Influence of biomechanical models on joint kinematics and kinetics in baseball pitching

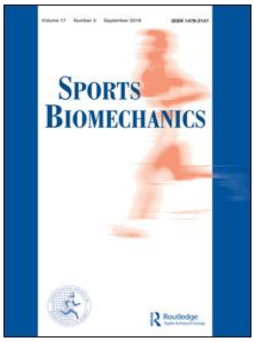
Gasparutto, Xavier; van der Graaff, Erik; van der Helm, Frans C.T.; Veeger, Dirkjan H.E.J.

DOI
10.1080/14763141.2018.1523453

Publication date
2018

Document Version
Final published version

Published in
Sports Biomechanics



Routledge
Taylor & Francis Group

Sports Biomechanics

ISSN: 1476-3141 (Print) 1752-6116 (Online) Journal homepage: http://www.tandfonline.com/loi/rspb20

Influence of biomechanical models on joint kinematics and kinetics in baseball pitching

Xavier Gasparutto, Erik van der Graaff, Frans C. T. van der Helm & Dirkjan H. E. J. Veeger

To cite this article: Xavier Gasparutto, Erik van der Graaff, Frans C. T. van der Helm & Dirkjan H. E. J. Veeger (2018): Influence of biomechanical models on joint kinematics and kinetics in baseball pitching, Sports Biomechanics, DOI: 10.1080/14763141.2018.1523453

To link to this article: https://doi.org/10.1080/14763141.2018.1523453

SPORTS BIOMECHANICS
https://doi.org/10.1080/14763141.2018.1523453

ARTICLE

OPEN ACCESS

Influence of biomechanical models on joint kinematics and kinetics in baseball pitching

Xavier Gasparutto[a], Erik van der Graaff[a,b], Frans C. T. van der Helm[a] and Dirkjan H. E. J. Veeger[a,b]

[a]Department of BioMechanical Engineering, Delft University of Technology, Delft, The Netherlands; [b]Department of Human Movement Sciences, Vrije Universiteit Amsterdam, Amsterdam, The Netherlands

ABSTRACT
In baseball pitching, biomechanical parameters have been linked to ball velocity and potential injury risk. However, although the features of a biomechanical model have a significant influence on the kinematics and kinetics of a motion, this influence have not been assessed for pitching. The aim of this study was to evaluate the choice of the trunk and shoulder features, by comparing two models using the same input. The models differed in thoraco-humeral joint definition (moving or fixed with the thorax), joint centre estimation, values of the inertial parameters and computational framework. One professional pitcher participated in the study. We found that the different features of the biomechanical models have a substantial influence on the kinematics and kinetics of the pitchers. With a fixed thoraco-humeral joint the peak average thorax angular velocity was delayed and underestimated by 17% and the shoulder internal rotation velocity was overestimated by 7%. The use of a thoraco-humeral joint fixed to the thorax will lead to an overestimation of the rotational power at the shoulder and will neglect the power produced by the forward and upward translation of the shoulder girdle. These findings have direct implications for the interpretation of shoulder muscle contributions to the pitch.

ARTICLE HISTORY
Received 27 July 2017
Accepted 3 September 2018

KEYWORDS
Inverse dynamics; modelling; overhand throw; shoulder; trunk

Introduction

Baseball pitching is one of the most studied motions in sport biomechanics, with many studies aiming to identify the biomechanical variables that influence performance and the risk of injury (Fortenbaugh, Fleisig, & Andrews, 2009; Oyama, 2012; Weber, Kontaxis, Brien, & Bedi, 2014; Whiteley, 2007). However, these studies did not all use the same biomechanical model. Different studies might have important differences in their biomechanical models, and so produce substantial differences in the estimated kinematics and kinetics of a given motion.

For a biomechanical model of pitching, we identified four important features (1) the thoraco-humeral joint model that can be either fixed with the thorax if the thorax is defined by the hip joint centres and acromio-clavicular joints (Aguinaldo & Chambers,

CONTACT Dirkjan H. E. J. Veeger h.e.j.veeger@tudelft.nl
Supplemental data for this article can be accessed here.

