MODELLING OF SCAPULAR MOTION INFLUENCES THE ESTIMATED GLENOHUMERAL JOINT TORQUES DURING VOLLEYBALL HITTING

Dhruv Gupta[1], Lotte Steynen[1], Gaelle Pano[1], Aimilia Rousaki[1], Nele Heylen[1], and Gert Vande Broek[1]

Department of Movement Sciences, KU Leuven, Leuven, Belgium[1]

This study aims to understand how much different ways of modelling the scapular motion in musculoskeletal models influence the estimated inverse dynamics torques at the glenohumeral joint in simulations of volleyball hitting. We collected hitting motion data of one advanced volleyball player and compared the scapular motion and inverse dynamics torques at the glenohumeral joint estimated using three different musculoskeletal models that model the scapular motion in different ways. We found root mean squared differences of up to 50 Nm in inverse dynamics torques estimated from the different models and coupling between scapula and humerus can affect these estimates. Highest marker tracking errors were observed in the model with lowest number of independent scapular degrees of freedom suggesting scapular degrees of freedom should be carefully chosen.

KEYWORDS: scapula, shoulder injury, simulation, biomechanics

INTRODUCTION: Shoulder injuries, primarily overuse injuries, lead to the highest absence from sport for volleyball players (6.29 $\pm$ 9.4 weeks) (Verhagen et al., 2004). Each hit involves high acceleration of the arm during the power generating phase, a high-speed collision with the ball, and rapid deceleration of the fast-moving arm (Gupta et al., 2021a), putting tremendous loads on the glenohumeral joint, the joint that connects the scapula to the humerus. Because the shoulder joint is highly complex, these injuries manifest in different ways, such as rotator cuff muscle/tendon injuries, suprascapular neuropathy, and impingement injuries (Seminati & Minetti, 2013). One reason for the diversity in injuries could be the variety of loads at the glenohumeral joint caused by rapid acceleration, collision, and deceleration. Another could be the variety of hitting techniques involved in volleyball hitting. Hence, it is crucial to study shoulder loads in different hitting phases and different hitting conditions.

Musculoskeletal modelling and simulation methods provide us with a non-invasive method to estimate the loads on the shoulder. An accurate estimation of joint loads using these methods depend on how well the musculoskeletal model captures the complexity of human motion. There are multiple musculoskeletal models of the upper limbs (Saul et al., 2015; McFarland et al., 2019; Seth et al., 2019, Wu et al., 2016). The primary difference between them is the number of independent degrees of freedom they use to model the motion of the scapula, which can be a critical element in accurately estimating the loads at the glenohumeral joint. The performance of these models has never been compared before for fast movements. This study aims to understand how much the difference in modelling the scapular motion influences the estimated shoulder joint loads for volleyball hitting. For one participant, we compared error in tracking of markers on the scapula and the inverse dynamics torques at the glenohumeral joint estimated by three different musculoskeletal models during volleyball hitting. We compared the model estimates for different phases of volleyball hitting under different hitting conditions.

METHODS: Participant and Laboratory Setup: We collected volleyball hitting data of one university-level male player (71 kg, 1.805 m, right-handed outside hitter) under three different hitting conditions. The study was approved by the Privacy and Ethics Assessment (PRET) of KU Leuven (study G-2023-6679). The data were collected in the Movement Analyses Laboratory of Leuven at KU Leuven. To simulate game conditions, we placed a volleyball net at the standard height. We built a mechanism where the ball can be placed at any height above the net within a meter distance from the net. The ball was held in place by two foam pads on either side. The ball was gently placed between the foam pads, allowing for minimal compression from the foam pads. The ball was connected to a light bungee to avoid damage to the motion capture cameras. To ensure that the bungee did not interfere with the hitting motion, we set up a system such that the bungee only engaged once the ball had already travelled at least 0.3 m after being hit. The ball was placed at the participant's preferred height and distance from the net. The location of the ball was kept constant for all hitting trials.

The participant wore the Vicon full-body Plug-In Gait template. Additional retro-reflective markers were placed on the hitting arm (three on the hand, four on the forearm, and two on the upper arm). To capture the movement of the scapula, we placed a 3-marker cluster on the acromion process, where two markers sit on the anterior and posterior ends of the acromion process, and one is placed on a 31 mm vertical stick in between them (Wells et al., 2017). Similar methods have been used to capture scapular motion (Richardson et al., 2016). To capture the motion of the ball, we placed eight markers on it on the side facing away from the participant. Marker data was captured using 10 Vicon MX cameras and 3 Vicon Vero cameras at 250 Hz. The lab allowed for up to six meters of approach distance before the ball was hit.

