Kinematic Perturbation in the Flexion-Extension Axis for Two Lumbar Rigs During a High Impact Jump Task

Marc R. Portus, David G. Lloyd, Bruce C. Elliott, and Neil L. Trama

The measurement of lumbar spine motion is an important step for injury prevention research during complex and high impact activities, such as cricket fast bowling or javelin throwing. This study examined the performance of two designs of a lumbar rig, previously used in gait research, during a controlled high impact bench jump task. An 8-camera retro-reflective motion analysis system was used to track the lumbar rig. Eleven athletes completed the task wearing the two different lumbar rig designs. Flexion extension data were analyzed using a fast Fourier transformation to assess the signal power of these data during the impact phase of the jump. The lumbar rig featuring an increased and pliable base of support recorded moderately less signal power through the 0–60 Hz spectrum, with statistically less magnitudes at the 0–5 Hz ($p = .039$), 5–10 Hz ($p = .005$) and 10–20 Hz ($p = .006$) frequency bins. A lumbar rig of this design would seem likely to provide less noisy lumbar motion data during high impact tasks.

Keywords: lumbar spine, measurement, apparatus, Fourier analysis, vibration, noise

The relationship between lower back injuries and cricket fast bowling has been well documented from biomechanical (e.g., Elliott and Khangure, 2002) and epidemiological (e.g., Orchard et al., 2002) perspectives. Epidemiological data show that lower back injuries present a significant health risk for cricketers and require lengthy rehabilitations (Foster et al., 1989; Stretch, 1995; Orchard et al., 2002; Ranson et al., 2005). A number of biomechanical studies have linked the excessive twisting action of the upper trunk, inherent in the mixed technique, with lower back injuries such as spondylolysis, increased rates of disc degeneration, muscle strains and ligament sprains (Foster et al., 1989; Elliott et al., 1992; Burnett et al., 1996; Portus et al., 2004). These studies suggest that the etiology of these injuries is mechanically based, though other factors such as the bowler’s physical attributes and bowling workloads are also important (Foster et al., 1989; Dennis et al., 2003). Burnett et al. (1998) and Ranson et al. (2008) have described the three-dimensional (3D) kinematics of the lower thorax in fast bowlers and reported varying magnitudes and somewhat independent motion with respect to more global measures such as pelvis and upper trunk kinematics.

This would suggest knowledge of the pathomechanics of lower back injuries in fast bowlers will be enhanced by quantifying the motion of the lumbar spine, the site of many debilitating injuries for fast bowlers (Engstrom & Walker, 2007; Foster et al., 1989; Orchard et al., 2002; Ranson et al., 2005). The measurement of lumbar spine kinematics, particularly in dynamic activities like fast bowling and javelin throwing, does, however, present a challenge due to soft tissue movement relative to the underlying lumbar vertebrae, the lumbar spine not being a rigid segment and occlusion of markers due to trunk rotation and arm movement. Electromagnetic systems (e.g., Burnett et al., 1998) the lumbar motion monitor (e.g., Marras et al., 1993) and the lumbar rig (e.g., Schache et al., 2002) have been previously used to describe lumbar motion. Each of these methods has strengths and weaknesses and suit particular applications better than others. For dynamic high impact activities lightweight devices are ideal, reducing the encumbrance on the athlete as they run, jump, flex, rotate and extend their trunk in activities such as cricket fast bowling or javelin throwing. While skin mounted markers offer the most lightweight solution, pilot testing in our 8-camera laboratory showed that marker occlusion during cricket fast bowling was overcome using markers mounted on a lumbar rig.

The lumbar rig consists of a lightweight triad of retro-reflective markers projecting dorsally from the trunk and is strapped securely around the torso of the participant. Retro-reflective optical motion analysis systems, such as Vicon, have been used to track the rig, enabling the quantification of thoraco-lumbar-pelvic kinematics in...
running (Schache et al., 2002). Due to its low mass and reduced encumbrance to the athlete, it is a potentially useful approach to measuring lumbar motion in dynamic sporting activities.

