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Kinematic and EMG characteristics of simple shoulder movements with proprioception and visual feedback

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Abstract

The objective of this study was to determine if simple, shoulder movements use the dual control hypothesis strategy, previously demonstrated with elbow movements, and to see if this strategy also applies in the absence of visual feedback. Twenty subjects were seated with their right arm abducted to 90° and externally rotated in the scapular plane. Subjects internally rotated to a target position using a custom shoulder wheel at three different speeds with and without visual feedback. Kinematics were collected with a motion analysis system and electromyographic (EMG) recordings of the pectoralis major (PECT), infraspinatus (INFRA), anterior and posterior (ADELT, PDELT) deltoid muscles were used to evaluate muscle activity patterns during movements. Kinematics changed as movement speed increased with less accuracy ($p < 0.01$). Greater EMG activity was observed in the PECT, PDELT, and INFRA with shorter durations for the ADELT, PDELT and INFRA. Movements with only kinesthetic feedback were less accurate ($p < 0.01$) and performed faster ($p < 0.01$) than movements with visual feedback. EMG activity suggests no major difference in CNS control strategies in movements with and without visual feedback. Greater resolution with visual feedback enables the implementation of a dual control strategy, allowing greater movement velocity while maintaining accuracy.

Keywords: Upper extremity; EMG; Kinematics; Kinesthesia

1. Introduction

Identifying central nervous system (CNS) control strategies of human movement through the study of movement kinematic, kinetics and muscle function through electromyography (EMG) is difficult due to the varied movements that humans perform. The understanding of CNS movement control strategies are developed from the study of relatively simple, uni-planar movements performed in a gravity eliminated position, most commonly simple elbow flexion and extension movements [7,19,20]. Gottlieb and Corcos [7,19,20] designed studies of elbow flexion and extension movements to reduce the number of factors that the CNS needs to control during the movement. However, CNS strategies developed from the study of elbow flexion and extension movements need to be applied to other joints and different types of movements in order to expand these theories. Elbow flexion and extension movements performed while studying movement control strategies were usually completed with visual feedback to ensure movement accuracy, however, it is not known

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if obscured visual feedback, forcing subjects to rely on kinesthetic feedback, could also alter CNS control of movements [7,19,20]. Understanding CNS control of movement strategies of *typical* human movements would then permit investigation of abnormal strategies encountered with injury or pathology.

1.1. Dual control hypothesis theory of motor control

Gottlieb and Corcos [7,19–21,8] formulated a dual control hypothesis for voluntary elbow movements that is comprised of two movement strategies: the speed "speed insensitive" (SI) and the "speed sensitive" (SS) strategies. The SI strategy is characterized by a relatively constant initial intensity of muscle amplitude, across multiple movements, while duration of muscle activity is modulated across movements [19,20]. The SS strategy is characterized by a relatively constant duration of muscle activity, across multiple movements, while amplitude of muscle activity is modulated [7]. They contend that it is the goal of the movement that dictates the strategy, which in turn drives the observable muscle activation patterns (EMG activity), joint torques, and movement kinematics [19]. The dual control hypothesis is an example of a cortical control strategy that results in CNS activation of the alpha motorneurons that is based on the patterns of EMG activity and the ensuing kinematics.

The SI strategy dictates that the initial pulse amplitude is selected considering the entire task, where initial alpha motorneuron activity is independent of changes that occur later in the movement task. The initial pulse amplitude is insensitive to changes in task variables, such as load and distance where changes in movement speed are modulated by the task [19,20]. The speed insensitive strategy is described as the default strategy and is dictated by the goal of the movement [20]. The term "insensitive" is used to indicate that in these movements, selected by the demands of a task, the characteristics of the movement remains invariant regardless of different movement speeds.

The SS strategy is the result of initial modulation of the initial amplitude of the alpha motorneuron pulses, where the duration of the pulse activity remains constant [7,20]. Similar pulse intensities can produce a variety of movement speeds relative to an external load; therefore, it can be learned quickly how to account for various inertial loads in order to achieve an optimum speed. The SS strategy requires the initial task duration to remain constant where an increased movement speed is the result of larger initial pulse from the alpha motorneuron driving muscle activity. Gottlieb and Corcos [7] use measures like burst of muscle activity, torque rise time and acceleration to identify these movement strategies. The SS strategy is initiated when an arbitrary speed makes it impossible to accomplish a task and is similar to other theories of pulse amplitude modulation [17]. The SS strategy is different than the SI strategy because SS movements are initiated where speed is the primary purpose of the task, while with the SI strategy, although speed is affected by the task, the initiation of the movement is not.

These studies were performed on the elbow, in a gravity-eliminated position and included visual feedback during the task. Kinematic and EMG patterns to identify both the SS and SI strategies with elbow movements have not been replicated in other joint movements. The dual control hypothesis of a SS and SI strategy could be further supported if observed at joints other than the elbow. This study attempted to examine simple shoulder movements in a manner similar to the elbow movements studied by Gottlieb and Corcos [7,19,20].

1.2. Proprioception

Proprioception can be described as afferent information arising from peripheral mechanoreceptors that contribute to postural control, joint stability and conscious sensation of movement [37]. Bastain used the term kinesthesia to describe conscious and unconscious aspects of both position and motion sense [2]. Conscious awareness is necessary for clinical measures of both joint position sense (JPS), the ability to identify a previously presented joint position, and motion sense, the ability to identify the onset of passive motion. Both JPS and motion sense are considered sub-modalities of conscious proprioception [33]. Unfortunately, terms such as proprioception and kinesthesia are used interchangeably in the literature with no differentiation between conscious and unconscious CNS mechanisms involved with this phenomenon during clinical testing of JPS or motion sense. This study used JPS to evaluate shoulder proprioception, which includes not only the conscious ability to recognize efferent position signals but also the efferent motor response necessary to move to the desired position. We are sensitive to the influence of unconscious motor commands on the conscious proprioceptive mechanisms (movement accuracy) associated with active movements during JPS testing, but make no attempt to differentiate between conscious and unconscious CNS mechanisms.

Visual and proprioceptive feedback are both used by the CNS in order to ensure accurate placement of the upper extremity. When vision is obscured, the CNS has to rely almost exclusively on proprioceptive feedback in an attempt to maintain accurate upper extremity placement and is believed to be of considerable functional significance [18,16]. The glenohumeral joint is at the base of the upper extremity, derives most of its stability from muscular activity, is very mobile and can function at high velocities. Generally, shoulder proprioception studies have been performed passively and at slow velocities, neglecting CNS control strategies for active movements in response to proprioception [1,5,28,39,43]. There has been little or no reported difference in the ability in shoulder JPS when performed either passively or actively [34,41]. However, multi-joint upper extremity active movements have demonstrated greater proprioceptive acuity compared to passive upper extremity movements [31]. In addition, there is scant evidence of how muscular activity, during active movements, can influence proprioceptive feedback to the CNS. Afferent feedback such as arm position and movement velocity in conjunction with visual feedback influence elbow movement accuracy [9,10]. There have been no studies to evaluate the possible CNS control mechanisms behind active movements that rely solely on proprioceptive feedback.

