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The Relationship Between Biceps Activity and Shoulder Eccentric External Rotation Torque

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1. Introduction

Electromyography (EMG) and handheld dynamometry (HHD) are increasingly accessible clinical tools, and clinicians are beginning to use them to assess eccentric shoulder loading for individual-level decision-making such as return-to-play (Johansson et al., 2015). For example, Cools et al. (2016) published HHD strength reference values for overhead athletes, enabling clinical assessment of how an individual compares to normative data. However, the biceps' contribution during eccentric external rotation, thought to help decelerate the arm after throwing, remains unclear, and little is known about the feasibility of pairing EMG with HHD during a standardized shoulder eccentric external rotator test or how to interpret results without reference values. Therefore, the primary aim of this study is to quantify the association between mean and peak biceps EMG activity and eccentric shoulder external rotation torque using clinically available instrumentation.

2. Methods

A convenience sample of nineteen Division 1 softball players were enrolled for this study. Participants were free from shoulder injury and actively participating in team activities. The participants were seated with their shoulder at 90° abduction and external rotation, and their elbow at 90°. Their elbow was placed on a pad to ensure proper positioning and a therapist lightly placed their hand on the participant's shoulder as a cue to maintain proper shoulder positioning. T performed maximal eccentric external rotation against a handheld dynamometer, moving from 90° external rotation to 0° over three seconds, timed with a metronome. This method has been shown to be reliable, result in valid angular velocity (Johansson et al., 2015) and has subsequently been used in research (Cools et al., 2016; Johansson et al., 2025). A practice trial was performed, followed by two recorded trials. Testing was then repeated on the contralateral arm, with limb order randomized by a coin flip. Forearm length was used to derive torque, which was normalized to body mass (kg).

During testing, surface EMG was recorded from the biceps and using wireless Ultium EMG sensors (Noraxon USA Inc., Scottsdale, AZ; MR4.0 software). The skin was cleaned with an alcohol wipe, and the electrodes were placed parallel to the muscle fibers on the mid-belly (Delagi, 1980). The participants performed a two to three-second maximal voluntary isometric contraction (MVIC) of the biceps by exerting maximum effort against an immovable object with the elbow at approximately 90° in a seated position. The biceps signal was normalized to the MVIC signal. EMG was sampled at 2000 Hz, band-pass filtered (4th-order Butterworth, 10 and 500 Hz), rectified, and smoothed using a 50 ms root-mean-square moving window (Gurney et al., 2016). Onset and offset were identified from video-recorded visible motion, with synchronized inertial measurement unit plots from the EMG sensors used to inform the best estimate of movement start and end.

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3. Results and discussion

Two separate linear mixed regression models included mean EMG and maximum EMG on normalized torque (Nm/kg) and side from each trial. An interaction between the predictors was considered, and a random intercepts model was used to account for repeated measures. The outcome was log-transformed, and model diagnostics were performed to account for model fit. Partial Eta^2 is reported for the effect size (ES) for the interaction term. In contrast, ES for the beta coefficients is reported as the percent EMG change for a positive 1-standard-deviation change in torque.

For both models, there was a significant interaction between normalized torque and side tested (mean EMG: $p < 0.01$, ES = 0.13; max EMG: $p < 0.01$, ES = 0.14). For the dominant arm, as torque increased by .1 Nm/kg, mean normalized EMG decreased by 3.0% ($p = 0.88$, ES = decrease by 3.2%) and maximum normalized EMG decreased by 7.3% ($p = 0.70$, ES = decrease by 7.9%). For the non-dominant arm, as torque increased by 0.1 Nm/kg, mean normalized EMG increased by 55% ($p < 0.01$, ES = increase by 62%) and max normalized EMG increased by 50% ($p = 0.01$, ES = increase by 56%).

For mean and maximum EMG, the rate of change in the non-dominant arm compared to the dominant arm is 1.60 ($p < 0.01$) and 1.62 ($p < 0.01$), respectively.

Biceps activation during eccentric external rotation was side-dependent. In the dominant arm, greater eccentric rotation torque was not associated with higher biceps EMG, whereas in the non-dominant arm, biceps EMG increased with torque. While the reason for the difference in sides is unknown, perhaps the dominant arm has better motor control than the non-dominant arm and can therefore better isolate activation of the external rotators, minimzing recruitment of the biceps. These findings provide an initial, clinically feasible approach for assessing biceps co-activation during eccentric shoulder loading.

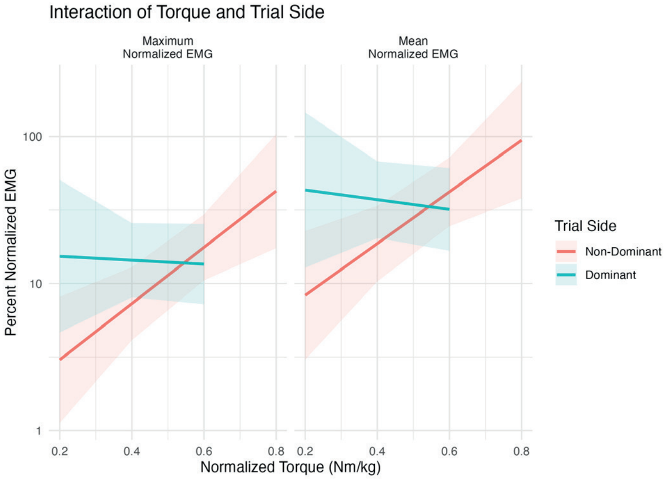

Figure1. Relationship between normalized torque and biceps EMG. EMG is on a log-scale.

4. Conclusions

These results help clinicians interpret EMG and HHD eccentric external rotation by showing that higher torque does not necessarily imply greater biceps activation in the dominant throwing arm, and that side-specific references may be needed. Further, while clinical assumptions that the biceps muscle is active during the follow-through of the throwing motion may be true, it is not necessarily a function of high shoulder eccentric external rotation torque production.

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Conflict of Interest Statement

None.

Contributor Roles

JW: Conceptualization, methodology, Data collection and analysis writing original draft; AB, OR: conceptualization; MR: conceptualization, data collection; JS data collection, methodology, supervision, writing – review & editing.

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