Field based quantification of trunk and pelvis motion in cricket bowling: a wearable technology approach

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This thesis is presented for the degree of Doctor of Philosophy of The University of Western Australia
School of Human Sciences (Exercise and Sport Science)
August 2019
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The research involving human data reported in this thesis was assessed and approved by The University of Western Australia Human Research Ethics Committee. Approval #: RA/4/1/2593. Written patient consent has been received and archived for the research involving patient data reported in this thesis.

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24th January 2019
Gold-standard, three-dimensional (3D) retro-reflective motion capture is associated with high measurement accuracy but suffers from inherently poor ecological validity. Systems are also expensive and require specific training to utilise effectively. The primary aim of this thesis was to explore field-based motion capture alternatives to 3D retro-reflective motion capture, with a focus on measurement of trunk and pelvis kinematics. Cricket fast bowling was used as the main reference task for all studies. Bowling is a complex skill that involves high-speed, multi-planar movements of the trunk and pelvis through large ranges of motion. Cricket fast bowlers are also susceptible to debilitating lumbar injuries. Previous research has linked bowling biomechanical factors to lumbar injury, many of which involve pelvis, lumbar and/or thorax kinematics. The major application of this research was to provide biomechanists and coaches with relevant information on the measurement validity of two-dimensional (2D) video and magneto-inertial measurement unit (MIMU) based analyses of dynamic trunk and pelvis motion during the cricket bowling action. A secondary application was to assess the suitability of custom 2D video and MIMUs as lumbar injury risk screening tools for cricket fast bowlers.

**Study 1** assessed the validity of 2D video based measurements of joint and segment angles associated with lumbar injury in fast bowlers. Nineteen fast-medium bowlers had their bowling actions simultaneously recorded by multiple 2D video cameras and a 3D retro-reflective motion capture system. Measures of thorax lateral flexion angle, pelvis rotation angle, hip flexion-extension angle, and knee flexion-extension angle were calculated by digitising the 2D video and were compared with anatomically modelled retro-reflective based motion capture data. The measurement validity of 2D video based measurements of lumbar injury risk factors were dependent on the variable being assessed, with intra-class correlation values ranging from 0.38-0.97 and root mean square difference (RMSD) values ranging from 4.3-9.4°. Body segment and joint modelling dissimilarities between the video and retro-reflective based approaches were likely the main cause of measurement differences for all variables, although video based measures of higher-velocity movements, and movements that crossed multiple planes of motion, produced the poorest agreement with the retro-reflective derived angles. It appears that some kinematic factors associated with fast bowler lumbar injury, such as pelvis rotation angle and knee flexion-extension angle at ball release, can be measured from 2D video. Those variables produced excellent intra-class correlations and small RMSD values when compared with 3D retro-reflective based anatomical model equivalent outputs. However, 2D video based methods do not appear to be wholly suitable for highly valid, comprehensive analyses of bowling biomechanics.
Magneto-inertial measurement units have been identified as a cost-effective and portable 3D motion capture tool with the potential to overcome the shortcomings of 2D video motion capture when measuring dynamic, sport-specific movements of the trunk and pelvis. Sport-specific MIMU measurement validity studies focusing on the trunk and pelvis are scarce. A key determinant of valid MIMU measurement involves aligning or calibrating MIMUs with body segments so that the MIMU measures are anatomically and functionally relevant. Anatomical and functional calibration approaches have been described in the literature. Anatomical alignment of MIMUs to body segments can involve simple visual estimation, or utilisation of calibration pointing devices to identify anatomical landmarks more precisely. Functional MIMU calibration methods, which involve moving joints through ranges of motion, appear to be practical, have good reliability and improve measurement validity. However, they have not been sufficiently investigated for the thorax or lumbar segments. The impact of functional calibrations on segment and joint angles during high-speed, multi-planar tasks has also not been documented. **Study 2** assessed the concurrent measurement validity of five different joint angle calculation methods, three of which incorporated functionally-defined axes. Functional axes of rotation were defined and applied to three different methods of calculating thorax and lumbar global segment angles, thorax-to-pelvis relative angles and lumbar-to-pelvis relative angles. Two non-functional methods (Grood and Suntay joint coordinate system and a basic Euler angle decomposition) were also used to calculate the same relative and global angles. A 3D retro-reflective motion capture system and a MIMU system concurrently recorded the movements of 10 fast-medium bowlers as they performed uni-planar trunk range of motion (ROM) trials and dynamic bowling trials. The 3D retro-reflective modelled joint angles were used as the criterion measure. The results indicated that MIMU joint angle calculation methods that incorporated functionally-defined segment axes did not significantly improve the concurrent validity of MIMU derived thorax and lumbar joint angles when compared with the non-functional methods. It may not be necessary to employ functional MIMU calibration methods when quantifying dynamic thorax and lumbar kinematics in cricket bowling, with simple approaches adequate (e.g., basic Euler angle decomposition).

Sensor specifications and sensor fusion algorithms can vary considerably between commercial MIMU systems. Previous studies have shown that these discrepancies can cause significant differences to sensor orientation estimation during static and low-velocity movement conditions. **Study 3** compared the measurement validity of two similar commercial MIMU systems during fast bowling and trunk ROM tasks. Ten fast-medium bowlers participated in the study. Data were collected separately from each MIMU system but concurrently with 3D retro-reflective motion capture data that acted as the criterion measure. Aligned retro-reflective technical coordinate systems overlaid the MIMUs to facilitate equitable joint angle comparisons between systems. Joint and segment angles were expressed via ZXY-ordered Euler angle decompositions for both MIMU systems and the retro-reflective system (z = medio-lateral, x = anterio-posterior, y = vertical).
Functional calibrations were not utilised following the findings of study 2. Results showed that both MIMU systems were capable of validly measuring uni-planar ROM tasks when assessed against the retro-reflective derived angle outputs. The RMSD values were less than 3 ±1.5° for thorax and lumbar global angles; thorax-to-pelvis and lumbar-to-pelvis relative angles. Overall, the RMSD values during the bowling trials compared well with previous sport-specific MIMU research, although one of the MIMU systems produced RMSDs as high as 7.4±1.9°, which may be considered a functionally relevant difference. The second MIMU system tested did not return RMSDs above 5° for any investigated variable, suggesting it is capable of measuring valid trunk and pelvis motion during cricket bowling. Despite the overall similarities between the MIMU systems, there were some small but significant differences to the RMSD values. The likely cause of these differences were dissimilarities in sensor fusion algorithm parameters and system sampling rates (100 Hz vs 75 Hz). The findings of this study indicate that commercial MIMU systems, with comparable sensor specifications and sensor fusion approaches, can produce significantly different measurements, especially when quantifying segment and joint angles during high-speed, sport-specific tasks.

A large proportion of fast bowling research has reported upper thorax/shoulder, thorax, lumbar, or pelvis kinematics, often by employing 3D retro-reflective anatomically based modelling. Study 4 investigated whether MIMU based measurements can be used as a substitute for 3D retro-reflective anatomically referenced joint angles when assessing thorax, lumbar and pelvis kinematics during fast bowling. Ten fast-medium bowlers had their bowling actions simultaneously recorded by 3D retro-reflective motion capture and MIMUs. Retro-reflective data were then modelled and referenced to anatomical standards, as is the norm in the majority of 3D retro-reflective based sport-specific research. The MIMU data were obtained from the MIMU system that displayed the superior measurement validity in study 3, with ZXY-ordered Euler angle decomposition used to calculate relative and global angles. One-dimensional statistical parametric mapping (SPM1D) showed that 19 of the 21 variables assessed had periods of significant difference during the bowling delivery stride. The mean RMSD value for all variables was 13.9±6.4°, indicating that functionally relevant differences existed. However, there did not appear to be a systematic linear offset between the MIMU and retro-reflective derived angle outputs. Discrepancies to segment coordinate system orientations, soft tissue artefact errors and the lack of skeletal rigidity of certain segments (e.g., thorax) were likely the primary causes of significant differences observed between MIMU and retro-reflective derived outputs. The findings suggest that MIMU based measures of thorax, lumbar and pelvis kinematics should not be directly compared with data in the existing bowling literature. The establishment of MIMU derived specific lumbar injury risk kinematic thresholds are necessary before MIMUs can be utilised as an effective injury risk screening tool by cricket coaches and researchers.
This research provides information on the suitability of portable and cost-effective motion capture technologies for measuring trunk and pelvis angles during high-speed, multi-planar tasks. The outcomes of these studies may serve to provide insight surrounding the validity of field based tools and facilitate advances in field-based biomechanical analysis techniques.
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I, Jacqueline Alderson certify that the student’s statements regarding their contribution to each of the works listed above are correct.

As all co-authors’ signatures could not be obtained, I hereby authorise inclusion of the co-authored work in the thesis.

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<td>0D/1D/2D/3D</td>
<td>Zero/one/two/three dimensions (dimensional)</td>
</tr>
<tr>
<td>ACL</td>
<td>Anterior Cruciate Ligament</td>
</tr>
<tr>
<td>ANOVA</td>
<td>Analysis of variance</td>
</tr>
<tr>
<td>ASIS</td>
<td>Anterior Superior Iliac Spine</td>
</tr>
<tr>
<td>BCE</td>
<td>Before common era</td>
</tr>
<tr>
<td>C7</td>
<td>Seventh thoracic vertebra</td>
</tr>
<tr>
<td>CAST</td>
<td>Calibrated anatomical systems technique</td>
</tr>
<tr>
<td>CI</td>
<td>Confidence interval</td>
</tr>
<tr>
<td>cm</td>
<td>Centimetre</td>
</tr>
<tr>
<td>et al.</td>
<td>et alii (‘and others’)</td>
</tr>
<tr>
<td>e.g.</td>
<td>exempli gratia (‘for example’)</td>
</tr>
<tr>
<td>FA</td>
<td>Functional axis</td>
</tr>
<tr>
<td>F-E or flex.-ext.</td>
<td>Flexion-extension</td>
</tr>
<tr>
<td>FFC</td>
<td>Front foot contact of the cricket bowling delivery stride</td>
</tr>
<tr>
<td>g</td>
<td>Acceleration due to gravity</td>
</tr>
<tr>
<td>GS</td>
<td>Grood and Suntay (1983) joint coordinate system</td>
</tr>
<tr>
<td>Hz</td>
<td>Hertz</td>
</tr>
<tr>
<td>ICC</td>
<td>Intra-class correlation</td>
</tr>
<tr>
<td>i.e.</td>
<td>id est (‘that is’)</td>
</tr>
<tr>
<td>MIMU</td>
<td>Magneto-inertial measurement unit</td>
</tr>
<tr>
<td>kg</td>
<td>Kilogram</td>
</tr>
<tr>
<td>km/h</td>
<td>Kilometres per hour</td>
</tr>
<tr>
<td>L1-L5</td>
<td>Lumbar vertebra (1-5)</td>
</tr>
<tr>
<td>Lat.</td>
<td>Lateral</td>
</tr>
<tr>
<td>Term</td>
<td>Description</td>
</tr>
<tr>
<td>--------------</td>
<td>------------------------------------------</td>
</tr>
<tr>
<td>LF or lat. flex.</td>
<td>Lateral flexion</td>
</tr>
<tr>
<td>m</td>
<td>Metre</td>
</tr>
<tr>
<td>MAD</td>
<td>Mean absolute difference</td>
</tr>
<tr>
<td>Max.</td>
<td>Maximum</td>
</tr>
<tr>
<td>Med.</td>
<td>Medial</td>
</tr>
<tr>
<td>Min.</td>
<td>Minimum</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>ms</td>
<td>Millisecond</td>
</tr>
<tr>
<td>m/s</td>
<td>Metres per second</td>
</tr>
<tr>
<td>m/s²</td>
<td>Metres per second squared</td>
</tr>
<tr>
<td>n</td>
<td>Number (of sample)</td>
</tr>
<tr>
<td>N</td>
<td>Newton</td>
</tr>
<tr>
<td>p</td>
<td>Calculated probability</td>
</tr>
<tr>
<td>PSIS</td>
<td>Posterior Superior Iliac Spine</td>
</tr>
<tr>
<td>r</td>
<td>Correlation coefficient</td>
</tr>
<tr>
<td>RMSD</td>
<td>Root mean square difference</td>
</tr>
<tr>
<td>ROM</td>
<td>Range of motion</td>
</tr>
<tr>
<td>Rot.</td>
<td>Rotation</td>
</tr>
<tr>
<td>RRanat</td>
<td>Retro-reflective anatomically modelled coordinate system</td>
</tr>
<tr>
<td>RRtech</td>
<td>Retro-reflective technical coordinate system</td>
</tr>
<tr>
<td>s</td>
<td>Second</td>
</tr>
<tr>
<td>S1-S5</td>
<td>Sacral vertebra (1-5)</td>
</tr>
<tr>
<td>Sag.</td>
<td>Sagittal</td>
</tr>
<tr>
<td>SD</td>
<td>Standard deviation</td>
</tr>
<tr>
<td>SPM</td>
<td>Statistical parametric mapping</td>
</tr>
<tr>
<td>SPM1D</td>
<td>One-dimensional statistical parametric mapping</td>
</tr>
<tr>
<td>STA</td>
<td>Soft tissue artefact</td>
</tr>
<tr>
<td>-----</td>
<td>---------------------</td>
</tr>
<tr>
<td>T1 – T12</td>
<td>Thoracic vertebra (1-12)</td>
</tr>
<tr>
<td>tmax</td>
<td>Peak SPM t-trace value</td>
</tr>
<tr>
<td>Trans.</td>
<td>Transverse</td>
</tr>
<tr>
<td>UK</td>
<td>United Kingdom</td>
</tr>
<tr>
<td>USA</td>
<td>United States of America</td>
</tr>
<tr>
<td>ZXY</td>
<td>ZXY order of rotations</td>
</tr>
<tr>
<td>α</td>
<td>Alpha</td>
</tr>
<tr>
<td>Δ</td>
<td>Change</td>
</tr>
<tr>
<td>°</td>
<td>Degrees</td>
</tr>
<tr>
<td>°/s</td>
<td>Degrees per second</td>
</tr>
<tr>
<td>%</td>
<td>Percent</td>
</tr>
<tr>
<td>&lt;</td>
<td>Less than</td>
</tr>
<tr>
<td>&gt;</td>
<td>Greater than</td>
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<td>±</td>
<td>Plus/minus</td>
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Since the seminal photographic motion capture studies from Marey (1873, 1887) and Muybridge (1878) in the 19th century, researchers have been constantly striving to improve the practicality, validity and reliability of human movement analysis methods. Two-dimensional (2D) film was the pre-eminent motion capture technology for much of the 20th century, before it was eventually superseded by analog, and then digital, video. Major improvements to computing technology also facilitated the development of vision-based three-dimensional (3D) motion capture technologies. From a measurement validity and accuracy perspective, the current gold-standard of motion analysis within sports biomechanics is 3D retro-reflective based anatomical modelling (van der Kruk and Reijne, 2018; Windolf et al., 2008). This approach enables biomechanists to effectively analyse the complex, high-speed motions, often exhibited in sports. Retro-reflective motion capture involves placing reflective markers on body segments and recording their positions throughout a calibrated 3D volume. The modelling procedures employed in conjunction with retro-reflective motion capture allow researchers to construct individual-specific and functionally relevant biomechanical models, which, among other functions, incorporate equations for approximation of 3D segment and joint angles. Advanced modelling innovations, such as functionally calibrated axes of rotation (Besier et al., 2003) and the calibrated anatomical systems technique (Cappozzo et al., 1996) have been incorporated into retro-reflective biomechanical models in an effort to address the two major sources of measurement error: anatomical landmark misidentification and soft tissue artefact (Cappozzo et al., 2005; Della Croce et al., 2005; Leardini et al., 2005). Unfortunately, 3D retro-reflective motion capture systems are cumbersome and expensive, and require specialised training for effective use. They are generally unsuitable for field-based motion capture, meaning biomechanical data are usually collected in a controlled laboratory environment. Whilst this may not be especially concerning for clinical analyses such as gait, many sporting actions are performed in environments that are drastically different to indoor laboratories (e.g., playing fields, swimming pools). Consequently, ecological validity is a major limitation of the 3D retro-reflective motion capture and modelling approach.

Motion capture technologies that are suitable for field-based use, and that also demonstrate high measurement validity and reliability, are valuable tools to biomechanists. Two-dimensional video is an example of a motion capture technology that is suitable for field-based use. Unlike retro-reflective systems, digital video cameras are affordable, user-friendly and easily portable, meaning they are more appropriate for use outside of the laboratory. However, the measurement validity of 2D video-based digitisation is questionable, particularly for complex, multi-planar movements.
Measurement validity refers to the degree to which a measurement approach succeeds in quantifying what it is designed to measure. For human movement analysis, measurement validity assessments usually involve comparison with retro-reflective motion capture criterion measurements. Compared with retro-reflective based modelling, video-based digitising employs fairly simplistic approaches to measuring complex 3D movements. Two-dimensional vectors are manually digitised to represent body segments, with the angles created between these vectors (2D vector angles) measured as approximations of joint angles. Measuring complex 3D motions from a 2D perspective means that perspective error is also a major limitation of 2D video-based movement analysis. Studies have shown that 2D video-based measurement validity for sport-specific tasks varies depending on the movement and body segment/joint being analysed (Aginsky and Noakes, 2010; Elliott et al., 2007; Hanley et al., 2018; Lopes et al., 2018).

The proliferation of wearable technology over the last decade has seen magneto-inertial measurement units (MIMUs) become an increasingly utilised sports analytical tool (Camomilla et al., 2018). A MIMU incorporates tri-axial accelerometers, gyroscopes and magnetometers. The data from the three sensor types are fused by filtering algorithms which compute the 3D orientation of the unit. The units are highly portable and, unlike 2D video, facilitate 3D measurement of complex motion. They are also unaffected by the occlusion issues that impact all vision-based motion capture techniques. Unfortunately, there are many factors that can impact the measurement validity of MIMU systems. Commercial MIMU manufacturers generally don’t disclose information regarding the specific structure of sensor fusion algorithms, which can be a major concern for researchers. Calibrating MIMUs to body segments so that MIMU measurements are functionally meaningful is also a key consideration. Anatomical and functional approaches to MIMU-to-segment calibration have been described. Anatomical MIMU calibration can involve simple visual alignment to a segment or more precise alignment with anatomical landmarks using customised calibration instruments (Picerno et al., 2008). Functional approaches involve axes of rotation being calculated after the joint of interest has been moved through a functional range of motion (Picerno, 2017). There is currently no consensus on the best approach to MIMU sensor-to-segment calibration. The studies that have reported on calibration procedures have almost exclusively done so for the upper (Cutti et al., 2008; de Vries et al., 2010; Galinski and Dehez, 2012; Ligorio et al., 2017) and lower limbs (Cutti et al., 2010; Favre et al., 2009; O’Donovan et al., 2007; Picerno et al., 2008).

Previous research has shown that MIMU measurements can vary considerably depending on the commercial MIMU model utilised, with other factors, such as the sensor fusion algorithm employed, also capable of influencing measurements (Bergamini et al., 2014; Filippeschi et al., 2017; Lebel et al., 2013, 2015; Ligorio et al., 2016; Picerno et al., 2011). These studies have primarily been conducted during static or low-velocity movement conditions, with a comparison of
measurement accuracy during higher-velocity motion reported only in robotic-based studies (Lebel et al., 2013, 2015).

Two-dimensional video cameras and MIMUs have characteristics that are well suited to field-based motion capture, including portability, affordability and ease-of-use. However, the measurement validity for both modalities can be variable and dependent on multiple factors. A context where both technologies could potentially be implemented is for cricket fast bowling biomechanical analysis. Cricket fast bowling is a complex skill that involves high-speed, multi-planar movements of the trunk and pelvis through large ranges of motion. Cricket fast bowlers have a lumbar spondylolysis (bone stress injury to the pars interarticularis) prevalence of between 11 and 67% (Annear et al., 1992; Crewe et al., 2012; Elliott et al., 1992; Foster et al., 1989; Gregory et al., 2004; Hardcastle et al., 1992); considerably higher than prevalence in the general population (4-6%) (Beutler et al., 2003; Fredrickson et al., 1984). Fast bowlers also commonly display accelerated intervertebral disc degeneration. The presence of degeneration in a cohort of young bowlers was reported to increase by 37% in under three years (Burnett et al., 1996).

Perhaps unsurprisingly given the dynamic nature of the movement, fast bowling biomechanics have been strongly linked with onset of lumbar injury (Forrest et al., 2017; Johnson et al., 2012; Olivier et al., 2016). The majority of injury research has employed 3D retro-reflective motion capture to analyse bowling biomechanics. Numerous kinematic factors have been associated with lumbar injury risk, with many of these variables involving thorax, lumbar and pelvis angles. Despite the high risk of lumbar injury among fast bowlers, bowling biomechanical screening processes have surprisingly not been well explored. It is unknown whether 2D video or MIMUs could be validly used to measure the kinematics associated with fast bowler lumbar injury outside of the confines of a laboratory.

This thesis aimed to investigate alternative, field-based motion capture options, with a particular focus on the measurement validity of such systems. Two-dimensional video digitisation and MIMU-based model angle outputs were compared with gold-standard, 3D retro-reflective based biomechanical model outputs of trunk, lumbar and pelvis angles during complex, high-speed sporting motions. Cricket fast bowling provided the key reference task for all of the studies contained within the thesis. Information is also provided on the suitability of 2D video and MIMU systems as potential field-based lumbar injury risk screening tools for cricket fast bowlers.

1.1 Statement of the Problem

1) Alternative methods of measuring high-speed, multi-planar movements, such as cricket fast bowling, have not been well validated in the literature. It is unknown whether 2D video
or MIMU-based measurements of thorax, lumbar and pelvis angles during the bowling action are reflective of angles derived using gold-standard 3D retro-reflective anatomically based models.

2) Functional calibration of MIMUs to body segments has been achieved for the upper and lower limbs but has not been sufficiently explored for the trunk (thorax and lumbar) segment. The effect of incorporating functionally defined axes of rotation in MIMU angle calculation methods has also not been thoroughly investigated for high-speed movement tasks.

3) Measurement comparisons of MIMU systems have focused primarily on static or low-velocity movement conditions. A comparison of commercial MIMU system measurements during sport-specific, high-velocity movements has not been well documented.

1.2 Research Aims

The general aim of this thesis was to provide information on the suitability of field-based motion capture technologies for measuring thorax, lumbar and pelvis angles during high-speed, multi-planar movements. Specifically, the research aimed to answer four questions: 1) Can digitised 2D video be used to validly measure fast bowling kinematics (segment and joint angles) associated with lumbar injury risk? 2) What impact does MIMU functional calibration have on thorax and lumbar joint angles for uni-planar and high-speed, multi-planar movement tasks. 3) Does measurement validity significantly vary between comparable commercial MIMU systems when quantifying angles of the thorax and lumbar segments during lower-speed uni-planar range of motion tasks and high-speed, complex sporting motions? 4) What is the nature and magnitude of the measurement differences between MIMU-based modelling and retro-reflective based anatomical modelling when assessing thorax, lumbar and pelvis angles during fast bowling?

1.3 Significance of the Research

This research serves to broaden our current understanding of field-based approaches to human motion capture for dynamic, sport-specific contexts by exploring a range of different factors which may impact the measurement validity of motion analysis. The research is highly applicable to biomechanists, with the information provided implementable for all MIMU and video based analyses of trunk and pelvis kinematics. Insight is also specifically provided on the suitability of 2D video and MIMUs as portable, cost-effective options to kinematic-based screening of lumbar injury kinematic risk factors in cricket fast bowlers.
1.4 Thesis Outline

This thesis comprises of a series of papers investigating field-based motion capture options measuring thorax, lumbar and pelvis kinematics, during high-velocity, sporting tasks. The experimental research is presented in chapters 4 to 7 and follows chapters which review the relevant literature (chapter 2) and expand on the methodologies utilised in the thesis (chapter 3). Chapter 4 assesses the measurement validity of 2D video-based analysis when quantifying kinematics associated with lumbar injury risk in cricket fast bowlers. Chapters 5 and 6 explore multiple factors that may affect measurement validity of MIMU-based modelling. Chapter 5 evaluates the impact of functional MIMU calibrations on thorax and lumbar angle measurements during uni-planar and high-speed multi-planar tasks. Chapter 6 compares two similar commercial MIMU systems and determines whether their measurements of trunk kinematics vary significantly during high-speed, multi-planar sporting motions. Chapter 7 assesses the relationship between MIMU and anatomically modelled retro-reflective outputs of thorax, lumbar and pelvis kinematics during cricket fast bowling.

Linking statements are provided between each of the experimental research chapters to clearly outline the reasoning of each investigation and to connect to the previously presented studies in the thesis.

1.4.1 Chapter 2: Review of the Literature

The literature review provides a thorough overview of the previously published research on 2D video motion capture, 3D retro-reflective motion capture and MIMU based motion capture, specifically focusing on sporting biomechanical analysis. The associated modelling techniques are also explored for each modality. The chapter concludes with a brief overview of cricket bowling biomechanical research and the link between fast bowling and lumbar injury.

1.4.2 Chapter 3: Extended Methods

This chapter includes some additional information on the research design and procedures employed throughout the thesis.

1.4.3 Chapter 4: Measuring Bowling Kinematics for Two-Dimensional Video

The aims of this study were to:

- Determine whether multi-plane 2D video of fast bowling can be digitised to validly measure an array of different fast bowling segment and joint angles associated with lumbar injury risk.
• Evaluate the suitability of 2D video motion analysis as a kinematic-based screening tool for lumbar injury risk in cricket fast bowlers.

The hypotheses tested were:

• That discrete measurements of trunk, pelvis, hip and knee angles recorded from multi-plane 2D video would be comparable with anatomically modelled 3D retro-reflective outputs of equivalent discrete variables during the fast bowling action.
• That 2D video motion analysis could be used as effective screening tool for identifying lumbar injury risk in cricket fast bowlers.

1.4.4 Chapter 5: Functionally Calibrated MIMU-based Thorax and Lumbar Measurements

The aims of this study were to:

• Evaluate the impact of MIMU functional calibration on thorax and lumbar joint angles for uni-planar and multi-planar, high-speed tasks.
• Determine whether MIMU functional calibration significantly improves measurement validity of MIMUs compared with simple anatomical calibration, when measuring thorax and lumbar angles during uni-planar range of movement tasks and cricket fast bowling.
• Ascertain whether a MIMU-based ZXY-ordered Euler angle decomposition for thorax and lumbar angles would produce greater measurement differences than MIMU angle calculations incorporating functional axes when both approaches were compared with 3D retro-reflective data modelled via the same angle calculation processes.

The hypotheses tested were:

• That MIMU functional calibration would significantly improve MIMU measurement validity when compared with simple anatomical calibration for measurements of thorax and lumbar angles during uni-planar range of movement tasks and cricket fast bowling.
• That a MIMU-based ZXY-ordered Euler angle decomposition for thorax and lumbar angles would produce greater measurement differences than functionally calibrated angle calculations when both methods were compared with 3D retro-reflective derived angles.

1.4.5 Chapter 6: Comparison of Trunk Angle Measures from Two Commercial Systems

The aims of this study were to:

• Assess the concurrent measurement validity of two commercial MIMU systems by comparing thorax and lumbar angles from both systems to gold-standard, 3D retro-reflective motion data angles.
• Determine whether measurement validity significantly varied between the two MIMU systems when measuring thorax and lumbar angles during uni-planar trunk range of motion tasks and cricket fast bowling.
• Ascertained whether MIMU measurement validity is dependent on the velocity and complexity of the movement task being captured.

The hypotheses tested were:

• That concurrent measurement validity of two similar, commercial MIMU systems would not significantly vary when thorax and lumbar angle measurements from both systems were compared with 3D retro-reflective angle outputs.
• That thorax and lumbar angle measurement validity of both MIMU systems would reduce as movement velocity and complexity increase.

1.4.6 Chapter 7: MIMU-based Measurement of Trunk and Pelvis Angles during Cricket Bowling

The aims of this study were to:

• Quantify the measurement differences between MIMUs and 3D retro-reflective based anatomical modelling when assessing thorax, lumbar and pelvis angles during fast bowling.
• Determine whether there is a systematic linear offset between the measurements of MIMUs and 3D retro-reflective based anatomical modelling when assessing thorax, lumbar and pelvis angles during fast bowling.
• Assess whether MIMU data can be compared with 3D retro-reflective data in the previous fast bowling literature.

The hypotheses tested were:

• There would be significant measurement differences between MIMUs and 3D retro-reflective based anatomical modelling when assessing thorax, lumbar and pelvis angles during fast bowling.
• That there would be no evidence of a systematic linear offset between the measurements of MIMUs and 3D retro-reflective based anatomical modelling when assessing thorax, lumbar and pelvis angles during fast bowling.
• That MIMU derived angles would not be comparable with 3D retro-reflective derived angles in the previous fast bowling literature.
1.4.7 Chapter 8: Synthesis of Findings and Conclusions

This chapter synthesises the findings of the thesis, with reference to the aims and hypotheses for each study. Recommendations are also presented for future research on field-based motion capture solutions.

1.4.8 Chapter 9: Supplementary Materials

This chapter includes supplementary data for some studies. Copies of the research ethical approvals, study information sheets and participant consent forms are provided. Details and copies of peer-review conference abstracts are also included.

1.5 Limitations and Delimitations

1.5.1 Limitations

The reader should be cognisant of the following limitations when evaluating the results of this research:

1.5.1.1 Chapters 4, 5, 6 and 7

- Cricket bowling is highly specific sporting skill. Results concerning this task may not translate to all other sporting contexts.
- Three-dimensional retro-reflective motion capture angles were used as the criterion measure for all chapters. Though this technology is the current motion capture gold-standard in the sports biomechanics field, it should not be considered a ground-truth measure. Consequently, the studies present concurrent measurement validity information for 2D video (Chapter 4) and MIMUs (Chapters 5, 6 and 7), rather than assessments of measurement accuracy, which would require a ground-truth joint angle measure (e.g., high speed fluoroscopy).
- All data were collected within a laboratory environment for the purpose of this criterion-based validity research. It is likely that movements performed in a laboratory may not completely reflect movement patterns observed in standard sporting environments.

1.5.1.2 Chapter 4

- The video cameras utilised in this study were a combination of high-speed (250 Hz) cameras precisely synchronised to the retro-reflective motion capture system and standard speed (50 Hz) cameras. These inconsistencies are acknowledged in the study and measures from the two types of video camera were not directly compared.
1.5.1.3  Chapters 5, 6 and 7

- The MIMU data presented in these studies were collected in an indoor motion analysis laboratory. The study also necessitated use of force platforms which may have provided some ferromagnetic interference to magnetometer data.
- Gyroscope saturation was not specifically assessed for the MIMU data, although no evidence of this problem was identified in the data presented in the thesis.
- There was a limitation to the amount of MIMU units available for the research presented in this thesis. This consequently limited the number segments that could be analysed. It is one of the main reasons that knee and hip angles are only measured in chapter 4 and not the subsequent chapters.

1.5.1.4  Chapter 5

- There was not an angle calculation method that could be considered a gold-standard or ground-truth for this study. Consequently, each MIMU angle calculation method was compared only with the same calculation method applied to retro-reflective data, allowing for direct between model comparisons.

1.5.1.5  Chapter 6

- The datasets for both of the analysed MIMU systems did not contain entirely the same cohort of participants, due to participant and MIMU system availability. Efforts were made to match the two datasets by including participants of comparable height and body mass.
- It was not possible to collect data from the two MIMU systems simultaneously without introducing variation to error sources associated with MIMU placement and soft tissue movement artefact.

1.5.2  Delimitations

The following delimitations were imposed on the research presented in the thesis:

1.5.2.1  Chapters 4, 5, 6 and 7

- The research in this thesis was concerned primarily with the measurement validity of different motion capture modalities. The measurement reliability of these methods was not specifically assessed.
- The main reference task for all studies was cricket fast bowling. The findings that involve this movement may not translate to all other movement tasks. Trunk range of motion tasks were also assessed in Chapters 5 and 6 and provide some comparison.
- The thorax, lumbar and pelvis segments were the key focus of this thesis. Findings may not be relevant to other body segments and joint angles.
1.6 References


Chapter 1: Introduction


2.1 Introduction to human motion analysis

The depth of humanity’s understanding of human motion has always been a reflection of the motion capture modalities available at a given time. Aristotle provided one of the first known written references to the analysis of human movement:

“If a man were to walk on the ground alongside a wall with a reed dipped in ink attached to his head the line traced by the reed would not be straight but zig-zag, because it goes lower when he bends and higher when he stands upright and raises himself.” – Aristotle (384-322 BCE) (Baker, 2007)

Unfortunately, Aristotle’s musings were never scientifically tested during his lifetime, primarily due to the fact that Galileo Galilei did not describe the modern scientific method until the 1620s. Clearly our analyses of human motion have become progressively more sophisticated, with advancements particularly evident in the past two centuries. The Weber brothers’ trailblazing observational work on human motion (Weber and Weber, 1836), were followed by the early motion capture studies from Jules Etienne Marey (1873; 1887) and Edward Muybridge (1878) later in the 19th century. The two contemporaries greatly progressed the field of biomechanics by developing different techniques to capture movements that were too fast for the human eye to track. Both men famously showed that there was a period during the trot of a horse that all four hooves were off the ground, a fact previously unperceivable by the human eye. Marey did this initially via use of pressure sensors attached to the horses’ hooves (1873), before Muybridge utilised a series of rudimentary photographic cameras to capture images of the entire trot cycle (1878). Inspired by Muybridge’s work, Marey went on to refine photographic technology by introducing a shutter to the camera so that multiple images could be captured on the same photographic plate, producing what became known as the chronophotograph (Marey, 1887). Figure 2.1 shows images taken from Marey’s 1887 publication on chronophotography and provides one of the earliest examples of human sporting movements being visually captured. Marey continued to develop photographic techniques after the invention of photographic film which eventually led to a version of the cine camera, although the Lumiè re brothers are generally credited with this invention (Baker, 2007). Otto Fischer and Willhelm Braune (1895) performed the first three-dimensional (3D) analysis of human motion in the early 1890s. Their method involved strapping Geissler tubes to the body prior to walking in a darkened room, where movements were captured by multiple cameras with continuous exposures (Baker, 2007). Despite this early example of 3D motion analysis, two-
dimensional (2D) film-based analysis of movement continued to be the pre-eminent method of motion capture for much of the 20th century, with more sophisticated methods of 3D analysis limited by the available technology. Thankfully, the 1970s witnessed considerable advancements in computer technology which enabled analysis of 2D and 3D human motion to become increasingly accurate, less cumbersome and more time-efficient (Baker, 2007). Camera-based 2D and 3D motion capture systems remain cornerstones of modern biomechanics. However, 21st century developments in wearable technologies, such as magneto-inertial measurement units (MIMUs), have provided biomechanists with exciting new motion capture alternatives.

![Figure 2.1: E.J. Marey's early chronophotographs of an individual running and jumping. Adapted from "Photography of moving objects, and the study of animal movement by chrono-photography" by E. Marey, 1887, Scientific American Supplement, 579, pp 9245.](image)

Since initial investigations in the 1800s, biomechanical analyses have been undertaken for almost any conceivable movement task and context, yet analysis of athlete biomechanics during sporting tasks remains one of the great dilemmas, especially from a validity and reliability perspective. Accurately capturing and modelling the complex, high-velocity movements observed in sporting activity provides a unique challenge. Ecological validity is also a major concern, particularly for movements that are performed in vastly different environments to a laboratory setting, such as on a field or in water.
This chapter will review multiple methodological approaches to sports biomechanical analysis. An overview of gold-standard, 3D retro-reflective motion capture and 2D digital video motion capture will be followed by a discussion of MIMUs and their potential applications. The associated modelling and measurement techniques for these technologies will also be examined, with the measurement of the thorax, lumbar and pelvis segments given particular attention. The review concludes with a discussion of cricket fast bowling: a complex sporting task that involves dynamic trunk and pelvis motion. Cricket fast bowling biomechanics will be briefly reviewed, including their association with lumbar injury risk. Finally, a brief evaluation of the use of motion capture techniques in cricket bowling research will also be presented.

2.2 Two-Dimensional Video Motion Capture

Two-dimensional film was the pre- eminent image-based motion capture method for the majority of the 20th century before it was superseded by analog, and then digital, video. Whilst 3D retro-reflective motion capture is the current motion capture gold-standard (van der Kruk and Reijne, 2018), digital video analysis remains commonly employed in many contexts; from clinical settings to sporting fields. This review will focus solely on sporting applications. Video footage can be used to provide almost immediate qualitative biomechanical feedback to athletes (Wilson, 2008). Commercial software packages are also available that enable quantification of kinematics from 2D video via simple digitising processes. Simple 2D vectors are utilised to represent body segments, with angles between the vectors (2D vector angles) measured to approximate a joint angle. Kinematic markers are often placed over joints prior to data collection to facilitate easier identification of joint positions during digitising. Regardless of whether simple qualitative inspection or systematic quantitative analysis of video footage is the intended purpose, 2D video capture provides a cost-effective, portable, and user-friendly solution to motion capture (Pueo, 2016).

Despite its practicality, 2D video based biomechanical analysis has considerable limitations, especially with regard to measurement validity and reliability. In simple terms, measurement validity refers to the degree to which a measurement approach succeeds in quantifying what it is designed to measure (Morgan et al., 2001). True measurement accuracy is frequently unquantifiable in human movement analysis as genuine ground-truth measures are often unattainable. Consequently, accepted gold-standard measurement systems (e.g., 3D retro-reflective motion capture and modelling) are instead used as the criterion measure for new or alternative measurement modalities. To make this process equitable, measurements are generally taken by both modalities simultaneously, hence the term ‘concurrent validity’ (Morgan et al., 2001). Measurement reliability is also a consideration for motion analysis. Reliability refers to the
consistency of a measurement, regardless of accuracy or validity. Inter-tester and intra-tester reliability are of particular interest for video-based analysis. Though studies have shown video-based kinematic analysis can be reliable (Portus et al., 2006; Weir et al., 2019), it often suffers from poor measurement validity because there are many possible sources of measurement error. These sources of error will be briefly discussed below, with potential solutions also provided.

Sources of 2D video based measurement error can be broadly categorised into system (camera) or modelling/digitising error (Elliott and Alderson, 2007). Selecting appropriate camera specifications and settings for a data collection is important for reducing potential sources of error. Standard video camera sampling rates are usually between 50 Hz and 60 Hz (images per second) (Garhammer and Newton, 2013). A standard sampling rate may be adequate for kinematic analysis of slower tasks but will likely be insufficient for high-velocity movements, such as those exhibited during dynamic sporting skills (e.g., throwing, running, jumping, kicking and so on). Attempting to measure kinematics continuously at an insufficient sampling rate for the speed of the movement can lead to temporal aliasing of kinematic waveforms, resulting in misinterpretation. Identifying critical events or measuring discrete values can also be inaccurate at insufficient sampling rates (Garhammer and Newton, 2013). Utilising cameras with higher sampling frequencies (> 100 Hz) is recommended for high-speed sporting motions (Pueo, 2016). Similarly, selecting higher shutter speeds for objects or body segments moving at high velocities is important to minimise motion blur (Pueo, 2016). Ambient light conditions, aperture settings and camera focus should also facilitate a clear, well-lit image. Another consideration for a successful video camera set-up is camera positioning. Perspective error or ‘planar crosstalk’ (Elliott and Alderson, 2007) is a major cause of error when kinematic measurements are taken from a camera that is not in plane with the movement direction. Camera placement therefore demands careful deliberation prior to data collection. Utilising two or more cameras to capture movement from multiple planes (e.g., sagittal and coronal) can increase the likelihood that movements will occur in plane with a camera.

As previously described, joint angles are typically measured from 2D video by calculating the angle between simple 2D vectors used to represent body segments. The endpoints of the vectors are usually approximated joint centre positions. Markers are often placed on the skin directly over joints to assist with the identifying of joint centres. Unfortunately, skin movement over joints can lead to significant changes to the position of the marker relative to the true joint centre or axis (Leardini et al., 2005). Measurement errors that are caused by movement of soft tissue over joints and bones are known as soft tissue artefact (STA). It has been repeatedly shown that STA is the largest cause of measurement error in marker-based motion capture systems (Leardini et al., 2005). Soft tissue artefact will be discussed in more detail in section 2.3.1. The alternative to placing skin markers over joints is to simply visually approximate the joint axis for each individual video frame.
Whilst this may somewhat negate the effect of STA, accurately and reliably locating joint centres or anatomical landmarks can be difficult, especially for joints such as the hip (Hanley et al., 2018). Misidentification of joint axes or landmarks by only a few millimetres has the potential to alter vector orientations and resultant kinematic outputs (Della Croce et al., 2005). Athlete clothing can also occlude joints which further complicates the joint centre identification process.

