about the constrained throwing motion in two dimension (2D) using the upper extremity alone (Hirashima et al. 2003a).

However, all of the studies mentioned in the preceding text considered two-joint or at most three-joint movements constrained in a 2D plane. Therefore it is unclear how the CNS coordinates multiple joints during unconstrained overarm throws with three and more DOFs in three-dimensional (3D) space to achieve high ball speed with accuracy. Recently it has been debated whether skilled ball throwers utilize the interaction torque to accelerate the wrist flexion during unconstrained ball throwing in 3D space (Debicki et al. 2004; Gray et al. 2006; Hore et al. 2005a). This has not been demonstrated because there was no method that could determine the contributions of the muscle, gravity, and interaction torques to an angular acceleration about a joint coordinate axis (e.g., shoulder internal/external rotation axis, elbow extension/flexion axis, or wrist flexion/extension axis) in 3D movements. Although many studies have investigated the profiles of the muscle torque (resultant joint torque) during various 3D movements such as walking, running, kicking, reaching, throwing, and tennis stroke (Aleshinsky and Zatsiorsky 1978; Bahlmann and Knudson 2003; Elliott et al. 2003; Eng and Winter 1995; Fleisig et al. 1995, 1999; Kim and Eng 2004; Lacquaniti et al. 1986; Nonome et al. 2002; Shimansky et al. 2004), only a few studies have examined the interaction torque during 3D movements (Feltner and Dapena 1989; Hong et al. 2001; Kadota et al. 2004). These studies still have two problems with regard to examining the interaction torque’s contribution to a

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joint angular acceleration such as shoulder internal rotation. The first problem is that their analyses were designed to elucidate the effect of the interaction torque on the “segment rotation” with respect to the ground but not on the “joint rotation” with respect to the adjacent proximal segment. Examining the joint rotation rather than the segment rotation is necessary to assess the contribution of the muscle actions evoked by the CNS because muscles act directly on the angular acceleration of the attached joint (Sainburg et al. 1995; Springings et al. 1994). The second problem is that the researchers decomposed the torques such as muscle and interaction torques into three orthogonal joint coordinate axes. This method is certainly useful to aid understanding of the musculoskeletal requirement for achieving the movements or the mechanism of sport injuries (Fleisig et al. 1995, 1999). However, a torque about a joint coordinate axis generally produces the joint angular accelerations not only about the joint coordinate axis but also about the other two axes because the joint coordinate axes generally do not correspond to the principal axes of inertia for the entire kinematic chain distal to the joint in 3D movements (for more details, see Hirashima et al. 2007). Therefore the torques about a joint coordinate axis cannot fully explain the mechanical cause of the angular acceleration about the joint coordinate axis in 3D movements. To overcome these problems, we recently developed the new method called the “non-orthogonal torque decomposition method” (Hirashima et al. 2007). This method can calculate the muscle, gravity, and interaction torques that produce a joint rotation only about a joint coordinate axis. Thus the new method enables us to clarify how an angular acceleration about a joint coordinate axis (e.g., shoulder internal rotation) is generated by the muscle, gravity, and interaction torques.

The purpose of this study was to apply this method to unconstrained ball throwing by skilled throwers and to examine whether they utilize the interaction torque at each joint rotation. In this study, we focused on the dynamic analysis on shoulder internal rotation, elbow extension, and wrist flexion. This is because previous studies on the baseball pitching have demonstrated that large angular velocities were found at these three joint rotations at the ball-release time (Barrentine et al. 1998; Escamilla et al. 1998; Fleisig et al. 1999; Matsuo et al. 2001; Sakurai et al. 1993) and that they were the major contributors to the ball velocity at the ball-release time (Miyanishi et al. 1996). Specifically we tested the hypothesis that skilled individuals coordinate a large number of DOF so that they can utilize the interaction torques to generate large angular velocities at the three critical joint rotations.

Our second objective was to clarify how skilled ball throwers adjust the movement speed in ball-throwing tasks. A debate exists about whether the motions of different speeds are planned based on the speed-invariant strategy in which joint rotations of different speeds have the same joint angles and amplitudes but are scaled in time (Hollerbach and Flash 1982). Although speed-invariant kinematics has been observed in relatively slow two-joint planar movements (Atkeson and Hollerbach 1985; Lacquaniti and Soechting 1982), it has not been found pointing with the trunk (Pigeon et al. 2003; Pozzo et al. 2002; Thomas et al. 2003) and skilled fast ball-throwing movements (Hore et al. 2005b). Hore et al. (2005b) concluded that as the movement task progressively changed from a relatively slow two-joint planar task to a very fast 3D overarm throw, differences in the underlying joint rotations in fast and slow movements became increasingly apparent. Considering that the interaction torque is extremely influential in fast movements, these results imply that the CNS adopts the strategy in which the interaction torque is employed rather than the speed-invariant kinematic strategy. Therefore we examined the manner in which the CNS utilizes the interaction torque in planning the movements of different speeds by instructing participants to throw balls at three different speeds. We hypothesized that skilled ball throwers would increase the joint angular velocities with the gain in ball speed by increasing the interaction torques.

We also examined the manner in which the participants increased the interaction torques. The interaction torque at a certain joint eventually originates from the muscle torque at other joints. Therefore if the study participants increase the muscle torques at some leading joints, no need exists to increase the muscle torques at other joints. We predicted that the muscle torque at the trunk or shoulder would increase with ball speed but that this would not occur at the elbow or wrist. We discuss the merit of this strategy for generating a high speed at the endpoint with accuracy, considering the nature of the signal-dependent noise in human muscles. Finally, we discuss the critical problem in neural control of multi-joint movements in 3D space.