2009; Fleisig, 1994; Naito, Takagi, & Maruyama, 2011; Roach & Lieberman, 2014) or moving with respect to the thorax if the thorax is defined by thoracic markers (Gasparutto, Van Der Graaff, Van Der Helm, & Veeger, 2016; Naito, Takagi, Yamada, Hashimoto, & Maruyama, 2014; Takagi et al., 2014), (2) the body segment inertial parameters (Ae, Tang, & Yokoi, 1992; Clauser, Mc Conville, & Young, 1969; Dempster, 1955; Dumas, Cheze, & Verriest, 2007), (3) the joint centres estimation (Ae et al., 1992; Dempster, 1955; Dillman, Fleisig, & Andrews, 1993; Dumas et al., 2007) and (4) the computational framework (Feltner & Dapena, 1986; Gasparutto et al., 2016; Naito & Maruyama, 2008). In studies of human gait, the features of a biomechanical model are known to affect the estimation of the kinematics and kinetics of the lower limb (Dumas, Nicol, & Chèze, 2007; Pearsall & Costigan, 1999; Rao, Amarantini, Berton, & Favier, 2006; Reinbolt, Haftka, Chmielewski, & Fregly, 2007; Stagni, Leardini, Cappozzo, Benedetti, & Cappello, 2000). To our knowledge, the influence of the features of a biomechanical model has not been quantified for baseball pitching, and so the consistency of the results obtained from the various models used in the literature has not been verified. It is reasonable to assume that modelling assumptions will also have a significant influence on the kinematics and kinetics for a highly dynamic motion as pitching. What is at stake is that significant differences in the estimated kinematics and kinetics obtained by different models for the same motion could result in contradictory conclusions, specifically for studies using correlations between kinematic and kinetic parameters. This could lead to incorrect and even potentially harmful recommendations to the coaches and pitchers. In addition, contradictory observations between two different studies could be due to different modelling assumptions more than differences in the pitching motion itself. To gain confidence in the recommendations from the scientific community, it is necessary to understand the influence of the choice for different features. Therefore, the aim of this study is to evaluate quantitatively the effect of the trunk and shoulder features by comparing the kinematics and kinetics obtained by multiple models for the same input pitching motion.

Two different models were selected. The first model was previously developed by the authors of the present study (Gasparutto et al., 2016) and based on the work of Dumas et al. (Dumas et al., 2007). The second model was developed by Fleisig et al. (Fleisig, 1994; Zheng, Fleisig, Barrentine, & Andrews, 2004) and is one of the most used models in biomechanical studies of baseball pitching. These two models use different regression equations to determine the inertial parameters and joint centres and have different mechanical frameworks but the main difference concerns the definition of the thorax segment and shoulder joint model. The Gasparutto model estimates the thorax motion with markers on the thorax only whilst the Fleisig model uses the estimated shoulder joint centres and hip joint centres to represent the thorax. As a consequence the Fleisig model merges the scapular girdle motion with the thorax motion, whereas, the Gasparutto model separates between the thorax motion and the displacement of the scapular girdle relative to the thorax. This leads to the following limitations: (a) if the thorax is flexed but the shoulders stay above the hips, the Fleisig model will not capture this flexion, (b) if the scapular girdle is moved forward, the Fleisig model will interpret that motion as an axial rotation of the thorax. Although this feature is bound to have an effect on thorax and arm kinematics, its influence was never assessed previously. Based on the limitations of the Fleisig model, we hypothesise that the merged thoraco-humeral joint model will show reduced thorax flexion and tilt and increased axial rotation leading to increased shoulder angles, angular velocities and actions when compared to the moving thoraco-humeral joint model developed by Gasparutto et al. (2016).

Methods

One professional right-handed baseball pitcher, with 5 years of MLB experience, participated in the study (height: 1.98 m, weight: 101.2 kg). After having been informed of the aims and procedures of the experiment, the player signed an informed consent form. The Faculty of Human Movement Sciences' local ethical committee approved this research project.

Equipment

A motion capture system consisting of eight motion capture beams was used to track active skin markers (4 Optotrack Certus, 4 Optotrack 3020, Northern Digital Inc., Waterloo, Ontario, Canada). The pitching mound consisted of a two-part wooden pitching mound that was taped to two forceplates (Vrije Universiteit, Amsterdam, The Netherlands,1.08 x 1.08 m, 200 Hz, (Ibrahim, Faber, Kingma, & van Dieën, 2017)). The standing part of the pitching mound included a pitching rubber and the stepping part had a downward slope of 5.5 degrees as recommended in the Major League Baseball regulations. A high-speed camera (Casio EX-ZR 1000, Casio Computer CO., LTD., Tokyo, Japan) was used to film the mound from the right lateral side at a frame rate of 240 Hz.

A net with a rectangular pitching target was placed at 10 m from the mound. As the strike zone is at 18 m during a game, the target was scaled to the usual strike zone. The pitches were marked 'strike' if they reached the target and 'ball' if they were out of the target. A speed gun (Stalker Pro II Speed Sensor Radar, Applied Concepts, Inc./ Stalker Radar, Plano, Texas, United States) placed behind the net was used to record the maximal ball speed of every pitch in miles per hour (mph).

Measurement procedure

The pitcher had a one-hour warm-up with his physical trainer, as in game conditions. The pitcher was then equipped with 24 active markers; 18 were placed on anatomical points on the head, upper limbs and lower limbs and 2 clusters of 3 active markers on the thorax and pelvis respectively.

Once equipped with markers, the pitcher performed as many warm-up throws as he wished from the pitching mound. When ready, the pitcher performed from the mound three fastballs with six active markers on the upper limb and then eight fastballs with 24 active markers on the full body. The acquisition was done at the maximal acquisition frequency: 170 Hz with six markers and 90 Hz with 24 markers.

The pitcher was instructed to throw as fast as he could while attempting to hit the pitching target. After the throwing procedure, eight anatomical points of the pelvis and thorax were acquired with a pointer. The main markers used in this study are depicted on Figure 1.