It is important to note that STA and anatomical landmark misidentification also affect other vision-based motion capture methods, including the motion capture gold-standard, 3D retro-reflective motion capture (Della Croce et al., 2005; Leardini et al., 2005). However, perspective error (planar crosstalk) is a problem that is specific to 2D video motion analysis. Previous research has shown that 2D video analysis can produce relatively valid measures for sporting kinematics that occur entirely in-plane with a camera (Dicesare et al., 2014; Hanley et al., 2018; Weir et al., 2019). For example, knee flexion-extension during gait can be measured from a sagittal plane camera. However, multi-planar movements are common during sporting tasks, especially of the upper body, and are more difficult to measure using this simple approach. As already alluded to, perspective error or planar crosstalk frequently occurs when 2D methods are employed to measure complex 3D movements. An example of planar crosstalk is often observed in cricket bowlers who are accused of an illegal bowling action. Illegal bowling actions are characterised by excessive elbow extension ranges (> 15°). Some bowlers present with large elbow carry angles (fixed elbow abduction) which can easily be misinterpreted as elbow flexion-extension when viewed or recorded non-planar positions (Aginsky and Noakes, 2010; Elliott et al., 2005, 2007; Elliott and Alderson, 2007; Lloyd et al., 2000). This phenomenon has resulted in the ruling that all official analyses of elbow flexion-extension during cricket bowling are to be undertaken with 3D retro-reflective motion capture, illustrating a major limitation of 2D video-based motion analysis.

2.2.1 Two-Dimensional Video as a Screening Tool

Despite its limitations, video based 2D analysis may be suitable for large scale injury risk screening programmes due to its accessibility, portability and usability. The last decade has witnessed multiple attempts to establish 2D video based injury risk screening methods, particularly in relation to anterior cruciate ligament (ACL) injury (Dicesare et al., 2014; Dingenen et al., 2015a, 2015b, 2014; Hewett et al., 2009; Lopes et al., 2018; McLean et al., 2005; Schurr et al., 2017; Sorenson et al., 2015; Weir et al., 2019). A recent meta-analysis of such studies (Lopes et al., 2018) reported that utilising 2D video based methods to calculate frontal plane lower limb and trunk kinematics during dynamic tasks is reliable but measurement validity varies depending on the sporting task. During single-leg squats, the authors reported no correlation between 2D knee frontal plane projection angles and 3D knee frontal plane angles ($r = 0.13, p = 0.09$) but moderate-good correlations were documented for the same angles during vertical drop jumps ($r = 0.62, p < 0.001$).
In other studies, 2D measures of trunk lateral flexion have been moderately correlated \((r = 0.62, p = 0.003)\) with 3D measures during jump landings (Dicesare et al., 2014). Furthermore, 2D measures of trunk lateral flexion during side-stepping tasks have also shown to be reliable (Weir et al., 2019). Whilst not related to ACL injury risk, a recent study of knee flexion-extension angles during race walking reported small root mean square differences (RMSD) \((4\pm2^\circ)\) between manually digitised video based angles and 3D retro-reflective based anatomically modelled angles (Hanley et al., 2018). Video-based screening in cricket bowling has focused on measuring elbow flexion-extension angles. As explained above, despite it being potentially possible to reliably record elbow angles during bowling (Portus et al., 2006), perspective errors make valid measurement difficult (Aginsky and Noakes, 2010; Elliott et al., 2005, 2007; Elliott and Alderson, 2007; Lloyd et al., 2000). Collectively, these screening studies suggest that 2D video-based measurements of sport-specific kinematics can reflect 3D retro-reflective anatomical modelled values but it appears that 2D video-based measurement validity is task and joint/segment dependent.

2.3 Three-Dimensional Retro-Reflective Motion Capture

Three-dimensional retro-reflective motion capture is accepted as the current non-invasive, gold-standard of motion capture due to its sub-millimetre measurement accuracy within 3D space (Elliott et al., 2007; van der Kruk and Reijne, 2018; Windolf et al., 2008). Modelling procedures applied to retro-reflective data are extensive and have high anatomical specificity. Retro-reflective motion capture involves the tracking of kinematic marker positions within a calibrated 3D volume. Multiple near infra-red cameras emit light which is reflected off the markers and captured by the cameras. Markers are usually affixed to the skin overlaying specific anatomical landmarks or body segments. The marker positions are then utilised in biomechanical models, ensuring segment and joint kinematics are specific to the individual’s anatomy.

There are many advantages of utilising 3D retro-reflective systems for sporting applications. The systems are capable of much higher sampling rates (150 Hz to 400 Hz) than standard 2D video cameras, whilst perspective errors are also not a concern for 3D retro-reflective base modelling. However, optimal camera positioning and calibration remains crucial for accurate measurement of marker positions (Chiari et al., 2005).

Whilst 3D retro-reflective systems are the current motion capture gold-standard for measurement validity and accuracy, they do suffer from major practical limitations which have prevented the systems from becoming entirely ubiquitous. Firstly, they are expensive and require specialised training to utilise effectively. This greatly restricts the availability of 3D retro-reflective based analysis to athletes and coaches, especially those below the elite levels of sport. Similar to video-
based motion capture, marker-based motion capture is affected by occlusion issues, especially during highly-dynamic movements. Retro-reflective systems are also generally restricted to laboratory environments due to their lack of portability, restrictive capture volumes and sensitivity to daylight (van der Kruk and Reijne, 2018). Hence, there are serious concerns about the ecological validity of data that are obtained via 3D retro-reflective motion capture. This is particularly relevant for sports biomechanics, as most athletes compete or perform in environments that are vastly different to a biomechanics laboratory. Additionally, the need to affix markers on the body may also have an impact on an athlete’s movement patterns (Elliott and Alderson, 2007). Despite these limitations, there is currently no comparable field-based alternative to 3D retro-reflective motion capture.

2.3.1 Three-Dimensional Retro-Reflective Based Modelling

Easily portable motion capture solutions have been largely unable to replicate the anatomical and functional specificity of 3D retro-reflective based anatomical modelling. Once marker positions have been recorded by a retro-reflective system, data are modelled so that biomechanical variables of interest can be measured or estimated. A mathematical model can be defined as:

“...an abstraction of a real-life scenario, system or event that uses mathematical language to describe and predict the behaviour, dynamics and evolution of said scenario, system or event. Mathematical modelling is thus the step-by-step process of performing this abstraction from real lift scenarios and formulas we can use to infer their characteristics” (Vargas, 2017).

This section will focus on the measurement of segment and joint kinematics via 3D retro-reflective based modelling, with a brief discussion on the strengths and weaknesses of current relevant modelling techniques. Particular comment will be made on trunk and pelvis modelling procedures where appropriate.

In biomechanical modelling, body segments are considered links in a kinematic chain. Body segments are comprised of bone overlayed with soft tissue (e.g., muscles, tendons, skin, and adipose tissue). Most biomechanical models work under the assumption that body segments are rigid entities, which means classical Newtonian mechanics can be employed to solve kinematic equations (Cappozzo et al., 2005). Calculating a segment’s position relative to another segment within a kinematic chain requires each segment to be represented as a bone-embedded reference frame (segment coordinate system). To estimate a joint angle, the orientation of a segment (child) is calculated relative to another segment (parent), whilst both segment reference frames exist within a common global reference frame (global coordinate system).
Euler angle decomposition (rotations) can be used to calculate the orientation of a segment relative to another segment (relative angle) or the orientation of a segment relative to the global reference frame (global angle). Euler rotations decompose a 3D angle into separate measures for each of the three planes of motion. This makes Euler angles easily interpretable from a functional point of view. However, the order in which these components are defined can affect the measurement values (order effect). Euler angle decompositions often occur in order of decreasing range of motion (Cole et al., 1993; Godwin et al., 2009). A common Euler angle rotation order in biomechanics is ZXY, where the z-axis is medio-lateral (flexion-extension), x-axis is anterio-posterior (abduction/adduction) and y-axis vertical (axial rotation). Joint coordinate systems that utilise a common ‘floating’ axis that is not embedded in either segment (parent and child) are not affected by the order of rotations (Grood and Suntay, 1983). Hence, the Grood and Suntay joint coordinate system is recommended by the International Society of Biomechanics (Wu et al., 2002, 2005).

Whilst the order effect is not a concern in this method per se, measurements obtained from joint coordinate systems are affected by which axis (X, Y or Z) is defined as the floating axis (Woltring, 1994).

Another disadvantage of Euler angle decomposition is that it is susceptible to measurement singularities (gimbal lock) (Rau et al., 2000; Woltring, 1994). In these instances, the resulting angles can fluctuate by 180° even though the joint movement is only small. Quaternions are an alternative means of defining orientation that are not affected by the order of rotations or singularity issues (Brückner et al., 2014). Unfortunately, they are not easily interpretable from a functional perspective and are consequently rarely used in retro-reflective based modelling.

The assumption that body segments are entirely rigid is clearly not realistic. Whilst soft tissue structures don’t adhere to this assumption, in some cases, the skeletal structure of a modelled segment can also violate the rigid body notion. This is a particularly relevant for segments that are comprised of multiple smaller bones and joints, such as the hands, feet and spine. Valid and reliable modelling of these structures, in particular the spine, is a major challenge facing biomechanists (Crewe et al., 2013a; Kudo et al., 2017). In many cases, spinal regions (e.g., thoracic, lumbar) are modelled as single segments for simplicity (Crewe et al., 2013a; Wu et al., 2005). Cadaveric studies have reported thoracic spine range of motion between T1 and T12 of 28° for flexion-extension, 36° for lateral flexion, and 45° for rotation (Borkowski et al., 2016). This illustrates one of the problems with modelling spinal regions as single segments, especially for movement tasks that involve large spinal ranges of motion.

Though some segments may be prone to errors related to skeletal rigidity, the two major causes of measurement error associated with retro-reflective based anatomical modelling are anatomical landmark misidentification and STA (Cappozzo et al., 2005). Anatomical landmark identification can
vary considerably within and between testers. Even small misplacements of reflective-markers on landmarks can alter segment or joint vector orientations significantly, with the final consequence being an effect on joint or segment kinematics (Della Croce et al., 2005). Naturally, some anatomical landmarks are more difficult to identify than others, which means this form of error may affect different joints or segments disparately (Della Croce et al., 2005).

As previously mentioned, STA relates to the movement of skin-mounted markers relative to the underlying bone or joints they represent. Markers that are placed directly on joints are affected primarily by skin deformation and translation (sliding), whilst soft tissue components such as muscle, tendon, and adipose tissue are more likely to affect markers placed away from segment endpoints. Soft tissue components can also be categorised as passive (skin, adipose tissue, connective tissue) or active (muscle, tendon), with their distribution along a segment highly irregular and segment specific. The various types of soft tissue all contribute independently to STA during movement (Leardini et al., 2005). Their contributions will also vary depending on the movement task and movement velocity. For example, running will likely cause different trunk STA when compared with a low-velocity trunk range of motion task (Kudo et al., 2017). Unsurprisingly, STA is also highly individualistic, due to the appreciable body composition variations in the population. Crucially, it is currently not possible to completely remove STA via digital filtering techniques. Soft tissue and bone possess similar frequencies during movement, making it difficult to differentiate between them (Fuller et al., 1997; Leardini et al., 2005). The variation in STA within and between individuals is also a major problem when attempting to mitigate the effects of STA on kinematic data by filtering. The consequences of STA are somewhat similar to those associated with anatomical landmark misidentification but the nature and size of the errors will arguably be more variable during movement tasks. Reflective-marker movement due to soft tissue deformation has the potential to dynamically alter segment and joint vectors during movement (Leardini et al., 2005). This means that kinematic data can often contain movements that are not functionally realistic, such as segment protractions or joint dislocations. It has been concluded that STA is the greatest single cause of measurement error in kinematic marker data (Fuller et al., 1997; Leardini et al., 2005).

Efforts to completely remove STA and landmark identification errors from kinematic data have so far been unsuccessful (Leardini et al., 2005). However, advanced modelling techniques have been developed in an effort to mitigate these errors. Two of these innovations will now be discussed.

2.3.1.1 Calibrated Anatomical Systems Technique

Minimising the number of markers placed on participants can be beneficial, as it increases participant comfort, decreases participant preparation duration and also reduces the computational requirements on retro-reflective based systems. The simplest method of modelling
joint centres/axes involves placing markers directly on landmarks at segment endpoints (e.g., lateral and medial femoral or humeral epicondyles). Joint centres are then approximated by taking the mid-point of the two markers. Unfortunately, markers positioned over joints are known to be highly susceptible to skin deformation and sliding (Cappello et al., 1997; Cappozzo et al., 1995). A marker may be placed correctly on an anatomical landmark at a given joint position but this does not mean the marker will remain on this landmark during dynamic motion. This is especially likely for movements that involve large joint ranges of motion, such as those often seen in dynamic sporting activities.

The calibrated anatomical systems technique (CAST) was developed in an effort to address the skin movement artefact problem for marker-based motion capture (Cappozzo et al., 1995). Biomechanical models that employ CAST incorporate clusters of three or more non-linear markers within participant marker-sets. Marker clusters are often placed on rigid or semi-rigid baseplates, which are then mounted on segments away from locations that are highly susceptible to heavy skin deformation and sliding (i.e., away from joints) (Cappozzo et al., 1996). The markers in a cluster are used to create technical reference frames (coordinate systems) that are considered bone-embedded. Virtual marker positions can then be stored and recreated within this technical frame, providing the marker cluster is placed on the same segment as the virtual marker’s targeted position. Marker positions will often be defined in a static calibration trial, with the markers then virtually recreated in dynamic trials. The CAST has several advantages, the most prominent of which is transferring marker positions from areas affected by severe skin deformation and sliding. Shifting markers away from joint axes also has the potential to improve participant comfort and reduce the risk of marker occlusion. For example, a cluster of markers on the lateral thigh are less likely to be occluded by the opposite lower limb than markers on the medial femoral epicondyle.

Models employing CAST for both upper (Boser et al., 2018; Campbell et al., 2009a, 2009b; Chin et al., 2009; Eftaxiopoulou et al., 2013; Schmidt et al., 1999) and lower body modelling (Besier et al., 2003; Cappozzo et al., 1997; Collins et al., 2009; Dempsey et al., 2007; Kainz et al., 2017) are commonplace in biomechanics laboratories around the world. The CAST-based modelling techniques reported by Campbell and colleagues (2009a, 2009b) for estimation of the glenohumeral joint centre are particularly relevant to this thesis. Their recommended method involves utilising regression equations and multiple technical clusters. Single markers positioned on the anterior and posterior shoulder are also utilised in the regression equation. When compared with magnetic resonance imaging (MRI), their method was shown to produce small joint centre location errors (13±1.5 mm) and good intra-tester (6±4 mm) and inter-tester reliability (6±3 mm) (Campbell et al., 2009b). The estimated glenohumeral centre is stored in two technical reference frames created from marker clusters placed on the acromion and lateral aspect of the upper arm.
The mean stored position from both technical frames is used as the glenohumeral joint centre in dynamic trials (Campbell et al., 2009a). Figure 2.2 displays the placement of markers recommended by Campbell and colleagues (2009a, 2009b).

Though CAST is an effective tool for attenuating skin deformation and sliding based STA, it is in no way a complete solution to STA. The obvious flaw in the CAST method is that STA is will be present at all locations of a segment, regardless of how close to joint axes markers are placed. Segmental deformation due to muscles, tendons, and adipose tissue is more likely to occur away from joint axes (Leardini et al., 2005; Stagni et al., 2005). Essentially, CAST prioritises skin sliding and deformation over other sources of STA. For this reason, many biomechanics laboratories continue to employ traditional marker sets and models which do not incorporate CAST and instead rely on placing markers directly on anatomical landmarks. However, it should be noted that more recent innovations involving multi-body kinematics optimisation have shown promise in mitigating the effects of STA on kinematic measurements (Bonnet et al., 2017; Camomilla et al., 2017; Duprey et al., 2017; Leardini et al., 2017) and may provide a pathway forwards.

Figure 2.2: Placement of markers utilised in the modelling of the glenohumeral joint centre, as per Campbell and colleagues (2009a, 2009b).

2.3.1.2 Functionally Defined Joint Centres and Joint Axes

Though CAST appears capable of mitigating STA errors associated with skin movement over joints, it does not address the other major source of error in 3D retro-reflective based modelling: anatomical landmark misidentification. As already highlighted, misplacement of markers on anatomical landmarks can reduce the measurement validity of a biomechanical model by altering segment and joint vector orientations to positions that are not reflective of the body’s true
positioning. The placement of markers on landmarks is also known to vary considerably within and between individuals (Della Croce et al., 2005), leading to poor measurement reliability. Numerical methods can be used to determine joint centres and axes, negating the need to precisely locate anatomical landmarks (Besier et al., 2003; Chen et al., 2013; Chin et al., 2009; Kainz et al., 2015, 2017; Leardini et al., 1999; Robinson and Vanreentghem, 2012; Van Campen et al., 2011). Once computed, the location of joint centres and orientations of axes of rotation are stored within a technical coordinate system and recreated within dynamic trials. Joint centres are able to be mathematically estimated for ball-and-socket joints like the hip (Besier et al., 2003; Kainz et al., 2015, 2017; Leardini et al., 1999), with axes also definable for hinge joints, such as the knee (Besier et al., 2003; Chen et al., 2013; Robinson and Vanreentghem, 2012; Van Campen et al., 2011).

Functional methods are commonly utilised to define joint centres and axes. Optimization equations are often used to define hip joint centres with mean helical axes calculated for hinge joints, such as the knee (Besier et al., 2003). As an example, to calculate a joint helical axis, markers are placed on two articulating segments (e.g., thigh and shank). The joint is then repeatedly moved, either actively or passively, through a range of motion whilst the movement of the markers recorded. During data processing the instantaneous helical axes can be calculated which represent the rotation and translation motion between the two body segments. The mean or median helical axis can then be computed and adopted as the joint axis (Besier et al., 2003; Woltring, 1994). Helical axis approaches have been applied primarily to the knee (Besier et al., 2003; Chen et al., 2013; Robinson and Vanreentghem, 2012; Van Campen et al., 2011) and elbow joints (Chin et al., 2009).

Similar to CAST, functional axes and joint centres are examples of advanced modelling techniques that have been developed to address STA and anatomical landmark misidentification respectively. Despite these innovations, it must be remembered that 3D retro-reflective based motion analysis is still affected by measurement errors and has limited ecological validity.

2.4 Magneto-inertial Measurement Units

In 1965, Gordon Moore made a famous prediction that the number of transistors placed on an integrated circuit would double roughly every two years, facilitating a two-fold increase in computing speed (Moore, 1965). This prophecy also means that the size of an integrated circuit will be able to halve approximately every two years, whilst maintaining computing speed. This helps to explain the boom in the wearable technology industry in recent years. Small wearable devices are now able to capture a wide range of data types with constantly improving levels of fidelity. A MIMU is a type of wearable technology that estimates 3D orientation. This makes MIMUs a particularly attractive proposition for biomechanists, as they allow 3D motion capture to take place outside of the traditional constraints of the laboratory, improving ecological validity in the process. Unlike
vision-based systems, MIMUs are not hampered by visual occlusion issues, which are particularly problematic when capturing dynamic sporting motions using passive retro-reflective markers. The confined capture volumes associated with 3D retro-reflective and 2D video based motion capture technologies are also not as relevant for MIMU systems, which often allow data to be captured onboard a unit (Iosa et al., 2016). Finally, MIMUs are also much more affordable, portable and user-friendly than 3D retro-reflective motion capture systems. It is apparent that MIMUs may provide a practical, field-based motion analysis solution but the measurement validity and accuracy of MIMU systems still requires further examination.

A MIMU combines tri-axial accelerometers, gyroscopes and magnetometers to estimate 3D orientation during all movement conditions. Accelerometers provide information on the sum of gravitational and inertial linear accelerations experienced by the sensor. The static orientation of a MIMU can be solely measured by an accelerometer. Accelerometers sense gravitational acceleration, therefore their signals are proportional to the deviation of its axis from the vertical direction (Picerno, 2017). By employing simple trigonometry, the static orientation of a MIMU with respect to gravity can be determined solely from three orthogonally-arranged accelerometers. This orientation representation is often referred to as ‘attitude’ or ‘inclination’, which is a combination of roll and pitch angles (Picerno, 2017). Gyroscopes are utilised in MIMUs to provide information on the MIMU’s orientation during dynamic conditions. This is achieved by numerical integration of the angular velocity measured by the gyroscopes. Unfortunately, the integration process is impacted by a form of measurement error known as drift, which increases over time and consequently makes solely-gyroscope based measurements of orientation unreliable. Corrections for drift can be undertaken by resetting the orientation of the MIMU at known movement conditions. A common practice is to use the accelerometers’ readings to reset angular displacement when the MIMU is static (Iosa et al., 2016; Picerno, 2017). This requires the signals from gyroscopes and accelerometers to be combined, which is known as ‘sensor fusion’. Sensor fusion is discussed further in section 2.4.1. As already alluded to, accelerometers are capable of correcting a MIMU’s orientation (as computed via numerical integration of angular velocity) but only with respect to gravity (i.e., only for roll and pitch angles). Magnetometers are incorporated into MIMUs to provide information about the vertical orientation (heading or yaw angle). They are capable of sensing the local magnetic north, which is used as a reference to reset drift about the vertical axis. Hence, utilising the three sensor types means it is possible for the complete orientation of a MIMU in 3D space to be obtained (Iosa et al., 2016; Picerno, 2017).

For practicality, gravity and magnetic fields are assumed constant within a measurement volume. This is true for gravity, although the magnetic field vector can be affected by the presence of ferromagnetic objects. This is especially problematic for indoor motion capture, as many objects or
appliances will provide ferromagnetic interference. This is one explanation for why larger errors are often observed for MIMU heading measurements than MIMU attitude measurements (Bergamini et al., 2014).

The specifications of sensor components can vary considerably between commercial MIMU systems. The capacities of accelerometers, gyroscopes and magnetometers, as well as the system sampling rate, are usually dependant on the MIMU system’s cost and intended usage (e.g., sport-specific analysis, clinical gait assessments). Analysis of high-velocity movements requires increased accelerometer and gyroscope specifications in order to avoid sensor saturation and subsequent data ‘clipping’ (Nam et al., 2014; Wells et al., 2018; Zhang et al., 2015). Aliasing of kinematic data during high-speed movement is also a potential issue if a MIMU’s sampling rate is insufficient to capture the movement velocity.

2.4.1 Sensor Fusion

Sensor fusion refers to the combining of data from the three types of sensors imbedded in MIMUs. As described above, this allows for a more robust calculation of the MIMU’s orientation. Two main approaches to sensor fusion have been proposed in the literature: stochastic filtering and complementary filtering (Bergamini et al., 2014). Stochastic filtering often involves Kalman filter (Kalman, 1960) based algorithms:

“These algorithms use knowledge of the expected dynamics of a system to predict future system states given both the current state and a set of control inputs” (pp. 7828, Lopez-Nava and Munoz-Melendez, 2016)

Complementary filtering is also commonly employed. This method fuses multiple noisy measurements that have complementary spectral characteristics. Complementary filters utilise only the part of the signal frequency spectrum that contains useful data, meaning the characteristics of the noise present in the process are not required to be included in the algorithm (Bergamini et al., 2014; Mahony et al., 2008). It is important to note that the precise structure of Kalman and complementary filter algorithms can vary between MIMU systems, with the specific algorithms often undisclosed by MIMU manufacturers (i.e., held as proprietary knowledge). There are also some general differences between the two approaches to sensor fusion that should be highlighted. The two approaches will vary in their treatment of gyroscope drift, magnetic disturbances and inertial acceleration. The algorithms can also describe orientation differently (Euler angles, rotation matrix, unit quaternions) and employ magnetometer data differently in the estimation of heading and attitude angles (Bergamini et al., 2014). The parameters utilised in Kalman filters can also vary depending on the MIMU system (Sabatini, 2011).
A study that compared complementary and Kalman filtering approaches concluded that there may not be significant discrepancies in measurement validity for locomotion and manual tasks when the two filtering methods are applied to the same MIMU system (Bergamini et al., 2014). The MIMU was placed on the lower back for the walking task and on the forearm for the manual task. The study clearly showed that both fusion approaches (Kalman and complementary) facilitate superior accuracy when compared with simple integration of gyroscope data, for the reasons outlined in section 2.4. In slight contrast to those findings, another study found that utilising different Kalman filter algorithms can lead to different MIMU orientation measures when both algorithms are applied to the sample MIMU system (Ligorio et al., 2016). The authors reported small but statistically significant measurement differences when a MIMU was positioned on the lower back during a clinical task (‘timed-up-and-go’ test).

2.4.2 Magneto-inertial Measurement Unit Based Modelling

Affixing MIMUs to body segments can allow estimations of segment and joint kinematics to be calculated. Similar to retro-reflective based modelling, the rigid segment assumption is employed for MIMU based modelling. A vital step when utilising MIMUs for functionally meaningful measurement of human movement is the aligning of MIMUs to body segments, commonly referred to as sensor-to-segment alignment (Picerno, 2017). This process ensures that the movement of a segment-mounted MIMU will represent the body segment kinematics in a functionally meaningful way. Many commercial MIMU systems allow simple calculation of sensor-to-sensor 3D angles, often expressed as Euler angles. However, for sensor-to-sensor angles to be functionally meaningful and truly representative of joint kinematics, both of the MIMUs must be correctly aligned to their respective segments and also expressed with respect to the same absolute reference frame (Picerno, 2017). Once MIMUs are aligned to a segment, joint kinematics can be estimated by multiplying the transposed rotation matrices of the two MIMUs, which allows a joint orientation matrix to be obtained. Joint kinematics are then calculated, usually by decomposing the joint orientation matrix into three consecutive rotations following a specified rotation order, as is usually done in retro-reflective based modelling (Picerno, 2017). Though Euler angle decomposition is frequently used to define MIMU orientations, quaternions are also regularly utilised for this purpose (Lopez-Nava and Munoz-Melendez, 2016). They are often seen as a more robust solution to MIMU orientation estimation as they are not affected by gimbal lock or order of rotation concerns (Brückner et al., 2014).

Sensor-to-segment alignment is usually achieved by either anatomical or functional calibration processes (Picerno, 2017). Anatomical calibration involves positioning the MIMU so that its coordinate system axes are aligned with the anatomical axes of the body segment. Often this is done imprecisely by simply visually aligning the MIMU with the orientation of the segment’s
skeletal structure. For instance, MIMUs are generally positioned on the spine so that the vertical axis of the MIMU’s coordinate system is parallel to the orientation of the spinal column. For segments of the limbs (e.g., the shank), MIMUs are typically positioned so that one of the MIMU’s axes is parallel with the long axis of the bone. More precise anatomical calibration techniques have been proposed (Picerno et al., 2008), although these approaches involve utilising additional purpose-built calibration instruments (pointers) which can be considerably more time-consuming than the simple visual estimation. Anatomical calibration also relies on accurate identification of anatomical landmarks, which, as discussed in section 2.3.1, is a major source of kinematic measurement error.

Functional calibration processes involve the estimation of individual-specific axes of rotation. The calibration processes are often similar to those used in retro-reflective based modelling (discussed in section 2.3.1.2). The major advantage of functional calibration is that it negates the need to precisely identify anatomical landmark locations. Many MIMU functional calibration methods have been described for the lower (Cutti et al., 2010; Favre et al., 2009; O’Donovan et al., 2007) and upper limbs (Cutti et al., 2008; de Vries et al., 2010; Galinski and Dehez, 2012; Ligorio et al., 2017). High angle measurement repeatability has been documented for the ankle (O’Donovan et al., 2007), knee (Cutti et al., 2010; Favre et al., 2009) and elbow (Ligorio et al., 2017) when MIMU functional calibrations are used. From a concurrent validity perspective, RMSDs of 1-8° have been reported when functionally calibrated MIMU-based angles were compared with angles calculated from 3D retro-reflective motion capture (Favre et al., 2009; Ligorio et al., 2017; O’Donovan et al., 2007). When compared with anatomically calibrated MIMU joint angles, functional calibration improved concurrent validity for knee (Favre et al., 2009) and elbow (Ligorio et al., 2017) joint angle measurements.

De Vries and colleagues (2010) also utilised MIMU functional calibration methods to define the thorax segment, which is of most relevance to this thesis. The aim of their study was to employ functional methods to define MIMU based anatomical coordinate systems (reference frames) for the thorax and upper limb. They tested which axes of rotation provided the lowest intra- and inter-tester variation. In doing so, they defined primary and secondary functional movements for the thorax (as well as the upper arm, forearm and hand) based on the lowest dispersion (highest repeatability) of rotation axis orientations. The researchers also compared the orientation of the MIMU based segment coordinate system with a segment coordinate system defined by retro-reflective markers placed on specific anatomical landmarks. They reported angle differences of 6.4±4.7° between the orientations of the retro-reflective based thorax coordinate system and the MIMU based thorax coordinate system. Though these findings are useful, the impact of different coordinate system orientations on angle outputs during complex movements was not reported.
The aforementioned studies assessed the impact of functional MIMU calibration on kinematics during low velocity, clinical movements (Cutti et al., 2008; Galinski and Dehez, 2012; Ligorio et al., 2017; O’Donovan et al., 2007) or walking gait (Cutti et al., 2010; Favre et al., 2009), with de Vries and colleagues (2010) investigating functional calibrations in relation to segment coordinate system orientations. The impact of MIMU functional calibrations on joint and segment angles has not been assessed for high-speed, sport-specific tasks, such as cricket bowling. Functional MIMU calibration has also not been explored for the lumbar segment nor fully validated for the thorax segment.

2.4.3 Measuring Trunk and Pelvis Kinematics with Magneto-inertial Measurement Units

Magneto-inertial measurement units have been utilised extensively in sporting, clinical and occupational settings to quantify, identify or categorise aspects of human movement. Their use has become more widespread within the last decade as wearable technology has improved and sensor sizes have decreased. Camomilla and colleagues (2018) recently provided a comprehensive systematic review of MIMU use in sports. Figure 2.3 is from their systematic review and clearly displays the growing popularity of MIMU based sport research.

![Figure 2.3: Distribution over time of magneto-inertial measurement unit papers focused on sport. Adapted from “Trends Supporting the In-Field Use of Wearable Inertial Sensors for Sport Performance Evaluation: A Systematic Review” by V. Camomilla, E. Bergamini, S. Fantozzi and G. Vannozzi, 2018, Sensors, 18, pp 8.](image)

Camomilla and colleagues (2018) stratified the MIMU sport studies into four categories: (1) technique analysis, (2) intensity measures for match analysis, (3) motor capacity assessment, and (4) activity classification. The current literature review is almost solely concerned with technique analysis and is particularly focused on quantification of segment and joint angles involving the thorax, lumbar and pelvis segments.

Whilst many MIMU studies have reported spinal or trunk kinematics, the majority of these studies analysed range of motion (Ha et al., 2013; Hajibozorgi and Arjmand, 2016; Lee et al., 2003; Mjøsund et al., 2017; Najafi et al., 2015; Narimani and Arjmand, 2018; Tafazzol et al., 2014;
Theobald et al., 2012), posture (Gleadhill et al., 2016; Wong and Wong, 2008), or low-velocity clinical assessments involving the trunk (Bauer et al., 2015; Charry et al., 2011; Faber et al., 2009; Fleron et al., 2018; Godwin et al., 2009; Grimpampi et al., 2013; Lee and Park, 2011; Mazzà et al., 2012; Plamondon et al., 2007; Schall Jr. et al., 2015; Walgaard et al., 2016). A smaller sample of studies have measured trunk kinematics in sport-specific contexts, including sprinting/running (Bergamini et al., 2013; Strohrmann et al., 2012), cycling (Zhang et al., 2015, 2013), golf (Watanabe & Hokari 2006), swimming (Bächlin and Tröster, 2012), discus (Brice et al., 2018; Ganter et al., 2010), snowboarding (Zihajehzadeh et al., 2015), trampolining (Helten et al., 2011) and cricket bowling (Senington et al., 2018).

Many of the studies listed above did not report a sport-specific MIMU measurement validity analysis by comparing with the current motion capture gold standard: 3D retro-reflective motion capture (Bächlin and Tröster, 2012; Ganter et al., 2010; Helten et al., 2011; Senington et al., 2018; Strohrmann et al., 2012; Watanabe and Hokari, 2006; Zihajehzadeh et al., 2015). Measurement validity has been determined for many commercial and customised MIMU systems in the literature (Camomilla et al., 2018; Lopez-Nava and Munoz-Melendez, 2016), however the varying results in these studies suggest that measurement validity may be impacted by many factors including, but not limited to: the type of MIMU system; movement condition or task; body region; sensor-to-segment calibration process; sensor fusion method; angle calculation method; testing environment; and the criterion measure (Bergamini et al., 2014; Filippeschi et al., 2017; Lebel et al., 2015, 2013; Ligorio et al., 2016; Pasciuto et al., 2015; Picerno et al., 2011).

Four studies have documented MIMU measurement validity for trunk kinematics during sport-specific tasks (Bergamini et al., 2013; Brice et al., 2018; Zhang et al., 2013, 2015). Bergamini and colleagues (2013) measured trunk inclination during sprint starts. The researchers positioned MIMUs on the second lumbar vertebra and aligned the orientation of the MIMU with the orientation of the trunk local coordinate system. They also placed markers on the MIMU, which facilitated the creation of a technical coordinate system aligned to MIMU’s coordinate system. Placing markers on the MIMU ensured that soft tissue artefact errors would affect the MIMU and the technical coordinate system similarly. The authors reported RMSD values of 3±2° when separately comparing the MIMU based inclination measurements with measures derived from the technical coordinate system and the trunk segment coordinate system. Zhang and colleagues (2013, 2015) measured athlete trunk position whilst mounted on a bicycle. They placed a MIMU on the sixth thoracic vertebra and compared its measurements to those derived from a retro-reflective defined trunk segment coordinate system. The authors reported RMSD values between 1.6±1.1° and 5.8±1.4° depending on the plane of motion and the sensor fusion algorithm used. Brice and colleagues (2018) provided measurement validity data for discus throwing, which arguably involves
more dynamic and multi-planar trunk and pelvis movements than sprinting or cycling. The authors chose not to align the MIMUs with technical coordinate systems or segment coordinate systems, instead comparing directly to retro-reflective anatomical model outputs. They documented MIMU measurement validity for torso and pelvis orientation, as well as shoulder-pelvis separation angle. The RMSD values for shoulder-pelvis separation angles (9.5° to the left, 12.5° to the right) were significantly poorer than for segment orientations (pelvis = 6.3°, torso = 9.3°). It has been noted that relative MIMU angles will often show larger measurement errors than global MIMU orientation measurements, due to the need of two MIMUs to define the same global reference coordinate system (Lebel et al., 2013; Picerno et al., 2011). Brice and colleagues (2018) also suggested that modelling differences and discrepancies in local coordinate system orientations between the MIMU and retro-reflective systems contributed to the higher RMSD values reported. However, they could not precisely quantify the contribution of these factors to the measurement differences.

The advantages of employing MIMUs to capture sporting motions are apparent. Unlike vision-based systems, such as 2D video and 3D retro-reflective motion capture, MIMUs are not impacted by occlusion issues, which can make recording of dynamic, complex movements difficult. They are also highly portable and are less restricted by the confined capture volumes that limit 3D retro-reflective motion capture. These factors make MIMUs seemingly ideal for field-based 3D motion capture. However, the MIMU literature suggests that MIMU measurement validity is highly dependent on multiple variables, including, but not limited to: the MIMU model; the sensor fusion algorithm; the MIMU calibration method; the criterion measure; the movement task; and the body segments/joints of interest. This topic has not been exhaustively investigated, especially in regards to high-speed, multi-planar sporting motions.

### 2.5 Cricket Fast Bowling: Biomechanics and Lumbar Injury

The role of the bowler in a cricket team is to bowl the ball down the pitch towards the opposition batter, with the aims of dismissing the batter and minimising opposition scoring. To achieve these aims a fast bowler will attempt to produce high ball release velocities (> 130 km/h), whilst maintaining bowling accuracy and control of the cricket ball’s unique aerodynamic characteristics. To assist with producing high ball release speeds, fast bowlers perform dynamic, multi-planar movements of the trunk during the bowling delivery stride. These movements are concomitant with substantial lumbar spinal loading (Bayne et al., 2016; Crewe et al., 2013b; Ferdinands et al., 2009). Consequently, there are high rates of lumbar injury amongst fast bowlers, particularly for young age groups (Crewe et al., 2012; Elliott et al., 1992; Johnson et al., 2012; Orchard et al., 2002, 2006, 2016; Ranson et al., 2005; Stretch, 2014). Lumbar spondylolysis is a serious bone stress injury
to the pars interarticularis that can often require extended recovery periods. The prevalence of lumbar spondylolysis has been estimated between 11 and 67% amongst fast bowlers (Annear et al., 1992; Crewe et al., 2012; Elliott et al., 1992; Foster et al., 1989; Gregory et al., 2004; Hardcastle et al., 1992). In comparison, studies of the general population have documented a spondylolysis incidence of 4-6% (Beutler et al., 2003; Fredrickson et al., 1984), with prevalence in the elite sporting population approximately 8% (Soler and Caldero, 2000).

Risk of severe lumbar intervertebral disc degeneration also appears to be heightened among fast bowlers (Burnett et al., 1996; Crewe et al., 2012; Ranson et al., 2005; Siemionow et al., 2011; Teraguchi et al., 2014). Studies have suggested that a third of professional fast bowlers have severe disc degeneration (Ranson et al., 2005), whilst presence of degeneration in young bowlers has been shown to increase by 37% in under three years of continued bowling (Burnett et al., 1996). Of further concern are the considerable recovery periods associated with these injuries, which can range from weeks to years dependent on severity. Epidemiological studies have listed lumbar bone stress injuries as the most prevalent injury suffered by Australian elite cricketers (1.9% of players unavailable at a given point in time) over a 10 year period, despite the injury being only the seventh highest in terms of incidence (3.2 new injuries per 100 players per year), indicating that the time spent not playing the sport was substantial (Orchard et al., 2016). It is therefore unsurprising that a large proportion of cricket research has focused on understanding the precise cause of fast bowler lumbar injuries and constructing informed injury prevention strategies. There is consensus in the literature that fast bowler lumbar injury aetiology is a complex interaction of three general categories of risk factor (Forrest et al., 2017; Johnson et al., 2012; Olivier et al., 2016b). Intrinsic risk factors predispose the athlete to injury, whilst bowling biomechanics and bowling workload (a measure of overuse and intensity) impose the loads that will cause an injury to occur. This review will focus solely on bowling biomechanics but this does not diminish the importance of the other two factors to lumbar injury aetiology.

2.5.1 Biomechanical Risk Factors to Lumbar Injury

Early research on fast bowling biomechanics by Elliott and Foster (1984) focused on developing a classification for different bowling styles. They described two main classifications: front-on action and side-on action. The front-on action is characterised by the bowler’s back foot of the delivery stride pointing down the pitch with the pelvis and shoulders orientated across the pitch. In a side-on action the back foot is positioned across the pitch with the pelvis and shoulders side-on to the batter. Subsequent studies also described a third type of classification, termed the mixed action, which comprises elements of both the side-on and front-on actions (Elliott et al., 1992; Foster et al., 1989; Portus et al., 2000). Multiple studies have suggested that the mixed bowling action is the most common bowling style adopted by fast bowlers (Ferdinands et al., 2010; Hardcastle et al.,
1992; Ranson et al., 2008, 2009). It is characterised by a lack of axial alignment between the shoulders and pelvis at back foot contact, which can be measured in the transverse plane and is often termed the shoulder-pelvis separation angle. A large separation angle (> 30°) has been associated with lumbar soft tissue injury (Portus et al., 2004).

Increased shoulder counter-rotation is also a feature of the mixed bowling action. Shoulder counter-rotation refers to the transverse plane movement of the shoulders and upper thorax away from the bowling direction between the points of back foot contact and front foot contact of the delivery stride (Figure 2.4) (Elliott et al., 1992; Portus et al., 2000). Foster and colleagues (1989) were the first group to suggest that increased shoulder counter-rotation range was linked with lower back injury. Many subsequent studies have also associated shoulder counter-rotation and/or mixed bowling actions with lumbar injury (Annear et al., 1992; Burnett et al., 1996; Elliott and Khangure, 2002; Elliott et al., 1992, 1993; Hardcastle et al., 1992; Portus et al., 2004) or increased lumbar kinetic loading (Crewe et al., 2013b). A shoulder counter-rotation range of greater than 30° has been suggested as a high risk threshold for fast bowlers (Portus et al., 2004). Worthy of note, however, are the multiple studies that have reported no link between lumbar injury and shoulder counter-rotation (Bayne et al., 2016; Dennis et al., 2008; Stuelcken et al., 2010). Further, despite the correlational evidence from previous research, the exact role of shoulder counter-rotation in lumbar injury causation remains an ongoing debate (Glazier, 2010; Ranson et al., 2008b). Elliott (2000) emphasised that shoulder counter-rotation was a measure of global trunk alignment not a measure of lumbar torsional stress. Shoulder counter-rotation occurs prior to front foot contact, whereas bowlers experience the highest lumbar loads between front foot contact and ball release (Crewe et al., 2013b; Ferdinands et al., 2009), well after shoulder counter-rotation ceases. Crewe and colleagues (2011) did report a strong correlation between shoulder counter-rotation and lumbar rotation range \( (r = 0.628, p < 0.01) \) between front foot contact and ball release. However, Senington and colleagues (2018) suggested the relationship between shoulder counter-rotation and lumbar kinematics was weaker, although they only assessed the period between back foot contact and front foot contact of the delivery stride and utilised MIMUs as their motion capture modality. It seems likely that large shoulder counter-rotation ranges and poor alignment of the shoulders and pelvis at back foot contact may cause fast bowlers to adopt positions or perform movements later in the bowling delivery stride that are potentially harmful to the lumbar spine.
Figure 2.4: A depiction of counter-rotation during the bowling action. The bowler’s shoulders counter-rotate between (A) and (B) before rotating towards the batter between (B) and (C). Adapted from “Technique factors related to ball release speed and trunk injuries in high performance cricket fast bowlers” by M. Portus, B. Mason, B. Elliott, M. Pfitzner and R. Done, 2004, *Sports Biomechanics*, 3(2), pp 265.