**METHODS**

**Study participants**

Nine right-handed male volunteers participated in this study. Eight were varsity baseball players (mean age: 20.4 yr) and one had been a professional baseball player (42 yr). Two of them were pitchers. In accordance with the Declaration of Helsinki, the participants were clearly explained the experimental procedure and submitted written informed consent. The experimental procedure was approved by the Ethical Committee of the Graduate School of Arts and Sciences of the University of Tokyo. Two participants were excluded from the analysis because their maximum ball speed did not exceed 25 m/s (90 km/h).

**Experimental setup and tasks**

We instructed the study participants to throw a baseball (136 g and 72 mm diam) in the natural 3D pitching motion with no constraint. They were instructed to throw a straight ball with the right (dominant) hand aiming at a target. The horizontal distance between the target and the marker line at which the participants readied themselves was 9.2 m. The center of the target was 0.9 m above the floor. The target board was a 1 × 1-m square.

Throws were made under three different speed conditions: slow–accurate, medium–accurate, and fast–accurate. After warm-up, the participants practiced the three speed conditions. First, they were instructed to throw a ball as fast as possible without largely changing the trajectory of the ball after the ball is released from the finger. This speed was set as the slow condition. The intermediate speed was set as the medium condition. After each practice trial, the participants were informed of the ball speed detected by the radar gun so that they could learn the three speed conditions. After the practice trials, they conducted recording trials. Each of the three conditions was randomly presented once to form one bout. Three bouts were repeated. Thus under each condition, the participants made three throws, and each individual made a total of nine throws. After each trial, they were informed of the
ball speed and the distance between the center of the target and the hitting position. The trial was repeated when the ball did not hit the target board.

Recording the movements

Eleven spherical reflective markers (1.5 cm diam) were used to identify anatomical landmarks for the purpose of digitization (Fig. 1). The motion of the reflective markers was recorded at 200 Hz by six high-speed video cameras (HAS-200R, Ditect). The 3D time-position data of each marker were obtained by means of the direct linear transformation method (Abdel-Aziz and Karara 1971). The data were smoothed by applying the bidirectional fourth-order Butterworth low-pass filter. The cutoff frequency was calculated by residual analysis (Winter 1990). The motion data were resampled at 1,000 Hz by spline interpolation.

We directly measured the total body weight and the lengths of upper arm, forearm, and hand of each participant. Using the measured anthropometric data and body segment parameters for Japanese reported by Ae et al. (1990), we conducted the following kinematic and dynamic analyses.

Kinematic analysis

ANGULAR VELOCITY. We analyzed the throwing movements with a four-segment model (trunk, upper arm, forearm, and hand plus ball). A rotatory motion of a segment can be described in terms of “segment rotation” relative to the ground or “joint rotation” relative to the adjacent proximal segment. In this paper, we describe the trunk rotation in terms of the “segment rotation” and the rotation of the other distal segments in terms of the “joint rotation” (i.e., shoulder, elbow, and wrist). We calculated the segment angular velocity vector of the trunk ($\omega_0$) and the joint angular velocity vectors of the shoulder ($\theta_1$), elbow ($\theta_2$), and wrist ($\theta_3$). To investigate the anatomical angular velocity (e.g., shoulder internal rotation), we divided them as follows

\[
\begin{align*}
\dot{\theta}_1 &= \dot{\theta}_{1-IE} + \dot{\theta}_{1-ED} + \dot{\theta}_{1-k} \\
\dot{\theta}_2 &= \dot{\theta}_{2-PS} + \dot{\theta}_{2-FE} + \dot{\theta}_{2-UR} \\
\dot{\theta}_3 &= \dot{\theta}_{3-UR} + \dot{\theta}_{3-ED} + \dot{\theta}_{3-IE} \\
\end{align*}
\]

Trunk angular velocity ($\omega_0$) consists of the anteroposterior lean ($\omega_{0-AP}$), mediolateral tilt ($\omega_{0-ML}$), and twist ($\omega_{0-TW}$). Shoulder joint angular velocity ($\dot{\theta}_1$) consists of the internal/external rotation ($\dot{\theta}_{1-IE}$), elevation/depression ($\dot{\theta}_{1-ED}$), and third axis rotation ($\dot{\theta}_{1-k}$). Elbow joint angular velocity ($\dot{\theta}_2$) consists of the extension/flexion ($\dot{\theta}_{2-FE}$) and pronation/supination ($\dot{\theta}_{2-PS}$). Wrist joint angular velocity ($\dot{\theta}_3$) consists of the flexion/extension ($\dot{\theta}_{3-IE}$) and ulnar/radial deviation ($\dot{\theta}_{3-UR}$).

