MEASURING ELBOW KINEMATICS IN CRICKET BOWLING

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MEASURING ELBOW KINEMATICS IN CRICKET BOWLING

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Whatever I was able to acquire in my life by way of acts visible to all, that is, to win my own transparency, I owe to a kind of special courage Poetry gave me: to be wind for the kite and kite for the wind, even when the sky is missing. I’m not playing with words. I mean the movement you discover being written in an “instant,” when you can open it and make it last. When, in fact, Sorrow becomes Grace and Grace Angel; Joy Alone and Sister Joy with white, long pleats over the void, a void full of bird dew, basil breeze and a hiss of resonant Paradise.

The little Mariner

Odysseas Elytis

(trans: Olga Broumas)
Abstract

In the sport of cricket the objective of the ‘no-ball’ law is to allow no performance advantage through elbow extension during ball delivery. Since the advent of high-speed video photography it has been revealed that some straightening occurs in bowlers who have actions that are traditionally considered in accordance with the law. Measuring the three-dimensional movement of the elbow is vital when assessing bowling legality in cricket. However, the elbow joint is a complex structure with a remarkable range of motion and tracking its movement through skin-based techniques can be highly erroneous due to the thick layer of skin overlying the joint.

Within this work, a biomechanical model was mathematically developed and experimentally validated to assess bowling legality in cricket. The new model meets all of the specifications of a measurement method to be used in sports-related biomechanical studies for non-invasive measurement of joint kinematics at high speeds whilst allowing for the subject to move freely within a large volume. The model was compared with existing methods via a series of sensitivity analyses and was found to significantly improve repeatability compared to available elbow measurement techniques particularly in measuring subtle elbow rotations, such as elbow abduction and forearm pronation. In addition this model can be easily implemented within the existing experimental protocol for assessing bowling legality in cricket as proposed by the England and Wales Cricket Board and will be used in future clinical and sport-related studies.
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<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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</thead>
<tbody>
<tr>
<td>ADL</td>
<td>Activity of daily living</td>
</tr>
<tr>
<td>AnatM</td>
<td>Anatomical Based Kinematic model 1 where the positions of the two elbow epicondyles are digitised</td>
</tr>
<tr>
<td>AnatM2</td>
<td>Anatomical Based Kinematic model 2 where markers are placed directly onto the two elbow epicondyles</td>
</tr>
<tr>
<td>BR</td>
<td>Ball Release</td>
</tr>
<tr>
<td>CA</td>
<td>Carrying Angle</td>
</tr>
<tr>
<td>CoR</td>
<td>Centre of rotation</td>
</tr>
<tr>
<td>CMC</td>
<td>Coefficient of Multiple Correlations</td>
</tr>
<tr>
<td>DoF</td>
<td>Degrees of freedom</td>
</tr>
<tr>
<td>ECB</td>
<td>English and Wales Cricket Board</td>
</tr>
<tr>
<td>FHA</td>
<td>Finite Helical Axis</td>
</tr>
<tr>
<td>FuncM1</td>
<td>Functional Based Kinematic model 1</td>
</tr>
<tr>
<td>FuncM2</td>
<td>Functional Based Kinematic model 2 whereby a common flexion axis is shared between the upper arm and forearm</td>
</tr>
<tr>
<td>GH</td>
<td>Glenohumeral centre of rotation</td>
</tr>
<tr>
<td>ICC</td>
<td>International Cricket Council</td>
</tr>
<tr>
<td>IHA</td>
<td>Instantaneous Helical Axis</td>
</tr>
<tr>
<td>ISB</td>
<td>International society of biomechanics</td>
</tr>
<tr>
<td>LE</td>
<td>Lateral epicondyle</td>
</tr>
<tr>
<td>MCC</td>
<td>Marylebone Cricket Club</td>
</tr>
<tr>
<td>ME</td>
<td>Medial epicondyle</td>
</tr>
<tr>
<td>RoM</td>
<td>Range of motion</td>
</tr>
<tr>
<td>STA</td>
<td>Soft Tissue Artefact</td>
</tr>
<tr>
<td>UH</td>
<td>Upper Arm Horizontal</td>
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</table>
1.1 Motivation

In the sport of cricket the objective of the ‘no-ball’ law is to restrain the bowler’s arm in a straight position to allow no performance advantage through elbow flexion or extension during ball delivery. Since the advent of high-speed video photography it has been revealed that some straightening occurs in bowlers who have actions that are traditionally considered in accordance with the law. For this reason the International Cricket Council (ICC) in the 1990s decided to allow a small amount of elbow extension to occur, which in 2004 was extended to $15^\circ$ for elbow extension for all types of bowlers.

Measuring the movement of the elbow is vital when assessing bowling legality in cricket. The methods that are currently used to measure the 3D motion of the bowling arm and conclude whether a bowling action is legitimate or not, are optical motion tracking systems, where markers are attached to the bowling arm of bowlers whose bowling action is subsequently recorded. Different laboratories have employed different models to analyse these actions but as the accuracy of the predictions of these models depends highly on the accuracy and repeatability of the input data the results from these laboratories may not be directly comparable.

Although elbow kinematics have been studied in order to define the functional range of motion in activities of daily living (ADL’s) (Sardelli et al., 2011) and compare normal and pathological elbows, its function during active dynamic activities is not yet fully understood. This is reflected in the design of elbow replacement implants where clinical results have been disappointing so far (Paraskevas et al., 2004). The difficulties in assessing elbow kinematics arise from a number of factors including the absence of a standard repeatable motion as in gait for the lower-limb and the large range of motion of the joint especially under highly dynamic activities which is obscured at the same time by overlying soft tissue obstructing access to bone movements. Some of these issues have been addressed in part in the literature and several compensation techniques have been proposed. However, no method currently exists that can accurately and precisely measure the dynamic movement of the elbow.
1.2 Main aim and thesis layout

The aim of this research is to build a biomechanical model to measure the 3D in-vivo kinematics of the elbow joint during cricket bowling which can be employed in cases where the legality of the action of a player is challenged, ensuring that the bowling review process is conducted in a fair and consistent manner with respect to both the data acquisition and analysis. This model can be later employed to examine the scientific evidence behind new regulations of the International Cricket Council (ICC) but also in clinical studies to measure the range of motion of healthy and pathological elbows.

The realisation of the aims of this study demanded the understanding of the function and motion of the elbow joint therefore, in Chapters 2, 3 and 4 a thorough literature review on the anatomy and physiology of elbow and the role of the joint during cricket bowling are presented, with focus on the kinematics of the joint. The mathematical model is then developed to achieve the aim stated above, and the method is finally tested and the errors quantified in a series of studies presented in Chapters 5 - 7.

In detail, this thesis is divided into eight chapters:

Chapter 1: Introduction

This chapter describes the motivation behind this work and the main aim of the project.

Chapter 2: Clinical Biomechanics of the Elbow - Literature Review

In order to understand the movement of the joint during cricket bowling a literature review of the elbow’s anatomy and function is presented in this chapter.

Chapter 3: The biomechanics of cricket bowling

In this chapter a review of the current knowledge on the biomechanics of the elbow during cricket bowling is conducted.
Chapter 4: Tracking Elbow Movement - Literature Review

The review consists of a brief summary of the available measurement techniques of joint kinematics with a focus on tracking elbow movement under dynamic activities.

Chapter 5: Kinematic Model of the Elbow Joint

In this chapter the development of a motion derived kinematic model is mathematically described and experimentally validated.

Chapter 6: Sensitivity Analysis

In this chapter the model’s sensitivity to anatomical based axes, single and double anatomical landmark calibration, markers placed directly onto the epicondyle is assessed.

Chapter 7: The carrying angle

As the presence of the carrying angle has been directly associated with the illusion of a throw in cricket bowling the aim of this chapter is to measure the carrying angle of cricket bowlers and evaluate its variability during flexion and pronation of the elbow joint.

Chapter 8: Summary, Conclusions and Future Work

The final chapter is a summary of the main results, limitations of the study, possible areas of application and areas of future work. A protocol is also proposed for the bowling review process which can be used in cases where the legality of the action of a player is challenged, ensuring that the Bowling Review Process is conducted in a fair and consistent manner with respect to both the data acquisition and analysis.
The main focus of this thesis is to develop a method to accurately measure the three-dimensional dynamic movement of the elbow joint during cricket bowling. In order to understand the movement of the joint in bowling a literature review of the current knowledge on the clinical biomechanics of the elbow joint is presented in this chapter.
2.1 Introduction

The elbow joint is a complex and highly mobile joint located in the middle of the arm. The elbow acts as a mechanical link within the upper extremity transferring forces and assisting both the shoulder and wrist in positioning the hand in space (Lockard, 2006). In sports activities and especially in overhand throwing motions the elbow is subject to and the subject of significant generated force (Fornalski et al., 2003). Appropriate throwing mechanics can enable an athlete to achieve maximum performance minimising the risk of injury at the same time (Ferdinands and Kersting, 2007). In cricket bowling elbow biomechanics plays a crucial role in the delivery swing as the laws of the game prohibit elbow flexion-extension during the arm acceleration of bowling.

Knowledge of both the anatomy and biomechanics of the elbow is of vital importance, and may provide essential information in the analysis of clinical disorders, athletic injuries and rehabilitation protocols as loss of elbow function can severely affect activities of daily living (Fornalski et al., 2003). The literature review presented in this chapter will provide a brief introduction to elbow anatomy and physiology with the main focus being the biomechanics of the articulations of the joint and the components that contribute to joint stability.

2.2 Anatomy and Physiology of the Elbow Joint

The elbow joint is a hinge joint that consists of three bones; the humerus, radius and ulna constrained at three articulation: humeroulnar, humeroradial and proximal radioulnar. In order to produce the motions necessary for normal functioning of the elbow complex the three articulations act along with muscle forces and ligament constraints in a way that allows the elbow its range of motion. Stability is conferred to the elbow joint equally by its bony architecture and ligaments (Lockard, 2006). A brief description of the elbow's main features, range of motion and stability is presented in this chapter.
2.2.1 Elbow Bones

2.2.1.1 Humerus

The humerus is the longest and largest bone of the upper limb; it articulates proximally with the scapula at the glenohumeral joint and distally with two bones, the ulna and the radius at the elbow joint (Tortora and Derrickson 2006).

The proximal end of the humerus features a rounded head which forms approximately one third of a sphere and articulates directly with a small shallow depression on the scapula, called the glenoid fossa to form the glenohumeral joint as shown in Figure 2-1 (Tortora and Derrickson 2006). Just distal to the head is the anatomical neck which is visible as an oblique groove and separates the head from the two tubercles (Lockard, 2006). The greater tubercle is a lateral projection distal to the anatomical neck while, the lesser tubercle is an anterior projection (Tortora and Derrickson 2006). Between the two tubercles there is a bicipital groove. Distal to the tubercles is the narrow surgical neck that is a constriction in the humerus where the head tapers to the shaft (Tortora and Derrickson 2006).

Figure 2-1: The glenohumeral joint (Modified from: http://biologycorner.com/anatomy/)
The shaft of the humerus is roughly cylindrical at its proximal end but it gradually becomes triangular until it is flattened at its distal end (Figure 2-2) (Tortora and Derrickson 2006). The shaft has two prominent features: the deltoid tuberosity and the radial groove that lie laterally and posteriorly respectively. The deltoid tuberosity is a roughened v-shaped area at the middle of the shaft that serves as a point of attachment for the tendons of the deltoid muscle (Tortora and Derrickson 2006). The radial groove serves as a point of attachment for the radial nerve and deep artery of the arm (Tortora and Derrickson 2006).

The distal end of the humerus resembles a triangle with the medial column terminating in the medial epicondyle, medial condyle, and trochlea and the lateral column terminating in the lateral epicondyle, lateral condyle, and the capitulum (Lockard, 2006). The two epicondyles are rough projections on either side of the distal end of the humerus that can be easily palpated and serve as attachment points for the collateral
ligament complexes (Tortora and Derrickson 2006). The distal humerus continues beyond the epicondyles to terminate in two articular surfaces: the trochlea and the capitulum. The capitulum is an articular surface, hemispherically shaped on the lateral aspect of the bone and the trochlea a pulley-like structure on the medial aspect of the bone (Guerra and Timmerman, 1996). Superior to the capitulum the radial fossa can be identified while just superior to the articular surface of the trochlea is an anterior depression termed the coronoid fossa.

### 2.2.1.2 Ulna

The ulna, usually referred to as the stabilising bone of the forearm, is located on the medial aspect and is the longer of the two bones of the forearm (Lockard, 2006). As shown in Figure 2-3 its proximal end has two prominent projections that form the walls of the trochlear notch; the olecranon process and the coronoid process (Guerra and Timmerman, 1996). On the lateral side of the coronoid process is a shallow depression termed the radial notch. Just inferior to the coronoid process is the ulnar tuberosity to which the biceps brachii muscle attaches (Tortora and Derrickson 2006).

At its narrow distal end, the ulna consists of a rounded head with a conical styloid process located on the posterior side separated from the wrist by a disc of fibrocartilage (Tortora and Derrickson 2006). The ulnar styloid process provides attachment for the ulnar collateral ligament to the wrist (Tortora and Derrickson 2006).

### 2.2.1.3 Radius

The radius is the shorter of the two forearm bones and is located at the lateral aspect of the forearm (Tortora and Derrickson 2006). Its proximal end consists of a cylindrical head that articulates with the capitulum of the humerus and the radial notch of the ulna (Tortora and Derrickson 2006), a constricted neck and a projection from the medial surface called the radial tuberosity. The radial tuberosity provides a site of attachment for the tendons of the biceps brachii muscle (Guerra and Timmerman, 1996). The radial head’s lateral aspect terminates distally at the radial styloid process that is considerably larger than the ulna and that extends further distally providing attachment for the
brachioradialis muscle and for the radial collateral ligament to the wrist (Tortora and Derrickson 2006).

2.2.2 Elbow Joints

The elbow joint complex consists of three articulations: the humeroulnar, the humeroradial and the proximal radioulnar joints. The humeroulnar joint occurs between the articulation of the trochlea of the humerus and the trochlear notch of the ulna. The humeroradial joint occurs between the capitulum of the humerus and the posterior aspect of the radial head. Finally, the ulna and radius articulate directly at their proximal and distal ends forming the proximal and distal radioulnar joints respectively (Lockard, 2006; Tortora and Derrickson 2006). In terms of physiology both the humeroulnar and the humeroradial joints are described as modified hinge joints permitting motion only in one plane while, the proximal radioulnar joint is described as a pivot joint allowing for rotation of the forearm and hand (Fornalski et al., 2003).
2.2.3 Elbow Movement and Range of Motion

Two main movements as shown in Figure 2-4 are possible at the elbow joint; flexion-extension and pronation-supination. Flexion-extension occurs at the humeroulnar and humeroradial joints and can be defined as the bending and straightening of the elbow about an axis that passes through the centers of the trochlea and capitulum (Lockard 2006). Because the capitulum lies on the anterior aspect of the distal humerus, during extension the radius articulates with only a portion of the capitulum but contact increases in flexion (Lockard, 2006). Studies have estimated that the humeroradial articulation bears about 60% of axial loads placed across the elbow during extension (Andrews et al., 1993; Fox et al., 1995). An abnormal movement beyond the normal limit of extension, is called hyperextension (Tortora and Derrickson 2006). Hyperextension of the elbow is usually prevented by the arrangement of ligaments and the anatomical alignment of bones (Tortora and Derrickson 2006).

Figure 2-4: Elbow movements: (A) Flexion-extension and (B) Pronation-supination (Luttgens and Hamilton 1977)
Flexion-extension movements usually occur in the sagittal plane. Clinical studies of elbow joint movement have shown that slight mobility of the humeroradial and humeroulnar joints also occurs at the frontal and transverse planes, in the forms of abduction-adduction, and medio-lateral rotation about the ulna, respectively (Lockard, 2006).

Pronation-supination can be defined as the action of turning the forearm and occurs between the radius and the ulna. Supination is the movement of the forearm at the proximal and distal radioulnar joints in which the palm is turned anteriorly from the anatomical position while pronation is the movement of the forearm where the palm is turned posteriorly from the anatomical position as the distal end of the radius crosses over the distal end of the ulna (Lockard, 2006). However, pronation can occur without displacement of the hand suggesting that the distal ulna moves radially as the radius rotates around it in order to keep the hand in a fixed position (Lockard, 2006). Anatomically, both pronation and supination movements occur about an axis passing from the centre of the radial head to the fovea of the distal ulna (Holdsworth and Mossad, 1990). With respect to the humerus, external rotation of the ulna occurs with supination and internal rotation occurs with pronation (Pomianowski et al., 2001).

The normal range of motion at the elbow joint is 140° of flexion from full extension or from -5° of hyperextension and from 75° of pronation to 85° of supination (Guerra and Timmerman, 1996). However, the functional range of motion, that is the range of motion required to perform the activities of daily living, has been reported to range from 30° to 130° of flexion and 50° of supination and pronation (Holdsworth and Mossad, 1984; Morrey and An, 1983).

2.2.3.1 The Carrying Angle

When the human arm is completely extended with the palm facing up then the humerus and the paired radius and ulna are not perfectly aligned. This deviation from a straight line that occurs in the direction of the thumb is referred to as the ‘carrying angle’ (Steel and Tomlison, 1958) as shown in Figure 2-5. Anatomically, this valgus angulation is
present because the trochlea extends farther distally than the capitellum (Zampagni et al., 2008b).

The carrying angle varies as a function of age and gender with females reported to have on average 3° to 4° more than males. The normal distribution of this angle varies greatly and averages 10° valgus in male subjects and 13° valgus in female subjects (Van Roy et al., 2005). The role of the carrying angle is quite significant in throwing athletes since they are reported to have increased carrying angles in the elbow joint of their bowling arm adapting in that way to repetitive stress (Zampagni et al., 2008). Increasing carrying angle may lead to elbow instability and pain in throwing activities and predispose to risk of elbow dislocation (Chang et al., 2008). Cricket bowlers have also been reported to have large carrying angles (Aginsky and Noakes, 2010). Because of the presence of such a large carrying angle when viewing the delivery from only one position the visual illusion of a ‘throw’ (or, illegal movement of the elbow) may be created. The latter is of great importance when assessing bowling legality in cricket and will be discussed in greater detail in Chapter 7.
2.2.4 Active and Passive Stabilisers of the Elbow Joint

Given the complexity of the elbow, the shape of the articulating surfaces and the capsuloligamentous structures along with muscle length and strength are critical factors in determining normal movement of the joint.

The articular geometry of the radial head and the olecranon, and coronoid processes are the primary stabilisers of the elbow against varus and valgus stress at less than 20° and more than 120° of flexion (Guerra and Timmerman, 1996). The groove formed between the trochlea and capitulum articulates directly with the rim of the radial head thus contributing to stability of the humeroradial articulation (Lockard, 2006). The distal end of the humerus fits perfectly with the proximal end of the ulna restricting gliding of the joint in the medial and lateral directions during flexion-extension (Lockard, 2006). Articular contact within the joint during movement is highly affected by the position of the forearm as the amount of translation that accompanies rotation varies and the weight-bearing loads are not equally distributed across the articular surfaces (Lockard, 2006). During extension the radial head has no contact with the capitulum while in flexion as the radial head moves proximally contact with the capitulum increases. Forearm rotation also affects the contact between the radius and ulna in the sense that contact decreases during supination and increases during pronation (Gordon et al., 2006; Lockard, 2006).

In the dynamic range of motion (20–120° of flexion), soft tissue structures are responsible for primary stability of the elbow. The passive soft tissue stabilisers of the elbow joint are primarily the collateral ligament complexes as well as the anterior joint capsule and the interosseous membrane (IOM) (Seiber et al., 2009).

The joint capsule consists of dense fibrous connective tissue made of bundles of collagen fibres surrounding the joint extending from the margin of the articular surface of the humerus to the radial notch of the ulna and the circumference of the head of the radius. Projecting between the radius and ulna into the cavity is a crescentic fold of synovial membrane that suggests the division of the elbow into the humeroradial and
humeroulnar joints (Nielsen and Olsen, 1999). The elbow capsule is loose both anteriorly and posteriorly to allow for full elbow extension and flexion.

Ligaments are bands of fibrous, dense connective tissue. Capsular ligaments are thin bands of tissue whose primary aim is to retain the synovial fluid within the capsule acting as mechanical reinforcements. Extra-capsular ligaments are usually thick, tough bands of tissue binding the bones together, preventing their displacement during motion. Finally, intra-capsular ligaments provide stability like the extra-capsular ones but permit a far larger range of motion (Lockard, 2006).

The elbow has ligamentous structures on both its medial and lateral aspects. The ulnar or medial collateral ligament (MCL) complex consists of three distinct structures: the anterior, posterior, and transverse ligaments as shown in Figure 2-6. The anterior medial collateral ligament is a thick, discrete band with parallel fibers originating from the medial epicondyle and inserting directly onto the medial aspect of the coronoid process. It is the primary stabiliser of the elbow providing approximately 70% of the valgus stress. The posterior ligament is a fan-shaped capsular ligament that originates from the medial epicondyle and inserts onto the medial margin of the trochlea notch and is best defined when the elbow is flexed at 90° (Guerra and Timmerman, 1996). The transverse ligament also known as the ligament of Cooper (Safran and Baillargeon, 2005), originates from the medial olecranon and inserts directly onto the inferior medial aspect of the coronoid process. Even though the transverse ligament does impart some medial support to the trochlea notch it is not considered a major contributor to elbow stability (Guerra and Timmerman, 1996; Lockard, 2006).

Similarly to the MCL, the lateral collateral ligament (LCL) complex is composed of four bundles: the annular ligament, the radial collateral ligament, the accessory lateral collateral ligament and the lateral ulnar collateral complex. The LCL complex protects the joint from varus loads throughout the range of movement restraining excessive postero-lateral rotation of the forearm on the humerus (Lockard, 2006). The annular ligament is a strong band of tissue that encircles the radial head and attaches to the anterior and posterior margins of the radial notch on the ulna maintaining the
articulation between the proximal ulna and radius (Guerra and Timmerman, 1996). The lateral ulnar collateral ligament extends from the lateral epicondyle to the ulna instability in the postero-lateral direction (Guerra and Timmerman, 1996). Finally, the radial collateral ligament is a short and narrow fibrous band that runs from a depression below the lateral epicondyle to the annular ligament (Guerra and Timmerman, 1996).

Clinical studies in human cadavers have shown that the structures that resist varus-valgus forces contribute different amounts depending on the angle of elbow flexion. For instance when a valgus force is applied to the elbow with the joint flexed to 90° then the medial collateral ligament (MCL) provides 54% of the valgus stability, the osseous structures provide 36% and the capsule 10%. In full extension, valgus stability is equally distributed among the ulnar collateral ligament, the bony articulations, and the capsule (Safran and Baillargeon, 2005). On the other hand, resistance to varus stress is conferred 75% by the bony articular structures, 9% by the lateral collateral ligament (LCL) and 13% by the anterior capsule while, at full extension almost half of the stability is conferred from the soft-tissue constraints (the LCL and capsule) and half is conferred by the osseous structures (Safran and Baillargeon, 2005).

The IOM is a fibrous sheet of connective tissue running from the proximal radius to the distal ulna. The IOM serves as an anatomic anchor for the forearm muscle attachments and guides and restrains the motion between the radius and ulna during pronation and
supination of the hand (Shepard et al., 2001). Histology studies reveal that the IOM has a thickened central third region resembling a ligament in architecture as well as tissue material properties comparable to those of the patellar tendon suggesting that it is capable of carrying high forces (Shepard et al., 2001).

Studies suggest that the position of the forearm also affects the stability of the elbow though the effect of forearm rotation on valgus laxity and stability of the elbow is still not fully understood. Several studies report that elbow laxity to both varus and valgus loads is greatest in pronation and 20° of elbow flexion, while others believe that it is greatest when the forearm is in neutral rotation and between 70° and 90° of flexion (Lockard, 2006; Safran and Baillargeon, 2005; Seiber et al., 2009).

Dynamic stability in the elbow joint is provided by 4 main muscle groups that traverse the elbow which are shown in Figure 2-7. The biceps brachii, brachioradialis, and brachialis muscles are the major flexors while the triceps and anconeus muscles are the major extensors of the elbow joint (Morrey and An, 1983). When the biceps, brachialis, and triceps contract they produce joint compression forces. The directions of these joint forces change with changes in joint angle or position thus preventing dislocation of the joint under different types of loading (Lockard, 2006). For instance, when muscular contraction occurs with the elbow in full extension then the joint compression forces are mainly directed anteriorly whereas when contraction occurs in a flexed position then the forces are directed posteriorly (Lockard, 2006). Supination at the elbow is achieved by the supinator and biceps brachii and pronation through the pronator quadratus, pronator teres, and flexor carpi radialis muscles (Lockard, 2006).

On the medial aspect of the joint the pronator teres, flexor digitorum superficialis, flexor carpi ulnaris, and flexor carpi radialis resist valgus force irrespective of forearm rotation by applying a varus moment to the joint (Safran and Baillargeon, 2005). On the lateral aspect of the joint the extensor digitorum comminus, extensor carpi radialis brevis, longus, anconeus, and extensor carpi ulnaris play a role in varus stability of the elbow by producing a valgus moment (Murray et al., 1995).
2.3 Elbow Injuries

Elbow injuries can be classified into acute or chronic overuse injuries. Acute injuries are associated with direct trauma such as fractures and dislocations; most commonly of the radial head or olecranon process, as well as acute ligamentous avulsions or tendon ruptures (Badia et al., 2008). Chronic overuse injuries are the result of repetitive overload and include several types of tendinosis (micro-tears that damage the tendon at a cellular level), inflammation of the soft tissues such as tendonitis, epicondylitis and muscular injury (Frostick et al., 1999). Chronic injuries are quite common among overhead throwers and the type of injury depends on the type of the activity (Frostick et al., 1999). Repetitive elbow flexion can cause biceps tendinosis or anterior capsule strain whereas any activity that involves forceful elbow extension can cause triceps tendinosis.
or posterior impingement syndrome. Enthesopathies such as lateral and medial epicondylitis are frequently associated with racquet sports or gymnastics while valgus stress injuries, posterior impingement and degenerative conditions such as chondromalacia, loose bodies, and osteochondritis dissecans (OCD) that mainly affect the articular cartilage within the joint are common among overhead throwers (Frostick et al., 1999). Post-traumatic stiffness of the elbow frequently follows after most of these injuries. In this case the capacity of the joint’s capsule to receive injected fluid is significantly reduced resulting in a reduced arc of motion (Ralphs and Benjamin, 1994).

2.4 Summary

The elbow joint is a complex structure with a remarkable capability in both its range of motion as well as the forces capable of bearing. Even though it is largely considered as a single joint the elbow is in fact the interface of relative motion between three bones; the humerus, radius and ulna, constrained at three articulations; the humeroulnar, humeroradial and radioulnar joints. The elbow’s unique anatomy includes the bony geometry, articulation and soft tissue structures described in this chapter significantly contribute to the joint’s stability. During passive and dynamic activities stability is essential for the joint in order to maintain its function as the mechanical link between the hand, wrist and the shoulder.

Elbow biomechanics and kinematics in overhand throwing are quite complex due to the speed and number of joint actions involved. During throwing the elbow is highly stressed in order to generate the rotations needed for sport performance while, the passive and active stabilisers of the joint are loaded in order to control these rapid motions. It becomes obvious that a sound understanding of the anatomy and biomechanics of the elbow is essential in order to understand the kinematics of the joint during cricket bowling that will be described in detail in the following chapters.
Chapter 3  The biomechanics of cricket bowling – Literature Review

The main focus of this thesis is the measurement and analysis of cricket bowling. Therefore, in this chapter a review of current knowledge on the biomechanics of cricket bowling is conducted.
Chapter 3  The biomechanics of cricket bowling – Literature Review

3.1  Introduction

The game of cricket is followed with passion in many different parts of the world and over 120 cricket-playing nations are formally recognised by the International Cricket Council (ICC). This international recognition and the expectations of cricket supporters ensure a high pressure upon players, especially upon cricket bowlers (Orchard et al., 2009).

Bowlers are required to propel the ball repeatedly, with speed and accuracy while their upper limb segment (the bowling arm) moves in accordance with the laws set by the Marylebone Cricket Club (MCC), the custodians of the Laws of the Game, in order to maintain the integrity of the game.

The literature review presented in this chapter will provide a brief introduction to the biomechanics of cricket bowling with the main focus being the definition and interpretation of the ‘no-ball’ law and the challenges in assessing the legality of what constitutes a fair delivery.

3.2  Cricket bowling

Balls in cricket are delivered in groups of 6, called overs and bowlers are not allowed to bowl 2 overs in a row to ensure they all get a minimum short rest break after each over (Orchard et al., 2009) and to provide variety to the players and spectators. In one-day matches each bowler is allowed to bowl a maximum of 10 overs with a maximum of 50 overs to be bowled in each team’s single innings. The more recently introduced 20 over matches allow bowlers a maximum of four overs. In First-class matches, which consist of two innings per team, each game can last for 4 or 5 days and there are unlimited overs allowed depending on the time it takes to dismiss the opposition so bowlers can be required to bowl about 50 overs or even more (Orchard et al., 2009). The rules of two-innings games are known as the Laws of Cricket and are maintained by the MCC (MCC, 2010). For Test matches and One Day Internationals (ODI) additional Standard Playing Conditions for Test matches and One Day Internationals augment these laws. (MCC, 2010)
Bowlers can be generally classified into two broad categories, the fast or "pace" bowlers whose aim is to bowl quicker than the batsman's reaction speed, and the spin bowlers or "spinners" that bowl slower deliveries that bounce in unpredictable ways (Dennis et al., 2003). In between the fast bowlers and the spinners are the "medium pacers" who rely on persistent accuracy to try and contain the rate of scoring.

Typically, there are four different phases of a delivery as shown in Figure 3-1 the set up, the unfold, the delivery and the follow through (Woolmer et al. 2008). The bowler reaches the delivery stride by means of a "run-up". Fast bowlers tend to take quite a long run-up as they need momentum to propel the ball, while spin bowlers start with a very slow run-up taking no more than a couple of steps before bowling (Orchard et al., 2009). During the run up phase the front arm extends high and the body turns sideways with the ball kept next to the bowler’s chin and pointing towards the wickets (Woolmer et al. 2008). The run-up ends as the bowler leaps into the air, with the back arched and the head behind a high left arm that is bent towards the right shoulder in preparation for the unfold phase (Bartlett et al., 1996).

During the unfold phase the back foot impacts at the ground and the front foot comes down and forwards. The weight of the bowler lies on the front foot with the body leaning away from the batsman and the arm starts to extend. The delivery phase starts as the front foot strikes the ground. Once the elbow of the bowling arm is fully straightened the other arm is pulled downwards facilitating the circular swing of the bowling arm about the shoulder joint. The elbow reaches peak extension at ball release. Several authors tend to subdivide the delivery stride in four different stages: back foot strike, mid-delivery stride, front foot strike and ball release. Finally, the elbow and trunk flex forward during the follow-through phase that initiates when the bent right leg steps past the front leg (Woolmer et al. 2008).
3.2.1 Fast Bowling

Fast bowlers can deliver the ball at a speed up to 45 m/s and rely on the ball’s speed to dismiss the batsman, by forcing them to react very quickly. Some fast bowlers also make use of the seam of the ball so that it swings during flight and bounces in unpredictable ways. The fast bowling motion involves a run-up and hurling movement with a straight elbow (Orchard et al., 2009). During the delivery stride the fast bowler’s trunk must flex, extend laterally, bend and rotate in a short period of time, whilst the body absorbs ground reaction forces which can be as high as six times their body weight (Bartlett et al., 1996). Fast bowling is classified into three main techniques; the front-on, side-on and mixed (Orchard et al., 2009).

In the front-on technique the bowler starts with a high run-up speed and at back-foot impact adopts a set-up position with the hips and shoulders in alignment with each other but rotated approximately 20° or more towards the batter (Wilk et al. 2008). No major deviation from this position occurs until after the front-foot impacts (Portus et al., 2000). The side-on technique on the other hand, involves a relatively lower run-up speed than the front-on technique with the bowler at back-foot impact adopting a set-up position with the hips and shoulders in alignment with each other but rotated approximately 20° or more towards the batter.
position with the hips and shoulders in alignment with each other pointing perpendicular to the batter (Portus et al., 2000). The mixed technique as shown in Figure 3-2 is a mixture of the side-on and front-on techniques and is characterized by the bowler initially adopting a front-on foot and shoulder orientation at back foot strike followed by a realignment of the shoulders to a more side-on position during the delivery stride (Bartlett et al., 1996).

![Figure 3-2: (a) The side-on technique (b) the front-on technique (c) the mixed technique: the hips are more front-on than the shoulders (d) the mixed technique: the hips are more side-on than the shoulders (Portus et al., 2000)](image)

3.2.2 Spin bowling

Spin bowlers aim to deliver the ball with rapid rotation in order for it to bounce in unpredictable ways once it hits the pitch and make it difficult for the batsman to hit the ball. Spinners use either predominant wrist or finger motion to impart spin to the ball and change the line of the ball off the pitch. They can be divided into four broad categories as shown in Figure 3-3; off–spin bowlers are right handed bowlers with a finger spin technique, leg–spinners are right handed bowlers with a wrist spin technique, left-arm orthodox spinners are left-handed bowlers with finger spin technique and left-arm unorthodox spinners that are left-handed spinners with wrist spin technique (Woolmer et al. 2008).