The ability of the CNS to modulate upper extremity movements under various types of peripheral feedback and across different movement speeds is not clearly understood. It is important to identify strategies that contribute to upper extremity movement control in order to identify CNS variables that could impact these movements. We used the dual control hypothesis of a SS and SI strategy to compare simple shoulder movements, at three different movement speeds, with visual feedback to movements that rely solely on kinesthetic feedback. The purpose of this study was twofold: (1) to determine if shoulder movements involve similar SI and SS strategies that have been demonstrated at the elbow and (2) to determine if these SI and SS strategies apply to these shoulder movements in the absence of visual feedback and how these strategies impacts movement accuracy. Evidence of both the SI and SS strategy during shoulder movements, similar to those demonstrated at the elbow, would suggest a broader application of the dual control hypothesis for movement control.

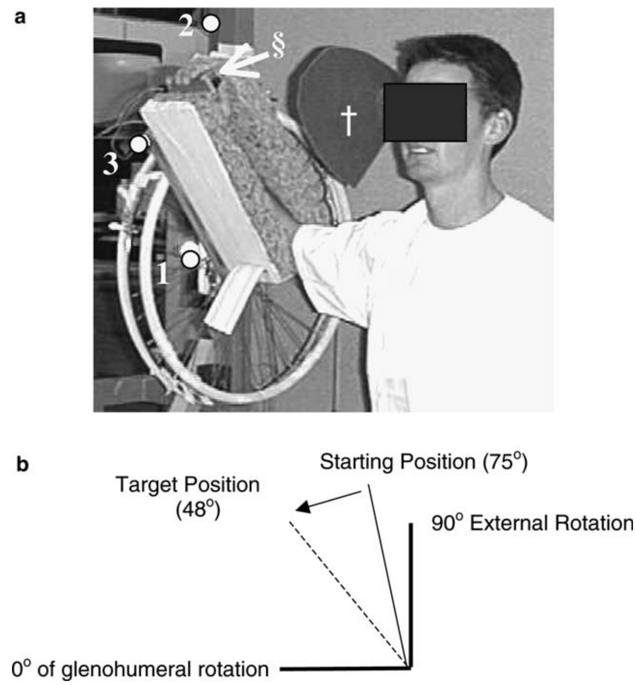

Fig. 1. (a) Subject positioned in shoulder wheel. Direct visualization of shoulder wheel and arm is prevented by the subject wearing goggles and shield (†). Retro-reflective markers used to measure shoulder angular position and velocity are located at the axis of rotation (1), on the stationary arm (2) and the movable arm (3) of the shoulder wheel. A thumb switch located in right hand (§) and is used to indicate when subject feels they have arrived at the target position. The glenohumeral joint is maintained at 90° abduction in the scapular plane (i.e., 30° anterior to frontal plane). (b) The starting position was 75° of external rotation; the glenohumeral joint was internally rotated (27°) to the target position of 48° of external rotation.

2. Methods

2.1. Subjects

Subjects were recruited from the general student population at the University of Kentucky (Table 1). Subjects were screened by a physical therapist. This screening included a brief medical history; assessment of glenohumeral range of motion (internal and external rotation), upper extremity muscle strength and upper extremity sensation. These tests were used to exclude individuals with recent shoulder injury or neurovascular compromise and to determine the dominant extremity. The institutional ethics committee approved this study; all subjects provided informed written consent. The subjects' dominant upper extremity was defined as the extremity with which they would prefer to throw a ball.

2.2. Shoulder wheel apparatus

Subjects sat upright and abducted their shoulder 90° and horizontally adducted 30°, to function in the plane of the scapula. The elbow was flexed 90° with forearm in a neutral position (half way between pronation and supination). The forearm was then inserted into a custom made shoulder wheel, and stabilized with a com-

Table 1
Subject demographic information, means (±SD)

Subjects ($n = 20$)	Age (years)	Height (cm)	Weight (kg)	Range of motion (°)		Subjects right-hand dominant
				Internal rotation	External rotation	
20	27.2 (3.3)	173.2 (18.1)	70.8 (14.5)	51.7 (15.2)	82.5 (13.7)	20

pressive sleeve around the forearm (Fig. 1(a)). The axis of rotation of the glenohumeral joint was visually aligned with the axis of rotation of the shoulder wheel. The sleeve on the movable arm of the shoulder wheel was padded to evenly distributed pressure throughout the forearm in order to reduce cutaneous feedback. Cutaneous feedback is known to enhance proprioceptive feedback even though sensitivity to stimuli can vary throughout the upper extremity [13]. To prevent injury, buttresses were attached to the rim of the shoulder wheel to limit external rotation to a maximum of 75° (relative to starting position) and prevent internal rotation beyond neutral or 0°. The buttress that limited external rotation also served as a consistent starting position for all subjects. The shoulder wheel allowed subjects to move from a position of 75° of external rotation through a range of internal rotation to a fixed target 27° from the starting point (Fig. 1(b)). Three retro-reflective markers were located on the shoulder wheel. One marker was coincident with the axis of rotation of the shoulder wheel. The second marker was attached on the stationary arm vertical to the axis of rotation of the shoulder wheel. The third marker was placed on the movable arm of the shoulder wheel, approximating the location of the hand when subjects were in the device (Fig. 1(a)). The movable arm represented the forearm position relative to the stationary arm and provided a measure of upper extremity kinematics as an estimate of glenohumeral motion [1,28,39,34,4,42]. A retractable third buttress was used to consistently position the movable arm of the shoulder wheel 27° from the starting position, indicating the target position. This buttress was moved out of the way during data collection so that it did not interfere with movement of the movable arm during the data collection trials.

2.3. Procedure

In this study, movements that included direct visual feedback are referred to as the visual-feedback (VF) condition, and movement conditions performed in the absence of visual feedback were referred to as the proprioceptive feedback (PF) condition. We do, however, acknowledge the presence of kinesthetic feedback during the VF condition; for clarity we will use the dominance of vision during the VF trials to identify these movement sequences. Subjects were seated as described above and were asked to move the forearm, enclosed within the movable arm of the shoulder wheel, to the target position from the starting position for both the VF and PF conditions (Fig. 1). The internal rotation movement is similar to previous shoulder proprioception studies [1,28,39].