CONTRIBUTION OF EACH ANGULAR VELOCITY TO THE TRANSLATIONAL VELOCITY OF THE FINGERTIP. To determine the contribution of each angular velocity to the translational velocity of the fingertip ($v_F$) at the time of ball release, we expressed $v_F$ as the sum of the contributions from the angular velocity vectors using a previously published method (Feltner and Nelson 1996; Miyaniishi et al. 1996; Spriggins et al. 1994)

\[
|v_F| = v_C \cdot u_i + (\omega_{0-AP} \times r_i) \cdot u_i + (\omega_{0-ML} \times r_i) \cdot u_i + (\omega_{0-TW} \times r_i) \cdot u_i + (\dot{\theta}_{1-IE} \times r_i) \cdot u_i + (\dot{\theta}_{1-ED} \times r_i) \cdot u_i + (\dot{\theta}_{1-k} \times r_i) \cdot u_i + (\dot{\theta}_{2-PS} \times r_i) \cdot u_i + (\dot{\theta}_{2-FE} \times r_i) \cdot u_i + (\dot{\theta}_{3-UR} \times r_i) \cdot u_i + (\dot{\theta}_{3-ED} \times r_i) \cdot u_i + (\dot{\theta}_{3-IE} \times r_i) \cdot u_i
\]

where $v_F$ is the translational velocity vector of the point C (see Fig. 1), $u_i$ is the unit vector of the $v_F$, $r_i$ is the vector from the point C to the fingertip, and $r_i$ is the vector from the joint i to the fingertip. Each bar in Fig. 2 shows the translational velocity of the fingertip produced by each component. For example, under the fast condition, the shoulder internal rotation (II) produced $\sim 10$ m/s of the total fingertip speed of 26.2 m/s.

Intersegmental dynamic analysis

We employed inverse dynamics and calculated the time series of the net torque vector (NET), gravity torque vector (GRA), muscle torque vector (MUS), and interaction torque vector (INT) for the shoulder, elbow, and wrist joints. Complete mathematical equations were presented in our previous paper (Hirashima et al. 2007). In-depth explanations about these four torques have been presented in several reports (Bastian et al. 1996; Cooper et al. 2000; Ghez et al. 1996; chapt. 5 in Zatsiorsky 2002). The NET is the torque required to generate the observed joint angular acceleration and is the sum of other components: $NET = MUS + INT + GRA$. The MUS includes not only the active muscle torque but also the passive torque from

\[\text{The subscript number (i) is used in such a manner that a segment and its proximal joint have the same number (i = \theta): trunk: 1; upper arm or shoulder: 2; forearm or elbow: 3; hand or wrist.}\]

\[\text{The ball speed is orthogonal to both the internal/external axis and elevation/depression axis. Because the rotation about the third axis generally does not have anatomical meaning, we call this the third axis.}\]
viscoelastic elements such as ligaments, articular capsules and other connective tissues. The INT is the torque that is dependent on the angular velocities and accelerations of all joints and segments except for the joint angular acceleration of the focused joint rotation. The GRA is the term describing the gravitational acceleration.

Nonorthogonal torque decomposition

Our point was not to merely calculate the four types of torque vectors (NET, MUS, INT, GRA) but to understand how each anatomical rotation is produced by MUS, INT, and GRA. The difficult point in relating the torque and 3D joint rotation is that a joint rotation about a joint coordinate axis (e.g., shoulder internal/external rotation axis) cannot be fully explained by the torque about the joint coordinate axis alone; the joint coordinate axes do not correspond to the principal axes of inertia for the entire kinematic chain distal to the focused joint (Hirashima et al. 2007). Therefore the previous 3D kinetic analysis that decomposed the torque into three orthogonal joint coordinate axes is not helpful to reveal the mechanical cause of a joint rotation about each joint coordinate axis, although the orthogonal decomposition can explain the musculoskeletal requirement for rotation about each joint coordinate axis, (Hirashima et al. 2007). Therefore the previous 3D axis) cannot be fully explained by the torque about the joint coordinate axis. To this end, we first specified the axis of the joint rotational motion (e.g., shoulder internal/external rotation axis), it is necessary to calculate the MUS, INT, and GRA vectors that affect the joint rotation only about one joint coordinate axis. We call the specified axis the “effective axis.” For example, we calculated the angular acceleration of shoulder internal rotation at time t (\(\dot{\theta}^{\text{INT}}_{\text{1-IE}}(t)\)) caused by the eINT_{1-IE}(t) as follows

\[
\dot{\theta}^{\text{INT}}_{\text{1-IE}}(t) = \frac{\text{eINT}_{\text{1-IE}}(t)}{\text{eINT}_{\text{1-IE}}^{-1}}(t)\dot{t}(t)
\]

The integral of the \(\dot{\theta}^{\text{INT}}_{\text{1-IE}}(T_R)\) during a certain time period indicates the amount of change in the angular velocity produced by the INT during the time period. Therefore the integral of the \(\dot{\theta}^{\text{INT}}_{\text{1-IE}}(t)\) from \(T_{vo}\) time when angular velocity is 0°/s) to \(T_R\) (ball-release time) calculated as follows

\[
\dot{\theta}^{\text{INT}}_{\text{1-IE}}(T_R) = \left[ \int_{T_{vo}}^{T_R} \dot{\theta}^{\text{INT}}_{\text{1-IE}}(t) dt \right]
\]

represents the contribution of the INT to the angular velocity of shoulder internal rotation at ball release. Similarly \(\dot{\theta}^{\text{MUS}}_{\text{1-IE}}(T_R)\) and \(\dot{\theta}^{\text{GRA}}_{\text{1-IE}}(T_R)\), calculated as follows

\[
\dot{\theta}^{\text{MUS}}_{\text{1-IE}}(T_R) = \left[ \int_{T_{vo}}^{T_R} \dot{\theta}^{\text{MUS}}_{\text{1-IE}}(t) dt \right]
\]

\[
\dot{\theta}^{\text{GRA}}_{\text{1-IE}}(T_R) = \left[ \int_{T_{vo}}^{T_R} \dot{\theta}^{\text{GRA}}_{\text{1-IE}}(t) dt \right]
\]

represent the contributions of the MUS and GRA to the angular velocity of shoulder internal rotation at ball release.
relationship $\dot{\theta}_{1-IE}(T_R) = \theta_{MUS}(T_R) + \dot{\theta}_{INT}(T_R) + \dot{\theta}_{GRA}(T_R)$ exists, we can determine the contribution of the muscle, gravity, and interaction torques to the angular velocity of shoulder internal rotation at ball release $\dot{\theta}_{1-IE}(T_R)$ by examining $\theta_{MUS}(T_R)$, $\dot{\theta}_{INT}(T_R)$, and $\dot{\theta}_{GRA}(T_R)$. We used the same procedure for elbow extension/flexion and wrist flexion/extension.