A consideration when using a lumbar rig for biomechanical analyses is the high impact nature of some sporting activities. Mean vertical ground reaction forces (GRF) of 5–7.3 bodyweights in the fast bowling delivery stride of cricketers have been reported (Elliott et al., 1992; Portus et al., 2004), whereas Hunter et al. (2004) reported relatively lower GRF of 3.2 bodyweights for running. As the lumbar rig projects dorsally from the lumbar spine, sudden changes of momentum of the trunk during high impact activities may lead the lumbar rig to move independently of the spine, resulting in exaggerated and erroneous lumbar acceleration. These high impact forces occur predominantly in the vertical direction; as such the flexion-extension data from the lumbar rig would be most susceptible to error. Drop jumps from a bench have been shown to record vertical ground reaction forces in the order of 8–12 bodyweights (Elvin et al., 2007), hence are a controllable high impact task to assess the merits of lumbar rig stability during high impact activities.

The aim of this study was to assess flexion-extension data from two lumbar rig designs during a controlled high impact bench drop jump. Both lumbar rigs were modified from previous validated designs in gait research (Whittle & Levine, 1999; Schache et al., 2002a). The first lumbar rig was based on Schache et al. (2002) (Rigid Rig); the second featured a pliable base of support at the interface with the lower back (Semi-Rigid Rig). Due to the increased base of support and pliability of the Semi-Rigid Rig, it was hypothesized that this design would suffer less perturbations around the flexion-extension axis during the high impact bench jump task.

Methods

Participants and Ethics

Eleven junior high performance male cricket fast bowlers participated in the study (mean age 16.0 ± 1.1 years; height 1.8 ± 0.05 m; mass 74 ± 8.6 kg). Ethical clearance was obtained from the home institutions Ethical committee and written, informed consent was granted by the athletes and/or their legal guardian.

Design of Lumbar Rigs

A key consideration during construction was increased durability to withstand the “wear and tear” that would be expected during dynamic high impact activities, as opposed to walking, jogging or running. The lumbar rigs were computer numerical control (CNC) milled from extruded sheet ultra high molecular weight polyethylene (UHMWPE) using a Bridgeport vertical mill, with 3D code generated using Mastercam software. This manufacturing process eliminated potential sources of intra-rig movement artifact due to construction articulations. The Rigid Rig featured a UHMWPE base that traveled 100 mm across the lumbar region in both lateral directions.

The Semi-Rigid Rig featured a lightweight semirigid base of support that spanned 100 mm in both lateral directions, 22 mm inferiorly and 70 mm superiorly from the rig origin (that is its interface with the lumbar spine; Figures 1 and 2). A key aim of the Semi-Rigid Rig design was improved conformity for a multitude of back shapes and sizes.

Lumbar Rig Placement and Coordinate System

As many lower back injuries are to the lower vertebral areas, the lumbar rigs were placed between the spinous process of the second and third lumbar vertebrae. The rigs had a padded Velcro strap which was secured firmly around the torso of the participant, so that when the bowler moved their trunk the lumbar rig followed the movement of the lower back (Figure 2C). The 3D coordinate system of the lumbar rigs was defined by the three points in space, represented by the posteriorly mounted retro-reflective markers (Figure 3). A virtual point was created at the midpoint between the left lumbar marker (LUML) and the right lumbar marker (LUMR). This was the origin for the lumbar rig coordinate system (LumOrig). The first defining line and x-axis of the rig’s coordinate system was a vector running from the middle lumbar rig marker (LUMM) through the LumOrig. The second defining line was a vector running from the LUML marker to the LUMR marker. The cross product of this defining line with the x-axis provided the vertical y-axis, while the cross product of the x- and y-axes provided the z-axis, running from LumOrig toward LUMR (Figure 3). Bodybuilder for Biomechanics software (Oxford Metrics, U.K.) was used to construct this custom written lumbar rig coordinate system. Flexion-extension of the lumbar rig was described relative to the laboratory coordinate system z-axis (global x-axis running posterior-anterior, global y-axis running vertically and global z-axis running laterally across the laboratory).

Participants performed a high impact jump from a bench that was 50 cm high at a target landing area 40 cm anterior to the bench. Each participant was instructed to leave the bench with both feet together with their trunk in a relatively upright posture (i.e., no excessive flexion at the hip was permitted), to look forward, minimize arm swinging movements and land simultaneously with both feet. Instructions for landing were to resume an upright anatomical position as quickly as could be achieved, while maintaining balance and control. Hip and knee flexion to assist absorption of landing forces was permitted; however, only trials where the anatomical position was assumed within one second after initial contact were used for analysis. Foot landing strategies (e.g., heel first, forefoot first, flat) were self selected during task familiarization, identified by the researchers and monitored during data collection to ensure participants used the same technique across all trials.
Figure 1 — The two lumbar rigs that were attached to the lumbar region of study participants. The Semi-Rigid lumbar rig (A) featured a lightweight Corflute material which was designed to minimize high impact flexion-extension perturbations that the Rigid Rig (B) was suspected to experience. The Semi-Rigid lumbar rig also featured a more pliable support base at the lateral aspects to increase fit for a variety of trunk morphologies.