A video camera and monitor, independent from the motion analysis system, provided visual feedback of movement accuracy. The camera was located 3 m from the shoulder wheel apparatus, oriented with its lens-to-object axis parallel to the rotational axis of the shoulder wheel. The video monitor (38.1 cm diagonal) for this separate camera was placed 2 m directly in front of the subject to provide real time two-dimensional visual feedback of the retro-reflective markers of the shoulder wheel and target during the VF condition. The video monitor was covered to eliminate the visual feedback of the retro-reflective markers and targets during the PF condition. In order to provide kinesthetic information about limb and target location for the PF trials, subjects were passively moved from the starting position to the target position, approximating the movable buttress, held in that position for 10 s and instructed that this was the target position. Subjects were then passively returned to the starting position and awaited instructions to move to the target as described below. Subjects wore goggles that blocked peripheral vision preventing direct visualization of the arm, forearm, hand and shoulder wheel movement for both the VF and PF conditions (Fig. 1(a)).

Subjects moved to the target at three different speeds (slow, natural, and fast) for both conditions (VF and PF). Speed of the movement was controlled with verbal instructions prior to every trial. For the slow movements, subjects were instructed to move "slowly and accurately, but to complete the movement within three seconds" in an attempt to assure some consistency of the task. Subjects were instructed to move "at a natural speed, as accurately as possible", or "as fast and as accurately as possible" for the natural and fast movement speeds, respectively. A total of 48 trials were performed where each trial was randomly assigned one of the six types of movements (VF-slow, VF-natural, VF-fast, PF-slow, PF-natural, PF-fast), yielding eight trials per test condition. This procedure took approximately one and a half hours to complete. Breaks were provided after a block of 10–20 repetitions, or upon the subject's request, and typically lasted 5–10 min. These frequent breaks were used in order to minimize the influence of muscular and mental fatigue.

2.4. Kinematics

Kinematic data were recorded by three high-speed Falcon Cameras (Motion Analysis Inc., Santa Rosa, CA) positioned in a semicircle around the shoulder wheel, approximately 5 m away and elevated approximately 4 m above the floor, with a sampling rate of 240 Hz. Three-dimensional coordinate data for the three retro-reflective markers were determined using the direct linear transformation (DLT) as modified by Motion Analysis Corporation Software (Motion Analysis Inc., Santa Rosa, CA) [23]. Coordinate data were smoothed with a fourth order, zero-lag, low-pass Butterworth filter with a cutoff frequency of 6 Hz [44]. Standard trigonometry was then utilized to calculate the angle formed by the stationary and movable arm of the shoulder wheel in each frame. Displacement of the shoulder wheel's movable arm relative to the stationary arm enabled the determination of the instantaneous position of the movable arm. The instantaneous angular velocity of the movable arm was calculated from angular position data using central differences. With the forearm fixed in the shoulder wheel, the displacements and velocities were representative of glenohumeral joint internal rotation.

A thumb switch, initiating a DC voltage, was synchronized with the motion capture system. This analog signal was sampled at 960 Hz, converted to a digital signal and stored on a PC. Subjects were instructed to activate the thumb switch, which served as an indicator for when subjects thought they were at the target position, only after they had stopped at the target. Thus, final location of shoulder wheel was determined by comparing the final position of the retro-reflective marker of the movable arm of the shoulder wheel when its angular velocity was nearest zero, immediately prior to activation of the thumb switch.

2.5. Electromyography (EMG)

Prior to electrode placement, the skin approximating electrode location was cleaned with isopropyl alcohol. The EMG signals were pre-amplified (×35) at the skin with on site, solid state pre-amplifiers (Therapeutics Unlimited Inc., Iowa City, IA) in rectangular silver-silver chloride bipolar electrodes (2 cm center-to-center inter-electrode spacing, input impedance >25 MΩ at DC and >15 MΩ at 100 Hz, noise <2.0 μV RMS, common mode rejection 87 dB at 60 Hz). Electrodes were placed in the middle third of the muscle and orientated parallel to the muscle fiber orientation for the following muscles; sternal portions of the pectoralis major (PECT), anterior deltoid (ADELT) and posterior deltoid muscles (PDELT) [11].

A custom made stainless steel, Teflon™ coated fine-wire electrode was inserted with a sterilized hypodermic needle into the middle third of the infraspinatus muscle belly (INFRA) [11]. The hypodermic needle was immediately withdrawn leaving the fine-wire electrode within the muscle. A reference electrode was placed on the acromiom process of the right shoulder. Electrode placement was confirmed with manual muscle testing and EMG signal inspection on an oscilloscope.

All EMG data were sampled (960 Hz), amplified (1000×), and band-pass filtered (10–1000 Hz) during collection. The EMG data underwent an analog to digital conversion (16 bit) and was stored on a PC. An initial "quiet" file of data were collected with subjects' arms resting at their sides and stored for subsequent analysis; these data represent baseline EMG activity and were used to determine a threshold for muscle activation during movement trials. EMG activity was then recorded for all four muscles during a maximum voluntary isometric contraction (MVIC). Subjects were instructed to push maximally against manual resistance, using standard manual muscle testing techniques [25]. The EMG data for this MVIC were collected for one second of maximal effort and stored for future processing.

The EMG signals were synchronized with the kinematic data and analyzed to determine duration, amplitude and initial burst EMG activity using a custom program in Matlab (Mathworks Inc., Natick, MA). EMG data for each trial were full-wave rectified and filtered using a fourth order low-pass Butterworth filter with a cutoff frequency of 10 Hz, to create a linear envelope of the data [27]. The threshold voltage (Vo), to establish the onset of muscle activity, was determined to be five standard deviations above the mean of the "quiet file" [12,22]. The onset of the muscle activity was evaluated by both visual inspection of the signal and an algorithm that compared discrete data point values (Vi) in a point-by-point fashion to the threshold voltage. When the mean voltage of a 25 ms window of data immediately adjacent to the point (50 ms total) that exceeded Vo, the initial data point value was considered to represent the onset of muscle activity (Vi_1). Termination of muscle activity was defined when the second discrete data point value (Vi_2) of a 25 ms window of data immediately adjacent to the point (50 ms total) dropped below Vo. The duration of the muscle activity was then determined to be the time from Vi_1 to Vi_2. Visual inspection was used to eliminate significant pre-movement artifact and provide links between multiple bursts that were clearly within the movement times of individual trials.

