

LUMBAR KINEMATICS AND KINETICS OF YOUNG AUSTRALIAN FAST BOWLERS

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Lower back injuries are a serious concern for cricket fast bowlers. As lumbar loading is the causal mechanism of such injuries, the purpose of this study was to find relationships between lumbar loads and selected kinematic variables. Thirteen young fast bowlers (17.4 ± 1.9 years) were tested with a 3-D motion analysis system (200 Hz). Kinematics and lumbar spine kinetics were calculated about the L5/S1 joint during the arm acceleration phase. The largest kinetic values were the lumbar axial forces and lumbar flexion moments. Maximum lumbar spine moments were associated with several kinematic variables such as front knee angle, pelvic and thoracic rotation at ball release, and shoulder counter-rotation. Modifying bowling kinematics may reduce lumbar loads and reduce the potential for lower back injuries.

KEYWORDS: cricket bowling, lumbar spine, counter-rotation, lumbar kinetics, fast bowling.

INTRODUCTION: Fast bowling in cricket requires the bowler to rapidly flex, laterally bend and rotate the lumbar spine in order to produce ball speeds up to 45 m s^{-1} . These movement patterns are considered to play a role in the development of lower back injuries. Intervertebral disc abnormalities and soft tissue injuries in the lumbar region are often observed in fast bowling populations, but the most serious condition in terms of lost playing time involves fractures to the pars interarticularis, particularly of the L4/L5 or L5/S1 vertebrae (Elliott et al., 1993; Portus et al., 2004; Orchard, 2006). The problem is of great concern to cricket administrators and coaches, because young bowlers are the most at risk group (Portus et al., 2007).

Researchers have studied various kinematic factors for associations with lumbar injury incidence. A number of studies have found there to be an increased risk of lumbar spine injury when shoulder counter-rotation, a preliminary rotation of the shoulder girdle in the horizontal plane away from the direction of bowling, is in excess of 30° or 40° (Elliott, 2000; Portus et al., 2004). Pelvic-shoulder separation angle at back foot angle has been associated with a moderate increase in soft tissue injury (Portus et al., 2004). In addition, bowlers with back injuries may utilise greater ranges of lateral flexion of the lumbar spine during delivery stride (Portus et al., 2007). In terms of identifying the causal mechanisms of lumbar injury, kinetic calculations are required. Ferdinands et al. (2009) tested 21 fast bowlers of premier grade level and above and found that large flexion, rotation and lateral bending moments were placed on the spine when displaced towards the end of its available range of motion. However, there has been no study to date that has investigated the association between kinematic variables and lumbar spine loads.

The established method of determining lumbar injury risk in fast bowlers is mostly based on shoulder counter-rotation, which is only a kinematic measure. Hence, shoulder counter-rotation is not a causal mechanism of lumbar injury. By establishing the kinematic correlates of lumbar spine loads, it may be possible to develop a more accurate assessment of lumbar injury risk. The identification of these kinematic characteristics may have implications for the development of safer bowling techniques, particularly with respect to younger bowlers. Therefore, the purpose of this study was to use three-dimensional motion analysis and inverse dynamics to investigate the relationships between kinematics and lumbar spine kinetics within an elite sample of young fast bowlers. The hypothesis is that there are associations between lumbar loads and kinematic variables, particularly with those kinematic variables that have been associated with an increased incidence of lumbar injury.

METHOD: Thirteen young fast bowlers (17.4 ± 1.9 years) were recruited from the Cricket New South Wales development squad. The trials were performed in a biomechanics

laboratory, which permitted a full length run-up. A 14-camera Cortex Motion Analysis System (Version 1.0, Motion Analysis Corporation Ltd., USA) was used to capture three-dimensional (3D) motion (200 Hz) and force plate (1000 Hz) data on 20 trials for each bowler while front and rear foot contact was made on two Kistler force plates. Each subject was instructed to bowl at maximum effort as in match conditions. Five trials in which the ball landed within a 'good length' area demarcated by two white lines 13 m and 19 m from the stumps at the bowler's end were selected for analysis. Subjects also rated their performance from 0 to 10 using an analogue performance scale. The video capture volume encompassed the back foot contact, front foot contact, ball release and follow through phases of the bowling action. The Cortex system was calibrated according to the manufacturer's recommendations resulting in a residual error of marker position of less than 1 mm.