**Statistics**

Repeated-measures one-way ANOVAs were performed to assess the effect of speed on $\theta_{MUS}(T_R)$, $\dot{\theta}_{INT}(T_R)$, and $\theta_{1-IE}(T_R)$ ($P < 0.05$). Tukey’s post hoc multiple comparison test was used to determine the increase or decrease in the values of these variables between speeds ($P < 0.05$).

**RESULTS**

**Ball speed and fingertip speed**

First we determined whether the study participants could throw the balls at three different speeds. Repeated-measures ANOVA revealed a significant main effect of the speed condition on the ball speed $[F(2,12) = 50.4]$. Tukey’s post hoc multiple comparison test indicated significant differences for all comparisons among slow (21.1 $\pm$ 2.24 m/s), medium (24.1 $\pm$ 1.94 m/s), and fast (28.1 $\pm$ 1.76 m/s) speeds. The fingertip speed at the ball-release time was also investigated. Repeated-measures ANOVA revealed a significant main effect of the speed condition on the fingertip speed $[F(2,12) = 68.4]$. Tukey’s post hoc multiple comparison test indicated significant differences for all comparisons among slow (20.3 $\pm$ 2.10 m/s), medium (23.3 $\pm$ 1.72 m/s), and fast (26.2 $\pm$ 1.28 m/s) speeds. Hereafter, we consider the speed adjustment of the ball as the speed adjustment of the fingertip.

**Contributions of each motion to the fingertip speed**

Each bar in Fig. 2 shows the translational velocities of the fingertip produced by each motion, which were calculated by Eq. 1. A repeated-measures ANOVA was performed separately for each motion. Significant main effects of the speed condition were observed in the translational velocity of the trunk (C), twist of the trunk (TW), shoulder internal rotation (IE), and wrist flexion (FE), and a nearly significant effect was observed in the elbow extension (EF, $P = 0.060$). Hereafter, we primarily examine the shoulder internal rotation, elbow extension, and wrist flexion.

**Shoulder internal rotation**

**KINEMATICS.** Figure 3, A and B (thick line), shows the angular velocity and angular acceleration of the shoulder internal rotation during the three speed conditions (slow, medium, and fast) of a single representative trial. The ball-release time was set to 0 ms. Initially, for all the conditions, the shoulder rotated externally ($\sim T_{vo}$) then rotated internally as the time of ball release approached ($T_{vo} - T_R$). The internal angular velocity at the time of the ball release increased with the ball speed (Fig. 3A). The internal angular velocity at ball release averaged across all participants is shown in Fig. 7C, far right. A repeated-measures ANOVA revealed that it significantly increased with the ball speed $[F(2,12) = 26.8]$.

**DYNAMICS.** Figure 3C shows how the eNETIE (thick line), which is required to generate the angular acceleration of the shoulder internal rotation, is generated by the eMUSIE (thin line) and eINTIE (dotted line). The eGRAIE is not shown because it is very small relative to the other torques. Therefore eNETIE can be practically expressed as the sum of eMUSIE and eINTIE. Before $T_{vo}$, a phase (backward phase) occurred in which the shoulder was externally rotated (Fig. 3A) but was internally accelerated (Fig. 3B, thick line). The internal eNETIE in this phase was produced by the internal eMUSIE that overcame the external eINTIE (Fig. 3C). After $T_{vo}$ (forward phase), eINTIE suddenly turned to the internal direction and produced the internal eNETIE in cooperation with eMUSIE. The magnitude of the internal eMUSIE and internal eINTIE increased with the ball speed. As a result, the magnitude of internal eNETIE also increased with the ball speed.

Figure 3B shows the three angular accelerations $\ddot{\theta}_{MUS}(\text{thick line})$, $\ddot{\theta}_{INT}(\text{dotted line})$, and $\ddot{\theta}_{GRA}(\text{thick line})$ caused by the three effective torques eMUSIE, eINTIE, and eNETIE, respectively. They show a similar tendency to the corresponding effective torques. To understand the mechanical cause of the angular velocity at the time of ball release, we integrated the

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3 $i-r = 1-IE$: shoulder internal/external rotation, 2-EF: elbow extension/flexion, 3-FE: wrist flexion/extension.
The magnitudes of the 3 shoulder muscle torques orthogonally decomposed into 3 joint coordinate axes. —, average of the fast 3 trials of the same subject as shown in Fig. 3. · · · , average ±1SD of the 3 trials.

MAGNITUDE OF TORQUE. Here it should be noted that the magnitude of the torque decomposed into nonorthogonal effective axes cannot be compared with the magnitude of the torque decomposed into orthogonal joint coordinate axes. To compare the magnitude of shoulder muscle torque observed in the present study to that reported by Fleisig et al. (1995), we also showed the shoulder muscle torque decomposed into orthogonal joint coordinate axes (MUSIE, MUSDE, MUSk in Fig. 4). They overall showed similar peak values and time-series profiles as those of professional baseball players (see Fig. 3 in Fleisig et al. 1995). Figure 4 also indicates that trial-to-trial variability of throwing movements was very small.