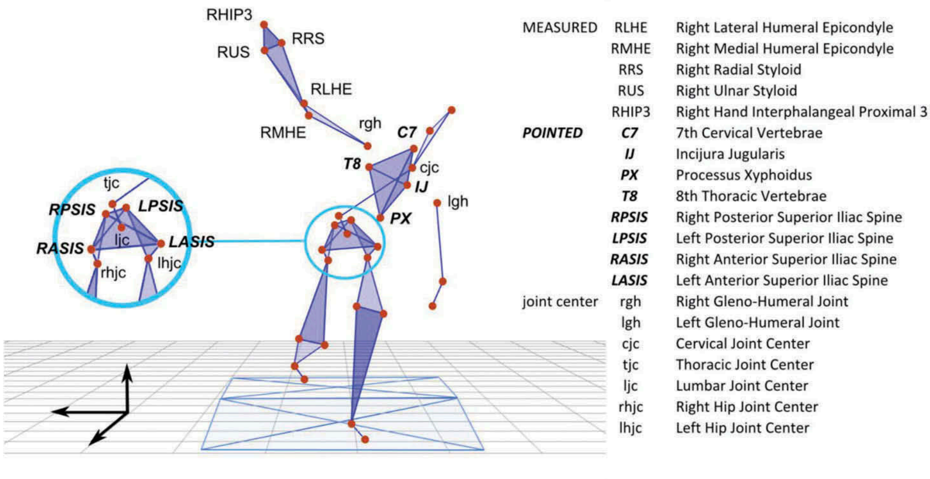


Figure 1. Markers and optimised length (blue lines).

Data processing

Data processing was done with Matlab2014a (The MathWorks, Inc., Natick, Massachusetts, United States) and with the use of Mokka and of the BTK Matlab API (Barre & Armand, 2014). The five fastest strikes were used for the analysis.

Marker trajectories and ground reaction forces

The marker trajectories were interpolated and upsampled to 170 Hz with a validated upsampling method (see Supplementary Materials). The upsampled marker trajectories were filtered with a fourth order Butterworth low-pass filter with a cutting frequency of 12.5 Hz during the stride and follow-through and with a cutting frequency of 25 Hz during the arm cocking, the arm acceleration and the arm deceleration phases.

The thorax and pelvis anatomical points were reconstructed based on the pointing procedure and the pelvis and thorax cluster motion. The joint centres (shoulder, elbow, wrist, hip) were defined with regression equations (Dumas, Cheze et al., 2007). To avoid inconsistencies in segment length due to the soft tissue artefacts and the interpolation of the trajectories, a quasi-static multibody optimisation (Lu & O'Connor, 1999), detailed in the Supplementary Materials, was performed with the 'fmincon' function of Matlab2014a.

The synchronised ground reaction force was downsampled to 170 Hz and corrected for the mound weight. The time of foot contact was defined as the time when the ground reaction force under the stride foot was larger than 10 N.

The synchronised high speed camera was used to identify the time of ball release. It was defined as the first frame when the ball is not visibly touched by the hand.

Models

The Fleisig model and the Gasparutto model differed in four features: (a) the thoraco-humeral joint that was fixed or moving, (b) the inertial parameters based on (Dempster,

1955) or (Dumas, Cheze et al., 2007), (c) the joint centres estimation based on (Dillman et al., 1993) or (Dumas, Cheze et al., 2007) and the mechanical framework based on (Feltner & Dapena, 1986) or (Dumas, Aissaoui, & de Guise, 2004; Gasparutto et al., 2016). These points are summarised in Table 1 and details of the thoraco-humeral joint models and mechanical framework are given below.

Thoraco-humeral joint

The Gasparutto model features a 'moving' thoraco-humeral joint. The thorax is defined by thorax markers (deepest point of Incisura Jugularis, Processus Xyphoideus, Processus Spinosus of the 7th cervical vertebrae, Processus Spinosus of the 8th thoracic vertebrae, see Figure 1) according to the recommendation of the International Society of Biomechanics (ISB) (Wu et al., 2005) and the scapular girdle motion is modelled by allowing the gleno-humeral joint (GH) to translate with respect to the thorax in three directions (forward/backward, upward/downward, lateral/medial). The Fleisig model features a 'fixed' thoraco-humeral joint. The thorax is defined by the right and left hip joint centres and the right and left GH. The GH displacements are defined with respect to the midpoint of the right and left GH and are limited to the lateral-medial direction. A forward/backward motion of the scapular girdle will be interpreted as an axial rotation of the thorax. Likewise an upward/downward motion of the scapular girdle will be interpreted as a lateral tilt of the thorax. The geometric differences between the Gasparutto model and the Fleisig model are depicted in Figure 2.

Kinematics and kinetics

The joint angles represent the angles of the distal segment with respect to the proximal one and the net joint forces and moments represent the net forces and moments of the proximal segment on the distal segment. When needed, the sign of the angles from the Fleisig model and the zero angle position of the joints were modified to match the recommendation of the ISB and be compatible with the other models. The displacements and velocities of GH in the thorax were computed with respect to the cervical joint centre for the Gasparutto model and with respect to the mid-point of the right and left GH for the Fleisig model (1994).

Table 1. Overview of the biomechanical features of the models.