The amplitude of muscle activity was defined as the area under the linear envelope during the duration of muscle activity. The area under the linear envelope was calculated using the trapezoidal estimation technique. Magnitude of EMG activity for the linear envelope was calculated over the duration of EMG activity and reported as %MVIC [38]. Muscle burst activity (Q30) was calculated by measuring the magnitude of the linear envelope of the EMG signal over the initial 30 ms of EMG muscle activity, and was based on Gottlieb and Corcos [7,19–21] estimate of muscle burst activity. The Q30 activity was also reported as %MVIC.

The mean values of EMG durations, linear envelopes and Q30 activity from the eight trials for each test condition were analyzed for each muscle and then averaged to calculate an overall mean for each test condition.

2.6. Kinematic normalization

To compare kinematic movement characteristics, each trial was divided into four phases: acceleration,

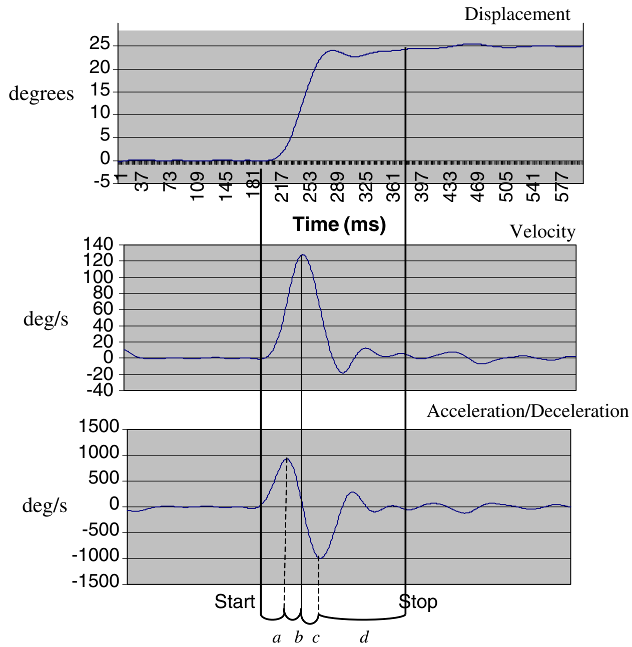

Fig. 2. Kinematic normalization of movement: Phases are divided by movement time (start to stop) for normalization: (a) acceleration phase; onset of movement to peak acceleration; (b) velocity phase; from peak acceleration to peak velocity; (c) deceleration phase; from peak velocity to peak deceleration; (d) error correction phase; from peak deceleration to end of movement.

velocity, deceleration and error correction (Fig. 2). Phases were defined by maximum kinematic values associated with each movement relative to the whole movement time to allow comparison of the contributions to each movement [36,3]. The acceleration phase was determined from onset of movement to peak acceleration (Fig. 2(a)). The velocity phase was from peak acceleration to peak velocity (Fig. 2(b)), the deceleration phase was from peak velocity to peak negative acceleration (Fig. 2(c)) and the error correction phase was the remainder of movement time (Fig. 2(d)). It is important to note that the acceleration phase was only calculated up to peak acceleration, although acceleration continued to peak velocity. Similarly, negative acceleration continued after the peak negative acceleration in the deceleration phase. The four kinematic phases were then divided by the total movement time for phase normalization. This procedure is commonly used in gait analysis and enabled the detection of kinematic differences between test conditions, regardless of subtle variations in movement times [40].

2.7. Statistical analysis

Dependent variables for this analysis were the movement accuracy, normalized kinematic temporal parameters (acceleration, velocity, deceleration and error correction phases), the magnitudes of the kinematics (acceleration, velocity and deceleration) and EMG parameters (duration, amplitude and burst) [35]. Independent variables included the test conditions (VF, PF) and the three different movement speeds (slow, natural, fast). A 2×3 ANOVA with repeated measures (SPSS Inc., Chicago, IL), was performed on all the dependent variables; level of significance set at $\alpha = 0.05$. Post hoc pairwise comparisons, with a Bonferroni correction to α, were used to determine differences among movement speeds.

3. Results

3.1. Kinematics characteristics with change in movement speed condition

With no reported interactions between the movement speed and feedback conditions for any of the kinematic characteristics (movement phases and kinematic magnitudes), there were, however, significant differences between the duration of the time normalized kinematic phases across all three movement speeds (slow, natural, fast). As subjects attempted to move faster, there were significant increases in duration of the normalized acceleration ($F_{(2,38)} = 136.5$, $p < 0.01$), velocity ($F_{(2,38)} = 25.6$, $p < 0.01$), and deceleration phases ($F_{(2,38)} = 16.2$, $p < 0.01$). The error correction phase demonstrated a significant decrease ($F_{(2,38)} = 35.1$, $p < 0.01$) in duration as movement speed increased. Post hoc pairwise comparisons among the three movement speeds, demonstrated significant differences ($p < 0.01$) in the time normalized duration of the kinematic phases between the slow and fast movement speeds and between the slow and natural movement speeds; there was no signif-

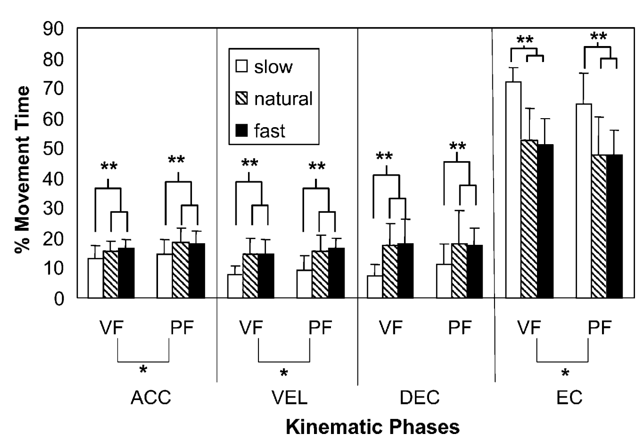

Fig. 3. Comparison of the duration of acceleration (ACC), velocity (VEL), deceleration (DEC), and error correction (EC) phases for all three movement speeds with visual feedback (VF) and proprioceptive feedback (PF). * indicates significant difference ($\alpha < 0.05$) between the VF and PF. **Post hoc comparisons among the three movement speeds indicates significant difference ($\alpha < 0.0167$) between the slow and natural, and slow and fast movement speeds.