Several variations of the four categories do exist according to how the ball leaves the spinner’s hand and in which direction the ball turns the most popular of which include the following: the “googly” where the action of the fingers and the wrist during the delivery is the same and the ball is released out of the back of the hand starting to spin.
Chapter 3  The biomechanics of cricket bowling – Literature Review

The “top-spinner” is the easiest variation of leg-spin technique where the position of the wrist points to the side and makes the ball spin upon release in a straight line towards the batsman.

In an “off-break” delivery the bowler’s fingers are used to rotate the ball as much as possible before release while, in a “flipper” delivery the ball gets squeezed between the thumb and fingers in such a way that it spins backwards and skids on low and fast after hitting the pitch. Finally, the “doosra” is an off-break delivery with the wrist turned around so that the top of the wrist faces down the pitch (Woolmer et al. 2008).

![Diagram of spin bowling techniques](image)

Figure 3-3: Types of spin deliveries (Woolmer et al. 2008)

In summary, it can be seen that there are many different bowlers with great diversity of styles and techniques. This indicates that the movement of the elbow joint can significantly differ between bowlers suggesting that each type of bowling technique should be independently assessed.

3.3  The biomechanics of cricket bowling

The bowling action is a highly dynamic and flexible movement built on three main foundations; momentum to carry the bowler to the point of delivery, balance for the bowler to be in control of their movement during the delivery stride and timing to control the flight and length of the delivery (Woolmer et al. 2008).
Chapter 3 The biomechanics of cricket bowling – Literature Review

Momentum adds to the ball’s speed and this is one of the main purposes of generating run-up speed (Ferdinands et al., 2009; Woolmer et al. 2008). Especially for fast bowlers, this is achieved by using the ground reaction forces to initially slow their lower body, and then use the front leg as a lever with pivot at the foot (Ferdinands and Kersting, 2007). Another critical feature in bowling technique is the role of trunk flexion in the generation of ball release speed. Studies have demonstrated that a correlation between trunk flexion speed and ball release speed exists. Looking at the delivery swing it can be seen that as the bowler leaps into the air the trunk is initially rotated and extended away from the batsman but as the bowler lands and moves from back to front foot strike the trunk begins to rotate and flex forwards pulling the bowling arm with it (Ferdinands and Kersting, 2007). Bowlers with a larger and leaner upper torso tend to bowl faster than those with a smaller and less lean upper torso possibly because they tend to approach the bowling crease faster (Portus et al., 2000).

During bowling the elbow joint rotates about 90° as a result of humeral rotation during the bowler’s delivery action (Aginsky and Noakes, 2010). Because the elbow remains straight for most of the delivery stride external rotation of the shoulder is about 150° a value significantly smaller when compared to the maximum external rotation during baseball pitching that has been reported to be as much as 160° to 178° (Brown et al., 1988). This might be because maximising external rotation when the elbow is extended does not give a biomechanical advantage and the elbow is flexed in baseball pitching, but not normally so in cricket bowling. Once maximum external rotation has been reached, the shoulder quickly internally rotates to accelerate the arm (Wilk et al. 2008). Clinical studies have demonstrated that cricketers who regularly bowl or throw over arm have significantly less internal and greater external, dominant to non-dominant glenohumeral rotation (Giles and Musa, 2008). Recent studies have revealed that some straightening of the elbow does occur and differences in the range of elbow flexion-extension have been reported among different type of bowling techniques (Elliott et al., 2002; Portus et al., 2006).

Shoulder counter-rotation is defined as the change in the shoulder alignment angle between two points; (1) back foot contact to (2) the most side-on shoulder alignment
Chapter 3  The biomechanics of cricket bowling – Literature Review

(Figure 3-4) during the delivery (Ranson et al., 2008). Bowlers with greater shoulder counter-rotation and those that adopt a front-on shoulder alignment at back-foot impact have been found to experience a significantly greater elbow flexion-extension range during the delivery to those who have a more side-on orientation at the same point in the delivery action (Roca et al., 2006).

Figure 3-4: Shoulder counter-rotation from (1) back foot strike, (2) minimum shoulder angle

3.4 The ‘no ball’ law

Within the game of cricket a ‘no ball’ has been defined in many different ways. The first version of the ‘no ball’ law - set in 1810 and amended in 1816- introduced for the first time the clause governing action as follows: “The ball must be bowled (not thrown or jerked) and be delivered underhand with the hand below the elbow, but if the ball be jerked or the arm extended below the elbow, or the arm extended from the body horizontally when the ball is delivered, the umpire shall call ‘No Ball’”. By 1896 MCC granted either umpire the power to call ‘No Ball’ if not satisfied with the fairness of the delivery. Between 1810 and the final revision of the laws that took place in 2000 many different versions of the wording were trialed. The driving force behind this were the changes that occurred in the bowling style itself such as the introduction of the over-arm delivery technique in the beginning of the 19th century.

The most notable changes occurred in 1965 and 2000. Before 1965 the wording merely said “For a delivery to be fair the ball must be bowled, not thrown or jerked: if either
umpire be not entirely satisfied of the absolute fairness of a delivery in this respect, he shall call and signal ‘No Ball’ instantly upon delivery”. An experimental Law tested in 1965 stated: “A ball shall be deemed to be thrown if, in the opinion of either umpire, the bowling arm, having been bent at the elbow, whether the wrist is backward of the elbow or not, is suddenly straightened immediately prior to the instant of delivery. The bowler shall nevertheless be at liberty to use the wrist freely in the delivery action”. This is the first time that the elbow is mentioned in the laws and the importance of it remaining straight during the delivery is highlighted. What is not clear from this definition is from which phase of the delivery swing is the elbow supposed to remain straight. The first definition of ‘The Horizontal Law’ however, was as follows: “The ball shall be bowled, not thrown or jerked. That is to say that when, on the final swing, the bowler’s arm reaches the horizontal, it shall be fully extended from shoulder to wrist until the ball is released. This does not preclude the use of the wrist.”

The next major revision of the Laws was the 1980 Code, and prior to the introduction of the 2000 Code, this stated “For a delivery to be fair the ball must be bowled not thrown – see Note A below. If either umpire is not entirely satisfied with the absolute fairness of a delivery in this respect he shall call and signal ‘No Ball’ instantly upon delivery.” Note A went on to state “Definition of a throw. A ball shall be deemed to have been thrown if, in the opinion of either umpire, the process of straightening the bowling arm, whether it is partial or complete, takes place during that part of the delivery swing which directly precedes the ball leaving the hand. This definition shall not debar a bowler from the use of the wrist in the delivery swing.” This definition provides a better definition of the role of the elbow in the delivery swing.

The final revision of the laws took place in 2000. This version states: “A ball is fairly delivered in respect of the arm if once the bowler’s arm has reached the level of the shoulder in the delivery swing, the elbow joint is not straightened partially or completely from that point until the ball has left the hand. This definition shall not debar the bowler from flexing or rotating the wrist in the delivery swing.”
3.4.1 Definition of a fair delivery

The objective of the no-ball law (Law 24, MCC, “Laws of the Game of Cricket”, 2000 code, 3rd edition, 2008) is to control the bowler’s arm so that there is no straightening or bending of the elbow during ball delivery. The 2000 version of the law was formulated to prevent bowlers from taking an unfair advantage thus, restraining their arm in a fixed-elbow position so as to minimise the contribution of elbow extension to the ball’s speed (Portus et al., 2006). Internal rotation of the forearm may also contribute to ball speed, but this is not addressed in the current formulation of the law. Since the advent of high-speed video photography it has been revealed that some straightening occurs in bowlers who have actions that are traditionally considered in accordance with the law. Even though this is the most explicit description of the law so far since 2000 there has been much debate about the wording of this law and its implementation on the field of play.

In the 1990s The ICC, who administer the international game of cricket, decided to allow a certain amount of elbow extension to occur. This was set at 10° for fast bowlers, 7.5° for medium pace bowlers and 5° for spin bowlers. One of the major studies was carried out by Portus et al., (2003, 2006) whereby 21 fast bowlers (19 international bowlers and 2 first-class bowlers) were tested under match conditions using two synchronized high-speed (250 Hz) video cameras. Results showed that 41% of all the deliveries analysed exceed the 10° threshold for fast bowlers. These measurements were taken outdoors so bowlers had the advantage of being able to play under match conditions analogous to those in real games but this comes with two major disadvantages when analysing the data; firstly, the joint centers had to be manually digitised over their clothes and secondly, researchers were not able to establish joint coordinate systems based on anatomical landmarks so only the absolute change in angle between the upper arm and forearm segments could be calculated. Nevertheless, on the basis of these results, they recommended an increase in the allowable change in elbow extension angle from 10° to 15° for fast bowlers (Ferdinands and Kersting, 2007).

Ferdinands and Kersting (2007) used an eight-camera motion analysis system (240 Hz) to measure the elbow angles of 42 bowlers of the major types: fast, fast-medium, medium,
slow, and spin. They did not find any correlation between change in elbow extension angle and ball speed. Based on these results they suggested a 15° elbow extension angle limit for all bowling types. These results were accepted by the ICC sub-committee meeting in Dubai (25–26 October 2004), so the limit was extended to allow for 15° of elbow extension for all types of bowlers with any elbow hyperextension not included in the 15° threshold. This was a remarkable departure from previous recommendations that gave spin bowlers an extra 10°. It seems that there were a number of drivers for these changing values. The first is that objectively some elbow extension does occur in bowlers (Portus et al., 2006). The second reason is that the measurement of elbow movement is not without difficulty and current techniques that are used to measure elbow movement may not be accurate enough to quantify very small changes in elbow angle.

3.4.2 Influence of the elbow anatomy on the visual perception of “throwing”

Mutiah Muralitharan, a Sri Lankan cricketer has been “called for throwing” on two tours in Australia (1995-1996 and 1999) and has been the subject of much controversy ever since (Lloyd et al., 2000). After Muralitharan had been called for throwing in 1995-1996, the International Cricket Council’s Expert Bowling Panel viewed video tapes of Muralitharan taken from different angles and considered his action legitimate but he was once again called for throwing in 1999 (Lloyd et al., 2000). It has been reported that Muralitharan has a restricted range of motion in his elbow joint, which means that at full extension his elbow is bent (Lloyd et al., 2000). In this case when bowling with a fully extended arm, his elbow may appear either to extend or flex depending on the angle of viewing (Lloyd et al., 2000). Consequently, two-dimensional TV images have proven to be deceptive in the case of Muttiah Muralitharan but also in the cases of other highly-rated bowlers such as Michael Holding and Shoaib Akhtar.

The main reason why the naked eye and even the use of normal video footage are not always capable of making an accurate determination of bowling legality is the clinical presence of elbow abnormalities such as elbow hyperextension and large carrying angles.
in cricket bowlers (Aginsky and Noakes, 2010). As it has been shown Muttiah Muralitharan has a restricted range of motion in his elbow joint so that at full extension his elbow is bent (Lloyd et al., 2000). On the other end of the scale there is Shoaib Akhtar who has prominent elbow hyperextension. In both cases the elbow may appear to extend during the delivery swing beyond the 15° limit though this is not the case. Cricket bowlers have also been reported to have large “carrying angle” above 15° when the average “carrying angle” for men has previously been reported to range between 6.8° and 12.5° (Aginsky and Noakes, 2010). As Aginsky and Noakes (2010) and Portus et al. (2003) showed, rotation of the elbow joint causes the plane in which the elbow moves to change throughout the delivery action. This movement varies between bowlers but the movement of elbow flexion-extension can be viewed only when the viewer’s eyes are at exactly 90° to the plane of elbow joint movement. Because of the presence of a large “carrying angle” when viewing the delivery from only one position the visual illusion of a “throw” may be created (Aginsky and Noakes, 2010).

The controversy has heightened with some of the most famous and the two fastest bowlers in the world having been reported for throwing and the clinical presence of elbow abnormalities in bowlers being frequently reported. The 15° limit was set so that the naked eye can be in a position to distinguish between legal actions and throws. The problem with this is that in order to be fair to the bowler there should be some allowance for error in measurement as it is extremely difficult to measure the elbow flexion to within even a few degrees. Allowing for instance for an error of ±3° effectively means that it would be possible for a straightening of 18° to be allowable. This further complicates things as bowlers without bone abnormalities can now directly benefit from these few degrees of extension. An example of how this can be used is the “doosra” delivery.

Initially invented by Saqlain Mustaq a Pakistani off-spinner, the “doosra” is a delivery that behaves like an “off-break” but is delivered with much more wrist and elbow movement rather than finger action. In the hands of bowlers like Harbajhan Singh and Muttiah Muralitharan this delivery has proven to be very effective in dismissing batsmen but recent studies have showed that in order to use the “doosra” type of delivery to
impart spin to the ball, there needs to be considerable flexion of the elbow (Chin et al., 2009).

### 3.5 Injuries

The definition of a cricket injury is one that either (1) prevents a player from being fully available for selection in a major match or (2) during a major match, causes a player to be unable to bat, bowl, or keep wicket when required (Orchard et al., 2009). The most common bowling injuries reported are thigh and hamstring muscle strains, abdominal strains, and lumbar injuries (Orchard et al., 2009).

A study by Orchard et al. (2009) found an association between injury and higher workloads and showed that when comparing injured and non-injured players, the latter tended to have previously bowled more overs. The same study showed also that this increase in injury due to higher workload may be somewhat delayed for a short period of time. This result is quite important as it may shed light on the high rate of injury reported for One Day International (ODI) cricket matches played in Australia. These matches have the highest injury incidence of all matches even though the bowlers are limited to 10 overs. However, it is possible that the high rate of ODI injuries in Australia is in fact an artefact, reflecting high workloads from a few weeks earlier in the main Test match series (Orchard et al., 2009).

The fast bowling injury profile includes lumbar spine, lower limb, and shoulder injuries, but relatively few elbow injuries (Orchard et al., 2009). Injuries to the lower back such as pedicle sclerosis, spondylolysis and spondylolisthesis of the lumbar spine have become a major concern for bowlers. Three main factors are associated with lower back injury; high compressive loads, rotational stress and backward arching (Bartlett, 2003). Clinical data suggest that certain movements of the trunk such as shoulder alignment, counter-rotation (Figure 3-4), an increased separation angle (difference between shoulder alignment and hip alignment) and increased trunk lateral flexion (Figure 3-5) during bowling predispose fast bowlers to back injuries (Elliott, 2000; Wallis et al., 2002) but it is believed that most injuries are probably caused by a combination of these movements (Elliott, 2000). Fast bowlers using a front-on technique, associated more with an ‘open’
shoulder alignment at back-foot impact, show an increased shoulder counter-rotation which may increase their predisposition to lower back injury (Portus et al., 2000). Another example of this is the mixed technique where the realignment of the shoulders from a front-on to a more side-on position during the delivery stride in conjunction with the excessive shoulder counter-rotation that takes place in the transverse plane, are believed to increase the torsional load experienced by the lumbar spine (Burnett et al., 1995; Portus et al., 2000). This may justify the significantly higher incidence of lower back injuries among fast bowlers who adopt a mixed technique (Orchard et al., 2009).

Figure 3-5: Lateral flexion in a young fast bowler prior to ball release (red lines indicate the angles between the shoulders and the pelvis to the vertical)

The effectiveness of physical guidance on reducing these selected bowling characteristics has been investigated by Wallis et al (2002). A harness, a brace that consists of a belt and a strap fixed from the belt to the non-throwing arm (Figure 3-6) was worn by bowlers to restrict the movement of the shoulders during the delivery stride. Results showed that there were no significant modifications to technique after the coaching intervention when bowlers were once again assessed without the harness. This is possibly because the improved position at back-foot strike combined with shoulder counter-rotation led to an increased negative shoulder separation angle at front-foot strike (Bartlett, 2003).
The majority of bowling injuries are related to the lower back in fast bowlers as the incidence of injuries in fast bowling is greater than in spin bowling (Elliott, 2000; Orchard et al., 2006; Portus et al., 2000). Recent studies however, have showed that many injuries occur at the throwing shoulder especially in wrist spinners (Gregory et al., 2002). The injuries to the shoulder in cricket bowling include rotator cuff sprain and subacromial impingement (Elliott et al., 2002; Wallis et al., 2002).

Cricket today demands greater physical effort from players and there is evidence of an increase in overuse injuries (Orchard and James, 2003). However, the literature reviewed shows that research into the nature of these injuries has been limited and merely focused on fast bowling. Even though fast bowlers are indeed the most likely to be injured, no long term investigation has been carried out to identify injury patterns and there is lack of data associated with injuries in spinners. A good understanding of the causes of the injury, and the severity and the nature of it, is crucial and would enable for a better diagnostic and prevention techniques. This further suggests that a biomechanical model able to measure the 3D \textit{in-vivo} kinematics of the elbow joint during bowling can aid in the understanding of the causes of injury and if further improved can also lead to many improvements in the treatment of elbow and shoulder pathology.
Chapter 3  The biomechanics of cricket bowling – Literature Review

3.6 Summary

Cricket bowling is a highly dynamic and flexible movement with great diversity of styles and techniques to help the bowler dismiss the batsman. These include bowlers who utilise the ball’s speed varying the pace, length and the line of delivery and bowlers who aim to spin the ball in different ways as it travels through the air or to change the line of the ball off the pitch. Independently of their technique or style all bowlers are required to deliver the ball while their bowling arm is in accordance with the laws of the game.

The objective of the “no-ball” law is to restrain the bowler’s arm in a fixed position during ball delivery minimising the contribution of elbow extension and upper arm internal rotation in the ball’s speed. Bowlers that are found to straighten their arm during this action are ‘called for throwing’ and are penalized. However, since the advent of high-speed video photography it has been revealed that some straightening occurs in bowlers who have actions that are traditionally considered in accordance with the law and the role of elbow extension during the delivery action has been under debate ever since. The ICC in the 1990s decided to allow a small amount of elbow extension to occur; this was extended in 2004 to 15° for elbow extension for all types of bowlers.

Biomechanical assessments of the bowling action show the need for a more rigorous determination of bowling legality; specific to elbow kinematic measures such as extension, hyperextension, abduction and forearm rotation as the match-based data that the 15° extension limit was based on are limited only to elbow straightening data.
In the previous chapter a literature review on the biomechanics of the elbow joint in cricket bowling was presented. As a main thesis aim is to develop a model to measure elbow kinematics during bowling, a literature review on tracking elbow movement under dynamic activities is presented.
4.1 Introduction

Measurement of elbow kinematics as described in Section 2.2.3 is not without difficulty because of the large range of motion of the joint which is obscured at the same time by overlying soft tissue. Motion analysis of a joint requires a kinematic model to describe the joint rotations. For a kinematic model to be accurate several parameters need to be taken into account such as the geometry of the articulating bones, the path of motion and the time over which the movement occurs (Hill et al., 2007). In order to calculate joint rotations the majority of these models employ mathematical techniques and optimisation methods (Hill et al., 2007). It becomes obvious that these models rely on the repeatability of the input data as well as to the mathematical assumptions made (Hill et al., 2007). Even though several data acquisition methods and mathematical techniques have been proposed over the past few years an optimal solution is yet to be described.

In order to employ a kinematic model to study the movement of the elbow joint in cricket bowling input data must first be acquired from a set of certain movements with a high level of accuracy and then a mathematical model needs to be employed to calculate joint rotations. In this chapter a literature review on measuring elbow kinematics will be conducted with the main focus being the challenges in measuring elbow motion and the optimisation methods proposed to solve the problem.

4.2 Elbow Measurement Techniques

A number of techniques have been developed to measure joint kinematics, varying from imaging techniques (Goto et al., 2004), motion analysis tracking systems (Lloyd et al., 2000; Roca et al., 2006) and cinematography (Rosecrance and Giuliani, 1991), to bone pins (Karduna et al., 2001), electrogoniometry (Barker et al., 1996), video-fluoroscopy (Baltzopoulou, 1995) and accelerometry (Bach et al., 1994). However, not all of the techniques proposed are accurate enough or suitable for elbow measurements. In this section the most commonly used techniques and the challenges faced on measuring dynamic movements of the elbow are presented.
4.2.1 Non-invasive dynamic measuring techniques

4.2.1.1 Motion Tracking Systems

Electromagnetic and optical motion tracking are the most frequently systems used in biomechanical research. Both systems have been successfully employed to measure joint kinematics under different conditions ranging from activities of daily living to sports biomechanics (Cutti et al., 2008; Elliott et al., 2007; Lloyd et al., 2000; Meskers et al., 1998; Roca et al., 2006).

Electromagnetic systems (Figure 4-1a) enable the direct measurement of the position and orientation of multiple sensors. Their main advantage is that they are relatively inexpensive, not subject to marker obscurement problems and require only one sensor per segment (Mills et al., 2007). However, the systems’ main drawbacks include their susceptibility to magnetic interference from metal objects and the limited capture...
volume. Finally, the wires connecting the sensors to the receiver can limit the range of the activity performed during certain tasks.

The main principal behind optical motion systems (Figure 4-1b) is that by viewing the same points/markers by multiple cameras simultaneously it is possible to measure the precise 3D positional data of each marker. However, these systems are sensitive to light conditions and any obstacle between the marker and the camera can seriously degrade the tracker's performance. This is the reason why they can normally only be used indoors with the environment carefully designed prior to any data capture (Elliott and Alderson, 2007).

Both motion tracking systems have the advantages of being fully dynamic and suitable for sport applications as they allow the subjects to move naturally. However these systems can be prone to skin movement artefact as in regions where the soft tissue is thick the sensors/markers are likely to reflect and combination of skin deformation and bone movement rather than the actual bone movement (Anglin and Wyss, 2000; Lundberg, 1996).

**4.2.1.2 Cinematography**

Video–based motion tracking systems can track and record any motion or movement in the camera's field-of-view. A minimum of two high speed cameras are required to record the activity and in most cases direct linear transformation algorithms are employed to convert sets of two-dimensional image coordinates into three-dimensional images. These systems can be used outdoors to measure both joint kinematics and kinetics and are quite flexible as they do not require markers onto the subject’s body. However, visible area constraints compromise the accuracy of these systems when used outdoors and lack of anatomical landmarks means that only absolute changes in angle between two adjacent segments can be calculated (Portus et al., 2003) thus these systems cannot be used for complex joint movements where all three rotations need to be recorded and the different rotations delineated taking into account possible cross-talk.
4.2.1.3 Accelerometer and rate gyroscope measurement of kinematics

Few authors have attempted to combine accelerometers and rate gyroscopes to measure the kinematics and kinetics of joints. These systems are relatively inexpensive, light weight and easy to mount onto the skin. Movement quality is mainly characterised using power spectral analysis of acceleration signals of the two segments moving relative to one another (Bach et al., 1994; Mayagoitia et al., 2002; Wixted et al. 2010). These systems constitute a promising design and have been effective in measuring knee kinematics during sit-stand trials and walking (Boonstra et al., 2006; Williamson and Andrews, 2001). However, the attachment of the device to soft tissue is an inherent source of error and the analysis of the accelerometer signal can be problematic mainly due to excessive noise in the raw data (Aminian and Najafi, 2004). To allow capture of outdoor and long-term movements these sensors need to be combined with a portable data logger able to accurately and wirelessly transmit the raw data.

4.3 Challenges in measuring elbow kinematics during cricket bowling

As discussed in Chapter 3 elbow biomechanics during cricket bowling are quite complex due to the speed and number of joint actions involved. Non invasive measuring techniques have been used to assess elbow kinematics during the delivery stride but the accuracy of these methods is still questionable as to whether they truly reflect the joint’s movement. Several researchers have tried to address the problems of these techniques so that reliable measurements of joint movements can be obtained.

4.3.1 Skin movement artefacts

Soft tissues artefacts (STA) constitute a major source of error in the calculation of joint kinematics and dynamics due to the relative displacement between the markers/sensors and the underlying bone, mostly associated with the interposition of both passive and active soft tissues (Cappozzo et al., 1996; Cutti et al., 2006) which can cause an underestimation (Meskers et al., 1998) or an overestimation (Karduna et al., 2001) of the actual bone movement.
The exact magnitude of STA in kinematic calculations has been difficult to determine (Peters et al., 2010). Unlike any instrumental errors, skin artifacts originate from the same motion as the segment and have the same frequency content as that of the underlying bone. This makes separation of the noise from bone movements using standard filtering techniques an impossible task (Cappozzo et al., 1996; Leardini et al., 2005). However, several studies employing bone pins (Lafortune et al., 1992), percutaneous trackers (Holden et al., 1997), external fixators (Cappozzo et al., 1996), and Roentgen photogrammetry (Maslen and Ackland, 1994) have showed that the influence of skin artefact is directly associated with the speed and nature of the movement performed but also with the physical characteristics of individuals and marker placement (Cappello et al., 2005; Cappozzo et al., 1996; Cutti et al., 2005; Leardini et al., 2005). For this reason studies have focused on introducing compensation techniques to reduce the magnitude of these variables onto the STA.

### 4.3.1.1 Marker Location

The choice of marker placement to define the segments, especially the ones of the upper extremity has been variable among researchers, particularly with reference to the humerus (Anglin and Wyss, 2000). Rab et al. (2002) placed their markers over prominent bony landmarks of the upper limb which were easily palpable thus placement was reproducible. In addition the markers were placed where subcutaneous tissue was thin in order to minimise marker movement. These bony landmark positions were later on included by the International Society of Biomechanics (ISB) who issued a protocol on marker placement and joint definitions in 2005 (Wu et al., 2005).

Schmidt et al. (1999) used two sets of markers, a static and a dynamic set, in their upper limb model. The static set was defined from 5 markers placed on the bony landmarks in order to identify joint centres (JCs). The dynamic set consisted of 3 collinear markers place on the upper arm, forearm and hand and these were used to measure the movements of the segments through all their degrees of freedom. The location of the anatomical landmarks and related JCs calculated from the initial static marker set were defined and held relative to the position and orientation of these dynamic markers in
the calibration trial. This allowed for the static marker set to be removed during movement such that they did not interfere with the motion. Cappozzo et al., (1995) proposed a similar compensation method to calibrate the positions of certain anatomical landmarks which are either not practical for use in dynamic experiments or can introduce high errors relative to the tip of a pointer of known dimensions. Although this method was originally proposed for the lower limb it has also been adopted in several upper limb studies (Cutti et al., 2005; Stokdijk et al., 1999). In this method the anatomical landmark positions were defined relative to a technical frame on the same segment in a static trial and then used for the remainder of the experiment. However, in both Schmidt’s and Cappozo’s suggestions either the dynamic marker set or the technical frame still follow to some extent the movement of the muscles on the segments they are attached to, which in some cases do not follow well the movement of the underlying bones (Cutti et al., 2005). It is important therefore, to place them on the segment in positions that have less significant skin-to-bone movement. With respect to the upper arm, Williams, (1996) and Anglin and Wyss (2000) suggested that three markers placed directly on the deltoid insertion, the medial head of triceps and the origin of the brachioradialis muscle could be chosen for the technical frame of the humerus (Figure 4-2a). Cappozzo et al, (1995) on the other hand proposed the use of light-weight frames with a minimum of 3 markers each (Figure 4-2b), placed onto the segment of interest. The use of cluster of markers in either rigid or deformable frames has shown several advantages such as easier marker mounting on subjects, optimal selection of cluster location and reduced number of cameras needed for data collection (Cappello et al., 1997).

### 4.3.1.2 Activity performed

The speed and nature of the activity performed has been shown to affect the amount of STA in kinematic measurements in a number of studies (Cappello et al., 2005; Cutti et al., 2005; Fuller et al., 1997; Peters et al., 2010). Fuller et al., (1997) and Cappello et al., (1997) investigated the STA of the lower limb during cycling and walking activities with the latter choosing two calibration postures to digitise anatomical landmarks, but showed little difference in the effect of STA at both the greater trochanter and the knee.
epicondyles. Stagni et al., (2006) and Cappozzo et al. (1996) also looking at STA propagation in the lower limb showed that maximal errors are encountered when a segment undergoes movement, or when a marker location is on a joint line (Peters et al., 2010).

Manal et al., (1999) evaluated eleven marker sets that differed in the location they were attached to the shank and the method used to attach them to the segment, by tracking the motion of the tibia during natural cadence walking. Results showed that a marker set consisting of four markers attached to a rigid shell and positioned over the distal lateral shank yielded the best estimate of tibial rotation (Manal et al., 2000). Similar studies to the upper body have showed that placement of either markers or clusters onto the upper arm is largely affected by STA during humeral internal-external rotation (Cutti et al., 2005).

Apart from trying to optimise the position of the markers/clusters in an attempt to minimise the errors caused by the relative skin movement other compensation techniques that optimise the relative movement (orientation and position) between segments based on joint constraints (Biryukova et al., 2000; Cutti et al., 2008; Schmidt et
al., 1999) have also been successfully suggested. These techniques include double anatomical landmark calibration (Cappello et al., 1997) and functional based joint axes (Cheze et al., 1998; Cutti et al., 2008; Stokdijk et al., 1999) which are task and subject specific and require a number of additional markers and data collecting trials. However, they have been found to perform better during dynamic activities (Besier et al., 2003; Stagni et al., 2006). Finally, minimisation techniques such as global optimisation (Lu and O'Connor, 1999) have also been trialed. These techniques try to minimise the STA by means of a predefined kinematic model of the body with specific constraints at the joints of interest. However, these methods do not take into account the variability between subjects or the differences between tasks and tend to perform poorer than compensation techniques (Stagni et al., 2009).

4.4 Summary

Several non–invasive measuring techniques exist that are able to capture dynamic activities and quantify several kinematic and kinetic properties of joints. However, as described in Section 4.2.1 all of these techniques are associated with errors which may potentially affect the reliability of the input data. As cricket bowling constitutes a highly dynamic activity taking place in a large volume (Section 3.2) the two main methods that are currently used to measure the 3D motion of the bowling arm are video–based and optical motion capture systems with the latter have been shown to be more accurate than video-based systems inside the laboratory environment (Elliott and Alderson, 2007). It is without a doubt preferable to record data under match conditions and it is possible that combining accelerometers and rate gyroscopes or using novel video based motion based systems will be able to be used in the near future, providing that the accuracy of the data reaches an acceptable level.

The studies looking at STA in joint kinematics have reached different conclusions as described in Section 4.3.1. However, most of the studies agree that even though it is very difficult to quantify STA, its effect in human motion analysis highly depends upon marker location, the activity performed and the instrumented segments (Kontaxis et al., 2009). With respect to the upper limb segments and the elbow joint which are of
interest in this study, compensation techniques such as single and double anatomical landmark calibration of the epicondyles and functional based joint axes definitions are believed to perform better and improve the reliability of the input data. Caution however, should be exercised when mounting markers or clusters of markers onto the segment of the upper arm as studies have identified that this segment is largely affected by STA during humeral axial rotation (Cutti et al., 2005). Most of the variables presented in this chapter will be taken into account in Chapter 5 where the development of a kinematic model to measure elbow motion during cricket bowling will be described. Furthermore, having identified the most critical sources of error in motion analysis the accuracy of the kinetic model will be assessed in Chapter 6 through a series of sensitivity studies in order to identify the most critical variables that can affect the reliability and repeatability of the proposed model.
Chapter 5  Kinematic Model of the Elbow Joint

In the previous chapter a literature review on capturing and analysing elbow motion during dynamic activities was presented. In this chapter a kinematic model to measure elbow biomechanics during cricket bowling is presented.
Chapter 5 Kinematic Model of the Elbow Joint

5.1 Kinematic models of the elbow joint

As described in the Chapter 4, the methods that are currently used to measure the 3D motion of the bowling arm are optical motion tracking systems, where skin markers are attached to the bowling arm of bowlers whose bowling action is subsequently recorded. The general principle behind motion analysis is that, by viewing the same points/markers by multiple cameras simultaneously it is possible to measure the precise 3D positional data of each marker. Based on these data the joint rotations between segments are then to be computed from clusters of markers. Other kinematic parameters such as velocity and acceleration are also determined from the derivation of these marker coordinates.

Several authors (Anglin and Wyss, 2000; Lundberg, 1996) have questioned the accuracy of these systems mainly due to the fact that in regions where the soft tissue is thick or moves independently from the bone underneath high errors are introduced with the markers reflecting the skin deformation rather than the actual bone movement (Lundberg, 1996). Nevertheless, these systems have been shown to be suitable for fast dynamic movements (Lloyd et al., 2000).

Different techniques have been developed for measuring joint kinematics using optical motion tracking systems and can be distinguished into two broad categories: the anatomical methods which take into account the relative movements of the markers of a cluster attached to a body segment (Cappozzo et al., 1995; Cheze et al., 1998; Spoor and Veldpaus, 1980) and the functional methods which optimise the relative movement (orientation and position) between segments based on joint constraints (Biryukova et al., 2000; Cutti et al., 2008; Schmidt et al., 1999).

5.1.1 Aims and Objectives

The aim of this study is to develop a kinematic model using a functional approach to measure elbow movement during cricket bowling and examine the scientific evidence behind new regulations of the International Cricket Council (ICC). All the steps of the
model development; characteristics of the participant group, data collection, modeling process and data analysis are presented in this chapter.