	Thoraco-Humeral Joint	Body Segment Inertial Parameters	Joint Centres	Mechanical Framework
Main Models				
Fleisig	Lumped (Fleisig, 1994)	(Dempster, 1955)	(Dillman et al., 1993)	(Feltner & Dapena, 1986)
Gasparutto	Moving (Gasparutto et al., 2016)	(Dumas, Cheze et al., 2007)	(Dumas, Cheze et al., 2007)	(Dumas et al., 2004; Gasparutto et al., 2016)
Intermediate Models				
Fixed	Lumped (Fleisig, 1994)	(Dumas, Cheze et al., 2007)	(Dumas, Cheze et al., 2007)	(Dumas et al., 2004; Gasparutto et al., 2016)
Inertial	Moving (Gasparutto et al., 2016)	(Dempster, 1955)	(Dumas, Cheze et al., 2007)	(Dumas et al., 2004; Gasparutto et al., 2016)
Joint Centre	Moving (Gasparutto et al., 2016)	(Dumas, Cheze et al., 2007)	(Dillman et al., 1993)	(Dumas et al., 2004; Gasparutto et al., 2016)
Framework	Lumped (Fleisig, 1994)	(Dempster, 1955)	(Dillman et al., 1993)	(Dumas et al., 2004; Gasparutto et al., 2016)

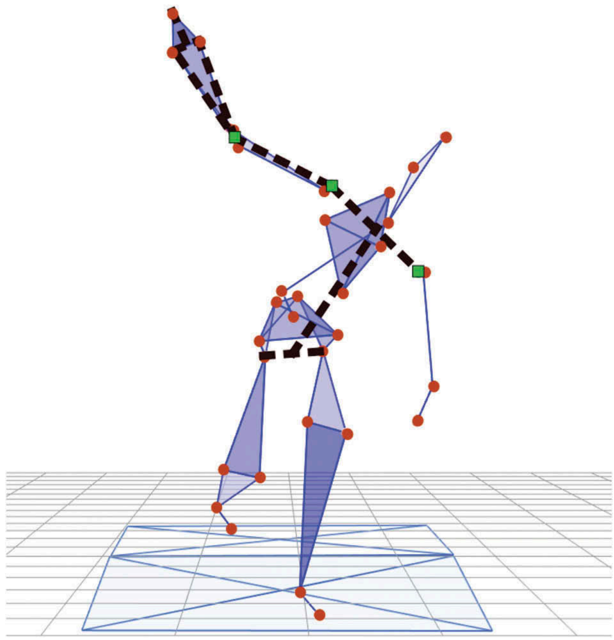


Figure 2. Geometric models of the Gasparutto model (blue mesh, spherical markers) and of the Fleisig model (black, dashed lines, square markers) at ball release.

The joint kinetics were estimated by the inverse dynamics methods described by Dumas et al. (2004) for the Gasparutto model and by the inverse dynamics method described by Zheng et al. (2004) for the Fleisig model. The net joint forces and moments were then projected in the corresponding joint coordinate systems as explained in Gasparutto et al. (2016).

Intermediate models

To identify the influence of each feature separately, four intermediate models were built. Three intermediate models were the Gasparutto model with one biomechanical feature of the Fleisig model. The last intermediate model was the Fleisig model implemented in the mechanical framework of the Gasparutto model. Thus, by comparing these models to the Gasparutto model we could identify the influence of each feature separately. The features of the intermediate models are detailed in Table 1.

The 'fixed' intermediate model was the Gasparutto model with the thoraco-humeral joint of the Fleisig model, the 'inertial' intermediate model was the Gasparutto model with the inertial parameters from the Fleisig model, the 'joint' intermediate model was the Gasparutto model with the joint centres from the Fleisig model and the 'framework' intermediate model was the Fleisig model expressed within the Gasparutto computational framework, that is the Gasparutto model with the joint centres, inertial parameters and thorax from the Fleisig model. These definitions are regrouped in Table 1.

Model comparison

The joint kinematics and joint kinetics estimated by the six models were compared for parameters relevant to baseball pitching (Fortenbaugh et al., 2009).

The selected kinematic parameters were: the shoulder horizontal abduction angle at foot contact, the shoulder maximal external rotation angle, the elbow flexion angle at maximal shoulder external rotation, the peak norm of the thorax angular velocity, the shoulder peak internal rotation velocity and the elbow peak extension velocity. The time series of the thorax angles, thorax angular velocities, GH translation and GH linear velocities were also compared and reported as they represent the main difference between the two selected models.

Regarding the kinetics, the selected parameters were the peak shoulder pulling force, the peak of the norm of the elbow net force, the peak of the norm of the shoulder net force, the peak elbow adduction moment, the peak shoulder external-rotation moment and the peak shoulder internal-rotation moment.

The mean difference between the parameters from the Gasparutto model and the Fleisig and intermediate models were computed as well as the timing difference for the peak values. A two-tailed paired *t*-test was performed to account for the statistical significance of the differences.

Results

The mean ball velocity was 39.7 ± 0.2 m/s (88.8 ± 0.4 mph).