Experimental protocol: The participant performed a warmup of 5 minutes of running at a self-selected speed on a treadmill, followed by five maximal reach jumps and his preferred pre-game warm-up (Gupta et al., 2021b). We collected hitting data in three different hitting conditions that are most common in volleyball hitting. First is the forward condition, where the participant hits in the direction of approach. Second is lateral condition, where the participant hits to the right (away from the shoulder). Third is the medial condition, where the participant hits to the left (over the shoulder). To ensure that the foam pads do not block the ball's movement in the medial and lateral conditions, the whole ball-holding mechanism was rotated along the vertical axis such that the intended direction of the ball's movement is parallel to the surfaces of the pads. We collected three trials in each hitting condition.

Musculoskeletal modelling and simulations: We selected three musculoskeletal models based on the different ways that they model the scapular motion and free availability in the open-source software OpenSim. In the first model, the Upper Extremity Dynamic Model (UEDM), the motion of the scapula is coupled to the motion of the humerus (Saul et al., 2015; McFarland et al., 2019). This model provides no independent motion to the scapula. The second model, the Scapulothoracic Joint Model (STJM), restricts the scapula to move over an ellipsoid that represents the thorax (Seth et al., 2019). This model has four degrees of freedom for the scapula: elevation, abduction over the thoracic ellipsoid, rotation of the scapula about the axis perpendicular to the thoracic ellipsoid, and rotation of the scapula that lifts the medial order off the thoracic ellipsoid. The independent movement of the scapula allows this model to capture movements like a shrug (Seth et al., 2019). The third model, Wu's Shoulder Model (WSM), treats the scapula as a rigid body connected to the clavicle by a 3-degree of freedom ball-and-socket joint (Wu et al., 2016). Unlike the STJM, the WSM does not restrict the movement of the scapula over a given surface. In OpenSim 4.3, we scaled the generic models to the size of the participant and ran inverse kinematics followed by inverse dynamics. The weights on marker placement in the scaling step and the marker tracking in the inverse kinematics step were kept constant across models. During the inverse dynamics step, we accounted for the force from the ball due to impact. We applied forces and moments that were equal and opposite to the ones experienced by the ball at the centre of the palm. The forces and moments were estimated based on the mass and kinematics of the ball, while assuming the ball to be a rigid body. A similar method using the same assumption has been used before (Jurkojc et al., 2017).

Analyses: We analysed the data from the start of the power-generating phase, that is, start of forward motion of elbow to hit the ball, until the end of the follow-through phase, that is, 30 ms after peak shoulder torque (Gupta et al., 2021a). To assess the accuracy of the motion of the scapula, we computed the error in the positions of the three markers on the acromion cluster. We computed the root mean square error (RMSE) in the marker locations for each model in each hitting condition across the markers and trials. We computed the resultant of inverse dynamics torque estimates along the humeral degrees of freedom. Glenohumeral joint centre serves as the origin of the humerus reference frame. We compared the torque from WSM to those from UEDM and STJM. We computed RMSE and cross-correlation (R^2) values for each hitting condition and for each phase of movement across trials. We used WSM as reference only because it gave the smallest marker location error. This does not imply that it is the most accurate. Since we analysed only one participant's data, we ran no statistics.

RESULTS: We found that the errors in the acromion marker positions were the highest with the UEDM model. The error was lower in the follow-through phase compared to the power-generating phase (Figure 1). We also found differences in the estimated inverse dynamics glenohumeral joint torques among the three models (Figure 2). Table 1 reports the RMSE and R^2 values comparing the inverse dynamics estimates from the WSM to the other models.