5.1.2 The functional axis

Under continuous movement of the elbow joint the forearm may be viewed to have a translation velocity along and a rotation velocity about a continuously changing line in space. This instantaneous axis of rotation is often called helical axis (IHA) or screw axis (SDA) (Woltring et al., 1985). The total amounts of translation and rotation can be defined as the time integrals of the instantaneous translation and rotation velocities at the instantaneous axis from a given reference time (Woltring et al., 1985). Various uses of the screw or helical axis have previously been reported in the literature in an attempt to accurately describe the kinematics for both the intact and the unstable elbow (Bottlang et al., 2000; London, 1981; Stokdijk et al., 1999), knee (Bull and Amis, 1998; Johnson et al., 2004) and spine (Kettler et al., 2004). The main advantage of the IHA is that its orientation remains invariant regardless of the reference co-ordinate axes used (An and Chao, 1984), however this definition comes with the disadvantage that the axis cannot be defined for pure translations and is highly dependent on experimental variability and rotation increment spacing between axis calculations (Woltring et al., 1985).

If referred to anatomical co-ordinate systems, the instantaneous axis of rotation can give a significant perception of movement evolution. For slow movements or short displacements, instantaneous helical axes can be averaged over large displacements with the resulting mean axis called finite helical axis (FHA). The main difference between the definitions of the FHA and IHA is that the finite axis describes a motion step while an instantaneous axis describes a motion at infinitesimal time (An and Chao, 1984). Despite these differences, most definitions of the IHA and FHA have been shown to improve the repeatability of kinematic and kinetic data in cadaveric studies. However, the sensitivity of the method for true in vivo data collection is still an issue.
Multiple methods and filtering techniques have been employed to calculate the axis parameters and improve the accuracy of the method; most notably Cheze et al. (1998) determined that for the elbow FHA measurement errors decreased for angular displacements greater than 25° while, Johnson et al. (2004) showed that rotation about the knee screw axis is seen to be repeatable, accurate and time step increment insensitive but at the same time displacement along the axis is highly dependent on time step increment sizing, with smaller rotation angles between calculations producing more accuracy.

5.1.2.1 The IHAs at the elbow joint

The relative distance between the segments of the upper arm and forearm during movement is used in order to determine the axes of rotation at the elbow joint. The positions of the IHAs vary depending upon the joint angle between the two segments. For this reason subjects need to perform several repetitive movements either actively (Cutti et al., 2008; Stokdijk et al., 1999) or passively (Bottlang et al., 2000; Veeger et al., 2006) as the elbow explores most of its range of motion (RoM). Stokdijk et al. (1999) asked subjects to actively perform elbow flexion extension while seated on a chair with the right elbow extended and the forearm in neutral position. From the initial position, five cycles of elbow flexion and extension were performed at four different velocities. In a later study, Stokdijk et al. (2000) used only the middle 50° from each movement suggesting that alterations in velocity in the extreme flexion and extension positions can cause inaccurate results. To obtain the instantaneous helical axes Cutti et al. (2008) asked subjects to flex-extend their elbow up to 130° five times, keeping a constant pronation and the humerus alongside the body, while Biriykova et al. (2000) instructed subjects to actively perform flexion and extension tasks with their forearm in neutral position.

Regardless of the motion performed most of these studies suggest that the flexion/extension (F/E) axis does not lie along a line parallel to the line joining the medial and lateral epicondyles as it is normally defined in anatomical axes but deviates by approximately 6.0° (±2.6°) from this (Chin 2009; Duck et al., 2003). These results support
previous clinical studies whereby the elbow F/E axis is described as the line running through the centre of the trochlea and the capitulum humeri (Deland et al., 1987; Fick and Lyons, 1997; Morrey and Chao, 1976; Youm et al., 1979).

In addition to the F/E axis several studies have sought to quantify the instantaneous pronation/supination (P/S) axes utilising the same methodology as that for the F/E determination. Most notably, Veeger et al. (1997) determined the P/S axis in vitro and Biriykova et al. (2000) determined the P/S axis in vivo but they both reported that the functional axis runs through the radial head and lies close to the lateral epicondyle crossing the F/E axis at an angle of 88.9° (± 5.1°).

5.2 Materials and Methods

5.2.1 Equipment and lab-set up

An optical motion tracking system (Vicon, Oxford, UK) was used at an acquisition rate of 200 Hz to track reflective markers attached to the skin of the bowlers. Two video recorders were also used at an acquisition rate of 100 Hz to record the delivery swing and to allow synchronisation with the opto-reflective data. The camera locations were selected as shown in Figure 5-1 in order to maximise the spatial volume of recording. This allowed for good detection of the reflective markers at each frame of the video.

Calibration of all the cameras was completed prior to data acquisition and an accuracy of ±0.2 mm was always obtained. The reflective markers are spheres of 14 mm in diameter on plastic bases and were attached to bony landmarks on the head, thorax, humerus, forearm and hand of the bowlers using double sided tape.
5.2.2 Marker Placement

Six markers were placed on the thorax; on the left and right acromion processes, the 7th cervical vertebra, the 8th thoracic vertebra, the xiphoid process and the suprasternal notch. In the bowling arm two markers were placed directly onto the anterior and posterior surfaces of the shoulder, two on the lateral and medial aspects of the wrist and finally, one marker was placed on the bowler’s hand as shown in Table 5-1 and Figure 5-2.

Figure 5-1: (a) Viewing positions to detect elbow flexion-extension angle during the delivery swing (A): Directions from which elbow flexion-extension can be detected when the upper arm reaches the level of the shoulder (B): Directions from which elbow flexion-extension can be detected prior to ball release (C): Directions from which elbow flexion-extension can be detected at ball release (Modified from: Aginsky and Noakes (2008)) and (b) Cameras arranged within the MCC Indoor School at Lord’s cricket ground
Table 5-1: Anatomical landmarks used in elbow tracking

<table>
<thead>
<tr>
<th>Segments</th>
<th>Markers</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>H: Head</td>
</tr>
<tr>
<td></td>
<td>HR: Head Right</td>
</tr>
<tr>
<td></td>
<td>HL: Head Left</td>
</tr>
<tr>
<td>Thorax</td>
<td>IJ: Suprasternal Notch</td>
</tr>
<tr>
<td></td>
<td>C7: Spinal Process of the 7th cervical vertebra</td>
</tr>
<tr>
<td></td>
<td>T8: Spinal Process of the 8th thoracic vertebra</td>
</tr>
<tr>
<td></td>
<td>XP: Xiphoid Process</td>
</tr>
<tr>
<td></td>
<td>RA: Right Acromion</td>
</tr>
<tr>
<td></td>
<td>LA: Left Acromion</td>
</tr>
<tr>
<td>Upper Arm</td>
<td>PSH: Posterior Surface of the Shoulder</td>
</tr>
<tr>
<td></td>
<td>ASH: Anterior Surface of the Shoulder</td>
</tr>
<tr>
<td></td>
<td>P1: Marker 1, Cluster P</td>
</tr>
<tr>
<td></td>
<td>P2: Marker 2, Cluster P</td>
</tr>
<tr>
<td></td>
<td>P3: Marker 3, Cluster P</td>
</tr>
<tr>
<td></td>
<td>D1: Marker 1, Cluster D</td>
</tr>
<tr>
<td></td>
<td>D2: Marker 2, Cluster D</td>
</tr>
<tr>
<td></td>
<td>D3: Marker 3, Cluster D</td>
</tr>
<tr>
<td>Forearm</td>
<td>FA1: Forearm 1</td>
</tr>
<tr>
<td></td>
<td>FA2: Forearm 2</td>
</tr>
<tr>
<td></td>
<td>FA3: Forearm 3</td>
</tr>
<tr>
<td></td>
<td>US: Medial Aspect of the Wrist (the most caudal-medial point on the ulnar styloid)</td>
</tr>
<tr>
<td></td>
<td>RS: Lateral Aspect of the Wrist (the most caudal-lateral point on the radial styloid)</td>
</tr>
<tr>
<td>Hand</td>
<td>HA1: Hand</td>
</tr>
</tbody>
</table>
Three clusters, with three markers each, mounted on light-weight frames (Figure 5-3) were attached to the upper arm and forearm of the bowling arm.

Cluster P was placed centrally onto the upper arm such that two markers were positioned parallel to the humeral long axis and one in the middle and Cluster D was positioned a few centimetres above the olecranon process as shown Figure 5-2. The clusters on the upper arm were used to replace some of the digitised anatomical landmarks during dynamic trials. The forearm cluster was also placed centrally onto the
forearm and finally, three markers were attached on a head band which was placed on the head of the subjects for the detection of the head.

### 5.2.3 Experimental Protocol

A total of 9 male bowlers, 4 medium pace, 3 fast bowlers and 2 spinners from the MCC Young Cricketers and staff participated in this study. Their mean age was 21 years (range: 16-31 years) and mean body mass index was 22.3 (range: 19 - 27.8). All testing took place at the MCC’s indoor cricket school which allowed for a 19-yard run-up for bowlers. Each bowler was asked to warm up on their own prior to data capture.

Initially, a static trial was taken of the marker setup with the subject standing and the arms hanging next to the body with the palms facing forwards (Willis 2005). Following this, two more static trials were obtained to calibrate the lateral (LE) and medial (ME) epicondyles. In each static trial the tip of the wand was positioned by the observer on each epicondyle (Figure 5-4). The post-processing and analysis of this trial is explained in Section 5.2.4.1.

A dynamic trial was captured in order to estimate the glenohumeral centre of rotation (GH). In this trial each subject was seated on a backless stool in the middle of the capture volume. The subject was asked to move their humerus in all three rotational degrees of freedom in a random manner, with their elbow flexed at 90° and at constant pronation. During this trial the observer tracked the movements of the scapula using a scapula locator as shown in Figure 5-6.

A dynamic trial was captured in order to estimate the IHA of elbow flexion-extension during a single joint movement of pure flexion. In this trial the subject was instructed to flex and extend their elbow up to 140° for five times, keeping a constant pronation and the humerus alongside the body.

A dynamic trial of a pure pronation-supination movement was captured in order to estimate the instantaneous helical axis of the forearm helical axes. During this trial each subject was instructed to fully pronate and supinate their forearm, keeping the elbow
flexed at 90° and the humerus once again alongside the body. The post-processing and analysis of these trials is explained in Section 5.2.6.

After digitisation, each bowler completed a total of 16 deliveries, from which the first 4 were not included in the analysis to allow for the bowlers to reach match pace. From the remaining deliveries, 6 successful deliveries were collected for analysis as proposed by the ECB regulations (ECB 2010). These trials were selected on the basis of markers visibility throughout the delivery. Analysis consisted of calculating the extension of the elbow at the point that the upper arm reached the level of shoulder until ball release as described in Section 5.2.10. This definition is in accordance with the relevant law in cricket that is associated with ‘no-ball’.

5.2.4 Data Analysis

5.2.4.1 Digitisation of the elbow epicondyles

In the case of the elbow joint the largest skin deformation is observed around the humeral epicondyles because of the loose skin surrounding the joint area and placement of markers around this area is highly discouraged (Cappozzo et al., 1996; Leardini et al., 2005). This problem becomes more prominent in individuals with joint abnormalities such as elbow hyperextension (Roca et al., 2006). In order to account for this error the positions of the two elbow epicondyles were digitised relative to other markers that are fixed elsewhere on the two body segments – the upper arm and the forearm through a procedure usually termed as anatomical landmark calibration (Cappozzo et al., 1995). The position of these landmarks during dynamic trials is expressed with respect to an upper-arm ‘cluster of markers’.

The position of the elbow epicondyles was defined relative to the position of the technical co-ordinate system of a pointer’s triad in ‘static’ trials before data collection. The tip of the calibration wand as shown in Figure 5-4 was placed onto the anatomical landmark of each epicondyle with the elbow flexed at 90° and a co-ordinate frame was defined using three markers on the wand as follows;
O_w: The origin coincides with marker C on the wand

X_w: The line connecting marker A and marker C, pointing towards A.

\[ X_w = \frac{(A - C)}{||A - C||} = \frac{\vec{AC}}{||\vec{AC}||} \] (5.1)

Y_w: The line connecting marker B to marker C, pointing towards B.

\[ Y_w = \frac{(B - C)}{||B - C||} = \frac{\vec{BC}}{||\vec{BC}||} \] (5.2)

Z_w: The line perpendicular to the X_w and Y_w axes, pointing upwards.

\[ Z_w = \frac{X_w \times Y_w}{||X_w \times Y_w||} \] (5.3)

Figure 5-4: Calibrating the position of the lateral epicondyle (LE). A co-ordinate frame of the wand is defined using markers C, A and B. Vectors a and b show the distance of the tip of the wand from marker C (Shaheen 2010)

The wand’s technical co-ordinate frame was then used to define the position of each epicondyle based on the known distances between the marker C on the wand and the landmark position. The global positions of the lateral (LE) and medial (ME) epicondyles were calculated using the following equations:
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\[
\overline{LE}_L = \overline{C} - a \cdot \overline{X}_w - b \cdot \overline{Z}_w \quad (5.4)
\]

\[
\overline{ME}_L = \overline{C} - a \cdot \overline{X}_w - b \cdot \overline{Z}_w \quad (5.5)
\]

During data capture of the bowling action the position of each epicondyle was reconstructed with respect to a local technical frame on the upper arm (Cappozzo et al., 1995; Roca et al., 2006).

\(O_{hp}\): The origin coincides with \(P_1\)

\(Y_{hp}\): The line connecting \(P_1\) and \(P_2\), pointing towards \(P_2\).

\[
y_{hp} = \frac{(P2 - P1)}{\|P2 - P1\|} \quad (5.6)
\]

\(X_{hp}\): The line perpendicular to the plane formed by \(P_1\), \(P_2\) and \(P_3\), pointing forward.

\[
x_{hp} = \frac{(P2 - P1) \times (P3 - P1)}{\|P2 - P1\| \times (P3 - P1)} \quad (5.7)
\]

\(Z_{hp}\): The line perpendicular to the \(X_{hp}\) and \(Y_{hp}\) axes, pointing to the right.

\[
z_{hp} = \frac{X_{hp} \times Y_{hp}}{\|X_{hp} \times Y_{hp}\|} \quad (5.8)
\]

The position of each epicondyle was then transformed from the global frame to the technical frame of Cluster \(P\) using the following equation:

\[
\overline{LE}_P = \overline{LE}_L \cdot T_p \quad (5.9)
\]

\[
\overline{ME}_P = \overline{ME}_L \cdot T_p \quad (5.10)
\]

Where \(T_p\) is the transformation matrix from the global co-ordinate frame to the technical frame of Cluster \(P\):

\[
T_p = \begin{bmatrix}
x_{hp} & y_{hp} & z_{hp} \\
x_x & y_x & z_x \\
x_y & y_y & z_y \\
x_z & y_z & z_z
\end{bmatrix} \quad (5.11)
\]
In subsequent dynamic trials the LE and ME coordinates were defined relative to the local technical frame of Cluster P and were used in the definition of the humeral anatomical co-ordinate frame as described in Section 5.2.6.

Figure 5-5: Transforming the LEL from the global co-ordinate frame \((X_G, Y_G, Z_G)\) to a local point LEP on the technical frame \((X_h, Y_h, Z_h)\) of the upper arm as defined by the clusters P

### 5.2.5 Glenohumeral Joint centre

The centre of rotation of the glenohumeral joint (GH) was defined using a functional method (Gamage and Lasenby, 2002). The method of Gamage and Lasenby (2002) was chosen as it has been reported to perform better than other sphere-fitting functional methods under the same testing environment (Camomilla et al., 2006; Cereatti et al., 2004; Lempereur et al., 2010). The GH was estimated by capturing the motion of the markers on the humerus, with the elbow flexed at 90°, as it explores most of its possible range of motion (RoM) in all three rotational degrees of freedom in a random manner, in relation to three markers on a scapula locator (Johnson et al., 1993). The scapula locator used to measure the position of the scapula (Johnson et al., 1993) is a tripod device with three pins which can be adjusted so that it can be placed onto three easily palpable bony...
landmarks on the scapula: the acromial angle, root of the scapular spine and inferior angle (Figure 5-6).

A description of how the GH centre was calculated is given in Appendix I.

Figure 5-6: Locator applied on the bowler to locate his scapula during movement

5.2.6 The instantaneous helical axis

The global positions of the bony landmarks in every arm position were used for the construction of a local co-ordinate system (LCS) on the humerus following the ISB recommendations on the definitions of joint co-ordinate systems (Wu et al., 2005).

The origin of the humeral co-ordinate system was the glenohumeral joint centre (GH). The $Y_{h1}$ axis was the line connecting the midpoint of the two elbow epicondyles (EC) to the glenohumeral joint centre (GH) pointing superiorly, the $X_{h1}$ axis was perpendicular to the plane formed by the $Y_{h1}$ axis and the line connecting the two elbow epicondyles (LE
and ME) pointing anteriorly. Finally the $Z_{h1}$ axis was the common line perpendicular to the Y and X axes (pointing laterally).

$O_{h1}$: The origin coincides with GH

$Y_{h1}$: The line connecting GH and EC, pointing towards GH.

$$Y_{h1} = \frac{(GH - EC)}{\|GH - EC\|}$$

(5.12)

$X_{h1}$: The line perpendicular to the plane formed by the Y axis and the line connecting the two elbow epicondyles (ME and LE).

$$X_{h1} = \frac{(GH - EC) \times (LE - ME)}{\|(GH - EC) \times (LE - ME)\|}$$

(5.13)

$Z_{h1}$: The line perpendicular to the $Y_{h1}$ and $X_{h1}$ axes, pointing to the right.

$$Z_{h1} = \frac{Y_{h1} \times X_{h1}}{\|Y_{h1} \times X_{h1}\|}$$

(5.14)

The LCS of the humerus and the forearm cluster were used to determine the rotations of the forearm with respect to the upper arm. This was performed by calculating the position of each forearm marker from the global coordinate frame to the technical frame of humerus using the following equation:

$$\overrightarrow{FA1_c} = \overrightarrow{FA1_G} \cdot T_H$$

(5.15)

where $T_H$ is the transformation matrix from the global coordinate frame to the humeral coordinate frame.

$$T_H = \begin{bmatrix}
X_H & Y_H & Z_H \\
\end{bmatrix} = \begin{bmatrix}
x_x & x_y & x_z \\
y_x & y_y & y_z \\
z_x & z_y & z_z \\
\end{bmatrix}$$

(5.16)

The linear velocity and linear acceleration of the three markers (FA1, FA2 and FA3) onto the forearm cluster were calculated using numerical differentiation (Woltring, 1991).
From the positions and orientations of the forearm cluster markers to the humeral co-ordinate frame instantaneous helical axes (IHAs) were calculated. Each IHA was calculated in least squares sense from landmark motion and described by a position vector \( \overline{P} \) and a unit direction vector \( \overline{n} \).

\[
\overline{P} = \begin{bmatrix} P_x & P_y & P_z \end{bmatrix} \tag{5.17}
\]
\[
\overline{n} = \begin{bmatrix} n_x & n_y & n_z \end{bmatrix} \tag{5.18}
\]

Following this the angular acceleration was calculated by deriving the angular velocity vector \( \omega \). The position \( \overline{P} \) and unit direction \( \overline{n} \) vectors were subsequently calculated based on algorithms according to Woltring (1980):

\[
\omega = \sqrt{\overline{\omega}^T \cdot \overline{\omega}} \tag{5.19}
\]
\[
\overline{n} = \frac{\overline{\omega}}{\omega} \tag{5.20}
\]
\[
\nu = \dot{p}^T \cdot \overline{n} \tag{5.21}
\]
\[
\overline{P} = p + \frac{\overline{\omega} \cdot \dot{p}^T}{\omega^2} \tag{5.22}
\]

where \( \nu \) is the translational velocity, \( p \) the position vector of each landmark in the humeral co-ordinate frame and \( \dot{p} \) is its derivative. Both the mean flexion-extension F/E and pronation-supination P/S axes were calculated employing the same methodology for the two single joint movements of pure flexion and pronation respectively.

The position and orientation vectors of the IHAs for the dynamic trial were used to compute \( p \) and \( n \) of the optimal IHA for each subject (Stokdijk et al., 1999).

\[
\overline{P}_{\text{optimal}} = Q^{-1} \cdot \frac{1}{N} \sum_{i=1}^{N} Q_i \cdot \overline{P}_i \tag{5.23}
\]
\[
Q_i = I - n_i^T \cdot n_i \tag{5.24}
\]
\[
Q = \frac{1}{N} \sum_{i=1}^{N} Q_i \tag{5.25}
\]
Figure 5-7: 3D representation of the bony landmarks of the upper arm and forearm with the elbow joint bent (Glenohumeral rotation centre (GH), medial and lateral epicondyles (ME and LE), ulnar and radial styloid processes (US and RS)) and the calculated IHA (black arrows), mean optimal pivot points (P) and axes (red arrows) for one subject as generated during (a) a motion of pure flexion-extension and (b) a movement of pure pronation-supination.
As a low angular velocity (under 0.25 rad/sec) can lead to inaccurate calculation of angular acceleration (Stokdijk et al., 1999) and cause outliers, therefore, only the middle 60° of every controlled movement were taken into account (Stokdijk et al., 2000) when calculating the optimal position vectors $P_{\text{F/E}}$ and $P_{\text{P/S}}$ for flexion-extension and pronation-supination axes respectively. Analogous to the calculation of $P_{\text{optimal}}$ an optimal unit direction vector ($n_{\text{optimal}}$) was calculated (Stokdijk et al., 1999). During the bowling action the elbow F/E functional axis is expressed in the LCS of the humerus and the P/S axis is then expressed in the LCS of the forearm I.

The origin of the local co-ordinate frame of the forearm I coincided with the marker on the ulnar styloid (US), the $Y_{F1}$ axis was the line joining the ulnar styloid and the midpoint of the two elbow epicondyles pointing superiorly, the $X_{F1}$ axis was perpendicular to the plane formed by the midpoint of the two epicondyles and the two markers the ulnar and radial styloids. Finally, the $Z_{F1}$ axis was defined as the common perpendicular to the $Y_{F1}$ and $X_{F1}$ axes (Wu et al., 2005)

$O_{F1}$: The origin coincides with US

$Y_{F1}$: The line connecting US and EC, pointing towards EC.

$$Y_{F1} = \frac{(EC - US)}{\|EC - US\|}$$

(5.26)

$X_{F1}$: The line perpendicular to the plane formed by the $Y$ axis and the line connecting the two markers the ulnar and radial styloids (US and RS).

$$X_{F1} = \frac{(EC - US) \times (US - RS)}{\|(EC - US) \times (US - RS)\|}$$

(5.27)

$Z_{F1}$: The line perpendicular to the $Y_{F}$ and $X_{F}$ axes, pointing to the right.

$$Z_{F} = \frac{Y_{F1} \times X_{F1}}{\|Y_{F1} \times X_{F1}\|}$$

(5.28)
5.2.7 Segmental Co-ordinate Frames

**Upper arm**

The origin of the upper arm ($O_u$) co-ordinate system was the glenohumeral joint centre (GH). The $Z_u$ axis was defined parallel to the optimal F/E axis $n_{F/E}$ (positive laterally), the $X_u$ axis was perpendicular to this axis and the line connecting the optimal pivot point for flexion ($P_{F/E}$) and the glenohumeral joint centre (GH) (positive anteriorly), and the $Y_u$ axis was the common line perpendicular to the $Z_u$ and $X_u$ axes (positive superiorly).

$O_u$: The origin coincides with GH

$Z_u$: The line along the functional F/E axis

$$Z_u = \frac{n_{F/E}}{\|n_{F/E}\|}$$

(5.29)

$X_u$: The line perpendicular to the plane formed by the $Z$ axis and the line connecting $P_{F/E}$ and GH

$$X_u = \frac{Z_u \times (P_{F/E} - GH)}{\|Z_u \times (P_{F/E} - GH)\|}$$

(5.30)

$Y_u$: The line perpendicular to the $Z_u$ and $X_u$ axes

$$Y_u = \frac{Z_u \times X_u}{\|Z_u \times X_u\|}$$

(5.31)

**Forearm**

The origin of the local co-ordinate frame of the forearm coincided with the ulnar styloid process, the $Y_f$ axis was defined parallel to the optimal P/S axis $n_{P/S}$ (pointing superiorly), and the $X_f$ axis was perpendicular to the plane formed by the $Y_f$ axis and the two markers the ulnar and radial styloids. Finally, the $Z_f$ axis was once again defined as the common perpendicular to the $Y_f$ and $X_f$ axes.
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**O_P:** The origin coincides with the marker on the ulnar styloid process

**Y_F:** The line along the functional P/S axis

\[ Y_F = \frac{\vec{n}_{P/S}}{\|\vec{n}_{P/S}\|} \]  

(5.32)

**X_F:** The line perpendicular to the plane formed by the Y axis and the line connecting the two markers the ulnar and radial styloids (US and RS).

\[ X_F = \frac{Y_F \times (US - RS)}{\|Y_F \times (US - RS)\|} \]  

(5.33)

**Z_F:** The common perpendicular to the Y_F and X_F axes.

\[ Z_F = \frac{Y_F \times X_F}{\|Y_F \times X_F\|} \]  

(5.34)

**5.2.8 Joint Rotations**

Elbow rotations were calculated using Euler angles with a z - x’ - y’’ Cardan sequence, where rotations about z, y, and x are the flexion-extension (flexion positive), pronation-supination (pronation positive) and adduction-abduction (adduction positive) as shown in Figure 5-8. Euler rotations are a means of representing the spatial orientation of any frame in space as a composition of rotations from a reference frame (Elliott et al., 2007) and are the most frequently used in the analysis of upper body kinematics (Cutti et al., 2008; Meskers et al., 1998; Wu et al., 2002). Euler angles are dependent on the sequence of the rotations. Even though the carrying angle is not an independent degree of freedom, studies suggest that when calculating elbow rotations between the segments of the upper arm and forearm it must be considered as the second rotation following flexion/extension but preceding pronation/supination (Anglin and Wyss, 2000).

A description of how Euler rotations are calculated is given in Appendix II.
5.2.9 Data Normalisation and Interpolation

To free the data from the effects caused by variations in speed across different trials within the same subject as well as between different subjects and hence make it more suitable for comparison purposes, the data were normalised to 100% of the bowling action (Figure 5-9) that was defined from 20 ms prior to upper arm horizontal to 20 ms after ball release. This is in accordance with the English Cricket Board’s (ECB) regulations for the ‘Review of Bowlers Reported with Suspected Illegal Bowling Actions’ (ECB, 2010).
To account for small variations within each bowler in terms of speed, a piecewise cubic polynomial approximation (R2007a, The MathWorks, Natick, USA) was used to interpolate between the data points (Figure 5-10).

The piecewise cubic polynomial approximation divides the interval between the data points into subintervals and provides the piecewise polynomial form of a cubic Hermite interpolant for each interval so that the shape of the data is preserved.

5.2.10 Means and Standard Deviations

The data for each bowler were averaged over the 6 best trials, and presented as the mean elbow angle and standard deviation as this is in accordance with the regulations for the review of bowlers reported with suspected illegal bowling action (ICC, 2005).
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The standard deviation is a measure of the spread of the distribution about the mean and was used to measure the variability between the different trials.

\[
\sigma = \sqrt{\frac{1}{N-1} \sum_{i=1}^{N} (x_i - \bar{x})^2}
\]  

(5.35)

Where \(\sigma\) is the standard deviation, \(N\) is the size of the sample and \(\bar{x}\) is the mean value given by the following equation:

\[
\bar{x} = \frac{1}{N} \sum_{i=1}^{N} x_i
\]  

(5.36)

For each subject the standard deviation (SD) was calculated as a measure of within trial variability of the elbow flexion angles.

Figure 5-11: Average (in black) flexion (+ve)/extension (-ve) angles for four trials of the same subject; the standard deviations between the trials are shown as error bars

5.2.11 Statistical analysis

The coefficient of multiple correlation (CMC) \(r\) is a statistical measure of waveform similarity that has been used to evaluate the overall similarity between waveforms in gait analysis (Kadaba et al., 1989) and shoulder and scapular motion (Amasay and Karduna, 2009; Meskers et al., 1998). The (CMC) takes values that range from 0 (dissimilar waveforms) to 1 (similar waveforms) and was originally proposed by Kadaba
in order to test the inter subject variability in two formulations, named *within-day* and *between-day* respectively.

In this study, the *within-day* (CMC) was used to evaluate the repeatability of six trials of elbow flexion/extension angles generated in each subject.

$$CMC = \frac{1 - \frac{\sum_{s=1}^{S} \sum_{i=1}^{W} \sum_{j=1}^{N} (Y_{sij} - \bar{Y}_s)^2}{SN(W - 1)}}{\frac{\sum_{s=1}^{S} \sum_{i=1}^{W} \sum_{j=1}^{N} (Y_{sij} - \bar{Y}_s)^2}{SN(W - 1)}}$$

(5.37)

Where $s$ is the number of the experimental days/sessions, $j$ is the % of the bowling action, $Y_{sij}$ is subject-specific elbow angle, $N$ is the total number of the % of the bowling action increments, $W$ is the total number of trials, $\bar{Y}_s$ is the average elbow angle at frame $j$, of the average waveform among the $W$ waveforms of session $s$ and finally, $\bar{Y}_s$ is the total average elbow angle.

For each plot, the protocol *within-day* reliability was interpreted as follows (Garofalo et al., 2009)

- 0.65<CMC<0.75 moderate
- 0.75<CMC<0.85 good
- 0.85<CMC<0.95 very good
- 0.95<CMC<1 excellent

In this study the *within-day* (CMC) was used as a measure of the inter-subject variability of the waveforms for every joint rotation whereas the standard deviation (SD) was used as a measure of the variability in the degree of joint rotation between two points: (1) the point that the upper arm reached the level of shoulder and (2) ball release between different trials.
5.3 Results and Discussion

Average elbow extension angles during the ball delivery were 13.8° (SD: 3.7°, Range 8.7° - 20.7°). Elbow angles for each bowler were averaged over 6 trials and presented in Figure 5-12 as the mean elbow angle and standard deviation. The within-day CMC was greater than 0.80 for all three rotations (Table 5-2) suggesting good consistency and repeatability of the kinematic patterns obtained in the six trials for each subject.

![Graph showing mean elbow flexion-extension angles for a fast bowler plotted against normalised % of bowling action (t=0.120 sec)](image)

Figure 5-12: Mean elbow flexion-extension angles for a fast bowler plotted against normalised % of bowling action (t=0.120 sec)

Mean elbow flexion-extension graphs for every bowler are presented in Appendix III. Results showed that all the bowlers but one that participated in the study were within the ICC threshold tolerance whilst, two bowlers were borderline in that they exceeded the 15° limit in individual trials. Medium pace bowlers showed on average 14.0° (2.3°) of elbow extension, 13.3° (2.8°) of adduction and 27.5° (4.7°) of pronation whereas fast bowlers showed 11.5° (3.8°), 9.8° (2.8°) and 26.2° (7.7°) respectively. Both these groups exceeded the first set of ICC regulations whereby fast bowlers were allowed 10° and medium pace bowlers 7.5° of elbow extension but were within the 2004 ICC regulations of 15°. All of the recorded bowling deliveries exceeded the original threshold of 5° for
spin bowlers. Even though the number of bowlers in each group does not allow for a statistical comparison between the groups it can be seen that fast and medium pace bowlers showed similar ranges of pronation and adduction while spin bowlers showed a greater range of pronation.