Some studies have reported that individuals who bowl with less flexion at the front leg knee joint have an increased risk of lower back injury (Elliott et al., 1992; Foster et al., 1989; Portus et al., 2004) or increased lumbar loading (Crewe et al., 2013b). However, other works have documented no relationship between the variables (Bayne et al., 2016; Dennis et al., 2008; Elliott et al., 1993; Olivier et al., 2016a). Interestingly, bowlers who deliver the ball with a more extended front knee also appear to release the ball at higher velocities (Crewe et al., 2013b; Loram et al., 2005; Middleton et al., 2016; Phillips et al., 2010; Worthington et al., 2013). This is likely attributed to increased peak braking forces, and decreased time to peak braking and vertical ground reaction forces (Portus et al., 2004). Perhaps unsurprisingly, Crewe and colleagues (Crewe et al., 2013b) also linked ball release speed to increased lumbar loading, indicating that fast bowlers may be more susceptible to lumbar injury.

Reduced hip flexion angle of the front leg in the delivery stride has been linked to injury by three studies (Bayne et al., 2016; Foster et al., 1989; Portus et al., 2004). A front leg hip angle of less than 45° at front foot contact may be hazardous for fast bowlers (Bayne et al., 2016). Similar to bowling with a more extended front knee, greater hip extension may mean that there is less attenuation of compressive forces following front foot contact (Bayne et al., 2016). A greater ball release height when normalised to standing height was linked to lumbar bony abnormalities in two of the early fast bowling studies (Elliott et al., 1992; Foster et al., 1989), which is logical, given the apparent relationship between injury and front limb extension. However, other studies have failed to find a link between injury and ball release height (Bayne et al., 2016; Dennis et al., 2008; Elliott et al., 1993).
Findings from more recent research have established thorax lateral flexion angle as one of the key biomechanical risk factors for fast bowler lumbar injury. Bayne and colleagues (2016) found that adolescent fast bowlers who sustained a lumbar injury during a season displayed 5° more thorax lateral flexion at front foot contact and 10° more at ball release than their non-injured counterparts. Thorax lateral flexion has also been associated with the mixed bowling action in professional fast bowlers (Ranson et al., 2008), lower back pain in elite female fast bowlers (Stuelcken et al., 2010) and ball release speed in elite bowlers (Phillips et al., 2010). Bayne et al. (2016) also discovered that bowlers who suffered a lumbar injury exhibited 10° more pelvis rotation by the point of ball release. It has been hypothesized by multiple authors that the combination of excessive pelvis rotation and thorax lateral flexion could be a key biomechanical factor in fast bowler lumbar injury aetiology (Bayne et al., 2016; Elliott, 2000; Glazier, 2010; Ranson et al., 2008).

All of the biomechanical risk factors previously discussed involve body segment motion separate to the lumbar region; the actual site of injury. The difficulty of recording spinal and lumbar motion during the bowling action is a major contributing factor to the dearth of fast bowling research that has measured lumbar kinematics or kinetics. Nonetheless, a small selection of studies have attempted to do so. In the first lumbar kinematics bowling study, Burnett and colleagues (1998) utilised an electromagnetic motion capture system and documented medium-large effect sizes that suggested that the mixed bowling action may be associated with increased lumbar extension at front foot contact and increased lumbar lateral flexion to the non-bowling arm side at ball release. Ranson and colleagues (2008) later reported medium effects sizes suggesting that bowlers with mixed actions had greater degrees of lower-trunk (lower thoracic and lumbar) lateral flexion and rotation at back foot impact of the bowling stride. However, a recent prospective study did not find any differences in lumbar kinematics between adolescent bowlers who suffered a lumbar injury during a season and those that did not (Bayne et al., 2016). The authors did show that injured bowlers exhibited significantly higher normalised peak lumbar flexion-extension moments, normalised peak lumbar lateral flexion moments, and normalised peak lumbar lateral flexion power than non-injured bowlers. These findings supported the hypothesis that lumbar lateral flexion is a key risk factor for lower back injury, which was postulated by Ferdinand and colleagues (2009) in the initial investigation of lumbar kinetics during bowling. A recent study also suggested that the upper (1st to 3rd lumbar vertebrae) and lower (4th and 5th lumbar vertebrae) portions of the lumbar segment contribute to bowling movements differently, with the upper lumbar segment producing greater ranges of extension, rotation and lateral flexion ipsilateral to the bowling arm and the lower lumbar segment responsible for greater flexion and contralateral lateral flexion ranges (Alway et al., 2018). The majority of spondylolysis cases affect the 4th or 5th lumbar vertebrae, with fast bowlers most often presenting with spondylolysis on the side contralateral to the bowling arm.
It appears findings from Alway and colleagues (2018) are congruent with the literature, providing more support for the hypothesis that excessive lateral flexion during bowling is a key risk factor for bony lumbar injury.

2.5.2 Cricket bowling biomechanical analysis

Cricket bowling biomechanical research has unsurprisingly followed similar motion analysis trends to the rest of sport biomechanics over the past few decades. Table 2.1 lists all the bowling biomechanical studies that have been cited in this review thus far. The motion capture modalities that were employed in those studies are also indicated. Whilst this chapter is not intended to be an exhaustive review of the literature, Table 2.1 is included to illustrate the progression of motion capture technology over the last three decades and the application of the technologies to cricket fast bowling research. The evolution of motion capture methods may help to explain some contradictions in the literature. For example, mixed bowling actions or shoulder counter-rotation were linked to lumbar injury by many studies that utilised 2D film or video analysis (Burnett et al., 1996; Elliott and Khangure, 2002; Elliott et al., 1992, 1993; Portus et al., 2004), whilst this link has been far more tenuous in studies that have utilised 3D retro-reflective motion capture. Conversely, thorax lateral flexion has only been linked to lumbar injury since the introduction of retro-reflective motion capture to bowling research (Bayne et al., 2016; Phillips et al., 2010; Ranson et al., 2008; Stuelcken et al., 2010). This trend may be purely coincidental, although it is concerning that there is a paucity of published research which has examined the concurrent measurement validity of 2D and 3D motion capture approaches to bowling biomechanical analysis.

The recent boom in wearable technology has witnessed MIMUs become more frequently used by cricket researchers, although the majority of these studies have focused on upper limb kinematics (Spratford et al., 2015; Wells et al., 2018; Wixted et al., 2011a, 2011b) or bowling event identification (McNamara et al., 2015; Rowlands et al., 2009). Four recent MIMU based studies could be considered relevant to lumbar injury risk. Callaghan and colleagues (2018) attempted to replicate bowling ground reaction force profiles from MIMU data, although their findings suggested this is not easily achievable using linear approaches. McNamara and colleagues (2018) utilised accelerometer and gyroscope data to quantify bowling frequency and intensity with moderate success. McGrath and colleagues (2018) employed MIMUs and machine-learning to successfully quantify bowling frequency; a key component of bowling workload. Senington and colleagues (2018) used MIMUs to examine shoulder counter-rotation and shoulder-pelvis separation and their link to lumbar kinematics. Unfortunately, the authors did not provide a comprehensive sport-specific investigation of MIMU measurement validity for the variables they assessed. It is unknown whether MIMU based measurements of shoulder counter-rotation, shoulder-pelvis separation and
lumbar kinematics are reflective of 2D video-based or 3D retro-reflective based measures of the same variables.

Table 2.1: Summary of fast bowling biomechanical studies previously cited in this literature review. Studies are categorised by the motion capture technology used to analyse fast bowling biomechanics, with capture frequencies (Hz) mentioned in brackets. sag. = sagittal camera, trans. = transverse camera

<table>
<thead>
<tr>
<th>High-Speed Film</th>
<th>Analog Video</th>
<th>Digital Video</th>
<th>3D Retro-Reflective Motion Capture</th>
<th>Other</th>
</tr>
</thead>
<tbody>
<tr>
<td>Foster et al. 1989 (200 Hz sagittal. camera, 100 Hz transverse camera)</td>
<td>Portus et al. 2000 (50 Hz)</td>
<td>Loram et al. 2005 (250 Hz)</td>
<td>Ranson et al. 2008a (120 Hz)</td>
<td>Burnett et al. 1998 (Electromagnetic system, 120 Hz)</td>
</tr>
<tr>
<td>Elliott et al. 1992 (200 Hz sag., 100 Hz trans.)</td>
<td>Elliott &amp; Khangure 2002 (60 z)</td>
<td>Dennis et al. 2008 (25 Hz)</td>
<td>Ranson et al. 2008b (120 Hz)</td>
<td>Senington et al. 2018 (MIMUs, 100 Hz)</td>
</tr>
<tr>
<td>Elliott et al. 1993 (200 Hz sag., 100 Hz trans.)</td>
<td>Portus et al. 2004 (50 Hz)</td>
<td>Olivier et al. 2016a (85 Hz)</td>
<td>Ranson et al. 2009 (300 Hz)</td>
<td></td>
</tr>
<tr>
<td>Burnett et al. 1996 (200 Hz sag., 100 Hz trans.)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Portus et al. 2004 (100 Hz)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

2.6 Summary

Three-dimensional retro-reflective motion analysis is currently considered the gold-standard of motion capture. The technology provides good measurement accuracy (van der Kruk and Reijne, 2018; Windolf et al., 2008), whilst retro-reflective based biomechanical modelling is also anatomically and functionally relevant. Unfortunately, retro-reflective systems are expensive and are generally unsuitable for outdoor motion capture, which makes them less than ideal for
analysing sporting motions in ecologically valid environments. Two-dimensional video digitising and MIMU-based modelling are motion analysis alternatives that address some of the shortcomings of retro-reflective systems. However, the measurement validity of both approaches has not been comprehensively explored, particularly for complex, high-speed sporting tasks and movements involving the trunk and pelvis.

An example of sport-specific, dynamic movement is cricket fast bowling. Multiple bowling kinematic variables are associated with increased risk of lumbar injury in cricket fast bowlers. Many of these factors concern angles of the thorax, lumbar and pelvis segments. The research examining fast bowling lumbar injury has primarily adopted retro-reflective motion capture to record bowling biomechanics. Utilising 2D video or MIMUs to measure segment and joint angles during bowling would enable coaches and researchers to more easily analyse bowling biomechanics in the field. However, it is generally unknown how the measurements for these alternative technologies compare with 3D retro-reflective derived measures.
2.7 References


Chapter 2: Review of the Literature


Chapter 2: Review of the Literature


This chapter provides additional information on data collection and processing procedures for all studies included in this thesis. The chapter concludes with a description of one-dimensional statistical parametric mapping (SPM1D); a data analysis technique employed in chapters 5, 6 and 7.

3.1 Data Collection and Processing for Chapter 4

Nineteen male fast-medium bowlers (16.6±3.3 years, 182.5±9.5 cm, 72.2±12.9 kg) currently playing district or community level cricket were recruited for the study presented in chapter 4. Informed consent was obtained from all participants or a parent/guardian for participants under the age of 18. Ethical approval was granted by the University of Western Australia Human Research Ethics Committee (RA/4/1/2593).

3.1.1 Motion Capture Systems and Laboratory Preparation

Three-dimensional (3D) retro-reflective trajectory data and multiple-plane, two-dimensional (2D) video data were collected concurrently in the indoor Sports Biomechanics Laboratory at the University of Western Australia.

A 22-camera, 250 Hz Vicon retro-reflective 3D motion capture system (Oxford Metrics, Oxford, UK), comprised of MX and T-series cameras, acted as the criterion measurement modality. Vicon cameras were calibrated with a Vicon Active Wand (Oxford Metrics, Oxford, UK) (Figure 3.1) according to the calibration procedures of Vicon motion systems. Successful camera calibration was indicated by image error residuals from each camera being below 0.30 pixels. An AMTI (Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA) force plate (2000 Hz) was also used to capture ground contact of the bowlers’ front foot. Vicon and force plate data were recorded with Vicon Nexus 1.8 software (Oxford Metrics, Oxford, UK). A 3D global coordinate system was also created for the laboratory. This involved capturing a static frame of the Vicon Active Wand (Oxford Metrics, Oxford, UK) positioned on the corner of the force plate.

Video footage was captured in the transverse and sagittal (cricket bowling arm side) planes by two 250Hz Vicon Bonita 2D video cameras (Vicon, Oxford Metrics, Oxford, UK) that were synchronised to the 3D system. Two Sony Handycam HDR-CX700 50 Hz video cameras (Sony Corporation, Tokyo, Japan) were positioned coronally behind the bowler’s run up and in the sagittal plane on the non-bowling arm side. To reduce image distortion, all camera shutter-speeds were set to the maximum possible speed (1/600th-1/1000th second) given the ambient light conditions.
In an effort to assimilate the laboratory to a cricket pitch, a bowling crease was marked on the force plate, with wickets placed at standard positions for a cricket pitch. Participants were also afforded enough space for a full bowling approach/run-up.

3.1.2 Participant Preparation

Approximately 50 reflective markers (14 mm) were attached to the participants’ skin using double-sided tape, prior to data collection. Of relevance to chapter 4 were markers affixed to the trunk, pelvis, and lower limbs (Bayne et al. 2016; Campbell et al. 2009b; Crewe et al. 2013). Placement of all relevant markers are described in Table 3.1, with Figure 3.2 displaying the marker positions. Single retro-reflective markers were placed on anatomical landmarks, with technical marker clusters mounted bilaterally on semi-rigid plates placed on the thigh and shank.
Table 3.1: Descriptions of reflective marker placement for chapter 4.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Marker Name</th>
<th>Marker Location</th>
</tr>
</thead>
<tbody>
<tr>
<td>Trunk</td>
<td>CLAV</td>
<td>Single marker positioned on the incisura jugularis (suprasternal or jugular notch).</td>
</tr>
<tr>
<td></td>
<td>STRN</td>
<td>Single marker positioned on the xiphoid process of the sternum.</td>
</tr>
<tr>
<td></td>
<td>C7</td>
<td>Single marker positioned on the spinous process of the seventh cervical vertebra.</td>
</tr>
<tr>
<td></td>
<td>T10</td>
<td>Single marker positioned on the spinous process of the tenth thoracic vertebra.</td>
</tr>
<tr>
<td></td>
<td>L5</td>
<td>Single marker positioned on the spinous process of the fifth lumbar vertebra. This marker was only used for video digitisation.</td>
</tr>
<tr>
<td>Pelvis</td>
<td>RASI, LASI, RPSI, LPSI</td>
<td>Single markers placed bilaterally on the posterior and anterior superior iliac spines.</td>
</tr>
<tr>
<td>Thigh</td>
<td>RTH1, RTH2, RTH3, LTH1, LTH2, LTH3</td>
<td>T-shaped marker clusters were positioned bilaterally on the lateral aspect of the mid-thigh with the long bar of the semi-rigid baseplate parallel to the femur and the short bar extending anteriorly/ventrally. Markers were used to create thigh technical coordinate systems.</td>
</tr>
<tr>
<td>Shank</td>
<td>RTB1, RTB2, RTB3, LTB1, LTB2, LTB3</td>
<td>T-shaped marker clusters were positioned bilaterally with the outer border of the long bar of the semi-rigid baseplate overlaying the medial border of the lower tibia. The short bar of the baseplate extended laterally and wrapped around the anterior border of the tibia. Markers were used to create shank technical coordinate systems.</td>
</tr>
<tr>
<td></td>
<td>RLMAL, RMMAL, LLMAL, LMMAL</td>
<td>Single markers placed bilaterally on the lateral malleolus of the fibula and the medial malleolus of the tibia.</td>
</tr>
</tbody>
</table>

Figure 3.2: Marker positions for chapter 4, shown in red. Markers on C7, T10, L5 and PSIS are not shown.

3.1.3 Data Collection Protocol

The data collection protocol commenced with a static anatomical A-pose calibration trial to allow static marker positions to be recorded. A 6-marker pointer calibration tool (Figure 3.3) was used to identify medial and lateral femoral epicondyles of both lower limbs in a series of static calibration
trials. Dynamic functional range of motion (ROM) trials were also collected for the knee and hip joints and were later used to determine bilateral knee axes of rotation and hip joint centres. The calibration trials were then followed by a self-directed warm-up period before participants were instructed to bowl 12 deliveries at match-level intensity.

**Figure 3.3:** The six-marker pointer calibration tool being used to point at the lateral epicondyle of the femur (chapter 4).

### 3.1.4 Data Processing

Three ‘full length’ deliveries were randomly selected for processing from each participant’s bowling trials. From these trials, a series of discrete variables were measured via 3D retro-reflective modelling and 2D video based digitising. All variables measured had been previously associated with lumbar injury in cricket fast bowling (further discussed in chapter 4).

#### 3.1.4.1 Retro-Reflective Data Processing

Retro-reflective trajectory data processing was conducted with Vicon Nexus 1.8 software (Oxford Metrics, Oxford, UK) and involved (1) marker labelling, (2) gap filling of trajectory data, (3) digital filtering, (4) biomechanical modelling, and (5) identification of key events.

Markers were labelled according to a customised Vicon skeleton template (.vst file), with marker gaps filled via the Vicon Nexus 1.8 (Oxford Metrics, Oxford, UK) in-built gap filling functions (spline fill or pattern fill).

Marker trajectories and ground reaction force data for the ROM and bowling trials were filtered using a fourth-order, zero-lag, low pass Butterworth filter (14 Hz cut-off). Cut-off frequencies were selected following a residual analysis of the data (Winter, 1990).

Retro-reflective data were modelled using a customised Vicon Bodybuilder model. The calibrated anatomical systems technique (CAST) (Cappozzo et al., 1996) was employed to store the static,
pointer-defined femoral epicondyle positions (Figure 3.3) relative to the thigh technical coordinate system. The epicondyle positions were then recreated in dynamic trials. Functional methods were used to define the bilateral hip joint centres and knee axes (Besier et al., 2003).

The descriptions of anatomical segment local coordinate systems are found in Table 3.2. The first axis listed for each segment was defined by a vector parallel to the first defining line. The second axis was perpendicular to the first and second defining lines. The third axis was perpendicular to the first and second axes. X-axes were anterio-posterior, y-axes were vertical (axial rotation), and z-axes were medio-lateral. Retro-reflective derived joint angles were computed by employing the Grood and Suntay (1983) joint coordinate system, with a ZXY-order of rotations.

**Table 3.2: Segment origins, defining lines and axis orders used to define body segments during retro-reflective data processing for chapter 4. R = right, L = left, med. = medial, lat. = lateral**

<table>
<thead>
<tr>
<th>Body Segment</th>
<th>Segment Origin</th>
<th>Defining Line 1</th>
<th>Defining Line 2</th>
<th>Axis Order</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>Mid-point of C7 and jugular notch</td>
<td>Mid-point of T10 and xiphoid process → Segment origin</td>
<td>C7 → Jugular notch</td>
<td>YZX</td>
</tr>
<tr>
<td>Pelvis</td>
<td>Mid-point of R ASIS and L ASIS</td>
<td>L ASIS → R ASIS</td>
<td>Segment origin → Mid-point of R PSIS and L PSIS</td>
<td>ZYX</td>
</tr>
<tr>
<td>Right Thigh</td>
<td>R knee joint centre</td>
<td>R knee joint centre → R hip joint centre</td>
<td>R lat. femoral epicondyle → R med. femoral epicondyle</td>
<td>YXZ</td>
</tr>
<tr>
<td>Left Thigh</td>
<td>L knee joint centre</td>
<td>L knee joint centre → L hip joint centre</td>
<td>L med. femoral epicondyle → L lat. femoral epicondyle</td>
<td>YXZ</td>
</tr>
<tr>
<td>Right Shank</td>
<td>R ankle joint centre</td>
<td>R ankle joint centre → R knee joint centre</td>
<td>R lat. malleolus → R med. malleolus</td>
<td>YXZ</td>
</tr>
<tr>
<td>Left Shank</td>
<td>L ankle joint centre</td>
<td>L ankle joint centre → L knee joint centre</td>
<td>L med. malleolus → L lat. malleolus</td>
<td>YXZ</td>
</tr>
</tbody>
</table>

Key event markers were placed using Vicon Nexus (Oxford Metrics, Oxford, UK) at the time-points of front foot contact and ball release during the bowling trials. Foot contact was determined from the force plate data, with a 20 N vertical force threshold used for this purpose. Ball release frame was determined from the sagittal Vicon Bonita Video Camera (Oxford Metrics, Oxford, UK) by identifying the first frame when the ball was no longer in contact with the hand. A customised MATLAB program (Mathworks, Natick, Massachusetts, USA) was used to output all relevant discrete 3D retro-reflective variables.

### 3.1.4.2 Video Data Processing

SiliconCoach Pro 7 (The Tarn Group, Dunedin, New Zealand) was employed to calculate 2D angles from the video footage. Descriptions and depictions of the 2D video-based measurement methods are found in Table 3.3. All 2D measurements were repeated three times, with the mean value used. Intra-tester (intra-class correlation = 0.936- 0.998) and inter-tester (intra-class correlation = 0.662-0.949) measurement reliability had been previously determined by a similar study (Weir et al., 2019).
Table 3.3: Methods used to calculate angles of interest from video data (chapter 4).

<table>
<thead>
<tr>
<th>Variable</th>
<th>Measurement Method</th>
<th>Convention</th>
<th>Image</th>
</tr>
</thead>
</table>
| Trunk lateral flexion angle at front foot contact of the bowling stride (°) | *Line 1:* Vertical vector from L5.  
*Line 2:* Vector from L5 to C7.  
*Measurement:* Global vertical angle between line 1 and line 2. | Away from bowling arm side = positive value (+)  
Towards bowling arm side = negative value (-) | ![Image](60) |
| Trunk lateral flexion angle at ball release (°)       | Same as trunk lateral flexion angle at front foot contact.                         | Same as trunk lateral flexion angle at front foot contact.                | ![Image](60) |
| Pelvis rotation angle at front foot contact of the bowling stride (°) | *Line 1:* Horizontal vector from left PSIS (for right arm bowler),  
*Line 2:* Vector from left PSIS to right PSIS (for right arm bowler),  
*Measurement:* Global horizontal angle between line 1 and line 2. | Rotation past horizontal vector = positive value (+)  
Rotation not past horizontal vector = negative value (-) | ![Image](60) |
### Front-leg hip flexion-extension angle at front foot contact of the bowling stride (°)

**Line 1:** Vector from estimated knee joint centre to estimated hip joint centre.

**Line 2:** Vector from estimated hip joint centre to iliac crest.

**Measurement:** Relative angle between line 1 and line 2. Computed value is subtracted from 180° (e.g., 180°-134° = 46° of hip flexion).

<table>
<thead>
<tr>
<th>Hip flexion</th>
<th>Hip extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>positive value (+)</td>
<td>negative value (-)</td>
</tr>
</tbody>
</table>

### Front-leg knee flexion-extension angle at front foot contact of the bowling stride (°)

**Line 1:** Vector from estimated ankle joint centre to estimated knee joint centre.

**Line 2:** Vector from estimated knee joint centre to estimated hip joint centre.

**Measurement:** Relative angle between line 1 and line 2. Computed value is subtracted from 180° (e.g., 180°-163° = 17° of knee flexion).

<table>
<thead>
<tr>
<th>Knee flexion</th>
<th>Knee hyper-extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>positive value (+)</td>
<td>negative value (-)</td>
</tr>
</tbody>
</table>

### Front-leg knee flexion-extension angle at ball release (°)

Same as knee flexion-extension angle at front foot contact.

| Same as knee flexion-extension angle at front foot contact. |

---

### 3.2 Data Collection and Processing for Chapters 5, 6 and 7

The data utilised in chapters 5, 6 and 7 was obtained from a single cohort. This section details the data collection and processing methodologies that were common for the three studies. Study specific procedures are addressed in the various chapters. Clarification is also given on the data subsets used in the three studies.

#### 3.2.1 Participants

Fast-medium bowlers were recruited from female state cricket squads, and from male district and community level cricket squads. The full participant cohort incorporated 4 female bowlers (172.5±5.5 cm, 66±4.7 kg, 19±3.7 years) and 16 male bowlers (182.5±5.6 cm, 80.7±11.3 kg, 24±7.9 years). Informed consent was granted by all participants or a parent/guardian if a participant was
under the age of 18. Ethical approval was provided by the University of Western Australia Human Research Ethics Committee (RA/4/1/2593) and the Curtin University Ethics Office (HRE2016-0472).

3.2.2 Motion Capture Systems and Laboratory Preparation

Data collection took place in the indoor Motion Analysis Laboratory at Curtin University, Perth, Western Australia. Three-dimensional retro-reflective data were captured by a 20-camera, 300 Hz Vicon motion capture system (Oxford Metrics, Oxford, UK), comprising a combination of MX and T-series cameras. Vicon cameras were calibrated using the same procedures described for chapter 4. A 300 Hz Vicon Bonita video camera (Oxford Metrics, Oxford, UK) was synchronised to the Vicon retro-reflective system. Two AMTI (Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA) force plates (1800 Hz) also concurrently collected force data within Vicon Nexus 2.5 software (Oxford Metrics, Oxford, UK).

Data were separately collected from three magneto-inertial measurement unit (MIMU) systems. Unfortunately, data from one of the systems (a prototype) was found to be unusable during data processing due to a manufacture issue. The two remaining MIMU systems were the Xsens Mtw Awinda system (Xsens Technologies B.V., Enschede, Netherlands) (75 Hz sampling frequency, ±2000°/s gyroscope, ±160 m/s² accelerometer, ±1.9 Gauss magnetometer) and the Noraxon MyoMOTION Research Pro system (Noraxon USA Inc., Scottsdale, Arizona, USA) (100 Hz sampling frequency, ±2000°/s gyroscope, ±16 g accelerometer, ±1.9 Gauss magnetometer). Unfortunately, it was not possible to match the sample rates of the two MIMU systems because the systems did not allow such flexibility. Xsens MIMU data were captured by MT Manager 4.2.1 software (Xsens Technologies B.V., Enschede, Netherlands). Noraxon MIMU data were captured by MyoRESEARCH 3.8.2 software (Noraxon USA Inc., Scottsdale, Arizona, USA).

Synthetic turf was laid in the laboratory to represent a cricket pitch. The laboratory included two large doors at either end, which were opened to allow participants adequate room to run-up and bowl a cricket ball. A bowling crease was marked on one of the force plates, with wickets also placed at standard positions for a cricket pitch.

3.2.3 Participant Preparation

Three MIMUs from each system were attached to the trunk and pelvis prior to commencement of data collection. A thorax MIMU was positioned so that its cranial edge was between the spinous processes of the seventh cervical (C7) and first thoracic (T1) vertebrae. A lumbar MIMU was affixed over the spinous process of the first lumbar vertebra (L1). The pelvis MIMU was placed with its caudal edge on the spinous process of the second sacral vertebra (S2).
Thirty-eight reflective-markers (10-14 mm) were affixed to participants’ upper arms, shoulders, trunk and pelvis (Bayne et al. 2016; Campbell et al. 2009b; Crewe et al. 2013). Marker and MIMU positions are displayed in Figure 3.4, with marker locations described in Table 3.4. Reflective markers were placed 1) on anatomical landmarks; 2) on body segments as technical marker clusters, mounted on semi-rigid plates; and 3) on MIMUs as technical marker clusters, mounted on rigid plates (Figure 3.5). Markers were affixed on the MIMUs to facilitate modelling of retro-reflective technical coordinate systems (RRtech) during data processing. The marker-plates were attached to the MIMUs using velcro, with the orientation of the markers aligned with the axes of the MIMU case as closely as possible. The coordinate system of each MIMU was also rotated during data processing so that it was more precisely aligned with the RRtech (described below).

3.2.4 Data Collection Protocol

Retro-reflective data functioned as a criterion measure and were collected simultaneously with MIMU data. The two MIMU systems were used separately because the MIMUs were required to be placed on specific anatomical landmarks. Participants completed the data collection protocol separately for each of the MIMU systems. On completion of the protocol for one MIMU model, the MIMUs and overlaid rigid-marker plates were switched whilst all other reflective markers remained on the body. Unfortunately, participant and MIMU system availability meant that not all participants were able to complete the protocol for both MIMU systems. A total of 10 participants performed the protocol for the Xsens system, with 14 participants completing the protocol for the Noraxon MIMUs.

The data collection protocol commenced with a static anatomical-pose calibration trial, so that static marker positions could be captured. Each individual then completed five moderately-paced trunk ROM trials. Standing ROM trials were collected for flexion-extension and lateral flexion, before sitting ROM trials were collected for all three planes of motion (flexion-extension, lateral flexion and rotation). For each movement plane, participants moved to the end of their range and repeated the movement five times in both directions. Participants then performed two clinical movement tasks (sit-to-stand and pen pick-up tests) and five side-steps (cutting manoeuvres) which are otherwise irrelevant to this thesis. Finally, participants were instructed to perform five cricket bowling trials at match-like intensity. Two bowling trials were later selected for further analysis. Only two trials were analysed because not all of the MIMU data for two participants captured correctly due to a limited capture range/volume. This was not discovered until during the data processing phase. For participants that had more than two trials, the chosen trials were those that displayed the least amount of reflective marker occlusion.
Figure 3.4: Retro-reflective marker set affixed to the shoulders, trunk and pelvis for chapters 5, 6 and 7. Location of MIMUs and overlaying marker technical clusters also indicated in blue.

Table 3.4: Descriptions of reflective marker placement for chapters 5, 6 and 7.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Marker Name</th>
<th>Marker Location</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulders/Upper Arms</td>
<td>RACR1, RACR2, RACR3</td>
<td>Marker clusters were positioned so that the lateral ridge of the semi-rigid baseplate overlaid the lateral ridge of the acromion processes.</td>
</tr>
<tr>
<td></td>
<td>RASH, RPSH, LASH, LPSH</td>
<td>Single markers were positioned on the anterior and posterior aspects of the shoulders, in a plane with the visually estimated glenohumeral joint centres.</td>
</tr>
<tr>
<td></td>
<td>RUA1, RUA2, RUA3</td>
<td>T-shaped marker clusters were positioned on the lateral aspect of the arms with the long bar of the semi-rigid baseplate parallel to the humerus and the short bar extending towards the lateral head of the biceps.</td>
</tr>
<tr>
<td></td>
<td>LUA1, LUA2, LUA3</td>
<td></td>
</tr>
<tr>
<td>Thorax</td>
<td>CLAV</td>
<td>Single marker positioned on the incisura jugularis (suprasternal or jugular notch).</td>
</tr>
<tr>
<td></td>
<td>STRN</td>
<td>Single marker positioned on the xiphoid process of the sternum.</td>
</tr>
<tr>
<td></td>
<td>T10</td>
<td>Single marker positioned on the spinous process of the tenth thoracic vertebra.</td>
</tr>
<tr>
<td></td>
<td>C7MIMU1, C7MIMU2, C7MIMU3</td>
<td>Markers arranged as shown in Figure 1.4 on a rigid marker plate. Placed over the thorax MIMU.</td>
</tr>
<tr>
<td>Lumbar</td>
<td>L3</td>
<td>Single marker positioned on the spinous process of the third lumbar vertebra.</td>
</tr>
<tr>
<td></td>
<td>L5</td>
<td>Single marker positioned on the spinous process of the fifth lumbar vertebra.</td>
</tr>
<tr>
<td></td>
<td>RLL, LLL</td>
<td>Single markers placed either side of the fourth lumbar vertebrae, approximately midway between the spinous process and the most lateral aspect of the trunk.</td>
</tr>
<tr>
<td></td>
<td>L1MIMU1, L1MIMU2, L1MIMU3</td>
<td>Markers arranged as shown in Figure 1.4 on a rigid marker plate. Placed over the lumbar MIMU.</td>
</tr>
<tr>
<td>Pelvis</td>
<td>RASI, LASI, RPSI, LPSI</td>
<td>Single markers placed bilaterally on the posterior and anterior superior iliac spines.</td>
</tr>
<tr>
<td></td>
<td>RILC, LILC</td>
<td>Single markers placed on the most superior points of the iliac crests.</td>
</tr>
<tr>
<td></td>
<td>S2MIMU1, S2MIMU2, S2MIMU3</td>
<td>Markers arranged as shown in Figure 1.4 on a rigid marker plate. Placed over the pelvis MIMU.</td>
</tr>
</tbody>
</table>
3.2.5 Data Processing

3.2.5.1 Retro-Reflective Data Processing

Retro-reflective trajectory data processing was conducted with Nexus 2.5 software (Oxford Metrics, Oxford, UK) and followed the same general pipeline as for chapter 4.

Markers were labelled according to a customised Vicon skeleton template (.vst file), with marker gaps filled via the Vicon Nexus 2.5 (Oxford Metrics, Oxford, UK) in-built gap filling functions (spline fill, rigid-body fill or pattern fill).

Marker trajectories and force plate data for the ROM and bowling trials were filtered using a fourth-order, zero-lag, low pass Butterworth filter. Cut-off frequencies of 9 Hz and 15 Hz were selected for ROM and bowling trials respectively, following a residual analysis of the data (Winter 1990).

Retro-reflective data were modelled using a custom Vicon Bodybuilder model (Oxford Metrics, Oxford, UK). The markers overlaying each of the MIMUs were modelled by using the mean position of the three markers as the origin of the RRtech. The first and second defining lines were from marker-2 to marker-1 and marker-2 and marker-3 respectively. The first defining line was parallel with the y-axis (vertical), with the x-axis the cross-product of the first and second defining lines, and the z-axis orthogonal to the y and x axes (Figure 3.5). The triad on the lumbar MIMU was used to create a marker representing the first lumbar vertebra (L1) by calculating the mid-point between two of the triad markers. A similar process was employed to estimate the position of the seventh cervical vertebra (C7) from the thorax MIMU triad.

![Image](image_url)

Figure 3.5: IMU and affixed rigid retro-reflective marker plate utilised in chapters 5, 6 and 7. Markers (M1, M2, M3), defining lines (D1, D2), and coordinate system axes (X, Y, Z) are indicated. X-axis is anterior-posterior. Y-axis is vertical. Z-axis is medio-lateral.

The method described by Campbell and colleagues (Campbell et al., 2009a, 2009b) was used to estimate bilateral shoulder joint centres, with CAST (Cappozzo et al., 1996) employed to store the position of the joint centre in two separate technical coordinate systems (upper arm and
The shoulder joint centre was recreated in dynamic trials but taking the mean stored position from both technical coordinate systems.

The descriptions of anatomical segment local coordinate systems are found in Table 3.5. The first axis listed for each segment was defined by a vector parallel to the first defining line. The second axis was perpendicular to the first and second defining lines. The third axis was perpendicular to the first and second axes. X-axes were antero-posterior, y-axes were vertical (axial rotation), and z-axes were medio-lateral.

Retro-reflective derived joint angles were computed by employing the Grood and Suntay (1983) joint coordinate system, with a ZXY-order of rotations. Chapter 5 also assessed the measurement validity of other different joint angle calculation methods but these will not be detailed here.

Global angles were calculated for each of the RRtechs and anatomical segment coordinate systems, relative to the global coordinate system. Relative angles were calculated between the RRtechs and between anatomical segments. All of the angle conventions utilised in chapters 5, 6 and 7 are described in Table 3.6.

Table 3.5: Segment origins, defining lines and axis orders used to define body segments during retro-reflective modelling for chapters 5, 6 and 7.

<table>
<thead>
<tr>
<th>Body Segment</th>
<th>Segment Origin</th>
<th>Defining Line 1</th>
<th>Defining Line 2</th>
<th>Axis Order</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>Mid-point of C7 and jugular notch</td>
<td>Mid-point of T10 and xiphoid process → Segment origin</td>
<td>C7 → Jugular notch</td>
<td>YZX</td>
</tr>
<tr>
<td>Upper Thorax/Shoulders</td>
<td>Mid-point of right and left shoulder joint centres</td>
<td>Left shoulder joint centre → Right shoulder joint centre</td>
<td>Jugular notch → Segment origin</td>
<td>ZYX</td>
</tr>
<tr>
<td>Lumbar</td>
<td>L5</td>
<td>L5 → L1</td>
<td>Right lower lumbar → Left lower lumbar</td>
<td>YXZ</td>
</tr>
<tr>
<td>Pelvis</td>
<td>Mid-point of Right and Left ASIS</td>
<td>Left ASIS → Right ASIS</td>
<td>Segment origin → Mid-point of Right and Left PSIS</td>
<td>ZYX</td>
</tr>
</tbody>
</table>

Using Vicon Nexus (Oxford Metrics, Oxford, UK), event markers were placed at the time-points of back foot contact, front foot contact and ball release during the bowling trials. Foot contact was determined from the force plate data, with a vertical force reading of greater than 20 N indicating a foot-strike. Ball release frame was determined from the sagittal Vicon Bonita Video Camera (Oxford Metrics, Oxford, UK) by identifying the first frame when the ball was no longer in contact with the hand. Event markers were also placed at the start and end of a ROM cycle. The start and end of the phases were defined from the thorax RRtech to pelvis RRtech angle data. Zero degrees (neutral trunk position) indicated the beginning and end of the phase.
Table 3.6: Anatomically derived retro-reflective angles and equivalent angles derived from retro-reflective technical coordinate systems (RRtech) overlaying MIMUs (chapters 5, 6 and 7).

<table>
<thead>
<tr>
<th>Anatomically Derived Retro-Reflective Angle</th>
<th>RRtech Derived Angle</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulders global angle</td>
<td>Thorax RRtech global angle</td>
</tr>
<tr>
<td>Shoulders-to-pelvis relative angle</td>
<td>Thorax RRtech to pelvis RRtech relative angle</td>
</tr>
<tr>
<td>Thorax global angle</td>
<td>Thorax RRtech global angle</td>
</tr>
<tr>
<td>Thorax-to-pelvis relative angle</td>
<td>Thorax RRtech to pelvis RRtech relative angle</td>
</tr>
<tr>
<td>Lumbar global angle</td>
<td>Lumbar RRtech global angle</td>
</tr>
<tr>
<td>Lumbar-to-pelvis relative angle</td>
<td>Lumbar RRtech to pelvis RRtech relative angle</td>
</tr>
<tr>
<td>Pelvis global angle</td>
<td>Pelvis RRtech global angle</td>
</tr>
</tbody>
</table>

3.2.5.2 Magneto-inertial Measurement Unit Data Processing

The Xsens and Noraxon MIMU free acceleration and orientation data were output for each trial as a .csv file using the Xsens MT Manager software (Xsens Technologies B.V., Enschede, Netherlands) and myoRESEARCH software (Noraxon USA Inc., Scottsdale, Arizona, USA) respectively. Sensor orientations were estimated by the manufacturers’ inbuilt filtering algorithms and were expressed in quaternions. Both MIMU systems utilised Kalman filter-based sensor fusion algorithms to estimate orientation angles, although the exact algorithms used by each system were not disclosed in the manufacturers’ documentation.

3.2.5.3 Additional Data Processing

Additional data processing was performed by a customised LabVIEW 2017 program (National Instruments Corp, Austin, Texas, USA) (Figure 3.6). The program (1) temporally synchronised MIMU and retro-reflective data; (2) spatially aligned the coordinate systems of each MIMU with the accompanying RRtech; (3) calculated joint angle values; and (4) compiled and output values for all trials. These steps are described below. For the study presented in chapter 5, the program also computed functional axes for the thorax and lumbar segments, and calculated angles using five different methods. Those procedures are described in detail in chapter 5.

The data from both MIMU systems were temporally synchronised with the retro-reflective data for each trial. Temporal synchronisation for the Xsens ROM trials was achieved by cross-correlating the angular component of the axis-angle representation of the thorax MIMU and its accompanying RRtech. The lumbar MIMU and RRtech were used instead for the sitting flexion-extension ROM trials. For the Noraxon MIMU ROM trials, cross-correlating the orientation quaternions for the thorax MIMU and RRtech was found to be more suitable. Both the Noraxon and Xsens bowling trials were temporally synchronised by cross-correlating the orientation quaternions. Cross-correlation was utilised because it enabled time-lag between the MIMU and retro-reflective data to be calculated and accounted for.
Figure 3.6: Interface for the customised LabVIEW 2017 program (National Instruments Corp, Austin, Texas, USA) utilised for chapters 5, 6 and 7.

To spatially align the Xsens and Noraxon MIMU coordinate systems with their paired RRtech, each MIMU coordinate system was rotated -90° about the z-axis and then 90° about the y-axis. The Noraxon coordinate systems were then also rotated 180° about the z axis, as they utilised a different coordinate system orientation to the Xsens sensors. A further refinement process utilised data from the seated trunk flexion-extension ROM trial. It involved calculating the orientation difference between the MIMU coordinate systems and RRtech orientations and then accounting for this difference. For each MIMU/RRtech pair, the orientation difference was calculated by multiplying the RRtech quaternion by the inverse of the MIMU quaternion. The difference in their orientations at each time-point was averaged over the course of the trial, with this correction value then applied to the MIMU data. Once these alignment procedures were completed, it was possible to calculate global angles by comparing the RRtech or the MIMU orientations with the retro-reflective 3D global reference frame.