Elbow extension

KINEMATICS. Figure 5, A and B (thick line), shows the angular velocity and angular acceleration of the elbow extension under three speed conditions of the same trial, as shown in Fig. 3. Initially, the elbow was flexed (−T₀) and extended as the ball-release time approached (T₀ −TR). The elbow extension angular velocity at the time of the ball release increased with the ball speed (Fig. 5A). The elbow angular velocity at the time of the ball release averaged across all participants is shown in the rightmost column of Fig. 7C. Repeated-measures ANOVAs were performed to assess the effect of speed on the two variables. The results revealed significant main effects of speed on ˙θ₁IEF (TR) [F(2,12) = 15.4] and ˙θ₂IEF(T₀) [F(2,12) = 15.3]. The results of Tukey’s post hoc multiple comparison tests are shown in Fig. 7C. These results indicated that skilled throwers increased the shoulder internal angular velocity at ball release by increasing both eMUSIE and eINTIE.

Wrist flexion

KINEMATICS. Figure 6, A and B (thick line), shows the angular velocity and angular acceleration of the wrist flexion during three speed conditions of the same trial as shown in Fig. 3. The wrist began to flex at ~50 ms before the time of ball release. The flexion angular velocity at the time of ball release increased with the ball speed (Fig. 5A). The elbow angular velocity at the time of the ball release averaged across all participants is shown in the rightmost column of Fig. 7C. Repeated-measures ANOVA revealed that it increased significantly with the ball speed [F(2,12) = 16.3].

DYNAMICS. Figure 5C shows how the eNETEF (thick line) is generated by eMUSEF (thin line) and eINTEF (dotted line). Overall, the profile of eNETEF was very similar to that of eINTEF, which indicated that the elbow extension angular acceleration was primarily produced by the eINTEF. The elbow extension eINTEF increased with the ball speed; hence, the elbow extension eNETEF increased with the ball speed.

To assess these observations quantitatively, we calculated the elbow extension angular velocity at the ball-release time (θ₂IEF(T₀) and θ₂IEF(T₁)) produced by eMUSEF and eINTEF (Fig. 7B). Repeated-measures ANOVAs were performed to assess the effect of speed on the two variables, revealing that θ₂IEF(T₁) [F(2,12) = 21.8] significantly increased with the ball speed, whereas θ₂IEF(T₀) [F(2,12) = 15.3] decreased significantly. The results of Tukey’s post hoc multiple comparison tests are shown in Fig. 7B. These results indicated that skilled throwers increased the elbow extension angular velocity at the time of ball release by increasing eINTEF alone.
creased with the ball speed (Fig. 6A). The flexion angular velocity at the time of ball release averaged across all participants is shown in the rightmost column of Fig. 7A. A repeated-measures ANOVA revealed that it significantly increased with the ball speed \( F(2,12) = 10.4 \).

DYNAMICS. Figure 6C shows how the eNET\(_{FE}\) (thick line) is generated by eMUS\(_{FE}\) (thin line) and eINT\(_{FE}\) (dotted line). Before \( T_0 \), a phase (backward phase) occurred in which the wrist was extended (Fig. 6A) but was accelerated in flexion (Fig. 6B). The flexion eNET\(_{FE}\) in this phase was produced by the flexion eMUS\(_{FE}\) (Fig. 6C). After \( T_0 \) (forward phase), eINT\(_{FE}\) acted in the flexion direction, whereas eMUS\(_{FE}\) acted in the extension direction. The flexion eINT\(_{FE}\) increased with the ball speed. As a result, the magnitude of the flexion eNET\(_{FE}\) increased with the ball speed.

To assess these observations quantitatively, we calculated the wrist flexion angular velocity at the ball-release time (\( \theta_{A-INT}^{\text{MUS}}(T_R) \) and \( \theta_{A-INT}^{\text{MUS}}(T_R) \)) produced by eMUS\(_{FE}\) and eINT\(_{FE}\) (Fig. 7A). Repeated-measures ANOVAs were performed to assess the effect of speed on the two variables, revealing that \( \theta_{A-INT}^{\text{MUS}}(T_R) \) \( F(2,12) = 7.69 \) significantly increased with the ball speed, whereas \( \theta_{A-INT}^{\text{MUS}}(T_R) \) \( F(2,12) = 3.77, P = 0.053 \) decreased almost significantly. The results of Tukey’s post hoc multiple comparison tests are shown in Fig. 7A. These results indicated that skilled throwers increased the wrist flexion angular velocity at the time of ball release by increasing eINT\(_{FE}\) alone.

**DISCUSSION**

This study investigates the manner in which skilled throwers generate high ball speed by coordinating a large number of DOFs in 3D space. Kinematic analysis revealed that, as observed by the previous studies (Miyanishi et al. 1996), the fingertip velocity at the ball-release time was mainly produced by the four DOFs (i.e., trunk leftward rotation, shoulder internal rotation, elbow extension, and wrist flexion), although much more DOFs are involved in the throwing motion. In addition, we found that the throwers controlled the ball velocity by adjusting the angular velocities of these four DOFs. The major purpose of this study was to investigate how skilled throwers generate large angular velocities at these critical DOFs by coordinating the muscle and interaction torques. Because using the interaction torque is a beneficial strategy for ball-throwing movements in which the largest possible speed at the hand is required, it can be hypothesized that skilled throwers adopt this strategy at the critical DOFs. We conducted the dynamic analysis on the shoulder internal rotation, elbow extension, and wrist flexion and found that skilled throwers utilized the interaction torques to generate large angular velocities for all the three joint rotations.