Table 5-2: Mean elbow angles, *within-day* (CMC) *r* values of elbow kinematics during bowling

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>At upper arm horizontal</th>
<th>Mean (SD)</th>
<th>Flexion-Extension</th>
<th>Mean (SD)</th>
<th>Adduction-Abduction</th>
<th>Mean (SD)</th>
<th>Pronation-Supination</th>
<th>Mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Med</td>
<td>12.4 (1.1)</td>
<td>15.8 (1.8)</td>
<td>0.912</td>
<td>12.5 (1.9)</td>
<td>0.873</td>
<td>27.1 (1.1)</td>
<td>0.894</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>Med</td>
<td>4.5 (1.6)</td>
<td>14.4 (3.5)</td>
<td>0.940</td>
<td>9.6 (1.7)</td>
<td>0.920</td>
<td>30.8 (4.6)</td>
<td>0.877</td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>Fast</td>
<td>10.7 (1.4)</td>
<td>10.0 (1.8)</td>
<td>0.862</td>
<td>6.5 (0.5)</td>
<td>0.882</td>
<td>26.7 (5.6)</td>
<td>0.980</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>Fast</td>
<td>0.0 (2.2)</td>
<td>8.7 (0.7)</td>
<td>0.974</td>
<td>11.5 (2.7)</td>
<td>0.953</td>
<td>18.3 (5.5)</td>
<td>0.977</td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>Spin</td>
<td>42.0 (0.7)</td>
<td>20.7 (2.1)</td>
<td>0.861</td>
<td>20.5 (7.5)</td>
<td>0.925</td>
<td>38.0 (7.6)</td>
<td>0.972</td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>Fast</td>
<td>26.0 (1.7)</td>
<td>15.9 (1.5)</td>
<td>0.879</td>
<td>11.3 (2.7)</td>
<td>0.882</td>
<td>33.6 (2.9)</td>
<td>0.948</td>
<td></td>
</tr>
<tr>
<td>7</td>
<td>Med</td>
<td>4.9 (3.9)</td>
<td>10.7 (2.0)</td>
<td>0.864</td>
<td>15.7 (1.3)</td>
<td>0.967</td>
<td>31.1 (5.7)</td>
<td>0.908</td>
<td></td>
</tr>
<tr>
<td>8</td>
<td>Med</td>
<td>32.0 (2.8)</td>
<td>15.2 (2.5)</td>
<td>0.910</td>
<td>15.2 (3.1)</td>
<td>0.975</td>
<td>21.0 (5.2)</td>
<td>0.945</td>
<td></td>
</tr>
<tr>
<td>9</td>
<td>Spin</td>
<td>-5.5 (1.2)</td>
<td>13.2 (2.0)</td>
<td>0.985</td>
<td>5.1 (0.7)</td>
<td>0.879</td>
<td>58.5 (5.7)</td>
<td>0.979</td>
<td></td>
</tr>
<tr>
<td>Mean</td>
<td></td>
<td>13.8 (3.7)</td>
<td>0.91 (0.05)</td>
<td>12.0 (4.8)</td>
<td>0.92 (0.04)</td>
<td>31.7 (11.7)</td>
<td>0.94 (0.04)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
With respect to the elbow kinematics during bowling, flexion-extension graphs revealed that in the legal action, the bowler’s arm can either flex (Figure 5-13a), maintain the same angle (Figure 5-13b), or extend prior to ball release (Figure 5-13c).

---

**Figure 5-13:** Mean elbow flexion-extension angles for three bowlers plotted against normalised % of bowling action
Three bowlers bowled with a bent arm exceeding 20° of flexion, two of which had mean elbow extension angles of 15° and one was a thrower. If we look closer at the thrower (Figure 5-14) it can be seen that he had elbow extension angles of 20.1°, 23.9°, 21.9°, 20.7°, 17.7° and 19.7°. Flexion-extension graphs revealed that when his upper arm reached the level of the shoulder he already had a bent arm at approximately 42° and subsequently extended up to 25° at ball release. This is a significant amount of extension which can allow for humeral internal rotation to contribute to ball release velocity (Ferdinands and Kersting, 2007; Marshall and Ferdinands, 2005).

**Figure 5-14:** Elbow flexion-extension angles of five trials and their average (in black) for one spin bowler plotted against normalised % of bowling action
5.4 Summary

In this study a biomechanical model is presented in order to measure elbow kinematics in cricket bowling. A total of nine bowlers participated in the study and excluding one thrower the sample mean of elbow extension angle were within the 15° threshold, irrespective of bowling type or speed. The within-day CMC for all three rotations ranged from 0.80 - 0.99 suggesting that the angles collected during the delivery stride were reproducible between trials and that the proposed kinematic model based on a helical approach of the elbow flexion-extension and pronation-supination axes can be used to accurately measure bowling legality in cricket.

When assessing bowling legality different laboratories have employed different models to analyse these actions but as the accuracy of the predictions of these models depends highly on the accuracy and repeatability of the input data the results from these laboratories may not be directly comparable. The following chapter is therefore devoted to investigating the accuracy of the proposed functional based kinematic model when compared to anatomical based approaches through sensitivity analysis.

Finally, half of the bowlers that participated in this study had their elbows flexed at upper arm horizontal more than 10° and in almost one third of them the elbow flexion angle exceeded 20° (Table 5-2). The question that arises is that as the number of bowlers who tend to have a bent arm at upper arm horizontal is significant is it possible to discriminate between those who passively extend their joint due to anatomical factors and those who actively extend aiming at increasing the ball’s speed through release? This is a matter that will be addressed in Chapter 7 where anthropometric data for each bowler will be analysed and presented.
Chapter 6  Sensitivity analysis

In the previous chapter a functional based kinematic model to measure elbow biomechanics during cricket bowling was presented. In this chapter the model’s sensitivity to anatomical based axes, single and double anatomical landmark calibration, markers placed directly onto the epicondyle is assessed.
6.1 Introduction

Sensitivity analysis is the study of how the variation in the output data of a mathematical model can be affected, to different variables in the input data. This analysis is a way to predict how a change in the input parameters causes a change in the dynamic behaviour of the model. By showing how the model behaviour responds to changes in parameter values, sensitivity analysis is a useful tool in model building as well as in model evaluation (Saltelli et al., 2000).

This chapter presents a sensitivity analysis following the one-parameter at a time principle whereby the effect of each parameter is assessed independently from all other parameters. In all the cases the sensitivity was assessed quantitatively by calculating the average elbow angles, standard deviations and correlation coefficients. In this chapter the model sensitivity to anatomical based axes, elbow orientation and choice of uppers arm triad during single anatomical landmark calibration of the humeral epicondyles, double landmark calibration and skin markers onto the epicondyles are investigated as these are the most common methods employed in motion analysis studies as discussed in Chapter 4. Finally the model’s sensitivity to the identification of the point of ball release is also investigated.
6.2 Study I: Anatomical and functional based upper limb models

As discussed in Chapter 5 biomechanical models used to analyse upper-limb joint kinematics are distinguished into two categories; the anatomical based and the functional based. Both models employ the same fundamental principles in order to determine joint kinematics. The positional data of the markers in space are used to create technical/local co-ordinate systems (LCS) whereby joint centres and axes are defined relative to and which are subsequently used to define the segmental co-ordinate frames. Anatomical based models take into account the relative movement between the co-ordinate frames of two adjacent segments for the calculation of joint kinematics (Grood and Suntay, 1983) whereas, functional models optimise the relative movement between the of two adjacent co-ordinate frames based on joint constraints.

6.2.1 Aims and Objectives

The aim of this study was to compare the functional model presented in Chapter 5 (FuncM1) to an anatomical based model (AnatM) and to another functional based model proposed (FuncM2) by Chin (2009). To evaluate the proposed functional model when measuring elbow kinematics all three elbow rotations were compared during two single joint movements and six bowling trials in terms of range of motion and correlation coefficient.

6.2.2 Materials and Methods

6.2.2.1 Study Population

A total of 9 male bowlers, 7 medium - fast and 2 spin bowlers, from the MCC Young Cricketers and staff participated in this study. Their mean age was 21 years (range: 16-31 years) and mean body mass index was 22.3 (range: 19 - 27.8). All testing took place at the MCC’s indoor cricket school as described in Section 5.2.1.
6.2.2.2 Equipment and Lab set-up

An optical motion tracking system (Vicon, Oxford, UK) and two video recorders were used at an acquisition rate of 200 Hz and 100 Hz respectively (Section 5.2.1), to track reflective markers attached to the skin of the bowlers as described in Section 5.2.2.

6.2.2.3 Experimental Protocol

The experimental protocol used was the same for all subjects as described in Section 5.2.3 and is as follows:

- A static trial of the marker setup
- One static trial to digitise the LE position
- One static trial to digitise the ME position
- A dynamic trial to estimate the GH position

After digitisation every bowler performed five repeated measures of a pure flexion-extension movement, five repeated measures of a pure pronation-supination movement (Section 5.2.4) and total of 16 deliveries. It should be noted that each bowler was asked to warm up on their own prior to data capture.

6.2.2.4 Data analysis

Data analysis was completed as described in Section 5.2.4. The digitisation of the LE and ME positions were the same as described in Section 5.2.4.1 and the centre of rotation for the glenohumeral joint GH was estimated as described in Section 5.2.5.

For the anatomical based model (AnatM) the co-ordinate frames of the segments were defined according to the ISB recommendations (Wu et al., 2005). For the construction of the second definition of a functional kinematic model (FuncM2) elbow flexion-extension and pronation-supination helical axes were defined based on the relative movement of the forearm with respect to the upper arm as described in Section 5.2.6 and then co-ordinate frames of the adjacent segments were introduced as follows:
Chapter 6  
Sensitivity analysis

**Upper arm**

The origin of the upper arm (O_u) co-ordinate system was the pivot point for flexion (P_{F/E}). The Z_u axis was defined parallel to the optimal F/E axis n_{F/E} (positive laterally), the X_u axis was perpendicular to the Z_u axis and the line connecting the optimal pivot point for flexion (P_{F/E}) and the glenohumeral joint centre (GH) (positive anteriorly). The Y_u axis was the common line perpendicular to the Z_u and X_u axes (positive superiorly).

**Forearm**

Three forearm co-ordinate frames were defined, the first to calculate elbow flexion and extension and the second and third were used for the definition of pronation and supination.

**Forearm 1**

The origin of the first local co-ordinate frame of the forearm coincided with the midpoint between the radial and ulnar styloid processes termed the wrist joint centre (WC), the Z_{F1} axis was defined parallel to the optimal F/E axis n_{F/E} (positive superiorly) and the X_{F1} axis was perpendicular to the plane formed by the Z_{F1} axis and the line joining the wrist joint centre (WC) to the optimal pivot point for flexion (P_{F/E}). Finally, the Y_{F1} axis was defined as the common perpendicular to the Z_{F1} and X_{F1} axes.

**Forearm 2**

The origin of the local co-ordinate frame was the wrist joint centre (WC), the Y_{F2} axis was defined parallel to the optimal P/S axis n_{P/S} (positive superiorly) and the X_{F2} axis was defined perpendicular to the plane formed by Y_{F2} and the optimal F/E axis n_{F/E} (positive anterior), and finally, the Z_{F2} axis was defined as the common perpendicular to the Y_{F2} and X_{F2} axes (positive laterally).

**Forearm 3**

The origin of the co-ordinate frame coincided with the wrist joint centre, the Y_{F3} axis was defined parallel to the optimal P/S axis n_{P/S} and the X_{F3} axis was perpendicular to the
plane formed by the $Y_{F3}$ axis and the two markers the ulnar and radial styloids. Finally, the $Z_{F3}$ axis was once again defined as the common perpendicular to the $Y_{F3}$ and $X_{F3}$ axes.

The co-ordinate frames for all three kinematic models and the joint rotations sequence to calculate elbow angles are summarised in Table 6-1. Elbow rotations between the co-ordinate systems were calculated using Euler angles.

To free the data from the effects caused by variations in speed across different trials within the same bowler the data were normalised to 100% of the bowling action (defined from 20 ms prior to upper arm horizontal to 20 ms after ball release) and interpolated as described in Section 5.2.9.

### 6.2.2.5 Statistical analysis

The data for each bowler were averaged over the 6 best trials, and presented as the mean elbow angle and standard deviation (Section 1.2.10) which is in accordance with the regulations for the review of bowlers reported with suspected illegal bowling action (ICC, 2005). The *within-day* coefficient of multiple correlation (CMC) $r$ (Kadaba et al., 1989) was used to evaluate the repeatability of six trials of elbow extension angles generated in each subject. Statistical differences between methods were assessed using a repeated measures analysis of variance (ANOVA) while non parametric Friedman’s ANOVA was used to assess any differences between the CMC values. Statistical significance was set at an alpha level of 0.05.
### Chapter 6 Sensitivity analysis

#### Table 6-1: Upper arm and forearm segmental co-ordinate frame definitions and joint rotations for the three kinematic models

<table>
<thead>
<tr>
<th>Kinematic Model</th>
<th>Segment</th>
<th>Co-ordinate Frames</th>
<th>Rotations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper Arm</td>
<td></td>
<td>$Y_{hl} = (GH - EC) | [GH - EC]$</td>
<td>Sequence: $z - x' - y''$</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$X_{hl} = Y_{hl} \times (LE - ME) | [Y_{hl} \times LE - ME]$</td>
<td>1&lt;sup&gt;st&lt;/sup&gt;: Flexion-extension (flexion positive)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Z_{hl} = Y_{hl} \times X_{hl} | [Y_{hl} \times X_{hl}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Anatomical (AnatM)</td>
<td>$Y_{f1} = (EC - US) | [EC - US]$</td>
<td>2&lt;sup&gt;nd&lt;/sup&gt;: Adduction-abduction (adduction positive)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$X_{f1} = Y_{f1} \times (US - RS) | [Y_{f1} \times US - RS]$</td>
<td>3&lt;sup&gt;rd&lt;/sup&gt;: Pronation-supination (pronation positive)</td>
</tr>
<tr>
<td></td>
<td>Forearm</td>
<td>$Z_{f1} = Y_{f1} \times X_{f1} | [Y_{f1} \times X_{f1}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Y_{f} = n_{PS} | [n_{PS}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Functional (FuncM1)</td>
<td>$Z_{u} = n_{F/E} | [n_{F/E}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$X_{u} = Z_{u} \times (EC - GH) | [Z_{u} \times EC - GH]$</td>
<td>1&lt;sup&gt;st&lt;/sup&gt;: Flexion-extension (flexion positive)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Y_{u} = Z_{u} \times X_{u} | [Z_{u} \times X_{u}]$</td>
<td>2&lt;sup&gt;nd&lt;/sup&gt;: Adduction-abduction (adduction positive)</td>
</tr>
<tr>
<td></td>
<td>Forearm</td>
<td>$X_{f} = Y_{f} \times (US - RS) | [Y_{f} \times US - RS]$</td>
<td>3&lt;sup&gt;rd&lt;/sup&gt;: Pronation-supination (pronation positive)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Z_{f} = Y_{f} \times X_{f} | [Y_{f} \times X_{f}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Anatomical (AnatM)</td>
<td>$Y_{f1} = (EC - US) | [EC - US]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$X_{f1} = Y_{f1} \times (US - RS) | [Y_{f1} \times US - RS]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Forearm</td>
<td>$Z_{f1} = Y_{f1} \times X_{f1} | [Y_{f1} \times X_{f1}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Y_{f2} = n_{PS} | [n_{PS}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Functional (FuncM2)</td>
<td>$Z_{u} = n_{F/E} | [n_{F/E}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$X_{u} = Z_{u} \times (EC - GH) | [Z_{u} \times EC - GH]$</td>
<td>1&lt;sup&gt;st&lt;/sup&gt;: Flexion-extension (flexion positive)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Y_{u} = Z_{u} \times X_{u} | [Z_{u} \times X_{u}]$</td>
<td>2&lt;sup&gt;nd&lt;/sup&gt;: Adduction-abduction (adduction positive)</td>
</tr>
<tr>
<td></td>
<td>Forearm1</td>
<td>$X_{f1} = Z_{f1} \times (P_{F/S} - WC) | [Z_{f1} \times P_{F/S} - WC]$</td>
<td>3&lt;sup&gt;rd&lt;/sup&gt;: Pronation-supination (pronation positive)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Y_{f1} = Z_{f1} \times X_{f1} | [Z_{f1} \times X_{f1}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Forearm2</td>
<td>$Y_{f2} = n_{PS} | [n_{PS}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Z_{f2} = Y_{f2} \times X_{f2} | [Y_{f2} \times X_{f2}]$</td>
<td>1&lt;sup&gt;st&lt;/sup&gt;: Pronation-supination (pronation positive)</td>
</tr>
<tr>
<td></td>
<td>Forearm3</td>
<td>$Y_{f3} = n_{PS} | [n_{PS}]$</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Z_{f3} = Y_{f3} \times X_{f3} | [Y_{f3} \times X_{f3}]$</td>
<td>2&lt;sup&gt;nd&lt;/sup&gt;: Adduction-abduction (adduction positive)</td>
</tr>
<tr>
<td></td>
<td>Forearm2</td>
<td>$X_{f3} = Y_{f3} \times (US - RS) | [Y_{f3} \times US - RS]$</td>
<td>3&lt;sup&gt;rd&lt;/sup&gt;: Flexion-extension (flexion positive)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$Z_{f3} = Y_{f3} \times X_{f3} | [Y_{f3} \times X_{f3}]$</td>
<td></td>
</tr>
</tbody>
</table>
Chapter 6  Sensitivity analysis

6.2.3  Results

This study compared kinematic data obtained from one anatomical (AnatM) and two functional based models (FunM1, FuncM2) during two single joint movements and during cricket bowling. Statistical analysis showed that during a movement of pure flexion-extension all three models calculated the same amount of flexion while, the functional models calculated significantly less adduction and pronation than the anatomical based model (Table 6-2, p<0.01).

Table 6-2: 95% confidence intervals, mean ranges of motion and probability values (p-value) for all three rotations during a controlled movement of pure flexion – extension

<table>
<thead>
<tr>
<th>Kinematic Model</th>
<th>Elbow Rotations</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean (°)</td>
</tr>
<tr>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion-Extension</td>
<td></td>
</tr>
<tr>
<td>AnatM</td>
<td>135.71</td>
</tr>
<tr>
<td>FuncM1</td>
<td>137.48</td>
</tr>
<tr>
<td>FuncM2</td>
<td>137.16</td>
</tr>
<tr>
<td>Adduction-Abduction</td>
<td></td>
</tr>
<tr>
<td>AnatM</td>
<td>15.44</td>
</tr>
<tr>
<td>FuncM1</td>
<td>9.52</td>
</tr>
<tr>
<td>FuncM2</td>
<td>0.00</td>
</tr>
<tr>
<td>Pronation-Supination</td>
<td></td>
</tr>
<tr>
<td>AnatM</td>
<td>22.53</td>
</tr>
<tr>
<td>FuncM1</td>
<td>17.44</td>
</tr>
<tr>
<td>FuncM2</td>
<td>17.44</td>
</tr>
</tbody>
</table>

* The mean difference is significant at the 0.05 level.

** The mean difference is significant at the 0.01 level.

*** The mean difference is significant at the 0.001 level.
Chapter 6  Sensitivity analysis

The repeated measures ANOVA determined that during controlled movements of pure pronation-supination the range of the recorded motion for flexion and adduction differed between the three kinematic models whilst all three models calculated the same range of pronation (Table 6-3). Post hoc analysis revealed that the FuncM1 model measured a greater range of flexion than the AnatM while, the AnatM measured more adduction than FuncM1 and FuncM2.

Table 6-3: 95% confidence intervals, mean ranges of motion and probability values (p-value) for all thee rotations during a controlled movement of pure pronation-supination

<table>
<thead>
<tr>
<th>Kinematic Model</th>
<th>Elbow Rotations</th>
<th>95% Confidence Interval</th>
<th>Model (p-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean (°)</td>
<td>Std. Error</td>
<td>Lower Bound</td>
</tr>
<tr>
<td>Flexion-Extension</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AnatM</td>
<td>13.51</td>
<td>1.67</td>
<td>9.63</td>
</tr>
<tr>
<td>FuncM1</td>
<td>17.29</td>
<td>2.22</td>
<td>12.16</td>
</tr>
<tr>
<td>FuncM2</td>
<td>13.83</td>
<td>1.95</td>
<td>9.32</td>
</tr>
<tr>
<td>Adduction-Abduction</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AnatM</td>
<td>18.23</td>
<td>1.45</td>
<td>14.89</td>
</tr>
<tr>
<td>FuncM1</td>
<td>12.47</td>
<td>1.05</td>
<td>10.03</td>
</tr>
<tr>
<td>FuncM2</td>
<td>0.00</td>
<td>0.00</td>
<td>0.00</td>
</tr>
<tr>
<td>Pronation-Supination</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AnatM</td>
<td>133.68</td>
<td>5.04</td>
<td>122.06</td>
</tr>
<tr>
<td>FuncM1</td>
<td>133.64</td>
<td>5.30</td>
<td>121.43</td>
</tr>
<tr>
<td>FuncM2</td>
<td>133.64</td>
<td>5.30</td>
<td>121.43</td>
</tr>
</tbody>
</table>

* The mean difference is significant at the 0.05 level.

** The mean difference is significant at the 0.01 level.

*** The mean difference is significant at the 0.001 level.
Kinematic waveforms of all three rotations during the two single joint movements of pure flexion and pronation are shown in Figure 6-1 and Figure 6-2 respectively. These were identical between the two functional models for pronation (Figure 6-1c, Figure 6-2c) while, a mean offset of 10.0° (±6.0°) existed between the AnatM and the FuncM1&2 (Figure 6-2c). Furthermore, all three models had similar shaped graphs for flexion-extension though again a mean offset of 6.4° (±4.0°) existed between the AnatM and FuncM1 curves (Figure 6-1a). The greatest difference between the models appeared for adduction-abduction where FuncM1 measured significantly less motion than the AnatM while with the FuncM2 no adduction was measured.
Figure 6-1: Kinematic waveforms of one bowler during an active movement of pure flexion-extension
Figure 6-2: Kinematics of one bowler during an active movement of pure pronation-supination
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Average elbow angles during the ball delivery are presented in Table 6-4. No significant differences were found between the three kinematic models for flexion while the functional based models calculated significantly less pronation than the anatomical \((p<0.01)\). Significant differences between the models were also reported for adduction \((p<0.001)\), post hoc analysis however showed that this was due to FunM2 that does not measure any adduction. The mean within-day CMC was greater than 0.80 for all three rotations suggesting good consistency of the kinematic patterns obtained in the six trials for each subject (Figure 6-3).

Table 6-4: Mean elbow angles, within-day CMC, 95% confidence intervals and probability values (\(p\)-value) for all thee elbow rotations during cricket bowling

<table>
<thead>
<tr>
<th>Elbow Rotations</th>
<th>Kinematic Model</th>
<th>Mean (°)</th>
<th>Std. Error</th>
<th>Within-day CMC (SD)</th>
<th>95% Confidence Interval</th>
<th>Model ((p)-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flexion-Extension</td>
<td>AnatM</td>
<td>13.44</td>
<td>1.99</td>
<td>0.917 (0.08)</td>
<td>8.84 - 18.05</td>
<td>0.651</td>
</tr>
<tr>
<td></td>
<td>FuncM1</td>
<td>13.84</td>
<td>1.23</td>
<td>0.910 (0.05)</td>
<td>11.01 - 16.68</td>
<td></td>
</tr>
<tr>
<td></td>
<td>FuncM2</td>
<td>12.72</td>
<td>2.19</td>
<td>0.894 (0.13)</td>
<td>7.68 - 17.76</td>
<td></td>
</tr>
<tr>
<td>Adduction-Abduction</td>
<td>AnatM</td>
<td>10.89</td>
<td>1.50</td>
<td>0.894 (0.08)</td>
<td>7.42 - 14.35</td>
<td>0.000***</td>
</tr>
<tr>
<td></td>
<td>FuncM1</td>
<td>11.99</td>
<td>1.58</td>
<td>0.917 (0.04)</td>
<td>8.33 - 15.64</td>
<td></td>
</tr>
<tr>
<td></td>
<td>FuncM2</td>
<td>0.00</td>
<td>0.00</td>
<td>-</td>
<td>0.00 - 0.00</td>
<td></td>
</tr>
<tr>
<td>Pronation-Supination</td>
<td>AnatM</td>
<td>41.56</td>
<td>3.93</td>
<td>0.944 (0.04)</td>
<td>32.48 - 50.63</td>
<td>0.008**</td>
</tr>
<tr>
<td></td>
<td>FuncM1</td>
<td>31.68</td>
<td>3.91</td>
<td>0.942 (0.04)</td>
<td>22.65 - 40.70</td>
<td></td>
</tr>
<tr>
<td></td>
<td>FuncM2</td>
<td>31.68</td>
<td>3.91</td>
<td>0.942 (0.04)</td>
<td>22.65 - 40.70</td>
<td></td>
</tr>
</tbody>
</table>

** The mean difference is significant at the 0.01 level.

*** The mean difference is significant at the 0.001 level.
6.2.4 Discussion

This study tested the effect that different definitions of the elbow flexion-extension and pronation-supination axes may have on quantifying elbow kinematics during cricket bowling by comparing the repeatability of the kinematic data obtained from three models, one anatomical (AnatM) and two functional based (FuncM1 and FuncM2) during two single joint movements and during cricket bowling.

Functional and anatomical based axes definitions can be directly compared when looking at the AnatM and FuncM1 models. For both kinematic models co-ordinate frames were established in the same way and elbow rotations were calculated using Euler angles with the same rotation sequence. However, while the AnatM relied solely on the distance between boney landmarks for the definition of the flexion and pronation axes the FuncM1 employed functional axes determined by moving the joint through a functional range of motion. Results showed that during both single joint movements the FuncM1 was less prone to kinematic crosstalk as large ranges of adduction were recorded with the AnatM. Both models produced similar shaped graphs for flexion-extension however; an offset existed between their curves. This offset could be due to
slight location errors of the ME and LE landmarks (Chin 2009) but also because the transepicondylar axis of the AnatM and the helical axis of the FuncM1 are not parallel.

When comparing the two functional models even though, FuncM2 employed the same flexion-extension and pronation-supination helical axes as FuncM1 it measured no abduction by sharing a common flexion axis between the upper arm and forearm (Chin, 2009). Pronation-supination kinematic waveforms and recorded range of motion were identical between the two functional models for all tasks.

During ball delivery no statistical differences were found between the three kinematic models for flexion while the \textit{within-day} CMC was greater than 0.85 suggesting that the flexion angles collected during the delivery stride were reproducible between trials and that all three kinematic models showed similar intra-tester repeatability. The functional models however, calculated significantly less pronation than the anatomical. These results are in agreement with previous studies that have shown that the optimal helical knee flexion-extension axis and the transepicondylar axis produce similar kinematic and kinetic data (Besier et al., 2003; Churchill et al., 1998).
6.3 Study II: The effect of digitisation of the humeral epicondyles on quantifying elbow kinematics during cricket bowling

As described in Chapter 5 when working with anatomical co-ordinate frames for the elbow joint the largest skin deformation is observed around the humeral epicondyles because of the loose skin surrounding the joint area (Cappozzo et al., 1996; Elliott et al., 2007; Leardini et al., 2005). Cappozzo et al., (1995) proposed a method to calibrate the positions of certain anatomical landmarks which are either not practical for use in dynamic experiments or can introduce high errors, in their case the knee epicondyles and the great trochanter of the hip joint, relative to the tip of a pointer of known dimensions. The anatomical landmark positions could then be defined relative to a technical frame on the same segment in a static trial and then during dynamic trials their position is expressed with respect to an upper-arm ‘cluster of markers’. From this definition it follows that during each given dynamic trial the position of the humeral epicondyles is linked to the upper-arm cluster and thus affected by the same skin movement artefact that affects the cluster (Cutti et al., 2006). In Section 6.2 the two functional based kinematic models were compared with an anatomical based model where the positions of the two elbow epicondyles were digitised with the elbow bent at 90°. The aim of to this study is to compare four different elbow orientations and two upper arm clusters for digitising the humeral epicondyles and investigate their effect on the measurement of the elbow flexion-extension angle during cricket bowling.

6.3.1 Materials and Methods

6.3.1.1 Study Population

Seven male bowlers, 5 medium-fast and 2 spin bowlers, from the MCC Young Cricketers and staff participated in this study. Their mean age was 22 years (range: 16 - 31 years) and mean body mass index was 23.6 (range: 18.6 – 27.8).
6.3.1.2 Equipment and Lab set-up

All testing took place at the MCC’s indoor cricket school as described in Section 5.2.1. An optical motion tracking system (Vicon, Oxford, UK) was used at an acquisition rate of 200 Hz and two video recorders were also used at an acquisition rate of 100 Hz to record the delivery swing and to allow synchronisation with the opto-reflective data (Section 5.2.1). Reflective markers were attached to boney landmarks on the head, thorax, humerus, forearm and hand of the bowlers (Section 5.2.2).

6.3.1.3 Experimental Protocol

The experimental protocol used was as defined in Section 5.2.3:

- A static trial of the marker setup
- Four static trials to digitise the LE position
- Four static trials to digitise the ME position
- A dynamic trial to estimate the GH position

After digitisation every bowler performed five repeated measures of a pure flexion-extension movement and a total of 16 deliveries, from which the first 4 were not included in the analysis to allow for the bowlers to reach match pace and from the remaining deliveries, 6 successful deliveries were collected for analysis.

6.3.1.4 Data analysis

Most of the data analysis was completed in the same way as described in Section 5.2.4. The digitisation of the LE and ME positions was the same as described in Section 5.2.4.1 and the centre of rotation for the glenohumeral joint GH was estimated as described in Section 5.2.5. However, in this study four different elbow positions as shown in Figure 6-4 were investigated: elbow fully flexed (Position I), elbow flexed to 90° (Position II), full extension (Position III), elbow flexed with humerus internally rotated (Position IV). For every dynamic trial and each elbow orientation the digitised epicondyles were
reconstructed with respect to two upper arm triads, that of Cluster P and that of Cluster D following the exact same methodology as described in Section 5.2.4.1.

Figure 6-4: Determination of the elbow epicondyles; I) When the elbow is fully flexed, II) flexed at 90°, III) fully extended and IV) when the elbow is flexed with the humerus internally rotated

For the anatomical based model the co-ordinate frames of the segments were defined according to the ISB recommendations (Wu et al., 2005) as follows:

**Humerus**

\( \mathbf{O}_{h1} \): The origin coincides with GH

\( \mathbf{Y}_{h1} \): The line connecting GH and EC, pointing towards GH.

\[
Y_{h1} = \frac{(GH - EC)}{\|GH - EC\|}
\]  

(6.1)
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\( X_{h_1} \): The line perpendicular to the plane formed by the Y axis and the line connecting the two elbow epicondyles (ME and LE).

\[
X_{h_1} = \frac{(GH - EC) \times (LE - ME)}{\|GH - EC\| \times \|LE - ME\|}
\]  \hspace{1cm} (6.2)

\( Z_{h_1} \): The line perpendicular to the \( Y_{h_1} \) and \( X_{h_1} \) axes, pointing to the right.

\[
Z_{h_1} = \frac{Y_{h_1} \times X_{h_1}}{\|Y_{h_1} \times X_{h_1}\|}
\]  \hspace{1cm} (6.3)

**Forearm**

\( O_{F_1} \): The origin coincides with US

\( Y_{F_1} \): The line connecting US and EC, pointing towards EC.

\[
Y_{F_1} = \frac{EC - US}{\|EC - US\|}
\]  \hspace{1cm} (6.4)

\( X_{F_1} \): The line perpendicular to the plane formed by the Y axis and the line connecting the two markers the ulnar and radial styloids (US and RS).

\[
X_{F_1} = \frac{(EC - US) \times (US - RS)}{\|EC - US\| \times \|US - RS\|}
\]  \hspace{1cm} (6.5)

\( Z_{F_1} \): The line perpendicular to the \( Y_F \) and \( X_F \) axes, pointing to the right.

\[
Z_F = \frac{Y_{F_1} \times X_{F_1}}{\|Y_{F_1} \times X_{F_1}\|}
\]  \hspace{1cm} (6.6)

Elbow rotations between the two segments were calculated using Euler angles with a Cardan sequence (\( z \cdot x' \cdot y'' \)).

### 6.3.1.5  Statistical analysis

The data for each bowler were normalised against the frames of interest (20 ms prior to upper arm horizontal to 20 ms after ball release) and subsequently averaged over the 6 best trials, and presented as the mean elbow angle and standard deviation (Figure 6-5).
Statistical differences between methods were assessed using a repeated measures analysis of variance (ANOVA) test with 1 dependent variable (elbow angle) and 2 within-subject factors (elbow orientation during calibration (Positions I to IV) and choice of cluster (Clusters P and D). All of the statistical tests were carried out using SPSS (version 18.0, Chicago, USA).

The within-day coefficient of multiple correlation (CMC) $r$ (Kadaba et al., 1989) was used to evaluate the repeatability of six trials of elbow extension angles generated in each subject for every digitisation position (Positions I to IV) and the new CMC formulation (Ferrari et al., 2010) was calculated to measure the similarity among the waveforms.
acquired by the 4 different methods of digitisation for one cluster. This new CMC formulation measures the similarity among waveforms acquired by the different protocols within each gait-cycle and is cleared from biological variability of the subject’s kinematics, variability in the propagation of the soft-tissue artifact, and variability in the measurement system performance (Ferrari et al., 2010). Similar to the within-day coefficient of multiple correlation (CMC) the new formulation also takes values that range from 0 (dissimilar waveforms) to 1 (similar waveforms).

\[
\text{new CMC} = \sqrt{1 - \frac{\sum_{g=1}^{G} \sum_{i=1}^{P} \sum_{j=1}^{N} (Y_{gij} - \bar{Y}_{gj})^2/GN_g (P - 1)}{\sum_{g=1}^{G} \sum_{i=1}^{P} \sum_{j=1}^{N} (Y_{gij} - \bar{Y}_g)^2 / GPN_g (1 - 1)}}
\]

(6.7)

Where \( P \) is the number of waveforms acquired by the different methods/protocols, \( G \) is the total number of kinematic cycles in this case trials, \( Y_{gij} \) is the flexion angle at frame \( j \) of each waveform as provided by protocol \( p \) at kinematic cycle \( g \), \( \bar{Y}_{gj} = \frac{1}{P} \sum_{i=1}^{P} Y_{gij} \) is the average elbow angle at \( j \% \) of the bowling action of the average waveform among the \( P \) waveforms of gait-cycle \( g \) and finally, \( \bar{Y}_g \) is the total average elbow angle for the delivery action among its \( P \) waveforms. To account for null or complex CMC values which would be interpreted as a complete dissimilarity of the waveforms, the squared value the CMC termed coefficient of multiple determinations CMD (\( r^2 \)) was evaluated instead. Non parametric Friedman’s ANOVA was used to investigate any differences between the CMD values. Statistical significance was set at an alpha level of 0.05.
6.3.2 Results

The repeated measures ANOVA determined that during controlled movements of pure flexion-extension the mean range of motion (Table 6-5) differed between the four elbow positions for all three rotations (Table 6-6). Post hoc tests using the Bonferroni correction revealed that Position I measures a greater range of extension than Positions II and III while, Position IV measures more adduction and pronation than Positions I, II and III. The interaction effect between the position and the cluster was significant for flexion and pronation suggesting that differences between the positions of calibration are observed for both clusters. When the digitised epicondyles are defined relative to Cluster D smaller range of motion in flexion is measured regardless of the orientation of the joint during the calibration.