Global and relative angles for both MIMU systems were calculated via a ZXY-ordered Euler angle decomposition. Lateral flexion (x) and rotation (y) were negated, so that the right hand rule applied. Additional angle calculation methods are presented in chapter 5.

The MIMU and retro-reflective continuous angle data were time-normalised for the ROM and bowling trials. Back foot contact and ball-release were used to define the bowling stride phase (Bartlett, 1996). A single trunk ROM cycle was used as a ROM phase. Following time normalisation of data, all angle values were collated and written to Microsoft Excel spreadsheets (Microsoft
3.2.6 Data subsets

Chapter 5 utilised data from the Xsens MIMU system and the concurrently collected RRtech data (n=10, 178.9±7.1 cm, 77±13.3 kg, 24.4±7.8 years). As previously stated, multiple different methods were employed to calculate joint angles for this study. Anatomically modelled retro-reflective data was not utilised in chapter 5.

Chapter 6 presented data from a total of 16 different participants. It also utilised the Xsens MIMU and RRtech data (n=10, 178.9±7.1 cm, 77±13.3 kg, 24.4±7.8 years). The measurement validity of the Xsens data was compared with a subset of the Noraxon data (n=10, 184.5±5.3 cm, 80±9.3 kg, 21.6±3.9 years) and the accompanying RRtech data. Participant and MIMU system availability meant the datasets for each system contained data from different people, with four participants appearing in both datasets. Anatomically modelled retro-reflective data were not analysed for chapter 6.

Chapter 7 utilised the Noraxon MIMU data subset (n=10, 184.5±5.3 cm, 80±9.3 kg, 21.6±3.9 years) but compared this data with the concurrently collected, anatomically modelled retro-reflective data. The RRtech data was not utilised for chapter 7.

3.3 One-Dimensional Statistical Parametric Mapping

One-dimensional statistical parametric mapping (SPM1D) was used to assess for significant differences (p < 0.05) between time-normalised MIMU and retro-reflective angle waveforms in chapters 5, 6 and 7. This data analysis technique, based on random field theory, has been successfully utilised to identify significant differences between sets of time-varying kinematic and force data (Donnelly et al., 2017; Pataky et al., 2013). Alternative approaches, such as only considering designated scalar values (e.g., maximum, minimum, etc.) or considering all data points simultaneously with time-independent statistics (e.g., correlation coefficients), attempt to analyse one-dimensional vectors by producing a zero-dimensional scalar statistic. The major strength of SPM1D is its ability to analyse complex waveforms in a one-dimensional manner. This enables periods of significant difference during a movement task to be dissociated from periods of similarity. However, SPM1D relies upon group means and standard deviations and can consequently be insensitive to small intra-trial differences between measures. Hence, discrete root mean square difference was also used in this thesis to quantify intra-trial differences between
measures. Worthy of note is that all continuous data presented in this thesis was visually inspected for outliers prior to being incorporated into the SPM1D analyses.

For this thesis, publicly available SPM1D source code (Pataky, 2018) was incorporated into customized MATLAB scripts (Mathworks, Natick, Massachusetts, USA). Each SPM1D analysis produces a critical t-statistic (threshold), the value of which is determined by the ‘temporal smoothness’ of the data and a predefined alpha value (i.e., $\alpha = 0.05$). A time-varying t-statistic ($t$-trace ($t$)) is then calculated at each time-point during the phase. The $t$-trace exceeding the critical t-threshold is indicative of a significant difference between the two sets of data. A supra-threshold period is accompanied by a $p$-value and an ‘extent’. An extent represents the duration of a supra-threshold period and is usually presented as a percentage of the movement phase (Pataky et al., 2013; Wells et al., 2017). Multiple supra-threshold periods (extents) can be present for each movement phase. These elements are displayed in Figure 3.7.

![Figure 3.7](image)

**Figure 3.7:** An example of one-dimensional statistical parametric mapping analysis presented in this thesis. The plot on the left displays the two angle waveforms of interest during the movement phase (mean curves are presented with standard deviation clouds). The plot on the right shows the time-varying t-statistic or t-trace ($t$), with significant supra-threshold periods (extents) clearly labelled when the t-trace exceeds the critical t-statistic threshold.
3.4 References


Chapter 4: Measuring Bowling Kinematics from Two-Dimensional Video

**Cottam, D. S.**., Bayne, H., Elliott, B. C., Donnelly, C. J., & Alderson, J. A. Does two-dimensional video data similarly reflect three-dimensional kinematics associated with lumbar injury risk in cricket fast bowlers?

*Prepared for submission*

Intended Submission Target: *Sports Biomechanics*

Intended Submission Date: April 2019.

The following abstract, resulting from the work in this chapter, was submitted as a peer-reviewed abstract and subsequently presented as an oral presentation at the 34th International Conference of Biomechanics in Sports in Tsukuba, Japan, 18-22 July, 2016. A copy of this abstract is available in 9.8: Appendix H.

4.1 Abstract

Many studies utilising three-dimensional (3D) motion capture have linked fast bowling biomechanics with lumbar injury. Unfortunately, most bowlers are unable to access 3D motion analysis services. This study aimed to assess whether multiple-plane, two-dimensional (2D) video analysis is a valid field-based alternative to 3D motion capture for measurement of bowling joint and segment angles associated with lumbar injury. It was hypothesised that bowling data captured using 2D video would be a valid method of assessing 3D kinematics associated with lumbar injury. Nineteen fast-medium bowlers were simultaneously recorded by 3D retro-reflective motion capture and multiple-plane 2D video. Intra-class correlation (ICC) and root mean square difference (RMSD) assessed differences between discrete 3D and 2D measures. All variables produced RMSDs between 4.3° and 9.4°. The ICC values varied from 0.38 to 0.97 (p<0.001). The 2D measures of pelvis rotation angle (ICC=0.91) and front-knee flexion-extension angle (ICC=0.97) at ball release were excellently correlated with 3D values. The validity of 2D measurements of lumbar injury risk factors depends on the variable being assessed, as front-leg knee (ICC=0.38) and hip (ICC=0.58) flexion-extension at front foot contact were poorly-moderately correlated with the 3D data. The information presented should be utilised to improve lumbar injury risk screening processes.
Chapter 4: Measuring Bowling Kinematics from Two-Dimensional Video

4.2 Introduction

Cricket fast bowling is a dynamic and multifaceted skill that places a large amount of stress on the body, particularly the lumbar region of the back (Burnett et al., 1998; Ferdinands et al., 2009; Foster et al., 1989). Prevalence of bone stress injuries to the lumbar pars interarticularis (termed spondylolysis) have been reported to be 11-67% of fast bowlers (Annear et al., 1992; Crewe et al., 2012; Elliott et al., 1992; Foster et al., 1989; Gregory et al., 2004; Hardcastle et al., 1992), which is much higher than the 4-6% of people affected by this injury in the general population (Beutler et al., 2003; Fredrickson et al., 1984) or the non-cricketer elite athlete population (Soler and Caldero, 2000). Intervertebral disc degeneration has also been shown to be accelerated in fast bowlers when compared with the general population (Burnett et al., 1996; Crewe et al., 2012; Ranson et al., 2005; Siemionow et al., 2011; Teraguchi et al., 2014). Young fast bowlers appear to be especially susceptible to such conditions. Engstrom and Walker (2007) showed that adolescent (13-17 year old) fast bowlers had a spondylolysis incidence of 24% over a four year period. Crewe and colleagues (2012) reported that a third of 15-19 year old asymptomatic fast bowlers had at least one pars interarticularis abnormality when scanned by magnetic resonance imaging, and a further 6 of 25 individuals went on to develop bone stress injuries in a single cricket season (Bayne et al., 2016). The rates of spondylolysis and disc degeneration, considerable rehabilitation periods, and long-term implications of these conditions has meant that a large proportion of cricket research has focused on improving understanding of their complex aetiologies and developing successful injury prevention strategies.

There is substantial evidence that bowling biomechanics play a role in lumbar injury onset (Forrest et al., 2017; Johnson et al., 2012; Olivier et al., 2016). The majority of these biomechanical factors relate to the bowling delivery stride, which is the period from back foot contact to ball release. Early fast bowling research identified large shoulder counter-rotation ranges and a transverse plane misalignment of the hips and shoulders at back foot contact of the delivery stride as injury risk variables of interest (Annear et al., 1992; Elliott et al., 1993, 1992; Foster et al., 1989; Hardcastle et al., 1992). Both of these variables occur prior to the bowler’s front foot making contact with the ground, whereas bowlers have been shown to experience the highest lumbar loads between front foot contact (FFC) and ball release (Crewe et al., 2013; Ferdinands et al., 2009). Consequently, the exact role of shoulder counter-rotation and hip-shoulder misalignment in lumbar injury causation has been debated (Glazier, 2010; Ranson et al., 2008). It seems likely that large shoulder counter-rotation ranges and poor axial alignment of the shoulders and hips at back foot contact may cause fast bowlers to adopt postures or perform movements following FFC that are potentially harmful to the lumbar spine. It is perhaps unsurprising then that kinematic variables occurring after FFC have
also been associated with lumbar injury. Individuals who bowl with less flexion at the front-leg knee joint during the delivery stride may have an increased risk of lower back injury (Elliott et al., 1992; Foster et al., 1989; Portus et al., 2004) or increased lumbar loading (Crewe et al., 2013). Similarly, reduced hip flexion of the front leg has also been linked to injury (Bayne et al., 2016; Foster et al., 1989; Portus et al., 2004). A greater ball release height when normalised to standing height was related to lumbar bony abnormalities by two of the early fast bowling studies (Elliott et al., 1992; Foster et al., 1989), which is logical given the apparent relationship between injury and front limb extension. More recent research has suggested that thorax lateral flexion angle may be a key biomechanical risk factor for fast bowler lumbar injury. Bayne and colleagues (2016) found that adolescent fast bowlers who sustained a lumbar injury during a season displayed approximately 5° more thorax lateral flexion at FFC and approximately 10° more lateral flexion at ball release. Thorax lateral flexion has also been associated with the mixed bowling action in professional fast bowlers (Ranson et al., 2008), lumbar pain in elite female fast bowlers (Stuelcken et al., 2010), and ball release speed in elite bowlers (Phillips et al., 2010). Bayne et al. (2016) discovered that bowlers who suffered a lumbar injury exhibited 10° more pelvis rotation by the point of ball release. It has been hypothesized by multiple authors that the combination of excessive pelvis rotation and thorax lateral flexion could be a key biomechanical factor in fast bowler lumbar injury aetiology (Bayne et al., 2016; Elliott, 2000; Glazier, 2010; Ranson et al., 2008).

Many of the injury risk kinematic variables mentioned have been identified by studies that used three-dimensional (3D) retro-reflective motion analysis to capture positional data prior to the calculation of bowling kinematics. Retro-reflective systems are capable of sub-millimetre accuracy (van der Kruk and Reijne, 2018; Windolf et al., 2008) but they are typically used in laboratories and are generally unsuitable for field-based data capture. They are also expensive and require specialised training to use effectively. For these reasons, 3D retro-reflective-based kinematic analyses are not readily available to most cricketers with the exception of those at the elite levels. This is concerning due to the prevalence of lumbar injury amongst adolescent bowlers. There has been a lack of research that has attempted to close the gap between the lab and the field by validating field-based motion capture technologies for bowling biomechanical analysis. Educating cricket coaches on the effective use of tools that are portable, cost-effective and user-friendly may allow lumbar injury risk screening of fast bowlers to become common at all levels of the sport.

Multiple plane, two-dimensional (2D) video analysis is an alternative solution to the lumbar injury risk screening conundrum. Video cameras are affordable and user-friendly, however the measurement approaches used in 2D video analysis are different to those employed for 3D motion capture. Retro-reflective based modelling involves the placement of many reflective markers on anatomical landmarks or segments, which allows 3D body segments and joints to be modelled with
anatomical and functional applicability. Joint angles can then be calculated by using 3D joint coordinate systems and Euler angle decompositions. Additional techniques such as functional joint calibrations (Besier et al., 2003) and the calibrated anatomical system technique (CAST) (Cappozzo et al., 1995) are often used in an effort to mitigate errors associated with misidentification of joint centres and soft tissue artefacts due to skin deformation over joints. In comparison with these approaches, 2D video based digitising is rudimentary. Markers are often placed directly over joints, meaning skin-based movement artefact can be a major source of error. Using video digitising software, simple 2D vectors are used to represent body segments, with angles between vectors measured to approximate a 3D joint angle. Placement of cameras is also paramount, with ‘planar crosstalk’ a significant obstacle for valid measurement of 2D kinematics from video (Elliott and Alderson, 2007).

Two-dimensional video analysis can be used successfully for field-based motion analysis if the user is cognisant of the method’s limitations. Measures such as shoulder counter-rotation were initially identified by cricket studies that used 2D video as the motion capture modality (Foster et al., 1989), whilst many cricket coaches already use video to provide qualitative feedback to players (Elliott and Khangure, 2002; Ranson et al., 2009; Wallis et al., 2002). Other cricket bowling research has suggested small reliability errors of approximately 3° can be expected when calculating elbow flexion-extension angles from high-speed 2D video (Portus et al., 2006). In different sporting contexts, moderate-high intra-class correlations (ICC) have been reported between 2D video and 3D retro-reflective derived measures for trunk and lower limb variables during side-stepping (Weir et al., 2019). These examples suggest that 2D video may be a viable field-based alternative to 3D motion capture for fast bowler lumbar injury risk screening. This research aimed to determine whether digitised 2D video can be used to validly measure fast bowling kinematics associated with lumbar injury risk, via assessment against a 3D retro-reflective approach. It was hypothesised that it would be possible to validly assess lumbar injury risk factors involving the trunk and lower limb using 2D multiple-plane video analysis.

4.3 Methods

4.3.1 Participants

Nineteen male fast-medium bowlers (16.6±3.3 years, 182.5±9.5 cm, 72.2±12.9 kg) currently playing district or community level cricket were recruited for the criterion-based validation study. Informed consent was obtained from all participants or a parent/guardian for participants under the age of 18. Ethical approval was granted by the University of Western Australia Human Research Ethics Committee (RA/4/1/2593).
4.3.2 Data Collection

Three-dimensional retro-reflective trajectory data and multiple-plane 2D video data were concurrently collected in the Sports Biomechanics Laboratory at the University of Western Australia. Trajectory data were captured by a 22-camera Vicon motion analysis system (Vicon, Oxford Metrics, Oxford, UK), consisting of 10 T-series and 12 MX-series cameras sampling at 250 Hz. Prior to data collection, a customized reflective marker set was affixed to the lower limbs and trunk of the participants (Bayne et al., 2016; Dempsey et al., 2007). The marker set consisted of single markers and three-marker technical clusters affixed to semi-rigid plates. Single markers were placed bilaterally on the medial and lateral malleoli of the ankle. Technical clusters were placed bilaterally on the antero-medial shank and the lateral aspect of the thigh, as per the calibrated anatomical systems technique (Cappozzo et al., 1995). Single markers were also placed on the right and left anterior superior iliac spines (ASIS), and the right and left posterior superior iliac spines (PSIS). For the trunk, single markers were placed on the seventh cervical vertebra (C7), the tenth thoracic vertebra (T10), xiphoid process and jugular notch of the sternum. A 6-marker pointer calibration tool was used to identify medial and lateral femoral epicondyles of both lower limbs in a series of static calibration trials. Dynamic functional range of motion trials were collected for the knees and hip joints and were later used to determine the knee axes of rotation and hip joint centres bilaterally (Besier et al., 2003).

Video footage was captured in the transverse and sagittal (bowling arm side) planes by two Vicon Bonita 2D video cameras operating at 250 Hz (Vicon, Oxford Metrics, Oxford, UK) that were synchronised to the 3D system. The sagittal camera was later used to identify the ball release frame. Two Sony Handycam HDR-CX700 50 Hz video cameras (Sony Corporation, Tokyo, Japan) were positioned coronally behind the bowler’s run up and in the sagittal plane on the non-bowling arm side. To reduce image distortion, all camera shutter-speeds were set to the maximum possible speed (1/600th-1/1000th second) given the ambient light conditions.

The laboratory was also implemented with a 1.2 m x 1.2 m AMTI force plate (Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA) which sampled at 2000 Hz and was synchronised to the 3D motion analysis system. The force plate was used to identify FFC of the delivery stride, with a vertical force measure greater than 20 N indicating a foot-strike. In an effort to assimilate the laboratory to a cricket pitch, a bowling crease was marked on the force plate, with wickets placed at standard positions for a cricket pitch. A collection of static and dynamic calibration trials was followed by a self-directed warm-up period before participants were instructed to bowl 12 deliveries at match-level intensity.
4.3.3 Data Processing

Three ‘full length’ deliveries were randomly selected for processing from each participant’s 12 bowling trials. From these trials, a series of discrete variables were measured using both 3D modelling and 2D video digitising. All the variables measured had been previously associated with lumbar injury. The variables measured were thorax lateral flexion angle at FFC and ball release (Bayne et al., 2016; Ranson et al., 2008; Stuelcken et al., 2010), pelvis rotation angle at ball release (Bayne et al., 2016), front-leg hip flexion angle at FFC (Bayne et al., 2016; Foster et al., 1989; Portus et al., 2004), and front-leg knee flexion angle at FFC and ball release (Crewe et al., 2013; Elliott et al., 1992; Foster et al., 1989; Portus et al., 2004). Ball release height (Elliott et al., 1992; Foster et al., 1989) was also measured from the 2D video.

Three-dimensional trajectory data were labelled, filtered and modelled in Vicon Nexus 1.8 software (Oxford Metrics, Oxford, UK). Marker trajectories were filtered using a fourth-order, zero-lag, low pass Butterworth filter, with a cut-off frequency set at 14 Hz, after a residual analysis was performed (Winter, 1990). Retro-reflective data were modelled by a customised Vicon Bodybuilder model (Oxford Metrics, Oxford, UK). For dynamic trials, pointer-defined femoral epicondyle positions were expressed in the thigh technical coordinate systems, as per the calibrated anatomical systems technique (Cappozzo et al., 1995). Functional methods were used to define the bilateral hip joint centres and knee axes (Besier et al., 2003). The body segment origins, defining lines and axis orders are presented in Table 4.1. The first listed axis for each body segment was created by a vector parallel to the first defining line. The second listed axis was the cross-product of the first and second defining lines. The third axis for each segment was the cross-product of the first and second axes. X-axes were antero-posterior, Y-axes were vertical (axial rotation), and Z-axes were medio-lateral. The joint coordinate system described by Grood and Suntay (1983) was used for joint angle calculation. A customised MATLAB program (Mathworks, Natick, Massachusetts, USA) output all relevant 3D retro-reflective based variables.
Table 4.1: Segment origins, defining lines and axis orders used to define body segments during retroreflective data processing. R = right, L = left, med. = medial, lat. = lateral

<table>
<thead>
<tr>
<th>Body Segment</th>
<th>Segment Origin</th>
<th>Defining Line 1</th>
<th>Defining Line 2</th>
<th>Axis Order</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>Mid-point of C7 and jugular notch</td>
<td>Mid-point of T10 and xiphoid process → Segment origin</td>
<td>C7 → Jugular notch</td>
<td>YZX</td>
</tr>
<tr>
<td>Pelvis</td>
<td>Mid-point of R ASIS and L ASIS</td>
<td>L ASIS → R ASIS</td>
<td>Segment origin → Mid-point of R PSIS and L PSIS</td>
<td>ZYX</td>
</tr>
<tr>
<td>Right Thigh</td>
<td>R knee joint centre</td>
<td>R knee joint centre → R hip joint centre</td>
<td>R lat. femoral epicondyle → R med. femoral epicondyle</td>
<td>YXZ</td>
</tr>
<tr>
<td>Left Thigh</td>
<td>L knee joint centre</td>
<td>L knee joint centre → L hip joint centre</td>
<td>L med. femoral epicondyle → L lat. femoral epicondyle</td>
<td>YXZ</td>
</tr>
<tr>
<td>Right Shank</td>
<td>R ankle joint centre</td>
<td>R ankle joint centre → R knee joint centre</td>
<td>R lat. malleolus → R med. malleolus</td>
<td>YXZ</td>
</tr>
<tr>
<td>Left Shank</td>
<td>L ankle joint centre</td>
<td>L ankle joint centre → L knee joint centre</td>
<td>L med. malleolus → L lat. malleolus</td>
<td>YXZ</td>
</tr>
</tbody>
</table>

SiliconCoach Pro 7 (The Tarn Group, Dunedin, New Zealand) was used to calculate 2D angles from the video footage. Depictions of the 2D variables are displayed in Figure 4.1. Thorax lateral flexion angle was measured from the coronal video (50 Hz) at FFC and ball release, using markers on C7 and L5. A measure of 0° was considered vertical/upright with a positive value indicating the bowler was leaning towards the non-bowling arm side. Pelvis rotation was calculated at ball release using the transverse video (250 Hz) and the PSIS markers. A 0° angle indicated the pelvis was exactly parallel to the bowling crease (i.e., a front-on position), with a positive value indicating rotation past a front-on position. Front-leg hip flexion-extension angle at FFC and front-leg knee flexion-extension angle at FFC and ball release were calculated from the non-bowling arm side sagittal video (50 Hz). The iliac crest, head of the femur, lateral condyle of the femur and the lateral malleoli were used for these angle measurements. Ball release height was calculated from the sagittal video on the bowling arm side (250 Hz). All 2D video-based measurements were repeated three times, with the mean value used and the standard deviation calculated. Intra-tester (intra-class correlation = 0.936-0.998) and inter-tester (intra-class correlation = 0.662-0.949) measurement reliability had been previously determined by a similar study (Weir et al., 2019).

Figure 4.1: (1) Thorax lateral flexion at FFC, (2) thorax lateral flexion at ball release, (3) pelvis rotation at ball release, (4) hip flexion at FFC, (5) knee flexion at FFC, and (6) knee flexion at ball release.
4.3.4 Data analysis

To compare the 2D and 3D kinematic values from the 57 trials assessed (19 participants x 3 trials), intra-class correlations (ICC) and 95% confidence intervals were computed by a single rater, absolute agreement, two-way random effects model (Koo and Li, 2016). Values of less than 0.5 were considered poor, 0.5-0.75 was moderate, 0.75-0.9 was good, and >0.9 was excellent (Koo and Li, 2016). An achieved power of 0.92 was calculated using an alpha level of 0.05 and an effect size of 0.6.

Means and standard deviations were calculated for the 2D and 3D derived angles for each variable. Mean absolute difference (MAD) and root mean square difference (RMSD) between the two methods were also computed for each variable.

A one-tailed, bivariate, Pearson correlation was used to assess for a relationship between 2D ball release height (normalised to participant standing height) and two other variables: 3D thorax lateral flexion angle at ball release and 3D knee flexion-extension angle at ball release. Correlation coefficient thresholds were categorized as small (0.1), moderate (0.3), large (0.5), very large (0.7) or extremely large (0.9) (Hopkins et al., 2009).

Version 20 of SPSS statistics (IBM, Armonk, New York, USA) was used to calculate correlations and descriptive statistics, while a customised MATLAB program (Mathworks, Natick, Massachusetts, USA) calculated MAD and RMSD.

4.4 Results

The ICC values and 95% confidence intervals (CI) are displayed in Table 4.2. Excellent ICC values (ICC > 0.9, p < 0.001) were found for the 2D and 3D measures of front-leg knee flexion-extension angle at ball release (ICC = 0.97, 95% CI = 0.94-0.98) and pelvis rotation angle at ball release (ICC = 0.91, 95% CI = 0.35-0.97), however the 95% confidence interval for pelvis rotation angle ranged from poor-excellent. A moderate-good ICC was calculated for thorax lateral flexion angle at ball release (ICC = 0.73, 95% CI = 0.57-0.83, p < 0.001). Poor-moderate intra-class correlations (ICC < 0.75, p < 0.001) were present for thorax lateral flexion angle at FFC (ICC = 0.65, 95% CI = 0.02-0.86), front-leg hip flexion angle at FFC (ICC = 0.58, 95% CI = 0.38-0.73), and front-leg knee flexion-extension angle at FFC (ICC = 0.38, 95% CI = 0.07-0.61).
Table 4.2: Intra-class correlations and 95% confidence intervals for 2D and 3D injury risk variables.

<table>
<thead>
<tr>
<th>Angle (°)</th>
<th>Intra-Class Correlation</th>
<th>95% Confidence Interval</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Lower Bound</td>
</tr>
<tr>
<td>Thorax lateral flexion at FFC</td>
<td>0.65*</td>
<td>0.02</td>
</tr>
<tr>
<td>Thorax lateral flexion at ball release</td>
<td>0.73*</td>
<td>0.57</td>
</tr>
<tr>
<td>Pelvis rotation at ball release</td>
<td>0.91*</td>
<td>0.35</td>
</tr>
<tr>
<td>Front-leg hip flexion at FFC</td>
<td>0.58*</td>
<td>0.38</td>
</tr>
<tr>
<td>Front-leg knee flexion-extension at FFC</td>
<td>0.38*</td>
<td>0.07</td>
</tr>
<tr>
<td>Front-leg knee flexion-extension at ball release</td>
<td>0.97*</td>
<td>0.94</td>
</tr>
</tbody>
</table>

*Significant at $p < 0.001$

Means and standard deviations for the 2D digitised and 3D modelled variables are displayed in Figure 4.2, along with the MAD and RMSD values for each variable. Thorax lateral flexion angle at FFC ($2D = 12.6 \pm 8.3^\circ$, $3D = 18.5 \pm 8.6^\circ$, MAD = $-5.9 \pm 5.3^\circ$, RMSD = 7.9$) was generally underestimated from the 2D video when compared with the 3D data. This was not the case for the thorax lateral flexion angle at ball release, with smaller measurement differences found ($2D = 46.2 \pm 7.5^\circ$, $3D = 47.8 \pm 7.8^\circ$, MAD = $-1.6 \pm 5.5^\circ$, RMSD = 5.7). Pelvis rotation angle at ball release ($2D = 11.8 \pm 14.5^\circ$, $3D = 7.2 \pm 13.3^\circ$, MAD = 4.6$) was often slightly overestimated from the 2D video.

Large measurement differences were seen for front-leg hip flexion angle at FFC ($2D = 43.6 \pm 8.2^\circ$, $3D = 43.3 \pm 9.9^\circ$, MAD = 0.4$) and front-leg knee flexion-extension angle at FFC ($2D = 17.9 \pm 6.5^\circ$, $3D = 12.6 \pm 8.4^\circ$, MAD = 5.3$). However, much smaller differences were produced for front-leg knee flexion-extension angle at ball release ($2D = 40.8 \pm 16.7^\circ$, $3D = 39.2 \pm 16.5^\circ$, MAD = 1.6$).

A large Pearson correlation of -0.67 ($p < 0.001$) was also found between normalised 2D ball release height and 3D thorax lateral flexion angle at ball release, however there was not a significant relationship between normalised 2D ball release height and front-leg knee flexion-extension angle at ball release ($r = 0.12$, $p = 0.18$).

As each 2D video-based measure was repeated three times per trial, it is also important to note the variability of those measurements. The within-trial angle measure standard deviations ranged between 1.43$^\circ$ (pelvis rotation angle at ball release) and 2.99$^\circ$ (thorax lateral flexion angle at FFC), whilst normalised ball release height (0.004) was measured particularly consistently.
4.5 Discussion

This study aimed to assess whether kinematic data measured from 2D video can be used to validly measure lumbar injury risk factors during cricket fast bowling. In general, the results suggest that 2D video may be a useful field-based alternative to 3D motion capture but should only be used when the user has an awareness of its limitations. The RMSD values ranged between 4.3° and 9.4° which suggests that measurement differences between 3D retro-reflective motion analysis and 2D video digitisation may be functionally significant (Bayne et al. 2016), despite the differences being comparable with studies which measured concurrent validity of alternative motion capture technologies during similar sporting tasks (Brice et al., 2018; Weir et al., 2019).

The wide range of ICC values (0.38-0.97) indicates that some of the variables assessed can be confidently measured using appropriately positioned 2D video cameras, whilst others do not satisfactorily reflect 3D measurements, irrespective of camera placement. The 2D measurement of pelvis rotation angle at ball release compared well with its 3D equivalent (ICC = 0.91, 95% CI = 0.35-0.97, MAD = 4.6±3.8°, RMSD = 5.9°). The large 95% confidence interval can likely be explained by large standard deviations for both 3D and 2D measures. In accepting the results for this variable, it must be acknowledged that pelvis rotation was measured from a transverse video camera, which sampled at 250 Hz, whilst the other variables assessed were calculated from cameras sampling at...
50 Hz (due to camera availability). The transverse camera was also synchronised to the 3D motion
capture system, which meant that FFC and ball release were temporally aligned between systems.
It is likely that synchronisation and increased sampling rate significantly improved the concurrent
validity between the 2D and 3D systems.

Fast bowling research in the last decade has identified thorax lateral flexion as a key indicator of
lumbar injury risk (Bayne et al., 2016; Ranson et al., 2008; Stuelcken et al., 2010). The results of this
study indicate that, if forced to rely on 2D video analysis, measuring thorax lateral flexion at ball
release (ICC = 0.73, MAD = -1.6±5.5°, RMSD = 5.7°) may be a more prudent practice than attempting
to measure the angle at FFC (ICC = 0.65, MAD = -5.9±5.3°, RMSD = 7.9°). We speculate that the
underestimation of thorax lateral flexion at FFC (MAD = -5.9±5.3°) when using the 2D video was
due to the bowlers’ trunks being in a semi-rotated position at this point of the delivery stride. This
meant the measurement was taken out of plane with the coronal camera. In comparison, bowlers
are in a relatively front-on position by ball release and their trunks are closer to a coronal plane.
Positioning an additional camera perpendicular to the orientation of a bowler’s trunk at FFC would
likely reduce the measurement error. Unfortunately, technique variability among fast bowlers
means determining the exact required position of the camera would be individual-specific and
tedious.

The 2D measurement of front-leg hip flexion-extension at FFC (ICC = 0.58, MAD = 0.4±8.3°, RMSD =
8.2°) may have been affected by misidentification of the hip joint location in the video. Participant
clothing makes it particularly difficult to locate joint positions, which has been noted in previous
studies (Portus et al., 2006). Front-leg knee flexion-extension 2D angle measurement at FFC was
also dissimilar to data recorded using 3D analysis (ICC = 0.38, MAD = 5.3±7.8°, RMSD = 9.4°),
although the same 2D angle measured at ball release was comparable; returning the smallest
differences to the 3D data for any of the variables assessed (ICC = 0.97, MAD = 1.6±4.0°, RMSD =
4.3°). The knee joint of the front-leg often moves with high angular velocities immediately following
FFC (King et al., 2016). Differences in sampling rates between the 3D system (250 Hz) and 2D video
(50 Hz) may have meant that the FFC frame was not well temporally matched between the
systems, leading to discrepancies in measurements of knee and hip angle at FFC. The fast bowlers’
front knee angle is relatively constant by the point of ball release, likely explaining the smaller
measurement differences at this time point.

As previously mentioned, it is likely that sampling rate discrepancies heavily influenced the results
for this study. Cricket fast bowling is a particularly dynamic skill, which involves high velocity
segmental motion. When possible, cricket coaches and biomechanists should utilise video cameras
with higher sampling rates for fast bowling biomechanical analysis. Using cameras with sampling
rates above 100 Hz is advisable. This will increase the likelihood that key events such as FFC and
ball release are captured accurately. If higher-speed cameras are unavailable, then capturing additional trials may allow coaches to select those trials which have frames nearest to key events.

In addition to the findings already discussed, a large correlation ($r = -0.67$, $p < 0.001$) between 2D ball release height (normalised to standing height) and 3D thorax lateral flexion angle at ball release suggests that release height may be a simple surrogate measure of thorax lateral flexion for coaches to measure when using 2D video analysis. The relationship between increased thorax lateral flexion and a lower ball release height is somewhat contrary to the previously postulated hypothesis that a greater ball release height is associated with more front-leg knee extension at ball release and a consequent increase to injury risk (Elliott et al., 1992; Foster et al., 1989). There was no significant relationship found between front-leg knee angle and ball release height in this study ($r = 0.12$, $p = 0.18$).

A limitation of this study was that only concurrent measurement validity was assessed, not intra-tester and inter-tester reliability. However, Weir and colleagues (2019) documented these factors for similar kinematic variables during side-stepping. As fast bowling could be considered a more dynamic task than side-stepping, it is recommended that future research specifically assesses the intra-tester and inter-tester reliability of the variables presented in this paper.

The overall findings of this research indicate that 2D video analysis can be used by coaches and biomechanists to measure kinematics associated with lumbar injury, however it should be done with a degree of caution.

The following summary points may be useful to coaches, players and other cricket researchers when attempting to use 2D video to analyse fast bowling biomechanics:

- Video cameras with higher frame rates (>100 Hz) are preferable for cricket fast bowling kinematic analysis.
- If high-speed cameras are unavailable then collecting additional trials and selecting those with frames nearest to key events (e.g., ball release) would be prudent.
- Quantifying movements that occur out of plane with a video camera should not be attempted.
- A 2D measurement of thorax lateral flexion may be more valid at ball release than at FFC.
- Measuring ball release height from a 2D camera is simple and may be reflective of thorax lateral flexion angle.
4.6 Conclusion

Two-dimensional video analysis may be used for field-based kinematic screening of fast bowlers for lumbar injury risk, however coaches and biomechanists should be aware that concurrent validity between data collected from 2D video and 3D retro-reflective motion capture varies greatly depending on the variable being assessed and respective camera sampling rates. Differences in system sampling rates, non-optimal video camera placement, and the obvious dissimilarities to modelling and angle measurement practices appear to be the main source of measurement differences between 3D retro-reflective motion analysis and 2D video digitisation of cricket bowling kinematics.
4.7 References


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Chapter 5: Functionally Calibrated MIMU-based Thorax and Lumbar Measurements

Cottam, D. S., Campbell, A. C., Davey, P. C., Kent, P., Elliott, B. C., & Alderson, J. A. Functional calibration does not improve the concurrent validity of MIMU-based thorax and lumbar angle measurements when compared with retro-reflective motion capture.

Prepared for submission

Intended Submission Target: Medical & Biological Engineering & Computing

Intended Submission Date: March 2019.
Research Linking Statement

The study presented in Chapter 4: found that some joint and segment angles associated with lumbar injury risk in cricket bowlers can be validly measured from two-dimensional (2D) video. However, 2D video does not appear suitable for comprehensive analyses of bowling biomechanics, especially when measurements of trunk kinematics are required. Simplistic digitising methods are one of the main limitations of 2D video analysis, with high-speed movements that cross multiple planes of motion particularly difficult to quantify from 2D video. Magneto-inertial measurement units (MIMUs) were identified as a portable alternative to 2D video and three-dimensional (3D) retro-reflective motion capture. Crucially, MIMUs are able to quantify 3D orientation. They are also not affected by marker/landmark occlusion issues and are capable of higher sampling rates than standard 2D video, making them seemingly suitable for field-based analysis of sporting tasks. Unfortunately, MIMU measurement validity remains a concern. There is a dearth of sport-specific MIMU measurement validity studies focusing on the trunk. A key determinant of MIMU measurement involves the calibration of MIMUs with body segments so that the MIMU measures are functionally meaningful. Anatomical and functional calibration methods have been documented in the literature. Anatomically aligning MIMUs with segments can be achieved by simple visual estimation, with more precise anatomical calibration made possible by using purpose-built calibration instruments to identify anatomical landmarks. Functional MIMU calibration approaches involve moving joints through ranges of motion in order to calculate axes of rotation. This technique appears to be practical and reliable whilst also capable of improving measurement validity. The functional approach has not been sufficiently explored for the thorax and lumbar segments. The impact of functional calibrations on the measurement validity of segment and joint angles during high-speed, complex sporting tasks has also not been reported. This study aimed to assess whether incorporating functional axes in angle calculation methods could significantly improve concurrent validity of thorax and lumbar angle measurements during slow, uni-planar movements and dynamic, sport-specific tasks.
5.1 Abstract

Commercial magneto-inertial measurement unit (MIMU) systems often allow calculation of simple sensor-to-sensor Euler angles, though this process does not address sensor-to-segment alignment which is important for functionally meaningful MIMU kinematics. Functional sensor-to-segment calibrations have improved concurrent validity for elbow and knee angle measurements, although they haven’t been comprehensively investigated in relation to the trunk or sport-specific movements. This study aimed to determine the influence of MIMU functional calibration on thorax and lumbar joint angles for uni-planar and multi-planar, sport-specific tasks. It was hypothesised that simple MIMU Euler angle decomposition would produce larger differences than functionally calibrated angles when both approaches were compared with concurrently collected 3D retro-reflective motion capture data. Ten fast-medium cricket bowlers were simultaneously recorded by MIMUs and retro-reflective motion capture. Five joint angle calculation methods were assessed, with three incorporating functionally-defined axes. Statistical parametric mapping and root mean squared differences (RMSD) quantified measurement differences. Statistical parametric mapping found no significant differences between MIMU and retro-reflective data for any of the calculation methods. The RMSDs for the functionally calibrated methods were not significantly smaller than the non-functional RMSDs. Functional calibration may be unnecessary for MIMU-based measurement of thorax and lumbar joint angles, with simple Euler angle decomposition adequate.
5.2 Introduction

The rapid evolution of wearable technologies over the last decade has provided biomechanists with alternatives to traditional lab-based technologies, such as three-dimensional (3D) retro-reflective motion capture systems. The magneto-inertial measurement unit (MIMU) is an attractive wearable device to biomechanists because it provides 3D orientation data, whilst being portable and relatively user friendly. Magneto-inertial measurement units incorporate three tri-axial sensors: accelerometers, gyroscopes and magnetometers. Data from the sensors can be fused using filtering algorithms, such as Kalman filters (Kalman, 1960), to facilitate 3D orientation estimation. Affixing and aligning MIMUs to body segments allows the orientation of the segment to be approximated. Magneto-inertial sensor measurement of joint kinematics involves MIMUs being placed on proximal and distal segments, with the relative orientation between the MIMUs expressed with respect to a reference frame (Picerno, 2017). Commercially available MIMU manufacturer software often allows calculation of sensor-to-sensor angles via Euler angle decomposition (Picerno, 2017). Euler angle decomposition is the simplest way of representing clinically interpretable MIMU-based joint angles. Joint Euler angle rotations often occur in order of decreasing range of motion (Cole et al., 1993; Godwin et al., 2009). For many joints, a ZXY decomposition order is recommended, where Z is the medio-lateral (flexion-extension) axis, X is the antero-posterior (adduction/abduction) axis and Y is the vertical (axial rotation) axis (Cole et al., 1993; Wu et al., 2005, 2002).

The alignment of a sensor to the relevant body segment is also required for functionally interpretable MIMU angles but is not addressed by simple Euler angle decomposition. Anatomical and functional calibration procedures have been described (Picerno, 2017) to overcome this problem. Anatomical alignment is based on static calibration poses and can incorporate additional calibration implements (Picerno et al., 2008). Utilising calibration pointers or rigs can be time-consuming and is not ideal for in-field motion capture. Functional alignment involves the estimation of joint rotation axes from simple uni-planar movement trials (Picerno, 2017). High repeatability of angle measurement has been reported for the ankle (O’Donovan et al., 2007), knee (Cutti et al., 2010; Favre et al., 2009) and elbow (Ligorio et al., 2017) when functionally-defined MIMU joint axes are used. Root mean squared differences (RMSD) of 1-8° have been documented when functionally calibrated MIMU-based angles were compared with angles calculated from reference motion capture systems (Favre et al., 2009; Ligorio et al., 2017; O’Donovan et al., 2007). Another study also reported angle differences of 6.4±4.7° between the functionally defined MIMU coordinate system of the thorax and the 3D retro-reflective, anatomically specific, thorax segment coordinate system (de Vries et al., 2010). Though this is a useful finding, de Vries and colleagues (2010) only assessed the differences in segment coordinate system orientations and did not
investigate the impact of these differences on measurement of downstream joint angles during dynamic movements. When compared with anatomically calibrated MIMU joint angles, functional calibration has improved concurrent validity for knee (Favre et al., 2009) and elbow (Ligorio et al., 2017) joint angle measurements. The mentioned studies assessed low velocity, clinical movements (de Vries et al., 2010; Ligorio et al., 2017; O’Donovan et al., 2007) or walking gait (Cutti et al., 2010; Favre et al., 2009). The concurrent validity of employing functional calibration for MIMU joint kinematics has not been assessed for dynamic, sport-specific tasks. Functional calibration has also not been explored for the lumbar segment nor fully validated for the thorax segment.

A small sample of studies (Bergamini et al., 2013; Brice et al., 2018; Zhang et al., 2015, 2013) have reported concurrent validity for sport-specific MIMU trunk kinematics by comparing against the gold-standard of motion capture: 3D retro-reflective motion capture. Three of the four studies only reported global trunk inclination or orientation (Bergamini et al., 2013; Zhang et al., 2015, 2013). Brice and colleagues (2018) measured shoulder-pelvis separation angle during discus throwing using MIMUs but compared the angles against retro-reflective measures that incorporated anatomically defined body segments. They reported RMSDs of 6-13° between the MIMU-based measures and retro-reflective based measures. It is likely that discrepancies between MIMU and anatomical coordinate system orientations (Chardonnens et al., 2012) and inconsistencies in soft tissue movement artefacts (Brice et al., 2018; Camomilla et al., 2018; Wong and Wong, 2008) heavily contributed to the measurement differences they reported.