Bench jumps were performed in a Biomechanics laboratory, while being filmed (120 Hz) by an 8-camera retro-reflective Vicon motion analysis system (Oxford Metrics, U.K.). Each participant performed four bench jumps, two jumps with each lumbar rig attached. Raw flexion-extension data were analyzed from five frames before foot contact with the ground until the participant was in an upright stable posture. Marker trajectory data and video of the performance were inspected during data collection to ensure each participant used a consistent landing strategy between trials and when wearing the different lumbar rigs. Twenty-two trials with each lumbar rig design (44 trials in total) were analyzed. Raw unfiltered data were exported from Bodybuilder into Microsoft Excel (Microsoft, U.S.A.) for further analysis.

A power spectrum analysis was used to investigate the signal power in flexion-extension data curves through a frequency spectrum of 0–60 Hz. A discrete Fourier transform (Burden and Faires, 1985) was used to perform the power spectrum analysis from the raw flexion-extension data for each lumbar rig through the bench jump impact phase. The power spectrums for each trial were interpolated using a cubic spline (Burden and Faires, 1985) from which the average signal power in 5 frequency bands were determined: 0 to 5 Hz; 5 to 10 Hz; 10 to 20 Hz; 20 to 30 Hz; and 30 to 60 Hz. From pilot testing we observed that most of the higher frequency power responses were in the 0–20 Hz range, particularly in the 0–10 Hz range, hence more frequency bins in these ranges. The 60 Hz maximum frequency was set by the 120 Hz data acquisition rate, i.e., 1/2 of 120 Hz. The frequency bin approach also facilitated statistical analysis.

As higher magnitudes of lumbar rig flexion influenced the power spectrum analysis—that is if the lumbar rig was more flexed more power was calculated—all lumbar rig flexion data curves were reset at zero at the commencement of the power spectrum analysis phase (5 frames before initial foot contact). The mean and sum power values of the two trials for each lumbar rig were calculated to provide descriptive data for each bowler. Paired t tests were employed (SPSS Inc., U.S.A.) to test for statistically significant differences for each harmonic band for each lumbar rig, and since the hypothesis was set a priori, and we specified the direction of the differences (Semi-Rigid design would have smaller power in harmonic bands compared with the Rigid design), statistical significance for paired t tests was set at 0.05 with one-sided tests. In one-sided tests it was acceptable to use $p < .10$ (twice the alpha for two-sided tests); however, statistical significance was maintained at $p < .05$ to help avoid the chance of type I error. Cohen’s effect sizes (Cohen, 1988) were also calculated for each harmonic band and Hopkins (2002) classification system adopted for descriptors of effect sizes: $\geq 0.0$ represent a trivial difference; $\geq 0.2$ a small difference; $\geq 1.2$ a moderate difference; $\geq 2.0$ a very large difference.

Results

The mean and sum signal powers were higher when wearing the Rigid Rig for all 11 bowlers, though the magnitude of this difference for 1 bowler was negligible (Figure 4).
Overall the Semi-Rigid Rig had “moderately” less signal power in the 0–60 Hz frequency spectrum (ES = 0.72, t = -2.87, p = .017). When signal power was analyzed through the 0–5, 5–10, 10–20, 20–30 and 30–60 Hz frequency spectrums, the Semi-Rigid Rig recorded less signal power in all spectrums except the 30–60Hz spectrum. Statistically significant differences were observed in the 0–5 Hz, 5–10 Hz and 10–20 Hz spectrums, where the Semi-Rigid Rig recorded “small”, “large” and “large” amounts less signal power, respectively. A nonsignificant difference was at the 20–30 Hz spectrum where the Semi-Rigid Rig had a “moderate” amount less signal power than the Rigid Rig. In the 30–60 Hz spectrum both rigs recorded similar powers. Statistically, the Semi-Rigid Rig recorded significantly less signal power in the 0–5, 5–10 and 10–20 frequencies than the Rigid Rig (Table 1).

