THE EFFECT OF MODEL COMPLEXITY ON TRACKING CRICKET FAST BOWLERS USING INVERSE KINEMATICS IN OPENSIM

Peter Sinclair, René Ferdinands, Andrew Greene and Max Stuelcken
Faculty of Health Sciences, The University of Sydney, Sydney, Australia

The purpose of this study was to investigate the effect of model complexity on the measurement of segmental kinematics. Three-dimensional marker coordinates from 13 cricket fast bowlers were used to drive an inverse kinematics routine in OpenSim using a 20-segment, whole-body model with the shoulder joint centres of rotation fixed within a rigid trunk segment. Adding shoulder joint translation to the model improved the fit of marker coordinates through the inverse kinematics routine and significantly changed joint angles throughout the body. Switching the rigid trunk for a model with anatomically appropriate levels of lumbar movement did not further improve the overall accuracy of marker coordinate reconstruction; however the shoulder extension angle was affected by approximately five degrees.

KEY WORDS: cricket, bowling, OpenSim, modelling, kinematics.

INTRODUCTION: Measurement of three dimensional (3D) kinematics requires underlying body movements to be inferred from direct measurement of reflective markers placed onto the skin. This can potentially cause errors in the reconstruction of movement owing to movement in the skin with respect to the underlying bones (Andersen, Benoit, Damsgaard, Ramsey & Rasmussen, 2010). Defining a kinematic model from directly measured markers can potentially lead to a model that does not conform to either the actual movement of the body, or to the model requirements for further analysis. For example, hinge joints and constant length segments are assumptions of common inverse dynamics calculations. Inverse kinematics potentially solves this problem by starting with a model that obeys all assumptions of the analysis.

OpenSim is a musculoskeletal modelling package allowing users to build and analyse 3D models of human movement (Simbios, Stanford University, California). As well as allowing analysis of underlying muscle kinematics and kinetics, the OpenSim package contains an Inverse Kinematics routine for fitting a musculoskeletal model to the measured marker kinematics (Delp, Anderson, Arnold, Loan, Habib, John, Guendelman & Thelen, 2007). A further benefit of the OpenSim is the support from a community of researchers that share resources online. This enables new users to start with existing models and adapt them to new purposes.

Inverse kinematics involves manipulating the position of a segmental model so that the position of markers embedded in the model match those measured experimentally throughout a movement. The accuracy of inverse kinematics routines can be affected by the complexity of the underlying skeletal model (Andersen et al., 2010) since an overly simplistic model will not be able to reproduce all components of in-vivo movement. On the other hand, having too complex a model will increase computational cost, making assumptions that introduce errors of a magnitude comparable to those of an over simplified model.

In the cricket fast bowling action the trunk must undergo flexion, lateral flexion and axial rotation (Stuelcken, Ferdinands & Sinclair, 2010) as the arm rapidly circumducts to release the ball with high speed and accuracy. It seems likely that this complex movement requires a complex musculoskeletal model to adequately represent the underlying movement. Such a model has not previously been reported for use with OpenSim.

The aim of this study is to develop an OpenSim model allowing complex shoulder and trunk movements, and to explore the effect of model complexity on the subsequent measurement of 3D kinematics.

METHODS: Thirteen fast bowlers (17.1 ±0.9 y, 85.8 ±10.8 kg, 1.91 ±0.05 m) were recruited from the Cricket New South Wales development squad. Forty-five 15 mm retroreflective markers were attached to the skin over bony landmarks as outlined by Ferdinands, Kersting & Marshall (2009). Each subject was instructed to bowl at maximum effort and the fastest trial from each subject was selected for analysis (mean velocity=33.3 ±2.4 m/s). Three-dimensional motion data (200 Hz) were captured using a 14-camera Cortex Motion Analysis System (Version 1.0, Motion Analysis Corporation Ltd., USA) and a recursive fourth-order low-pass Butterworth filter was used to smooth the coordinates at cut-off frequencies (8-15 Hz) determined from residual analyses. Data were analysed from 0.03 s before rear foot contact to 0.03 s after ball release.

The inverse kinematics routine of OpenSim 2.4.0 was used to fit three different models to the measured marker coordinates. The first model, RIGID, consists of a 20-segment, whole body model with 37 degrees of freedom (Hamner, Seth & Delp, 2010). This model is available in the public domain (simtk.org/home/opensim) and consists of three segment arms, five segment legs, pelvis, and single segment trunk/head/neck. The pelvis is connected to the rigid thorax/head by a three degree of freedom joint, with shoulder joints also connected to the trunk with three degree of freedom joints with fixed joint centres. The second model, SHOULDER, was an adaptation of Hamner's model to incorporate an additional three degrees of freedom of shoulder translation for each arm. This enabled the shoulder joint centre to translate with respect to the trunk, simulating movement of the pectoral girdle. The third model, BACK, incorporated a flexible lumbar segment developed by Christophy, Faruk Senan, Lotz & O'Reilly (2012). For this model, amounts of flexion, lateral flexion and axial rotation are partitioned between the lumbar vertebrae to provide anatomically appropriate amounts of movement between each pair of vertebrae. The original model by Christophy et al. (2012) consisted only of pelvis and vertebral segments. The present study, however, incorporated these components into the whole body model of Hamner et al. (2010), whilst also including the additional degrees of freedom of shoulder translation. While this new model incorporates an additional five segments for the lumbar vertebrae, no additional degrees of freedom were required because of constraints partitioning the amount of rotation between each vertebrae.