Table 6-5: Mean ranges of motion and standard deviations for all thee rotations during a controlled movement of flexion – extension

<table>
<thead>
<tr>
<th>Cluster</th>
<th>Position</th>
<th>Elbow Rotations</th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Flexion - Extension</td>
<td>Adduction - Abduction</td>
<td>Pronation – Supination</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Mean (°)</td>
<td>Std. Deviation</td>
<td>Mean (°)</td>
<td>Std. Deviation</td>
<td>Mean (°)</td>
</tr>
<tr>
<td>D</td>
<td>PI</td>
<td>135.76</td>
<td>5.76</td>
<td>18.98</td>
<td>7.67</td>
<td>19.41</td>
</tr>
<tr>
<td></td>
<td>PII</td>
<td>130.71</td>
<td>5.89</td>
<td>17.68</td>
<td>6.05</td>
<td>20.35</td>
</tr>
<tr>
<td></td>
<td>PIII</td>
<td>128.90</td>
<td>5.77</td>
<td>18.45</td>
<td>6.08</td>
<td>20.77</td>
</tr>
<tr>
<td></td>
<td>PIV</td>
<td>136.80</td>
<td>7.63</td>
<td>38.48</td>
<td>10.14</td>
<td>64.91</td>
</tr>
<tr>
<td>P</td>
<td>PI</td>
<td>142.83</td>
<td>6.46</td>
<td>17.53</td>
<td>7.50</td>
<td>17.81</td>
</tr>
<tr>
<td></td>
<td>PII</td>
<td>136.80</td>
<td>7.63</td>
<td>15.10</td>
<td>5.95</td>
<td>16.93</td>
</tr>
<tr>
<td></td>
<td>PIII</td>
<td>138.68</td>
<td>6.94</td>
<td>19.37</td>
<td>9.25</td>
<td>27.03</td>
</tr>
<tr>
<td></td>
<td>PIV</td>
<td>138.68</td>
<td>10.55</td>
<td>38.06</td>
<td>7.36</td>
<td>77.93</td>
</tr>
</tbody>
</table>
Chapter 6

Sensitivity analysis

Table 6-6: The probability values (p-value) from the repeated-measures analysis of variance (ANOVA) tests for all three elbow rotations for both tasks

<table>
<thead>
<tr>
<th>Elbow Rotations</th>
<th>Flexion - Extension</th>
<th>Adduction - Abduction</th>
<th>Pronation - Supination</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pure Flexion</td>
<td>0.007**</td>
<td>0.667</td>
<td>0.104</td>
</tr>
<tr>
<td>Bowling</td>
<td>0.231</td>
<td>0.104</td>
<td>0.153</td>
</tr>
<tr>
<td>Pure Flexion</td>
<td>0.007**</td>
<td>0.013*</td>
<td>0.001**</td>
</tr>
<tr>
<td>Bowling</td>
<td>0.667</td>
<td>0.944</td>
<td>0.351</td>
</tr>
</tbody>
</table>

* The mean difference is significant at the 0.05 level.

** The mean difference is significant at the 0.01 level.

*** The mean difference is significant at the 0.001 level.

Average elbow extension angles during the ball delivery were 11° (SD: 7°, Range 0.3° - 25.5°). Elbow angles for each bowler were averaged over 6 trials, and presented in as the mean elbow angle and standard deviation. The calculated extension angles were 13.5° (± 6.3°) for Position II and 13.7° (± 6.6°) for Position III; these were greater (p<0.05) than those calculated for positions I and IV, in which the changes were 9.5° (± 8.4°) and 7.5° (± 5.5°) respectively. The within-day CMC was greater than 0.80 for all three rotations (Table 6-7) suggesting good consistency of the kinematic patterns obtained in the six trials for each subject. No significant differences were found between the two clusters for the same positions of digitisation. The mean new CMD values for Cluster D were 0.5 (± 0.2) for abduction and 0.37 (± 0.17) for pronation and were significantly greater from Cluster P 0.25 (± 0.27) and -0.19 (± 0.04) (χ² (1) =6.000, p=0.014 and χ² (1) =7.000, p=0.008) respectively.
Table 6-7: Elbow angles, within-day CMC (r) and new formulation CMD (r²) values of elbow kinematics during bowling for all four positions of digitisation and both clusters

<table>
<thead>
<tr>
<th>Cluster</th>
<th>Elbow Rotations (°)</th>
<th>Flexion - Extension</th>
<th>Adduction - Abduction</th>
<th>Pronation - Supination</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean (SD)</td>
<td>Within-day CMC</td>
<td>New CMD</td>
<td>Mean (SD)</td>
</tr>
<tr>
<td></td>
<td>D</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PI</td>
<td>10.70 (3.08)</td>
<td>0.92 (0.03)</td>
<td>0.55 (0.28)</td>
<td>9.11 (0.72)</td>
</tr>
<tr>
<td>PII</td>
<td>13.95 (2.26)</td>
<td>0.91 (0.03)</td>
<td>8.41 (0.39)</td>
<td>0.83 (0.16)</td>
</tr>
<tr>
<td>PIII</td>
<td>14.55 (2.17)</td>
<td>0.91 (0.03)</td>
<td>8.41 (0.56)</td>
<td>0.85 (0.13)</td>
</tr>
<tr>
<td>PIV</td>
<td>8.40 (2.45)</td>
<td>0.89 (0.04)</td>
<td>13.41 (1.71)</td>
<td>0.90 (0.09)</td>
</tr>
<tr>
<td></td>
<td>P</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PI</td>
<td>8.22 (3.42)</td>
<td>0.92 (0.04)</td>
<td>0.51 (0.26)</td>
<td>11.85 (2.23)</td>
</tr>
<tr>
<td>PII</td>
<td>13.08 (2.64)</td>
<td>0.92 (0.03)</td>
<td>11.05 (1.91)</td>
<td>0.89 (0.09)</td>
</tr>
<tr>
<td>PIII</td>
<td>12.75 (2.83)</td>
<td>0.91 (0.05)</td>
<td>10.90 (1.67)</td>
<td>0.90 (0.09)</td>
</tr>
<tr>
<td>PIV</td>
<td>6.48 (1.72)</td>
<td>0.90 (0.07)</td>
<td>15.72 (1.90)</td>
<td>0.92 (0.06)</td>
</tr>
</tbody>
</table>

* The mean difference is significant at the 0.05 level.

** The mean difference is significant at the 0.01 level.

### 6.3.3 Discussion

One of the primary aims of this study was to test the effect that the orientation of the elbow at the calibration process may have onto the average elbow angles during cricket bowling. Elbow rotations for two movements were calculated when digitising the humeral epicondyles at four different elbow positions (Positions I to IV). During a motion of pure flexion–extension, Position IV was found to measure considerably more abduction and pronation than the other three positions which suggests that the relative...
movement between the clusters and underlying bones significantly affects the digitised landmarks at this elbow orientation. During ball delivery the within-day CMC for flexion ranged from 0.90 - 0.94 suggesting that the curves obtained in the six trials for each subject were reproducible between trials for all positions of digitisation. However, when the humeral epicondyles were digitised with the elbow fully flexed (Position I) or with the elbow flexed and the humerus internally rotated (Position IV) average elbow angles were significantly lower. Looking closer at the results it can be seen that the statistical differences reported in the mean extension values are not related to the range of the recorded motion but to the formulation of the ICC regulations whereby elbow hyperextension is not included in the 15° tolerance threshold (Figure 6-6) with positions I and IV measuring more hyperextension than Positions II and III.

![Graph showing elbow flexion/extension rotations](image)

**Figure 6-6:** Elbow flexion/extension rotations of the same subject normalised against % of bowling action for Positions III and IV

Two upper arm clusters were also trialed in this study to replace the digitised anatomical landmarks during dynamic trials; Cluster P placed centrally onto the upper arm and Cluster D placed a few centimeters distal to the olecranon process. Statistical analysis for the two movements showed no significant differences in the mean angles reported for the same position of digitisation. However, the new CMD values for each rotation, determined to measure the similarity among the waveforms acquired by the 4 different...
positions of digitisation for each cluster, were found to be significantly greater for Cluster D than for Cluster P for abduction and pronation. This difference can be justified if one considers that during dynamic movements the most critical source of error is the rigid motion of the cluster with respect to the underlying bone (Cutti et al., 2005) as the digitised landmarks are affected by the same skin artefact (STA) that affects the cluster they are related to. The technical anatomical frame of Cluster P follows to some extent the movement of the biceps muscles that does not follow well the bones during extended humeral rotation (Cutti et al., 2005). Looking at the biomechanics of cricket bowling, studies have shown that the elbow joint rotates about 90° as a result of humeral rotation during the movement of shoulder circumduction produced by the bowler’s delivery action (Aginsky and Noakes, 2010; Portus et al., 2006). This concludes that digitising the humeral epicondyles with the joint flexed at 90° (Position II) or in full extension (Position III) and relating them to a distal upper arm cluster (Cluster D) improves the accuracy of the measurements especially when measuring changes in pronation and adduction as the propagation of STA mainly affects the joints characterised by a small range of motion (Cappello et al., 2005).
6.4 Study III: Single and Double anatomical landmark calibration

Soft tissue artefact and anatomical landmark misplacement have been recognized as the most critical sources of error in motion analysis (Cappello et al., 2005; Cutti et al., 2005; Della et al., 2005; Donati et al., 2007). The thick layer of skin covering the elbow makes it difficult to determine the joint’s position during motion and has led to the development of a number of measurement techniques to improve the accuracy of the data. In the previous study such a method was described where the positions of the elbow epicondyles were digitised in order to reduce the problem of soft-tissue deformation. However, the manual handling of the wand by an observer means that this method can be prone to errors due to wrong landmark placement during the calibration process (Della Croce et al., 2005).

Cappello et al. (1997) originally proposed a multiple anatomical landmark calibration method to compensate for soft tissue artefact in lower limb kinematics during cycling. The main idea behind this new technique derived from the fact that the soft tissues around the calibrated anatomical landmarks tend to move with respect to the underlying bone following a quasi-linear closed loop (Cappello et al., 1997). In this method landmarks are calibrated twice, once with the closest joint flexed and once when extended, at the maximum expected values during the specific task of interest. By interpolating between the two positions, the skin sliding can then be effectively compensated (Della Croce et al., 2005). This method was validated in two studies that showed that double anatomical landmark calibration improved the accuracy in knee kinematics during dynamic activities (Cappello et al., 2005; Stagni et al., 2005).

The aim of this study was to compare the effect that single and double anatomical landmark calibration have on elbow rotations during cricket bowling.
6.4.1 Materials and Methods

6.4.1.1 Study Population and Lab set-up

Seven male bowlers, 5 medium-fast and 2 spin bowlers, from the MCC Young Cricketers and staff participated in this study. Their mean age was 22 years (range: 16 - 31 years) and mean body mass index was 23.6 (range: 18.6 – 27.8). Testing took place at the MCC’s indoor cricket school. An optical motion tracking system (Vicon, Oxford, UK) at an acquisition rate of 200 Hz and two video recorders at an acquisition rate of 100 Hz were also used to record the delivery (Section 5.2.1) and track the reflective markers attached on the skin of the bowlers (Section 5.2.2).

6.4.1.2 Experimental Protocol and Data Analysis

The experimental protocol described in Section 5.2.4 was also employed in this study. Each bowler was asked to warm up prior to data capture and six bowling deliveries were collected for analysis.

Data analysis was carried out as described in Section 1.2.4.1. Single calibration of the LE and ME was performed with the joint flexed at 90° and with the joint in full extension as described in Section 6.3.1.4. Double calibration of each landmark was then performed through linear interpolation of the distal upper arm triad and digitised landmark positions between the two reference configurations (Figure 6-7), assuming time as the independent variable (Cappello et al., 1997).

Segmental co-ordinate systems were defined for the upper arm and forearm following the ISB recommendations on the definitions of joint co-ordinate systems (Section 6.3.1.4) and elbow rotations were calculated using Euler angles with a z-x’-y’’ sequence. The kinematic data for each bowler were subsequently normalised against the frames of interest (20 ms prior to upper arm horizontal to 20 ms after ball release), averaged over the 6 trials and presented as the mean elbow angle and standard deviation (Section 6.3.1.5).
6.4.1.3 Statistical analysis

The within-day coefficient of multiple correlation (CMC) $r$ (Kadaba et al., 1989) was used to evaluate the repeatability of six trials of elbow angles generated in each subject. To explore the differences between the two methods (single and double calibration) paired samples t-tests were conducted. Differences in the means were considered statistically significant for $p$ values <0.05. All of the statistical tests were carried out using SPSS (version 18.0, Chicago, USA).

6.4.2 Results

Both models produced similar kinematic waveforms for flexion (Figure 6-8) and high intra-tester repeatability during ball delivery with a within-day CMC greater than 0.80 for all three elbow rotations (Table 6-8). No significant differences were found in the calculated elbow angles between the models (Table 6-8).
### Table 6-8: Mean elbow angles, within-day CMC and probability values (p-value) of elbow kinematics during bowling for both methods of digitisation

<table>
<thead>
<tr>
<th>Method</th>
<th>Elbow Rotations(°)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean (°)</td>
</tr>
<tr>
<td>Flexion-Extension</td>
<td></td>
</tr>
<tr>
<td>Single Calibration</td>
<td>13.97</td>
</tr>
<tr>
<td>Double Calibration</td>
<td>13.53</td>
</tr>
<tr>
<td>Adduction-Abduction</td>
<td></td>
</tr>
<tr>
<td>Single Calibration</td>
<td>11.05</td>
</tr>
<tr>
<td>Double Calibration</td>
<td>8.84</td>
</tr>
<tr>
<td>Pronation-Supination</td>
<td></td>
</tr>
<tr>
<td>Single Calibration</td>
<td>41.01</td>
</tr>
<tr>
<td>Double Calibration</td>
<td>39.17</td>
</tr>
</tbody>
</table>

---

![Graph showing % Bowling Motion vs Flexion Angle](image-url)
6.4.3 Discussion

In this study the single and double calibration methods were used to calculate elbow rotations during cricket bowling. Results showed that the intra-tester repeatability of the methods was similar and no significant differences were found in the calculated elbow angles between the models.

In Section 6.3 it was shown that digitising the humeral epicondyles with the joint flexed at 90° or in full extension improves the accuracy of the measurements. In this study double calibration was performed between these two positions and compared to a single calibration with the elbow flexed at 90°. It is possible that by choosing different elbow orientations during the calibration process differences in the repeatability of the methods would be noted. Previous studies that have looked at knee kinematics during dynamic activities have shown that the double calibration improves the accuracy in knee kinematics during dynamic limiting the propagation of STA to knee kinematics due to anatomical landmark misplacement (Stagni et al., 2006) so double calibration could be used to improve the reliability of bowling kinematics to account for possible cases of inexperienced examiners or bowlers with elbow deformities, but this was not assessed here.
6.5 Study IV: The effect of placing markers onto the humeral epicondyles

The choice of marker placement to define the segments of the upper extremity has been variable, particularly with respect to the humerus and a reason of great controversy between researchers (Cappozzo et al., 1995; Cutti et al., 2005; Della Croce et al., 2005; Leardini et al., 2005; Rho et al., 1995). In the previous studies two different methods for digitising the humeral epicondyles were presented. In this study the effect of placing two markers directly onto the elbow epicondyles on measuring elbow kinematics during cricket bowling is presented.

6.5.1 Materials and Methods

Eight male bowlers, 6 medium - fast and 2 spin bowlers, from the MCC Young Cricketers and staff with a mean age of 21 years (range: 16 - 31 yrs) and mean body mass index of 22.3 (range: 19 - 28) participated in the study. Testing took place at the MCC’s indoor cricket school by means of an optical system (Vicon, Oxford, UK) at an acquisition rate of 200 Hz and of two video recorders at an acquisition rate of 100 Hz (Section 5.2.1). The experimental protocol described in Section 5.2.4 was also employed in this study for the construction of the functional based model (FuncM1) as described in Section 5.2.6.

Two reflective markers were attached to the humeral epicondyles with the bowler’s arm in an upright position. Each bowler was subsequently requested to bowl a total of 10 deliveries. Based on the marker positions co-ordinate frames of a second anatomical based model (AnatM2) were defined according to the ISB recommendations (Section 6.3.1.4) and elbow rotations were calculated using Euler angles with a z′ - x′ - y′′ sequence (Wu et al., 2005).

Data analysis was carried out as described in Section 1.2.4.1. The data for each bowler were normalised against the frames of interest (20 ms prior to upper arm horizontal to 20 ms after ball release) and subsequently averaged over 6 trials and presented as the mean elbow angle and standard deviation.
Chapter 6  Sensitivity analysis

The within-day coefficient of multiple correlation (CMC) $r$ (Kadaba et al., 1989) was used to evaluate the repeatability of the kinematic waveforms produced with each model. Differences in the mean elbow angles between the two models (FuncM1 and AnatM2) were assessed using paired samples t-tests. Non parametric Friedman’s analysis was used to investigate any differences between the CMC values. Statistical significance was set for $p$ values <0.05. Statistical tests were carried out using SPSS (version 18.0, Chicago, USA).

6.5.2 Results

Mean elbow angles during bowling, within-day CMC and probability values for both models are presented in Table 6-9.

Table 6-9: Mean elbow angles, within-day CMC and probability values ($p$-value) of elbow kinematics during bowling for both methods

<table>
<thead>
<tr>
<th>Elbow Rotations (*)</th>
<th>Method</th>
<th>Mean (°)</th>
<th>Std. Error</th>
<th>Model ($p$-value)</th>
<th>Within-day CMC (SD)</th>
<th>CMC $X^2$ ($p$-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flexion-Extension</td>
<td>FuncM1</td>
<td>13.84</td>
<td>1.39</td>
<td>0.828</td>
<td>0.90 (0.05)</td>
<td>0.50</td>
</tr>
<tr>
<td></td>
<td>AnatM2</td>
<td>14.18</td>
<td>2.39</td>
<td></td>
<td>0.80 (0.18)</td>
<td></td>
</tr>
<tr>
<td>Adduction-Abduction</td>
<td>FuncM1</td>
<td>12.28</td>
<td>1.76</td>
<td>0.010*</td>
<td>0.92 (0.04)</td>
<td>4.50</td>
</tr>
<tr>
<td></td>
<td>AnatM2</td>
<td>8.00</td>
<td>1.50</td>
<td></td>
<td>0.51 (0.39)</td>
<td></td>
</tr>
<tr>
<td>Pronation-Supination</td>
<td>FuncM1</td>
<td>31.78</td>
<td>4.43</td>
<td>0.699</td>
<td>0.95 (0.03)</td>
<td>8.00</td>
</tr>
<tr>
<td></td>
<td>AnatM2</td>
<td>30.14</td>
<td>4.43</td>
<td></td>
<td>0.63 (0.39)</td>
<td></td>
</tr>
</tbody>
</table>

* The mean difference is significant at the 0.05 level.

** The mean difference is significant at the 0.01 level.
There were no significant differences in the calculated flexion-extension and pronation-supination angles between the models the AnatM2 however, calculated a significantly smaller range of adduction than the FuncM1. The mean within-day CMC values for the FuncM1 model were 0.92 (± 0.04) for adduction and 0.95 (± 0.03) for pronation and were significantly greater from the AnatM2 0.51 (± 0.39) ($\chi^2 (1) = 4.50, p=0.034$) and 0.63 (± 0.39) ($\chi^2 (1) = 8.00, p=0.005$) respectively.

6.5.3 Discussion

In this study the effect of placing two markers directly onto the humeral epicondyles was investigated. Results showed that during the delivery stride both models calculated the same degree of elbow extension the within-day CMC values however indicate that the FuncM1 model shows higher intra-tester repeatability than the AnatM2 in the cases of pronation and adduction. As kinematic cross talk; a phenomenon where one joint rotation is being interpreted as another, has been recognised as one of the major sources of error in joint kinematics (Cutti et al., 2008; Piazza and Cavanagh, 2000) it is proposed that a functional approach will produce more reproducible between trials elbow angles during bowling.
6.6 Study V: The effect of one frame

As discussed in Chapter 3 the objective of the ‘no ball’ law is to control the bowler’s arm so that there is no straightening or bending of the elbow during ball delivery. For bowlers that have been called for throwing the degree of flexion-extension of the elbow is measured between two points: (1) the point that the upper arm reaches the level of shoulder and (2) ball release as shown in Figure 6-9. From this definition it becomes evident that identifying these two points can be of paramount importance when assessing the legality of a single bowl.

![Figure 6-9: Identifying (a) upper arm horizontal and (b) ball release for one bowler](image)

In this study two video recorders at an acquisition rate of 100 Hz synchronised with the optical motion tracking system were used to identify these two points. Each camera was set at exactly 90° to the plane of elbow joint movement that was of interest. Even though, it is relatively easy to determine the frame at which the upper arm is horizontal and parallel to the ground, identifying ball release has been proved to be challenging. The point of ball release is generally defined as the first frame that the ball and hand are not in contact but because of differences in the ball’s speed and bowling style it is not always easy to identify. In this study the effect that one frame may have on the average elbow extension angles was investigated.
6.7 Materials and Methods

The same study population, equipment, lab set up, data capture and analysis as described in Section 5.2 for the construction of a functional based kinematic model (FuncM1) were employed. The degree of elbow extension was measured from the point that the upper arm reached the level of shoulder until: (1) one frame before ball release, (2) ball release and (3) one frame after the ball has left the bowler’s hand. Based on the frequency at which the optical motion system operated each frame was 0.005 sec. Statistical differences between methods were assessed using a repeated measures (ANOVA) test with the significance set at an alpha level of 0.05.

6.7.1 Results

Average elbow angles during the ball are presented in Table 6-10. The repeated measures ANOVA showed that the mean range of elbow extension differed between the three methods. When compared to the degree of elbow extension predicted with the FuncM1 from the point the upper arm is horizontal to ball release post hoc analysis revealed that excluding 1 frame measures a smaller range of extension of 1.1° (±1.0°) while, including 1 frame after ball release measures a greater range of extension of 1.7° (±2.2°) on average.

Table 6-10: Mean elbow angles, 95% confidence intervals and probability values (p-value) for flexion-extension during cricket bowling

<table>
<thead>
<tr>
<th>Method</th>
<th>Mean (°)</th>
<th>Std. Error</th>
<th>95% Confidence Interval</th>
<th>Model (p-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lower Bound</td>
<td>Upper Bound</td>
</tr>
<tr>
<td>Flexion-Extension</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 frame before BR</td>
<td>12.73</td>
<td>1.44</td>
<td>9.41</td>
<td>16.06</td>
</tr>
<tr>
<td>FuncM1</td>
<td>13.84</td>
<td>1.23</td>
<td>11.01</td>
<td>16.68</td>
</tr>
<tr>
<td>1 frame after BR</td>
<td>15.50</td>
<td>1.28</td>
<td>12.55</td>
<td>18.45</td>
</tr>
</tbody>
</table>
6.7.2 Discussion

In this study the effect that one frame may have onto the average elbow flexion-extension angles was investigated and results showed that significant differences in the mean elbow extension angles were reported when including or excluding one frame from the analysis.

In order to detect the frames of interest (when the upper arm reaches the level of shoulder and ball release) two video cameras running at 100 Hz were used. The first video recorder was positioned square to the bowler when the bowler’s arm was horizontal and the second video recorder was placed in front of the bowler at the instant of ball release (Aginsky and Noakes, 2010). This allowed the examiner to have an optimal view of the two most critical points of the delivery swing. Furthermore, the video cameras were synchronised with the optical motion tracking system which run at 200 Hz which effectively meant that every frame of video recording corresponded to two frames of motion tracking.

Differences in the degree of flexion-extension can be explained looking closer to elbow kinematics during bowling as described in Section 5.3. Flexion-extension graphs revealed that bowlers can either flex their elbow through release or have a small elbow extension angle prior to ball release. These changes in the last frames of the delivery show that the point of release is crucial when determining bowl legality and may underestimate or overestimate changes in the recorded motion.
6.8 Summary

Different models have been used to measure the bowling motion and assess the legality of a delivery. This chapter investigated the sensitivity of the proposed functional based kinematic model (FuncM1) to anatomical based axes (AnatM), elbow orientation and choice of upper arm triad during single anatomical landmark calibration, double calibration and markers onto the humeral epicondyles (AnatM2) as well as the effect that one frame has on quantifying elbow kinematics during cricket bowling. In all the cases sensitivity was assessed quantitatively through indirect measurement approaches by calculating the average elbow angles and correlation coefficients.

The current anatomical based model AnatM was found to be sensitive to the position of the elbow joint during the digitisation process and the choice of upper arm cluster. Results showed that digitising the humeral epicondyles with the joint flexed at 90° or in full extension and relating them to a distal upper arm cluster, close to the joint improves the accuracy of the measurements. The effect of double calibration of the humeral epicondyles on joint angles was also investigated and results showed that the intra-tester repeatability of the methods was similar to that of the single calibration with no significant differences in the calculated elbow angles between the models reported. Anatomical landmark misplacement however, has been recognised as the most critical source of error in motion analysis. As previous studies have shown that this method is highly effective in limiting the propagation of STA to joint kinematics due to anatomical landmark misplacement (Stagni et al., 2006) double calibration should be used when employing anatomical based models, to account for possible cases of inexperienced examiners or bowlers with elbow deformities.

Within this study the effect of placing two markers directly onto the humeral epicondyles (AnatM2) was also investigated. Even though both models calculated the same degree of elbow extension, the FuncM1 model showed higher intra-tester repeatability than the AnatM2 in the cases of pronation and adduction producing more reproducible between trials elbow angles during bowling.
To conclude, both the functional and anatomical based kinematic models showed similar intra-tester repeatability and could therefore be used in assessing bowl legality. However, a functional based approach that is not dependant on the accurate identification of anatomical landmarks and is therefore less affected by errors due to landmark misplacement is recommended to assess bowl legality. To this end the importance of correct identification of the point of ball release should also be highlighted as it can significantly change the results.
Chapter 7  The carrying angle in cricket bowlers

In the previous chapters functional and anatomical based kinematic models to measure elbow kinematics during bowling were presented. As the presence of the carrying angle has been directly associated with the illusion of a throw in cricket bowling the aim of this chapter is to measure and calculate the carrying angle of cricket bowlers and evaluate its variability during flexion of the elbow joint using in-vivo kinematic data.
Chapter 7  The carrying angle in cricket bowlers

7.1 Introduction

When the human arm is completely extended with the palm facing up then the upper arm and forearm are not perfectly aligned. This deviation from a straight line that occurs in the direction of the thumb is referred to as the ‘carrying angle’ (Steel and Tomlison, 1958) and is most apparent with the elbow completely extended and the forearm supinated (Zampagni et al., 2008b). In clinical practice, the carrying angle is assessed through radiographs or goniometers and substantial differences have been reported (Van Roy et al., 2005). This variability in the reported values of the carrying angle is most apparent in differences in the design of elbow replacement implants where clinical results have been disappointing so far.

The role of the carrying angle is quite significant in throwing athletes since they are reported to have increased carrying angles in the elbow joint of their bowling arm adapting in that way to repetitive stress (Zampagni et al., 2008b). This adaptation may lead to elbow instability and pain in throwing activities and predispose to risk of elbow dislocation (Chang et al., 2008). Furthermore the presence of a large carrying angle in cricket bowling has been widely associated with the optical illusion of the throw (Aginsky and Noakes, 2010) as discussed in Section 3.4.2.

The aim of this study is to calculate the carrying angle of cricket bowlers and evaluate its variability during active dynamic movements of the elbow.

7.2 Functional anatomical context of the carrying angle

As described in Chapter 2 (Section 2.2.3.1) the valgus angulation in the human arm, termed as the ‘carrying angle’, is present because the trochlea extends farther distally than does the capitulum (Zampagni et al., 2008b). Anatomically, the carrying angle is divided by the transverse axis of the elbow joint in two smaller angles; the superior angle, referred to as ‘brachial angle’ is limited by the longitudinal brachial axis and the transverse axis of the elbow while, the inferior one termed the ‘ulnar angle’ is limited by the longitudinal antebrachial axis and the transverse axis of the elbow (Paraskevas et al., 2004). Usually, these two angles are approximately equal with the brachial angle ranging
between 70°-89° and the ulnar angle 77°-95°. During elbow flexion however, if the brachial angle is greater, then the forearm tends to deflect medially and if the ulnar angle is less then the forearm tends to deflect laterally (Paraskevas et al., 2004).

Figure 7-1: (a) Determination of the brachial (a) and ulnar (b) angle of the carrying angle of the elbow joint in posteroanterior radiographs (Paraskevas et al., 2004) and (b) posterior aspect of the elbow

7.2.1 Assessment of the carrying angle

The carrying angle is typically between 5° and 15° for men and 10° to 25° for women (Anglin and Wyss, 2000) and is usually estimated in full elbow extension by radiographs (London, 1981; Steel and Tomlison, 1958) or a protractor goniometer (Amis and Miller, 1982; Paraskevas et al., 2004). However, a number of different measuring techniques such as optical and electromagnetic motion tracking systems (Gordon et al., 2006; Zampagni et al., 2008a) and imaging techniques (Park and Kim, 2009) have also been used to assess the carrying angle (Figure 7-2).
Figure 7-2: Estimation of the carrying angle using: (a) a manual goniometer (Chang et al., 2008), (b) a clinical goniometer with an electromagnetic tracking system (Van Roy et al., 2005), (c) a radiograph (Alsubael and Hegazy 2010) and (d) 3-D reconstruction of MR images (Goto et al., 2004)

The carrying angle of the elbow is usually estimated from the angle formed by the longitudinal axes of the humerus and the forearm with the elbow fully extended and the
forearm fully supinated (Van Roy et al., 2005). However, substantial differences in the mean values of the carrying angle have been reported and there has also been much debate among researchers as to the effect of gender and age in the carrying angle. Potter (1895) was one of the first to measure the carrying angle in humans. He used a simple hinged board and kept the forearm fully extended and supinated, an arrangement also employed by Mall (1905), Nagel (1907) and Fick (1911). They all suggested differences in the mean angle between genders, with women having a significantly higher carrying angle than men that ranged from 1.8° to 5.8°.

Steel and Tomilson (1958) were the first to define the carrying angle in terms of bony landmarks. They x-rayed the human arm with the forearm fully extended and supinated and calculated the trochlear angle as the acute angle between the tangent line to the medial aspect of the humerus passing through the medial epicondyle and the line connecting the medial epicondyle and the most medial aspect of the ulna as shown in Figure 7-3c. They reported a mean angle of 19° and found no statistical differences between genders. Amis and Miller (1982) defined the carrying angle as the angle formed between the longitudinal axes of the upper arm and forearm (Figure 7-3a) whilst, London et al., (1981) calculated the carrying angle formed by the longitudinal axis of the ulna and a line perpendicular to the axis of elbow flexion (Figure 7-3e). Shiba et al., (1988) reported the angle formed by the longitudinal axis of the ulna and a perpendicular line to the line passing through the elbow epicondyles (Figure 7-3d).

Paraskevas et al., (2003) evaluated the carrying angle as the supplementary angle between the longitudinal axis of the humerus and that of the forearm in 600 subjects and reported a mean angle of 12.9° (±5.9°). Furthermore, the carrying angle was found to change with skeletal growth and maturity. This finding has also been documented in studies of the carrying angle in children and young adults whereby, the carrying angles of both dominant and non-dominant arms were found to be higher in the volunteers older than 14 years than of those younger than 14 years; with females ranking higher than males in both groups (Golden et al., 2007; Yilmaz et al., 2005).
Goto et al., (2004) studied the humeroulnar and humeroradial joints non-invasively in 3 healthy volunteers using magnetic resonance imaging and image reconstruction via bone registration algorithms. They reported a mean carrying angle of 18.64° evaluated as the abduction - adduction angle of the long axis of the ulna with respect to the instantaneous axes of rotation of the humeroulnar joint. Finally, Park and Kim, (2009) evaluated the carrying angle as the angle between two longitudinal axes formed from the centrelines of the humerus and the ulna using three-dimensional CT data (Figure 7-3f). They subsequently reported a mean value of 20.7° (±3.6°) and compared these values to those measured by simple radiograph which were 16.3° (±3.2°) without statistical difference.