Functionally calibrated MIMU joint angles for the ankle (O’Donovan et al., 2007), knee (Cutti et al., 2010; Favre et al., 2009) and elbow (Ligorio et al., 2017) have been shown to be repeatable and concurrently valid, when assessed against reference motion capture data. There is a small sample of sport-specific trunk MIMU studies (Bergamini et al., 2013; Brice et al., 2018; Zhang et al., 2015, 2013) but the efficacy of MIMU functional calibration for the trunk has not yet been fully assessed. Therefore, the aim of this study was to determine the impact of MIMU functional calibration on thorax and lumbar joint angles for moderate speed, uni-planar tasks and high-speed, multi-planar tasks. This was assessed by comparing MIMU derived angles with concurrently collected 3D retro-reflective data for uni-planar trunk range of motion (ROM) tasks, and a multi-planar, high-speed, sport-specific task: cricket fast bowling (Burnett et al., 1998; Ferdinands et al., 2009). It was hypothesised that a MIMU-based ZXY Euler angle decomposition for thorax and lumbar angles would produce greater measurement differences than functionally calibrated angles when both methods were compared with 3D retro-reflective derived angles.
5.3 Methods

5.3.1 Participants

Seven male (181.1±6.9 cm, 81.4±13.4 kg, 27±7.7 years) and three female (173.8±5.5 cm, 66.7±4.9 kg, 18.3±4.2 years) cricket fast or fast-medium bowlers were recruited from community or state squads for the criterion-related validation study. Informed consent was obtained from all participants or a parent/guardian for participants under the age of 18. Ethical approval was granted by the University of Western Australia Human Research Ethics Committee (RA/4/1/2593) and the Curtin University Ethics Office (HRE2016-0472). All research was conducted in accordance to the principles of the Helsinki Declaration.

5.3.2 Data Collection

Magneto-inertial sensor data were concurrently collected with retro-reflective trajectory data to enable the MIMUs to be assessed against a criterion measure. Data collection was completed in the indoor Motion Analysis Laboratory at Curtin University, Western Australia. Three XSens Mtw Awinda MIMUs (Xsens Technologies B.V., Enschede, Netherlands) (75 Hz sampling frequency, ±2000°/s gyroscope, ±160 m/s² accelerometer, ±1.9 Gauss magnetometer) were placed on the thorax, lower-back, and pelvis of all participants. The thorax sensor’s cranial edge was positioned between the spinous processes of the seventh cervical (C7) and first thoracic (T1) vertebrae. The lower-back MIMU was placed on the spinous process of the first lumbar vertebra (L1). The pelvis MIMU was positioned with its caudal edge on the spinous process of the second sacral vertebra. MIMU data were captured by the manufacturer’s software (MT Manager 4.2.1).

Retro-reflective marker trajectories were recorded by 20 MX and T-series Vicon cameras (Oxford Metrics, Oxford, UK), sampling at 300 Hz. Each MIMU was affixed with a three-marker rigid plate (Figure 5.1), facilitating creation of a retro-reflective technical coordinate system (RRtech). The marker-plates were attached to the MIMUs using velcro, with the orientation of the markers aligned with the axes of the MIMU case as closely as possible. The coordinate system of each MIMU was further adjusted to match the RRtech orientation during data post-processing (detailed below). Alignment procedures ensured that differences in segment coordinate system orientations did not impact comparisons between the MIMU and retro-reflective systems. It also ensured that movement artefact error would be consistent between MIMU and retro-reflective systems. The RRtech was created by using the mean position of the three markers as the origin of the coordinate system. The first and second defining lines were from marker-2 to marker-1 and marker-3 and marker-2 respectively. The first defining line was parallel with the y-axis (vertical), with the x-axis
the cross-product of the first and second defining lines, and the z-axis orthogonal to the y and x axes (Figure 5.1).

Figure 5.1: IMU and affixed rigid retro-reflective marker plate. Markers (M1, M2, M3), defining lines (D1, D2), and coordinate system axes (X, Y, Z) are indicated. X-axis is antero-posterior. Y-axis is vertical. Z-axis is medio-lateral.

Two AMTI (Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA) force plates (1800 Hz) captured back foot contact and front foot contact of the bowling delivery stride. A 300 Hz Vicon Bonita video camera (Oxford Metrics, Oxford, UK), synchronised to the retro-reflective system, was positioned sagittally and used to identify the ball release frame during data processing. The first frame when the ball was no longer in contact with the hand was determined as ball release.

Participants completed a series of uni-planar ROM tasks. Each individual performed five moderately paced, trunk flexion-extension movements to end of movement range whilst standing, followed by five standing left and right trunk lateral flexion movements. Participants then repeated the tasks whilst seated before also completing five seated left and right trunk rotation ROM cycles. Five successful bowling trials (multi-planar movement) performed at match-like intensity were also recorded.

5.3.3 Data Processing

Retro-reflective trajectory data were labelled, filtered and modelled in Vicon Nexus 2.5 software (Oxford Metrics, Oxford, UK). Marker trajectories for ROM and bowling trials were filtered using a fourth-order, zero-lag, low pass Butterworth filter. Cut-off frequencies of 9 Hz and 15 Hz were selected for ROM and bowling trials respectively, following a residual analysis (Winter, 1990). Retro-reflective data were modelled by a customised Vicon Bodybuilder model (Oxford Metrics, Oxford, UK).
The MIMU free acceleration and orientation data for each trial were output as a .csv file using the Xsens MT Manager software (Xsens Technologies B.V., Enschede, Netherlands). Sensor orientations were estimated by the manufacturer’s inbuilt Kalman filters and were expressed in quaternions.

Further processing was performed by a customised LabVIEW 2017 program (National Instruments Corp, Austin, Texas, USA). This program (1) spatially aligned the coordinate systems of each MIMU and the overlaying RRtech, (2) temporally synchronised MIMU and retro-reflective data, (3) calculated functional (helical) axes for each joint (relative and global) from the ROM trials, (4) calculated joint angle values using five different joint angle calculation methods, and (5) compiled and output all angle values for all trials. These steps are detailed below.

To spatially align the Xsens MIMU coordinate systems with their paired RRtech, each MIMU coordinate system was rotated -90° about the z-axis and then 90° about the y-axis. A further refinement process utilised data from the seated trunk flexion-extension ROM trial. It involved calculating the orientation difference between the MIMU coordinate systems and RRtech orientations and then accounting for this difference. For each MIMU/RRtech pair, the orientation difference was calculated by multiplying the RRtech quaternion by the inverse of the MIMU quaternion. The difference in their orientations at each time-point was averaged over the course of the trial, with this correction value then applied to the MIMU data. Once these alignment procedures were completed, it was possible to calculate global angles by comparing the RRtech or the MIMU orientations with the retro-reflective 3D global reference frame.

Temporal synchronisation for ROM trials was achieved by cross-correlating the angular component of the axis-angle representation of the thorax MIMU and its matching RRtech. The lumbar MIMU and RRtech were used instead for the sitting flexion-extension ROM trials. The cross-correlation enabled time-lag between the two systems to be calculated and accounted for. Bowling trials were synchronised by cross-correlating the thorax MIMU and RRtech orientations during the bowling action. Refinements involved using the acceleration profiles of the lumbar MIMU and RRtech.

Functional axes for each joint were determined by a two-step process. Firstly, an initial functional axis estimate was produced by using the vector representing the average angular velocity axis. Secondly, this was refined using an unconstrained optimisation function, incorporating the Broyden quasi-Newton method (Broyden, 1967). In this optimisation, the functional axis vector was rotated by all the orientations of the MIMU/RRtech joint and the cost of this functional axis was defined as the sum of the variance of the x, y and z components of the rotated vectors. The vector with the minimum cost was determined the optimal functional axis. If $Q_{\text{Joint}}$ is the joint quaternion and $FA$ is the functional axis, then this can be expressed mathematically by:
Chapter 5: Functionally Calibrated MIMU-based Thorax and Lumbar Measurements

\[ FA'_{i} = Q_{\text{Joint}i} \times FA \times Q_{\text{Joint}i}^{-1} \]

\[ FA = (0, FA_x, FA_y, FA_z) \]

\[ \min_{FA_x, FA_y, FA_z} \left( \sigma_{FA_x}^2 + \sigma_{FA_y}^2 + \sigma_{FA_z}^2 + (\|FA\| - 1)^2 \right) \]

**Equation 5.1:** Equation for determining the optimal functional axis for each joint.

This process was completed for each joint relative to the pelvis (S2) MIMU/RRtech, relative to the child segment and relative to the retro-reflective coordinate system global origin. The three seated ROM trials were used to calculate the flexion-extension, lateral flexion and rotation functional axes.

The five joint angle calculation methods were fully defined once the functional axes were determined for each joint. These angle calculation methods were applied to both the retro-reflective and MIMU data. The five calculation methods were termed ‘ZXY’, ‘GS’, ‘GSfunctional’, ‘Functional-1’, and ‘Functional-2’. Parent and child segments for all methods were created using the following framework:

\[ \text{Segment} = \langle \text{Origin point} \mid \text{Defining Line 1} \mid \text{Defining Line 2} \mid \text{Axis Order} \rangle \]

**Equation 5.2:** Definition framework for each parent and child segment

Parent and child segment definitions varied depending on the calculation method used. Table 5.1 displays axis definitions for YXZ, ZXY and XYZ ordered segments.

**Table 5.1:** Axis definitions for YXZ, ZXY and XYZ ordered segments.

<table>
<thead>
<tr>
<th>Axis Order</th>
<th>X</th>
<th>Y</th>
<th>Z</th>
</tr>
</thead>
<tbody>
<tr>
<td>YXZ</td>
<td>Defining Line 2 × Defining Line 1</td>
<td>Defining Line 1</td>
<td>-Y × X</td>
</tr>
<tr>
<td>ZXY</td>
<td>Defining Line 2 × Defining Line 1</td>
<td>Z × X</td>
<td>Defining Line 1</td>
</tr>
<tr>
<td>XYZ</td>
<td>Defining Line 1</td>
<td>Defining Line 2 × Defining Line 1</td>
<td>X × Y</td>
</tr>
</tbody>
</table>

The defining line definitions and order of axes for the child and parent segments varied for the five calculation methods (Table 5.2). The calculation methods used basic child and parent segments unless specified otherwise. These basic segments were defined by the marker positions of the RRtech (Figure 5.1).
Table 5.2: Defining line and axis orders for child and parent segments. The MIMU and retro-reflective systems utilised basic parent and child segments unless specified otherwise. FE = flexion-extension; LF = lateral flexion; ROT = rotation; FA = functional axis; M1, M2, M3 = marker 1, 2, 3; GS = Grood and Suntay joint coordinate system; GSF = Grood and Suntay joint coordinate system with functional axes.

<table>
<thead>
<tr>
<th></th>
<th>Defining Line 1</th>
<th>Defining Line 2</th>
<th>Axis Order</th>
</tr>
</thead>
<tbody>
<tr>
<td>PARENTBasic</td>
<td>M1-M2</td>
<td>M3-M2</td>
<td>YXZ</td>
</tr>
<tr>
<td>CHILDBasic</td>
<td>M1-M2</td>
<td>M3-M2</td>
<td>YXZ</td>
</tr>
<tr>
<td>CHILD_F-AFE</td>
<td>FA:FE</td>
<td>CHILDBasic-Y</td>
<td>ZXY</td>
</tr>
<tr>
<td>CHILD_F-LF</td>
<td>FA:LF</td>
<td>CHILDBasic-Z</td>
<td>XYZ</td>
</tr>
<tr>
<td>CHILD_F-ROT</td>
<td>FA:ROT</td>
<td>-CHILDBasic-Z</td>
<td>YXZ</td>
</tr>
<tr>
<td>CHILD_GSF</td>
<td>CHILDBasic-Y</td>
<td>-FA:FE</td>
<td>YXZ</td>
</tr>
<tr>
<td>PARENT_GSF</td>
<td>FA:FE</td>
<td>FA:ROT</td>
<td>ZXY</td>
</tr>
<tr>
<td>CHILD_GS</td>
<td>CHILDBasic-Y</td>
<td>-CHILDBasic-Z</td>
<td>YXZ</td>
</tr>
<tr>
<td>PARENT_GS</td>
<td>PARENTBasic-Z</td>
<td>PARENTBasic-Y</td>
<td>ZXY</td>
</tr>
</tbody>
</table>

Joints were created once parent and child segments were defined. The ZXY method was a ZXY-ordered Euler angle decomposition, which did not utilise functional axes. The joint for this method can be expressed as:

$$Joint = \left(Child_{Basic}^{-1} \times Parent_{Basic}\right)^{-1}$$

Equation 5.3: Joint definition for the ZXY Euler angle decomposition method.

Lateral flexion (x) and rotation (y) were negated for the ZXY method. The GS method was a Grood and Suntay (1983) joint coordinate system, which also did not incorporate functional axes (see Table 5.2). A limitation of the GS method is its relative unreliability when used to calculate angles close to zero degrees (Dabirrahmani and Hogg, 2017). In an effort to address this, the ‘atan2’ function was used in place of the flexion and rotation angle equations defined by Grood and Suntay. Rotation (y) was negated for the GS method.

The three remaining calculation methods all utilised the functional joint axes. The Grood and Suntay joint coordinate system was again employed for GSF, however in this case the flexion-extension functional axis was used to define the parent segment’s z-axis (Table 5.2). The ‘atan2’ function was again incorporated, with rotation (y) also negated.

The Functional-1 method used the functional axes when defining child segments for each plane of motion (Table 5.2). When calculating the flexion-extension joint angle, a child segment created using the flexion-extension functional axis as the segment z-axis was used. For calculating lateral flexion angle, the lateral flexion functional axis was used to define the x-axis of the child segment, and for calculating the rotation angle the rotation functional axis was used to define the y-axis of the child segment. The basic parent segment was used and the standard ZXY Euler angle decomposition order was maintained for all three calculations. If FA-FE is the flexion-extension
functional axis, then the joint used to calculate flexion-extension for the Functional-1 method can be expressed as:

\[
Joint = (Child_{FA-\text{FE}}^{-1} \times Parent_{\text{Basic}})^{-1}
\]

**Equation 5.4: Joint definition for the Functional-1 method.**

Finally, the Functional-2 method did not use the functional axes to define the child segment, instead calculating the angular difference between each functional axis and the corresponding basic child segment axis. This was expressed as the quaternion difference, which was then used to rotate the joint rotation quaternion, before the angles were decomposed as described in the Functional-1 method. Below is an example of how the flexion-extension joint was created for this method:

\[
Joint = (Child_{\text{Basic}}^{-1} \times Parent_{\text{Basic}})^{-1} \times JointAlign_{FA-\text{FE}}
\]

\[
JointAlign_{FA-\text{FE}} \text{ Axis} = Z_{Child_{\text{Basic}}} \times FAxis_{\text{FE}}
\]

\[
JointAlign_{FA-\text{FE}} \text{ Angle} = \arccos(Z_{Child_{\text{Basic}}} \cdot FAxis_{\text{FE}})
\]

**Equation 5.5: Joint definition for the Functional-2 method.**

### 5.3.4 Data Analysis

The five angle calculation methods were employed for 12 different variables. Global thorax angles (thorax IMU to global), global lumbar angles (lower-back MIMU to global), thorax-to-pelvis relative angles (thorax MIMU to pelvis MIMU), and lumbar-to-pelvis relative angles (lower-back MIMU to pelvis MIMU) were analysed across all three planes of motion (flexion-extension, lateral flexion, and rotation). These angles were assessed for the ROM trials and for the two bowling trials of best data quality (least marker occlusion) for each participant, which was determined by visual inspection of retro-reflective trajectory data. The MIMU angles were only compared with retro-reflective angles calculated using the same method. Bowling and ROM continuous data were both time-normalised to 101 data-points. Back foot contact (on one of the force plates) and ball-release were used to define the bowling stride phase. The thorax-to-pelvis RRtech angle was utilised for ROM phase definition, with a 0° measurement (neutral trunk position) indicating the start and end of the phase.

Continuous MIMU and retro-reflective angle waveforms were analysed using one-dimensional statistical parametric mapping (SPM1D) (Pataky et al., 2013). Publicly available SPM1D source code (Pataky, 2018) was incorporated into customized MATLAB scripts (Mathworks, Natick, Massachusetts, USA). The SPM1D analysis enabled grouped continuous data to be assessed for significant differences \(p < 0.05\) across an entire phase, rather than just discrete time-points.
However, SPM1D relies on group means and standard deviations and is not particularly sensitive to smaller differences between groups, particularly when there are large group standard deviations. To address this limitation, the customized MATLAB program (Mathworks, Natick, Massachusetts, USA) also calculated root mean square differences (RMSD) for each individual trial. Root mean square differences have been previously used in similar studies (Bergamini et al., 2013; Ligorio et al., 2017; Picerno et al., 2008; Wells et al., 2018). They were calculated at the minimum (RMSD-min) and maximum (RMSD-max) joint angle values during the phase. A third RMSD value was calculated at the time-point coinciding with the greatest difference between the MIMU and retro-reflective angles, as determined by the t-trace from the SPM1D analysis. This was termed RMSD-tmax. The RMSD-max and RMSD-min measures could occur at differing time-points for MIMU and retro-reflective data, whilst RMSD-tmax occurred at the same time-point for both systems. The RMSD values were categorised by trial type (ROM or bowling), and also sub-categorised by angle type (e.g., thorax-to-pelvis angle) and plane of motion (e.g., sagittal/flexion-extension). This involved calculating the mean value for all of the RMSD measures in a given category (e.g., the mean RMSD for all thorax-to-pelvis angles during the ROM trials). Assessing data in this way enabled trends to be easily identified and simplified the presentation of a large quantity of data.

SPSS statistics (IBM, Armonk, New York, USA) was used to calculate a one-way analysis of variance (ANOVA) with a partial Bonferroni post-hoc correction to assess whether there were significant differences ($p < 0.005$) between RMSD values of the ZXY calculation method and the other four methods (GS, GSfunctional, Functional-1, and Functional-2). Using the STATA ‘power oneway’ command (STATA Corp, College Station, Texas, USA), a power calculation was performed of the ANOVA procedure used. There were 200 data points in each of the five groups in each ANOVA for the ROM trials and 240 for the bowling trials. So, with an alpha level of 0.005, a minimum of 200 data points in each of the 5 groups and a within-group standard deviation as large as 3.0° (a variance of 9.0°), there was 80% power to detect a between-group difference as small as 0.2°.

5.4 Results

The SPM1D analysis produced no significant differences for any angle calculation methods across all trial types and angles.

The RMSD-max, RMSD-min, and RMSD-tmax values for each calculation method were averaged for ROM across all joints and all planes of motion (Figure 5.2). The values range from 2.1° to 3.6°. Despite some statistically significant main effects ($p < 0.05$), no significant differences (Bonferroni corrected $p < 0.005$) were found when the ZXY method was compared with the GS, GSfunctional, Functional-1 or Functional-2 methods at post-hoc analysis. There were no significant differences
found for the bowling trials across all joints and planes of motion (Figure 5.3). Bowling RMSD values (4.6° to 8.5°) were larger than for the ROM trials.

The RMSD values were also averaged for joint and movement type across all planes of motion. GSfunctional RMSD-min values (4.6±1.3°) were found to be approximately 2.5° greater than ZXY values (2.0±1.2°, p = 0.002) for ROM lumbar-to-pelvis angles following post-hoc testing.

Finally, the RMSD values were averaged for plane of motion and movement type across all joints. GSfunctional RMSD-max values (4.4±1.6°) were significantly greater than ZXY (2.7±1.2°, p = 0.004) for coronal plane movement (lateral flexion) in the ROM trials.

Statistical tables displaying full RMSD and ANOVA statistics can be found in the supplementary material for this paper in 9.1: Appendix A.
5.5 Discussion

This study appears to be one of the first studies to assess the efficacy of using functionally calibrated MIMU joint angle calculation methods for application to sport-specific thorax and lumbar data. Generally, the findings suggest that there are minimal differences between functionally calibrated joint angle calculation methods (GSfunctional, Functional-1, and Functional-2) and non-functionally calibrated calculation methods (ZXY and GS) when they are assessed against concurrently collected retro-reflective data. This is evident from the SPM1D analysis of the grouped continuous data, which found no significant differences between MIMU and retro-reflective data for any angle calculation method.

The RMSD analysis indicates that there are small differences (2.1° to 3.6°) between the MIMU-based measures and retro-reflective based measures when averaged across all ROM trials (Figure 5.2). These measurement differences are similar to previous research (Mjøsund et al., 2017; Najafi et al., 2015; Theobald et al., 2012). As expected, there were larger average RMSD values calculated for the bowling trials (4.6° to 8.5°) (Figure 5.3). Cricket fast bowling involves high-speed, multi-planar thorax and lumbar movements occurring through large ranges of motion (Bayne et al., 2016; Burnett et al., 1998), hence these values compare well with previous studies (Bergamini et al., 2013; Brice et al., 2018; Zhang et al., 2015), which analysed less dynamic, sport-specific movements (discus, sprinting and cycling). The 75 Hz sampling frequency of the MIMU system may have contributed to these slightly larger RMSD values, with a faster sampling rate more desirable for high-speed tasks.

It was hypothesised that incorporating functionally calibrated axes into MIMU thorax and lumbar joint angle calculations would produce smaller measurement differences than a ZXY Euler angle decomposition when both approaches were compared with 3D retro-reflective based joint angles. The data presented did not support that hypothesis. The ZXY method did not produce significantly larger RMSD values than any of the other four methods, three of which incorporated functional axes (GSfunctional, Functional-1 and Functional-2). That result suggests that there are no significant improvements to MIMU concurrent validity to be gained by employing angle calculation methods that incorporate functionally defined joint axes.

The GSfunctional RMSD values were found to be significantly larger than the ZXY RMSD values for lumbar-to-pelvis angles and coronal plane movements during the ROM tasks, indicating that incorporating functional axes can occasionally increase measurement differences to the motion capture gold-standard. The differences between the GSfunctional and ZXY methods in these instances were small (1.7° to 2.6°), hence the functional implications of this finding are likely inconsequential. It is possible that filtering and data cleaning discrepancies between the MIMU and
retro-reflective systems may have contributed to differences in the calculation of functional axes, which subsequently affected the angle outputs for the GSfunctional method.

A major limitation of this study was that MIMU angles were only compared with retro-reflective angles quantified using the same calculation method. This was decided because none of the five calculation methods analysed can be truly considered a ground-truth measure. Therefore, this study was limited because measurement accuracy could not be reported and this should be considered when interpreting all findings. Additionally, a reliability analysis of the calculation methods was not undertaken and the methods were only assessed for calculation of thorax and lumbar angles. Hence, further investigation on MIMU angle calculation methods is required.

The functional implications of the study are that MIMU functional calibration does not appear to be necessary for measurement of thorax and lumbar joint angles during uni-planar and multi-planar tasks. A ZXY Euler angle decomposition is a valid method of calculating thorax and lumbar joint angles, when assessed against concurrently collected 3D retro-reflective data.

5.6 Conclusion

Employing functional calibrations for MIMU-based thorax and lumbar joint angle calculations did not significantly decrease measurement differences when compared with concurrently collected 3D retro-reflective based data. A ZXY Euler angle decomposition appears to be appropriate for MIMU thorax and lumbar angle measurements for uni-planar and high-speed, sport-specific multi-planar movements.
5.7 References


Chapter 6: Comparison of Trunk Angle Measures from Two Commercial MIMU Systems

Cottam, D. S., Campbell, A. C., Davey, P. C., Kent, P., Elliott, B. C., & Alderson, J. A. Measurement of uni-planar and sport specific trunk motion using magneto-inertial measurement units: A comparison of two similar commercially available systems.

Prepared for submission

Intended Submission Target: Sensors

Intended Submission Date: June 2019.
Research Linking Statement

Chapter 5 explored the effects of magneto-inertial measurement unit (MIMU) functional calibrations on thorax and lumbar angle measurements during trunk range of motion tasks (ROM) and cricket bowling. It was found that incorporating functionally-defined axes of rotation in thorax and lumbar angle calculations did not significantly improve the concurrent validity of the MIMU measures when compared with the three-dimensional retro-reflective motion capture outputs. The take-away message of the study was that simple angle calculation methods, such as standard Euler angle decomposition, may be adequate for MIMU measurements of thorax and lumbar angles during uni-planar and high-speed, multi-planar tasks. Whilst it is possible that the findings of Chapter 5: may be applicable to all commercial MIMUs, previous research has shown that measurement validity, reliability and accuracy can vary considerably between commercial MIMU systems. Magneto-inertial sensor comparison studies have primarily focused on assessing measurements during static or low-velocity conditions. A comparison of MIMU system measurement validity during high-velocity, complex sporting movements such as cricket bowling has not been presented. This study compared the measurement validity of two similar commercially available MIMU systems when estimating thorax and lumbar angles during trunk ROM tasks and cricket fast bowling. Three-dimensional retro-reflective motion capture was used as the criterion measure. It is anticipated the information in this study can be used by consumers and researchers when considering MIMU application in clinical and sporting contexts.
6.1 Abstract

There is a range of magneto-inertial measurement unit (MIMU) systems currently commercially available, however sensor specifications and fusion methods vary considerably between systems. Different MIMUs have been compared during static or low-velocity conditions, with higher-velocity movements assessed in robotic-based studies. This study aimed to assess the concurrent validity of two commercial MIMU systems when measuring trunk angles during uni-planar range of motion (ROM) and cricket bowling, which involves high-speed, multi-planar movements. Both systems utilised comparable sensor specifications and Kalman filter fusion algorithms. The MIMU data were compared with concurrently captured three-dimensional retro-reflective data for 10 fast-medium bowlers. It was hypothesised that the MIMU measurement differences to criterion retro-reflective data would be similar for both MIMUs. Statistical parametric mapping and root mean squared differences (RMSD) were computed for both MIMU systems. Statistical parametric mapping showed no significant differences for both MIMU systems when compared with retro-reflective data outputs. The MIMU systems produced ROM RMSDs between 1.4±1.0° and 2.6±1.5°. One system displayed RMSDs between 4.6±1.4° and 7.4±1.9° during bowling, indicating functionally significant differences to retro-reflective data outputs. There were some small but statistically significant differences in RMSDs between the MIMU systems, suggesting that comparable MIMU systems can produce varying measurements during ROM and bowling tasks.
6.2 Introduction

Wearable technology has become progressively more integrated into everyday life within the last decade. Magneto-inertial measurement units (MIMUs) are a type of wearable technology that are of particular interest to biomechanists due to their ability to measure three-dimensional (3D) orientation outside of the traditional confines of the laboratory. Attaching and aligning MIMUs to body segments allows segmental kinematics to be approximated. Joint kinematics can also be calculated from magneto-inertial sensor data, although this requires further calibration and computation (Picerno, 2017).

Commercial MIMU models comprise three tri-axial sensors: accelerometers, gyroscopes, and magnetometers. The specifications of these sensor components can vary from one MIMU system to another, and are usually dependant on the system’s cost and intended application (e.g., sport-specific analysis, clinical gait assessments). Analysis of high-velocity movements requires increased accelerometer and gyroscope specifications in order to avoid sensor saturation and subsequent data ‘clipping’ (Nam et al., 2014; Wells et al., 2018; Zhang et al., 2015). Signal aliasing of high-speed movement is also a potential problem if a MIMU’s sampling rate is not sufficient to capture the speed of the movement. To estimate the 3D orientation of a MIMU in space, accelerometer, gyroscope and magnetometer data are fused via algorithms, which are usually incorporated into the manufacturer’s proprietary software. Kalman filters (Kalman, 1960) are the sensor fusion approach most commonly utilised by manufacturers (Lopez-Nava and Munoz-Melendez, 2016). Kalman filter algorithms predict future states from current information and a set of control parameters (Simon, 2001), however the exact algorithms and parameters used may vary between commercial MIMU systems (Lopez-Nava and Munoz-Melendez, 2016). Other data fusion methods, such as complementary filtering algorithms, vector observation algorithms and integration-based algorithms have also been used to estimate the orientation of MIMUs (Lopez-Nava and Munoz-Melendez, 2016). The precise algorithms and parameters used by manufacturers are usually undisclosed (deemed proprietary for commercial advantage), suggesting that there is potential for between system reliability and validity variability.

A small sample of studies have compared performance between various MIMU models. Picerno and colleagues (2011) assessed the static orientation accuracy of nine different MIMU models. A worst-case measurement discrepancy of 5.7° was found between the selected MIMU outputs. Measurement differences during movement have been classified as functionally insignificant below 5° (Intolo et al., 2010; Mjøsund et al., 2017), therefore it seems reasonable to suggest that static orientation differences should be smaller than that threshold in application. Lebel and colleagues (2013, 2015) examined the effect of velocity and time on the measurement accuracy of three
MIMU models. By utilising a specialised robotic Gimbal table capable of movement in all three planes of motion individually and simultaneously, they concluded that increasing the velocity or duration of movement can drastically increase measurement errors to well above 5° in some of the MIMUs, whilst others performed relatively consistently and within acceptable error ranges. Importantly, some of the measurement errors reported substantially varied from the error bandwidths stated by the manufacturers. Bergamini et al. (2014) assessed the measurement validity of three different approaches to MIMU orientation estimation in vivo. The methods were based on numerical integration, Kalman filtering and complementary filtering. Integration-based orientation estimation produced significantly larger ($p < 0.05$) heading (yaw angle) measurement errors than the other two filtering methods during gait and a manual task, with significantly larger ($p < 0.05$) attitude (pitch and roll) measurement errors also present during the manual task. Of concern, large heading measurement root mean square errors were also recorded during gait for the Kalman and complementary based filters ($\approx 20°$), albeit these were less than the errors shown for the integration based method ($>30°$). Heading and attitude errors for the Kalman and complementary based filters during the manual task were approximately 5° or less, whereas the integration based method led to errors of $\approx 10°$. A similar study examined the effect of different Kalman filter algorithms on MIMU orientation measures when both algorithms were applied to the same MIMU system (Ligorio et al., 2016). The authors reported small but statistically significant measurement differences between the two approaches for a MIMU positioned on the lower back during a ‘timed-up-and-go’ task. In this case, the largest measurement differences between the filtering approaches were seen for the attitude angles. A final study documented the effects of manipulating many MIMU system characteristics which included alterations to the sensor algorithm type and algorithm parameters; the sensor components used (e.g., no magnetometer); and the method used to calculate kinematics (e.g., Euler angles, quaternions) (Filippeschi et al., 2017). The five methods assessed in the study showed differences in measurement of uni-planar, upper limb motion. Measurement errors for elbow flexion-extension (39 mm to 89 mm), shoulder flexion-extension (83 mm to 214 mm), and shoulder adduction/abduction (66 mm to 272 mm) varied depending on the method employed.

The comparison studies mentioned have all indicated that MIMU based measurements can vary depending on the MIMU model or sensor fusion approach adopted. However, these studies focused primarily on measurement of static orientation (Picerno et al., 2011), uni-planar movement (Filippeschi et al., 2017; Lebel et al., 2013, 2015), or low-velocity tasks (Bergamini et al., 2014; Ligorio et al., 2016), despite increasing movement complexity and velocity being shown to increase MIMU measurement errors (Godwin et al., 2009; Lebel et al., 2013; Pasciuto et al., 2015). Lebel and colleagues (2013, 2015) also assessed multi-planar rotations and higher-speed movements, however these were performed by a robotic Gimbal table with a maximum angular velocity of
Angular velocities during sporting movements are often more than double this value (Burnett et al., 1998; Salter et al., 2007; Wells et al., 2018; Yeow et al., 2009), which raises further questions about the between system variation in MIMU-based measurement when quantifying high-speed, sport-specific movements.

This study aimed to assess the concurrent validity of two similar, commercial MIMU systems by comparing their measurements to gold-standard, retro-reflective motion capture data. Joint angle measurements from both MIMU systems were assessed during uni-planar trunk range of motion (ROM) tasks and cricket fast bowling, which involves high-speed, multi-planar trunk movements (Burnett et al., 1998; Ferdinands et al., 2009). It was hypothesised that measurement differences to the criterion retro-reflective motion capture system would be comparable for both MIMU systems. No functionally significant differences (greater than 5°) were expected between MIMU-based angles and the retro-reflective derived angle outputs.

6.3 Methods

6.3.1 Participants

Sixteen fast or medium cricket bowlers (181.2±6.6 cm, 76.9±10.7 kg, 22.6±6.9 years) were recruited from community or state cricket squads for this criterion-based validation study. Informed consent was granted by all participants or a parent/guardian if a participant was under the age of 18. Ethical approval was provided by the University of Western Australia Human Research Ethics Committee (RA/4/1/2593) and the Curtin University Ethics Office (HRE2016-0472).

6.3.2 Data Collection

Data from two MIMU systems were collected separately but each was collected concurrently with 3D retro-reflective trajectory data, which acted as the criterion measure. Data collection was completed in the indoor Motion Analysis Laboratory at Curtin University, Western Australia. The first MIMU system assessed was the Xsens Mtw Awinda system (Xsens Technologies B.V., Enschede, Netherlands) (75 Hz sampling frequency, ±2000°/s gyroscope, ±160 m/s² accelerometer, ±1.9 Gauss magnetometer). The second system was the Noraxon MyoMOTION Research Pro system (Noraxon USA Inc., Scottsdale, Arizona, USA) (100 Hz sampling frequency, ±2000°/s gyroscope, ±16 g accelerometer, ±1.9 Gauss magnetometer). Data were collected from 10 participants for both the Xsens (178.9±7.1 cm, 77±13.3 kg, 24.4±7.8 years) and Noraxon (184.5±5.3 cm, 80±9.3 kg, 21.6±3.9 years) systems. Participant and MIMU system availability meant not all participants were able to complete the data collection protocol for both MIMU systems. Therefore, the available dataset for each system contained data from different individuals, with four
participants appearing in both datasets. For each MIMU system, three sensors were placed on the thorax, lower-back and pelvis of each individual. The thorax sensor was positioned so that its cranial edge was between the spinous processes of the seventh cervical (C7) and first thoracic (T1) vertebrae. The lower-back sensor was affixed over the spinous process of the first lumbar vertebra (L1). The pelvis MIMU was placed with its caudal edge on the spinous process of the second sacral vertebra (S2). Xsens MIMU data were captured by MT Manager 4.2.1 software (Xsens Technologies B.V., Enschede, Netherlands). Noraxon MIMU data were captured by MyoRESEARCH 3.8.2 software (Noraxon USA Inc., Scottsdale, Arizona, USA).

A 20-camera Vicon motion analysis system (Oxford Metrics, Oxford, UK), comprised of MX and T-series cameras, recorded reflective marker trajectories at 300 Hz. Each MIMU affixed to the body was overlaid with a rigid plate attached with three reflective markers (Figure 6.1). This allowed retro-reflective technical coordinate systems (RRtech) to be created during data processing. The marker-plates were attached to the MIMUs using velcro, with the orientation of the markers aligned with the axes of the MIMU case as closely as possible. The coordinate system of each MIMU was also rotated during data processing for more precise alignment with the RRtech (explained below). This process ensured that soft tissue artefact errors for the MIMUs and reflective markers were comparable and that differences to coordinate system orientations would not impact the inter-system measurement comparison.

Two AMTI (Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA) force plates (1800 Hz) recorded front and back foot contact during the bowling delivery stride. A 300 Hz Vicon Bonita video camera (Oxford Metrics, Oxford, UK) was positioned sagittal to the bowling action. The camera was synchronised to the retro-reflective system and was used to identify the ball release frame.

Following participant preparation, each individual was instructed to perform five moderately-paced trunk ROM trials. Standing ROM trials were collected for flexion-extension and lateral flexion, before sitting ROM trials were collected for all three planes of motion (flexion-extension, lateral flexion and rotation). For each trial, participants moved to the end of their range and repeated the movement five times in both directions. Five bowling trials were then recorded, with participants instructed to bowl at match-like intensity.

6.3.3 Data Processing

The Xsens and Noraxon MIMU free acceleration and orientation data were output for each trial as a .csv file using the Xsens MT Manager software (Xsens Technologies B.V., Enschede, Netherlands) and myoRESEARCH software (Noraxon USA Inc., Scottsdale, Arizona, USA), respectively. Sensor orientations were estimated by the manufacturers’ inbuilt filtering algorithms and were expressed
in quaternions. Both MIMU systems utilised Kalman filters to estimate orientation angles, however the exact algorithms used by each system were not disclosed in the manufacturers’ documentation.

Reflective trajectories were labelled, filtered and modelled using Vicon Nexus 2.5 software (Oxford Metrics, Oxford, UK). All trajectories were filtered using a fourth-order, zero-lag, low pass Butterworth filter. Cut-off frequencies of 9 Hz and 15 Hz were selected for ROM and bowling trials respectively, following a residual analysis (Winter, 1990). Retro-reflective data were modelled by a customised Vicon Bodybuilder model (Oxford Metrics, Oxford, UK). Technical coordinate systems were defined in the same way for the thorax, lower-back and pelvis (Figure 6.1). The RRtech origin was the mean position of the three markers. The first defining line was a vector from marker-2 to marker-1, with the second defining line a vector from marker-3 to marker-2. The y-axis (vertical) was parallel to the first defining line. The x-axis was the cross-product of the first and second defining lines. The z-axis was perpendicular to the y and x axes (Figure 6.1).

![Figure 6.1](image)

**Figure 6.1**: IMU and affixed rigid retro-reflective marker plate. Markers (M1, M2, M3), defining lines (D1, D2), and coordinate system axes (X, Y, Z) are indicated. X-axis is anterio-posterior. Y-axis is vertical. Z-axis is medio-lateral.

A customised LabVIEW 2017 program (National Instruments Corp, Austin, Texas, USA) performed additional custom data processing. The program (1) temporally synchronised MIMU and retro-reflective data, (2) spatially aligned the coordinate systems of each MIMU with the accompanying RRtech, (3) calculated joint angle values, and (4) compiled and output all values for all trials. These processes are detailed below.

Both MIMU systems were temporally synchronised with the retro-reflective data for each trial. Temporal synchronisation for the Xsens ROM trials was achieved by cross-correlating the angular component of the axis-angle representation of the thorax MIMU and its accompanying RRtech. The lumbar MIMU and RRtech were used instead for the sitting flexion-extension ROM trials. For the Noraxon MIMU ROM trials, cross-correlating the orientation quaternions for the thorax MIMU and RRtech was found to be more suitable. Both the Noraxon and Xsens bowling trials were temporally
synchronised by cross-correlating the orientation quaternions. Cross-correlation was utilised because it enabled time-lag between the MIMU and retro-reflective data to be calculated and accounted for.

To spatially align the Xsens and Noraxon MIMU coordinate systems with their paired RRtech, each MIMU coordinate system was rotated -90° about the z-axis and then 90° about the y-axis. The Noraxon coordinate systems were then also rotated 180° about the z-axis, as they displayed a different coordinate system orientation to the Xsens sensors. There was also a further refinement process which utilised data from the seated trunk flexion-extension ROM trial. It involved calculating the orientation difference between the MIMU coordinate systems and RRtech orientations and then accounting for this difference. For each MIMU/RRtech pair, the orientation difference was calculated by multiplying the RRtech quaternion by the inverse of the MIMU quaternion. The difference in their orientations at each time-point was averaged over the course of the trial, with this correction value then applied to the MIMU data. Once these alignment procedures were completed, it was possible to calculate global angles by comparing the RRtech or the MIMU orientations with the retro-reflective 3D global reference frame.

Joint angles for both MIMU systems and their matching RRtechs were calculated via a ZXY-ordered Euler angle decomposition. Lateral flexion (x) and rotation (y) were negated, so that the right hand rule applied. Four different angles were calculated for each trial. Global thorax angles (thorax MIMU to global), global lumbar angles (lumbar MIMU to global), thorax-to-pelvis relative angles (thorax MIMU to pelvis MIMU), and lumbar-to-pelvis relative angles (lumbar MIMU to pelvis MIMU) were analysed across all three planes of motion separately (flexion-extension, lateral flexion, and rotation). Angles were calculated for the ROM trials and the two bowling trials of best data quality for each data collection session. The bowling trials were chosen following visual inspection of the trajectory data, with the trials showing the least reflective marker occlusion chosen for analysis.

MIMU and retro-reflective continuous angular data were time-normalised to 101 data-points for the ROM and bowling trials. The beginning and end of the ROM phase was defined by a 0° thorax-to-pelvis RRtech angle (neutral trunk position). Back foot contact and ball-release were used to define the bowling stride phase (Bartlett, 1996).