Discussion

Results show that the Semi-Rigid Rig was significantly less susceptible to excessive perturbation in the 0 to 20 Hz frequency spectrum during a high impact activity. The design of this lumbar rig was specifically created to reduce excessive noise following an impact; results indicate the Semi-Rigid rig achieved this aim. The 5–20Hz spectrum was where the greatest discrepancies occurred between the two lumbar rigs, suggesting that it is in this area of the frequency spectrum where the true motion of the lumbar spine could be contaminated with noisy data when wearing the Rigid Rig during a high vertical force. Pilot fast-Fourier transform analysis of fast bowling front foot vertical ground reaction forces in our laboratory indicated dominant frequencies ranged from 4 to 33 Hz with a mean ± SD of 10 ± 7 Hz. This would suggest the Semi-Rigid rig would provide less perturbed lumbar motion data during a high impact task, such as cricket fast bowling.

A lumbar rig design featuring a firm but pliable and lightweight base plate significantly reduced flexion-extension movement artifact after controlled high impact bench jumps. This design is likely to be more suitable when lumbar spine kinematics and kinetics need to be measured during high impact activities. This is due to its increased base of support and semi rigid pliability, so it can better conform to a variety of lower back morphologies (e.g., larger and small muscle masses). This design should be considered by researchers needing to measure the kinematics or
kinetics of the lumbar spine during dynamic, high impact activities, such as cricket fast bowling and javelin throwing.

There are a number of limitations to this study that the reader should consider. Due to technical issues our force plates were unfortunately not available during the entire data collection phase of this study. This meant we were not able to assess impact forces across the 2 lumbar rig bench jump conditions. However we are confident that we employed a well controlled study design and task execution was consistent, as detailed in the methods section. The second limitation was we would have liked to have conducted more bench jump trials for each fast bowler to assess any variability in the bench jump task. As this study was part of a bigger research program we were unable to obtain more time from our participants. Again we are confident that we have faithfully assessed the relative merits of the two lumbar rigs as we used consistent criteria for task execution in order for a trial to qualify for inclusion in the study. Although the Semi-Rigid rig was designed for a variety of back morphologies, this study had a small adolescent athletic male cohort, meaning these results may not be generalizable to the greater adolescent, female or adult male populations. Researchers should assess the device thoroughly with the specific population and task to be studied. Finally, the frequency bins were qualitatively chosen and may have influenced the results. However, these were selected to best represent

Table 1  Signal powers (Mean ± SD) and statistical results for the Semi-Rigid and Rigid lumbar rigs in the bench jump task (120 Hz); ES = effect size

<table>
<thead>
<tr>
<th>Frequency Bin (Hz)</th>
<th>Signal Power</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Semi-Rigid Rig</td>
<td>Rigid Rig</td>
<td>ES</td>
<td>ES Rating</td>
<td>t</td>
</tr>
<tr>
<td>0–5</td>
<td>3843 ± 3127</td>
<td>5752 ± 4034</td>
<td>0.53</td>
<td>Small</td>
<td>-2.4</td>
</tr>
<tr>
<td>5–10</td>
<td>262 ± 165</td>
<td>513 ± 224</td>
<td>1.29</td>
<td>Large</td>
<td>-3.6</td>
</tr>
<tr>
<td>10–20</td>
<td>112 ± 60</td>
<td>559 ± 459</td>
<td>1.72</td>
<td>Large</td>
<td>-3.5</td>
</tr>
<tr>
<td>20–30</td>
<td>22 ± 16</td>
<td>64 ± 84</td>
<td>0.85</td>
<td>Moderate</td>
<td>-2.0</td>
</tr>
<tr>
<td>30–60</td>
<td>19 ± 26</td>
<td>18 ± 12</td>
<td>-0.07</td>
<td>Trivial</td>
<td>0.2</td>
</tr>
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Figure 4 — Fourier analysis signal power for 0–60 Hz frequencies for participants when wearing the Semi-Rigid and rigid lumbar rig during the bench jump task.
the continuous power-frequency curves of pilot data, so the affects should be minimal.

In conclusion, the Semi-Rigid lumbar rig was demonstrated to suffer less noise from a high impact bench jump task. Its lightweight Semi-Rigid design provides a good alternative for researchers wishing to quantify lumbar motion in high impact dynamic sporting activities, particularly with motion analysis systems with fewer cameras (n ~ 10).

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References