Table 2
Peak amplitude (means ± SD) associated with each temporal phase during different speeds and different types of feedback

		Movement speed			Main effect-feedback
		Slow	Natural	Fast	
ACC (°/s^2)	VF	130.2 ± 23.2	178.0 ± 37.2	499.7 ± 143.5	269.3 ± 49.8
	PF	154.0 ± 32.9	200.9 ± 49.6	497.7 ± 141.3	284.2 ± 47.4
Main effect-speed*		142.1 ± 28.5	189.5 ± 43.6	498.7 ± 142.4	
VEL (°/s)	VF	24.9 ± 8.0	43.4 ± 23.7	92.1 ± 26.7	53.5 ± 6.8**
	PF	31.5 ± 7.6	52.1 ± 29.2	96.3 ± 20.1	60.0 ± 10.2
Main effect-speed*		28.2 ± 7.4	47.8 ± 26.9	94.2 ± 22.9	
DEC (°/s^2)	VF	−150.6 ± 42.6	−177.6 ± 52.2	−505.4 ± 156.1	−277.9 ± 62.5
	PF	−150.6 ± 37.5	−197.1 ± 47.1	−491.3 ± 167.5	−279.6 ± 60.3
Main effect-speed*		−150.6 ± 40.9	−187.3 ± 49.9	−498.4 ± 160.8	

* Significant differences ($p < 0.05$) among all speeds (slow to natural, slow to fast and natural to fast) for acceleration, velocity and deceleration phases.
** Significant difference ($p < 0.05$) between VF and PF movements.

Table 3
Duration (means ± SD) of EMG activity during voluntary shoulder movements

		Movement speed			Main effect-feedback
		Slow	Natural	Fast	
ADELT	VF	2615.2 ± 203.5	2363.8 ± 197.9	2364.8 ± 143.9	2247.9 ± 172.1
	PF	2711.1 ± 184.7	2502.8 ± 175.7	2359.2 ± 151.8	2524.3 ± 162.0
Main effect-speed		2663.1 ± 190.1*	2433.0 ± 181.3	2362.0 ± 140.6**	
PECT	VF	1390.5 ± 280.8	1300.1 ± 248.3	1544.9 ± 229.8	1411.8 ± 239.1
	PF	1524.6 ± 262.1	1385.1 ± 267.6	1502.7 ± 249.2	1470.8 ± 254.3
Main effect-speed		1457.5 ± 268.3	1342.6 ± 253.3	1523.8 ± 234.2	
PDELT	VF	1841.1 ± 220.9	2261.2 ± 243.6	1677.0 ± 211.9	1926.4 ± 196.8
	PF	2028.3 ± 244.2	1736.6 ± 185.6	1952.5 ± 206.1	1905.8 ± 180.3
Main effect-speed		1934.7 ± 206.7	1989.9 ± 187.3#	1814.7 ± 175.9	
INFRA	VF	2749.5 ± 217.1	958.9 ± 275.6	2072.7 ± 200.9	1827.1 ± 173.8
	PF	936.1 ± 160.5	2172.1 ± 183.9	955.7 ± 153.9	1188.2 ± 139.2†
Main effect-speed		2522.9 ± 158.2*	2189.0 ± 198.0	2283.7 ± 158.2**	

* Significant difference ($p < 0.017$) between slow and natural movement speed.
** Significant difference ($p < 0.017$) between the slow and fast movement speed.
# Significant difference ($p < 0.017$) between the natural and fast movement speed.
† Significant difference ($p < 0.05$) between movements with VF and PF.

icant difference in kinematic time normalized phases between the natural and fast movement speeds (Fig. 3).

The peak acceleration ($F_{(2,38)} = 108.0$, $p < 0.01$), velocity ($F_{(2,38)} = 109.8$, $p < 0.01$) and deceleration ($F_{(2,38)} = 86.1$, $p < 0.01$) increased significantly as movement speed increased. Post hoc pairwise comparisons among the three movement speeds indicated significant differences among all three movement conditions for peak acceleration, velocity, and deceleration (Table 2).

3.2. EMG characteristics with change in movement speed condition

There was a significant interaction ($F_{(2,38)} = 5.28$, $p < 0.01$) between the feedback condition and the movement speed for ADELT linear envelope. Post hoc test of the interaction indicated no significant differences among movement speeds, therefore, main differences are between the PF and VF conditions (Table 4). Duration of ADELT EMG activity shortened significantly ($F_{(2,38)} = 6.2$, $p < 0.01$) as movement speed increased. Post hoc pairwise comparisons demonstrated significance when the slow speed was compared to the natural ($p < 0.01$) and fast ($p < 0.01$) movement speeds (Table 3).

Linear envelope of the PECT significantly increased ($F_{(2,32)} = 7.3$, $p < 0.01$) as movement speed increased. Post hoc comparisons demonstrated significance between slow and natural ($p = 0.015$) and natural and fast ($p = 0.011$) movement speeds (Table 4).

The PDELT demonstrated significant differences among the three movement speeds for duration ($F_{(2,36)} = 3.9$, $p = 0.03$), linear envelope ($F_{(2,38)} = 5.9$, $p < 0.01$), and Q30 ($F_{(2,38)} = 4.98$, $p = 0.012$) activity. Post hoc pair-wise comparisons demonstrated significant decrease

Table 4
Linear Envelopes (%MVIC ± SD) of EMG activity during voluntary shoulder movements

		Movement speed			Main effect-feedback
		Slow	Natural	Fast	
ADELT	VF	19.1 ± 3.9	11.8 ± 2.4	17.2 ± 3.7	16.1 ± 3.0
	PF	9.8 ± 1.9	18.6 ± 3.9	11.7 ± 2.1	13.4 ± 2.3
Main effect-speed		14.5 ± 2.6	15.2 ± 2.8	14.4 ± 2.5	
PECT	VF	7.1 ± 1.9	6.7 ± 1.7	10.1 ± 2.3	8.1 ± 1.7
	PF	8.4 ± 1.7	7.1 ± 1.9	10.3 ± 2.3	8.5 ± 1.8
Main effect-speed		7.8 ± 1.8*	6.9 ± 1.7#	10.1 ± 2.0	
PDELT	VF	8.9 ± 2.2	7.9 ± 2.1	12.1 ± 3.4	9.6 ± 2.5†
	PF	10.1 ± 2.6	11.2 ± 2.1	14.8 ± 4.3	12.0 ± 3.1
Main effect-speed		9.5 ± 2.4	9.5 ± 2.3#	13.4 ± 3.8**	
INFRA	VF	20.3 ± 4.1	10.9 ± 2.8	21.0 ± 3.9	16.4 ± 2.8
	PF	8.4 ± 2.1	23.2 ± 6.1	9.8 ± 2.9	17.8 ± 2.9
Main effect-speed		12.5 ± 4.2*	16.3 ± 6.4	14.9 ± 5.2**	

* Significant difference ($p < 0.017$) between slow and natural movement speed.
** Significant difference ($p < 0.017$) between the slow and fast movement speed.
# Significant difference ($p < 0.017$) between the natural and fast movement speed.
† Significant difference ($p < 0.05$) between movements with VF and PF.