Two-way Anova with main effects of marker set (6) and model (3) was used to analyse the RMS differences between measured marker coordinates and those embedded in each segment during inverse kinematics fitting across the whole time period. One-way multivariate Anova with the main effect of model (3) was used to investigate kinematic variables of shoulder translation and lumbar, shoulder, elbow, hip, knee and ankle joint angles at the point of time when the bowling arm passed vertical in the RIGID model.

RESULTS: There was a significant main effect for model in the error between measured marker positions and those from the inverse kinematics routine (Table 1, $p<0.001$). Tests of within-subjects contrasts revealed that RIGID produced higher overall marker errors than the other models ($p=0.000$), while the total error for SHOULDER and BACK were similar to each other ($p=0.149$).

Table 1: Difference between measured marker positions and inverse kinematics for the three models under investigation (mean ±SD).

Marker set	RIGID *	SHOULDER	BACK
All Markers (n=45)	25.0 ±3.6	16.8 ±3.4	17.4 ±4.4
Right Arm (n=9)	27.0 ±3.2	16.7 ±3.5	16.6 ±3.5
Shoulders (n=6) #	43.3 ±4.2	26.1 ±3.8	26.0 ±3.6
Pelvis (n=3)	30.0 ±2.8	14.8 ±2.9	16.6 ±3.4
Legs (n=20) #	16.1 ±2.9	13.0 ±3.4	14.6 ±6.1
Trunk (n=4)	26.5 ±3.4	18.1 ±3.0	17.3 ±2.9

* Model is statistically different to the BACK model (p≤0.01).

\# Marker set is significantly different to the mean of all markers (p≤0.01).

There was also a significant main effect for marker set (p=0.000). Six markers positioned on the shoulders produced higher errors than the total averaged across all markers, while the errors for leg markers were less than the total (p=0.000). Marker errors on the arm, pelvis and trunk did not differ from the total marker set (p=0.780, 0.524 and 0.410 respectively). There was a significant interaction between the main effects of model and marker set (p=0.000), indicating that adding shoulder translation and lumbar movement to the RIGID model had more effect on some markers than others.

Freeing up the shoulder joint centre to translate allowed a movement of 11.1 mm, 15.5 and 4.4 mm respectively in the anterior, vertical and lateral directions (Table 2). Results for BACK produced similar results to SHOULDER (p≥0.021).

Table 2: Kinematic variables from the point in time when the arm passed vertical in the saggital plane, calculated using each of the three models under investigation (mean ±SD).

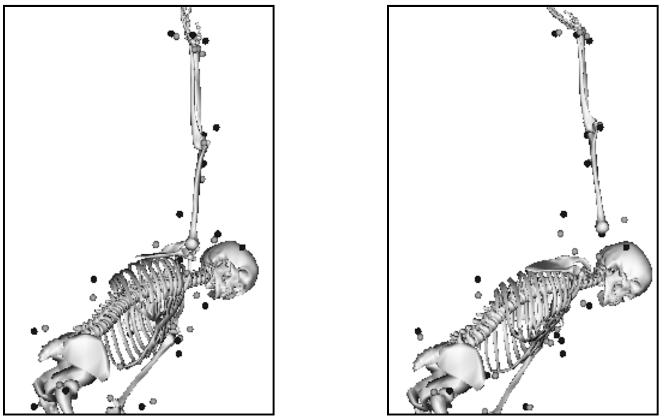
	RIGID		SHOULDER		BACK	
Shoulder anterior translation (m)	0.0*	±0.0	0.111	±0.026	0.110	±0.026
Shoulder vertical translation (m)	0.0*	±0.0	0.155	±0.019	0.148	±0.021
Shoulder lateral translation (m)	0.0	±0.0	0.044	±0.024	0.042	±0.023
Hip Flexion (°)	24.2	±11.6	24.8	±12.6	26.8	±13.2
Hip Abduction (°)	-0.2*	±5.6	3.1	±7.6	2.6	±7.8
Knee flexion (°)	62.8*	±17.3	69.4	±17.7	69.9	±17.9
Ankle plantarflexion (°)	30.4*	±12.2	26.5	±11.7	26.8	±11.9
Lumbar extension (°)	32.8	±10.6	34.7	±9.4	32.6	±9.4
Lumbar lateral flexion (°)	32.8*	±7.5	19.0	±7.0	21.1	±7.4
Lumbar rotation (°)	38.6*	±8.2	20.5	±6.7	17.8	±7.3
Shoulder extension (°)	141.3*	±13.5	112.8#	±13.0	107.9	±13.3
Shoulder abduction (°)	56.5*	±6.5	47.4	±10.7	48.3	±10.6
Elbow flexion (°)	8.2*	±13.1	28.0	±15.6	28.3	±15.8

* RIGID model is statistically different to the BACK model (p≤0.01).
\# SHOULDER model is statistically different to the BACK model (p≤0.01).