6.3.4 Data Analysis

The angular data from each MIMU system were compared with RRtech angles for each trial. Continuous angle waveforms were assessed for significant differences (p < 0.05) via one-dimensional statistical parametric mapping (SPM1D) (Pataky et al., 2013). The SPM1D analysis enabled grouped data to be analysed across an entire phase, rather than discrete time-points. Each
plane of motion was considered separately for ease of interpretability and relevance to previous clinical research. Publicly available SPM1D source code (Pataky, 2018) was incorporated into customized MATLAB scripts (Mathworks, Natick, Massachusetts, USA). The MATLAB scripts also calculated root mean square difference (RMSD) at three time-points for each trial: (1) the maximum (RMSD-max), (2) minimum (RMSD-min) and (3) the time-point coinciding with the peak t-trace value from the SPM1D analysis (RMSD-tmax). RMSD-tmax represented the largest difference between the MIMU and retro-reflective data during each phase. The RMSD-tmax measure occurred at the same time-point for both systems, whilst RMSD-max and RMSD-min could occur at differing time-points for the Noraxon and Xsens data.

SPSS statistics (IBM, Armonk, New York, USA) was used to perform an independent samples t-test which assessed whether there were significant differences ($p < 0.05$) between the Xsens and Noraxon RMSD values. The RMSD values were categorised by trial type (ROM or bowling), and also sub-categorised by angle type (e.g., thorax-to-pelvis angle) and plane of motion (e.g., sagittal/flexion-extension). This involved calculating the mean RMSD value for all of the RMSD measures in a given category (e.g., the mean RMSD for all thorax-to-pelvis angles during the ROM trials). Assessing the data in this way enabled trends to be easily identified and also reduced the volume of data presented.

Using G*Power 3.1.6 (Faul et al., 2007), a statistical power calculation was performed. The number of data-points incorporated in the t-tests varied between 30 and 240 depending on how the data was categorised. Using an effect size of 0.75 and an alpha level of 0.05, a power of 0.81 was calculated for the minimum group size (30 data-points).

### 6.4 Results

The results from the SPM1D analyses showed no significant differences for both MIMU systems (Xsens and Noraxon) compared with the retro-reflective derived angles for both movement tasks. The SPM1D figures are displayed in section 9.2: Appendix B.

When analysing all RMSD values calculated from the ROM trials, the Xsens values were significantly larger than the Noraxon values for RMSD-max (Xsens = 2.6±1.5°, Noraxon = 1.4±1.0°, $t = 2.80, p < 0.01$) and RMSD-tmax (Xsens = 2.5±1.4°, Noraxon = 1.7±0.7°, $t = 2.23, p = 0.03$). Larger differences were observed between the two systems during the bowling trials for RMSD-max (Xsens = 4.6±1.4°, Noraxon = 3.2±0.9°, $t = 2.97, p < 0.01$), RMSD-min (Xsens = 5.8±1.7°, Noraxon = 3.6±1.2°, $t = 3.69, p < 0.01$), and RMSD-tmax (Xsens = 7.4±1.9°, Noraxon = 4.0±1.3°, $t = 5.14, p < 0.01$), with Xsens mean RMSD values exceeding 5° at RMSD-min and RMSD-tmax. The mean RMSDs for the ROM and bowling trials are displayed in Figure 6.2.
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Figure 6.2: Root mean square differences (RMSD) for Xsens and Noraxon MIMU derived angles when compared with retro-reflective based angles. The mean RMSDs are presented separately for the ROM and bowling trials. * = Xsens and Noraxon RMSDs significantly different, with individual p-values indicated.

When RMSD values were analysed for each of the four angles individually (thorax global, thorax-to-pelvis, lumbar global, or lumbar-to-pelvis), Noraxon ROM thorax-to-pelvis RMSD values were found to be approximately 2° less than the Xsens values (Figure 6.3), with significant differences observed at RMSD-max (Xsens = 3.6±1.7°, Noraxon = 1.6±0.7°, t = 2.48, p = 0.04) and RMSD-tmax (Xsens = 3.5±1.7°, Noraxon = 1.6±0.7°, t = 2.29, p = 0.05). Significant differences were also found for thorax-to-pelvis angle during the bowling trials (Figure 6.3) for RMSD-max (Xsens = 5.0±1.0°, Noraxon = 3.0±0.4°, t = 3.07, p = 0.04), RMSD-min (Xsens = 6.4±1.0°, Noraxon = 4.3±0.1°, t = 3.94, p = 0.02), and RMSD-tmax (Xsens = 7.8±1.6°, Noraxon = 3.4±0.6°, t = 4.43, p = 0.01).

Figure 6.3: Root mean square differences (RMSD) for Xsens and Noraxon MIMU derived angles when compared with retro-reflective based angles. The mean Thorax-to-Pelvis RMSDs are presented separately for the ROM and bowling trials. * = Xsens and Noraxon RMSDs significantly different, with individual p-values indicated.
Finally, significant differences were found between the Noraxon and Xsens RMSD values when the discrete data were analysed separately for each plane of motion. The transverse plane (rotation) ROM RMSD-tmax values were significantly lower for the Noraxon data than the Xsens data (Xsens = 4.0±0.6°, Noraxon = 2.3±0.3°, t = 5.21, p < 0.01). Similar differences were also found for transverse plane measures during the bowling trials for RMSD-max (Xsens = 6.1±1.0°, Noraxon = 3.7±0.7°, t = 3.95, p < 0.01) and RMSD-min (Xsens = 6.5±0.8°, Noraxon = 4.0±1.5°, t = 2.93, p = 0.03). Transverse plane (rotation) RMSD measures are presented in Figure 6.4. In addition to the significant differences to rotation RMSD measures, Xsens RMSD-tmax values were also larger than those for Noraxon for sagittal plane (flexion-extension) movements during bowling trials (Xsens = 7.7±1.4°, Noraxon = 3.4±1.1°, t = 4.85, p < 0.01).

![Figure 6.4: Root mean square differences (RMSD) for Xsens and Noraxon MIMU derived angles when compared with retro-reflective based angles. The mean transverse plane (rotation) RMSDs are presented separately for the ROM and bowling trials. * = Xsens and Noraxon RMSDs significantly different, with individual p-values indicated.](image)

### 6.5 Discussion

Overall, both of the MIMU systems analysed in this study displayed acceptable validity when estimating thorax and lumbar joint kinematics during trunk uni-planar and high-speed, multi-planar motion (Bayne et al. 2016; Brice et al. 2018; Mjøsund et al., 2017). This is evident from the SPM1D analysis, where continuous data from the Xsens and Noraxon MIMU systems were found to be similar to the concurrently collected retro-reflective angle outputs. This supports the view that some of the greatest causes of measurement differences between MIMU systems and retro-reflective based modelling are dissimilarities in coordinate system orientations and soft tissue movement artefacts (Brice et al., 2018; Camomilla et al., 2018; Chardonnens et al., 2012). Aligning MIMU coordinate systems with retro-reflective coordinate systems, as was done in this study, enables an
equitable comparison between technologies, as differences to soft tissue artefacts or segment coordinate system orientations will not affect measurements.

Root mean square differences were used to measure task-specific differences in this study. Unsurprisingly, RMSD values for both MIMU systems were smaller for the slower, uni-planar trunk ROM trials than the bowling trials (Figure 6.2), which involved high-speed, multi-planar trunk movements. Increasing velocity (Lebel et al., 2013; Pasciuto et al., 2015) or movement complexity (Godwin et al., 2009) has been shown to be associated with a reduction in MIMU measurement accuracy. For both MIMU systems, the mean ROM RMSD values ranged from 1.4° to 2.6°, whereas the bowling trials produced RMSDs of 3.2° to 7.4°. It should be noted that only the Xsens system produced values greater than 5°. The ROM RMSD values of less than 3° are comparable with the findings of previous research which examined uni-planar movements of robotic gimbal tables (Lebel et al., 2013) or human participants (Mjøsund et al., 2017). Measurement differences under 5° are likely to be functionally inconsequential (Intolo et al., 2010; Mjøsund et al., 2017), which provides further support for MIMU-based measurement of trunk ROM. The bowling trial RMSD values also compare well with previous sport-specific MIMU research that has quantified trunk angles (Bergamini et al., 2013; Brice et al., 2018; Zhang et al., 2015), especially when it is considered that cricket bowling involves particularly high-speed trunk movements through large ranges of motion (Bayne et al., 2016; Burnett et al., 1998). The RMSD value differences between the ROM and bowling trials appear to be attributable to the increase in movement velocity and complexity during cricket bowling.

Noraxon MIMU RMSD values were significantly smaller ($p < 0.05$) than the Xsens MIMU RMSD values in some instances (Figure 6.2, Figure 6.3 and Figure 6.4). It should be noted that the statistically significant differences found in the ROM trials were between 0.8° and 2°, which is likely to be functionally irrelevant. The differences between the Xsens and Noraxon bowling RMSDs ranged from 1.4° to 4.4°, which could be considered to be approaching functional significance, especially when it is considered that only the Xsens system produced RMSDs above 5°.

The Noraxon thorax-to-pelvis RMSDs were significantly smaller than the Xsens thorax-to-pelvis RMSDs for five of the six discrete comparisons (Figure 6.3). Thorax-to-pelvis was the only one of the four angles assessed to show significant differences to RMSD values between the MIMU systems. This is somewhat expected, as calculating relative angles with MIMUs has been previously shown to be less accurate than calculating absolute (global) angles (Brice et al., 2018; Lebel et al., 2013). Similarly, previous work has also suggested that MIMUs are susceptible to measurement errors when estimating axial rotation angles (Bergamini et al., 2014; Brice et al., 2018; Theobald et al., 2012), which likely explains the larger Xsens MIMU RMSD values recorded for transverse plane measurements in this study (Figure 6.4).
The two MIMU systems assessed in this study had almost identical tri-axial sensor specifications: accelerometers (Noraxon: 16 g, Xsens: ±160 m/s²), gyroscopes (Noraxon: ±2000°/s, Xsens: ±2000°/s) and magnetometers (Noraxon: ±1.9 Gauss, Xsens: ±1.9 Gauss). It appears the two major differences between the MIMU systems were the sampling rate (Noraxon: 100 Hz, Xsens: 75 Hz), and probable variations to proprietary sensor fusion algorithms and parameters. We speculate that the slightly higher sampling rate for the Noraxon system may have been beneficial for the measurement of kinematics during the fast bowling trials, especially when it is considered that the entire bowling phase typically lasted for approximately a third of a second. Despite both systems employing a Kalman filter approach to sensor fusion, there were almost certainly differences to fusion algorithms, and algorithm parameters and weightings, which likely impacted resultant angle measurements. Previous research comparing MIMU models and fusion methods has illustrated the impact that varied sensor fusion approaches can have on MIMU measurement accuracy (Bergamini et al., 2014; Filippeschi et al., 2017; Ligorio et al., 2016). The current research supplements that literature and provides further support that even when MIMU systems utilise similar general filtering approaches (i.e., Kalman filtering) variance to angle measurement outputs remain likely.

A limitation of this study was that the data from both MIMU systems were compared with retro-reflective motion capture data, which cannot be considered a genuine ground-truth measure despite its reported high reliability and accuracy (van der Kruk and Reijne, 2018; Windolf et al., 2008). Subsequently, when interpreting the findings of this study, one must be aware that concurrent validity is not necessarily reflective of measurement accuracy. Also worthy of note, was that the two groups of 10 participants were not entirely the same for both MIMU systems, however the best efforts were made to match participant characteristics given the data sample available. It was also not possible for the two MIMU systems to be worn simultaneously on the same anatomical landmarks, regardless of participant groups.

The functional implications of this study are that the Xsens Mtw Awinda and the Noraxon MyoMOTION Research Pro MIMUs both appear capable of measuring concurrently valid trunk motion for the uni-planar ROM tasks tested. Overall, the angle measurements during high-speed, multi-planar bowling movements also compared well with previous research. However, the Noraxon system may be more suitable for quantification of such sport-specific movement, with the Xsens system producing some measurement differences above 5° when compared with the criterion retro-reflective derived angle outputs. The findings of this study appear to outline the impact sensor fusion algorithms and parameters can have on MIMU measurements, even when the MIMU systems utilise the same general approaches (i.e., Kalman filtering). Higher MIMU system sampling rates are also beneficial when measuring sport-specific tasks.
6.6 Conclusion

The two MIMU systems assessed in this study produced trunk angle measures that can be considered to have acceptable concurrent validity for the tasks assessed, especially for uni-planar ROM movements. Both MIMU systems produced RMSD values of less than 3° when compared against 3D retro-reflective motion capture angle data. Measurement differences during high-speed, multi-planar movements were generally greater for both systems, with one system producing mean RMSD values greater than 5° at various time-points of the bowling delivery stride, possibly indicating functionally significant measurement errors. It appears that even slight differences to commercial MIMU sensor fusion algorithms and sampling rates can have a significant impact on MIMU measurement validity. Consumers should be cognisant of the measurement variations that can occur between commercial MIMU systems when collecting and reporting data, especially for sport-specific activities.
6.7 References


Chapter 6: Comparison of Trunk Angle Measures from Two Commercial MIMU Systems


Chapter 7: MIMU-based Measurement of Trunk and Pelvis Angles during Cricket Bowling

Cottam, D. S., Campbell, A. C., Davey, P. C., Kent, P., Elliott, B. C., Bayne, H., & Alderson, J. A.

Measurement of thorax, lumbar and pelvis motion during cricket fast bowling using magneto-inertial measurement units: A comparison with three-dimensional retro-reflective anatomical modelling.

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Research Linking Statement

Chapters 5 and 6 investigated selected factors capable of influencing MIMU measurement validity. These factors included: the sensor-to-segment calibration approach (chapter 5), angle calculation method (chapter 5), movement task complexity and velocity (chapters 5 and 6), and the commercial MIMU system (chapter 6). Concurrent measurement validity was assessed in the two previous chapters by comparing MIMU measures with outputs derived from 3D retro-reflective technical marker clusters placed on the MIMUs. This process ensured that MIMU and retro-reflective segment coordinate system orientations could be easily aligned and to account for soft tissue artefact, by assuming errors would affect both modalities similarly. Theoretically, this facilitated a more equitable comparison of the two technologies. However, retro-reflective modelling typically incorporates markers placed on anatomical landmarks, an approach generally not adopted in MIMU positioning. It is acknowledged that placing sensors and markers on different segment locations means it is likely that soft tissue artefact and landmark identification errors will have varying effects on measurements. Additionally, while retro-reflective anatomically based modelling has been employed in the majority of historical cricket fast bowling biomechanical research, it is currently unknown how MIMU-derived trunk and pelvis measures compare with anatomically modelled thorax, lumbar and pelvis angles during bowling. This study aimed to investigate the nature of the relationship between MIMU and retro-reflective derived trunk and pelvis angle measures during fast bowling. The findings of chapters 5 and 6 were practically applied so the best available data collection and processing procedures were utilised. Ideally only small overall differences would be present between the retro-reflective and MIMU based trunk and pelvis angle outputs, or a systematic linear relationship between measurements would also be desirable. This would enable MIMUs to be used in the field as an ecologically valid bowling analysis tool, with their measurements directly applicable to historical injury causation data in the bowling literature. An effective field-based screening tool could potentially enable early identification of increased lumbar injury risk in fast bowlers at all levels of the sport.
7.1 Abstract

Previous fast bowling research has identified multiple lumbar injury risk factors involving thorax, lower-back and pelvis kinematics. Research has generally employed three-dimensional retro-reflective based anatomical modelling to record bowling biomechanics. Magneto-inertial measurement units (MIMUs) could provide a more accessible, portable alternative to retro-reflective motion capture methods. This study aimed to assess differences between MIMU systems and retro-reflective based anatomical modelling when quantifying thorax, lumbar and pelvis angles during bowling. It was hypothesised that MIMU measures would be significantly different to retro-reflective based anatomically modelled measures. The movements of 10 fast-medium bowlers were simultaneously recorded by MIMUs and retro-reflective motion capture. One-dimensional statistical parametric mapping (SPM1D) and root mean square differences (RMSD) were used to compare the two approaches. Results from the SPM1D analysis showed significant differences for 19 of the 21 angles. On average, the measures were significantly different for 41.4±27.7% of the bowling stride. The mean RMSD value for all variables was 13.9±6.4°. Discrepancies to segment coordinate system orientations and movement artefact errors likely contributed to measurement differences. These findings demonstrate that MIMU measurements of thorax, lumbar and pelvis kinematics during bowling should not be compared with retro-reflective data reported in the bowling literature.
7.2 Introduction

Lumbar spondylolysis and intervertebral disc degeneration are potentially debilitating conditions that often require extensive recovery periods. Spondylolysis is disproportionately common amongst cricket fast bowlers, with prevalence estimated to be 11-67% (Annear et al., 1992; Crewe et al., 2012; Elliott et al., 1992; Foster et al., 1989; Gregory et al., 2004; Hardcastle et al., 1992). This is considerably higher than the prevalence in the general population or other athletic populations, where 4-6% and 8% of people are affected respectively (Beutler et al., 2003; Fredrickson et al., 1984; Soler and Caldero, 2000). Fast bowlers also appear to be susceptible to severe and premature intervertebral disc degeneration (Burnett et al., 1996; Crewe et al., 2012; Ranson et al., 2005; Siemionow et al., 2011; Teraguchi et al., 2014). One population of professional fast bowlers (n=36) reported disc degeneration in 61% of individuals, with 33% of the population classified as having severe degeneration (Ranson et al., 2005). Another study found the presence of disc degeneration in young fast bowlers had increased from 21% to 58% over 2.7 years (Burnett et al., 1996). The severity and frequency of lumbar injuries in fast bowlers has resulted in a large proportion of cricket research concentrating on establishing aetiological mechanisms of this injury.

There is consensus that bowling biomechanical factors contribute to the high rates of lumbar injury amongst fast bowlers (Forrest et al., 2017; Johnson et al., 2012; Olivier et al., 2016; Orchard et al., 2016). The vast majority of these factors occur during the bowling delivery stride, which is the period from back foot contact to ball release. Early investigations of fast bowling biomechanics and lumbar injury concluded that excessive shoulder counter-rotation ranges (rotation of the shoulders and upper thorax away from the bowling direction) and large shoulder-pelvis separation angles (shoulders-to-pelvis relative transverse plane angle) early in the bowling delivery stride were potentially hazardous for the lower back (Annear et al., 1992; Elliott et al., 1992, 1993; Foster et al., 1989; Hardcastle et al., 1992). These early studies were limited by the motion capture technologies available at the time, which made it difficult to directly quantify spinal kinematics, leading researchers to explicitly focus on thorax and pelvis motion. More recent studies have proposed other kinematic lumbar injury risk factors, including large thorax or trunk lateral flexion ranges (Bayne et al., 2016; Phillips et al., 2010; Ranson et al., 2008; Stuelcken et al., 2010) and excessive pelvis rotation (Bayne et al., 2016; Elliott, 2000; Glazier, 2010; Ranson et al., 2008). Only a few studies have directly associated lumbar biomechanics during bowling with injury risk. The initial study on lumbar kinematics utilised a wearable electromagnetic device and suggested that large lumbar flexion-extension and lateral flexion angles during the bowling delivery stride may be variables of interest (Burnett et al., 1998). Bayne and colleagues (2016) modelled lumbar kinematics and kinetics, and reported that bowlers who sustained a lumbar injury exhibited greater
peak flexion-extension and lateral flexion moments during the delivery stride, supporting previously stated hypotheses in the literature (Ferdinands et al., 2009).

The vast majority of the bowling studies mentioned employed either two-dimensional (2D) image-based motion capture systems (e.g., high speed film or digital video) or three-dimensional (3D) retro-reflective motion capture systems, with the research by Burnett and colleagues (1998) the obvious exception. Video based digitisation is an affordable and relatively portable motion analysis solution but it has limited measurement validity for high-speed, multi-planar movements, such as those exhibited during cricket bowling (Elliott and Alderson, 2007; Elliott et al., 2007). Contributing to these limitations are perspective errors, insufficient frame rates, occlusion issues, and simplistic 2D vector-based representations of body segments and joints. Conversely, 3D retro-reflective motion capture is reported to have a sub-millimetre measurement accuracy (van der Kruk and Reijne, 2018; Windolf et al., 2008). Retro-reflective systems are also capable of capturing multi-planar motion in 3D, which is a major limitation of 2D video motion analysis approaches. Retro-reflective based analysis involves the placement of reflective markers on anatomical landmarks or body segments, with the modelling of marker trajectories facilitating anatomically specific kinematic calculations. Solutions such as the calibrated anatomical systems technique (Cappozzo et al., 1995) and functional joint calibrations (Besier et al., 2003) have further improved the measurement validity of retro-reflective based modelling by mitigating skin movement artefacts over joints and joint centre misidentification errors. However, retro-reflective motion capture systems are expensive and require specific training to be utilised effectively. They are usually confined to indoor laboratories, lending the analysis approach to criticisms of poor ecological validity. Restricted capture volumes, lack of portability, marker occlusion issues, and sensitivity to ambient light conditions renders retro-reflective systems largely unsuitable for outdoor, field-based motion capture. Additionally, whilst 2D video is commonly used as a qualitative coaching tool within the cricket community (Elliott and Khangure, 2002; Ranson et al., 2009; Wallis et al., 2002), 3D retro-reflective motion analysis is not available to the vast majority of cricketers, except those at the elite levels. There is clearly a requirement within the cricketing community for a motion capture technology that provides valid measurements whilst being affordable and suitable for field-use.

Magneto-inertial measurement units (MIMUs) may be a solution to the bowling biomechanical analysis dilemma. A MIMU incorporates tri-axial accelerometers, gyroscopes and magnetometers. Data from the sensors can be fused by algorithms, such as Kalman filters (Kalman, 1960), which enables the 3D orientation of the unit to be computed. Placing MIMUs on body segments facilitates measurement of segment orientations. Three-dimensional joint angles can also be estimated by introducing further calibration and computation processes. Importantly, MIMUs are considerably
more portable, affordable and user-friendly than 3D retro-reflective motion capture systems. Unlike 3D and 2D camera-based systems, they are also not affected by occlusion issues or restrictive capture volumes.

It is important to recognise that individual MIMUs only have the capacity to measure the orientation of the unit they are housed within, hence they can only be expected to reflect the movement of the segment position they are affixed to. Conversely, retro-reflective based anatomical modelling utilises multiple markers placed on anatomical landmarks or body segments.

To allow an unbiased comparison between systems, previous research has placed retro-reflective markers on MIMUs, which allows retro-reflective technical coordinate systems to be created and then aligned with the MIMU coordinate systems. When this step is undertaken, MIMUs have shown the capability of validly measuring sport-specific trunk motion (Bergamini et al., 2013).

However, different coordinate system orientations will likely have an effect on kinematic measurements. It should also be expected that movement artefacts will affect the systems disparately if MIMUs are attached to the body at different locations to reflective markers (Brice et al., 2018; Camomilla et al., 2018; Wong and Wong, 2008). Recent research has documented large discrepancies between MIMU and retro-reflective based measures of shoulder-pelvis separation angle during discus throwing (Brice et al., 2018). The researchers chose not to align the MIMU and retro-reflective coordinate systems. Senington and colleagues (2018) recently utilised MIMUs in a bowling study which examined shoulder counter-rotation and shoulder-pelvis separation and their association with lumbar kinematics. However, the authors did not compare the MIMU measures with retro-reflective based data. Dissimilarities between MIMU and retro-reflective based systems means it is possible that their measurements may not be applicable to prior bowling research that employed 3D retro-reflective motion capture or 2D video based motion capture. Understanding the exact nature of any differences may allow a bridging of the gap between past research and the results of future analyses based on MIMU data.

Despite their apparent suitability for field-based motion capture, it is currently unknown whether MIMUs can be used as a substitute for retro-reflective anatomical modelling when assessing bowling kinematics associated with lumbar injury. This study aimed to understand the nature and magnitude of the measurement differences between MIMUs and retro-reflective based anatomical modelling when assessing thorax, lumbar and pelvis kinematics during cricket fast bowling. Focus was given to variables previously linked with lumbar injury: shoulder counter-rotation, shoulder-pelvis separation, thorax lateral flexion, pelvis rotation, lumbar flexion-extension, and lumbar lateral flexion. It was hypothesised that MIMU based measures of thorax, lumbar and pelvis kinematics during fast bowling would be significantly different to measures obtained via 3D retro-reflective based anatomical modelling.
Chapter 7: MIMU-based Measurement of Trunk and Pelvis Angles during Cricket Bowling

7.3 Methods

7.3.1 Participants

Ten male district or community level fast/medium bowlers (184.5±5.3 cm, 80±9.3 kg, 21.6±3.9 years) participated in this criterion-based validity study. All participants provided informed consent prior to commencing, with a parent/guardian also providing consent for participants under the age of 18. The University of Western Australia Human Research Ethics Committee (RA/4/1/2593) and the Curtin University Ethics Office (HRE2016-0472) provided ethical approval for the research.

7.3.2 Data Collection

Magneto-inertial data and 3D retro-reflective trajectory data were collected concurrently in the indoor Motion Analysis Laboratory at Curtin University, Western Australia. The Noraxon MyoMOTION Research Pro MIMU system (Noraxon USA Inc., Scottsdale, Arizona, USA) (100 Hz sampling frequency, ±2000°/s gyroscope, ±16 g accelerometer, ±1.9 Gauss magnetometer) was used to capture MIMU data. Three MIMUs were placed on the thorax, lumbar spine and pelvis of each participant prior to data collection (Figure 7.1). The thorax MIMU was placed with its cranial edge between the spinous processes of the seventh cervical (C7) and first thoracic (T1) vertebrae. The lumbar MIMU was positioned over the spinous process of the first lumbar vertebra (L1). The caudal edge of the pelvis MIMU was placed on the spinous process of the second sacral vertebra (S2). The MIMU data were captured by MyoRESEARCH 3.8.2 software (Noraxon USA Inc., Scottsdale, Arizona, USA).

Reflective marker trajectories were recorded at 300 Hz by a 20-camera Vicon motion analysis system (Oxford Metrics, Oxford, UK), consisting of MX and T-series cameras. Reflective markers were affixed to the pelvis, trunk and shoulders of each participant (Figure 7.1). Pelvis markers were placed bilaterally on the anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS) and iliac crest. Lumbar markers were positioned as per Crewe and colleagues (2013). They were placed on the spinous processes of the third (L3) and fifth lumbar (L5) vertebrae. Left and right lower lumbar markers were also placed either side of the fourth lumbar vertebra, approximately midway between the spinous process and the most lateral aspect of the trunk. Thorax markers were placed on the spinous process of the tenth thoracic vertebra (T10), the xiphoid process and jugular notch of the sternum. Shoulder and upper arm markers were positioned as per Campbell and colleagues (2009). Bilateral acromion marker triads affixed to semi-rigid plates were positioned with the plate’s lateral edge overlaying the lateral ridge of the acromion. Single markers were positioned on the posterior and anterior shoulder, in plane with the visually estimated glenohumeral joint centre. Triads attached to T-shaped semi-rigid plates, were also placed on both
arms, with the long bar of the plate parallel to the humerus and the short bar extending towards the lateral head of the biceps. Finally, marker triads were attached to rigid plates which overlaid each of the MIMUs. Whilst these triads are not the focus of this study, they were involved in the alignment of the global reference frames of the retro-reflective and MIMU systems during post-processing. The technical cluster marker positions were also used to create virtual markers during data processing which were then incorporated in body segment definitions. The triad on the lumbar MIMU was used to create a marker representing the first lumbar vertebra (L1) by calculating the mid-point between two of the triad markers. A similar process was employed to estimate the position of the seventh cervical vertebra (C7) from the thorax MIMU triad.

![Image of retro-reflective marker set affixed to the shoulders, trunk and pelvis. Location of MIMUs and overlaying marker technical clusters also indicated in blue.](image)

Back foot and front foot contact of the bowling delivery stride were recorded at 1800 Hz by two AMTI force plates (Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA). A 300 Hz Vicon Bonita video camera (Oxford Metrics, Oxford, UK) was utilised for identification of the ball release frame. It was positioned sagittal to the bowling motion and synchronised with the retro-reflective system. Synthetic turf and wickets were used to liken the laboratory to a cricket pitch.

Following preparation, participants were instructed to complete a series of different movement tasks, culminating in five bowling trials. Bowlers were requested to bowl at match-like intensity. The two bowling trials of best data quality for each participant were chosen for further analysis. This was determined by visual inspection of the retro-reflective data, with trials showing minimal marker occlusion chosen.
7.3.3 Data Processing

Each MIMU’s free acceleration and orientation data were output in a .csv file for each trial via the MyoRESEARCH software (Noraxon USA Inc., Scottsdale, Arizona, USA). Sensor orientations were calculated by the manufacturer’s inbuilt Kalman filters and were expressed in quaternions.

Retro-reflective data were processed using Vicon Nexus 2.5 software (Oxford Metrics, Oxford, UK). This involved labelling, filtering and modelling of marker trajectories. Markers were filtered by a fourth-order, zero-lag, low pass Butterworth filter, with a cut-off frequency of 15 Hz chosen after a residual analysis (Winter, 1990). Data was modelled by a customised Vicon Bodybuilder model (Oxford Metrics, Oxford, UK). Shoulder joint centres were defined as per Campbell and colleagues (2009). Body segment origins, defining lines and axis orders are presented in Table 7.1. The first axis listed for each segment was defined by a vector parallel to the first defining line. The second axis was perpendicular to the first and second defining lines. The third axis was perpendicular to the first and second axes. X-axes were antero-posterior, y-axes were vertical (axial rotation), and z-axes were medio-lateral. Joint angles were defined by employing the Grood and Suntay (1983) joint coordinate system. Global angles were calculated for the upper thorax/shoulders (henceforth referred to as ‘shoulder’), thorax, lumbar, and pelvis segments. Angles were also calculated for the shoulder, thorax and lumbar segments relative to the pelvis.

Table 7.1: Segment origins, defining lines and axis orders used to define body segments during retro-reflective modelling.

<table>
<thead>
<tr>
<th>Body Segment</th>
<th>Segment Origin</th>
<th>Defining Line 1</th>
<th>Defining Line 2</th>
<th>Axis Order</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>Mid-point of C7 and jugular notch</td>
<td>Mid-point of T10 and xiphoid process → Segment origin</td>
<td>C7 → Jugular notch</td>
<td>YZX</td>
</tr>
<tr>
<td>Upper Thorax/Shoulders</td>
<td>Mid-point of right and left shoulder joint centres</td>
<td>Left shoulder joint centre → Right shoulder joint centre</td>
<td>Jugular notch → Segment origin</td>
<td>ZYX</td>
</tr>
<tr>
<td>Lumbar</td>
<td>L5</td>
<td>L5 → L1</td>
<td>Right lower lumbar → Left lower lumbar</td>
<td>YXZ</td>
</tr>
<tr>
<td>Pelvis</td>
<td>Mid-point of Right and Left ASIS</td>
<td>Left ASIS → Right ASIS</td>
<td>Segment origin → Mid-point of Right and Left PSIS</td>
<td>ZYX</td>
</tr>
</tbody>
</table>

A customised LabVIEW 2017 program (National Instruments Corp, Austin, Texas, USA) was utilised for further processing and outputting of MIMU and retro-reflective data. The program temporally synchronised the data from both systems, calculated MIMU joint angles, and compiled and output all data for each trial.

An important step was aligning the global reference frames of the retro-reflective and MIMU systems, so that global angle measurements could be properly compared. This was done by utilising the marker triads positioned on top of the MIMUs. Each MIMU was first spatially aligned.
with a coordinate system created from its paired marker triad. This was achieved by rotating each MIMU coordinate system -90° about the z-axis, 90° about the y-axis and finally 180° about the z-axis. A further refinement process involved calculating the orientation difference between the MIMU coordinate systems and retro-reflective coordinate system orientations and then accounting for this difference. The orientation difference was calculated by multiplying the retro-reflective coordinate system quaternion by the inverse of the MIMU quaternion. The difference in their orientations at each time-point was averaged over the course of the trial, with this correction value then applied to the MIMU data. Once these alignment procedures were completed, it was possible to calculate global angles by comparing the RRtech or the MIMU orientations with the retro-reflective 3D global reference frame.

The MIMU data were temporally synchronised with the retro-reflective data by cross-correlating the orientation quaternions for the thorax MIMU and the overlaying technical coordinate systems. This enabled time-lag between the two systems to be calculated and accounted for. The MIMU joint angles were calculated by employing a ZXY-order Euler angle rotation. Lateral flexion (x) and rotation (y) were negated so that the right hand rule applied. Global angles were calculated for each of the three MIMUs (thorax, lumbar and pelvis), with thorax-to-pelvis and lumbar-to-pelvis relative angles also computed. The data from both motion capture systems were time-normalised to 101 data-points. Back foot contact and ball-release were used to define the bowling stride phase.

7.3.4 Data Analysis

Retro-reflective anatomical (RRanat) based angles were compared with the equivalent MIMU derived angles (Table 7.2) using a time-dependent, one-dimensional analysis approach and zero-dimensional analysis. Key variables that had been previously linked to lumbar injury were of particular interest (Table 7.2). One-dimensional statistical parametric mapping (SPM1D) (Pataky et al., 2013) was used to assess for significant differences (p < 0.05) between the MIMU and retro-reflective based measures during the entire bowling delivery stride phase (back foot contact to ball release). Publicly available SPM1D source code (Pataky, 2018) was incorporated into customized MATLAB scripts (Mathworks, Natick, Massachusetts, USA). The SPM1D approach is based on random field theory, and enables multiple time-normalised waveforms to be compared in a time-sensitive manner. Each comparison produces a critical t-statistic (threshold), the value of which is determined by the ‘temporal smoothness’ of the data and a predefined alpha value (i.e., α = 0.05). A time-varying t-statistic (t-trace) is then calculated at each time-point during the phase. The t-trace exceeding the critical t-threshold is indicative of a significant difference between the two sets of data. A supra-threshold period is accompanied by a p-value and an ‘extent’. An extent
represents the duration of a supra-threshold period and is usually presented as a percentage of the movement phase (Pataky et al., 2013; Wells et al., 2017). Multiple supra-threshold periods (extents) can be present for each movement phase.

Root mean square differences (RMSD) were also calculated at three discrete points for each trial. The maximum (RMSD-max) and minimum values (RMSD-min) were compared for both systems, with a third value termed RMSD-tmax also computed. The RMSD-tmax value was calculated at the time-point that coincided with the largest t-trace value (largest difference), as determined by the SPM1D analysis. RMSD-max and RMSD-min values could occur at different time-points for the MIMU and RRanat measures, whilst RMSD-tmax occurred at the same time-point. The RMSD analyses were incorporated into a customised MATLAB script (Mathworks, Natick, Massachusetts, USA).

Table 7.2: Retro-reflective based angles and MIMU based equivalents. Key variables that have been previously associated with fast bowler lumbar injury are also listed along with functional interpretation of the variables. Angles were calculated for all three planes of motion (flexion-extension, lateral flexion and axial rotation).

<table>
<thead>
<tr>
<th>Retro-reflective Derived Angle</th>
<th>MIMU Derived Angle</th>
<th>Lumbar Injury Risk Variables</th>
<th>Functional Interpretation of Lumbar Injury Risk Variables</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder global angle</td>
<td>Thorax MIMU global angle</td>
<td>Global shoulder rotation</td>
<td>Global rotation (alignment) of the shoulders (transverse plane)</td>
</tr>
<tr>
<td>Shoulder-to-pelvis relative angle</td>
<td>Thorax MIMU to pelvis MIMU relative angle</td>
<td>Shoulder-to-pelvis separation angle</td>
<td>Rotation of the shoulders relative to the pelvis (transverse plane)</td>
</tr>
<tr>
<td>Thorax global angle</td>
<td>Thorax MIMU global angle</td>
<td>Thorax global lateral flexion</td>
<td>Global lateral flexion of the thorax (frontal plane)</td>
</tr>
<tr>
<td>Thorax-to-pelvis relative angle</td>
<td>Thorax MIMU to pelvis MIMU relative angle</td>
<td>Thorax-to-pelvis lateral flexion</td>
<td>Lateral flexion of the thorax relative to the pelvis (frontal plane)</td>
</tr>
<tr>
<td>Lumbar global angle</td>
<td>Lumbar MIMU global angle</td>
<td>Lumbar global flexion-extension, Lumbar global lateral flexion</td>
<td>Global flexion-extension/lateral flexion of the lumbar segment (sagittal/frontal planes)</td>
</tr>
<tr>
<td>Lumbar-to-pelvis relative angle</td>
<td>Lumbar MIMU to pelvis MIMU relative angle</td>
<td>Lumbar-to-pelvis flexion-extension, Lumbar-to-pelvis lateral flexion</td>
<td>Flexion-extension/lateral flexion of the lumbar segment relative to the pelvis (sagittal/frontal planes)</td>
</tr>
<tr>
<td>Pelvis global angle</td>
<td>Pelvis MIMU global angle</td>
<td>Pelvis global rotation</td>
<td>Global rotation of the pelvis (transverse plane)</td>
</tr>
</tbody>
</table>

7.4 Results

The SPM1D analysis revealed that there were periods of significant differences ($p < 0.05$) between the MIMU and RRanat derived angles for almost all of the variables assessed. Table 7.3 summarises these differences. Figures 7.2 to 7.8 display waveforms and SPM1D visual representations for the seven 3D angle comparisons of the RRanat and MIMU derived data. Only two of the 21 variables showed no significant differences during the delivery stride. The average percentage of the bowling stride that was significantly different between the two outputs was 41.4±27.7%. The two variables
to produce no significant differences were shoulder global rotation (transverse plane) (Figure 7.2) and global pelvis rotation (Figure 7.8), which have both been associated with lumbar injury (Table 7.2).

**Table 7.3: A summary of the statistical parametric mapping extents of significant difference between the MIMU and retro-reflective based angle measures, expressed as percentages of the bowling delivery stride. Two values for a variable indicates two separate supra-threshold periods during the phase.**

<table>
<thead>
<tr>
<th>Angle (°)</th>
<th>Sagittal</th>
<th>Plane of Motion</th>
<th>Transverse</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder global angle</td>
<td>10%*</td>
<td>49%**</td>
<td>0%</td>
</tr>
<tr>
<td>Shoulder-to-pelvis relative angle</td>
<td>16%*</td>
<td>40%<strong>, 42%</strong></td>
<td>17%*</td>
</tr>
<tr>
<td>Thorax global angle</td>
<td>39%**, 19%*</td>
<td>100%**</td>
<td>14%*</td>
</tr>
<tr>
<td>Thorax-to-pelvis relative angle</td>
<td>46%**, 6%*</td>
<td>47%<strong>, 36%</strong></td>
<td>17%**</td>
</tr>
<tr>
<td>Lumbar global angle</td>
<td>3%*, 26%**</td>
<td>5%*</td>
<td>13%*</td>
</tr>
<tr>
<td>Lumbar-to-pelvis relative angle</td>
<td>35%<strong>, 27%</strong></td>
<td>44%**</td>
<td>51%**</td>
</tr>
<tr>
<td>Pelvis global angle</td>
<td>40%**</td>
<td>6%*, 36%**</td>
<td>0%</td>
</tr>
</tbody>
</table>

* = Significant at $p < 0.05$, ** = Significant at $p < 0.001$

Among the remaining injury risk factors, shoulder-to-pelvis separation angle (rotation) (extent = 17%, $p = 0.01$) (Figure 7.3) and lumbar global lateral flexion (extent = 5%, $p = 0.04$) (Figure 7.6) showed relatively short periods of significant difference but lateral flexion measures for the global thorax (extent = 100%, $p < 0.001$) (Figure 7.4), thorax-to-pelvis (extent 1 = 36%, $p < 0.001$; extent 2 = 47%, $p < 0.001$) (Figure 7.5) and lumbar-to-pelvis (extent = 44%, $p < 0.001$) (Figure 7.7) were all significantly different for large portions of the bowling stride. Likewise, lumbar global flexion-extension (extent 1 = 26%, $p = 0.001$; extent 2 = 3%, $p < 0.046$) (Figure 7.6) and lumbar-to-pelvis flexion-extension measures (extent 1 = 35%, $p = 0.001$; extent 2 = 27%, $p = 0.005$) (Figure 7.7) were significantly different for long periods throughout the delivery stride phase.
Figure 7.2: Angle waveforms and SPM1D analysis for shoulder global angle. Mean and standard deviation plots are shown on the left for antero-posterior tilt (sagittal plane), lateral tilt (frontal plane), and rotation (transverse plane), with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures, as well as extents and p-values for periods of significant difference.

Evaluation of all the SPM1D analyses revealed there were generally longer periods of difference for relative angles than global angles. Combined extents greater than 50% of the phase were present for shoulder-to-pelvis lateral tilt (frontal plane) (extent 1 = 42%, p < 0.001; extent 2 = 40%, p < 0.001) (Figure 7.3), thorax-to-pelvis flexion-extension (extent = 46%, p < 0.001; extent = 6%, p = 0.04) (Figure 7.5), thorax-to-pelvis lateral flexion (extent = 36%, p < 0.001; extent = 47%, p < 0.001) (Figure 7.5), lumbar-to-pelvis flexion-extension (extent = 35%, p = 0.001; extent = 27%, p = 0.001) (Figure 7.7), and lumbar-to-pelvis rotation (extent = 51%, p < 0.001) (Figure 7.7).
Figure 7.3: Angle waveforms and SPM1D analysis for shoulder-to-pelvis relative angle. Mean and standard deviation plots are shown on the left for anterio-posterior tilt (sagittal plane), lateral tilt (frontal plane), and rotation (transverse plane), with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures, as well as extents and p-values for periods of significant difference.