A review of the published literature reveals that there are substantial differences in the measuring techniques and definition of the carrying angle among researchers. These methodological differences, different concepts used to determine the carrying angle but
also variations in the populations studied may justify to some extent the produced conflicting data. In daily clinical practice the carrying angle is mainly assessed via a simple goniometer and defined as the deviation of the perceived long axis of the fully supinated forearm from the sagittal plane containing the long humeral axis (Figure 7-4a). Even though, this is the simplest way to define the carrying angle this definition can vary considerably between investigators and be prone to errors due to the fact that the centre line of the upper arm follows to some extent the biceps muscle (Van Roy et al., 2005). On the other hand when the carrying angle is assessed in radiological images or via motion tracking systems its definition is associated with bony reference points and in most cases is determined between the long axes of the humerus and the ulna (Figure 7-4b) instead of the long axes of the upper arm and forearm. This difference in the definition of the long axis of the forearm alone can cause a shift in the estimated carrying angle (Figure 7-4c).

Figure 7-4: The carrying angle defined as the acute angle between (a) the long axes of the upper arm and forearm and (b) the long axes of the upper arm and ulna and (c) the difference $\phi$ between the two definitions (a) and (b)

However, as shown in Figure 7-3 even when bony landmarks are employed several different procedures have been proposed to define these reference lines of interest. The tangent line to the medial aspect of the humerus passing through the medial epicondyle or the line joining two mid-points at the distal and proximal end of the humerus do not
necessarily coincide with each other or with the direction of the long axis of the upper arm. In the same way, the humeral axis defined as the line perpendicular to the axis of elbow flexion - with the latter being defined either as the line joining the two elbow epicondyles or as the line running through the centre of the trochlea and the capitulum humeri - will not be parallel to the longitudinal axis of the upper arm as perceived in clinical practice. With respect to the forearm it can be seen that the line connecting the medial epicondyle and the most medial aspect of the ulna or the line joining the mid-point at the distal end of the humerus to the ulnar styloid process or to the mid-point between the ulnar and radial styloid processes may not be parallel to the longitudinal axis of the forearm.

7.2.2 The carrying angle during elbow flexion

Amis et al. (1977) was the first to notice a sinusoidal decrease of the carrying angle as a function of increasing passive elbow flexion whilst Chao and Morrey (1978), Deland (1988), Van Roy (1993) and Zampagni (2008) all reported a linear decrease (Table 7-1). This change in the carrying angle during flexion can be anatomically explained by the spiral form of the humeral trochlear groove (Amis, 1977). During elbow flexion the inferior and anterior parts of the humeral trochlea articulate with the ulnar trochlear notch while during elbow extension the dorsal aspect of humeral trochlear groove partly articulates the ulnar notch (Van Roy et al., 2005). In other words, during extension of the forearm the carrying angle is normal, while during flexion the forearm is medial to the arm (Paraskevas et al., 2004).

Some studies also include an estimation of the carrying angle using kinematic data. Van Roy et al., (2005), showed that during flexion–extension movements a continuous decrease of the carrying angle exists as a function of elbow flexion with women showing higher values than men though this difference decreased as the elbow flexion increased. Shiba et al. (1988) found constantly changing elbow centers during flexion-extension movements. They suggested that a correlation exists between the transverse axis of elbow flexion and the line that connects the lateral and medial elbow epicondyles.
In a cadaveric study, Morrey and Chao (1978) reported that the carrying angle is greatest in valgus at full extension, diminishing during flexion, and becoming varus at full flexion. They also showed that slight medial and lateral rotation components exist at the beginning and in the final course of flexion – extension movement which suggests that elbow anthropometrics and three-dimensional joint kinematics are directly related. The significance of the carrying angle in the joint’s kinematics is such that even though, it is not an independent degree of freedom, in the absence of pathologies it must be considered as one in fact modeled as the second rotation, following elbow flexion but preced ing pronation (Anglin and Wyss, 2000).
<table>
<thead>
<tr>
<th>Dominant arm CA (Mean ±SD)</th>
<th>Measurement Technique</th>
<th>During flexion</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Males</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Potter (1895)</td>
<td>6.8°</td>
<td>Goniometer</td>
</tr>
<tr>
<td>Steel and Tomilson (1958)</td>
<td>18.4° (±3.4°)</td>
<td>Radiograph</td>
</tr>
<tr>
<td>Baughman et al., (1974)</td>
<td>11° (2°–21°)</td>
<td>Goniometer</td>
</tr>
<tr>
<td>Amis, (1977)</td>
<td>10.5° (7.0°-14.7°)</td>
<td>Radiograph</td>
</tr>
<tr>
<td>Chao and Morrey (1978)</td>
<td>10°</td>
<td>Radiograph</td>
</tr>
<tr>
<td>Shiba et al., (1988)</td>
<td>5° (2.5°-7.5°)</td>
<td>surface analytic methods</td>
</tr>
<tr>
<td>Van Roy et al., (2005)</td>
<td>11.6° (±3.2°)</td>
<td>Electromagnetic tracking system</td>
</tr>
<tr>
<td>Yilmaz et al., (2005)</td>
<td>10.47° (±3.75°)</td>
<td>Goniometer</td>
</tr>
<tr>
<td>Zampagni et al., (2008)</td>
<td>12.39° (±3.64°)</td>
<td>FaroArm electrogoniometer</td>
</tr>
<tr>
<td>Park and Kim, (2009)</td>
<td>19.8° (13.5°-26.4°)</td>
<td>3D CT</td>
</tr>
<tr>
<td>Females</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Potter (1895)</td>
<td>12.7°</td>
<td></td>
</tr>
<tr>
<td>Steel and Tomilson (1958)</td>
<td>19.3° (±4.7°)</td>
<td></td>
</tr>
<tr>
<td>Amis, (1977)</td>
<td>10°</td>
<td></td>
</tr>
<tr>
<td>Chao and Morrey (1978)</td>
<td>10°</td>
<td></td>
</tr>
<tr>
<td>Shiba et al., (1988)</td>
<td>13.5° (9.5°-17.5°)</td>
<td></td>
</tr>
<tr>
<td>Paraskeuas et al., (2003)</td>
<td>15.07° (±4.95°)</td>
<td></td>
</tr>
<tr>
<td>Van Roy et al., (2005)</td>
<td>16.7° (±2.7°)</td>
<td></td>
</tr>
<tr>
<td>Yilmaz et al., (2005)</td>
<td>12.03° (±3.21°)</td>
<td></td>
</tr>
<tr>
<td>Zampagni et al., (2008)</td>
<td>12.9° (±3.95°)</td>
<td></td>
</tr>
<tr>
<td>Park and Kim, (2009)</td>
<td>22.1° (15.9°-27.0°)</td>
<td></td>
</tr>
</tbody>
</table>
Chapter 7 The carrying angle in cricket bowlers

7.2.3 The carrying angle in cricket bowling

As discussed in Section 7.2.1 the value of the carrying angle reported in the literature is greatly variable, ranging between 7° and 19°. Even though several studies have suggested that throwing athletes tend to have increased valgus angulation in their bowling arm adapting in that way to repetitive stress, very few studies have measured the carrying angle in overhead throwers and mean values of 9° (±4°) (Werner et al., 2007) and 8.6 (±2.9°) (Kiyoshige et al., 2000) have been reported in collegiate baseball pitchers. This fits into the normal range reported elsewhere. Clinical studies have shown that repetitive overhead throwing imparts high valgus loads to the throwing arm (Werner et al., 2007) often leading to progressive structural changes such as an increased carrying angle (Chang et al., 2008; Werner et al., 2007). Increasing carrying angle (above 15°) has been widely associated with elbow instability and reduced function of elbow flexion (Chang et al., 2008) leading to pain and dislocation during throwing (Cain et al., 2003).

Cricket bowlers have also been reported to have large carrying angles (Aginsky and Noakes, 2010; Portus et al., 2004) while the presence of the large carrying angle has been widely associated with the visual illusion of a “throw” (Aginsky and Noakes, 2010). As discussed in Section 3.4 assessing bowling legality involves measuring elbow rotations during throwing so this study aims at analysing and presenting anthropometric data for each bowler in an attempt to distinguish between those who passively extend their joint due to anatomical factors and those who actively extend aiming at increasing the ball’s speed through release.
7.3 Materials and Methods

7.3.1 Data acquisition

7.3.1.1 Study Population

A total of 6 male right handed bowlers (5 medium-fast and 1 spinner) from the MCC Young Cricketers and staff participated in this study. Their mean age was 21 years (range: 16-31 years) and mean body mass index was 22.3 (range: 19 - 27.8).

7.3.1.2 Equipment and Lab set-up

Testing took place at the MCC’s indoor cricket school as described Section 5.2.1. An optical motion tracking system (Vicon, Oxford, UK) was used at an acquisition rate of 200 Hz (Section 5.2.1). Reflective markers were attached to boney landmarks on the head, thorax, humerus, forearm and hand of the bowlers (Section 5.2.2).

7.3.1.3 Experimental Protocol

The experimental protocol used was the same for all subjects as described in Section 5.2.3 and is as follows:

- A static trial of the marker setup
- One static trial to digitise the LE position
- One static trial to digitise the ME position
- One static trial with the subject holding their arm in full extension (Figure 7-6a)
- One static trial with the subject holding their arm at 90° of flexion (Figure 7-6b)
- A dynamic trial to estimate the GH position

After digitisation every bowler performed three repeated measures of a pure flexion-extension movement and three repeated measures of a pure pronation-supination movement.
7.3.2 Data analysis

Data analysis was completed as described in Section 5.2.4. The digitisation of the LE and ME positions was carried out as described in Section 5.2.4.1 and the centre of rotation for the glenohumeral joint GH was estimated as described in Section 5.2.4.

For the anatomical based model (AnatM) the co-ordinate frames of the segments were defined according to the ISB recommendations (Wu et al., 2002). For the construction of a functional kinematic model (FuncM1) elbow flexion-extension and pronation-supination helical axes were defined based on the relative movement of the forearm with respect to the upper arm as described in Section 5.2.6. The carrying angle was evaluated in two static trials (Figure 7-6) with the subject (a) holding their arm in full extension and (b) 90° of flexion, three dynamic trials where each subject was instructed to (c) actively flex and extend the elbow while keeping a constant supination and (d)
three repeated measures of a pure pronation-supination movement with the joint flexed at approximately 90°.

Elbow rotations were calculated using Euler sequence of $z \times' y''$ where $z$ is flexion-extension, $x$ is abduction-adduction $y$ is pronation-supination of the elbow. Differences in the elbow abduction angles between the two models (FuncM1 and AnatM) and between flexion angles (0° and 90°) for each model during the static trials were assessed using paired samples $t$-tests. To explore statistical differences in the reported abduction angles during dynamic movements a repeated measures analysis of variance (ANOVA) test was conducted for each kinematic model. Regression analysis was used to identify the relationship between the changes in abduction with flexion angle. Intra-subject repeatability for both kinematic models was assessed by comparing the standard deviation of the mean abduction angles at every 15° intervals of flexion for the functional range of motion (30° to 130° of elbow flexion). Similarly, the standard deviations of the mean abduction angle for each subject were compared at every 20° intervals of pronation for the functional range of motion (40° to 140° pronation). Statistical analysis was done using a repeated measures ANOVA test. Statistical significance was set for $p$ values <0.05. Statistical tests were carried out using SPSS (version 18.0, Chicago, USA).
7.4 Results

7.4.1 Measurements of carrying angles in extension

In full elbow extension the mean values of the carrying angle were 9.6° (±4.3°, range: 3.3° to 14.8°) for the anatomical based kinematic model (AnatM) and 6.9° (±3.9°, range: 2.5° to 12.5°) for the motion derived model (FuncM1) (Table 7-2). With the elbow joint flexed at approximately 90° the average carrying angle significantly decreased to -7.9° (±7.7°, range: 1.2°– -20°; \( p = 0.002 \); Table 7-2) for the AnatM model while remaining almost constant with the functional based FuncM1 (7.4°±4.0°; \( p = 0.720 \)).

Table 7-2: Carrying angle values (in degrees) of the elbow with the joint in full extension and 90° of flexion

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>AnatM</th>
<th>FuncM1</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0°</td>
<td>90°</td>
<td>0°</td>
</tr>
<tr>
<td>1 Med</td>
<td>10.8</td>
<td>-5.1</td>
<td>9.4</td>
</tr>
<tr>
<td>2 Fast</td>
<td>11.8</td>
<td>1.2</td>
<td>8.7</td>
</tr>
<tr>
<td>3 Spin</td>
<td>3.3</td>
<td>-8.0</td>
<td>5.1</td>
</tr>
<tr>
<td>4 Fast</td>
<td>11.4</td>
<td>-2.5</td>
<td>3.0</td>
</tr>
<tr>
<td>5 Med</td>
<td>5.6</td>
<td>-20.0</td>
<td>2.5</td>
</tr>
<tr>
<td>6 Med</td>
<td>14.7</td>
<td>-11.6</td>
<td>12.4</td>
</tr>
<tr>
<td>Mean (±SD)</td>
<td>9.6° (4.3°)</td>
<td>-7.7° (7.5°)</td>
<td>6.9° (3.9°)</td>
</tr>
</tbody>
</table>

**. The mean difference is significant at the 0.01 level.
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7.4.2  Measurements of carrying angles during flexion-extension movements

During dynamic trials of active flexion-extension movement changes in the carrying angle were evaluated for the functional range of motion from 30° to 130° of flexion and abduction values were statistically compared at every 15° of elbow flexion. Results showed that when employing an anatomical based model (AnatM) the carrying angle significantly decreases with flexion (Table 7-3) from 5.2° (± 5.4°) abduction in 30° of flexion to 8.9° (± 6.5°) adduction in 130° of flexion (p<0.001).

Table 7-3: Carrying angle values at different degrees of active elbow flexion (AnatM)

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>30° (p-value)</th>
<th>45° (p-value)</th>
<th>60° (p-value)</th>
<th>75° (p-value)</th>
<th>90° (p-value)</th>
<th>105° (p-value)</th>
<th>120° (p-value)</th>
<th>130° (p-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Med</td>
<td>5.3 (0.7)</td>
<td>3.4 (0.8)</td>
<td>1.8 (0.8)</td>
<td>0.5 (0.8)</td>
<td>-1.2 (0.5)</td>
<td>-1.8 (0.5)</td>
<td>-2.5 (0.6)</td>
<td>-3.1 (0.6)</td>
</tr>
<tr>
<td>2</td>
<td>Fast</td>
<td>6.6 (1.2)</td>
<td>3.1 (0.2)</td>
<td>-0.2 (0.3)</td>
<td>-3.0 (0.5)</td>
<td>-6.1 (0.8)</td>
<td>-8.3 (0.8)</td>
<td>-8.5 (0.7)</td>
<td>-7.7 (0.2)</td>
</tr>
<tr>
<td>3</td>
<td>Spin</td>
<td>4.4 (0.6)</td>
<td>4.5 (1.0)</td>
<td>3.2 (1.0)</td>
<td>2.5 (1.2)</td>
<td>1.1 (1.5)</td>
<td>-0.3 (1.3)</td>
<td>-1.4 (0.6)</td>
<td>-2.1 (0.5)</td>
</tr>
<tr>
<td>4</td>
<td>Fast</td>
<td>5.1 (0.5)</td>
<td>4.5 (0.8)</td>
<td>2.6 (0.5)</td>
<td>0.0 (0.5)</td>
<td>-3.4 (0.3)</td>
<td>-5.5 (0.3)</td>
<td>-7.4 (0.5)</td>
<td>-8.6 (1.0)</td>
</tr>
<tr>
<td>5</td>
<td>Med</td>
<td>-3.7 (3.2)</td>
<td>-8.2 (3.4)</td>
<td>-13.9 (3.6)</td>
<td>-19.0 (2.5)</td>
<td>-22.5 (1.8)</td>
<td>-23.3 (1.0)</td>
<td>-21.5 (0.7)</td>
<td>-19.7 (0.5)</td>
</tr>
<tr>
<td>6</td>
<td>Med</td>
<td>13.2 (2.5)</td>
<td>11.5 (1.7)</td>
<td>7.2 (0.9)</td>
<td>2.5 (0.7)</td>
<td>-1.9 (1.3)</td>
<td>-6.3 (1.1)</td>
<td>-11.4 (0.3)</td>
<td>-12.4 (2.6)</td>
</tr>
</tbody>
</table>

Mean (±SD) 5.2 (5.4) 3.2 (6.3) 0.0 (7.2) -2.8 (8.2) -5.8 (8.5) -7.6 (8.2) -7.6 (7.3) -8.8 (6.5) -8.9 (6.5)

***. The mean difference is significant at the 0.001 level.

Post-hoc analysis as shown in Table 7-4 showed that statistical differences between the abduction angles occur at flexion angles less than 100°.
### Table 7-4: Post hoc test comparing the abduction angles at specific flexion angles for the AnatM.

<table>
<thead>
<tr>
<th>Flexion Angle (°)</th>
<th>Mean Difference</th>
<th>Std. Error</th>
<th>(p-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>45</td>
<td>2.01</td>
<td>0.70</td>
<td>0.036*</td>
</tr>
<tr>
<td>60</td>
<td>5.03</td>
<td>1.34</td>
<td>0.013*</td>
</tr>
<tr>
<td>75</td>
<td>7.90</td>
<td>1.99</td>
<td>0.011*</td>
</tr>
<tr>
<td>90</td>
<td>10.81</td>
<td>2.35</td>
<td>0.006**</td>
</tr>
<tr>
<td>105</td>
<td>12.73</td>
<td>2.57</td>
<td>0.004**</td>
</tr>
<tr>
<td>120</td>
<td>13.93</td>
<td>2.80</td>
<td>0.004**</td>
</tr>
<tr>
<td>130</td>
<td>14.08</td>
<td>2.74</td>
<td>0.004**</td>
</tr>
<tr>
<td>45</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>60</td>
<td>3.01</td>
<td>0.71</td>
<td>0.008**</td>
</tr>
<tr>
<td>75</td>
<td>5.88</td>
<td>1.41</td>
<td>0.009**</td>
</tr>
<tr>
<td>90</td>
<td>8.80</td>
<td>1.81</td>
<td>0.005**</td>
</tr>
<tr>
<td>105</td>
<td>10.71</td>
<td>2.12</td>
<td>0.004**</td>
</tr>
<tr>
<td>120</td>
<td>11.91</td>
<td>2.55</td>
<td>0.005**</td>
</tr>
<tr>
<td>130</td>
<td>12.06</td>
<td>2.60</td>
<td>0.006**</td>
</tr>
<tr>
<td>60</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>75</td>
<td>2.86</td>
<td>0.72</td>
<td>0.011*</td>
</tr>
<tr>
<td>90</td>
<td>5.78</td>
<td>1.15</td>
<td>0.004**</td>
</tr>
<tr>
<td>105</td>
<td>7.70</td>
<td>1.54</td>
<td>0.004**</td>
</tr>
<tr>
<td>120</td>
<td>8.90</td>
<td>2.13</td>
<td>0.009**</td>
</tr>
<tr>
<td>130</td>
<td>9.05</td>
<td>2.31</td>
<td>0.011*</td>
</tr>
<tr>
<td>75</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>90</td>
<td>2.91</td>
<td>0.46</td>
<td>0.002**</td>
</tr>
<tr>
<td>105</td>
<td>4.83</td>
<td>0.95</td>
<td>0.004**</td>
</tr>
<tr>
<td>120</td>
<td>6.03</td>
<td>1.73</td>
<td>0.018*</td>
</tr>
<tr>
<td>130</td>
<td>6.18</td>
<td>2.02</td>
<td>0.028*</td>
</tr>
<tr>
<td>90</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>105</td>
<td>1.91</td>
<td>0.56</td>
<td>0.019*</td>
</tr>
<tr>
<td>120</td>
<td>3.11</td>
<td>1.44</td>
<td>0.084</td>
</tr>
<tr>
<td>130</td>
<td>3.26</td>
<td>1.80</td>
<td>0.130</td>
</tr>
<tr>
<td>105</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>120</td>
<td>1.20</td>
<td>0.93</td>
<td>0.253</td>
</tr>
<tr>
<td>130</td>
<td>1.35</td>
<td>1.34</td>
<td>0.361</td>
</tr>
<tr>
<td>120</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>130</td>
<td>0.15</td>
<td>0.48</td>
<td>0.769</td>
</tr>
</tbody>
</table>
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Regression analysis confirmed that a linear variation does exist between the carrying angle and the flexion angle (Figure 7-7a). The coefficient of determination $R^2$ was evaluated for each bowler for three line fits, a linear, an inverse and a quadratic as shown in (Table 7-5). Non parametric Friedman’s ANOVA showed that $R^2$ values were significantly different between the curves ($X^2(2)=12.000$, $p=0.002$) while, Post-hoc analysis with Wilcoxon Signed-Rank Tests identified that $R^2$ values were significantly lower for the inverse line fit when compared to the linear ($z=-2.207$, $p=0.027$) and the quadratic ($z=-2.201$, $p=0.028$).

Table 7-5: Coefficient of determination $R^2$ for every bowler for three line fits for the AnatM

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>Coefficient of determination $R^2$</th>
<th>$p$ - value</th>
<th>$X^2(2)$=12.00</th>
<th>$p=0.002^{**}$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Linear</td>
<td>Inverse</td>
<td>Quadratic</td>
<td>$R^2$</td>
<td>$Y=ax+b$</td>
</tr>
<tr>
<td>1</td>
<td>Med</td>
<td>0.976</td>
<td>0.937</td>
<td>0.997</td>
<td>$a=-0.08$</td>
</tr>
<tr>
<td>2</td>
<td>Fast</td>
<td>0.946</td>
<td>0.935</td>
<td>0.991</td>
<td>$a=-0.16$</td>
</tr>
<tr>
<td>3</td>
<td>Spin</td>
<td>0.978</td>
<td>0.777</td>
<td>0.985</td>
<td>$a=-0.07$</td>
</tr>
<tr>
<td>4</td>
<td>Fast</td>
<td>0.966</td>
<td>0.726</td>
<td>0.995</td>
<td>$a=-0.15$</td>
</tr>
<tr>
<td>5</td>
<td>Med</td>
<td>0.965</td>
<td>0.926</td>
<td>0.995</td>
<td>$a=-0.05$</td>
</tr>
<tr>
<td>6</td>
<td>Med</td>
<td>0.993</td>
<td>0.864</td>
<td>0.995</td>
<td>$a=-0.29$</td>
</tr>
<tr>
<td>Mean (±SD)</td>
<td>0.971</td>
<td>(±0.016)</td>
<td>0.861</td>
<td>(±0.090)</td>
<td>0.991</td>
</tr>
</tbody>
</table>
Figure 7-7: Scatter plot of three repeated measures and regression lines of the carrying angle plotted against flexion for one bowler using (a) anatomical based (AnatM) and (b) motion derived (FuncM1) kinematic models.

The valgus angulation did not appear to significantly decrease with flexion when using a functional based kinematic model (FuncM1) and remained almost constant and equal to the ulnar deviation angle (Figure 7-7b). Average abduction values ranged from 4.7° (± 3.5°) in 30° of flexion to 5.9° (± 1.8°) at 130° (Table 7-6).
### Table 7-6: Carrying angle values at different degrees of active elbow flexion (FuncM1)

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Dynamic Measurements</th>
<th>Flexion angle (p-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Med</td>
<td>8.6 (0.8)</td>
<td>8.3 (0.9)</td>
</tr>
<tr>
<td></td>
<td>8.4 (1.2)</td>
<td>7.6 (0.9)</td>
</tr>
<tr>
<td></td>
<td>7.5 (0.7)</td>
<td>7.7 (1.5)</td>
</tr>
<tr>
<td></td>
<td>7.1 (1.6)</td>
<td>6.2 (0.8)</td>
</tr>
<tr>
<td></td>
<td>0.118</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>9.4 (0.8)</td>
<td>9.9 (0.4)</td>
</tr>
<tr>
<td>Fast</td>
<td>9.8 (0.4)</td>
<td>9.7 (0.2)</td>
</tr>
<tr>
<td></td>
<td>8.8 (0.4)</td>
<td>7.8 (0.6)</td>
</tr>
<tr>
<td></td>
<td>7.4 (1.0)</td>
<td>7.5 (0.4)</td>
</tr>
<tr>
<td>3</td>
<td>2.3 (0.7)</td>
<td>4.6 (0.9)</td>
</tr>
<tr>
<td>Spin</td>
<td>6.0 (1.0)</td>
<td>6.4 (1.3)</td>
</tr>
<tr>
<td></td>
<td>6.9 (1.5)</td>
<td>7.2 (1.6)</td>
</tr>
<tr>
<td></td>
<td>6.7 (1.6)</td>
<td>6.7 (1.1)</td>
</tr>
<tr>
<td></td>
<td>6.6 (0.7)</td>
<td>6.6 (0.7)</td>
</tr>
<tr>
<td>4</td>
<td>0.8 (0.1)</td>
<td>3.4 (0.4)</td>
</tr>
<tr>
<td>Fast</td>
<td>5.6 (0.2)</td>
<td>6.3 (0.1)</td>
</tr>
<tr>
<td></td>
<td>5.9 (0.4)</td>
<td>5.2 (0.6)</td>
</tr>
<tr>
<td></td>
<td>3.7 (0.3)</td>
<td>3.3 (0.3)</td>
</tr>
<tr>
<td>5</td>
<td>2.6 (4.7)</td>
<td>4.0 (4.5)</td>
</tr>
<tr>
<td>Med</td>
<td>4.2 (4.5)</td>
<td>3.4 (2.3)</td>
</tr>
<tr>
<td></td>
<td>3.8 (1.1)</td>
<td>3.4 (1.1)</td>
</tr>
<tr>
<td></td>
<td>3.8 (0.4)</td>
<td>3.9 (0.3)</td>
</tr>
<tr>
<td>6</td>
<td>4.5 (0.5)</td>
<td>6.6 (1.6)</td>
</tr>
<tr>
<td>Med</td>
<td>8.0 (0.9)</td>
<td>8.9 (1.0)</td>
</tr>
<tr>
<td></td>
<td>9.1 (0.4)</td>
<td>8.8 (0.3)</td>
</tr>
<tr>
<td></td>
<td>7.4 (1.7)</td>
<td>7.7 (1.8)</td>
</tr>
<tr>
<td>Mean</td>
<td>4.7 (3.5)</td>
<td>6.1 (2.6)</td>
</tr>
<tr>
<td>(±SD)</td>
<td>7.0 (2.2)</td>
<td>7.0 (2.3)</td>
</tr>
<tr>
<td></td>
<td>7.0 (2.0)</td>
<td>6.7 (2.0)</td>
</tr>
<tr>
<td></td>
<td>6.0 (1.8)</td>
<td>5.9 (1.8)</td>
</tr>
</tbody>
</table>

Intra-subject variability during active flexion movements was assessed with the standard deviations for both the FuncM1 and for the AnatM compared at every 15° of flexion (Table 7-7); there was no difference between the intra-subject variability for both models (p=0.837). However, a paired t-test between the mean standard deviations for the 6 subjects showed that with the FuncM1 there was less inter-subject variability (Figure 7-8) (p<0.001).
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Table 7-7: Means, 95% confidence intervals, standard deviations of measurements and $p$-values of the intra-subject variability during active flexion movements

<table>
<thead>
<tr>
<th>Flexion angle</th>
<th>Mean</th>
<th>Std. Error</th>
<th>95% confidence intervals</th>
<th>p-values</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lower Bound</td>
<td>Upper Bound</td>
</tr>
<tr>
<td>30°</td>
<td>1.19</td>
<td>0.33</td>
<td>0.33</td>
<td>2.04</td>
</tr>
<tr>
<td>45°</td>
<td>1.30</td>
<td>0.42</td>
<td>0.19</td>
<td>2.40</td>
</tr>
<tr>
<td>60°</td>
<td>1.28</td>
<td>0.37</td>
<td>0.31</td>
<td>2.25</td>
</tr>
<tr>
<td>75°</td>
<td>1.07</td>
<td>0.24</td>
<td>0.43</td>
<td>1.69</td>
</tr>
<tr>
<td>90°</td>
<td>0.90</td>
<td>0.17</td>
<td>0.45</td>
<td>1.34</td>
</tr>
<tr>
<td>105°</td>
<td>0.80</td>
<td>0.17</td>
<td>0.34</td>
<td>1.25</td>
</tr>
<tr>
<td>120°</td>
<td>0.80</td>
<td>0.13</td>
<td>0.44</td>
<td>1.15</td>
</tr>
<tr>
<td>130°</td>
<td>0.63</td>
<td>0.07</td>
<td>0.43</td>
<td>0.83</td>
</tr>
</tbody>
</table>

Table 7-8: Paired t-test comparing the inter-subject variability between the kinematic models AnatM and FuncM1 at specific flexion angles

<table>
<thead>
<tr>
<th>Flexion angle</th>
<th>AnatM</th>
<th>FuncM1</th>
<th>Model (p-values)</th>
</tr>
</thead>
<tbody>
<tr>
<td>30°</td>
<td>5.31</td>
<td>3.28</td>
<td>0.00003***</td>
</tr>
<tr>
<td>45°</td>
<td>6.35</td>
<td>2.57</td>
<td></td>
</tr>
<tr>
<td>60°</td>
<td>7.24</td>
<td>2.07</td>
<td></td>
</tr>
<tr>
<td>75°</td>
<td>8.15</td>
<td>2.25</td>
<td></td>
</tr>
<tr>
<td>90°</td>
<td>8.53</td>
<td>1.99</td>
<td></td>
</tr>
<tr>
<td>105°</td>
<td>8.22</td>
<td>1.99</td>
<td></td>
</tr>
<tr>
<td>120°</td>
<td>7.29</td>
<td>1.77</td>
<td></td>
</tr>
<tr>
<td>130°</td>
<td>6.48</td>
<td>1.84</td>
<td></td>
</tr>
</tbody>
</table>

***. The mean difference is significant at the 0.001 level.
7.4.3 Measurement of adduction-abduction during forearm rotation

The influence of the range of forearm rotation to the abduction angle was also investigated in this study. Repeated measures analysis of variance (ANOVA) tests and post-hoc analysis showed that abduction angles were significantly different at every 20° of pronation. Table 7-9 and Table 7-10 summarise the mean abduction angles for each bowler at every 20° of elbow pronation when employing an anatomical (AnatM) and a motion derived (FuncM1) model respectively. For the AnatM model results showed that the values of abduction linearly increased ($R^2>87\%$) with pronation ranging from 4.5° ($\pm6.2°$) of adduction in 40° to 6.3° ($\pm6.4°$) of abduction in 140° pronation ($p<0.001$). Post-hoc analysis showed that abduction angles were significantly different at every 20° of pronation.
### Table 7-9: Carrying angle values at different degrees of active forearm rotation (AnatM)

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>40°</th>
<th>60°</th>
<th>80°</th>
<th>100°</th>
<th>120°</th>
<th>140°</th>
<th>R²</th>
<th>Pronation angle (p-value)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Med</td>
<td>0.1</td>
<td>1.6</td>
<td>2.8</td>
<td>4.5</td>
<td>5.9</td>
<td>6.0</td>
<td>0.950</td>
<td>0.000***</td>
</tr>
<tr>
<td></td>
<td></td>
<td>(1.1)</td>
<td>(1.5)</td>
<td>(2.2)</td>
<td>(2.1)</td>
<td>(1.6)</td>
<td>(1.1)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>Fast</td>
<td>-5.5</td>
<td>-1.8</td>
<td>-1.3</td>
<td>1.6</td>
<td>3.6</td>
<td>6.4</td>
<td>0.997</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>(1.9)</td>
<td>(1.3)</td>
<td>(1.2)</td>
<td>(1.3)</td>
<td>(1.4)</td>
<td>(0.9)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>Spin</td>
<td>0.8</td>
<td>3.8</td>
<td>7.2</td>
<td>10.0</td>
<td>12.7</td>
<td>14.7</td>
<td>0.988</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>(1.1)</td>
<td>(0.6)</td>
<td>(0.7)</td>
<td>(0.7)</td>
<td>(1.0)</td>
<td>(1.2)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>Fast</td>
<td>1.4</td>
<td>3.2</td>
<td>5.6</td>
<td>8.1</td>
<td>10.5</td>
<td>12.0</td>
<td>0.873</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>(1.4)</td>
<td>(1.2)</td>
<td>(1.1)</td>
<td>(1.5)</td>
<td>(2.0)</td>
<td>(1.9)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>Med</td>
<td>-10.2</td>
<td>-7.8</td>
<td>-4.9</td>
<td>-2.3</td>
<td>-0.3</td>
<td>1.2</td>
<td>0.983</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>(0.5)</td>
<td>(1.3)</td>
<td>(1.7)</td>
<td>(2.2)</td>
<td>(2.1)</td>
<td>(1.5)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>Med</td>
<td>-13.3</td>
<td>-10.8</td>
<td>-8.6</td>
<td>-6.1</td>
<td>-4.5</td>
<td>-2.3</td>
<td>0.998</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>(2.1)</td>
<td>(2.2)</td>
<td>(2.6)</td>
<td>(2.3)</td>
<td>(1.6)</td>
<td>(1.9)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean</td>
<td>(±SD)</td>
<td>-4.5</td>
<td>-2.0</td>
<td>0.1</td>
<td>2.6</td>
<td>4.6</td>
<td>6.3</td>
<td>0.965</td>
<td>0.048</td>
</tr>
<tr>
<td></td>
<td></td>
<td>(6.2)</td>
<td>(6.1)</td>
<td>(6.2)</td>
<td>(6.1)</td>
<td>(6.5)</td>
<td>(6.4)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

***. The mean difference is significant at the 0.001 level.

For the FuncM1 model the repeated measures ANOVA showed that the mean carrying angle differed with internal rotation. Post-hoc analysis however, with the Bonferroni correction applied, revealed that there were no significant differences between the abduction angles at different degrees of elbow pronation.
Table 7-10: Carrying angle values at different degrees of active forearm rotation (FuncM1)

<table>
<thead>
<tr>
<th>Bowler Type</th>
<th>Dynamic Measurements</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>40°</td>
</tr>
<tr>
<td>1 Med</td>
<td>4.6 (0.9)</td>
</tr>
<tr>
<td>2 Fast</td>
<td>11.3 (0.9)</td>
</tr>
<tr>
<td>3 Spin</td>
<td>6.4 (0.7)</td>
</tr>
<tr>
<td>4 Fast</td>
<td>4.1 (1.4)</td>
</tr>
<tr>
<td>5 Med</td>
<td>13.1 (1.0)</td>
</tr>
<tr>
<td>6 Med</td>
<td>0.7 (1.8)</td>
</tr>
<tr>
<td>Mean (±SD)</td>
<td>6.6 (4.6)</td>
</tr>
</tbody>
</table>

* The mean difference is significant at the 0.05 level.