There was a significant difference between the RIGID and BACK models in the measurement of every kinematic variable except lumbar and hip flexion (Table 2), while only shoulder extension demonstrated a significant difference between the SHOULDER and BACK models (4.9°, p=0.01). Choice of model significantly affected the processing time for the Inverse Kinematics routine in OpenSim (p=0.000). The average processing times were 107.9 ±23.9 s for RIGID, 6.5 ±1.7 s for SHOULDER and 500.4 ±104.6 s for BACK.

DISCUSSION: Allowing the shoulder joint centres to translate with respect to the trunk is anatomically appropriate given their connection to the trunk via the shoulder girdle. Having a rigidly attached shoulder model is a significant restriction and it is not surprising that markers placed on the shoulders produced the largest degree of error during the inverse kinematics routine. Shoulder markers still produced the largest degree of error even after changing to models allowing translation, although they had the greatest relative improvement of all marker sets.

What was perhaps less expected was the improvement in tracking pelvic markers after allowing shoulder translation. This improvement in the pelvis tracking came about through the rigid model having to select segment positions that minimised errors across all markers, including the shoulders. In order to minimise RMS marker error across all markers, including the shoulders, the pelvis was moved upward to follow the shoulders and improve the overall accuracy of reconstruction (Figure 1). This change in marker positions across the whole body had consequent effects on all joint angles. Even ankle angle, which should not be affected by changes in the shoulder or trunk anatomy, was significantly different between models.



a) RIGID b) SHOULDER

Figure 1: Inverse kinematics for the Rigid and Shoulder Translation models for one subject at a single point in time. Measured marker positions are illustrated in black, while those reconstructed using inverse kinematics are illustrated in grey.

The effect of model choice on inverse kinematics processing time was much greater than initially anticipated. Freeing up shoulder translation simplified the processing as each arm could be moved to an appropriate position without affecting other segments, thereby reducing processing time by more than fifteen times compared to the rigid model. By contrast, adding the additional lumbar movements increased processing time by a factor of more than 75, compared to the shoulder model.

CONCLUSION: Adding shoulder joint translation to the musculoskeletal model improved the fit of marker coordinates through the inverse kinematics routine and significantly changed all the investigated joint angles except lumbar and hip flexion. Of particular note were changes to the angles of joints well away from the arm; for example the ankle joint. Switching the rigid trunk for a model with anatomically appropriate levels of lumbar movement did not further improve the overall accuracy of marker coordinates; however there was a 5° change in shoulder extension angle. This study highlights the need for an appropriate level of complexity in developing a model for use with inverse kinematics. Tracking cricket bowling with the shoulder joint centre rigidly fixed to the trunk creates kinematic errors throughout the body. The further benefit from allowing movement of the lumbar vertebrae is less clear. Small differences in kinematics were found with the more complex model, but with greatly increased computational time. This additional time is likely to be of benefit, however, if researchers are specifically interested in the mechanics of the lumbar spine.

REFERENCES:

Andersen M.S., Benoit D.L., Damsgaard M, Ramsey D.K., & Rasmussen J. (2010). Do kinematic models reduce the effects of soft tissue artefacts in skin marker-based motion analysis? An in vivo study of knee kinematics. *Journal of Biomechanics*, 43, 268–273.

Christophy M., Faruk Senan N.A., Lotz J.C. & O'Reilly O.M. (2012). A Musculoskeletal model for the lumbar spine. *Biomechanics and Modeling in Mechanobiology*. 11,19-34.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E. & Thelen, D.G. (2007). OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement, *IEEE Transactions on Biomedical Engineering*, 54, 1940-1950.

Ferdinands, R.E., Kersting, U & Marshall, R.N. (2009). Three-dimensional lumbar segment kinetics of fast bowling in cricket. *Journal of Biomechanics*, 42, 1616-1621.

Hamner, S.R., Seth, A. & Delp, S.L. (2010). Muscle contributions to propulsion and support during running, *Journal of Biomechanics*, 43, 2709-2716.

Stuelcken M.C., Ferdinands R.E.D. & Sinclair P.J. (2010). Three-dimensional trunk kinematics and low back pain in elite female fast bowlers. *Journal of Applied Biomechanics*. 26, 52-61.