GH displacements (Figure 4)

The 'moving' thoraco-humeral models showed a backward displacement during the arm cocking phase. This was followed by a forward motion during the arm acceleration phase with a peak forward velocity of 1.5 m/s at ball release for the Gasparutto model and a peak forward velocity of 1.9 m/s between ball release and maximal internal rotation for the 'joint centre' intermediate model. GH continued with a forward motion during the follow-through. Regarding the upward motion, the 'moving' thoraco-humeral models showed small variations in position but an upward velocity during the acceleration and deceleration phases with a peak of 0.4 m/s and 0.7 m/s at ball release for the Gasparutto model and 'joint centre' intermediate model respectively. By definition, the fixed thoraco-humeral models showed no displacement and no velocity for these two degrees-of-freedom. All models presented some lateral-medial displacement and velocity.

Model comparison (Figure 3, Table 2)

When compared to the Gasparutto model the Fleisig model showed increased shoulder horizontal abduction at foot contact, reduced thorax angular velocity, increased elbow extension velocity, and increased elbow forces, elbow adduction moment and shoulder internal rotation moment. The peak thorax angular velocity and peak elbow extension velocity were also delayed by 15 ms and 11 ms respectively.

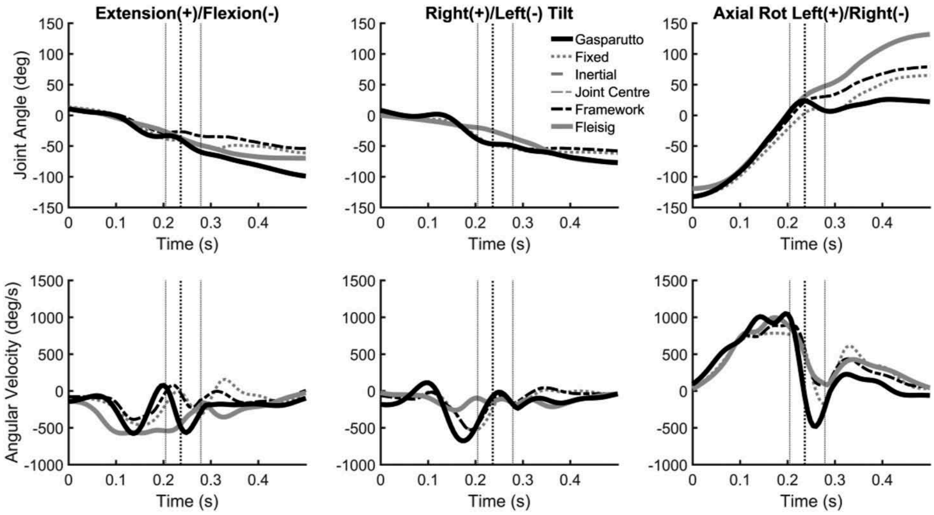


Figure 3. Thorax angles and angular velocities with respect to the global frame, t = 0 is foot contact and the vertical lines indicates maximal external rotation, ball release and maximal internal rotation.

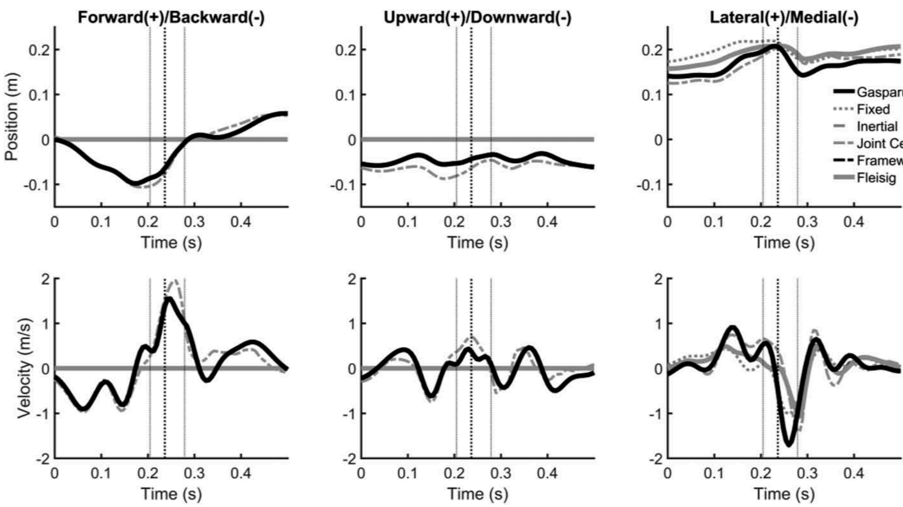


Figure 4. Gleno-humeral joint position and velocities with respect to the thorax, $t = 0$ is foot contact and the vertical lines indicates maximal external rotation, ball release and maximal internal rotation.

Based on the 'fixed' intermediate model results, the thoraco-humeral feature was found responsible for the reduced thorax angular velocity and showed as well an increased shoulder internal rotation velocity. The elbow extension velocity was mainly influenced by the joint centre estimation.

The net joint forces and moments were influenced by the inertial parameters and the joint centres but the largest influence was found for the mechanical framework feature.