Table 5
Q30- Muscle Burst Activity (%MVIC ± SD) of EMG activity during voluntary shoulder movements

		Movement speed			Main effect-feedback
		Slow	Natural	Fast	
ADELT	VF	1.57 ± 0.55	1.48 ± 0.46	1.61 ± 0.51	1.45 ± 0.49
	PF	1.42 ± 0.63	1.70 ± 0.44	1.53 ± 0.49	1.65 ± 0.51
Main effect-speed		1.53 ± 0.48	1.56 ± 0.43	1.57 ± 0.45	
PECT	VF	1.34 ± 0.17	1.67 ± 068	3.05 ± 0.15	2.09 ± 0.52
	PF	1.86 ± 0.5	1.57 ± 0.22	2.04 ± 0.68	1.91 ± 0.47
Main effect-speed		1.59 ± 0.31	1.61 ± 0.21	2.61 ± 0.23	
PDELT	VF	1.35 ± 0.36	1.22 ± 0.38	1.50 ± 0.39	1.32 ± 0.35
	PF	1.31 ± 0.38	1.36 ± 0.32	1.53 ± 0.35	1.35 ± 0.36
Main effect-speed		1.32 ± 0.37	1.26 ± 0.31	1.51 ± 0.38	
INFRA	VF	2.50 ± 0.55	2.26 ± 0.44	2.52 ± 0.51	2.37 ± 0.42†
	PF	2.82 ± 0.53	3.70 ± 0.083	2.41 ± 0.058	2.97 ± 0.56
Main effect-speed		2.62 ± 0.47	2.94 ± 0.58#	2.45 ± 0.41**	

** Significant difference ($p < 0.017$) between the slow and fast movement speed.
# Significant difference ($p < 0.017$) between the natural and fast movement speed.
† Significant difference ($p < 0.05$) between movements with VF and PF.

in EMG duration between the natural and fast movement speeds ($p = 0.014$) (Table 3). There was also a significant increase in the amplitude of the linear envelopes detected with post hoc analysis between the slow and fast ($p = 0.006$) and natural and fast ($p = 0.016$) movement speeds (Table 4). Likewise, significantly greater Q30 muscle activity was detected between post hoc testing slow and fast ($p = 0.007$) and natural and fast ($p = 0.002$) movement speeds (Table 5).

Significant interactions were demonstrated between the movement speeds and the type of feedback for duration ($F_{(2,38)} = 80.2$, $p < 0.01$), and linear envelope ($F_{(2,38)} = 19.6$, $p < 0.01$), for INFRA activity. These interactions demonstrated generally shorter durations of EMG activity for PF movements and greater EMG activity with faster movements. The INFRA demonstrated significant differences among the three movement speeds for EMG duration ($F_{(2,36)} = 7.4$, $p < 0.01$), and linear envelope ($F_{(2,36)} = 10.4$, $p < 0.01$) activity. Post hoc pairwise comparisons demonstrated significant differences in INFRA duration of EMG activity between the slow and natural ($p < 0.01$) and slow and fast ($p < 0.01$) movement speeds (Table 3). Post hoc differences were detected for linear envelope activity between the slow and natural ($p = 0.016$) movement speeds and the slow and fast ($p = 0.013$) movement speed. There was not a significant difference in Q30 activity among the three movement speeds (Table 4).

3.3. Kinematics characteristics with change in feedback

The duration of the normalized acceleration ($F_{(1,19)} = 13.7$, $p < 0.01$) and normalized velocity ($F_{(1,19)} = 6.3$, $p < 0.01$) phases were significantly greater in the PF condition with a concomitant decrease in the error correction ($F_{(1,19)} = 15.6$, $p < 0.01$) phase. There was no significant difference ($p = 0.2$) in the normalized deceleration phases between movements with the two feedback conditions (Fig. 3). The PF condition demonstrated significantly greater peak velocity ($F_{(1,19)} = 11.1$, $p < 0.01$), but there was no significant difference between the PF and VF condition for peak acceleration ($p = 0.17$) or deceleration ($p = 0.86$) (Table 2).

3.4. EMG characteristics with change in feedback

The PDELT demonstrated a larger linear envelope ($F_{(1,18)} = 5.6$, $p = 0.029$) for the PF condition compared to the VF condition (Table 4). The duration of EMG activity for the INFRA was significantly greater ($F_{(1,18)} = 12.9$, $p = 0.002$) in the VF condition versus the PF condition (Table 3). The INFRA Q30 activity was significantly greater ($F_{(1,18)} = 5.7$, $p = 0.028$) for the PF condition versus the VF condition (Table 5).

3.5. Accuracy of movements

There was significantly greater ($F_{(2,38)} = 6.7$, $p < 0.01$) movement accuracy observed across movement speeds. Post hoc pairwise comparisons demonstrated that significance between the slow and fast ($p = 0.023$) and between the natural and fast ($p = 0.002$) movement speeds. The ability to replicate target location was similar for the slow and natural movement speeds ($1.7 \pm 1.5°$ SD) compared to ($3.5 \pm 2.5°$ SD) the fast movement speeds. Movement accuracy was significantly greater ($F_{(1,19)} = 29.3$, $p < 0.01$) for movements with VF ($-0.1 \pm 0.1°$ SD) compared to movements with PF ($4.1 \pm 0.8°$ SD).

4. Discussion

Gottlieb and Gorcos [19,20] described a dual control hypothesis consisting of a speed insensitive (SI) and speed sensitive (SS) movement strategy for elbow movements. Where the SS control strategy is used when a given task requires rapid movements leading to greater burst of muscle activity and resultant kinematics, while the SI control strategy uses primarily duration of muscle activity as a control mechanism. The study presented here examined simple shoulder movement with visual or proprioceptive feedback and how the type of feedback appears to influence the movement strategy. Important to dual control hypothesis is how this strategy relates to the desired movement velocity.