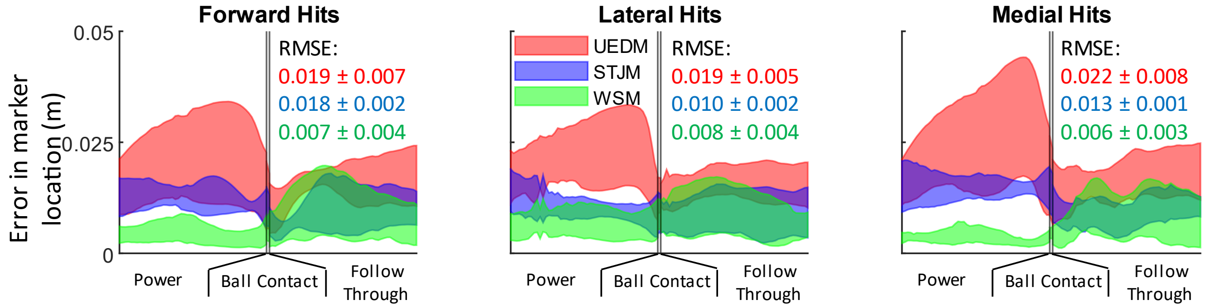

Figure 1: Mean ± standard deviation (SD) error in acromion marker locations compared to experimental data. RMSE is reported across hitting phases, trials, and acromion cluster markers.

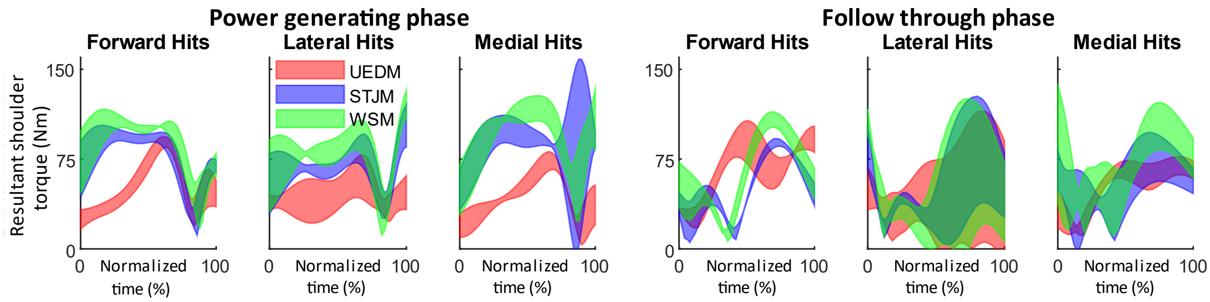

Figure 2: Mean ± SD inverse dynamics torques during power and follow-through phases.

Table 1: RMSE and R^2 values comparing resultant inverse dynamics torque estimated using WSM to UEDM and STJM. R^2 was not computed for ball contact phase since the phase is too short.

		Power generating phase			Ball contact phase			Follow through phase		
		Forward	Lateral	Medial	Forward	Lateral	Medial	Forward	Lateral	Medial
WSM vs UEDM	RMSE (Nm)	44 ± 4	38 ± 7	54 ± 7	49 ± 32	42 ± 21	90 ± 35	35 ± 4	21 ± 4	31 ± 7
	R^2	0.93 ± 0.02	0.97 ± 0.01	0.95 ± 0.01				0.88 ± 0.03	0.89 ± 0.07	0.94 ± 0.03
WSM vs STJM	RMSE (Nm)	13 ± 2	12 ± 3	26 ± 12	16 ± 8	39 ± 38	29 ± 3	19 ± 5	15 ± 1	25 ± 3
	R^2	0.99 ± 0.00	0.99 ± 0.00	0.96 ± 0.04				0.98 ± 0.01	0.95 ± 0.03	0.95 ± 0.04

DISCUSSION: We found that the way the scapular motion is modelled influences the estimated glenohumeral joint torques during volleyball hitting in all hitting phases and hitting conditions. We found that the torques estimated from the STJM and WSM were more similar in shape and magnitude, compared to those estimated from the UEDM (Figure 2 & Table 1). The differences between torques estimated from UEDM and WSM were in the order of 50 Nm, especially in the power-generating and ball contact phases (Table 1). We think that the differences stem from the coupling between the scapular and humeral degrees of freedom in UEDM. The coupling changes the interpretation of the glenohumeral joint torque. In STJM and WSM the torques about the humeral degrees of freedom only move the humerus keeping the glenohumeral joint centre location fixed. However in UEDM, the humeral movement is accompanied by scapular movement which changes the location of the glenohumeral joint centre itself, hence moving the reference frame itself about which the torques are computed.