Table 2. Mean parameters of the Gasparutto model and differences with the Fleisig model and intermediate models. Only the significant differences were reported in table ($p < 0.05$). FC stands for the Foot Contact event, MER stands for the event 'shoulder Maximal External Rotation', 't' is the time at which the considered parameter occurs, 'Δt' is the time difference in milliseconds between the models, 'mean Δ' is the mean difference of the considered parameter.

		Angles (deg)			Angular Velocity (deg/s)			Net Forces (N)			Net Moment (Nm)		
		Shoulder Horizontal. Abduction At FC	*Shoulder Maximal External Rotation*	*Elbow Flexion at MER*	*Peak Norm of Thorax Ang. Vel.*	*Peak Shoulder Internal Rotation*	*Peak Elbow Extension*	*Peak Shoulder Pulling Force*	*Peak Norm of Elbow net force*	*Peak Norm of Shoulder net force*	*Peak Elbow Adduction*	*Peak Shoulder Internal Rotation*	*Peak Shoulder External Rotation*
Results model Gasparutto	*T* (ms)	0	204	204	152	249	237	237	236	238	208	208	258
	Mean	–25	–177	83	1280	3939	–1879	1224	1058	1306	83	83	–36
	SD	±9	±1	±3	±21	±379	±147	±24	±40	±38	±1	±1	±7
Difference with model Gasparutto													
Fleisig	Δt (ms)	0	-	0	15	-	11	-	–4	-	3	3	-
	Mean Δ	12	-	4	–215	-	118	-	162	-	17	21	-
	p value	0.006	-	0.000	0.000	-	0.049	-	0.003	-	0.000	0.000	-
Fixed	Δt (ms)	-	0	-	29	0	-	-	-	-	-	-	-
	Mean Δ	-	5	-	–165	317	-	-	-	-	-	-	-
	p value	-	0.000	-	0.000	0.000	-	-	-	-	-	-	-
Inertial Parameters	Δt (ms)	-	-	-	-	-	-	1	3	1	-	3	–1
	Mean Δ	-	-	-	-	-	-	–27	–34	–50	-	3	–8
	p value	-	-	-	-	-	-	0.009	0.000	0.000	-	0.000	0.000
Joint Centre	Δt (ms)	-	-	0	-	0	–2	-	4	4	1	3	–2
	Mean Δ	-	-	–1	-	6	83	-	58	–47	2	8	–3
	p value	-	-	0.000	-	0.027	0.002	-	0.018	0.029	0.011	0.000	0.008
Mechanical Framework	Δt (ms)	0	0	0	28	–1	0	9	-	6	2	4	–1
	Mean Δ	13	–4	–1	-137	276	77	–43	-	–89	3	10	–10
	p value	0.005	0.003	0.000	0.000	0.000	0.005	0.021	-	0.002	0.002	0.000	0.002

Discussion and implications

This study aimed at understanding the influence of the choice of biomechanical model features on the analysis of pitching, especially for the thorax segment and the thoraco-humeral joint. Two models were compared: the Fleisig model (Fleisig, 1994; Zheng et al., 2004) and the Gasparutto model (Gasparutto et al., 2016). The influence of each feature was evaluated with four intermediate models. Although this paper studied the pitching motion of a skilled professional baseball player, the possibility of idiosyncrasies in the pitching technique of the participant cannot be excluded.

This study showed that there is a non-negligible displacement of GH during the pitch that a fixed thoraco-humeral joint cannot account for. The position of GH with respect to the thorax showed a backward displacement during the cocking phase followed by a peak forward and upward velocity occurring at ball release and a forward displacement during the follow-through, likely to be used to ease the deceleration of the upper limb.

It is important to note that the lateral-medial displacement of GH does not represent a shortening-lengthening of the clavicle but the displacement of GH around an arc of ellipsoid for the moving thoraco-humeral joints and the variation of distance between the right and left GH for the fixed thoraco-humeral joints. The 'fixed' model can only capture lateral-medial displacement of the scapular girdle. However, the lateral/medial position of GH for the fixed thoraco-humeral joint convention is equal to half the distance between the left and right GH and the point of reference to compute that position is moving with respect to the thorax which makes any interpretation difficult. Thus the fixed thoraco-humeral convention is not appropriate for the description of the scapular girdle motion and while not exactly modelling the role of the scapula, a moving thoraco-humeral joint should be preferred to get insight in the scapular girdle motion during pitching.

The assumption in the Fleisig model that the thorax is defined based on the shoulder joint centres and hip joint centres led to reduced thorax flexion angle, tilt angle, and peak thorax rotation velocities and consequently to an increased estimation of the peak shoulder internal rotation velocity. The increased thorax axial rotation after ball release was likely to be due to the forward motion of GH during the arm acceleration phase and follow-through. Indeed, the forward motion of the right GH was interpreted as thorax axial rotation by the fixed thoraco-humeral joint model. It is interesting to see that the timing of the peak thorax angular velocity and of the peak elbow extension velocity was changed. This could lead to significant differences when studying the kinetic chain. Regarding the kinetics, contrary to our initial hypothesis, the thoraco-humeral joint definition did not have any significant influence on the net joint forces and moments. This can be understood by the fact that the equations used to estimate the net moment and force at the shoulder only use the estimation of the GH position in the mound reference frame and not the thorax position and orientation. The thorax orientation was only used during the projection of the shoulder net joint moment and forces in the shoulder joint coordinate system.