Utilization of the interaction torque

It has been debated whether skilled individuals utilize the interaction torque to accelerate the wrist flexion in ball throw-
ing movements (Debicki et al. 2004; Gray et al. 2006; Hirashima et al. 2003a,b; Hore et al. 2005a). Our previous study on ball-throwing movements constrained in 2D with the upper extremity alone found that the study participants could not utilize the interaction torque to accelerate wrist flexion (Hirashima et al. 2003a). Our simulation study also showed that this was not due to a lack of participants' skill but rather to the mechanical properties of the wrist joint; the interaction torque for wrist flexion was never attained even when we activated the elbow and wrist muscles with various activation patterns (Hirashima et al. 2003b). Thus using the wrist was disadvantageous in obtaining large angular velocity, whereas this wrist property is advantageous for keeping the wrist joint stable and would lead to avoiding excessive wrist extension or flexion and to simplification of extrinsic finger control (Hirashima et al. 2003b; Koshland et al. 2000; Werremeyer and Cole 1997).

However, we found that in whole-body 3D throwing, skilled ball throwers utilized the beneficial interaction torque for wrist flexion (Fig. 6). This result suggests that skilled ball throwers change the dynamic context of the upper extremity motion by making use of the translational and rotational motion of the trunk. This idea is compatible with Debicki et al.'s (2004) idea that large forward translational deceleration of the forearm, which could accelerate the wrist flexion, occurred only in unconstrained 3D throws made by skilled participants.

The interaction torques at the elbow and shoulder also play a more important role in whole-body 3D throwing than in arm-only 2D throwing. Specifically, although the elbow extension was accelerated by the muscle torque more than by the interaction torque in arm-only 2D throwing (Hirashima et al. 2003a), the elbow was mostly accelerated by the interaction torque in whole-body 3D throwing. Although the shoulder was decelerated by the interaction torque in arm-only 2D throwing (Hirashima et al. 2003a), the shoulder internal rotation was accelerated by the interaction torque in whole-body 3D throwing.

The difference between the constrained 2D throwing and unconstrained 3D throwing underscores the importance of studying unconstrained 3D movements to reveal the control mechanism of highly skillful movements because it is likely that task constraints preclude skillful inter-joint coordination.

Adjustment of movement speed

Studies on speed adjustment in ball throwing have implications concerning the motor planning process by the CNS. Our previous study on constrained 2D throwing showed that the participants in that study imposed the speed adjustment on the elbow joint but not on the wrist joint because the beneficial interaction torque is available only at the elbow (Debicki et al. 2004; Hirashima et al. 2003a). Therefore we hypothesized that the CNS imposed the speed adjustment on the joints where the beneficial interaction torque can be obtained. The present study supports this idea by showing that skilled ball throwers imposed the speed adjustment on the wrist joint where the beneficial interaction torque was available in whole-body 3D throwing. This idea is also applicable to elbow extension and shoulder internal rotation.

Several studies have examined the speed adjustment strategy to investigate the motor planning process. Hollerbach and Flash (1982) proposed a speed-invariant strategy in which joint rotations of different speeds have the same joint angles and amplitudes but are scaled in time. This strategy simplifies the multi-joint control problem because the successful motor command for one speed can be used for another speed by time and amplitude scaling of the motor command. This results in the reduction of the cost for nonlinear inverse dynamics computation. The speed-invariant strategy has been found in relatively slow two-joint reaching movements (Atkeson and Hollerbach 1985; Lacquaniti and Soechting 1982) and sequential movements of single-joint movement (Carter and Shapiro 1984). However, the speed-invariant strategy has not been found when the movement task became more complex and faster, such as 3D reaching, reaching with trunk motion and overarm throws (Adamovich et al. 1999; Hore et al. 2005b; Klein Breteler 1998; Pigeon et al. 2003; Pozzo et al. 2002; Thomas et al. 2003). Hore et al. (2005b) predicted that skilled throwers use different types of inter-joint coordination for different speeds so that they can utilize the interaction torque more than when they adopt the speed-invariant strategy. This idea is strengthened by the present result that skilled ball throwers increased ball speed by increasing the beneficial interaction torques.

Hierarchical control

One more important question is how skilled ball throwers increase the interaction torque. The interaction torque at a certain joint eventually originates from the muscle torque at other joints. Therefore the increase in the interaction torque is caused by the increase in the muscle torque at other joints. The muscle torques at the shoulder, trunk, and lower extremity may be responsible for this role. In fact, the shoulder is the only arm joint in which the muscle torque increased with the ball speed. Because the translational motion and twist rotation of the trunk increased with ball speed (Figs. 2 and 8), we can conclude that the trunk and lower-extremity muscle torques also increased with ball speed. However, the interaction torque alone contributed to the adjustment in the angular velocity at the elbow and wrist joints. These results indicate that the increase in the muscle torque at the shoulder, trunk, and lower extremity not only increase their own angular velocities but also increase the elbow and wrist joint angular velocities through the interaction torque.