The UEDM also registered the highest acromion marker tracking error (Figure 1), exceeding the recommended 2 cm threshold (simtk.org). Limiting the degrees of freedom of the scapula limits its ability to get into certain positions, which can lead to a higher marker tracking error. However, marker tracking accuracy should not be used as the only criteria to judge a model's accuracy since skin artifacts influence marker locations. The contraction of the deltoids could cause additional artifacts, especially at high shoulder elevations. Indeed, we observed higher marker error for the UEDM in the power phase where the shoulder elevation is high (Figure 1). Moreover, physiologically, we know that the scapula moves over the thorax, and WSM may overfit the acromion marker cluster trajectories as it has the freedom to ignore this constraint.

Since there exist differences in the interpretation of glenohumeral joint torques between models and marker tracking errors are not the most reliable method to judge model performance, we propose running static optimization and comparing the estimated muscle activity to the measured muscle activity (EMG) to further validate the models for volleyball hitting. Additionally, comparing glenohumeral joint reaction forces between models will be more consistent rather than glenohumeral joint torques.

CONCLUSION: This study found that the level of complexity in the way the scapular motion is modelled affects the estimated inverse dynamics of glenohumeral joint torques. Additionally, having independent (STJM & WSM) versus coupled (UEDM) scapular degrees of freedom affects the estimated glenohumeral joint torques. Our results indicate that the number of independent and dependent scapular degrees of freedom should be carefully chosen and the results should be appropriately interpreted while simulating fast movements.

REFERENCES

Gupta, D., Donnelly, C. J., Jensen, J. L., & Reinbolt, J. A. (2021a). Goal-Oriented optimization of dynamic simulations to find a balance between performance enhancement and injury prevention during volleyball spiking. *Life*, *11*(7), 598.

Gupta, D., Jensen, J. L., & Abraham, L. D. (2021b). Biomechanics of hang-time in volleyball spike jumps. *Journal of Biomechanics*, *121*, 110380.

Jurkojć, J., Michnik, R., & Czapla, K. (2017). Mathematical modelling as a tool to assessment of loads in volleyball player's shoulder joint during spike. *Journal of sports sciences*, *35*(12), 1179-1186.

McFarland, D. C., Brynildsen, A. G., & Saul, K. R. (2020). Sensitivity of neuromechanical predictions to choice of glenohumeral stability modeling approach. *Journal of Applied Biomechanics*, *36*(4), 249-258.

Richardson, R. T., Nicholson, K. F., Rapp, E. A., Johnston, T. E., & Richards, J. G. (2016). A comparison of acromion marker cluster calibration methods for estimating scapular kinematics during upper extremity ergometry. *Journal of biomechanics*, *49*(7), 1255-1258.

Saul, K. R., Hu, X., Goehler, C. M., Vidt, M. E., Daly, M., Velisar, A., & Murray, W. M. (2015). Benchmarking of dynamic simulation predictions in two software platforms using an upper limb musculoskeletal model. *Computer methods in biomechanics and biomedical engineering*, *18*(13), 1445-1458.

Seminati, E., & Minetti, A. E. (2013). Overuse in volleyball training/practice: A review on shoulder and spine-related injuries. *European journal of sport science*, *13*(6), 732-743.

Seth, A., Dong, M., Matias, R., & Delp, S. (2019). Muscle contributions to upper-extremity movement and work from a musculoskeletal model of the human shoulder. *Frontiers in neurorobotics*, *13*, 90.

Verhagen, E. A. L. M., Van der Beek, A. J., Bouter, L. M., Bahr, R. M., & Van Mechelen, W. (2004). A one season prospective cohort study of volleyball injuries. *British journal of sports medicine*, *38*(4), 477-481.

Wells, D. J. M., Alderson, J. A., Dunne, J., Elliott, B. C., & Donnelly, C. J. (2017). Prescribing joint co-ordinates during model preparation to improve inverse kinematic estimates of elbow joint angles. *Journal of biomechanics*, *51*, 111-117.

Wu, W., Lee, P. V., Bryant, A. L., Galea, M., & Ackland, D. C. (2016). Subject-specific musculoskeletal modeling in the evaluation of shoulder muscle and joint function. *Journal of biomechanics*, *49*(15), 3626-3634.

ACKNOWLEDGEMENTS: The authors are grateful to Dr. Friedl De Groote for her inputs, Peter Smolders for his assistance in the laboratory, and OpenAIRE ID – 101068850 funding.