4.1. Effects of movement speed on kinematic characteristics

Subjects were required to move to the target position at three self-selected speeds and results were pooled across the two major types of feedback (VF and PF). The strategy employed when subjects moved to the target appears similar to the SS strategy. Increased movement speed resulted in significantly longer duration of the normalized acceleration, velocity and deceleration phases with associated increases in peak acceleration, velocity and deceleration phases. Corcos et al. [7] observed similar increases in initial joint torques or associated peak angular accelerations as elbow flexion movement speed increased leading to decreased movement times. The peak velocity of the natural movement speed was nearly double the slow movement; similarly the peak velocity of the fast movement was nearly double the natural movement, but there was no statistically significant difference in the kinematic time normalized phase duration between the natural and fast movement speeds. Schneider and Schmidt [36] used kinematic landmarks to identify separation of motor programs as a way of examining upper extremity control strategies during simple movements. The slow movement demonstrated significant difference in kinematic phases compared to the natural and fast movements. This suggests that the differences between the movements could be attributable to the peak acceleration, velocity, and deceleration regardless of changes in the duration of kinematic phases, similar to the SS strategy. While the slowest movement was the only one that demonstrate a difference from the two other movements, it was also the only one for which an attempt was made to limit the movement time (3 s).

The duration of the normalized acceleration, velocity and deceleration phases was significantly longer (% of movement time) for the fast movement compared to the slow movement. This coincides with movement accuracy, where there was significantly less accuracy with the fast ($3.5 \pm 2.5°$ SD) movements compared to both the slow ($1.7 \pm 1.5°$ SD) and natural ($1.7 \pm 1.5°$ SD) movements, as less of the movement was spent in the error correction phase. With multi-joint upper extremity movements, Gordon and Ghez [17] suggest a pulse amplitude modulation as an effective means to simplify movement control in the upper extremity, as a possible mechanism to enhance function and accuracy.

The kinematic characteristics of the movements in this study suggests a modulation of the movement as a result of changes in the magnitudes of acceleration, velocity, and deceleration and is consistent with the SS control strategy to increase speed in single and multi-joint upper extremity movements presented by other investigators [7,19,26,29]. The shoulder movements in this study were also more complex (joint architecture) than gravity eliminated simple elbow flexion and extension movements performed in prior studies. This is important, because similar CNS control strategies for different joints and different types of movements suggest a more robust CNS control strategy for modulating movement speed. It remains to be determined if this method of control applies to discrete movements of distal parts of the upper extremities or for lower extremity tasks.

4.2. Effects of movement speed on EMG characteristics

As movement speed increased a shorter duration of activity of the INFRA, PDELT, and ADELT was observed. With faster movement speeds we also observed larger linear envelopes of EMG activity for the INFRA, PDELT, and PECT and a larger Q30 activity, representative of muscle burst activity of the PDELT. While this list does not include all muscles, there does seem to be a trend of greater muscle activity, over a shorter period of time, with increased movement speed and a concomitant increase in peak acceleration, velocity and deceleration. This trend is not just for muscles initiating the internal rotation movement, but also deceleration or stopping of this movement and has also been observed with deceleration of elbow movements [24].

The glenohumeral joint is very complex with many muscles contributing to even simple movements and we only evaluated four of the many muscles that can influence shoulder movement. The most glaring muscle omission from this study is the subscapularis, the primary internal rotator of the humerus. However, the fine wire EMG techniques to study this muscle could have been a detriment to subject recruitment. The study presented here evaluates CNS control of a complex joint during a simple movement and our data appears consistent with previous studies of movement control at the elbow, in which the amplitude of EMG activity increased with movement speed, whether the muscle functioned as an agonist or antagonist to the movement task [7,19,20].

4.3. Accuracy of increased movement speed

As movement speed increased subjects were less accurate; this is consistent with Fitts' law, which dictates that accuracy of a movement decreases linearly as movement speed increases [14]. Fitts law describes the effect of target width and target distance on movement time. Larger targets and longer movements generally produce faster, but less accurate movements. In our study, there was less accuracy at fast movement speeds compared to the slow and natural movement speeds. There was, however, no difference in the accuracy between the slow and natural movement speeds. While movement speed nearly doubled between the slow and natural movement speed (28.2–47.8°/s) the accuracy did not change appreciably. Possible explanations for the lack of significance between the slow and natural movement speeds include the use of verbal instructions to control velocity as opposed to target width and the use of a gross motor shoulder movement as opposed to Fitts' tracking task which is a fine motor task.

As subjects moved to their target position direct visual feedback was necessary to maintain accuracy at the greater movement speeds, but as movement velocity increased Fitts' law dictated increased movement error. In summary, as movement speeds increased from slow to natural or fast, there was an increase in the duration of the normalized acceleration, velocity and deceleration phases, with a concomitant decrease in the error correction phase. Along with these increases there were significant increases in peak values across all three movement speeds of acceleration, velocity and deceleration phases, suggesting it is the peak values of acceleration, velocity and deceleration and not the duration of the movement time accounting for the changes in movement speeds, indicating the presence of the SS strategy. The EMG data are less clear but we believe that there is a trend for shorter but a greater burst of initial EMG activity to explain the increased movement speed. We contend that these EMG and kinematic characteristics (normalized kinematic phases and peak values) are consistent with the SS strategy to modulate movement speed [7].

4.4. Effects of feedback on kinematic characteristics

The tendency for a SI strategy during PF movements lies in significant differences among the duration of the kinematic phases with no concomitant changes in the magnitudes of acceleration, velocity and deceleration. As expected, subjects performed less accurately with PF compared to VF as the resolution of the visual system is probably greater than proprioceptive feedback alone, creating faster movements overall but a SI strategy based on modulations of the kinematic phase durations, as opposed to the magnitude of peak acceleration and deceleration.

Overall, PF movements demonstrated a greater peak velocity compared to movements with VF. Faster movement velocity for movements with PF, observed in this study, is consistent with prior studies evaluating upper extremity proprioception and movement accuracy [18,16,15]. While there were no significant differences in peak acceleration or deceleration between VF and PF movements, differences were found in the duration of the normalized acceleration, velocity and error correction phases. This evidence seems to suggest that duration modulation, similar to the SI strategy, versus amplitude modulation, or the SS strategy, is used for movements with PF, regardless of the movement velocity. This temporal modulation of movement is not only consistent with the previously described SI strategy but also Gordon and Ghez's [20,17] "pulse duration" modulation with active upper extremity movements.

Of particular interest was the lack of significant difference in the duration of the deceleration phases between the VF and PF movements. In contrast, there was a significant difference between the duration of the acceleration phases between the VF and PF movements. This could be suggestive of what Gordon et al. [18] describe as an initial feed-forward strategy early in the movement and a feedback strategy later in the movement. This strategy would stress the importance of peripheral receptors providing feedback to the CNS during the latter stages of a movement. Gottlieb et al. [19] also found that initial accelerating torque is not well correlated to peak decelerating torque. These factors suggest that the deceleration phase could be the separation between the feed-forward and feedback portions of the motor plan for this task, similar to the segmentation of a motor plan that Schneider and Schmidt [36] describe.