**Fig. 8.** Angular velocities of the trunk twist rotation (\(\omega_{TW}\)) under the slow (left), medium (middle), and fast conditions (right) of the same subject as shown in Fig. 3. —, average of 3 trials in each speed condition. •••, average ± 1 SD of the 3 trials.
This result is consistent with the recent findings about various 2D multi-joint movements such as reaching, drawing, and cyclical movements (Buchanan 2004; Dounskaia et al. 1998, 2002a,b; Galloway and Koslhand 2002; Ketcham et al. 2004; Levin et al. 2001). Dounskaia et al. (2000, 2005) summarized these results and proposed that the CNS organizes multi-joint movements by using hierarchical control as follows. There is one (leading) joint that creates a dynamic foundation for the entire limb motion. Acceleration at the leading joint is mainly produced by the muscle torque at the joint. Leading joint motion generates a powerful interaction torque at the other (subordinate) joints. The muscle torque at the subordinate joint regulates the interaction torque to fulfill the task demand. This idea is compatible with Bernstein's (1967) suggestion that the role of muscle activity is not only to accelerate the limb but also to control the intersegmental interaction. Dounskaia (2005) also proposed that the proximal joint tends to be a leading joint because proximal segments have higher inertia and more massive musculature than distal segments (see also Putnam 1991, 1993).

The present study extends these ideas by adding three new findings. First, more than two leading joints can occur in a complex 3D movement, such as ball-throwing. The EMG record (Hirashima et al. 2002) and trunk angular velocity profile (Fig. 8) suggest that the serratus anterior and external oblique muscles at the right side accelerate the scapula or upper trunk to rotate leftward from about –150 to –50 ms, indicating that the upper trunk has the leading role. After about –70 ms, the shoulder acted as a leading joint because the shoulder internal rotation was accelerated by the muscle torque (Fig. 3C). Thus the leading joint can switch from one joint to another with a certain overlap duration.

Second, a joint can have both leading and subordinate roles. The shoulder joint initially acted as a leading joint during the backward phase (Fig. 3, B and C) because the muscle torque mainly produced the acceleration for the shoulder internal rotation. On the other hand, it served as a subordinate joint during the forward phase because the interaction torque substantially contributed to internal acceleration.

The third finding is related to the sequence in which joints dynamically influence each other. Although previous studies have predominantly reported proximal-to-distal dynamic effects, our data provide an example of the opposite sequence of dynamic influence. Specifically, the interaction torque contributing to the shoulder internal rotation (about –30 ms in Fig. 3C) occurred later than that contributing to the elbow extension (about –100 ms in Fig. 5C). In addition to the evidence in dynamics, the distal-to-proximal sequence was also observed in joint kinematics. The shoulder internal rotation (about –30 ms in Fig. 3A) occurred later than the elbow extension (about –60 ms in Fig. 5A) (see also Feltner and Dapena 1986; Marshall and Elliott 2000). The occurrence of the distal-to-proximal sequence suggests the possibility that the distal joint rotation may sometimes have a strong backward influence on the proximal joint rotation. We predict that the strong effect from the distal to proximal joints can occur only at the longitudinal rotation during 3D movements. Because the moment of inertia around the longitudinal axis of the upper arm gets smaller as the elbow is extended from a right angle to full extension, we can interpret the distal-to-proximal phenomena as the sequence from a joint rotation with a large moment of inertia to a joint rotation with a small moment of inertia. Indeed it was when the elbow angle was extended from 110 to 150° (full extension equals 180°) that the interaction torque assisted the shoulder internal rotation. It thus follows that skilled throwers obtained the beneficial interaction torque at the shoulder internal rotation by appropriately configuring the upper limb posture.

In 3D movements with many joints, it is quite difficult to estimate the direct source of the interaction torque because the interaction torque at one joint rotation is influenced by joint angular velocities and accelerations of many other joint rotations and the 3D configuration of the segments. In future studies, it will be necessary to examine each mathematical term of the interaction torque to determine the direct sources of the interaction torques of shoulder internal rotation, elbow extension, and wrist flexion. At the present time, we tentatively conclude that the organization principle of skilled multijoint movements is the hierarchical control in which muscle torques at joints with relatively high moments of inertia create a dynamic foundation of the entire limb motion and generate a beneficial interaction torque at joints with relatively low moments of inertia. In the future, it will be interesting to examine whether unskilled ball throwers or cerebellar patients can use this hierarchical control.

Relationship between hierarchical control and movement accuracy

A ball-throwing movement requires not only high ball speed but also accuracy. The accuracy can be improved by reducing the variability of the motor command (Harris and Wolpert 1998) or by compensatory coordination of movement parameters (Kudo et al. 2000; McDonald et al. 1989; Scholz et al. 2000). Here we explain how hierarchical control contributes to reducing the variability of motor command.

Recently, studies have investigated the nature of the signal-dependent noise in motor command and its effect on the neural control of movement kinematics and muscle activation patterns (Hamilton et al. 2004; Harris and Wolpert 1998; Haruno and Wolpert 2005; Jones et al. 2002). Harris and Wolpert (1998) proposed that the shape of a trajectory of human movements is selected to minimize the variance of the final eye and arm positions. This minimum-variance model accurately predicts the trajectories of both saccades and arm movements. Hamilton et al. (2004) examined the signal-dependent noise of muscles with different strengths during an isometric force production task and found that the coefficient of variation of force decreases systematically as the muscle strength increases. This means that a given torque or force can be more accurately generated by a stronger muscle than a weaker muscle. Therefore the strategy of depending on the muscle torque only at the stronger proximal muscles to create a dynamic foundation for the entire limb motion and using the interaction torque at the distal joint is helpful to minimize the consequences of the signal-dependent noise under the condition that large motor output is required.