Mean inter-subject variability during active forearm rotation movements was significantly lower (p<0.001) for the FuncM1 model when compared to the AnatM (Table 7-11) whilst, no statistical differences were found when comparing the intra-subject variability between the two models at every 20° of pronation (p=0.09) (Table 7-12).
Table 7-11: Paired t-tests comparing the inter-subject variability between the kinematic models AnatM and FuncM1 at specific pronation angles

<table>
<thead>
<tr>
<th>Pronation angle (°)</th>
<th>AnatM</th>
<th>FuncM1</th>
<th>Model (p-values)</th>
</tr>
</thead>
<tbody>
<tr>
<td>40°</td>
<td>6.24</td>
<td>4.57</td>
<td>0.0002***</td>
</tr>
<tr>
<td>60°</td>
<td>6.07</td>
<td>4.81</td>
<td></td>
</tr>
<tr>
<td>80°</td>
<td>6.18</td>
<td>4.90</td>
<td></td>
</tr>
<tr>
<td>100°</td>
<td>6.15</td>
<td>5.44</td>
<td></td>
</tr>
<tr>
<td>120°</td>
<td>6.47</td>
<td>5.45</td>
<td></td>
</tr>
<tr>
<td>140°</td>
<td>6.36</td>
<td>5.11</td>
<td></td>
</tr>
</tbody>
</table>

***. The mean difference is significant at the 0.001 level.

Table 7-12: Means, 95% confidence intervals, standard deviations of measurements and p-values of the intra-subject variability during active movements of forearm rotation

<table>
<thead>
<tr>
<th>Pronation angle (°)</th>
<th>Mean</th>
<th>Std. Error</th>
<th>95% confidence intervals</th>
<th>p-values</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lower Bound</td>
<td>Upper Bound</td>
</tr>
<tr>
<td>40°</td>
<td>1.37</td>
<td>0.12</td>
<td>1.06</td>
<td>1.68</td>
</tr>
<tr>
<td>60°</td>
<td>1.11</td>
<td>0.23</td>
<td>0.50</td>
<td>1.73</td>
</tr>
<tr>
<td>80°</td>
<td>1.07</td>
<td>0.22</td>
<td>0.50</td>
<td>1.63</td>
</tr>
<tr>
<td>100°</td>
<td>1.18</td>
<td>0.25</td>
<td>0.53</td>
<td>1.82</td>
</tr>
<tr>
<td>120°</td>
<td>1.27</td>
<td>0.23</td>
<td>0.66</td>
<td>1.87</td>
</tr>
<tr>
<td>140°</td>
<td>1.34</td>
<td>0.17</td>
<td>0.90</td>
<td>1.77</td>
</tr>
</tbody>
</table>

In order to determine the best fit line for the AnatM model, the coefficient of determination $R^2$ was evaluated for each bowler for three possible line fits; a linear, an inverse and a quadratic. Non parametric Friedman’s ANOVA showed that $R^2$ values were significantly different for the three line fits $\chi^2(2) = 7.636$, $p = 0.022$ (Table 7-13) whilst Post-hoc analysis with Wilcoxon Signed-Rank Tests identified that $R^2$ were significantly lower for the inverse line when compared to the linear ($z=-1.922$, $p=0.046$) and the quadratic ($z=-2.201$, $p=0.028$).
### Table 7-13: Coefficient of determination $R^2$ for every bowler for three line fits for the AnatM

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>Coefficient of determination $R^2$</th>
<th>Linear $Y=ax+b$</th>
<th>Inverse $Y=(a/x)+b$</th>
<th>Quadratic $Y=ax^2+bx+c$</th>
<th>$p$-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>$R^2$</td>
<td>$a$</td>
<td>$b$</td>
<td>$R^2$</td>
<td>$a$</td>
</tr>
<tr>
<td>1</td>
<td>Med</td>
<td>0.950</td>
<td>0.07</td>
<td>-2.35</td>
<td>0.980</td>
<td>-427.3</td>
</tr>
<tr>
<td>2</td>
<td>Fast</td>
<td>0.997</td>
<td>0.11</td>
<td>-9.88</td>
<td>0.885</td>
<td>-632.8</td>
</tr>
<tr>
<td>3</td>
<td>Spin</td>
<td>0.988</td>
<td>0.14</td>
<td>-5.10</td>
<td>0.944</td>
<td>-826.8</td>
</tr>
<tr>
<td>4</td>
<td>Fast</td>
<td>0.880</td>
<td>0.11</td>
<td>12.43</td>
<td>0.834</td>
<td>-672.4</td>
</tr>
<tr>
<td>5</td>
<td>Med</td>
<td>0.983</td>
<td>0.15</td>
<td>-19.9</td>
<td>0.820</td>
<td>-834.6</td>
</tr>
<tr>
<td>6</td>
<td>Med</td>
<td>0.998</td>
<td>0.12</td>
<td>-19.9</td>
<td>0.894</td>
<td>-670.7</td>
</tr>
<tr>
<td>Mean (±SD)</td>
<td>0.966 (0.046)</td>
<td>0.893 (0.063)</td>
<td>0.975 (0.044)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
Figure 7-9: Scatter plot of three repeated measures and regression lines of the carrying angle plotted against pronation for one bowler using (a) anatomical based (AnatM) and (b) motion derived (FuncM1) kinematic models.
7.5 Discussion

7.5.1 Measurements of carrying angles in extension

In this study the carrying angle of the dominant arm was measured in 6 male club level bowlers by means of an optical motion tracking system when using an anatomical based (AnatM) and motion derived kinematic model (FuncM1). Average carrying angles of 9.6° (±4.3°) were reported in full elbow extension for the anatomical based kinematic model (AnatM) and 6.9° (±3.9°) for the motion derived (FuncM1) without statistical difference. Although a lack of a standardised protocol when assessing the carrying angle does not allow for direct comparisons between the different studies, it can be seen the values reported in this study lie within the published data whereby, the mean values of carrying angles reported in males have been shown to vary from 6.8° to 18.5° (Table 7-1). With the joint flexed at approximately 90° the average carrying angle significantly decreased to -7.7° (±7.5°) when quantified using the anatomical method.

7.5.2 Measurements of carrying angles during flexion-extension movements

Elbow kinematics as a function of flexion angle was evaluated during active dynamic trials using an anatomical based kinematic model (AnatM) and was found to decrease as a function of joint flexion, which is in accordance to previous studies presenting a similar trend (Chao and Morrey, 1978; Van Roy et al., 2005; Zampagni et al., 2008a). The maximum adduction values at 130° of flexion reported in this study were on average 5° greater than those previously reported (Chang et al., 2008; Van Roy et al., 2005). The motion derived kinematic model (FuncM1) revealed slight differences in the reported carrying angles during flexion and produced kinematic pathways with less variability between subjects. This is justified as the helical axis approach takes under consideration the changing conditions of the capsuloligamentous system surrounding the joint during active and passive dynamic movements. In that sense the FuncM1 represents a kinematic rather than an anthropometric parameter (Van Roy et al., 2005) although the
former is clearly dependent on the latter. Intra-subject variability was similar between the two models during active movements of pure flexion and pronation. Ferreira et al. (2011) showed that intra-subject variability of active flexion pathways was reduced with motion derived coordinate systems when compared to anatomically derived ones. However, in that study kinematic data were collected from human cadavers using computer-controlled servomotors and pneumatic actuators connected to the tendons to generate simulated active elbow flexion whilst for the anatomical based axes cadavers were disarticulated in order to digitise the anatomical landmarks of interest. In this study, subjects were asked to actively flex and extend their joint so it is possible that both models were affected to some extent by skin movement artefacts.

7.5.3 Measurements of adduction-abduction during forearm rotation

The range of adduction-abduction of the elbow during forearm rotation was also evaluated in this study. With the joint fixed at 90° of flexion there appears to be a reciprocal relationship when employing the AnatM model, between the range of pronation and that of abduction with the long axis of the forearm deviating medially about 11° on average. Taking under consideration however, that forearm rotation takes place about an axis passing through the radial head and centre of the distal end of the ulna the deviation reported in this study constitutes more of measure of medial rotation of the radius rather than a measure of the carrying angle. Average abduction angles predicted with the FuncM1 remained almost constant and equal to the ulnar deviation angle whereas the variability with joint motion in the AnatM based model can be attributed to differences in anatomy which do not coincide with joint motion patterns (Ferreira et al., 2011).

7.5.4 The carrying angle during cricket bowling

As discussed earlier in this chapter several studies have outlined the importance of the carrying angle in throwing athletes. In professional throwers it is not uncommon to find carrying angles greater than 15° (Cain et al., 2003). When assessing joint kinematics during sport activities, kinematic models and especially anatomical based ones may be
prone to kinematic cross talk whereby, one joint rotation is being interpreted as another (Section 4.3). In this study the abduction angle was evaluated in both static and dynamic trials and the carrying angle was found to linearly decrease with flexion when using an AnatM ($p<0.01$) but remained essentially constant throughout flexion when evaluated with the FuncM1 kinematic model. Furthermore, during forearm rotation the AnatM measured significantly more medial rotation of the radius than the FuncM1. This variation of the carrying angle with flexion might also help to justify to some of the differences in the mean carrying angles reported in the literature. Bearing this in mind it can be seen that when employing anatomical based kinematic models to look at joint rotations in the elbow flexion values should also be reported along with the carrying angle.

As shown in Table 7-14 the age of the participants in this study varied considerably ranging form 16 to 31 years. Even though, the carrying angle has been shown to change with age in women, men do not show a similar trend (Golden et al., 2007; Yilmaz et al., 2005). However, taking under consideration that overhead throwing does impart high valgus loads to the bowling arm (Werner et al., 2007) often leading to structural changes within the joint such as an increased carrying angle, bowlers 1, 2, and 6 could potentially develop a higher carrying angle. Bowler 6 in particular had the highest carrying angle among the subjects that participated in the study and as it can be seen from Figure 7-10 at the point which the upper arm reached the level of the shoulder had an already bent arm. During active and dynamic movements the bowler was not able to extend his upper arm beyond 19° of flexion. Even though, only one bowler had a large carrying angle that does not allow for a statistical comparison, when looking closer at his bowling action it can be seen that the FuncM1 predicted lower elbow extension angles than the AnatM model so that with the former his action is considered to be borderline with the ICC recommendations whilst with the latter he is considered a thrower. As this study showed that the FuncM1 is less affected by kinematic cross-talk than the AnatM it is believed that the former represents more closely the bowler’s action.
Table 7-14: Anthropometric data and mean elbow angles during bowling

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>Age (yrs)</th>
<th>Flexion angle (at upper arm horizontal)</th>
<th>Min flexion angle (during active flexion)</th>
<th>Carrying angle (CA)</th>
<th>FuncM1 Mean (SD)</th>
<th>AnatM Mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Med</td>
<td>24</td>
<td>4.5 (1.6)</td>
<td>14.0 (1.9)</td>
<td>10.8</td>
<td>14.4 (1.8)</td>
<td>10.1 (2.0)</td>
</tr>
<tr>
<td>2</td>
<td>Fast</td>
<td>19</td>
<td>0.0 (2.2)</td>
<td>0.0 (0.6)</td>
<td>11.8</td>
<td>8.7 (1.8)</td>
<td>7.2 (1.8)</td>
</tr>
<tr>
<td>3</td>
<td>Spin</td>
<td>31</td>
<td>42.0 (0.7)</td>
<td>18.0 (1.6)</td>
<td>3.3</td>
<td>20.7 (2.1)</td>
<td>21.3 (2.1)</td>
</tr>
<tr>
<td>4</td>
<td>Fast</td>
<td>29</td>
<td>26.0 (1.7)</td>
<td>10 (1.5)</td>
<td>11.4</td>
<td>15.9 (1.5)</td>
<td>21.9 (1.5)</td>
</tr>
<tr>
<td>5</td>
<td>Med</td>
<td>16</td>
<td>4.9 (3.9)</td>
<td>5.0 (4.4)</td>
<td>5.6</td>
<td>10.7 (2.0)</td>
<td>6.1 (2.3)</td>
</tr>
<tr>
<td>6</td>
<td>Med</td>
<td>16</td>
<td>32.2 (2.8)</td>
<td>19.0 (2.4)</td>
<td>14.7</td>
<td>15.2 (2.5)</td>
<td>18.0 (2.9)</td>
</tr>
<tr>
<td>Mean (±SD)</td>
<td></td>
<td></td>
<td>9.6 (4.3)</td>
<td>14.3 (4.2)</td>
<td>14.1 (7.2)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Figure 7-10: Bowler 6 at upper arm horizontal (a) side on and (b) front on view

Bowler 4 also had a carrying angle of about 11° which in several studies is considered as the upper limit for male subjects (Chao and Morrey, 1978; Potter, 1895; Van Roy et al.,
2005; Yilmaz et al., 2005) and a restricted range of motion in his elbow joint so that at full extension his elbow was considerably bent. As shown in Table 5-2 similar to Bowler 6 the FuncM1 predicted lower elbow extension angles than the AnatM, with his action considered borderline.

Even though, in the cases of bowlers 4 and 6 it can be argued that they passively extended their joint due to anatomical restrictions, Bowler 3 had a small carrying angle of 3.3° and a bent arm at upper arm horizontal of more than 20° above his minimum flexion angle. Both kinematic models showed during the delivery swing he exceeded the 15° threshold as set by the ICC.

7.6 Summary

The purpose of this study was to calculate the carrying angle of cricket bowlers and evaluate its variability during flexion of the elbow joint using in-vivo kinematic data. The carrying angle of the dominant arm was measured in 6 male club level bowlers using an optical motion tracking system employing an anatomical based and a motion derived kinematic model. Even though, there have been very few in-vivo studies to measure the carrying angle, the mean values reported in this study were within the range reported in the literature for both kinematic models. Elbow rotations calculated during dynamic movements using an anatomical derived (AnatM) model confirmed a linear decrease in the carrying angle with increased elbow flexion and increased medial rotation of the radius with increased pronation. The valgus angulations measured using the functional based kinematic model (FuncM1) did not appear to decrease with flexion and remained equal to the ulnar deviation angle during controlled forearm rotation. In conclusion, both models could be employed when looking at anthropometric data however, when assessing elbow kinematics during bowling a functional kinematic model was shown to decrease the variability between subjects. This shows that this model constitutes a promising tool for use in assessing elbow kinematics during dynamic activities especially in cases where joint abnormalities are present. Further discussion of the results shown in this chapter and an overall discussion of the thesis will be presented in Chapter 8.
In this thesis a method to measure the three-dimensional dynamic movement of the elbow joint during cricket bowling was mathematically developed and experimentally validated. In this Chapter an overall discussion of the thesis results, conclusions and recommendations for future studies are presented.
8.1 Summary of Results

The elbow joint is a complex structure with a remarkable capability in both its range of motion as well as the forces it is capable of bearing. In cricket bowling, elbow biomechanics plays a crucial role in the delivery swing as the laws of the game prohibit elbow flexion-extension during the arm acceleration phase of bowling. Since the advent of high-speed video photography however, it has been revealed that some straightening occurs in bowlers who have actions that are traditionally considered in accordance with the law.

There are many difficulties in measuring elbow kinematics especially under highly dynamic movements such as cricket bowling; these have so far led to different laboratories employing different models to analyse bowling legality. However, as the accuracy of the predictions of these models depends on the accuracy and repeatability of the input data, the results from these laboratories may not be directly comparable.

Within this work, a thorough study of the available elbow measurement techniques and the technical obstacles of accurately measuring the joints kinematics during bowling were presented. This has led to the development and validation of a biomechanical model to assess bowling legality in cricket. The new model meets all of the specifications of a measurement method to be used in sports-related biomechanical studies measuring joints kinematics at high speeds over the functional range of motion of the elbow whilst allowing for the subject to move freely within a large volume. In addition, this model can be easily implemented within the existing experimental protocol for assessing bowling legality in cricket as proposed by the England and Wales Cricket Board (ECB).
8.1.1 A functional based kinematic model of the elbow joint

In this study a biomechanical model based on a helical axis kinematic model of the elbow flexion-extension and pronation-supination axes was presented in order to measure elbow kinematics in cricket bowling. A total of nine bowlers participated in the study and excluding one thrower the sample mean of elbow extension angles were within the 15° threshold as set by the ICC in 2004 irrespective of bowling type or speed. However, both the fast and medium pace bowlers that participated in this study exceeded the first set of ICC regulations whereby fast bowlers were allowed 10° and medium pace bowlers 7.5° of elbow extension. The within-day coefficient of multiple correlations (CMC) was used to evaluate the intra-subject repeatability for one over of balls and results showed that the proposed model was highly repeatable for all three elbow rotations.

In order to validate the model and compare the accuracy of the method to existing techniques the studies described below were undertaken.

Study I: Anatomical and functional based upper limb models

This study investigated the effect that different definitions of the elbow flexion-extension and pronation-supination axes may have on quantifying elbow kinematics during cricket bowling by comparing the repeatability of the kinematic data obtained from three kinematic models, one anatomical (AnatM) and two functional (FuncM1 and FuncM2) during two single joint movements and cricket bowling.

All three models measured the same range of motion (RoM) for flexion and pronation during controlled single joint movements. However, the functional based models measured smaller overall RoM in adduction in comparison to the anatomical based suggesting that the former are less prone to kinematic crosstalk. The results during bowling showed that all three kinematic models produced similar kinematic waveforms and no statistical differences were found between them for extension angles. The functional models however, calculated significantly less pronation than the anatomical models.
Functional and anatomically-derived axes can be directly compared when comparing the AnatM to the FuncM1 models. During controlled single joint movements both models produced similar shaped graphs however, for flexion-extension an offset existed between their curves. Even though this offset can be attributed to the different locations and orientation of the flexion axes between the two models it is difficult to identify which model provides the most clinically representative data as lack of direct measurements do not allow for the actual movement between the markers and the underlying bones to be quantified. In this study the effect of the choice of kinematic model on joint angles was assessed through a series of indirect measures in terms of the recorded range of motion and consistency in the kinematic data.

A review of the available literature shows that both anatomical and motion derived coordinate frame definitions are associated with significant errors. Anatomical based models take into account the relative movement between the co-ordinate frames of two adjacent segments for the calculation of joint kinematics but have been widely associated with significant STA and landmark misplacement errors as discussed in Section 4.3.1 (Cappozzo et al., 1995; Cutti et al., 2005; Della Croce et al., 2005; Leardini et al., 2005; Peters et al., 2010). Functional models on the other hand optimise the relative movement between the two adjacent co-ordinate frames based on joint constraints. A number of studies have drawn attention to the method being susceptible to significant measurement noise (Cheze et al., 1998; Stokdijk et al., 1999; Woltring et al., 1985). Johnson et al. (2004) showed that although rotation about the knee helical axis seemed to be repeatable and time step insensitive, displacements were highly dependent on time step and increment size with smaller rotation angles between the calculations improving accuracy (Johnson et al., 2004). Duck et al. (2004) also suggested that the mean IHA should only be calculated at rotations of greater than 5° whilst Stokdijk et al. (2000) used only the middle 50° from each movement for the calculation of the elbow optimal helical axis. Previous investigations on knee kinematics however, have also shown that the optimal helical knee flexion-extension axis and the transepicondylar axis produce similar kinematic data (Besier et al., 2003; Churchill et al., 1998) and this study showed that this is also the case at the elbow joint.
Both motion derived kinematic models (FuncM1 and FuncM2) showed similar intra-subject repeatability for all three rotations, producing identical pronation-supination kinematic waveforms in particular. However, even though, FuncM2 employed the same flexion-extension and pronation-supination optimal helical axes as the FuncM1 it measured no abduction. This is because with the FuncM2 both the long axes of the humerus and that of the forearm were defined as being perpendicular to the flexion optimal helical axis.

Study II: The effect of digitisation of the humeral epicondyles on quantifying elbow kinematics during cricket bowling

The objective of this study was to look at the effect that the elbow orientation during calibration and the choice of upper arm tracking cluster have on the measurement of elbow kinematics. The mean elbow angles were compared for four different elbow positions and two upper arm clusters during a controlled movement of pure flexion-extension and during cricket bowling. Results showed that digitising the humeral epicondyles with the joint flexed at 90° or in full extension and relating them to a distal upper arm cluster, close to the joint are the most reproducible methods of assessing bowling action and should be considered when a bowler’s action is called into question.

With the elbow flexed and the humerus internally rotated during calibration, the kinematic model was found to measure significantly more abduction and pronation than the other three positions during movements of pure flexion suggesting that the relative movement between the clusters and underlying bones significantly affects the digitised landmarks at this elbow orientation.

With respect to the choice of the technical cluster employed to replace the digitised landmarks during active movements results showed that a cluster placed close to the elbow joint, in this case at the distal part of the upper arm, improved the accuracy of the measurements especially in the estimated abduction-adduction and pronation-supination angles which are confirmed to be the most sensitive to all experimental errors, including kinematic cross talk (Della et al., 2005) and soft tissue artifact (Leardini et al., 2005). This is because placing the cluster near the elbow joint avoids the skin...
deformation caused by the bicep contraction which causes most of the reported errors, especially during extended humeral rotation (Cutti et al., 2005).

**Study III: Single and Double anatomical landmark calibration**

The aim of this study was to compare the effect that single and double anatomical landmark calibration have on elbow rotations during cricket bowling. Results showed that the intra-tester repeatability of the two methods was similar and no significant differences were found in the calculated elbow angles between the models. In this study however, double calibration was performed between two static elbow orientations; with the joint in full extension and flexed at 90° and subsequently compared to a single calibration with the elbow flexed at 90°. As discussed in Study II these two elbow orientations were shown to improve the accuracy and repeatability of the measurements. Therefore, it is possible that by choosing different elbow orientations during the calibration process differences in the repeatability of the methods would be noted.

A number of studies have reported joint kinematics sensitivity to rotation axes and the precision of anatomical landmark determination (Della et al., 2005; Donati et al., 2008; Small et al., 1993) identifying landmark misplacement as one of the most critical sources of error. Anatomical landmark calibration has been shown to improve the repeatability of the kinematic data (Cappozzo et al., 1995) however, possible erroneous palpation of these areas during the digitisation processes is mitigated during dynamic activities. The latter can be easily caused by lack of experience of the investigator as well as when assessing the performance of bowlers with elbow deformities such as hyperextension and large carrying angle. As double calibration has been shown to compensate for STA in knee rotations and translations, even with large misplacement errors on the anatomical coordinates of the knee epicondyles (Stagni et al., 2006) it is therefore, suggested that when bowling legality is assessed using anatomical based kinematic models that double calibration should be performed to improve the reliability of joint kinematics and account for inexperienced researchers and bowlers with joint deformities.
Study IV: Markers onto the epicondyles

In this study the functional based kinematic model (FuncM1) was compared to an anatomical based kinematic model (AnatM2) in which the elbow flexion axis was defined by placing two markers directly onto the humeral epicondyles instead of digitising their positions as discussed in Study I. As in the case of every anatomically derived kinematic model the quality of subject marking depends on marking accuracy in identifying the correct landmark location but also on the inter-tester variability of repeated marking procedures (Benedetti et al., 1998; Rabuffetti et al., 2002). Both these issues depend on the skill of the researcher as well as the anatomical characteristics of the participants. Studies have shown that when skilled researchers performed the marking of anatomical landmarks on one subject then the inter-tester variability ranged from 11.5 to 24.8 mm while the intra-tester variability from 4.8 to 21.0 mm (Della Croce et al., 1999). Markers placed on the elbow epicondyles can be highly susceptible to skin movement artefacts especially as the joint reaches full extension. In order to improve the accuracy of the AnatM2 markers were placed on each subject with their joint in full extension. Results showed that both models calculated the same degree of elbow extension during bowling however, the FuncM1 model showed higher intra-subject repeatability than the AnatM2 in the estimated abduction-adduction and pronation-supination angles producing more reproducible elbow angles between trials. This clearly shows that a functional approach significantly improves the repeatability of elbow measurement techniques particularly in measuring subtle elbow rotations, such as elbow abduction and forearm pronation.

Study V: The effect of one frame

In this study the effect that one frame may have onto the average elbow flexion-extension angles was investigated. Looking at the flexion-extension graphs in Chapter 5 it can be seen that bowlers maintain a constant flexion angle near the point of upper arm horizontal but tend to extend or in some cases to slightly flex their joint through release, so that most of the variation in the reported RoM actually takes place during the last frames of the delivery. Results confirmed this observation and showed significant
differences in the mean elbow extension angles when including or excluding one of the final frames from the analysis. These changes in the last frames of the delivery show that the point of ball release is crucial when determining bowl legality under the current formulation of the laws and should be taken into account as it may underestimate or overestimate changes in the recorded motion.

The maximum acquisition rate of the video cameras used in this study were 100 Hz so in order for them to be synchronised with the optical motion tracking system the latter had to be operating at 200 Hz. Based on the findings from this study it is proposed that this is the minimum acquisition rate that both systems should be operating when assessing bowl legality in cricket.

Study VI: The carrying angle in cricket bowlers

The purpose of this study was to calculate the carrying angle of cricket bowlers and evaluate its variability during flexion of the elbow joint using *in-vivo* kinematic data. The carrying angle of the dominant arm was measured in 6 male club level bowlers and the mean values reported in this study were within the range reported in the literature when using both an anatomical (AnatM) and a functional based kinematic (FuncM1) model. Elbow rotations calculated during dynamic movements using the AnatM model showed a linear decrease in the carrying angle with increased elbow flexion while, when employing the FuncM1 the carrying angle did not appear to significantly decrease with flexion. The question again is whether it can be concluded which models predicts the most clinically representative data. From a robotics point of view the carrying angle should not change with flexion as it constitutes an intrinsic, geometric parameter (Cutti et al. 2006), from a clinical perspective however, as discussed in Section 2.2.3 studies have identified that during flexion-extension movement slight mobility of the humeroradial and humeroulnar joints does occur the frontal plane in the form of abduction-adduction about the ulna (Lockard, 2006). Based once again in indirect measures of joint rotation and as both kinematic models showed similar inter-subject reliability it is concluded that they could be both used when measuring the carrying angle at full elbow extension. Further research needs to be carried out in order to
determine which model is more clinically relevant during joint flexion-extension movements however, taking into consideration that when assessing joint kinematics the elbow is considered as a hinge joint allowing for movement in one plane, a functional kinematic model that minimises the range of the recorded adduction-abduction motion may be more representative during motion analysis.

In conclusion, both functional and anatomically-derived kinematic models could be employed when assessing elbow kinematics during bowling as both definitions produced similar flexion-extension kinematic waveforms and high intra-subject repeatability while no statistical differences were found between the models for extension angles. It should be noted however, that this study included a single experienced observer to obtain all the measurements and intra-tester repeatability was assessed only for 1 session and for a limited number of frames (from 20ms prior to upper arm horizontal to 20ms after the ball is released) so there were no systematic errors of the two models due to reapplication of markers that could have influenced the data. Improved repeatability of the functional approach to the anatomical might be more pronounced with examiners who are less experienced or with subjects who have large bony deformities at their elbow. All of the bowlers that participated in this study showed carrying angles within the ranges reported in the literature for the normal male population and for collegiate baseball pitchers however, in a study by Chin et al. (2009) it was shown that elite cricket players displayed larger amount of forearm abduction and fixed forearm flexion in comparison to high performance players.

The recommendation from this work is that a functional-based approach that is not dependent on the accurate identification of anatomical landmarks and is therefore less affected by errors due to landmark misplacements should be employed to determine bowling legality especially in cases where forearm rotation through release needs to be assessed. In case examiners wish to employ anatomical based models, this study showed that it is best to digitise the position of humeral epicondyles rather than place markers directly onto the skin. With respect to the digitisation process, it is recommended that double calibration of the humeral epicondyles should be performed in two static positions with the joint in full extension and flexed at 90° and defined relative to a
technical frame of distal upper arm triad closer to the joint during dynamic trials in order to improve the reliability of the data and reduce the effect that soft tissue artifacts may have onto the measured elbow angles.

8.2 Other Errors and limitations of the current study

In this study a kinematic model based on motion derived coordinate frames was developed and validated through experimental work and sensitivity analysis, in order to measure elbow kinematics in cricket bowling. As discussed in Chapter 4 even though, quantitative motion analysis techniques have provided an experimental framework to answer to a variety of problems related to joint kinematics there are still sources of error contributing to the measurements, justifying to some extent the practical differences in the protocols used by different researchers. The errors directly related to the measurement of the elbow motion during dynamic activities such as skin movement artefacts, observer handling errors of the calibration wand and the effect of 1 frame on the mean extension angles have been addressed in Chapter 6. However, other errors which may affect to some extent the quality of the model do exist and include:

- **System and modelling errors**: These errors are directly related with the optical motion tracking system used to capture the motion of the bowlers ranging from 0.1 - 0.4 mm. When compared to other error sources the system calibration errors were assumed to be of negligible magnitude however, these errors can be much higher when the subject moves away from the centre of the volume. In this study, an accuracy of ±0.2 mm was always obtained the cameras were strategically placed in the MCC indoor school allowing for the full delivery action as well as the initial ball flight to be captured whilst the frames of interest; from the point that the upper arm reached the level of shoulder to ball release always took place in the middle of the volume.

- **In vivo estimation of the glenohumeral joint centre**: Several functional algorithms have been proposed for determining the centre of the glenohumeral joint from the relative movement of adjacent body segments. However, considering the shoulder as a perfect ball and socket joint with a fixed joint
centre introduces errors to the measurements. In addition several studies have reported errors associated with the algorithms used in the estimation (Lempereur et al., 2010; Stokdijk et al., 2000). In this study the glenohumeral joint centre was estimated with the method of Gamage and Lasenby (2002) as this algorithm has been shown to return the smallest distance between the estimated joint centre and the centre of the humeral head (Lempereur et al., 2010).

- **Simplifications in joint rotations:** These errors are caused by the simplification of the shoulder joint into a ball and socket joint, the elbow joint into a hinge joint and interpolation errors.

- **Assessing intra and inter-subject repeatability through indirect measures:** The studies presented in this work concentrated on assessing the rotational errors of anatomical and functional based kinematic models on joint angles. Direct measures to report the actual movement of markers with respect to underlying bones have not been looked at within this work and can form the basis of future studies.

- **Number of subjects and type of bowlers:** All of the studies presented in this thesis have made use of a limited number of bowlers. This work could have benefited from the inclusion of a larger number of bowlers so as to distinguish between different types of bowling techniques but also from the inclusion of some bowlers with elbow joint abnormalities, where the transepicondylar axis is not the best representation of the flexion-extension axis of the elbow in order to truly assess the measurement errors between the different models.
8.3 Conclusions and recommendations for future work

The variability in the results from this study indicates that the type of marker placement and kinematic model chosen by the investigator to determine elbow angle can have a profound impact on the results obtained. This data can also provide a better understanding of the kinematic variability during cricket bowling and used to review critically some of the experimental protocols used when measuring the legality of a bowler’s action. Based on the findings from this research future advancements can be developed in the following areas:

- This research developed an upper limb kinematic model with functional axes at the elbow joint. Future research extending this work could look into employing functional axes at the wrist joint for other upper limb activities. However, for cricket bowling this is not required.

- As discussed earlier in Section 8.2 one of the main limitations of this study was the number of subjects which did not allow for direct comparisons between different types of bowlers. Future work should therefore include different types of bowlers and especially spin bowlers where quantitative analysis has been limited until now as well as bowlers with an elbow joint with significantly different anatomy such as fixed flexion deformation, extreme hyperextension, large carrying angle, and a significantly varying carrying angle with forearm pronation. Future research should also aim at exploring the contribution of segmental angular velocities to the ball’s speed as well as the amount of forearm rotation across different bowling techniques. The latter would be of paramount importance in reviewing critically the scientific evidence behind the ICC regulations and might influence the formulation of the laws of the game.