A significant difference was found for the peak shoulder internal rotation velocity between the Gasparutto model and the intermediate models with the fixed thoraco-humeral joint but not between the Gasparutto model and the Fleisig model. The intermediate models results suggest that this inconsistency comes from the differences in mechanical framework. The effect of the framework can also be clearly observed on the estimation of the peak elbow adduction moment and peak net joint force. Indeed, significant differences on these peaks were found between the Fleisig model and the Gasparutto model but not between the Gasparutto model and the intermediate model for the mechanical framework feature. As the Fleisig model and this intermediate model are identical apart from the biomechanical framework, the difference in the estimation of the elbow peak between Fleisig and Gasparutto can be explained by the difference of framework. This observation is supported by a previous study (Dumas et al., 2007) showing that the influence of the inverse dynamics method was 'at least of equivalent importance' than other modelling hypotheses.

Conclusion

The influence of the choice of trunk and shoulder features on the kinematics and kinetics of baseball pitching were quantified in this paper. These features regrouped the thoraco-humeral joint model, the joint centre location, the inertial parameters, and the computational framework. By comparing two main models and four intermediate models with one different feature at a time, we showed that all of the features had a significant influence on the kinematics and/or kinetics of the pitcher and we were able to identify the variability associated with each feature.

The Fleisig model is a simple and elegant model that allows for a reasonable estimate of the kinematics and kinetics of the upper limb. However, the use of GH and the hip joint centres to estimate the thorax orientation and translations leads to underestimations of the thorax angular velocity, overestimations of the shoulder internal rotation velocity, delayed timing of the peak thorax and elbow angular velocities and will make it impossible to estimate the GH displacement during the pitch. Thus, it might not be sufficiently detailed to study the shoulder girdle action during pitching and could lead to a large overestimation of the angular powers occurring at the shoulder while neglecting the power due to the forward and upward translation of the shoulder girdle. This has direct implications for the interpretation of shoulder muscle function during the pitch as it could lead as well to an overestimation of the role of the internal rotator of the shoulder in power generation. The 'moving' thoraco-humeral joint model was developed to tackle these issues and gain a deeper knowledge of the shoulder complex actions during pitching.

Acknowledgements

The authors would like to thank for their support Peter Hordijk, the technical team from the Vrije Universiteit Amsterdam (Leon Schutte, Franz-Joseph Halkes, Vincent Tuinder, Siro Otten and Hans Agricola) and Martijn Nijhoff. The authors also thank Dr Fleisig for sharing his PhD thesis.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

This work was supported by the Dutch Technology Foundation (STW) under grant number 12893

ORCID

Erik van der Graaff http://orcid.org/0000-0003-2487-8056
Dirkjan H. E. J. Veeger http://orcid.org/0000-0003-0292-6520

References

Ae, M., Tang, H. P., & Yokoi, T. (1992). Estimation of inertia properties of the body segments in Japanese athletes. *Biomechanisms*, *11*, 23–33. doi:10.3951/biomechanisms.11.23

Aguinaldo, A., & Chambers, H. (2009). Correlation of throwing mechanics with elbow valgus load in adult baseball pitchers. *The American Journal of Sports Medicine*, *37*, 2043–2048. doi:10.1177/0363546508331205

Barre, A., & Armand, S. (2014). Biomechanical toolKit: Open-source framework to visualize and process biomechanical data. *Computer Methods and Programs in Biomedicine*, *114*, 80–87. doi:10.1016/j.cmpb.2014.01.012

Clauser, C., McConville, J., & Young, J. (1969). *Weight, volume, and center of mass of segments of the human body* (Report No AD-710 622). Dayton, US-OH: Air Force Systems command Wright-Patterson Air Force Base.

Dempster, W. T. (1955). *Space requirements of the seated operator* (Report No 55-159). Dayton, US-OH: Wright Air Development Center, Air Research and Development Command, US Air Force, Wright-Patterson Air Force Base.

Dillman, C. J., Fleisig, G. S., & Andrews, J. R. (1993). Biomechanics of pitching with emphasis upon shoulder kinematics. *Journal of Orthopaedics & Sports Physical Therapy*, *18*, 402–408. doi:10.2519/jospt.1993.18.2.402

Dumas, R., Aissaoui, R., & de Guise, J. A. (2004). A 3D generic inverse dynamic method using wrench notation and quaternion algebra. *Computer Methods in Biomechanics and Biomedical Engineering*, *7*, 159–166. doi:10.1080/10255840410001727805

Dumas, R., Cheze, L., & Verriest, J. (2007). Adjustments to McConville et al. and Young et al. body segment inertial parameters. *Journal of Biomechanics*, *40*, 543–553. doi:10.1016/j.jbiomech.2006.02.013

Dumas, R., Nicol, E., & Chèze, L. (2007). Influence of 3D inverse dynamics method on the joint forces and moments during gait. *Journal of Biomedical Engineering*, *129*, 786–790. doi:10.1115/1.2768114

Feltner, M., & Dapena, J. (1986). Dynamics of the shoulder and elbow joints of the throwing arm during a baseball pitch. *International Journal of Sports Biomechanics*, *2*, 235–259. doi:10.1123/ijsb.2.4.235

Fleisig, G. S. (1994). *The biomechanics of baseball pitching* (Doctoral dissertation). The University of Alabama at Birmingham.