Neural control of 3D multijoint movements

To reveal important aspects of neural control of human movements, it is necessary to study complex 3D movements,
as suggested by Desmurget et al. (1997) and Cordo and Gurfinkel (2004). However, the study of 3D movements has been constrained by limited knowledge about the source of the complexity of 3D movements. Here we explain a critical difference between 2D and 3D movements in terms of neural control.

Let us consider the three-segment model with the upper arm, forearm, and hand. In the case of horizontal 2D movements, one principal axis of the inertia for the entire kinematic chain distal to the shoulder joint is always parallel to the shoulder joint axis irrespective of the elbow and wrist angles. Therefore if an ideal shoulder flexor muscle exists, it always functions to accelerate the shoulder into flexion irrespective of the elbow and wrist angles, although the magnitude of the acceleration is affected by the elbow and wrist angles.

In contrast, in the case of 3D movements, the principal axes of inertia for the entire kinematic chain distal to the shoulder joint generally deviate from the shoulder joint coordinate axes \( (\mathbf{i}_1, \mathbf{j}_1, \text{ and } \mathbf{k}_1) \) in Fig. 3B in Hirashima et al. (2007). Therefore the ideal shoulder flexor muscle described in the preceding text can produce angular acceleration not only about the flexion axis \( (\mathbf{k}_1) \) but also about the elevation axis \( (\mathbf{i}_1) \) and the internal rotation axis \( (\mathbf{j}_1) \). Furthermore the spatial relationship between the shoulder joint coordinate axes and principal axes of inertia of the entire kinematic chain changes depending on the elbow and wrist angles. Therefore a shoulder joint angular acceleration vector caused by the shoulder muscle changes depending on the elbow and wrist angles. This indicates that the CNS must determine the motor command for each muscle by considering the configuration of the entire kinematic chain. Thus the multijoint structure of the human body in 3D space requires the neural system to cope with not only the interaction torque but also the “variable function of the muscle.”

It should be noted that the variable function of the muscle stated here is simply derived from the mechanics of the 3D multijoint structure. Therefore this is different from postural dependence of muscle action that is derived from the 3D muscle anatomy reported by Buneo et al. (1997). They showed that the same muscle at the shoulder produces different directional torques depending on the shoulder angle relative to the trunk. This is derived from the anatomical phenomenon that the muscle geometry, including the line of muscle action, changes with the limb posture.

In summary, the effect of each muscle activation on the joint angular acceleration depends on two factors. First, the muscle torque vector produced by the muscle activation depends on the joint angle at which the muscle acts (anatomical factor). Second, the joint angular acceleration produced by the muscle torque depends on the joint angles at which the muscle does not act (mechanical factor). Therefore the CNS must consider the dual postural dependence of muscle function to determine the motor command for each muscle. It is totally unclear how the CNS copes with the dual complexity of the muscle function. One possible strategy that can simplify the neural control of rapid movements, such as reaching, pointing, and throwing, is minimizing the effect of the postural change during a motion; this strategy can reduce the change of muscle function. This would be one possible hypothesis for explaining why the final posture of 3D reaching movement depends on the starting location, which has been recently debated (Desmurget et al. 1998; Hermens and Gielen 2004; Soechting et al. 1995; Vetter et al. 2002). This strategy can not only reduce the burden for computing the motor command for each muscle in the motor planning process but also reduce the kinematic variability due to the execution noises (i.e., signal-dependent noise, constant noise, and temporal noise) (see van Beers et al. 2004).

The present study provides the framework to differentiate movement variables such as muscle activity, muscle torque, and angular acceleration. This framework is helpful to promote motor neuroscience at the neural level as well as the limb mechanics and behavior levels because neurophysiological studies have focused on what movement variables are encoded in the activity of primary motor cortex and premotor cortex neurons (Evarts 1968; Georgopoulos et al. 1982; Kakei et al. 1999, 2001; Scott and Kalaska 1997; Sergio and Kalaska 2003). Because the effect of muscle activation on the joint angular acceleration is fixed in 2D movements, it is impossible to dissociate the muscle activity from the joint rotation. In contrast, the two variables can be discriminated by examining the 3D movements with different postures because the effect of muscle activation on the joint angular acceleration changes depending on the posture in 3D movements. For example, let us consider the shoulder flexion movements with elbow flexed 0 or 90°. Different combinations of shoulder muscles activities are needed to produce the same shoulder flexion with different elbow angles. Thus elaborated experimental setup that can dissociate the movement variables by making use of the variable function of the muscle will be helpful to understand which movement variables are represented in individual neurons in motor areas.

**Conclusion**

We examined how skilled ball throwers coordinate a large number of DOFs in the presence of the interaction torque in 3D space. The study participants adopted the hierarchical control strategy in which muscle torques at the joints with relatively high moments of inertia create a dynamic foundation of the entire limb motion and generate a beneficial interaction torque at joints with relatively low moments of inertia. This hierarchical control would be helpful to minimize the consequences of signal-dependent noise under the condition that large motor output is required. We also provide a framework to study the neural control of 3D multijoint movements by demonstrating that the 3D multijoint structure requires the neural system to cope with not only the interaction torque but also the variable function of the muscle that depends on the configuration of the entire kinematic chain.

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**References**

Abdel-Aziz YI, Karara HM. Direct linear transformation from comparator coordinates into object space coordinates in close-range photogrammetry.