- In addition, several studies have suggested that through release a significant amount of shoulder axial rotation takes place however, up to this point shoulder kinematics during bowling have not been investigated. Current advances in tracking scapular motion during dynamic activities (Prinold et al., 2011; Shaheen
et al., 2011a; Shaheen et al., 2011b) could be incorporated within the current model to measure glenohumeral rotations during bowling.

- This research established a protocol able to measure elbow kinematics during bowling which can be used within a laboratory. However, as it is preferable to record data under match conditions future work should aim at effectively combining accelerometers and rate gyroscopes to measure joint angles during match conditions, providing that the accuracy of the data reaches the level of the protocol established in this study as defined in Appendix IV and also that an accurate way is established to identify ball release.
Reference List


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Presentations and Publications

Conference Presentations


Journal Publications


Appendix I: Estimation of the Centre of Rotation (CoR) of the Glenohumeral joint

Gamage and Lasenby proposed a functional method for estimating the Centre of Rotation (CoR) of any human joint. In this case this idea can be employed in order to define the CoR of the glenohumeral joint assuming that a set of vectors on a body rotates around a time varying axis of rotation with the fixed CoR while, the ‘tips’ of the vectors lie on co-centric spheres (Figure 1b).

The least squares cost function assuming there are $P$ markers and $N$, frames is:

$$C = \sum_{p=1}^{P} \sum_{k=1}^{N} \left[ (V_k^p - m^2) - (r^p)^2 \right]^2$$

Where $V_k^p$ represents the $p^{th}$ vector in the $k^{th}$ time instance, $m$ represents the centre of rotation and $r$ is the radius of the sphere.

In order to estimate $r^p$ and $m$ that minimise the cost function $C$ we must differentiate with respect to $r^p$ so:
\[ r^p = \sqrt{\frac{1}{N} \cdot \sum_{k=1}^{N} (V^p_k - m)^2} \]  

(2)

And differentiating with respect to \( m \) we arrive at the following:

\[ \sum_{p=1}^{P} \sum_{k=1}^{N} \left[ (V^p_k - m^2) \{ (V^p_k - m^2) - (r^p)^2 \} \right] = 0 \]  

(3)

Substituting Eq.(13) into (14):

\[ \sum_{p=1}^{P} \left[ (\overline{V}^p)^3 - \overline{V}^p \cdot (\overline{V}^p)^2 \right] = 2 \sum_{p=1}^{P} \left[ \frac{1}{N} \cdot \sum_{k=1}^{N} V^p_k (V^p_k \cdot m) - \overline{V}^p (m \cdot \overline{V}^p) \right] \]  

(4)

Where:

\[ \overline{V}^p = \frac{1}{N} \sum_{k=1}^{N} V^p_k \]  

(5)

\[ (\overline{V}^p)^2 = \frac{1}{N} \sum_{k=1}^{N} (V^p_k)^T V^p_k \]  

(6)

\[ (\overline{V}^p)^3 = \frac{1}{N} \sum_{k=1}^{N} V^p_k \cdot (V^p_k)^T V^p_k \]  

(7)

Eq. 15 can be expressed into the following simplified equation (Eq. 19) which can then be used to estimate the CoR \( m \).

\[ A \cdot m = b \]  

(8)

Where \( A \) is a \((3 \times 3)\) matrix given by:

\[ A = 2 \sum_{p=1}^{P} \left\{ \frac{1}{N} \cdot \sum_{k=1}^{N} V^p_k \cdot (V^p_k)^T - \left( \overline{V}^p \cdot (\overline{V}^p)^T \right) \right\} \]  

(9)

\[ b = \sum_{p=1}^{P} \left[ (\overline{V}^p)^3 - (\overline{V}^p) \cdot (\overline{V}^p)^2 \right] \]  

(10)
Appendix II: Euler Angles

For the description of the three dimensional movement of joints two coordinate systems are introduced. The \( i, j, k \) system is fixed representing the unit base vectors of the humeral coordinate frame while the \( I, J, K \) system rotates representing the unit base vectors of the anatomical coordinate frame of the forearm.

In this case the sequence of the elbow joint rotations is to rotate the anatomical coordinate frame of the forearm about the three humeral axes in the following succession: first rotate about the \( z \)-axis which is the flexion/extension axis by an angle \( \vartheta_3 \), then about the varus/valgus \( x \)-axis by \( \vartheta_2 \), and finally rotate about the pronation/supination \( y \)-axis by \( \vartheta_1 \). This sequence of rotations is equivalent to a rotation matrix \([R_{xyz}]\).

The Euler transformations for each rotation are given below:

\[
R = R_z(\vartheta_3)R_x(\vartheta_2)R_y(\vartheta_1)
\]  
(1)

\[
R_x = \begin{bmatrix}
1 & 0 & 0 \\
0 & \cos(\vartheta_1) & -\sin(\vartheta_1) \\
0 & \sin(\vartheta_1) & \cos(\vartheta_1)
\end{bmatrix}
\]  
(2)

\[
R_y = \begin{bmatrix}
\cos(\vartheta_2) & 0 & \sin(\vartheta_2) \\
0 & 1 & 0 \\
-\sin(\vartheta_2) & 0 & \cos(\vartheta_2)
\end{bmatrix}
\]  
(3)

\[
R_z = \begin{bmatrix}
\cos(\vartheta_3) & -\sin(\vartheta_3) & 0 \\
\sin(\vartheta_3) & \cos(\vartheta_3) & 0 \\
0 & 0 & 1
\end{bmatrix}
\]  
(4)

The transformation matrix corresponding to the Euler rotations \( \vartheta_3, \vartheta_2, \vartheta_1 \) in the sequence \( z, x', y'' \) is:

\[
[R] = \begin{bmatrix}
\cos \vartheta_3 & -\sin \vartheta_3 & 0 \\
\sin \vartheta_3 & \cos \vartheta_3 & 0 \\
0 & 0 & 1
\end{bmatrix}
\begin{bmatrix}
\cos \vartheta_2 & 0 & \sin \vartheta_2 \\
0 & 1 & 0 \\
-\sin \vartheta_2 & 0 & \cos \vartheta_2
\end{bmatrix}
\begin{bmatrix}
1 & 0 & 0 \\
0 & \cos \vartheta_1 & -\sin \vartheta_1 \\
0 & \sin \vartheta_1 & \cos \vartheta_1
\end{bmatrix}
\]  
(5)
### Appendix III: Mean flexion-extension elbow angles for all the Bowlers

Table 1: Mean elbow angles, *within-day* CMC values of elbow kinematics during bowling

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>Age</th>
<th>Elbow Flexion-Extension Angles (°)</th>
<th>At upper arm horizontal</th>
<th>Mean (SD)</th>
<th>Within – day CMC</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Med</td>
<td>24</td>
<td>12.4 (1.1)</td>
<td>15.8 (1.8)</td>
<td></td>
<td>0.912</td>
</tr>
<tr>
<td>2</td>
<td>Med</td>
<td>20</td>
<td>4.5 (1.6)</td>
<td>14.4 (3.5)</td>
<td></td>
<td>0.940</td>
</tr>
<tr>
<td>3</td>
<td>Fast</td>
<td>22</td>
<td>10.7 (1.4)</td>
<td>10.0 (1.8)</td>
<td></td>
<td>0.862</td>
</tr>
<tr>
<td>4</td>
<td>Fast</td>
<td>19</td>
<td>0.0 (2.2)</td>
<td>8.7 (0.7)</td>
<td></td>
<td>0.974</td>
</tr>
<tr>
<td>5</td>
<td>Spin</td>
<td>31</td>
<td>42.0 (0.7)</td>
<td>20.7 (2.1)</td>
<td></td>
<td>0.861</td>
</tr>
<tr>
<td>6</td>
<td>Fast</td>
<td>29</td>
<td>26.0 (1.7)</td>
<td>15.9 (1.5)</td>
<td></td>
<td>0.879</td>
</tr>
<tr>
<td>7</td>
<td>Med</td>
<td>16</td>
<td>4.9 (3.9)</td>
<td>10.7 (2.0)</td>
<td></td>
<td>0.864</td>
</tr>
<tr>
<td>8</td>
<td>Med</td>
<td>16</td>
<td>32.2 (2.8)</td>
<td>15.2 (2.5)</td>
<td></td>
<td>0.910</td>
</tr>
<tr>
<td>9</td>
<td>Spin</td>
<td>19</td>
<td>-5.5 (1.2)</td>
<td>13.2 (2.0)</td>
<td></td>
<td>0.985</td>
</tr>
<tr>
<td>Mean (SD)</td>
<td></td>
<td></td>
<td>13.8 (3.7)</td>
<td></td>
<td></td>
<td>0.91 (0.05)</td>
</tr>
</tbody>
</table>
MEASURING ELBOW KINEMATICS IN CRICKET BOWLING

Bowler 1

Bowler 2

Bowler 3
MEASURING ELBOW KINEMATICS IN CRICKET BOWLING

Bowler 4

Bowler 5

Bowler 6
Figure 1: Mean elbow flexion-extension angles for three bowlers plotted against normalised % of bowling action.
1. Introduction

In this thesis a method to measure the three-dimensional dynamic movement of the elbow joint during cricket bowling was mathematically developed and experimentally validated. The protocol presented in this section is largely based on the findings from this research but also on the suggestions from the England and Wales Cricket Board (ECB) for the review of bowlers with suspected illegal bowling actions (Bowling Review Process).

As the final revision of Law 24, (MCC, “Laws of the Game of Cricket”, 2000 code, 3rd edition, 2008) states: “A ball is fairly delivered in respect of the arm if once the bowler’s arm has reached the level of the shoulder in the delivery swing, the elbow joint is not straightened partially or completely from that point until the ball has left the hand. This definition shall not debar the bowler from flexing or rotating the wrist in the delivery swing.”

However, since the advent of high-speed video photography it has been revealed that some straightening occurs in bowlers who have actions that are traditionally considered in accordance with the law. The International Cricket Council (ICC) currently allows for 15° of elbow extension for all types of bowlers. It should be however noted that as elbow hyperextension and elbow abduction-adduction are considered involuntary movements that cannot be controlled by the bowler during throwing these are not taken into account.

It is proposed that this protocol should be employed in cases where the legality of the action of a player is challenged, ensuring that the Bowling Review Process is conducted in a fair and consistent manner with respect to both the data acquisition and analysis. The conclusion of the report should be that having analysed the action in the laboratory
and having compared it to the action as used in match situations, whether in the opinion of the expert, the action used by the bowler is within or exceeds the acceptable levels of elbow extension (ECB, 2010).

2. Data acquisition

2.1 Equipment and set up

All biomechanical assessment must be carried out with the player bowling off their normal full run-up, on a correct length cricket pitch (ECB, 2010). A minimum of six synchronised motion tracking cameras must be used at a minimum acquisition rate of 200 Hz to track reflective markers attached to the skin of the bowlers. Two video recorders must also be used at a minimum acquisition rate of 100 Hz to record the delivery swing and to allow synchronisation with the opto-reflective data.

Figure 1: Viewing positions to detect elbow flexion-extension angle during the delivery swing (A): Directions from which elbow flexion-extension can be detected when the upper arm reaches the level of the shoulder (B): Directions from which elbow flexion-extension can be detected prior to ball release (C): Directions from which elbow flexion-extension can be detected at ball release (Modified from: Aginsky and Noakes (2008))
Calibration of all the cameras must be completed prior to data acquisition and accuracy of at least ± 0.2 mm must be reached. A calibration wand, which is a rigid bar with markers, can be used in the calibration procedure. The camera locations must be selected as shown in Figure 1 in order to maximise the spatial volume of recording and allow for good detection of the reflective markers at each frame of the video. Each video camera must be placed at 90°±5° to the plane of elbow joint movement that is of interest to allow for the examiner to have an optimal view of the most critical points of the delivery swing. As proposed by ECB, (2010) ball release should be identified as the frame at which the upper arm is horizontal and parallel to the ground and ball release is to be identified as the first frame that the ball and hand are not in contact (Figure 2).

The video cameras must be synchronised with the optical motion tracking system so for instance when the optical motion tracking system operates at 200 Hz and the video cameras at 100 Hz this means that every frame of video recording corresponds to two frames of motion tracking.

**Figure 2: Identifying (a) upper arm horizontal and (b) ball release for one bowler**

### 2.2 Marker Placement

Six markers must be placed directly on the thorax; on the left and right acromion processes, the 7th cervical vertebra, the 8th thoracic vertebra, the xiphoid process and the suprasternal notch. In the bowling arm two markers must be placed on the lateral and medial aspects of the wrist and one marker on the bowler’s hand as shown in Table 1 and Figure 3. Two clusters, with a minimum of three markers each, mounted on lightweight frames need to be attached to the upper arm and forearm of the bowling arm.
The upper arm cluster must be positioned a few centimetres above the olecranon process and the forearm cluster centrally onto the forearm.

Table 1: Anatomical landmarks used in elbow tracking

<table>
<thead>
<tr>
<th>Segments</th>
<th>Markers</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>IJ: Suprasternal Notch,</td>
</tr>
<tr>
<td></td>
<td>C7: Spinal Process of the 7th cervical vertebra</td>
</tr>
<tr>
<td></td>
<td>T8: Spinal Process of the 8th thoracic vertebra</td>
</tr>
<tr>
<td></td>
<td>XP: Xiphoid Process</td>
</tr>
<tr>
<td></td>
<td>RA: Right Acromion</td>
</tr>
<tr>
<td></td>
<td>LA: Left Acromion</td>
</tr>
<tr>
<td>Upper Arm</td>
<td>P1: Upper arm Cluster Marker 1</td>
</tr>
<tr>
<td></td>
<td>P2: Upper arm Cluster Marker 2</td>
</tr>
<tr>
<td></td>
<td>P3: Upper arm Cluster Marker 3</td>
</tr>
<tr>
<td>Forearm</td>
<td>FA1: Forearm Cluster Marker 1</td>
</tr>
<tr>
<td></td>
<td>FA2: Forearm Cluster Marker 2</td>
</tr>
<tr>
<td></td>
<td>FA3: Forearm Cluster Marker 3</td>
</tr>
<tr>
<td></td>
<td>US: Medial Aspect of the Wrist (the most caudal-medial point on the ulnar styloid)</td>
</tr>
<tr>
<td></td>
<td>RS: Lateral Aspect of the Wrist (the most caudal-lateral point on the radial styloid)</td>
</tr>
<tr>
<td>Hand</td>
<td>HA1: Hand</td>
</tr>
</tbody>
</table>
Figure 3: Showing the subject marker set-up. The markers in blue are landmarks recommended by the ISB (International Society of Biomechanics (Wu et al. 2005). The markers in yellow are additional markers; used to help the identification of anatomical landmarks during dynamic trials. The markers in pink are mounted in light weight frames; used to replace anatomical markers during dynamic trials. Refer to Table 1 for descriptions of the landmarks.

2.3 Anthropometric Assessment (Both the bowling & non-bowling arms)

As proposed by the ICC and ECB anthropometric measurements need to be taken into account prior to data capture. These measurements are important to aid the examiner to discriminate between bowlers who passively extend their joint due to anatomical factors and those who actively extend aiming at increasing the ball’s speed through release.

Physical Measurements: Height, weight and assessment of the range of motion of the shoulder, elbow and wrist joints. The RoM for each joint should be measured actively according to the ISB recommendations (Wu et al. 2005).

Carrying angle: The carrying angle of the elbow of both arms needs to be estimated as the angle formed by the longitudinal axes of the humerus and the forearm with the elbow fully extended and the forearm fully supinated (Figure 4a). The long axis of the upper arm is to be defined as the axis perpendicular to the plane formed by the elbow flexion axis and the line joining the elbow and glenohumeral joint centres and the
forearm long axis is to be defined as the axis pointing from the ulnar styloid process towards the radial head.

Hyperextension Angle is defined as the angle formed by the longitudinal axes of the upper arm and forearm, in the sagittal plane (Figure 4b). In order for the hyperextension angle to be assessed two measurements of a pure flexion-extension movement are required; a passive measurement, were no force is applied and a second measurement where the bowler is required to extend his arm back as far as possible with ‘reasonable’ force applied (ECB, 2010). This can be achieved by asking the bowler to lean on each arm, applying in that sense a force of about one body weight to their joints (Figure 4c).

History of any previous injuries: Any history of pain or any previous injuries especially at the wrist, elbow and shoulder joints should be determined prior to testing (ECB, 2010).

2.4 Experimental Protocol

Each bowler needs to be asked to warm up on their own prior to data capture.

Initially, a static trial must be captured of the marker setup with the subject standing and the arms hanging next to the body with the palms facing forwards.
Two more static trials must be obtained with the elbow flexed to 90° to calibrate the lateral (LE) and medial (ME) epicondyles. The position of the epicondyles should be reported with reference to a triad of markers placed on the upper arm.

A dynamic trial is to be captured in order to estimate the glenohumeral centre of rotation (GH). A functional method is proposed to be used such as the one proposed by Gamage and Lasenby (2002) which has been reported to perform better than other sphere-fitting functional methods under the same testing environment.

A dynamic trial must be captured in order to estimate the IHAs of elbow flexion-extension during five repeated measures of a single joint movement of pure flexion. In this trial the subject is instructed to actively flex and extend their elbow for five times, keeping a constant pronation and the humerus alongside the body.

A dynamic trial of a pure pronation-supination movement must be captured in order to estimate the instantaneous helical axes of the forearm helical axis. During this trial each subject is instructed to fully pronate and supinate their forearm, keeping the elbow flexed at 90° and the humerus once again alongside the body.

Each bowler should complete a total of 22 deliveries, from which the first 4 should not be included in the analysis to allow the bowlers to reach match pace. From the remaining deliveries, fast and medium paced bowlers should deliver six ‘normal’ – good length deliveries, six ‘yorkers’ and six ‘bouncers’ and six deliveries of each spin variation for the leg or off-spin bowlers must also be recorded (ECB, 2010). Standard video cameras should be used to assess the position the ball lands, along with the amount of bounce and turn created with each delivery. If a hawk-eye system (Hawk-Eye Innovations Ltd, Winchester, Hampshire) is available it can also be used to determine the bounce and speed of the ball and the length of each delivery. This video should also be presented in the report (ECB, 2010).
3. Data analysis

3.1 Initial Trials

3.1.1 Digitisation of the elbow epicondyles

The position of the elbow epicondyles is defined relative to the position of the technical coordinate system of a pointer’s triad in ‘static’ trials before data collection. The tip of the calibration wand is placed onto the anatomical landmark of each epicondyle with the elbow flexed at 90° and a coordinate frame is defined using three markers on the wand as follows:

\[
X_w = \frac{(A - C)}{||A - C||} = \frac{\vec{AC}}{||AC||} \tag{11}
\]

\[
Y_w = \frac{(B - C)}{||B - C||} = \frac{\vec{BC}}{||BC||} \tag{12}
\]

\[
Z_w = \frac{X_w \times Y_w}{||X_w \times Y_w||} \tag{13}
\]

The wand’s technical coordinate frame is then used to define the position of each epicondyle based on the known distances between the marker C on the wand and the landmark position. The global positions of the lateral (LE) and medial (ME) epicondyles were calculated using the following equations:

\[
LE_L = \overline{C} - a \cdot X_w - b \cdot Z_w \tag{14}
\]

\[
ME_L = \overline{C} - a \cdot X_w - b \cdot Z_w \tag{15}
\]
Figure 5: Calibrating the position of the lateral epicondyle (LE). A coordinate frame of the wand is defined using markers C, A and B. Vectors a and b show the distance of the tip of the wand from marker C.

During data capture of the bowling action the position of each epicondyle will be reconstructed with respect to a local technical frame on the upper arm

\[
Y_{hp} = \frac{(P2 - P1)}{\left\| (P2 - P1) \right\|}
\]

\[
X_{hp} = \frac{(P2 - P1) \times (P3 - P1)}{\left\| (P2 - P1) \times (P3 - P1) \right\|}
\]

\[
Z_{hp} = \frac{X_{hp} \times Y_{hp}}{\left\| X_{hp} \times Y_{hp} \right\|}
\]

The position of each epicondyle is transformed from the global frame to the technical frame of Cluster P using the following equation:

\[
\overline{LE}_p = \overline{LE}_l \cdot T_p
\]

\[
\overline{ME}_p = \overline{ME}_l \cdot T_p
\]

Where \(T_p\) is the transformation matrix from the global coordinate frame to the technical frame of Cluster P:
In subsequent dynamic trials the LE and ME coordinates will be defined relative to the local technical frame of Cluster P.

### 3.1.2 The Glenohumeral Joint centre

The glenohumeral joint centre is estimated by capturing the motion of the markers on the humerus, with the elbow flexed at 90°, as it explores most of its possible range of motion (RoM) in all three rotational degrees of freedom in a random manner, in relation to three markers on a scapula locator. The scapula locator used to measure the position of the scapula is a tripod device with three pins which can be adjusted so that it can be placed onto three easily palpable bony landmarks on the scapula: the acromial angle, root of the scapular spine and inferior angle (Figure 6a).

Gamage and Lasenby proposed a functional method for estimating the Centre of Rotation (CoR) of any human joint. In this case this idea can be employed in order to define the CoR of the glenohumeral joint assuming that a set of vectors on a body rotates around a time varying axis of rotation with the fixed CoR while, the ‘tips’ of the vectors lie on co-centric spheres (Figure 6b).

The least squares cost function assuming there are \( P \) markers and \( N \), frames is:

\[
C = \sum_{p=1}^{P} \sum_{k=1}^{N} \left[ (V_k^p - m^2) - (r^p)^2 \right]^2
\]

(22)

Where \( V_k^p \) represents the \( p^{th} \) vector in the \( k^{th} \) time instance, \( m \) represents the centre of rotation and \( r \) is the radius of the sphere.
Figure 6: (a) Locator applied on the bowler to locate his scapula during movement and (b) definitions of the vectors $v$, $u$ and $m_o$. The marker is moving along a trajectory on the surface of a sphere with centre at $m_o$. The $(X, Y, Z)$ system is fixed representing the unit base vectors of the scapular coordinate frame (Halvorsen, 2003)

In order to estimate $r^p$ and $m$ that minimise the cost function $C$ we must differentiate with respect to $r^p$ so:

$$ r^p = \sqrt{\frac{1}{N} \sum_{k=1}^{N} (V^p_k - m)^2 } $$  \hspace{1cm} (23) $$

And differentiating with respect to $m$ we arrive at the following:

$$ \sum_{p=1}^{P} \sum_{k=1}^{N} \left[ (V^p_k - m^2) \{ (V^p_k - m^2) - (r^p)^2 \} \right] = 0 $$  \hspace{1cm} (24) $$

Substituting Eq.(13) into (14):

$$ \sum_{p=1}^{P} \left[ (\bar{V}^p)^2 - \bar{V}^p \cdot \overline{(V^p)^2} \right] = 2 \sum_{p=1}^{P} \left[ \frac{1}{N} \sum_{k=1}^{N} V^p_k (V^p_k \cdot m) - \bar{V}^p (m \cdot \bar{V}^p) \right] $$  \hspace{1cm} (25) $$

Where:

$$ \overline{V}^p = \frac{1}{N} \sum_{k=1}^{N} V^p_k $$  \hspace{1cm} (26) $$
\[
(V^p)^2 = \frac{1}{N} \sum_{k=1}^{N} (V^p_k)^T V^p_k 
\]

\[
(V^p)^3 = \frac{1}{N} \sum_{k=1}^{N} V^p_k \cdot (V^p_k)^T V^p_k 
\]

Using geometric algebra, Eq. 15 can be expressed into the following simplified equation (Eq. 19) which can then be used to estimate the CoR \( m \).

\[ A \cdot m = b \] (29)

Where \( A \) is a \((3 \times 3)\) matrix given by:

\[
A = 2 \sum_{p=1}^{P} \left\{ \frac{1}{N} \sum_{k=1}^{N} V^p_k \cdot (V^p_k)^T \right\} \left( V^p - (V^p)^T \right) 
\] (30)

\[
b = \sum_{p=1}^{P} \left[ (V^p)^3 - (V^p) \cdot (V^p)^2 \right] 
\] (31)

### 3.1.3 Elbow flexion and forearm rotation mean helical axes

The instantaneous helical axes for each movement are calculated based on algorithms according to Woltring (1980) and the optimal elbow flexion and forearm rotation mean helical axes are derived according to Stokdijk et al. (2000).

The global positions of the bony landmarks in every arm position are used for the construction of a local coordinate system (LCS) on the humerus following the ISB recommendations on the definitions of joint coordinate systems (Wu et al. 2005).

\[
Y_{hi} = \frac{(GH - EC)}{\|GH - EC\|} 
\] (32)

\[
X_{hi} = \frac{(GH - EC) \times (LE - ME)}{\|(GH - EC) \times (LE - ME)\|} 
\] (33)

\[
Z_{hi} = \frac{Y_{hi} \times X_{hi}}{\|Y_{hi} \times X_{hi}\|} 
\] (34)

The LCS of the humerus and the forearm cluster are used to determine the rotations of the forearm with respect to the upper arm.
\[ \overline{FA}_L = \overline{FA}_C \cdot T_H \]  

(35)

Where \( T_H \) is the transformation matrix from the global coordinate frame to the humeral coordinate frame

\[
T_H = \begin{bmatrix}
X_H \\
Y_H \\
Z_H
\end{bmatrix} = \begin{bmatrix}
x_x & y_x & z_x \\
x_y & y_y & z_y \\
x_z & y_z & z_z
\end{bmatrix}
\]

(36)

The linear velocity and linear acceleration of the three markers (FA1, FA2 and FA3) of the forearm cluster are calculated using numerical differentiation (Woltring, 1991).

From the positions and orientations of the forearm cluster markers to the humeral coordinate frame instantaneous helical axes (IHAs) are calculated. Each IHA was calculated in least squares sense from landmark motion and described by a position vector (P) and a unit direction vector (n).

\[
\overline{P} = \begin{bmatrix}
P_x \\
P_y \\
P_z
\end{bmatrix}
\]

(37)

\[
\overline{n} = \begin{bmatrix}
n_x \\
n_y \\
n_z
\end{bmatrix}
\]

(38)

The angular acceleration is calculated by deriving the angular velocity vector \( \omega \). The position (P) and unit direction (n) vectors are calculated based on algorithms according to Woltring (1980).

\[
\omega = \sqrt{\overline{\omega}^T \cdot \overline{\omega}}
\]

(39)

\[
\overline{n} = \frac{\overline{\omega}}{\omega}
\]

(40)

\[
v = \overline{\dot{p}}^T \cdot \overline{n}
\]

(41)

\[
\overline{P} = p + \frac{\overline{\omega}}{\omega^2} \cdot \overline{p}^T
\]

(42)

Where: \( v \) is the translational velocity, \( p \) the position vector of each landmark in the humeral coordinate frame and \( \dot{p} \) is its derivative.

Both the mean flexion-extension F/E and pronation-supination P/S axes are calculated employing the same methodology for the two single joint movements of pure flexion and pronation respectively.
The position and orientation vectors of the IHAs for the dynamic trial were used to compute \( P \) and \( n \) of the optimal IHA for each subject.

\[
P_{\text{optimal}} = Q^{-1} \cdot \frac{1}{N} \sum_{i=1}^{N} Q_i \cdot P_i
\]  

(43)

\[
Q_i = I - n_i \cdot n_i^T
\]

(44)

\[
Q = \frac{1}{N} \sum_{i=1}^{N} Q_i
\]

(45)

As a low angular velocity (under 0.25 rad/sec) can lead to inaccurate calculation of angular acceleration (Stokdijk et al. 1999) and cause outliers, it is proposed that only the middle 60° of every controlled movement are taken into account (Stokdijk et al. 2000) when calculating the optimal position vectors \( P_{F/E} \) and \( P_{P/S} \) for the flexion-extension and pronation-supination axes respectively. However, different methods can also be applied to minimise measurement noise. Indicatively a method proposed by Chin (2009) can also be used whereby the instantaneous helical axes for each movement are calculated at every time point that there is a displacement of greater than 25° in the following time point and a mean helical axis is defined as proposed by Besier et al. (2003) with IHA’s with orientation components greater than 2 standard deviations removed and the mean axis recalculated to obtain a finite helical axis.

Analogous to the calculation of \( P_{\text{optimal}} \) an optimal unit direction vector \( (n_{\text{optimal}}) \) is calculated. During the bowling action the elbow F/E functional axis is expressed in the LCS of the humerus and the P/S axis is then expressed in the LCS of the forearm cluster.

### 3.1.4 Bowling

Descriptive analysis during the match footage should compare the actions of the bowlers. These comparisons should be clearly presented in the report (ECB, 2010). Only bowls that match the performance of the bowler in terms of leg, arm and torso positioning, arm velocity, ball speed, bounce and delivery length should be included in the analysis. An experienced umpire should work along with the examiners to aid in the identification of such deliveries.
Analysis of the bowling action must consist of calculating the extension of the elbow from the point that the upper arm reached the level of shoulder until ball release. This definition is in accordance with the relevant law in cricket that is associated with ‘no-ball’.

Segmental coordinates are introduced for the upper arm and forearm as follows:

**Upper arm** $(X_u, Y_u, Z_u)$

\[
Z_u = \frac{n_{F/E}}{\|n_{F/E}\|} \quad (46)
\]

\[
X_u = \frac{Z_u \times (P_{EE} - GH)}{\|Z_u \times (P_{EE} - GH)\|} \quad (47)
\]

\[
Y_u = \frac{Z_u \times X_u}{\|Z_u \times X_u\|} \quad (48)
\]

**Forearm** $(X_f, Y_f, Z_f)$

\[
Y_f = \frac{n_{P/S}}{\|n_{P/S}\|} \quad (49)
\]

\[
X_f = \frac{Y_f \times (US - RS)}{\|Y_f \times (US - RS)\|} \quad (50)
\]

\[
Z_f = \frac{Y_f \times X_f}{\|Y_f \times X_f\|} \quad (51)
\]

Elbow rotations are calculated using Euler angles with a $z - x' - y''$ Cardan sequence, where rotations about $z$, $y$, and $x$ are the flexion-extension (flexion positive), pronation-supination (pronation positive) and adduction-abduction (adduction positive).

### 4. Presentation of Results

A summary of the results of the anthropometric assessment and each delivery must be presented – graphical or table format (Appendix A).

The data for each bowler must be averaged over the 6 trials of each type of delivery and presented as the mean elbow angle and standard deviation (ICC, 2005).
The standard deviation is a measure of the spread of the distribution about the mean and was used to measure the variability between the different digitisation methods.

\[
\sigma = \sqrt{\frac{1}{N-1} \sum_{i=1}^{N} (x_i - \bar{x})^2}
\]  

(52)

Where \(\sigma\) is the standard deviation, \(N\) is the size of the sample and \(\bar{x}\) is the mean value given by the following equation:

\[
\bar{x} = \frac{1}{N} \sum_{i=1}^{N} x_i
\]  

(53)

Elbow flexion-extension angles, shoulder, elbow, wrist and ball release speeds for each delivery as well as average elbow extension angles and standard deviations of the different types of delivery (e.g. Bouncer, Normal Length) must be displayed in a table format.

The six trials for each type of delivery must be plotted on the same graph along with their average and standard deviations in the form of error bars. To free the data from the effects caused by variations in speed across different trials within the same subject as well as between different subjects the data can be normalised to 100% of the bowling action defined from 20 ms prior to upper arm horizontal to 20 ms after ball release.

Apart from the flexion-extension graphs, the abduction-adduction curve and the hyperextension curve must also be presented separately in graphical form (ECB, 2010). In all the graphic representations the key positions of upper arm horizontal and ball release should be clearly marked in order to aid in the presentation of the results (ECB, 2010).
### Part A: Anthropometric Assessment

**Table 2: Physical Measurements for each bowler**

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Type</th>
<th>Height (cm)</th>
<th>Weight (Kg)</th>
<th>RoM (°)</th>
<th>Shoulder Joint</th>
<th>Elbow Joint</th>
<th>Wrist Joint</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Flexion/Extension</td>
<td>Abduction/Adduction</td>
<td>Internal/External Rotation</td>
</tr>
</tbody>
</table>
Table 3: Assessment of the carrying angle and the hyperextension angle

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Injuries</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carrying angle (°)</td>
<td>Bowling arm</td>
</tr>
<tr>
<td>Hyperextension angle (°)</td>
<td>No force applied</td>
</tr>
<tr>
<td></td>
<td>External force applied</td>
</tr>
</tbody>
</table>
Table 4: Elbow extension angles and shoulder, elbow, wrist and ball release speeds for each bowler and each delivery type

<table>
<thead>
<tr>
<th>Bowler</th>
<th>Delivery Type</th>
<th>Extension angle (°)</th>
<th>Speed (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Bowler 1</td>
<td></td>
<td>Shoulder</td>
</tr>
<tr>
<td></td>
<td>Bowler 2</td>
<td></td>
<td>Wrist</td>
</tr>
<tr>
<td></td>
<td>Bowler 3</td>
<td></td>
<td>Elbow</td>
</tr>
<tr>
<td></td>
<td>Bowler 4</td>
<td></td>
<td>Ball release</td>
</tr>
<tr>
<td></td>
<td>Bowler 5</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Bowler 6</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Mean (SD)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>