Motion analysis capture was performed on each subject wearing a full body marker set comprising forty-five 15 mm spherical markers, which were attached to bony landmarks (Ferdinands, 2009). Markers were located on the left and right sides of the body except for markers half-way between the posterior superior iliac spines (mid-PSIS), and on the 7th cervical vertebrae, supra-sternal notch, and the head. The positions of the anterior superior iliac spine (ASIS), mid-PSIS, and greater trochanter markers were used to calculate the hip joint centres. All other joint centres were calculated as the average position between two markers placed either medially and laterally or anteriorly and posteriorly on the joint. Exceptions were the position of the shoulder, mid-trunk, hip markers, and cricket ball. A recursive fourth-order low-pass Butterworth filter was used to smooth the motion analysis data. The cut-off frequencies (8 – 15 Hz) were determined from residual analyses.

The three-dimensional motion analysis data of the markers were imported into a 22 segment rigid body model of the cricket fast bowler in Kintrak (V.7.0, University of Calgary), which is a software programme designed to perform kinematics and inverse dynamics analysis using motion analysis and force plate data. Local segment coordinate systems of the rigid body model were defined for each of their respective segments. The lumbar spine segment (LSS) was defined as a single segment having its inferior end located half-way between the hip joint centres at the level of L5/S1. The superior endpoint was located at the mid-point between the markers on the suprasternal notch and T7.

Kinematic and kinetic data were calculated during the arm acceleration phase defined from the time of maximum vertical front foot ground reaction force to time of maximum hand velocity. Net joint torque was calculated about the inferior end of the lumbar spine segment for lateral bending, flexion/extension and rotation. Right lateral bending, extension and right rotation were defined as positive. Pearson's correlation coefficients were calculated in SPSS (Version 17, SPSS Inc.) to assess the relationship between selected kinematic and kinetic variables.

RESULTS: The sample had the following mean kinematic values: shoulder counter-rotation ($39.3 \pm 12.7^\circ$), pelvic-shoulder separation angle ($-23.1 \pm 8.2^\circ$), thoracic lateral bending ($41.5 \pm 8.5^\circ$), thoracic rotation ($119.2 \pm 14.6^\circ$), pelvic rotation ($107.4 \pm 13.2^\circ$), front knee flexion angle (-17.0 ± 6.5), stride angle ($5.0 \pm 5.8^\circ$) and bowling hand velocity ($23.8 \pm 1.2 \text{ m s}^{-1}$).

Forces and moments were expressed in terms of body weight (BW) and body weight x height (BW m). The highest ground reaction forces were the vertical components (5.3 ± 0.8 and $2.3 \pm 0.4 \text{ BW}$) at the front foot and back foot. The mean maximum lumbar forces were $8.0 \pm 1.2 \text{ BW}$ along the inferior-superior long axis, $1.5 \pm 0.6 \text{ BW}$ along the posterior-anterior axis and $0.4 \pm 0.5 \text{ BW}$ along the lateral-medial axis. The mean maximum lumbar torques were $3.1 \pm 0.5 \text{ BW m}$ (flexion), $0.9 \pm 0.4 \text{ BW m}$ (left lateral bending) and $0.2 \pm 0.2 \text{ BW m}$ (right rotation).

Table 1 shows the Pearson correlation coefficients between the lumbar spine kinetic and kinematic variables. There were also other kinematic variables that were not significantly correlated with any kinetics variables: kinematic crunch factor (thoracic lateral bending x pelvic rotation), stride length, and centre of mass horizontal and vertical velocities at back foot contact. Bowling hand velocity was not correlated with any of the above kinematic variables. Counter-rotation was correlated with stride angle ($r = -0.57$, $p = 0.042$). The

coefficients of variation of the kinematic and kinetic data were less than 8.0% and 15.2%, respectively.