Fortenbaugh, D., Fleisig, G. S., & Andrews, J. R. (2009). Baseball pitching biomechanics in relation to injury risk and performance. *Sports Health*, *1*, 314–320. doi:10.1177/1941738109338546

Gasparutto, X., Van Der Graaff, E., Van Der Helm, F. C. T., & Veeger, H. E. J. (2016). Elite athlete motor and loading actions on the upper limb in baseball pitching. *Procedia Engineering*, *147*, 181–185. doi:10.1016/j.proeng.2016.06.210

Ibrahim, R., Faber, G. S., Kingma, I., & van Dieën, J. H. (2017). Kinematic analysis of the drag flick in field hockey. *Sports Biomechanics*, *16*, 45–57. doi:10.1080/14763141.2016.1182207

Lu, T. W., & O'Connor, J. J. (1999). Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. *Journal of Biomechanics*, *32*, 129–134. doi:10.1016/S0021-9290(98)00158-4

Naito, K., & Maruyama, T. (2008). Contributions of the muscular torques and motion-dependent torques to generate rapid elbow extension during overhand baseball pitching. *Sports Engineering*, *11*, 47–56. doi:10.1007/s12283-008-0002-3

Naito, K., Takagi, H., & Maruyama, T. (2011). Mechanical work, efficiency and energy redistribution mechanisms in baseball pitching. *Sports Technology*, *4*, 48–64. doi:10.1080/19346182.2012.686502

Naito, K., Takagi, H., Yamada, N., Hashimoto, S., & Maruyama, T. (2014). Intersegmental dynamics of 3D upper arm and forearm longitudinal axis rotations during baseball pitching. *Human Movement Science*, *38*, 116–132. doi:10.1016/j.humov.2014.08.010

Oyama, S. (2012). Baseball pitching kinematics, joint loads, and injury prevention. *Journal of Sport and Health Science*, *1*, 80–91. doi:10.1016/j.jshs.2012.06.004

Pearsall, D. J., & Costigan, P. A. (1999). The effect of segment parameter error on gait analysis results. *Gait & Posture*, *9*, 173–183. doi:10.1016/S0966-6362(99)00011-9

Rao, G., Amarantini, D., Berton, E., & Favier, D. (2006). Influence of body segments' parameters estimation models on inverse dynamics solutions during gait. *Journal of Biomechanics*, *39*, 1531–1536. doi:10.1016/j.jbiomech.2005.07.032

Reinbolt, J. A., Haftka, R. T., Chmielewski, T. L., & Fregly, B. J. (2007). Are patient-specific joint and inertial parameters necessary for accurate inverse dynamics analyses of gait? *IEEE Transactions on Biomedical Engineering*, *54*, 782–793. doi:10.1109/TBME.2006.889187

Roach, N. T., & Lieberman, D. E. (2014). Upper body contributions to power generation during rapid, overhand throwing in humans. *The Journal of Experimental Biology*, *217*, 2139–2149. doi:10.1242/jeb.107482

Stagni, R., Leardini, A., Cappozzo, A., Benedetti, M. G., & Cappello, A. (2000). Effects of hip joint centre mislocation on gait analysis results. *Journal of Biomechanics*, *33*, 1479–1487. doi:10.1016/S0021-9290(00)00093-2

Takagi, Y., Oi, T., Tanaka, H., Inui, H., Fujioka, H., Tanaka, J., … Nobuhara, K. (2014). Increased horizontal shoulder abduction is associated with an increase in shoulder joint load in baseball pitching. *Journal of Shoulder and Elbow Surgery*, *23*, 1757–1762. doi:10.1016/j.jse.2014.04.001

Weber, A. E., Kontaxis, A., Brien, S. J. O., & Bedi, A. (2014). The biomechanics of throwing: Simplified and cogent. *Sports Medicine and Arthroscopy Review*, *22*, 72–79. doi:10.1097/JSA.0000000000000019

Whiteley, R. (2007). Baseball throwing mechanics as they relate to pathology and performance – a review. *Journal of Sport Science & Medecine*, *6*, 1–20. Retrieved from https://www.ncbi.nlm.nih.gov/pubmed/24149219

Wu, G., van der Helm, F. C. T., DirkJan Veeger, H. E. J., Makhsous, M., Van Roy, P., Anglin, C., … Buchholz, B. (2005). ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: Shoulder, elbow, wrist and hand. *Journal of Biomechanics*, *38*, 981–992. doi:10.1016/j.jbiomech.2004.05.042

Zheng, N., Fleisig, G. S., Barrentine, S., & Andrews, J. R. (2004). Biomechanics of Pitching. In G. K. Hung & J. M. Pallis (Eds.), *Biomedical engineering principles in sports* (pp. 209–256). Boston, MA: Springer US. doi:10.1007/978-1-4419-8887-4_9