Contrary to this trend were the thorax global measures (Figure 7.4), particularly for flexion-extension (extent = 39%, $p = 0.002$; extent = 19%, $p = 0.02$) and lateral flexion (extent = 100%, $p < 0.001$) measurements. Shoulder (Figure 7.2), lumbar (Figure 7.6) and pelvis global angles (Figure 7.8) had periods of significant difference in most instances, however they did not display cumulative extents greater than 50% for any measures.
Figure 7.4: Angle waveforms and SPM1D analysis for thorax global angle. Mean and standard deviation plots are shown on the left for flexion-extension (sagittal plane), lateral flexion (frontal plane), and rotation (transverse plane), with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures, as well as extents and p-values for periods of significant difference.

Inspecting the SPM1D analyses with a focus on plane of movement reveals that transverse plane (rotation) measures showed smaller SPM1D extent values than the other two planes of movement for four of the seven variables assessed. Additionally, six of the seven transverse plane analyses displayed extents of less than 18% of the phase. The one exception was lumbar-to-pelvis rotation (extent = 51%, p < 0.001) (Figure 7.7).
Figure 7.5: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative angle. Mean and standard deviation plots are shown on the left for flexion-extension (sagittal plane), lateral flexion (frontal plane), and rotation (transverse plane), with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures, as well as extents and p-values for periods of significant difference.

In contrast, MIMU based measures of coronal plane movements (lateral flexion or pelvis obliquity) were significantly different to the RRanat based measures for at least 44% of the delivery stride, with lumbar global lateral flexion (extent = 5%, p = 0.04) (Figure 7.6) the only exception. Similarly, sagittal plane (flexion-extension or anterio-posterior tilt) angle waveforms calculated by the MIMUs were significantly different to the RRanat based measures for more than 28% of the phase, with the two exceptions being shoulder global anterio-posterior tilt (sagittal plane) (extent = 10%, p = 0.04) (Figure 7.2) and shoulder-to-pelvis anterio-posterior tilt (sagittal plane) (extent = 16%, p = 0.01) (Figure 7.3).
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Figure 7.6: Angle waveforms and SPM1D analysis for lumbar global angle. Mean and standard deviation plots are shown on the left for flexion-extension (sagittal plane), lateral flexion (frontal plane), and rotation (transverse plane), with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures, as well as extents and p-values for periods of significant difference.

The RMSD-max, RMSD-min and RMSD-tmax values for all measures are shown in Figure 7.9. The RMSD values ranged from 3.3° to 28.2°, with an average RMSD value of 13.9±6.4°. Only four of the 63 RMSD values were less than 5°. In agreement with the SPM1D analysis, the average RMSD values for transverse plane movements (11.2±5.5°) were smaller than for sagittal (15.4±6.0°) or coronal plane movements (15.1±7.0°).
Figure 7.7: Angle waveforms and SPM1D analysis for lumbar-to-pelvis angle. Mean and standard deviation plots are shown on the left for flexion-extension (sagittal plane), lateral flexion (frontal plane), and rotation (transverse plane), with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures, as well as extents and p-values for periods of significant difference.

The RMSD values for thorax global angles were greater than 20° for five of nine measures, reflecting the SPM1D results. No other angle assessed displayed more than two RMSD values greater than 20°. Conversely, RMSD values for lumbar global angles were less than 10° on six occasions, with pelvis global RMSD values less than 10° for five discrete measures. When focusing on variables that have been previously linked with lumbar injury (Table 7.2), pelvis global rotation clearly showed the smallest differences between the RRanat and MIMU based measures (RMSD-max = 5.0°, RMSD-min = 4.2°, RMSD-tmax = 6.4°). Lumbar global lateral flexion values were also
less than 10° for all three RMSD discrete measures (RMSD-max = 5.7°, RMSD-min = 8.7°, RMSD-tmax = 9.9°).

Figure 7.8: Angle waveforms and SPM1D analysis for pelvis global angles. Mean and standard deviation plots are shown on the left for anterio-posterior tilt (sagittal plane), obliquity (frontal plane), and rotation (transverse plane), with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures, as well as extents and p-values for periods of significant difference.

Thorax global lateral flexion (RMSD-max = 22.4°, RMSD-min = 12.4°, RMSD-tmax = 24.8°) and thorax-to-pelvis lateral flexion (RMSD-max = 11.7°, RMSD-min = 13.9°, RMSD-tmax = 25.3°) were the injury risk factors to produce particularly large RMSD values. Contrary to the SPM1D analysis, considerable RMSD values were also calculated for shoulder global rotation (transverse plane) (RMSD-max = 10.3°, RMSD-min = 10.5°, RMSD-tmax = 15.2°). The remaining injury risk variables produced comparable RMSD values: lumbar global flexion-extension (RMSD-max = 9.7°, RMSD-min
= 8.2°, RMSD-tmax = 17.7°), lumbar-to-pelvis flexion-extension (RMSD-max = 9.5°, RMSD-min = 15.8°, RMSD-tmax = 15.5°), lumbar-to-pelvis lateral flexion (RMSD-max = 7.8°, RMSD-min = 10.7°, RMSD-tmax = 18.3°), and shoulders-to-pelvis rotation (RMSD-max = 9.5°, RMSD-min = 7.2°, RMSD-tmax = 14.1°).

Figure 7.9: Root mean square differences (RMSD) between RRanat and MIMU based angle measures. Values for all variables and planes of motion are presented.

7.5 Discussion

The overarching finding of this study is that MIMU based estimations of thorax, lumbar and pelvis kinematics during cricket bowling are not sufficiently representative of 3D retro-reflective based,
anatomically modelled (RRanat) measures. This was evident from the SPM1D analysis (Figures 7.2-7.8) where 19 of the 21 variables displayed significant differences during the bowling delivery stride (back foot contact to ball release). These differences persisted for an average of 41.4±27.7% of the delivery stride, with SPM1D analysis clearly showing that measurement differences were not consistent, as no evidence of systematic linear measurement offset was observed. The discrete RMSD data also supported these general findings, with 59 of 63 values above 5° (Figure 7.9). Measurement differences greater than 5° are likely functionally meaningful (Intolo et al., 2010; Mjøsund et al., 2017). Nine of the variables assessed were previously linked with lumbar injury risk in fast bowlers (Table 7.2). Of these variables, MIMUs appear capable of producing similar angles to RRanat for pelvis global rotation. The SPM1D analysis (Figure 7.8) produced no significant differences for that variable, with the RMSD values the smallest of any movement assessed (RMSD-max = 5.0°, RMSD-min = 4.2°, RMSD-tmax = 6.4°). Comparable RMSD values (6.4°) have been reported for pelvis rotation measurements in a MIMU based study of discus throwing (Brice et al., 2018). The shoulders global rotation SPM1D analysis (Figure 7.2) also showed similar findings during the delivery stride but the associated RMSD values were not indicative of good within-trial agreement (RMSD-max = 10.3°, RMSD-min = 10.5°, RMSD-tmax = 15.2°). The SPM1D and RMSD analyses suggest that MIMU based measurements of shoulders rotation, shoulders-to-pelvis rotation, thorax global lateral flexion, thorax-to-pelvis lateral flexion, lumbar global flexion-extension, lumbar global lateral flexion, lumbar-to-pelvis flexion-extension, and lumbar-to-pelvis lateral flexion are not adequately reflective of RRanat based measures. Consequently, MIMU based measures of these variables should not be compared with previous retro-reflective based bowling research, particularly when considered as an indicator of lumbar injury risk.

Aligning the coordinate systems of MIMUs with retro-reflective technical coordinate systems appears to be crucial for an unbiased assessment of MIMU concurrent measurement validity. Differing coordinate system orientations serve to escalate measurement differences between MIMU and retro-reflective based approaches. Placing reflective markers on different locations to MIMUs introduces inconsistencies in soft tissue movement artefacts, which also increases measurement differences between the systems (Brice et al., 2018; Camomilla et al., 2018; Wong and Wong, 2008), a position supported by the findings of the present study along with previous MIMU studies investigating trunk and pelvis motion. This study produced average RMSD values of 13.9±6.4° across all variables. These figures can be compared with RMSD values reported by a MIMU study of transverse plane kinematics during discus throwing (Brice et al., 2018), which documented RMSDs of 6.3° and 9.3° for pelvis and torso orientation respectively. Larger RMSD figures were calculated for shoulders-pelvis separation angles (9.5° to the left, 12.5° to the right). The study authors stated that their decision to not align MIMU and retro-reflective coordinate systems likely contributed to the functionally significant measurement differences. A MIMU study...
measuring trunk inclination during sprint starts (Bergamini et al., 2013) did align MIMU and retro-reflective coordinate systems and reported smaller RMSD values of 3±2°, illustrating the impact aligned coordinate systems can have on measurement concurrent validity between MIMUs and retro-reflective systems. The findings of the mentioned papers and the current study suggest that MIMU based modelling is a valid portable alternative to 3D retro-reflective based anatomical modelling, however measurements between the two motion analysis modalities are often significantly different and should not be directly compared.

A trend of the comparative data presented in this study was that global angle measurement differences were generally smaller than differences for relative angle measures. This was particularly evident following SPM1D analysis, with global measures typically displaying shorter periods of measurement differences during the bowling delivery stride. Previous studies have provided an explanation for this phenomenon by stating that measurement errors tend to increase when multiple MIMUs are required to define the same global reference coordinate system (compounding differences); a process required for measurement of relative joint angles using MIMUs (Lebel et al., 2013; Picerno et al., 2011).

Thorax global angles were the major exception to the trend outlined above, with large SPM1D extents found, especially for flexion-extension (extent = 39%, \( p = 0.002 \); extent = 19%, \( p = 0.02 \)) and lateral flexion (extent = 100%, \( p < 0.001 \)). Five of nine RMSD values for thorax global measurements were also larger than 20°. The y-axis of the RRanat segment coordinate system for the thorax was created by a vector from the midpoint of the xiphoid process and T10 to the midpoint of C7 and jugular notch. This vector clearly crosses many more vertebral levels than the thorax MIMU (positioned on C7-T1), which may have contributed to the large measurement differences between the two systems. The assumption employed when biomechanically modelling the human body is that segments are rigid entities. This is inherently maintained for MIMU based measurements because the MIMU itself is rigid. However, the assumption is clearly not realistic for retro-reflective based anatomical modelling. This is especially pertinent to the thorax segment in this study which incorporated 10 individual intervertebral joints. The skeletal non-rigidity of the thorax is emphasised by cadaveric studies which have calculated average thoracic spine range of motion between T1 and T12 at 28° for flexion-extension, 36° for lateral flexion, and 45° for rotation (Borkowski et al., 2016). Previous research has also shown that the trunk is highly susceptible to deformation during high-velocity movements (Kudo et al., 2017). The thorax segment was likely the least rigid body segment examined in this study, especially in regards to its skeletal structure. We speculate that this was a contributing factor to the large thorax angle measurement differences between the MIMU and retro-reflective system outputs.
Another noteworthy finding of this study was that RRanat and MIMU measurements were generally most similar for transverse plane movements (11.2±5.5°), rather than movements occurring in the sagittal (15.4±6.0°) or coronal planes (15.1±7.0°). The SPM1D analysis also showed that six of the seven transverse plane measures had extents of less than 18% of the bowling delivery stride. These findings somewhat contradict previous MIMU-based gait research, where the largest measurement errors were reported in the transverse plane (Bergamini et al., 2014; Picerno et al., 2008). Rotation movement ranges were smaller than flexion-extension or lateral flexion movement ranges for many joints or segments, which may help to explain these differences. It should also be highlighted that the two measurements that showed no significant differences in the SPM1D analysis were shoulders global rotation and pelvis global rotation. The RRanat segment coordinate system definitions for the pelvis and upper thorax (Table 7.1) were less likely to be affected by the non-rigidity of the spinal column. It is likely that the intervertebral movement within the thorax and lumbar segments may have caused the RRanat based measures to further diverge from the MIMU based measures.

Finally, it should be noted that the MIIMU system utilised in this study had a sampling frequency of 100 Hz, whilst the retro-reflective motion capture system captured at 300Hz. It is probable that the discrepancies in capture frequencies also contributed to measurement differences.

The functional implications of this study are that MIMU based measures of thorax, lumbar, and pelvis kinematics during fast bowling should not be compared with results from previous 3D retro-reflective based bowling research. There does not appear to be a systematic linear relationship between the measurements obtained from the two motion capture modalities. If MIMU-specific injury risk thresholds can be developed for established lumbar injury risk factors then it may be possible for MIMUs to function as a field-based injury risk screening tool for fast bowlers. Future research should progress towards this objective.

### 7.6 Conclusion

Employing MIMUs to measure thorax, lumbar and pelvis kinematics during cricket fast bowling is not equivalent to utilising 3D retro-reflective based anatomical modelling. Discrete and continuous analysis of kinematic variables during the bowling delivery stride showed significant differences between MIMU and retro-reflective based measures, with no clear systematic linear relationship evident. These differences do not seem to be indicative of MIMU system measurement error, with divergent segment coordinate system orientations, soft tissue artefact errors and the non-rigidity of body segments likely to be the primary contributors to measurement discrepancies between the systems. Future work should focus on developing MIMU-specific lumbar injury risk kinematic...
thresholds for fast bowlers. This would potentially allow coaches to utilise MIMUs as a field-based lumbar injury risk screening tool. Doing so could facilitate technique intervention prior to the onset of lumbar injury.
7.7 References


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Chapter 7: MIMU-based Measurement of Trunk and Pelvis Angles during Cricket Bowling


8.1 Summary of Findings

Three-dimensional (3D) retro-reflective motion capture is the current gold-standard of biomechanical analysis but the technology is heavily limited by ecological validity concerns, which are pertinent to sports biomechanical analysis. Two portable, cost-effective alternate technologies that have shown potential as field-based kinematic measurement tools are two-dimensional (2D) video cameras and magneto-inertial measurement units (MIMUs). However, the measurement validity of both modalities can be affected by a multitude of factors, which include the movement task and body regions being assessed. The research within this thesis has presented information on factors affecting MIMU and 2D video measurement validity of trunk and pelvis angles during cricket fast bowling; a high-speed, multi-planar sporting task. The criterion measure for all research was 3D retro-reflective motion capture and modelling. A summary of the key findings and conclusions from chapter 4 through chapter 7 are outlined emphasising the significance of the research.

8.1.1 Chapter 4: Measuring Bowling Kinematics for Two-Dimensional Video

Listed here are the hypotheses initially presented in chapter 1, with comment made on the major outcomes of the study:

- That discrete measurements of trunk, pelvis, hip and knee angles recorded from multi-plane 2D video would be comparable with anatomically modelled 3D retro-reflective outputs of equivalent discrete variables during the fast bowling action.
  - **Hypothesis rejected due to the measurements not being comparable.**

- That 2D video motion analysis could be used as effective screening tool for identifying lumbar injury risk in cricket fast bowlers.
  - **Hypothesis rejected as video analysis is inappropriate for comprehensive lumbar injury risk screening.**

The main outcome from chapter 4 (study 1) was that 2D video does not appear to be wholly suitable as a measurement tool for comprehensive kinematic analyses of cricket bowling biomechanics. This finding is especially relevant to lumbar injury risk screening for cricket fast bowlers. Thorax lateral flexion angles, pelvis rotation angles, hip flexion-extension angles, and knee flexion-extension angles during bowling were calculated at discrete points by digitising the 2D video and were compared with anatomically modelled retro-reflective motion capture data. Only the 2D
measures of pelvis rotation angle at ball release (intra-class correlation (ICC) = 0.91, root mean square difference (RMSD) = 5.9°) and knee flexion-extension at ball release (ICC = 0.97, RMSD = 4.3°) demonstrated reasonable measurement validity, with all other measures reporting poor to moderate ICCs (0.38-0.73) and RMSDs between 5.7° and 9.4°. These differences were primarily attributed to the large discrepancies in measurement methods of joint and segments angles between 2D video digitising and 3D retro-reflective anatomical modelling. Complex, high-speed movements that cross multiple planes of motion (e.g., trunk movement) appeared to be especially difficult to measure validly using a 2D video approach. Alternative approaches to field-based measurement of fast bowling kinematics should be sought for the analyses of fast bowler lumbar injury risk.

8.1.2 Chapter 5: Functionally Calibrated MIMU-based Thorax and Lumbar Measurements

Listed here are the hypotheses initially presented in chapter 1, with comment made on the major outcomes of the study:

- That MIMU functional calibration would significantly improve MIMU measurement validity when compared with simple anatomical calibration for measurements of thorax and lumbar angles during uni-planar range of movement tasks and cricket fast bowling.
  - **Hypothesis rejected as functional calibration did not improve MIMU measurement validity.**

- That an MIMU-based ZXY-ordered Euler angle decomposition for thorax and lumbar angles would produce greater measurement differences than functionally calibrated angle calculations when both methods were compared with 3D retro-reflective derived angles.
  - **Hypothesis rejected as ZXY Euler angles did not produce greater differences.**

Magneto-inertial measurement units are an alternative field-based 3D motion capture solution to 2D video. A key consideration is the calibration of MIMUs to body segments so that measurements are functionally meaningful (Picerno, 2017). The main conclusion from chapter 5 (study 2) was that incorporating functionally calibrated axes of rotation in MIMU angle calculation methods for the thorax and lumbar segments does not appear to improve measurement validity when assessed against the criterion measure (3D retro-reflective motion capture). From a concurrent validity perspective, it appears that visually aligning MIMUs with the spine produces similar measurement validity to functional axis calibration. Simple ZXY-ordered Euler angle decomposition does not produce less valid trunk measurements than calculation methods incorporating functional axes for
uni-planar trunk range of motion tasks and cricket fast bowling. It should be noted that none of the five angle calculation methods assessed (three involving functional axes, two with no functional axes) assessed produced significant differences to retro-reflective based data during the clinical (ROM) or dynamic (bowling) tasks when analysed using one-dimensional statistical parametric mapping (SPM1D). The practical implication of chapter 5 is that the functional movements performed in this study may not be necessary for MIMU calibration of the thorax and lumbar segments. Forgoing functional calibration simplifies MIMU data collection procedures and signal processing requirements for MIMU users.

8.1.3 Chapter 6: Comparison of Trunk Angle Measures from Two Commercial Systems

Listed here are the hypotheses initially presented in chapter 1, with comment made on the major outcomes of the study:

- That concurrent measurement validity of two similar, commercial MIMU systems would not significantly vary when thorax and lumbar angle measurements from both systems were compared with 3D retro-reflective angle outputs.
  - Hypothesis rejected as concurrent validity did vary in some instances.

- That thorax and lumbar angle measurement validity of both MIMU systems would reduce as movement velocity and complexity increase.
  - Hypothesis accepted

Previous research has found that commercial MIMU systems can produce varying measurements, with discrepancies in sensor fusion methods often attributed as a major cause of such differences (Bergamini et al., 2014; Filippeschi et al., 2017; Lebel et al., 2015, 2013; Ligorio et al., 2016; Picerno et al., 2011). Chapter 6 (study 3) revealed that differences can occur even when the MIMU systems assessed incorporate very similar sensor specifications and the same general sensor fusion approaches (Kalman filter based algorithms). It was also found that these differences can escalate for complex, high-speed movements, which supports previous research (Lebel et al., 2013). Nonetheless, both of the MIMU systems evaluated in study 3 showed good concurrent measurement validity assessed against 3D retro-reflective motion capture data during uni-planar trunk movements. Thorax global, lumbar global, thorax-to-pelvis and lumbar-to-pelvis angles all showed RMSDs of less than $3\pm1.5^\circ$ for both MIMU systems, although some small but statistically significant differences were computed between the two systems at discrete time-points. One of the MIMU models produced significantly larger, functionally meaningful differences during cricket bowling, with discrete RMSDs up to $7.4\pm1.9^\circ$ calculated. Despite the variation to RMSD values between the two systems, SPM1D indicated no significant differences for any variables when
compared with the criterion measure across the entire movement phases. It was concluded that variations to sensor fusion algorithms (such as the included algorithm parameters) and contrasting system sampling rates (100 Hz vs 75 Hz) were the likely causes of measurement differences. Users of commercial MIMU systems should be aware that even slight discrepancies between MIMU models can potentially alter trunk angle measurements, especially during high-speed, sport-specific tasks.

8.1.4 Chapter 7: MIMU-based Measurement of Trunk and Pelvis Angles during Cricket Bowling

Listed here are the hypotheses initially presented in chapter 1, with comment made on the major outcomes of the study:

- There would be significant measurement differences between MIMUs and 3D retro-reflective based anatomical modelling when assessing thorax, lumbar and pelvis angles during fast bowling.
  - **Hypothesis accepted**

- That there would be no evidence of a systematic linear offset between the measurements of MIMUs and 3D retro-reflective based anatomical modelling when assessing thorax, lumbar and pelvis angles during fast bowling.
  - **Hypothesis accepted**

- That MIMU derived angles would not be comparable with 3D retro-reflective derived angles in the previous fast bowling literature.
  - **Hypothesis accepted**

A large portion of cricket bowling biomechanical research has employed 3D retro-reflective motion capture to measure kinematics involving the thorax, lumbar and pelvis segments. Many of these variables have been associated with lumbar injury. Chapter 7 (study 4) was an applied cricket bowling study that compared MIMU measurements of thorax, lumbar and pelvis angles with anatomically modelled 3D retro-reflective outputs. It appears that MIMU based modelling is generally incomparable to retro-reflective based modelling when measuring those variables during the bowling delivery stride phase. The SPM1D analysis showed MIMU and retro-reflective measures were statistically different for 19 of the 21 variables across an average 41.4±27.7% of the bowling delivery stride. The discrete measurement difference was 13.9±6.4° when averaged for all angles assessed, which has significant functional implications. Furthermore, no systematic linear offset between the two forms of angle measurement was observed. Similar conclusions were made
in a recent MIMU study investigating the pelvis and upper-thorax during discus throwing (Brice et al., 2018). The consequence of these findings is that MIMU-based angle measures of the thorax, lumbar and pelvis segments during cricket bowling should not be compared with the data in the existing bowling literature. The major apparent sources of measurement difference were variations to segment coordinate system orientations and soft tissue artefact errors, caused by having the MIMUs and reflective markers placed on different segment locations in an attempt to measure a similar motion. The lack of skeletal rigidity for the thorax and lumbar segments may also have contributed to the measurement discrepancies. The important practical implication to arise from study 4 is that there is a requirement for lumbar injury risk kinematic thresholds to be established specifically for MIMU data. If this can be achieved, then it may be possible for MIMUs to be employed by cricket researchers and coaches as effective lumbar injury risk screening tools.

8.2 Synthesis of Findings

The studies presented in this thesis provide valuable information on a range topics. Insight has been provided on the measurement validity of 2D video, MIMUs and 3D retro-reflective motion capture, as well as the various modelling techniques employed in conjunction with these technologies. The research findings highlight the importance of conducting measurement validity assessments that are specific to movement tasks, body regions and motion capture modalities. The research is also highly relevant to cricket bowling and the identification of biomechanical risk factors of lumbar injury. This section will synthesise these findings and suggest possible future research directions for the cricket and wider biomechanics research fields.

8.2.1 Two-dimensional Video-Based Motion Capture and Digitising

Two-dimensional video-based motion capture is widely used by sport coaches to provide qualitative and quantitative feedback to athletes. Chapter 4 examined the measurement validity of 2D video-based digitisation to measure joint angles associated with lumbar injury risk in cricket bowlers. The findings may serve to remind one that even though 2D video is a practical and affordable motion capture tool, it should not be depended upon for valid, comprehensive biomechanical analyses of complex sporting movements, such as cricket bowling. Simplistic 2D vector-based measurement procedures and perspective errors are major limitations of video-based analysis (Elliott and Alderson, 2007; Lopes et al., 2018). Standard-speed video cameras are also hampered by lower-frame rates that drastically reduce the chance of accurately capturing discrete values (e.g., maxima, minima) or task-specific events (e.g., foot contact, ball release). With respect to digitising approaches, care should be taken to ensure that the orientation of 2D vectors match body segment positions as closely as possible. Joint or segment angles should not be measured if
the movement of interest is occurring out of plane with the camera. Correctly positioning cameras can reduce the risk of perspective errors for some variables, although complicated movements that cross multiple planes of motion (e.g., trunk movements during bowling) will be inherently challenging to measure from a 2D perspective.

8.2.2 Magneto-inertial Measurement Unit-Based Motion Capture and Modelling

Chapters 5, 6 and 7 investigated different aspects of MIMU motion capture and modelling, particularly in relation to thorax, lumbar and pelvis angle measurements during sport-specific tasks. It is generally apparent from the results in chapters 5 and 6 that MIMUs provide good measurement validity from a system or technology perspective. Results from chapter 7 suggest that their application in sport from a measurement validity/biomechanical modelling perspective remains debatable, especially for complex sporting movements like cricket bowling. It is important to remember that a MIMU is restricted to measurement of its own orientation. Therefore, valid MIMU-based measurement of human movement is highly dependent on the placement and calibration of MIMUs to the body segment of interest. It is unrealistic to expect that a single MIMU will always reflect the movement of an entire body segment, especially for segments such as the thorax, where segment rigidity assumptions are poorly adhered to during dynamic movements (Borkowski et al., 2016; Kudo et al., 2017). Consumers should also be aware that soft tissue artefact (STA) is highly variable, with factors such as the complexity and velocity of the movement task; the participant’s body composition; and the body region of interest all markedly affecting STA levels and subsequent MIMU measures (Kudo et al., 2017; Leardini et al., 2005). In an attempt to nullify the impact of these issues on measurement comparisons, the MIMU data recorded for chapters 5 and 6 were only compared with 3D retro-reflective measurements derived from markers placed directly on the MIMUs (Bergamini et al., 2013). This enabled MIMU and retro-reflective coordinate systems to be easily aligned and negated the influence of large variations to STA on angle measurement comparisons. This practice is highly recommended for all MIMU measurement validity studies, as it allows for measurement differences due to system dissimilarities to be somewhat dissociated from differences arising from modelling discrepancies.

There are varying recommendations in the MIMU literature regarding sensor-to-segment calibration procedures (Cutti et al., 2010, 2008; de Vries et al., 2010; Favre et al., 2009; Galinski and Dehez, 2012; Ligorio et al., 2017; O’Donovan et al., 2007; Picerno et al., 2008). Chapter 5 added to the existing information by showing that functional MIMU calibration procedures for the thorax and lumbar segments does not significantly improve the concurrent measurement validity of thorax and lumbar angle measures during uni-planar range of motion tasks or cricket bowling. It is important to note that the reliability of these approaches was not assessed. Nonetheless, these findings suggest that visually aligning MIMUs with the spine may be adequate for valid MIMU
measurement of the trunk. Consequently, MIMU users may be able to simplify data collection and data processing procedures by opting for visual anatomical alignment of MIMUs to the trunk and by continuing to utilise basic Euler angle decomposition to calculate joint and segment orientations.

It is important to re-emphasise a major limitation of study 5, which was the absence of a proper ground-truth angle calculation method. This meant that each MIMU angle calculation method was only compared to retro-reflective data modelled using the same angle calculation method. It is possible that one of the angle calculation methods that incorporated functional axes of rotation may produce values that are more reflective of the ‘true’ movement of the thorax and lumbar segments. An attempt could be made to investigate this hypothesis by comparing each of the angle calculation methods with anatomically modelled retro-reflective data. Unfortunately, this would introduce undesirable measurement errors (e.g., STA, landmark misidentification) to the comparison, making the approach somewhat futile as the source of differences would not be able to be isolated. Hence, whilst the findings of chapter 5 have clear practical applications they should be interpreted with a degree of caution.

There is a wide range of commercial MIMU systems currently available to consumers. Chapter 6 reiterated the importance of selecting a system that is well suited to the motion analysis task at hand. This is especially pertinent for athlete motion capture, as the measurement validity differences between the two MIMU models tested in chapter 6 were more likely to be functionally meaningful during cricket bowling than uni-planar trunk range of motion tasks. System characteristics such as the manufacturer’s sensor fusion algorithm (and incorporated algorithm parameters); sensor component specifications (accelerometer, gyroscopes and magnetometers speeds); and system sample rates should all be evaluated prior to selecting MIMU systems for sports performance analysis (Bergamini et al., 2014; Filippeschi et al., 2017; Lebel et al., 2015, 2013; Ligorio et al., 2016; Pasciuto et al., 2015; Picerno et al., 2011).

Regardless of the commentary above, MIMUs are a highly useful motion capture tool and are evidently suitable for field-based sports biomechanical analysis. Their portability and affordability is a major asset, as is their ability to overcome the occlusion and restrictive capture volume problems that restrain vision-based motion capture. Biomechanists should continue to validate MIMU measurements for a diverse range of applications. Doing so could potentially improve ecological validity and accuracy of field-based biomechanical analyses.

8.2.3 Modelling 3D Retro-Reflective Motion Capture Data

Whilst this thesis has focused primarily on alternative motion capture solutions, a few aspects of 3D retro-reflective motion capture should also be commented on. Firstly, it is important for researchers to be cognisant that retro-reflective based modelling is frequently not an entirely
truthful representation of human motion. Retro-reflective based modelling is affected by many sources of error, chiefly STA and anatomical landmark misidentification (Della Croce et al., 2005; Leardini et al., 2005). However, camera calibration issues; frequent marker occlusion; and choice of data filtering methods are also among the factors that can alter kinematic measurements (Cappozzo et al., 2005). Biomechanical modelling methodologies are also not ubiquitous in laboratories around the world. This complicates matters if anatomically modelled retro-reflective data are being used as a reference or criterion measure, as it was in this thesis. For instance, it is unknown whether placing markers on differing segment landmarks would have altered the findings of the research presented in chapters 4 and 7. Although well validated methods for hip and shoulder joint centre estimation were utilised, alternative approaches may have modified results relating to those joints in chapters 4 and 7 respectively.

It has been previously noted that the trunk segment is especially difficult to model because the spine is comprised of multiple vertebrae and joints, which all permit degrees of movements (Alway et al., 2018; Borkowski et al., 2016; Crewe et al., 2013). Lack of segment rigidity was likely a large contributor to the significant measurement differences displayed in chapter 7 between the retro-reflective and MIMU derived thorax and lumbar angles. Certainly, STA accounts for a large portion of this segment distortion (Kudo et al., 2017), however, potential skeletal movement within a modelled segment (e.g., thorax, lumbar) is also somewhat concerning. It may be appropriate for biomechanists to consider more nuanced segmental modelling for the trunk, particularly for high-speed, multi-planar sporting motions that involve large spinal ranges of motion, such as cricket bowling. Modelling smaller spinal segments that contain only two or three vertebra may mitigate the influence of spinal non-rigidity on kinematic outputs (Alway et al., 2018).

8.2.4 Assessments of Measurement Validity

This thesis also highlighted the importance of conducting highly-specific measurement validity assessments. The results from all four studies suggest that motion analysis measurement validity is variable and can depend on: the movement task; the body segment or joint of interest; the specific technologies assessed (e.g., MIMU model); the employed modelling methodologies; and the criterion measure. Researchers should not assume that a measurement modality has good validity if it has not been assessed in comparable circumstances to their intended application (e.g., similar movement speed, range of motion, degrees of freedom).

8.2.5 Lumbar Injury Risk Screening for Cricket Bowlers

The sport-specific application of this research was to cricket fast bowling and measurement of kinematics associated with fast bowler lumbar injury risk. The findings of chapter 4 suggest that, whilst 2D video is powerful coaching tool, it should not be relied upon as a platform for
comprehensive and valid quantitative analyses of fast bowling biomechanics, particularly in relation to identifying kinematic indicators of lumbar injury risk. Magneto-inertial measurement units appear to be a superior field-based prospect for measurement of trunk and pelvis angles during fast bowling. However, MIMU-based angle data for the thorax, lumbar and pelvis should not to be directly compared with data in prior fast bowling literature. This recommendation should be adhered to by cricket coaches and researchers. Failure to do so could lead to bowlers receiving feedback that is invalid, unnecessary or potentially detrimental to performance and injury risk. As such, cricket researchers should continue to investigate alternative means of lumbar injury risk screening or potentially work towards developing kinematic thresholds for injury risk that are specific to MIMU data.

8.2.6 Future Research Directions

Some future research directions relevant to this thesis are:

- Continue to enhance measurement validity research on MIMU-based biomechanical analysis by providing information that is movement task and body region specific.
- Report reliability for functional and anatomical sensor-to-segment calibration methods for the thorax and lumbar segments.
- Continue development of thorax and lumbar modelling methods, especially for sporting tasks that involve high-speed, multi-planar trunk movements.
- Conduct prospective lumbar injury research that utilises both 3D retro-reflective motion capture and MIMU motion capture to quantify bowling kinematics. This would help to facilitate development of MIMU-based kinematic thresholds of lumbar injury risk in fast bowlers.

8.3 Practical Applications

The applied knowledge that can be disseminated from the results of this thesis are:

- Two-dimensional video-based biomechanical analysis should not be viewed as a wholly valid, robust solution to kinematic-based lumbar injury risk screening for cricket fast bowlers.
- Overall, MIMUs appear to be a viable field-based alternative to 3D retro-reflective motion capture systems for measurement of trunk and pelvis angles during dynamic sporting tasks. However, modelling discrepancies mean that MIMU-based measures generally do not reflect anatomically modelled 3D retro-reflective motion capture data.
• Measurement validity of MIMU-based measures of thorax and lumbar angles reduces as movement velocity and complexity increases.

• It may not be necessary to employ functional calibration procedures for effective alignment of MIMUs to the thorax and lumbar segments. Simple visual alignment of MIMUs to the spine facilitates good measurement validity for thorax and lumbar angles during uni-planar and sport-specific tasks.

• Basic Euler angle decomposition appears to be acceptable for calculations of MIMU-based thorax and lumbar angles.

• Trunk angle measurements from commercial MIMU systems can vary significantly during sport-specific tasks, even when system specifications and sensor fusion approaches are comparable.

• There are functionally significant measurement differences between MIMU and retro-reflective based measures of thorax, lumbar and pelvis angles during the cricket bowling action. There is also no apparent systematic linear relationship between the measures.

• Magneto-inertial measurement unit derived measures of thorax, lumbar and pelvis angles during fast bowling should not be compared with data in the existing bowling literature, especially in relation to lumbar injury risk.


8.4 References


Chapter 9: Supplementary Materials

This chapter includes supplementary material for the thesis. Additional data are provided for chapter 5 (section 9.1) and chapter 6 (section 9.2). Evidence of ethical approval is also displayed (sections 9.3, 0, and 9.5), along with participant information sheets and consent forms for all studies. The thesis concludes by providing copies of the peer-reviewed conference proceedings that have arisen from this work (section 0).
### 9.1 Appendix A: Additional Results Tables for Chapter 5

Table 9.1: Root mean square difference (RMSD) averages and standard deviations for five angle calculation methods when averaged for each movement task across all joint angles and all planes of motion. ANOVA F and p-values also presented.

<table>
<thead>
<tr>
<th>Task</th>
<th>Event</th>
<th>Angle Calculation Method RMSD Mean and St Dev</th>
<th>ANOVA F (p-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>ZXY</td>
<td>GS</td>
</tr>
<tr>
<td>ROM</td>
<td>Max</td>
<td>2.6±1.5°</td>
<td>2.4±1.4°</td>
</tr>
<tr>
<td></td>
<td>Min</td>
<td>2.3±1.4°</td>
<td>2.2±1.5°</td>
</tr>
<tr>
<td></td>
<td>Tmax</td>
<td>2.5±1.4°</td>
<td>2.5±1.4°</td>
</tr>
<tr>
<td>Bowling</td>
<td>Max</td>
<td>4.6±1.4°</td>
<td>4.6±1.4°</td>
</tr>
<tr>
<td></td>
<td>Min</td>
<td>5.8±1.7°</td>
<td>5.8±1.7°</td>
</tr>
<tr>
<td></td>
<td>Tmax</td>
<td>7.4±1.9°</td>
<td>7.4±1.9°</td>
</tr>
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</table>
Table 9.2: Root mean square difference (RMSD) averages and standard deviations for five angle calculation methods when averaged for each movement task and joint angle across all planes of motion. ANOVA F and p-values also presented.

<table>
<thead>
<tr>
<th>Task</th>
<th>Joint Angle Event</th>
<th>Angle Calculation Method</th>
<th>RMSD Mean and St Dev</th>
<th>ANOVA F (p-value)</th>
</tr>
</thead>
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<td></td>
<td>ZXY</td>
<td>GS</td>
<td>GSfunctional</td>
</tr>
<tr>
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<td></td>
<td></td>
<td></td>
<td></td>
</tr>
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<td>ROM Thorax Global Max</td>
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<td>1.8±0.9°</td>
</tr>
<tr>
<td></td>
<td>Tmax</td>
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<td>2.3±0.9°</td>
<td>2.3±0.9°</td>
</tr>
<tr>
<td>Thorax-to-Pelvis Max</td>
<td>3.6±1.7° 3.6±1.7°</td>
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<td>2.8±1.2°</td>
<td>0.71 (0.595)</td>
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<td>2.9±1.2°</td>
</tr>
<tr>
<td></td>
<td>Tmax</td>
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<td>3.0±1.4°</td>
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<td>1.18 (0.349)</td>
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<td>2.9±0.5°</td>
<td>1.8±0.8°</td>
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<td>Lumbar-to-Pelvis Max</td>
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<td>Tmax</td>
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<td>2.4±1.3°</td>
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<td>Bowling Thorax Global Max</td>
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<td>9.1±7.0°</td>
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<td>Thorax-to-Pelvis Max</td>
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</tr>
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<td>4.7±2.1° 6.4±4.4°</td>
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<td>9.7±8.2°</td>
</tr>
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<td>Lumbar-to-Pelvis Max</td>
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<td>7.2±5.2°</td>
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<td>8.6±0.7°</td>
<td>7.7±3.4°</td>
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*Significant difference (p < 0.005) between ZXY and GSfunctional, as determined by Bonferroni post-hoc analysis.
Table 9.3: Root mean square difference (RMSD) averages and standard deviations for five angle calculation methods when averaged for each movement task and plane of motion across all joint angles. ANOVA F and p-values also presented.

<table>
<thead>
<tr>
<th>Task</th>
<th>Plane of Motion</th>
<th>Event</th>
<th>Angle Calculation Method</th>
<th>RMSD Mean and St Dev</th>
<th>ANOVA F (p-value)</th>
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<td>ZXY</td>
<td>GS</td>
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<td></td>
</tr>
<tr>
<td>ROM</td>
<td>Flexion-Extension</td>
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<td>1.8±1.6°</td>
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<td>2.4±1.5°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Min</td>
<td>1.5±1.3°</td>
<td>1.4±1.3°</td>
<td>2.6±1.2°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Tmax</td>
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<td>1.6±1.2°</td>
<td>2.7±1.2°</td>
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<tr>
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<td>2.7±1.2°*</td>
<td>4.4±1.6°*</td>
</tr>
<tr>
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<td></td>
<td>Min</td>
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</tr>
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<td>4.3±1.7°</td>
</tr>
<tr>
<td></td>
<td>Rotation</td>
<td>Max</td>
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<td>3.8±1.2°</td>
<td>4.4±1.3°</td>
</tr>
<tr>
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<td>Tmax</td>
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<td>4.0±0.6°</td>
<td>4.1±0.7°</td>
</tr>
<tr>
<td>Bowling</td>
<td>Flexion-Extension</td>
<td>Max</td>
<td>3.9±0.4°</td>
<td>3.9±0.3°</td>
<td>4.8±1.3°</td>
</tr>
<tr>
<td></td>
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<td>Min</td>
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<td>4.8±0.9°</td>
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</tr>
<tr>
<td></td>
<td></td>
<td>Tmax</td>
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<td>7.7±1.4°</td>
<td>7.8±1.6°</td>
</tr>
<tr>
<td></td>
<td>Lateral Flexion</td>
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<td>4.0±1.5°</td>
<td>4.0±1.5°</td>
<td>5.6±1.8°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Min</td>
<td>6.2±2.6°</td>
<td>6.2±2.6°</td>
<td>8.2±3.0°</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Tmax</td>
<td>6.6±2.3°</td>
<td>6.6±2.3°</td>
<td>6.6±1.7°</td>
</tr>
<tr>
<td></td>
<td>Rotation</td>
<td>Max</td>
<td>6.0±1.0°</td>
<td>6.0±1.0°</td>
<td>6.4±1.3°</td>
</tr>
<tr>
<td></td>
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<td>Min</td>
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<td>6.5±1.5°</td>
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<tr>
<td></td>
<td></td>
<td>Tmax</td>
<td>7.9±2.1°</td>
<td>7.9±2.1°</td>
<td>8.0±2.4°</td>
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</tbody>
</table>

*Significant difference (p < 0.005) between ZXY and GSfunctional, as determined by post-hoc analysis.
9.2 Appendix B: Statistical Parametric Mapping Figures for Chapter 6

Figure 9.1: Angle waveforms and SPM1D analysis for thorax global flexion-extension angles during the seated trunk flexion-extension range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.2: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative flexion-extension angles during the seated trunk flexion-extension range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.3: Angle waveforms and SPM1D analysis for lumbar global flexion-extension angles during the seated trunk flexion-extension range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.4: Angle waveforms and SPM1D analysis for lumbar-to-pelvis relative flexion-extension angles during the seated trunk flexion-extension range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.5: Angle waveforms and SPM1D analysis for thorax global lateral flexion angles during the seated trunk lateral flexion range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.6: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative lateral flexion angles during the seated trunk lateral flexion range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.7: Angle waveforms and SPM1D analysis for lumbar global lateral flexion angles during the seated trunk lateral flexion range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.8: Angle waveforms and SPM1D analysis for lumbar-to-pelvis relative lateral flexion angles during the seated trunk lateral flexion range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.9: Angle waveforms and SPM1D analysis for thorax global rotation angles during the seated trunk rotation range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.10: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative rotation angles during the seated trunk rotation range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.11: Angle waveforms and SPM1D analysis for lumbar global rotation angles during the seated trunk rotation range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.12: Angle waveforms and SPM1D analysis for lumbar-to-pelvis relative rotation angles during the seated trunk rotation range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.13: Angle waveforms and SPM1D analysis for thorax global flexion-extension angles during the standing trunk flexion-extension range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.14: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative flexion-extension angles during the standing trunk flexion-extension range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.15: Angle waveforms and SPM1D analysis for lumbar global flexion-extension angles during the standing trunk flexion-extension range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.16: Angle waveforms and SPM1D analysis for lumbar-to-pelvis relative flexion-extension angles during the standing trunk flexion-extension range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.17: Angle waveforms and SPM1D analysis for thorax global lateral flexion angles during the standing trunk lateral flexion range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Critical t-threshold = 3.20
Critical t-threshold = 3.25

Figure 9.18: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative lateral flexion angles during the standing trunk lateral flexion range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Critical t-threshold = 3.48
Critical t-threshold = 3.46
Figure 9.19: Angle waveforms and SPM1D analysis for lumbar global lateral flexion angles during the standing trunk lateral flexion range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.20: Angle waveforms and SPM1D analysis for lumbar-to-pelvis relative lateral flexion angles during the standing trunk lateral flexion range of motion task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.21: Angle waveforms and SPM1D analysis for thorax global flexion-extension angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.22: Angle waveforms and SPM1D analysis for thorax global lateral flexion angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.23: Angle waveforms and SPM1D analysis for thorax global rotation angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.24: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative flexion-extension angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.25: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative lateral flexion angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.26: Angle waveforms and SPM1D analysis for thorax-to-pelvis relative rotation angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.27: Angle waveforms and SPM1D analysis for lumbar global flexion-extension angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.28: Angle waveforms and SPM1D analysis for lumbar global lateral flexion angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.29: Angle waveforms and SPM1D analysis for lumbar global rotation angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.30: Angle waveforms and SPM1D analysis for lumbar-to-pelvis relative flexion-extension angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
Figure 9.31: Angle waveforms and SPM1D analysis for lumbar-to-pelvis relative lateral flexion angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.