Table 1. Pearson's correlation coefficients between lumbar spine kinetic and kinetic variables Thoracic and pelvic kinematics were calculated at ball release. Front knee angle was calculated at the time of maximum front foot ground reaction force. (Significance level, $p < 0.05$)

	Axial Force	Anterior-posterior Force	Medio-lateral Force	Flexion Moment	Rotation Moment	Lateral Bend Moment
Thoracic flexion		$r = -0.82$ $p < 0.001$			$r = 0.52$ $p = 0.067$	
Thoracic rotation		$r = -0.57$ $p = 0.040$		$r = 0.59$ $p = 0.033$	$r = 0.59$ $p = 0.034$	$r = -0.83$ $p < 0.001$
Thoracic lateral bend						$r = 0.59$ $p = 0.035$
Pelvic rotation		$r = -0.58$ $p = 0.037$		$r = 0.49$ $p = 0.093$	$r = 0.112$ $p = 0.056$	$r = -0.92$ $p = 0.001$
Counter-rotation			$r = 0.52$ $p = 0.067$		$r = 0.61$ $p = 0.028$	
Pelvic-shoulder separation						$r = -0.54$ $p = 0.053$
Front knee angle		$r = -0.82$ $p < 0.001$			$r = 0.91$ $p < 0.001$	$r = -0.81$ $p = 0.001$
Stride angle			$r^2 = -0.61$ $p = 0.028$			
Hand velocity			$r = 0.50$ $p = 0.082$			
Front foot GRF Z	$r = 0.59$ $p = 0.03$					

DISCUSSION: The aim of this study was to investigate the relationships between lumbar spine kinematics and kinetics in a sample of elite young fast bowlers. A relatively young cohort was selected because this age group has a high incidence of lumbar injury (Portus et al., 2007).

Interestingly, the kinematic variables that were most strongly correlated with lumbar spine kinetics were the pelvic and thoracic rotations at ball release. These variables were correlated with four kinetics variables: anterior-posterior force and all three lumbar spine moments. The position of the thorax in bowling and related motions such as throwing all require the pelvis and thorax to face the target at the time of release. However, there is no need for these segments to rotate beyond this as these variables were not correlated with bowling hand speed.

Conversely, the kinematic variables that have previously been associated with an increased incidence of lumbar injury had fewer associations with lumbar kinetics. Shoulder counter-rotation was strongly correlated with the lumbar spine rotation moment, but this moment was very small in magnitude and may not be of clinical significance. In addition, shoulder counter-rotation was strongly correlated with the medio-lateral force, but this may just result from an unbalanced position during delivery stride since counter-rotation was also correlated with a

closed stride angle. Pelvic shoulder separation angle was strongly correlated with the lumbar lateral bending moment. The crunch factor is used as an index of shear stress loads at the lumbar vertebrae. However, this factor was not associated with any lumbar spine moments.

The front leg acts as a shock absorber to attenuate the ground reaction forces upon front foot contact. Knee flexion angle had no effect on the attenuation of the large axial forces, which were strongly correlated with vertical ground reaction force acting through the front foot. However, the front knee flexion angle did have strong correlations with both the lumbar spine rotation and lateral bending moments and therefore has an important effect on lumbar spine loading.

In general, the data shows that there a number of kinematic variables associated with lumbar spinal loads. A combination of these variables can work additively to increase lumbar spine loading. For instance, the data suggests that a bowler landing with a more flexed front knee, large pelvic-shoulder separation angle and large range of lateral thoracic bending has three factors that would contribute to the generation of a lateral bending moment. This suggests that a risk of assessment of lumbar injury in bowling may need to consider multiple variables.

CONCLUSION: This study supports the hypothesis that lumbar loading in fast bowling is associated with kinematic variables. As the causal mechanisms of lumbar injury are ultimately linked to spinal loading, the identification of kinematic variables associated with such spinal loading can lead to an improved assessment of lumbar injury risk in young fast bowlers. However, a prospective longitudinal study is needed to compare a wide range of kinematic and lumbar spine kinetics variables to assess their predictive ability of lumbar spine injury. The researchers are currently evaluating the MRI scans taken of all bowlers towards the end of the cricket season to quantitatively assess their injury status. In addition, the data of another five bowlers will be analysed. Such research has the potential to yield an accurate multi-index assessment of lumbar injury risk. This would give coaches the ability to more accurately screen young bowlers for injury risk and also suggest changes to the kinematics of bowling techniques to reduce lumbar spine loads.

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