4.5. Effects of feedback on EMG characteristics

The relationship between EMG activity and kinematic parameters has been demonstrated in previous studies of upper extremity movements [7,19,20,29]. In our study, it is the evaluation of EMG activity and the kinematic phases that provides a more comprehensive examination of CNS control during movement. Despite variability associated with EMG activity, it is interesting that the INFRA muscle demonstrated different patterns of activity between the VF condition and the PF condition. The INFRA is ideally positioned, anatomically, to influence shoulder movements such as decelerating the arm following the voluntary internal rotation and may be thought of as a "kinesiological monitor" to provide important kinesthetic feedback to the CNS [32]. Trained athletes, for example, have demonstrated the capacity to prolong INFRA muscle latency to sudden internal rotation movements [4]. The INFRA may play an important role in controlling glenohumeral movements in the absence of visual feedback although further study of the role of INFRA with proprioceptive feedback is warranted.

4.6. Accuracy PF versus VF movements

Repositioning accuracy studies to evaluate shoulder proprioception report similar measures of error (1–6°) in reproducing targets to those that were observed in our study [1,28,39]. While these previous investigators used passive movement of the shoulder to replicate target position, our study used active movements to replicate target position. Rogol et al. [34] demonstrated no significant difference in testing the shoulder actively or passively when assessing shoulder kinesthesia. A SI strategy employed during movements without visual feedback could be the result of less resolution of the proprioceptive feedback compared to the visual feedback, regardless of peak velocity that was achieved during the PF condition. Gottlieb and Corcos [20] describe the SI strategy being the default strategy for movements until a time constraint is introduced, forcing a SS strategy. We contend that the SI strategy is used for slower movements regardless of type of feedback but as movement speed is increased PF does not permit the implementation of the SS strategy because of the limitations of the PF. However, the resolution of the visual system enables the implementation SS strategy and while also maintaining movement accuracy.

In summary, while movement speed was greater (peak velocity) for the PF movements, there was an increase in the duration of the normalized acceleration and velocity phases, with a concomitant decrease in the error correction phase and accuracy. However, along with the increased phase duration there were no significant increases in peak acceleration or deceleration. This suggests that it is the duration of kinematic phases, not the peak value, that account for differences in the strategies behind PF and VF movements. Lack of differences in EMG characteristics suggests no difference in movement strategies between these two types of movements, other than some changes in the PDELT.

5. Limitations

This study examined the performance of a simple internal rotation at the shoulder, in a restricted plane. However, the trunk and scapula were not fixed, where the scapulothoracic junction, the sternoclavicular joint and the acromiomclavicular joint could all contribute to the observed motion. It is unclear if performance of a movement in a restricted single plane, applies to unrestricted, three-dimensional movements. While it has been reported that restricted movements appear to bias a motor performance, it is not conclusive if movements restricted to a single degree of freedom have any impact on three-dimensional functional movements [35,6]. We also presented the visual feedback in two dimensions (video monitor), whereas visual feedback is usually in three dimensions. Restricting movements to a single plane and providing only two-dimensional feedback may limit the interpretations of the study of unrestricted human movement.

Calculation of the Q30 measure of muscle burst activity was slightly different than the calculation of muscle burst activity that was measured by the Gottlieb group [19]. In addition to integrating over the initial 30 ms, we also normalized by MVIC, as we were comparing among many subjects and could have deemed some of the outcomes insignificant in this analysis.

The use of verbal instructions to control movement velocity is common for studying upper extremity movements to a fixed target, however, these instructions are open to interpretation by each of the subjects and may contribute to the variability of the initial chosen EMG activity, acceleration and velocity and overall accuracy of the movements [7,19,20]. Subjects were free to choose the initial velocity based on the verbal instructions, however, there was no attempt to control the latter half of the movement where subjects could perform any number of error correction maneuvers in order to seek the final target. This could have dissociated the initial peak velocity from the accuracy required near the end of the movement, contributing to lack of significant difference in movement accuracy between the slow and natural movement speeds [14]. Improved accuracy is generally achieved by adjusting initial velocity in the beginning of motion in relation to minimizing error corrections later in the movement [26,30]. Training subjects to restrict error corrections later in the movement may have enhanced the direct relationship between movement velocity and accuracy [30].

Due to complexity of the CNS, it was the overall trend of the kinematic and EMG parameters that were used to allow us to speculate on possible CNS control mechanisms for simple shoulder movements at three different speeds, with and without vision. Simple shoulder movements, used in this study, were controlled like simple elbow movements using the dual control hypothesis (SI and SS), that has been defined by Gottlieb's group [7,19,20].

While we believe the results of this study support the dual control hypothesis for shoulder movements, with or without vision, there is clearly a need for further study of CNS control mechanism for more functional movements. In addition, movements that rely solely on PF appear to utilize an SI strategy with significant changes across kinematic phase durations, but nearly as many significant changes in the peak values of the kinematic phases. While it is important to understand possible CNS control strategies behind movements with various types of peripheral feedback, it appears that these strategies are not appreciably different than movements with visual feedback. Future studies are needed to determine if this CNS control strategy is consistent for multi-joint upper extremity movements and under a variety of conditions, for example, the addition of an external load.

6. Conclusion

The dual control hypothesis appears to be reasonable method of CNS control for the simple shoulder movements demonstrated in this study. This means that as subjects desired to increase their movement speed they employed a SS strategy to move to the target position. The presence of the dual strategies in shoulder movements in addition to elbow movements suggests a certain robustness of the dual control hypothesis, indicating that the CNS can use similar control strategies for most or all upper extremity movements. In addition, we also demonstrated that shoulder movements, in the absence of vision, employ a SI strategy regardless of the fact that movements were actually faster.

It has yet to be determined if lower extremity movements utilize similar CNS strategies that have been demonstrated in the upper extremity. Moreover, it is the interaction between the type of feedback and speed of movement on the implementation of the movement strategy that is important, whereby the visual system with greater resolution (less error) allows the implementation of a SS strategy, as opposed to the lower resolution proprioceptive feedback that limits CNS control to a SI strategy. Ideally, further investigation into the dual control hypothesis should focus on multi-joint movements and functional upper extremity movements, as this would have the greatest impact on identifying abnormal movement control as the result of injury or pathology.

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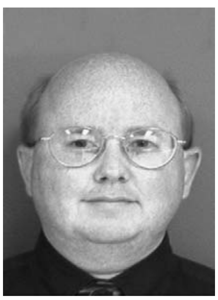

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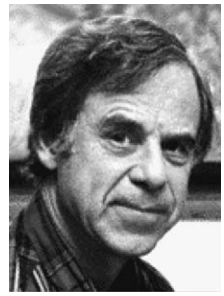

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