Figure 9.32: Angle waveforms and SPM1D analysis for lumbar-to-pelvis relative rotation angles during the bowling task. The MIMU data are compared with concurrently collected retro-reflective technical coordinate system (RRtech) data for both MIMU systems. Mean and standard deviation plots are shown on the left, with accompanying SPM1D visual representations on the right. Critical t-statistics are displayed on all SPM1D figures.
9.3 Appendix C: Ethical approval from the University of Western Australia Human Research Ethics Committee (Initial Approval)

Our Ref: RA/4/12593

Assistant Professor Jacqueline Alderson
Sport Science, Exercise & Health (School of)
MBDP: M408

Dear Professor Alderson

EXERCISE SCIENCE SUBCOMMITTEE - ETHICS APPROVAL

Lumbar spinal loading in junior cricket fast bowlers

Student(s): Helen Crewe - Masters - 2054175

Ethics approval for the above project has been granted by the Exercise Science Subcommittee from 04 November 2009 to 04 November 2014 in accordance with the requirements of the National Statement on Ethical Conduct in Human Research (National Statement) and the policies and procedures of The University of Western Australia.

You are reminded of the following requirements:

1. The application and all supporting documentation form the basis of the ethics approval and you must not depart from the research protocol that has been approved.
2. The Human Research Ethics Office must be approached for approval in advance for any requested amendments to the approved research protocol.
3. The Chief Investigator is required to report immediately to the Human Research Ethics Office any adverse or unexpected event or any other event that may impact on the ethics approval for the project.
4. The Chief Investigator must inform the Human Research Ethics Office as soon as practicable if a research project is discontinued before the expected date of completion, providing reasons.

Any conditions of ethics approval that have been imposed are listed below.

Special Conditions
None specified

The University of Western Australia is bound by the National Statement to monitor the progress of all approved projects until completion to ensure continued compliance with ethical standards and requirements.

Please note that the maximum period of ethics approval for this project is five (5) years from the date of this notification. However, ethics approval is conditional upon satisfactory progress reports being received by the designated renewal date for continuation of ethics approval. The Human Research Ethics Office will forward a request for a progress report communication for your completion approximately four weeks before this renewal date.

If your progress report is not then received by the renewal date, your ethics approval will expire, requiring that all activities relating to the project cease immediately.

If you have any queries please do not hesitate to contact Kate Kirk on (08) 6488 3703.

Please ensure that you quote the file reference – RA/4/12593 — and the associated project title in all future correspondence.
9.4 Appendix D: Ethical approval from the University of Western Australia Human Research Ethics Committee (Amendment Approval 1)

Our Ref: RA/4/1/2593

06 August 2014

Associate Professor Jacqueline Alderson
School of Sport Science, Exercise & Health
MBDP: M408

Dear Professor Alderson

HUMAN RESEARCH ETHICS OFFICE - AMENDMENT REQUEST APPROVED

Lumbar spinal loading in junior cricket fast bowlers

Student(s): Daniel Shane Cottam, Helen Louise Bayne, Jack William Halley

I confirm receipt of your correspondence requesting an amendment to the protocol for the above project.

Approval has been granted for the amendment as outlined in your correspondence and attachments (if any) subject to any conditions listed below.

The following is a brief description of the amendment and any conditions that apply:

1. Re-open the project till 1 August 2015
2. Increase the recruited participants to improve the statistical power of the study (from 40 bowlers to 60 bowlers).
3. Updated PIF
4. Include Jack Halley and Daniel Cottam to this project

If you have any questions, please contact the HREO at hreo.research@uwa.edu.au

Please ensure that you quote the file reference RA/4/1/2593 and the associated project title in all future correspondence.

Yours sincerely,

Dr Mark Dixon
Associate Director, Research Integrity
9.5 Appendix E: Ethical approval from the University of Western Australia Human Research Ethics Committee (Amendment Approval 2)

Our Ref: RA4/1/2663

27 January 2017

Associate Professor Jacqueline Alderson
School of Sport Science, Exercise and Health
MSDP: M203

Dear Professor Alderson

HUMAN RESEARCH ETHICS OFFICE—AMENDMENT REQUEST APPROVED

Lumbar spinal loading in young cricket fast bowlers

I confirm receipt of your correspondence requesting an amendment to the protocol for the above project. Approval has been granted for the amendment as outlined in your correspondence and attachments (if any) subject to any conditions listed below. The following is a brief description of the amendment and any conditions that apply:

1. Update personnel: Jonathan Stainor & Steven Kosovich
2. The participant age range be expanded to include 18-24 year old males
3. The use of intramuscular (fine-wire) electromyography (EMG) to collect muscle activation signals from participants whilst bowling and also whilst performing three neuromuscular assessments. Only participants aged 18-24 will be recruited for this study. Conducted by Daniel Cottam.
4. The use of surface EMG to collect muscle activation signals from participants whilst bowling.
5. The use of inertial sensors during bowling.
6. The project title be changed from Lumbar spinal loading in junior cricket fast bowlers to Lumbar spinal loading in young cricket fast bowlers.

If you have any queries, please contact the HEO at humanaethics@uwa.edu.au.

Please ensure that you quote the file reference RA4/1/2663 and the associated project title in all future correspondence.

Peter Millar
Acting Manager, Human Ethics

Name: Associate Professor Jacqueline Alderson
Faculty / School: School of Sport Science, Exercise and Health
Role: Chief Investigator
Student(s): Helen Grewe - Masters - 20014703, Daniel Cottam, Helen Bayne, Jonathan Stainor, Steven Kosovich
Appendix F: Information Sheet and Consent Form (Chapter 4)

**Cricket Bowling Three-Dimensional Analysis**

-Participant Information Sheet-

**Purpose**

The overall aim of this research will be to improve bowling performance and coaching practices in cricket and also to reduce the chance of injury during bowling.

**Procedures**

Three-dimensional (3D) biomechanical assessments, as well as video recordings, of your bowling action will be undertaken at the UWA Biomechanics Laboratory. On arrival at these testing sessions you will be shown around the lab and the procedures will be explained fully. You will then be asked to dress in running or cycling shorts, and shoes suitable for indoor bowling. You will be required to perform the assessment without a shirt. Small reflective markers will be placed on your skin using double-sided tape. You will be given chance to warm up in preparation for the bowling task. You will then be required to bowl approximately 20 deliveries at maximal effort.

Testing sessions will run for approximately two hours. You will need to organise transport to and from the Biomechanics Laboratory at UWA. A map and directions are provided with this information.

**Risks**

The risk of injury in the testing session will be similar to that of a cricket match or standard practice situation. The utmost care will be taken to avoid any injury occurrence. If at any stage of the testing you sustain an injury to any area or feel any pain, tightness you are highly encouraged to report this to the researchers.

Low Allergenic double-sided tape will be used to affix markers to your skin during the 3D analysis. This tape can cause minor skin irritation on some people, and can be momentarily painful when removed.

**Benefits**

We believe that this research will result in better understanding of correct coaching, training and performance practices for young fast bowlers. You will have the opportunity to view your bowling action as a 3D representation and also in slow-motion 3D camera footage.
Chapter 9: Supplementary Materials

Confidentiality

All data, video and images will be stored in a secure location at the School of Sport Science, Exercise & Health (UWA), Crawley, Western Australia. No names will be present in any reported results.

Participant Rights

Participation in this research is voluntary and you are free to withdraw from the study at any time without prejudice. You can withdraw for any reason and you do not need to justify your decision.

If you do withdraw we may wish to retain the data that we have recorded from you but only if you agree, otherwise your records will be destroyed.

Your participation in this study does not prejudice any right to compensation that you may have under statute of common law.

If you have any questions concerning the research at any time please feel free to ask the researchers about your concerns. Further information regarding this study may be obtained from the researchers.

The Human Research Ethics Committee at the University of Western Australia requires that all participants are informed that, if they have any complaint regarding the manner, in which a research project is conducted, it may be given to the researcher or, alternatively, to the Secretary, Human Research Ethics Committee, Registrar’s Office, University of Western Australia, 35 Stirling Highway, Crawley, WA 6009 (telephone number 6480-3703). All study participants will be provided with a copy of the Information Sheet and Consent Form for their personal records.
Cricket Bowling Three-Dimensional Analysis
-Participant Consent Form-

I have read the information provided and any questions I have asked have been answered to my satisfaction. I agree to participate in the bowling analysis, realising that I may withdraw at any time without reason and without prejudice. I understand that all information provided is treated as strictly confidential and will not be released by the investigator unless required to by law. I have been advised as to what data is being collected, what the purpose is, and what will be done with the data upon completion of the research. I agree that data gathered for the study may be published provided my name or other identifying information is not used.

I consent to the use of photographs and video recordings of my bowling action in scientific presentations:
YES [ ] NO [ ]

Participant Name ___________________________ Date ___________________________

Participant Signature ___________________________

The Human Research Ethics Committee at the University of Western Australia requires that all participants are informed that, if they have any complaint regarding the manner, in which a research project is conducted, it may be given to the researcher or, alternatively to the Secretary, Human Research Ethics Committee, Registrar’s Office, University of Western Australia, 35 Stirling Highway, Crawley, WA 6009 (telephone number 0488-5705). All study participants will be provided with a copy of the Information Sheet and Consent Form for their personal records.
9.7 Appendix G: Information Sheet and Consent Form (Chapters 5, 6 and 7)

Participant information sheet

Study title: How accurate are wearable sensors; are high cost systems necessary?

Investigators

Assoc. Professor Jacqueline Alderson 08 6488 5827 08 9366 1771
jacqueline.alderson@uwa.edu.au a.campbell@curtin.edu.au

Assoc. Professor Peter Kent 08 9366 4992

Mr Daniel Cottam 21470884@student.uwa.edu.au
daniel.cottam@research.uwa.edu.au

Mr Steven Kosovich

Invitation statement

This is an invitation to participate in a research study. Please take your time to read and understand the following information about why the study is being conducted and what it will involve. Do not hesitate to ask us if you need any clarification or if you would like more details. This information sheet will help you decide whether or not to take part in the study.

What is the research study about?

Wearable sensors are increasingly being marketed for their abilities to measure human movement. However, there is limited information on how accurately they can do this. Therefore, we want to perform a study that compares the accuracy of a selection of available small, lightweight wearable sensors for the measurement of trunk and lower limb movements during a range of activities. Cricket players place their trunk under a large amount of stress and most research that measures trunk and lower limb movements has been performed in laboratory environments. However, there would be a number of advantages if this could be measured in real-life circumstances in the field. Therefore, determining if these wearable sensors are accurate enough to be used in the field would have benefit to the cricket community.

Why am I being asked to take part?

We are recruiting male and female cricket players aged between 18 and 45 to take part in this investigation.

Do I have to take part in this research study?

Once you have read this information sheet we are happy to answer any questions you may have. It is then your decision whether or not to volunteer for this project. If you do decide to volunteer, you are free to withdraw from the data collection at any time.

What will I have to do if I take part?

If you agree to take part in this study, you will be asked to attend a single data collection session at the Curtin University Motion Analysis Laboratory. We will measure your height using a stadiometer and your mass using scales. You will then have four small sensors secured to your upper back, lower back, pelvis and lower leg using low allergenic double sided tape. Reflective markers will be secured on top of the sensors and also directly on your feet, pelvis, trunk and arms. During testing the reflective markers must be visible, thus we will ask you to attend data collection in shorts and running shoes. You will be required
to perform the assessment without a shirt, to allow all reflective markers to be seen. After you perform a series of static and dynamic calibration procedures you will be given time to warm up and familiarize yourself with wearing the sensors in the laboratory environment. Then we will ask you to perform a series of movements including trunk bending, sit-to-stand and bending over to pick up a light weight object. You will be asked to repeat each movement five times at a self-selected steady speed. Finally, you will be asked to perform a series of side-stepping trials and bowling trials. These will be repeated until we have five good trials. The entire data collection will be repeated for the three different sensor systems. We anticipate the data collection will take two hours to complete. A video camera will also be used throughout data collection for data processing purposes.

What are the possible risks, inconveniences and any discomfort?
There may be some discomfort when removing the sensors, as the adhesive is similar to a first-aid plaster. You may feel a bit fatigued at the end of the data collection, although the physical requirements are less than those of a typical training session.

What are the possible benefits of taking part?
If you take part in this study, we will provide you a report on your bowling technique. This will include details of mechanisms that have been linked with low back pain.

Will my taking part in this study be kept confidential?
When you attend data collection you will be assigned a participation number. Your data will thus be de-identified from the time of data collection. Only you will be provided a confidential report on your bowling action. All results published or presented will only report group level findings and no individual could be identified from those findings.

What will happen to the results of the research study?
The results of this study will be published in one or more scientific journals and presented at scientific conferences.

What happens next and who can I contact about the research?
If you are interested in volunteering for this project you can contact one of the researchers listed at the top of this form.

Ethics review and complaints
Curtin University Human Research Ethics Committee (HREC) has approved this study (HREC number XX/XXXX). Should you wish to discuss the study with someone not directly involved, in particular, any matters concerning the conduct of the study or your rights as a participant, or you wish to make a confidential complaint, you may contact the Ethics Officer on (08) 9266 9223 or the Manager, Research Integrity on (08) 9266 7093 or email hrec@curtin.edu.au.

Thank you for your interest in this study.
Study title: How accurate are wearable sensors: are high cost systems necessary?

CONSENT FORM

Principal investigator:
Co-investigator:

- I have read the participant information sheet and I understand its contents.
- I believe I understand the purpose, extent and possible risks of my involvement in this project.
- I voluntarily consent to take part in this research project and I can withdraw at any time of the study without complications.
- I have had an opportunity to ask questions and I am satisfied with the answers I have received.
- I understand that my data will be re-identified while confidentiality is maintained at all times to ensure my privacy.
- I understand that this project has been approved by Curtin University Human Research Ethics Committee and will be carried out in line with the National Statement on Ethical Conduct in Human Research (2007).
- I do consent to the storage and use of my information in future ethically-approved research projects related to this study.
- I understand I will receive a copy of this Information Sheet and Consent Form.

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Declaration by researcher: I have supplied an Information Sheet and Consent Form to the participant who has signed above, and believe that they understand the purpose, extent and possible risks of their involvement in this project.

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<th>Researcher Signature</th>
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9.8 Appendix H: Conference Proceedings

The following conference proceedings arose from the work of this thesis. Conference information is presented and followed by the peer-reviewed conference abstract.
Can field-based two-dimensional measures be used to assess three-dimensional lumbar injury risk factors in cricket fast bowlers?

Daniel Cottam\textsuperscript{1}, Helen Bayne\textsuperscript{2}, Bruce Elliott\textsuperscript{1}, Cyril J. Donnelly\textsuperscript{1}, Jacqueline Alderson\textsuperscript{1}

\textsuperscript{1}The School of Sport Science, Exercise and Health, University of Western Australia, Perth, Australia

\textsuperscript{2}Institute for Sport, Exercise Medicine and Lifestyle Research, Section Sports Medicine, Faculty of Health Sciences, University of Pretoria, South Africa

The following abstract arising from the work in Chapter 4 was accepted after peer-review and subsequently presented the 34\textsuperscript{th} International Conference of Biomechanics in Sports in Tsukuba, Japan, July 2016.

https://ojs.ub.uni-konstanz.de/cpa/article/view/7029 (Accessed 11/01/19)

CAN FIELD-BASED TWO-DIMENSIONAL MEASURES BE USED TO ASSESS THREE-DIMENSIONAL LUMBAR INJURY RISK FACTORS IN CRICKET FAST BOWLERS?

Daniel Cottam¹, Helen Bayne², Bruce Elliott¹, Cyril J. Donnelly¹ and Jacqueline Alderson¹

The University of Western Australia, Perth, Western Australia¹
Institute for Sport, Exercise Medicine and Lifestyle Research, Section Sports Medicine, Faculty of Health Sciences, University of Pretoria, South Africa²

Bayne et al. (2016) recently established a direct link between lumbar injury incidence and increased 3D measures of thorax lateral flexion, pelvis rotation and hip extension during the cricket fast bowling action. However, the majority of bowlers are not able to avail themselves to 3D biomechanical analysis. Therefore, we set out to ascertain whether it is appropriate to use 2D measures to assess 3D lumbar injury risk factors in fast bowlers. Nineteen fast-medium bowlers were simultaneously recorded by 3D motion capture and 2D video. Results showed that 2D thorax lateral flexion and pelvis rotation at ball release correlate particularly well with the 3D equivalents. The information presented may be practically applied by coaches to improve field-based lumbar injury risk screening processes.

KEY WORDS: screening, low back injury, thorax lateral flexion

INTRODUCTION: Debilitating lumbar injuries such as spondylolysis and intervertebral disc degeneration are prevalent amongst adolescent and young adult cricket fast bowlers (Johnson, Ferreira, & Hush, 2012). It is widely accepted that bowling technique is a major aetiological factor of injuries to the lumbar region (Bayne, Elliott, Campbell, & Alderson, 2016; Elliott, 2000; Elliott, Hardcastle, Burnett, & Foster, 1992; Foster, John, Elliott, Ackland, & Fitch, 1999; Johnson et al., 2012). Bayne and colleagues (2016) recently reported that adolescent fast bowlers who suffered a lumbar injury over the course of a cricket season exhibited increased levels of thorax lateral flexion (TLF) at front foot contact (FFC) and ball release (BR), pelvis rotation at BR, and hip extension at FFC (Bayne et al., 2016). In the same study, higher peak lumbar flexion-extension and lateral flexion moments were also reported in bowlers who suffered a lumbar injury. Though the link is clearly significant, the kinematic and kinetic variables were identified using a three-dimensional (3D) motion capture system. Retro-reflective motion capture is the current gold standard method of motion analysis, however it is not readily available to most cricket coaches and players, except those at the elite level of the game. Therefore, this research aimed to bridge part of the gap between current biomechanical knowledge of lumbar injury risk and practical application of this research. We hypothesised that it would be possible to reliably replicate the measurement of the four key kinematic variables previously mentioned via two-dimensional (2D) multiple-plane video analysis. Video cameras are comparatively affordable and user friendly, potentially allowing coaches greater opportunity to assess bowling actions for increased risk of lumbar injury before such an injury occurs.

METHODS: Nineteen male fast-medium bowlers (16.6±3.3 years, 182.5±9.5 cm, 72.2±12.9 kg) from district or community level cricket clubs consented to having their bowling actions recorded in the sports biomechanics laboratory at the University of Western Australia. All participants bowled 12 deliveries at match level intensity at a set of wickets, with three 'good' length balls selected for analysis. A 22-camera Vicon motion analysis system (Vicon, Oxford Metrics, Oxford, UK) was used to record 3D trajectory data. A customised, retro-reflective marker set (14mm diameter) and model was applied to the lower limbs, trunk, bowling arm and lumbar spine of all participants (Crewe, Campbell, Elliott, & Alderson, 2013; Dempsey et al., 2007). Static calibration trials collected medial and lateral malleoli positions, with 6-marker pointer calibration trials used to place virtual markers on the lateral and medial epicondyles of the bowling arm and medial and lateral femoral condyles of both lower limbs using the
calibrated anatomical systems technique (Cappozzo, Catani, Della Croce, & Leardini, 1995). Dynamic functional methods were used to determine joint axes of rotation for the bowling elbow and bilateral knee and hip joint centres (Besier, Sturts, Elderson, & Lloyd, 2003). A fourth-order, low-pass Butterworth filter (14Hz cut-off) was applied to the data, with cut-off determined via residual analysis (Winter, 2000).

Video footage was captured in the transverse and sagittal (bowling arm side) planes by two high-speed Vicon Bonita 2D video cameras (250Hz) synchronised to the 3D system. Two Sony Handycam HDR-CX700 50Hz video cameras (Sony Corporation, Tokyo, Japan) were positioned in a coronal plane behind the bowler’s run up and in a sagittal plane on the non-bowling arm side. All camera shutter-speeds were set to the maximum possible speed (1/6000th second) given the ambient light conditions to reduce blur. SiliconCoach Pro 7 (The Tam Group, Dunedin, New Zealand) was used to calculate 2D angles from the video footage. TLF angle was measured from the coronal video (50Hz) at FFC and BR, using markers on C7 and L5. A 0° value indicated a perfectly upright trunk, with a positive value indicating the bowler was leaning towards the non-bowling arm side. Pelvis rotation was calculated at BR using the transverse video (250Hz) and the posterior superior iliac spine markers. A 0° angle indicated the pelvis was exactly parallel to the bowling crease (i.e. front-on position), with a positive value indicating rotation past a perfectly front-on position. Front-leg hip flexion-extension angle at FFC and front-leg knee flexion-extension angle at FFC and BR were calculated from the non-bowling arm side sagittal video (50Hz). The iliac crest, head of the femur, lateral condyle of the femur and the lateral malleolus were used for these angle measurements. BR height was calculated from the sagittal video on the bowling arm side (250Hz). Knee flexion-extension and BR height have been associated with lumbar injury previously (Foster et al., 1995; Portus, Mason, Elliott, Pfistzer, & Done, 2004). All 2D measurements were repeated three times, with the mean value used. Intra-rater reliability had been previously determined by a similar study (Weir, Smale, Alderson, Elliott, & Donnelly, 2013). 2D kinematic variables are displayed in figure 1.

![Figure 1: (1) TLF at FFC, (2) TLF at BR, (3) pelvis rotation at BR, (4) hip flexion-extension at FFC, (5) knee flexion-extension at FFC, and (6) knee flexion-extension at BR.](image)

The front foot of each bowler landed on a 1.2m x 1.2m AMTI force plate (Advanced Mechanical Technology Inc., Watertown, MA) recording at 2000Hz, enabling the FFC event to be determined. BR was determined from the synchronized high-speed video (Wells, Donnelly, Dols, Elliott, & Alderson, 2015). An intra-class correlation (ICC) with absolute agreement was used to compare the 2D and 3D kinematic values from 57 trials. Average measurement difference (º) between the two methods was also calculated for each variable. A one-tailed, bivariate Pearson correlation was used to investigate the association between 2D BR height (normalised to participant standing height) and 3D TLF angle at BR.

RESULTS AND DISCUSSION: Overall, the 2D measurements correlated strongly with the 3D kinematic measurements. The ICC coefficients with absolute agreement and mean 2D-3D measurement differences are displayed in table 1. 2D measurements of TLF at FFC (ICC=0.65) and BR (ICC=0.73) returned strong absolute agreement ICC with the 3D angles. However, 2D TLF at FFC was on average 6.9±4.3° less than 3D TLF at FFC. In comparison, 2D TLF at BR was only 1.6±5.5° less than the 3D values. We speculate that the greater difference at FFC is due to the bowlers’ trunks being in a semi-rotated position at this point. This meant the measurement was taken slightly out of plane with the coronal plane camera.
In comparison, bowlers are in a relatively front-on position by the time they release the ball, making 2D measurement from a coronal camera fairly simple. The measurement error of 2D TLF at FFC may be partially rectified by placing an additional video camera behind the bowling crease at approximately 45°. However, the considerable variation in bowler trunk positions means this variable will be inherently more difficult to measure consistently than TLF at BR, regardless of camera position.

Table 1: 3D and 2D kinematic variable intra-class correlations with absolute agreement and mean measurement differences

<table>
<thead>
<tr>
<th>Kinematic Variable</th>
<th>2D-3D ICC with absolute agreement</th>
<th>Mean measurement difference (degrees)</th>
</tr>
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<tbody>
<tr>
<td>TLF at FFC</td>
<td>0.65*</td>
<td>5.9±5.3</td>
</tr>
<tr>
<td>TLF at BR</td>
<td>0.73*</td>
<td>1.6±5.5</td>
</tr>
<tr>
<td>Pelvis rotation at BR</td>
<td>0.91*</td>
<td>4.5±3.8</td>
</tr>
<tr>
<td>Front-leg hip flexion-extension at FFC</td>
<td>0.56*</td>
<td>0.4±8.3</td>
</tr>
<tr>
<td>Front-leg knee flexion-extension at FFC</td>
<td>0.38*</td>
<td>5.2±7.8</td>
</tr>
<tr>
<td>Front-leg knee flexion-extension at BR</td>
<td>0.97*</td>
<td>1.6±4.0</td>
</tr>
</tbody>
</table>

*Significant at p<0.001

A very strong correlation coefficient was found for pelvis rotation at ball release (ICC=0.91). There was a tendency for the 2D measurements (4.5±3.8°) to be greater than the 3D, but a relatively low standard deviation suggests good measurement repeatability. This was the only one of the four key variables to be measured from the 250Hz video. A major benefit of the 250Hz video was that FFC and BR could be matched exactly with the 3D data. It is probable that some of the 2D to 3D measurement differences in the other variables can be contributed to measurements being observed at slightly different time points (i.e. not precisely at BR or FFC) due to the 50Hz frame rate.

Front-leg hip flexion-extension at FFC produced a comparatively weak ICC (0.56) when compared with the other key kinematic variables identified by Bayne and colleagues (2016). It also returned the largest standard deviation in measurement difference (0.4±8.3°). Front-leg knee flexion was also measured at FFC and BR. Surprisingly, knee flexion at FFC produced only a moderate ICC of 0.38, whereas the same measurement at BR correlated extremely well with the 3D values (ICC=0.97). Average measurement difference was also much greater at FFC (5.2±7.8°) than at BR (1.6±4.0°). The comparatively weak correlations and large 2D to 3D measurement differences for both knee and hip flexion-extension at FFC may be due to the rapid knee flexion or extension that occurs just after a bowler’s front foot makes contact with the ground. As these 2D measurements were taken from the 50Hz video, it is likely that the FFC frame differences between the 50Hz video and the 250Hz 3D data have significantly impacted the hip and knee angle measurements at FFC. The flexion-extension angular velocity at the knee slows considerably by BR, hence the much stronger correlation between the 3D and 2D measurements.

A strong Pearson correlation of -0.67 (p<0.001) was found between 2D BR height and 3D TLF at BR. We suggest that coaches can utilise BR height as a secondary measurement of TLF at BR, when looking to identify bowlers at an increased risk of lumbar injury. A BR height less than 110% of standing height may suggest greater TLF. In this study, those bowlers (n=11) who averaged a BR height of less than 110% of their standing height had a mean 3D TLF angle at BR of 51.3±6.8°. Significantly (p=0.01), those who had a BR height greater than 110% of standing height (n=8), had a TLF angle of 42.5±4.1°. For TLF at BR, <50° is considered to be in the higher lumbar injury risk range (Bayne et al., 2016).

These findings suggest 2D video analysis is an appropriate method of lumbar injury risk measurement in fast bowlers. The following summary points may be useful to coaches, players and other cricket researchers:
1. 2D measurement of TLF is more repeatable at BR than at FFC.
2. 2D BR height <110% of standing height may also suggest increased levels of TLF.
3. An additional video camera placed behind the bowling crease at approximately 45° may facilitate a more accurate measurement of TLF at FFC than a coronal camera.
4. 2D pelvis rotation can be accurately measured from a transversely positioned camera.
5. Front-leg hip and knee flexion-extension 2D angle measurement may be impacted by the rapid flexion or extension at the knee following front-foot ground contact.
6. Video cameras with higher frame rates are preferable for measuring 2D angles.

CONCLUSION: Here we have presented evidence supporting the use of video-based 2D measures to assess 3D kinematic lumbar injury risk factors in cricket fast bowlers. The strong correlations and repeatability between the 2D and 3D measures suggest that multiple-plane video analysis is an appropriate and feasible method of measuring 3D kinematics associated with lumbar injury risk. We suggest that cricket coaches should practically apply these findings to lumbar injury risk screening protocols.

REFERENCES:
Can inertial measurement units be used to measure pelvis and thorax motion during cricket bowling?

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CAN INERTIAL MEASUREMENT UNITS BE USED TO MEASURE PELVIS AND THORAX MOTION DURING CRICKET BOWLING?

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Identifying lumbar injury risk amongst cricket bowlers is a challenge to those involved in the sport. Bowling technique injury risk factors concerning thoracic and pelvic motion have been identified by previous research that used three-dimensional (3D) retro-reflective (RR) motion analysis. Inertial measurement units (IMUs) are considered a feasible and more portable means of 3D motion analysis. However, the validity of IMU measurement of thorax and pelvis movement during bowling has not yet been fully determined. This study aimed to achieve this by comparing concurrent IMU and RR angle outputs. Results suggest that when RR coordinate systems are aligned with the IMUs' there are no significant differences in cricket bowling relevant angle outputs. However, some differences arise when IMUs are compared to the anatomically derived RR angle outputs typically used in 3D analysis.

KEY WORDS: Inertial sensors, lumbar injury, injury screening

INTRODUCTION: Cricket fast bowlers are significantly more likely to suffer debilitating lumbar injuries than the general population (Johnson, Ferreira, & Hush, 2012). Bowling technique factors have been previously related to injury incidence (Bayne, Elliott, Campbell, & Alderson, 2016; Olivier et al., 2016). Many of these factors involve thorax and pelvis kinematics. Excessive rotation of the shoulders away from the bowling direction (shoulder counter-rotation) and shoulder-pelvis separation (SPS) at the start of the delivery stride, were linked to injury incidence in early prospective studies on bowling (Foster, John, Elliott, Ackland, & Fitch, 1989). More recent prospective studies have associated thorax lateral flexion (TLF) contralateral to the bowling arm, excessive pelvis rotation, and reduced front-leg hip flexion with increased injury risk (Bayne et al., 2016). Injury risk thresholds for the majority of these variables were identified by studies that used retro-reflective (RR) three-dimensional (3D) motion analysis (Bayne et al., 2016). Though reported to have sub-millimetre accuracy (Windolf, Gotzen, & Morlock, 2008), RR motion capture is unavailable to most bowlers and has poor ecological validity. Consequently, alternative analysis technologies are an attractive proposition to cricket coaches for lumbar injury risk screening purposes. Inertial measurement units (IMUs) combine accelerometers, gyroscopes and magnetometers to measure 3D motion. Unlike RR systems, they are relatively affordable, portable and easy to use, making them ideal for field use. Unfortunately, IMU literature has typically focused on validating IMU angle measurement for simple motions such as uniplanar movement or gait (Lopez-Nava & Angelica, 2016). IMU measurement validity for thorax and pelvis kinematics is unknown for cricket bowling. Therefore, this research aimed to assess whether IMUs are able to validly measure the high-speed, multi-planar thoracic and pelvic movements exhibited during cricket bowling. Concurrent IMU and RR angle measures were compared during the bowling action. RR coordinate systems aligned to the IMU coordinate systems, as well as anatomically defined segment coordinate systems were created to allow dual comparison. It was hypothesised that when the 3D coordinate systems of the IMUs and RR systems were aligned there would be no significant differences in angle outputs. However, some differences were expected when IMU angles were compared to anatomically defined angles that are typical of RR 3D modelling.

METHODS: Seven asymptomatic male (183.2±7.7cm, 75.3±9.3kg, 26.1±8.6 years) and three asymptomatic female (173.8±6.2cm, 66.7±4.9kg, 18.3±4.2 years) state or club level
fast/medium bowlers agreed to participate in the study. Informed consent was obtained before data collection commenced. Data collection was completed in an indoor biomechanics laboratory. Ethical approval was granted prior to the commencement of the study. Three Xsens Mti Awinda model IMUs (Enschede, Netherlands) (75Hz sample, ± 2000 deg/s gyroscope, ±160m/s² accelerometer, ±1.9 Gauss magnetometer) were each placed on the thorax, pelvis, and shank. The thorax sensor was placed with its superior edge between the spinous processes of the 7th cervical (C7) and 1st thoracic (T1) vertebrae. The pelvis IMU had its inferior edge on the spinous process of the second sacral vertebrae, and the shank sensor was approximately 5cm superior to the lateral malleolus, ipsilateral to the bowling arm. IMU data was captured by the manufacturer’s software (MT Manager 4.2.1, Xsens Technologies).

RR marker trajectories were recorded using a 300Hz, 20 camera Vicon system (Oxford, UK). The IMUs were overlayed by three-marker rigid plates (figure 1), allowing creation of a RR technical coordinate system (RRtech). The RRtech orientation and the IMUs’ coordinate system (IMUtech) orientations were aligned. The RRtech origin was the mean position of the three markers, with the first and second defining lines being M2 to M1 and M2 to M3 respectively (figure 1). The first defining line was the y-axis, with the x-axis the cross-product of the first and second defining lines, and the z-axis perpendicular to the y and x axes. A customised marker set and model was used to anatomically model the shoulders, thorax and pelvis. (Campbell, Lloyd, Alderson, & Elliott, 2009; Dempsey et al., 2007). An upper thorax segment was created for measurement of shoulder counter-rotation (SCR) and SPS (Middleton, Foster, & Alderson, 2016). Anatomically defined segment coordinate systems are henceforth referred to as RRanat. Participants performed five bowling trials at match intensity, with the two trials of best data quality selected for analysis (determined by visual inspection). Two AMTI (Advanced Mechanical Technology Inc., Watertown, MA) force plates (1800Hz) captured back foot contact (BFC) and front foot contact (FFC) of the bowling stride. A 300Hz Vicon Bonita camera (Oxford, UK) synchronised with the RR system, was placed sagittal to a bowling crease marked on the second force plate. It was used to identify ball release (BR) and post-data collection. Participants performed a calf-raise at the start of each trial to facilitate post-processing temporal synchronisation. IMU data was processed using in-built Kalman filters to calculate IMU orientations from accelerations and angular velocities. Relative and absolute IMU angles were output as quaternions using the Xsens MT Manager software (Enschede, Netherlands). Prior to modelling, RR trajectories were filtered by a 4th order, low-pass Butterworth filter, with a 15Hz cut-off determined by residual analysis (Winter, 1990). BFC to FFC was termed the delivery step, with delivery stride from BFC to BR.

IMU and RR data was processed further and output by a customised program developed in LabVIEW 2017 (National Instruments Corp. Austin, Texas). Relative and global IMU angles were produced from the quaternion outputs from each individual IMU. Phases were time normalised to 101 data-points using cubic spline interpolation. RR and IMU joint angles were output as Euler angles by following the Grood and Suntay method (Grood & Suntay, 1983). One-dimensional statistical parametric mapping (1DSPM) incorporating a paired t-test (p=0.05) (Pataky, Robinson, & Vanrenterghem, 2013) was used to test angular differences between IMUtech and RRtech, and IMUtech and RRanat for the variables of interest. SCR (global upper thorax rotation angle) and SPS (relative rotation angle between the upper thorax and pelvis) were assessed during the delivery step. Pelvis rotation (global pelvis angle), TLF (global TLF angle), thorax-to-pelvis flexion-extension (relative angle between the thorax and pelvis), and thorax-to-pelvis lateral flexion (relative angle between the thorax and pelvis) were all assessed during the delivery stride.

RESULT AND DISCUSSION: The 1DSPM measurement comparison between IMUtech and RRtech showed no significant angle differences for any of the variables assessed. This suggests that when coordinate systems are comparable, IMUs have good 3D measurement validity compared with RR systems for dynamic, multi-planar movements. The 1DSPM
analysis of IMUtech vs RRanat did show significant differences for some variables. Shoulder rotation measurements were significantly different from 0-4% (p=0.045) and 92-100% (p=0.046) of the delivery step (Figure 2a), however SPS measures were not different during the same phase. Pelvis rotation measures were not significant different, however the other three delivery stride variables all displayed discrepancies. IMUs underestimated global TLF when compared to RRanat, with significant differences for 100% of the delivery stride (p<0.01) (Figure 2b). Significant differences were also seen for thorax-to-pelvis flexion-extension at 0-13% (p<0.01) and 30-88% (p=0.048) of the delivery stride (Figure 2c), and thorax-to-pelvis lateral flexion at 49-73% (p<0.01) and 75-96% (p<0.01) of the delivery stride (Figure 2d).

Figure 2: 1D SPM for IMUtech (black/grey) versus RRanat (red). Time-varying angles with standard deviation clouds plus SPM graphs are shown for factors with significant differences:
(a) SCR, (b) TLF, (c) thorax-to-pelvis flexion-extension, (d) thorax-to-pelvis lateral flexion.
Significance (p<0.05) was reached when the t-threshold was crossed on the SPM graph.

Given no significant differences were found between IMUtech and RRtech measures, our findings suggest that pelvis and thorax angle measurement differences between IMUs and RR systems are likely due to segment coordinate system modelling differences. The findings suggest that IMUs are capable of validly measuring dynamic, multi-planar movements, but the IMU movement may not reflect movement of the whole body segment. The inherent errors associated with RR motion capture, and IMU motion capture to a lesser extent, may also contribute to discrepancies in measurement. Marker placement errors and movement artefacts can contribute to misrepresentations of joint centre positions and segment orientations during anatomical modelling (Reinschmidt, Van den Bogert, Nigg, Lundberg, & Murphy, 1997; Taylor et al., 2005). The employed assumption that body segments are rigid entities is also not always accurate. Significant differences were displayed for IMUtech to RRanat comparisons involving the thorax segment. The thoracic spine is comprised of 12 vertebrae that do not move as a single unit, therefore expecting an IMU to represent the movement of the entire thorax may be unrealistic. The SJC were also used to define the upper thorax segment. SJC estimation is known to become less reliable during humeral elevation (Campbell, Alderson, Lloyd, & Elliott, 2009); a movement exhibited at high velocities during bowling. These examples may help to explain why significant differences were exhibited for IMUtech to RRanat comparisons relating to the thorax but not for pelvis rotation. The pelvis is an almost rigid segment and hence is likely to be less susceptible to some of the errors presented. The findings suggest that IMUs are capable of measuring high-speed, multi-planar movement validly, with measurement
differences between IMU and RR methods likely attributable to coordinate system differences. This may be an illustration of the inherent flaws of RR anatomical modelling techniques, or alternatively it may suggest IMU kinematics do not truly represent body segment motion. We speculate that both explanations contribute to the measurement differences reported.

Future work will evaluate dynamic, multi-planar measurement validity of IMUs from other manufacturers. IMU measurement validity for movements of other body segments during bowling is also of interest. It may be prudent for cricket researchers to establish new IMU-derived thresholds for the injury risk factors discussed in this paper. This would enable cricket coaches to screen bowlers for 3D lumbar injury risk factors by using IMUs, an affordable and portable kinematic measuring tool.

CONCLUSION: Here we have presented evidence supporting the use of IMUs to validly measure high-speed, multi-planar movements, such as those displayed during cricket bowling. Modelling differences between the IMU coordinate systems and the anatomically defined body segment coordinate systems used in RR 3D motion capture appear to be the main cause of statistical differences for 3D angular measures of thorax and pelvis motion during bowling. Nonetheless, these findings suggest that valid field-based 3D kinematic screening for lumbar injury risk factors in cricket bowlers is a sensible and feasible aim.

REFERENCES: