SEQUENCING OF SEGMENT KINETIC ENERGY IN THE GOLF SWING

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Abstract

The proximal-to-distal sequence has been associated with mechanical and muscular rewards which enable high speed to be produced at the distal end of a linked system. Although the proximal-to-distal sequence has frequently been examined using analyses of segment angular velocity, analyses of segment kinetic energy can provide the most appropriate means of exploring sequential movements. However, due to methodological complexities few studies have adopted this technique. Therefore, the aim of this thesis was to determine if a sequence of segment KE was evident in the golf swing. To enable segment kinetic energy to be calculated body segment inertial parameters were estimated for 17 rigid bodies using a 30 shape geometric model. Kinematic data were then collected using a sixteen-channel Polhemus Liberty electromagnetic tracking system sampling at 240 Hz and twelve, six-degrees-of-freedom electromagnetic sensors. Using this data, total kinetic energy, calculated as the sum of the translational, local rotational and remote rotational kinetic energies was then determined for four grouped segments (Lower Body, Upper Body, Arms and Club) in the downswing phase of the golf swing. The thesis then established that the data collection technique was capable of producing reliable measures of segment KE in the golf swing. Therefore, three further studies were performed which examined the effect of club type, playing standard and weight transfer style on the sequencing of segment KE in the golf swing. In all studies, peak total segment kinetic energy increased sequentially from the proximal to the distal segment. However, the timing of peak total segment kinetic energy did not follow a proximal-to-distal sequence. Instead, peak total segment kinetic energy for the Lower Body, Upper Body and Arms occurred simultaneously at approximately 74% of the downswing, significantly earlier in the downswing than peak total Club kinetic energy which occurred just before ball contact. For skilled golfers, the results suggested that peak translational and rotational kinetic energy increased sequentially from the Upper Body to the Club. Furthermore, when the driver was used, larger magnitudes of peak translational Arms and local rotational Upper Body kinetic energies were produced and peak translational Lower Body and Upper Body KE occurred significantly later in the downswing. The results also identified that highly skilled golfers generated significantly larger magnitudes of total Arms and Club kinetic energies than less skilled golfers. Furthermore, a sequence of translational KE from the Upper Body to the Club was only produced by highly skilled players. Finally, the results indicated peak translational and local rotational Lower Body, peak translational Upper Body and peak remote rotational Arms KE occurred significantly later for the Front Foot players. In conclusion, club type, playing standard and weight transfer style had little effect on the magnitude and timing of peak total segment kinetic energy. However, significant effects were identified on measures of translational and rotational components of peak segment KE. These findings should encourage golf coaches and researchers to use analyses of segment kinetic energy rather than analyses of segment angular velocity as they are sensitive to subtle changes in technique and consider the 3D translation and rotation of body segments. The findings also highlight the importance for future analyses to consider the different weight transfer styles that exist within the golf swing.

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Preface

The findings presented in this thesis have been peer reviewed as follows:

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List of Abbreviations

3D	Three-dimensional
ANOVA	Analysis of variance
BC	Ball contact
BSIP	Body segment inertial parameters
COM	Centre of mass
CP	Centre of pressure
ED	Early downswing
HSD	Honest significant difference
ICC	Intraclass correlation coefficient
ISA	Instantaneous screw axis
KE	Kinetic energy
LB	Late Backswing
LOA	Limits of agreement
M	Mean
MB	Mid backswing
MD	Minimal difference
MF	Mid follow-through
MOI	Moment of inertia
MT	Downswing movement time
RMS	Root mean square error
SD	Standard deviation
SEM	Standard error of measurement
SSC	Stretch shortening cycle
TA	Take away
TOB	Top of the backswing

Glossary of Terms

Delayed wrist release: Maintaining the wrist-cock (radial deviation) angle late into the downswing.

Golf swing hub: The centre of the upper trunk.

Kinetic link principle: High speed at the distal end of a linked system is achieved due to the sequential acceleration, deceleration and subsequent interactions between segments.

Optimal coordination of partial momenta: Optimal distal end velocity of a linked system is achieved when the angular velocities of the segments in the system peak simultaneously or in an order specific to an individual, not in a proximal-to-distal sequence.

Proximal-to-distal sequence: Movements characterised by sequential motions of the segments comprising the linked system, progressing from the most proximal to the most distal segment.

Rigid body: A section of the human body defined by a proximal and distal joint.

Segment: A group of rigid bodies.

Stretch shortening cycle: An active stretch (eccentric contraction) of a muscle followed by an immediate shortening (concentric contraction) of that same muscle which produces enhanced muscular force when compared with an isolated concentric.

Summation of speed principle: A linked system will only achieve maximum end point velocity if each segment starts its motion at the instant of greatest speed of the preceding segment and reaches a maximum speed greater than that of its predecessor.

Weight transfer: The movement of weight between the feet typically in the direction of the shot.

Wrist-cock angle: The radial deviation angle.

X-factor: A descriptive coaching term used to consider the separation between the axial rotation of the shoulders and the axial rotation of the pelvis.

X-factor stretch: An increase in the magnitude of upper torso and pelvis axial rotation separation that occurs at the start of the downswing.

1. Chapter I - Introduction

Golf is a unique sport that can be played and appreciated regardless of age, gender, socioeconomic group or athletic ability. The use of handicapping also allows player's of different abilities to compete against each other on a level playing field. Golf has become an increasingly popular sport and it was recently estimated that golf is practiced by 10-20% of the adult population in many countries with 67 million participant's worldwide (Zheng *et al.*, 2008).

The substantial growth in the popularity of golf, the complexity of the golf swing and an increase in prize money in the late 1980's has stimulated considerable golf science research (Farrally *et al.*, 2003). Amongst the scientific disciplines which have been applied to explore the human factors associated with the golf swing is biomechanics. Biomechanical analyses of the golf swing have been performed to improve golf performance and reduce the severity and risk of injury (Hume, Keogh and Reid, 2005). Although a vast amount of biomechanical golf swing analyses have been undertaken, the complex movement patterns associated with the golf swing are still not completely understood (Vena *et al.* 2011a).

One aspect of the golf swing which has received considerable attention in biomechanical research is the sequencing of motions that comprise the downswing. Commonly referred to as the timing of the swing, the sequencing of body segment motion in the golf swing has become an important theme in golf instruction (Cheetham *et al.*, 2008; Vena *et al.*, 2011a). It has been suggested that a proximal-to-distal sequence (Putnam, 1993) can ensure that clubhead velocity increases throughout the downswing and reaches a maximum at impact (Bunn, 1972; Cheetham *et al.*, 2008; Vena *et al.*, 2011b). To achieve maximum clubhead velocity, this concept suggests that the motion of the large proximal segments must be followed by faster motion of the smaller distal segments (Bunn, 1972). Each segment should also start its motion at the instant of greatest speed of the preceding segment and reach a speed greater than that of its predecessor (Putnam, 1993).

The proximal-to-distal sequence in the golf swing has been analysed by many authors using a multitude of different approaches and techniques (Burden, Grimshaw and Wallace, 1998; Cochran and Stobbs, 1968; Teu *et al.*, 2006; Kenny *et al.*, 2008). However, it has been suggested that analyses of segment kinetic energy (KE) provide the most appropriate quantification of the proximal-to-distal sequence (Anderson, Wright and Stefanyshyn, 2006; Ferdinands, 2011). This type of analysis has many advantages over conventional kinematic analyses; it is independent of direction, considers the inertial properties of segments, enables linear and angular components of movement to be considered separately and as a single entity and is much more sensitive to subtle changes in technique (Bechard, 2009; Ferdinands, Kerstig and Marshall, 2012; Slawinski *et al.*, 2010).

Early research by Budney and Bellow (1982) quantified KE of the club at impact but made no reference to the KE of body segments. More recently, Anderson, Wright and Stefanyshyn (2006) and Kenny *et al.* (2008) analysed the sequencing of segment KE in the golf swing. Both studies reported that peak magnitudes of segment KE increased sequentially from proximal to distal segments whilst the timing of peak segment KE did not follow the same sequential pattern. Instead, the three human based segments (Lower Body, Upper Body and Arms) peaked simultaneously and the club segment peaked significantly later in the swing, just before impact.

Although these studies examined the sequencing of segment KE in the golf swing, a comprehensive understanding has yet to be formulated. Anderson, Wright and Stefanyshyn, (2006) and Kenny *et al.* (2008) only considered the sequencing of total segment KE. An examination of the linear and angular components of segment KE would take advantage of the benefits associated with energetic analyses and provide a more detailed description of the complex movement patterns associated with the golf swing. The effects of playing standard and club type on the sequencing of segment KE were also not considered. Both studies examined the sequencing of segment KE for highly skilled golfers and only Kenny *et al.* (2008) considered the effect of club type. However, this analysis of club type only included a driver and a 7 iron and only one player was analysed. Further, more detailed analyses of the sequencing of KE are therefore required to provide a comprehensive understanding of the proximal-to-distal sequence in the golf swing.

1.1. Purpose of the thesis

The purpose of this thesis was to examine the sequencing of body segment movements in the golf swing. More specifically, based on a thorough critique of current literature relevant to the programme of study (Chapter II) the main aim of this thesis was to examine the sequencing of segment KE.

1.2. Structure of the report

To address the overall purpose, this thesis comprises seven further chapters which are structured as follows:

Chapter II provides a thorough, critical review of the literature relevant to the programme of research. The literature review considers the underlying principles of the proximal-to-distal sequence, before examining proximal-to-distal sequencing research in striking and throwing movements. Kinematic and kinetic analyses of the proximal-to-distal sequence in the golf swing are then explored, before a critical review of energetic analyses is provided.

Chapter III presents the methodology which is used in all subsequent studies to measure the sequencing of segment KE. This chapter describes the geometric modelling technique used to collect accurate and reliable body segment inertial parameters, before outlining the data collection procedure. Furthermore, this chapter explains the calculation of segment KE and identifies the data analysis protocol.

Chapter IV examines the reliability of the measurement of the magnitude and timing of peak segment KE in the golf swing. This assessment of test-retest reliability also provides measures with which future studies can determine the meaningfulness of differences in segment KE.

Chapter V examines whether a proximal-to-distal sequence of segment KE is evident in the golf swings of highly skilled players using three different clubs. Chapter V includes an analysis of the linear and angular components of segment KE as well as an analysis of total segment KE.

Chapter VI examines the effect of playing standard on the sequencing of segment KE in the golf swing. This chapter examines the segment KE sequencing profiles of three categories of golfers, classified according to their current playing standard (CONGU handicap) using three different clubs.

Chapter VII assesses the effect of weight transfer style on the sequencing of segment KE in the golf swing. By considering the centre of pressure in the direction of the shotf shot, two groups of golfers, a front foot group and a reverse group were established. Chapter VII then determines the impact of this swing style classification by examining whether different sequences of segment KE are exhibited by the two groups.

Chapter VIII summarises and discusses the findings reported in the thesis. This chapter identifies the practical implications the thesis findings will have in biomechanical golf swing research and golf coaching. Chapter VIII also highlights the limitations of this thesis and suggests future directions for the programme of research to progress. Finally, this chapter provides an overall conclusion for the programme of research.

2. Chapter II - Literature Review

2.1. Introduction

This literature review will introduce segmental sequencing in open kinetic chain activities. Then, the review will focus on segmental sequencing in the golf swing. Initially, underlying principles and descriptions of the proximal-to-distal sequence and traditional approaches to characterising this phenomenon will be outlined. Subsequently, these approaches will be reviewed and the effect of a proximal-to-distal sequence on performance will be examined.

The review will then introduce kinematic and kinetic approaches which have examined the proximal-to-distal sequence in the golf swing. The appropriateness of these methods will be discussed before the most pertinent findings are related to performance. Finally, the review will consider the analysis of segment KE sequencing in the golf swing. Relevant research will be summarised, before a rationale for the programme of research is provided.

2.2. Biomechanics of the golf swing

Golf is a fascinating sport because there are many different ways for players to perform good, effective swings (Adlington, 1996). While its purpose is simple, the golf swing can be extremely difficult to master. The golf swing is regarded as one of the most difficult sporting motions to execute given the challenging requirement to swing a relatively long club at a relatively small ball with high velocities (Lindsay, Mantrop, and Vandervoort, 2008). The golf swing also requires extremely complex temporal and spatial movement considerations (Bradshaw *et al.*, 2009).

Until the Royal Society Golf Group (Cochran and Stobbs, 1968) analysed mechanical properties underlying performance, little theoretical or empirical golf science research had been performed. However, the increase in the popularity of golf and the complexity of the golf swing stimulated considerable scientific research. The increase in prize money in the late 1980s also encouraged more scientific research to be performed (Farrally *et al.*, 2003). Another major step towards increasing scientific golf research was the creation of the World Scientific Congress of Golf in 1990. Since then, this organisation has been responsible for a substantial amount of golf science research (Vena *et al.,* 2011a).

Amongst the scientific disciplines which have been applied to explore golf swing technique is biomechanics. Biomechanics uses mechanical principles to understand movement (Hume, Keogh and Reid, 2005). Golf swing biomechanics therefore applies the principles of mechanics to the structure and function of the golfer (Hume, Keogh and Reid, 2005). Biomechanical analyses of the golf swing have been performed in an attempt to improve golf performance and reduce the severity and risk of injury (Hume, Keogh and Reid, 2005). To investigate the factors associated with performance and injury, biomechanists have used various approaches including the *in vivo* measurements and calculation of kinematics and kinetics (Chu, Sell, and Lephart, 2010). Kinematic analyses provide numerical data that describe movements, whilst kinetic analyses provide numerical data on the internal and external forces to examine the causes of movements (Hume, Keogh and Reid, 2005).

Although extensive biomechanical analyses have been performed, the complex movement patterns associated with the golf swing are still not completely understood. It has been suggested that biomechanical research has yet to make significant advances on the landmark work of Cochran and Stobbs (1968) (Farrally *et al.,* 2003). However, it has also been suggested that it is from the application of sound biomechanics that performance standards in golf are most likely to be increased (Farrally *et al.*, 2003; Hellstrom, 2009).

2.2.1. Improving golf swing performance using biomechanics

A primary aim of biomechanical research and analysis is to improve golf performance which is ultimately defined by lower scores (Hellstrom, 2009; Hume, Keogh and Reid, 2005). However, it is not always possible to perform biomechanical golf swing analyses on the golf course. In fact, research is habitually performed in a laboratory environment where it is not possible to directly measure golf performance. Therefore, in biomechanical research, golf performance has been classified using a variety of techniques including; clubhead velocity at impact (Ball and Best, 2007a; Myers *et al.*, 2008; Teu, *et al.*, 2006) post-impact ball velocity (Chu, Sell and Lephart, 2010; Myers *et al.*, 2008) and golfers handicap (Bradshaw *et al.*, 2009; Cheetham *et al.* 2008). Since one of the main determinants of ball speed and ultimately shot distance is the speed of the club immediately before impact the majority of golf swing research has classified performance in this way (Tinmark *et al.*, 2010).

Elite golf performance requires players to hit a variety of successful shots using two principal movements: swing and putting. The majority of biomechanical analyses have focussed on the complexities associated with the swing with limited research conducted on putting. This appears to be logical as, according to the Professional Golf Association (PGA) Tour statistics, the putting stroke only accounted for 35% of all strokes made during tournament rounds of golf in 2008 (PGA Tour, 2009). The difference in emphasis could also be explained by the fact that the swing is performed with 13 clubs compared with putting which is performed with just one (Farrally *et al.*, 2003).

To understand the factors that underpin elite golf performance, kinematic (Burden, Grimshaw and Wallace, 1998; Chu, Sell and Lephart, 2010; Lindsay, Horton, and Paley, 2002; Myers *et al.*, 2008; Tinmark *et al.*, 2010) and kinetic (Ball and Best, 2007a; Betzler *et al.*, 2006; MacKenzie and Sprigings, 2009; Miura, 2001) analyses have been performed. These analyses have associated variables such as; wrist flexion-extension angle (Zheng *et al.*, 2008; Chen, Inoue, and Shibara, 2007), left-hand grip strength (Brown *et al.*, 2011), delayed club release (Pickering and Vickers, 1999), hip rotation angle (Egret *et al.* 2006), shoulder motion (Mitchell *et al.*, 2003), trunk lateral bending (Chu, Sell and Lephart, 2010), lateral and upward shift of the hub (Miura, 2001), muscle torque coordination (MacKenzie and Sprigings, 2009) and weight transfer style (Ball and Best, 2007a) with elite golf performance. Furthermore, the x-Factor – a descriptive coaching term used to consider the separation between the rotation of the shoulders and the rotation of the pelvis – has frequently been examined in peer reviewed biomechanical studies. Numerous analyses have associated increased upper torso and pelvis separation at the top of the backswing with elite golf performance (Burden, Grimshaw, and Wallace, 2001; Cheetham *et al.*, 2001; Myers *et al.*, 2008). It has been reported that professional golfers (Cheetham *et al.*, 200; Zheng *et al.*, 2008) and golfers who generate high clubhead velocities (Chu, Sell and Lephart, 2010; Healy *et al.*, 2011; Myers *et al.*, 2008) produce larger upper torso and pelvis separation compared to amateur golfers and golfers who generate low clubhead velocities.

The sequential rotation of the pelvis and thorax during the downswing has also been the subject of a considerable amount of research (Burden, Grimshaw and Wallace 2001; Cheetham *et al.*, 2001; Horan *et al.*, 2010). By initiating the downswing with the rotation of the hips towards the target, it has been suggested that skilled golfers are able to increase the magnitude of upper torso and pelvis separation (x-factor stretch) (Fletcher and Hartwell, 2004; Horan *et al.*, 2010: McTeigue *et al.*, 1994). Furthermore, the sequential rotation of the hips and shoulders has enabled professional golfers to rotate faster than amateur golfers and subsequently allowed them to generate increased clubhead velocities (Cheetham *et al.*, 2001). These performance benefits associated with the x-factor stretch have been attributed to the enhanced utilisation of the stretch–shortening cycle (Cheetham *et al.*, 2001). The rapid rotation of the pelvis early in the downswing is believed to increase the eccentric stretching of muscles in the trunk before they concentrically contract subsequently increasing the power produced (Horan *et al.*, 2010).

Another variable which has provided the basis for numerous golf swing analyses is the delayed release of the club (Pickering and Vickers, 1999; Sprigings and Mackenzie, 2002; Sprigings and Neal, 2000). These analyses have suggested that maintaining the wrist-cock (radial deviation) angle late into the downswing increases wrist angular velocity later in the downswing which subsequently increases clubhead speed and ultimately shot distance (Jorgensen, 1970; Sprigings and Mackenzie, 2002). Optimally timed release of the club has been associated with advances of 1.6% (Sprigings and Mackenzie, 2002), 2.5% (Pickering and Vickers, 1999), 2.9% (Jorgensen, 1970) and 9% (Sprigings and Neal, 2000) in clubhead speed at impact. It has also been indicated that less skilled golfers release and accelerate their club earlier than highly skilled players (Robinson, 1994; Zheng *et al.*, 2008).

The proximal-to-distal sequence has also received considerable attention in biomechanical research. Since the landmark work by Cochran and Stobbs (1968), the proximal-to-distal sequence has become an important theme in golf swing instruction. Sequential motions associated with this movement pattern such as the x-factor stretch and delayed wrist release have been acknowledged as a possible means to ensure that clubhead velocity increase throughout the downswing and achieves its maximum at impact (Bunn, 1972; Burden, Grimshaw and Wallace 1998; Cheetham *et al.* 2008; Putnam, 1993; Vena *et al.* 2011a).

2.3. The proximal-to-distal sequence

In many sporting movements, the body can be modelled as an open linked system of rigid segments in which the distal end is able to move freely through space (Putnam, 1993). The theory underpinning the proximal-to-distal sequence states that, to produce high speed at the distal end, movements should be characterised by sequential motions of the segments comprising the system, progressing from the most proximal to the most distal segment (Morehouse & Cooper, 1950).

The proximal-to-distal sequence is often referred to as a whip-like action and can be easily understood when one considers the motion used to crack a whip (Fradet *et al.*, 2004; Hirashima *et al.*, 2008; Sprigings and Mackenzie, 2002). To do this, the handle is rapidly accelerated before being rapidly decelerated. This deceleration of the handle is the key to achieving transfer of speed through the whip. It causes each segment of the system to build on that of the previous segment and ultimately causes increased distal end speed.

2.3.1. Principles of the proximal-to-distal sequence

One of the most influential concepts underlying the theory of the proximal-to-distal sequence is the summation of speed principle proposed by Bunn (1972). This principle states that a linked system will only achieve maximum end point velocity if each segment starts its motion at the instant of greatest speed of the preceding segment and reaches a maximum speed greater than that of its predecessor. Since it was proposed, the summation of speed principle has become the focus of significant research scrutiny (Neal *et al.*, 2007). It has also become an influential concept in the explanation of how both translational and angular velocities can be transferred in a linked system (Anderson, Wright and Stefanyshyn, 2006).

The summation of speed has been considered essential for the generation of high speed in the distal segment of any open linked system (Putnam, 1993). However, it only provides a description of the proximal-to-distal sequence. To explain why the proximal-to-distal sequence is advantageous in open kinetic chain movements Kreighbaum and Barthels (1985) proposed the kinetic link principle. The kinetic link principle suggests that high speed at the distal segment is achieved due to the sequential acceleration, deceleration and subsequent interactions between segments (Putnam, 1993). The basic notion of the kinetic link principle suggests that an initial external torque applied to the proximal segments provides a linked system with momentum (Putnam, 1993; Hirashima *et al.*, 2007). This acts as a foundation for distal segment motion as the deceleration of a proximal segment causes the velocity of the remaining system to increase as it assumes that lost by the proximal segment (Welch *et al.*, 1995). The kinetic link principle suggests that it is beneficial interaction torques that enable momentum to be sequentially transferred through the linked system to the distal segment (Kreighbaum and Barthels, 1985).

Despite the support for the proximal-to-distal sequence provided by the summation of speed and kinetic link principles, Van Gheluwe and Hebbelinck (1985) proposed the principle of optimal coordination of partial momenta. This principle states that optimal distal end velocity of a linked system is reached when the angular velocities of the segments peak simultaneously or in an order specific to an individual, not in a proximal-to-distal sequence. It is also known that the speed of the distal end of any rotating body, pivoted at its proximal end is equal to the length of the body multiplied by the body's angular speed. Therefore, from a purely kinematic perspective, it appears logical that all segments with appropriate positioning should be rotating with maximum speed when maximum distal end speed is desired (Putnam, 1993).

Whilst these principles examine the proximal-to-distal sequence from a mechanical perspective, segments in a linked system are also connected via the musculoskeletal system. The musculoskeletal system is capable of accelerating and decelerating the segments through the application of muscular force (Welch *et al.*, 1995). Consequently, the stretch shortening cycle has also been recognised as an underlying principle of movements that adopt a proximal-to-distal sequence (Bunn, 1972). The stretch shortening cycle (SSC) can be defined as an active stretch (eccentric contraction) of a muscle followed by an immediate shortening (concentric contraction) of that same muscle (Nicol, Avela, and Komi, 2006).

It is believed that, as a consequence of the SSC, enhanced muscular force is produced as a result of the elastic energy generated and stored during the eccentric contraction (Komi, 2000). Compared with an isolated concentric muscle contraction, muscular output and subsequently distal segment speed can be increased by eccentric contractions and pre-stretching of muscles prior to concentric contraction (Finni *et al.*, 2003; Nicol, Avela and Komi, 2006). The effectiveness of the stretch-shortening cycle depends on the magnitude of the stretch imposed during the eccentric phase and on the rate of the stretch (Rodacki, Fowler, and Bennett, 2001). Therefore, it would appear that, optimising both the mechanical and physiological components of a linked system could therefore produce maximum velocity at the distal end (Lees, 2010; Welch *et al.*, 1995).

2.3.2. Proximal-to-distal sequence analysis techniques

The summation of speed principle, kinetic link principle and SSC provide a theoretical underpinning for the existence of the proximal-to-distal sequence. However, they do not provide evidence that it occurs in open kinetic chain movements. To examine the sequencing of body segments in such movements, kinematic, kinetic and energetic analysis techniques have all been used. Kinematic analyses have assessed body segment sequencing by examining linear velocities of segment endpoints, joint angular velocities and segment angular velocities (Marshall and Elliott 2000). Kinetic analyses have examined joint forces and joint moments (Hirashima, Kudo and Ohtsuki, 2003; Naito, Fukui, and Maruyama, 2010) whilst energetic analyses have assessed segment KE (Anderson, Wright and Stefanyshyn, 2006), joint power (Nesbit, 2005) and work (Sprigings and Mackenzie, 2002).

Kinematic analyses of body segment sequencing in terms of linear velocities have demonstrated that in open kinetic chain movements the distal ends of segments get progressively faster in a proximal-to-distal sequence (Fradet *et al.*, 2004). Describing movement in these terms has also provided a description of the instantaneous kinematic contributions of individual segments to distal segment speed (Elliott, Marshall, and Noffal, 1995). However, descriptions of segmental sequencing in these terms have limited value as they are unable to examine the sequencing of movements that are predominantly rotational (Putnam, 1993) - such as trunk motion in the golf swing.

Since we typically think of motion as a series of joint rotations, analyses of joint angular velocities can make it easy to visualise movements (Putnam, 1993). Furthermore, using equations outlined by Zajak and Gordon (1989), analyses of joint angular velocities can be used to examine sequential motions as it can lead to an explanation of how the motion at each joint contributes to the final speed of the distal segment (Herring and Chapman, 1992). However, this technique requires the angular velocity of the most proximal segment participating in the sequence, the angular velocities of the joints at the proximal end of the remaining segments and the linear acceleration of the proximal end of the linked system be known (Putnam, 1993).

Kinematic analyses in terms of segment angular velocities also provide clear descriptions of the proximal-to-distal sequence. Although it can be difficult to visualise motion from this data, it may be advantageous to describe motion in these terms (Putnam, 1993). In a similar fashion to the analysis of joint angular velocities, analyses of segment angular velocities can lead to an appreciation of how the motion of each segment ultimately contributes to the final speed of the distal segment in a linked system. In addition, analyses of segment angular velocities allow an appreciation of the causes of segment motion sequences to be determined even if the linear acceleration of the proximal end of the linked system is not known (Putnam, 1993).

Linear and angular kinematic analyses can all provide a description of motion. However to gain a comprehensive understanding of the mechanisms that produce body motion it has been suggested that a kinetic analysis would be more beneficial (Feltner and Dapena, 1989; Herring and Chapman, 1992). Kinetic analyses examine the factors responsible for the production of motion by calculating the joint forces and joint moments exerted on a segment (Nunome, *et al.*, 2006). Joint forces are the resultant of all muscular and non-muscular forces acting at or across a specific joint (Feltner and Dapena, 1989) whilst joint moments are the result of muscle forces, ligament forces and forces due to articular surface contact acting about a specified joint (Challis and Kerwin, 1996).

Joint motion in multi-limb systems not only result from joint forces and joint moments about its corresponding joint axis but also from the net moment originating from other joint rotations (motion-dependent moment) (Naito, Fukui, and Maruyama, 2010). Therefore, the examination of motion-dependent moments can provide further insight into the proximal-to-distal sequence. By re-organising the typical Newtonian equations of motion Putnam (1991) enabled the interaction between two adjacent segments (proximal thigh and distal shank) in two-dimensions to be examined. However, in most sporting movements the motion of any given joint is influenced by three-dimensional (3D) multi-joint motions of the entire body (Feltner and Dapena, 1989). Despite the complexity associated with motion-dependent interactions in 3D, dynamical models have been developed in kicking research which examined the cause-effect relationship between the motion of the standing leg, the trunk and the kicking leg joint motion (Naito, Fukui and Maruyama *et al.*, 2010; Nunome *et al.*, 2006).

Investigating movements using kinetic analyses can provide substantial information on the cause-effect relationship between the motions of numerous segments and provide an insight into the sequencing of motion (Robertson *et al.*, 2005). However, obtaining information using Newton's methods does not provide insight into overall changes in velocity or energy transfer and as such only yield a snapshot image of movement dynamics (Flanagan, 2014; Nesbit, 2005). Therefore, as the fundamental purpose of most open kinetic chain movements is to produce KE which can be transferred to the most distal segment, energetic analyses which examine factors responsible for the production of movement such as work, energy and power can provide further insight into complex sporting movements (Nesbit, 2005). As such, energetic analyses have been used to examine the sequencing of total segment KE during walking and running gait (Segers *et al.*, 2006), the sprint start (Slawinski *et al.*, 2010) and during the rowing stroke (Bechard *et al.*, 2009). Furthermore, energetic analyses have examined the mechanical sources of power that are responsible for the production of large distal segment KE in sporting movements such as the golf swing (Sprigings and Mackenzie, 2002) and during maximal speed cycling (Martin and Brown, 2009).

2.4. Proximal-to-distal sequencing in striking and throwing movements

The speed of the distal end of a linked system immediately before impact or release is one of the main determinants of performance in striking and throwing movements (Putnam, 1993). The proximal-to-distal sequence appears to provide logical mechanical and muscular explanations of how high speed can be developed at the distal segment of a linked system. Therefore, using the analysis techniques defined in the previous section, substantial research has been performed to provide evidence for the proximal-to-distal sequence in open kinetic chain movements.

Evidence from both kinematic and kinetic analyses suggests the proximal-to-distal sequence is essential for the generation of high speed in the distal segment of a linked system (Putnam 1993). Support for the proximal-to-distal sequence has been identified in many striking and throwing movements which attempt to maximise the speed of the most distal segment. A proximal-to-distal sequence has been identified in the swing phase of running and walking (Putnam 1991), baseball pitching (Feltner and Dapena, 1989), baseball batting (Cross, 2005), overarm throwing (Putnam, 1993;

Hirashima *et al.*, 2008), the tennis serve (Elliott *et al.*, 1986; Marshall and Elliott, 2000), javelin throwing (Mero *et al.*, 1994; Whiting, Gregor, and Halushka, 1991) and ball kicking (Dörge *et al.*, 2002; Nunome *et al.*, 2006; Khorasani *et al.*, 2009).

2.4.1. Kinematic analyses of striking and throwing movements

A kinematic chain, characterised by increased maximum linear velocity of segments in a proximal-to-distal sequence was originally identified in the tennis serve (Elliott, Marsh, and Blanksby, 1986). A proximal-to-distal sequence is also believed to be critical for obtaining fast release speeds in javelin throwing as the linear speeds of joint centres increased from the shoulder to the hand (Liu, Leigh, and Yu, 2010; Whiting, Gregor, and Halushka, 1991; Mero *et al.*, 1994). Mero *et al.* (1994) suggested that the orderly progression of peak linear speeds describes how KE may be transferred from the proximal shoulder segment to the distal javelin segment. In Javelin throwing research, it has also been suggested that upper extremity motion patterns that are inconsistent with the proximal-to-distal sequence cause shorter throwing distances to be produced (Whiting, Gregor, and Halushka, 1991).

Using a more complex analysis technique to examine segment angular velocities, a proximal-to-distal sequence from the trunk to the wrist has been reported in other striking and throwing movements. Woo and Chapman, (1994) reported that during the squash forehand a proximal-to-distal sequence was evident in both the timing and magnitude of peak linear velocities of segment end-points and peak angular velocities of the striking arm. Furthermore, in the tennis serve, Gordon and Dapena (2006) suggested that the rotation of the lower trunk was followed by the rotation of the upper trunk and the sequential motions of shoulder abduction, elbow extension, ulnar deviation rotation and wrist flexion. The results of these studies clearly suggest that, in striking and throwing movements, sequential motion for the upper extremities are critical for obtaining high speed at the distal end of a linked system.

The majority of kinematic research has found support for the proximal-to-distal sequence in open kinetic chain movements. However, the literature suggests that there are aspects of these movements where modifications are seen in the proximal-to-distal sequence. For example, in the squash forehand (Marshall and Elliott, 2000) and tennis serve (Elliott *et al.*, 1995; Gordon and Dapena, 2006; Marshall and Elliott, 2000) it has been suggested that, rotations around the longitudinal axes do not follow a proximal-to-distal sequence. More specifically, Elliott *et al.*, (1995) demonstrated that in the tennis serve, the internal rotation of the upper arm occurred late in the motion sequence and made a significant contribution to racquet velocity at impact. Marshall and Elliott, (2000) also suggested that in both striking movements, the internal rotation of the upper arm occurred simultaneously with or after wrist flexion, much later than predicted.

In handball throwing, the existence of the proximal-to-distal sequence has also been questioned. Fradet *et al.* (2004) only identified a proximal-to-distal sequence in the magnitude of joint angular velocities. The timing of peak angular velocity did not follow a sequential pattern as maximum angular velocity of the shoulder joint occurred after maximum angular velocity of the elbow. Similarly, Liu *et al.* (2010) reported that, for a group of elite javelin throwers, peak right shoulder internal rotation occurred after peak right elbow extension velocity. In these movements, it is possible that extending the elbow slightly before internal rotation of the shoulder can reduce the moment of inertia and enable athletes to achieve larger distal end angular velocities (Fleisig, *et al.*, 2003).

Kinematic striking and throwing research has also questioned the utilisation of sequential trunk rotation in the generation of distal end speed. An analysis of angular motion in the tennis serve suggested that maximum upper torso angular velocity preceded maximum pelvis angular velocity (Fleisig *et al.*, 2003). Furthermore in handball throwing it has been suggested that players throw with their arm and do not use the sequential motion of the trunk to generate high velocity at the distal segment (Fradet *et al.*, 2004). It was suggested that in an attempt to make their opponents commit an error, handball throwers adopt a non-sequential movement pattern (Fradet *et al.*, 2004).

In summary, the majority of striking and throwing research has provided support for the proximal-to-distal sequence. When high speed at the distal end of a linked system is required it would appear that the proximal-to-distal sequence is desirable. However the literature suggests that in certain striking and throwing movements, internal rotation of the upper arm occurs late in the movement, out of sequence. It has also been indicated that in more open movements such as the tennis serve or throwing in handball, in an attempt to make opponents commit an error, the rotation of the trunk also occurs out of sequence.

2.4.2. Kinetic analyses of striking and throwing movements

Whilst kinematic analyses provide a description of movement patterns, kinetic analyses, through the examination of joint torques and motion-dependent interaction torques can provide an understanding of the mechanics that cause movements (Hirashima *et al.*, 2008). Kinetic analyses of open kinetic chain multi-joint striking and throwing movements have provided evidence that proximal joints create a dynamic foundation for distal segment motion (Cross, 2005; Hirashima *et al.*, 2008; Putnam, 1993). The motion of the proximal segment, produced by large muscles, has been reported to produce powerful interaction torques which can be exploited by distal segments to generate large velocities (Gribble and Ostry, 1999; Hirashima *et al.*, 2008; Putnam, 1991).

Support for the kinetic chain and influence of motion-dependent interaction moments has been provided in kicking (Dörge, *et al.*, 2002; Nunome *et al.*, 2006; Putnam, 1993), overarm throwing (Hirashima, Kudo, and Ohtsuki, 2003; Hirashima *et al.*, 2007) and baseball pitching (Feltner and Dapena, 1989). In soccer kicking, it has been reported that as a result of the motion-dependent interaction moments, the angular velocity of the thigh, early in the forward swing was closely related to the angular acceleration of the lower leg (Dörge *et al.*, 2002). It has also been suggested that, as the angular velocity of the lower leg was still increasing after a backward knee muscle moment was exhibited, the acceleration of the lower leg is dominated by the interaction moment caused by the movement of the proximal segment (Nunome *et al.*, 2006).

Therefore, faster distal segment angular velocities have also been attributed to larger motion-dependent interaction moments (Dörge *et al.*, 2002b; Nunome *et al.*, 2006).

In a kinetic analysis of the torques exerted on the throwing arm during baseball pitching, the velocity of elbow extension prior to ball release was associated with the angular velocity of the upper arm and trunk and not with the torque produced by the elbow extensor muscles (Feltner and Dapena, 1989). In a 2D kinetic analysis of a throwing, it was also reported that elbow muscle torque was produced to assist the interaction torque produced by the motion of the shoulder (Hirashima, Kudo, and Ohtsuki, 2003). These 2D studies suggest that distal segment speed is created by the preceding motion of the proximal segments. However, these analyses assumed that there was no angular velocity about the longitudinal axis of each segment. The kinematic analyses reported in section 2.4.1 (Gordon and Dapena, 2006; Marshall and Elliott, 2000), suggested that such an assumption is clearly not justifiable in analyses of striking and throwing movements.

The utilisation of motion-dependent interaction torques in the development of distal segment speed has subsequently been reported in more comprehensive, 3D analyses of throwing activities (Hirashima *et al.*, 2007; Hirashima *et al.*, 2008). These 3D analyses also reported that the interaction torques produced by the angular velocity of the distal forearm segment contributed to the internal rotation of the shoulder joint late in the movement sequence (Hirashima *et al.*, 2008). This distal-to-proximal sequence was related to the moment of inertia of the upper arm around the longitudinal axis reducing as the elbow was extended from a right angle to full extension (Hirashima *et al.*, 2007). Therefore, it would appear that as a result of changes in system inertia particularly around the longitudinal axis, interaction torques produced by the distal segments have a positive effect on the motion of proximal segments (Hirashima *et al.*, 2008).

Kinetic analyses have also provided evidence that, to enable athletes to avoid injury, interaction torques produced by the distal segment are also utilized to decelerate the motion of proximal segments (Hore *et al.*, 2011). For example, in kicking, negative interaction torques produced by the movement of the lower leg act as a safety mechanism and enable the thigh segment to be decelerated late in the movement sequence (Hore *et al.*, 2011). Similarly, in 2D throwing, shoulder deceleration before ball release was related to the interaction torque produced by motion of the elbow and not from shoulder antagonist muscle activity (Hirashima, Kudo and Ohtsuki, 2003).

Whilst the role of the interaction torque produced by the distal segment appears to be unclear, the role of muscle torque produced by the distal segments has been well defined in kinetic striking and throwing literature. Research has suggested that distal segment muscle torques are predominantly employed to regulate the interaction torques produced by the proximal segments (Gribble and Ostry, 1999; Hirashima *et al.*, 2003; Pigeon *et al.*, 2003). 2D (Hirashima *et al.*, 2003) and 3D throwing (Hirashima *et al.*, 2007) analyses have reported that wrist joint muscle torques exactly counteracted the interaction torque produced by the proximal segment. In baseball batting, it has also been suggested that a backward wrist torque was applied late in the downswing to control the interaction torques produced by the proximal segments (Cross, 2005).

2.4.3. Summary of the proximal-to-distal sequence in striking and throwing movements

The proximal-to-distal sequence has been associated with both mechanical and muscular rewards which enable high speed at the distal end of a linked system to be produced. Kinematic research examining striking and throwing movements suggested that a proximal-to-distal sequence of linear and angular velocities is evident in a variety of movements. However, these studies also reported that internal rotation of the upper arm can occur late in the motion sequence, after elbow extension. It has also been suggested that in more open loop movements such as the tennis serve and handball throw, the rotation of the trunk can occur out of sequence in an attempt to make opponents commit an error.

Kinetic analyses of striking and throwing movements indicated that when a proximal-to-distal sequence is adopted the motion of the most proximal joint creates a dynamic foundation for the entire movement. The interaction torques produced by this motion enabled higher speeds at the distal end of a linked system to be produced. Similar to kinematic analyses, these kinetic analyses also suggested that as a result of the moment of inertia around the longitudinal axis reducing after the elbow extends internal shoulder rotation occurred later than expected in the motion sequence. The kinetic analyses also reported that the internal rotation of the shoulder was caused by an interaction torque produced by the rotation of the distal segment.

2.5. Proximal-to-distal sequencing in the golf swing

The following quote taken from Herring and Chapman (1992) attests to the durability of the proximal-to-distal sequence in the golf swing: *'I thought to digest the general rules of motion into verse which are these:*

All motions with the strongest joints performe
Lett the weaker second and perfect the same
The stronger joint its motion first most end
Before the nixt to move in the least intend'

- Thomas Kincaid, 9 February 1687

Little theoretical or empirical research supported this principle until Cochran and Stobbs (1968) suggested that the most effective golf swings adopt a proximal-to-distal sequence of body segments. Since this study, the sequencing of body segments has become an important theme in golf swing instruction and research as it is believed that it can ensure that clubhead velocity increases throughout the downswing and achieves its maximum at impact (Burden, Grimshaw and Wallace, 1998; Cheetham *et al.*, 2008; Vena *et al.*, 2011a).

2.5.1. Kinematic analyses of sequencing in the golf swing

The proximal-to-distal sequence requires several attributes to be evident in the golf swing; all body segments should accelerate and decelerate before impact whilst the club should reach peak velocity at impact (Herring and Chapman, 1992). The order of peak velocity should be proximal-to-distal and each velocity peak should be larger than that of the previous, more proximal segment (Cheetham *et al.*, 2008). In the golf swing, it has been observed that the lower body is already rotated past its starting position late in the downswing while the upper body lags behind (Lindsay, Mantrop and Vandervoort, 2008). This apparent coordination of movement not only gives the golf swing its unique appearance, but is likely to be essential for developing maximum clubhead speed at impact (Lindsay, Mantrop and Vandervoort, 2008).

The striking and throwing research demonstrated that kinematic analyses enable the timing, speed and acceleration of a movement to be critically reviewed. Therefore, kinematic analyses can enable golf swing technique and subsequently performance to be improved by providing a description of the underlying mechanics (Hume, Keogh and Reid, 2005). However, in golf swing research, conflicting results have been reported as a variety of different research protocols have been used to examine the coordination of body segments. It is also challenging to compare these studies as few consistent measures have been used to classify performance and segment motion.

Golf swing analyses have been performed by using three-dimensional coordinate data to construct one dimensional lines to represent the position of body segments and determining the time derivative of the relative angle between the projection of these lines onto the transverse plane and the line of shot (McTeigue *et al.*, 1994; Burden, Grimshaw and Wallace 1998; Myers *et al.*, 2008; Zheng *et al.*, 2008). This technique has predominantly been used to analyse the rotation of the hips and shoulders (McLaughlin and Best 1994). Using this technique, it has been reported that the downswing is initiated by the rotation of the hips towards the target, followed by the forward rotation of the shoulders (McTeigue *et al.* 1994; Burden, Grimshaw and Wallace 1998; Myers *et al.* 2008).

The sequential rotation at the start of the backswing which maximises the separation angle between the hips and shoulders is commonly referred to as the X-Factor stretch and is an established principle used by teaching professionals to maximise the length of the golf shot (Cheetham *et al.*, 2000). This increase in the length of shot has been associated with the muscular benefits of the stretch shortening cycle (Cheetham *et al.*, 2000; Myers *et al.,* 2008). The initial rotation of the pelvis increases the stretch on the large, powerful muscles in the trunk and enables the stretch shortening cycle to be utilised to increase the force of the concentric muscle contraction (McTeigue *et al.*, 1994; Welch *et al.*, 1995; Myers *et al.*, 2008). However, it has also been suggested that pelvis and shoulder separation beyond a certain point may have a negative impact on clubhead velocity (Komi, 2000). Excessive forward rotation of the hips or late initiation of shoulder rotation can place excessive stretch on trunk muscles and contribute to a reduction of muscular force (Joyce, Burnett and Ball, 2010; Welch *et al.*, 1995).

Research using angles created by projecting segments onto a global plane has provided support for the proximal-to-distal sequence (McTeigue *et al.*, 1994; Burden, Grimshaw and Wallace, 1998; Myers *et al.*, 2008). These sequential rotations suggest that golfers are able to take advantage of the mechanical benefits associated with the summation of speed and kinetic link principles as well as the muscular advantages associated with the SSC. However, this analysis technique only represents one component of segment rotation along a single imposed axis (Vena *et al.*, 2011a). As the golf swing is a complex three-dimensional movement it has been suggested that it is important for golf swing analyses to consider other components of trunk rotation such as trunk lateral bending and flexing as they can have a large impact on the resultant magnitude and timing of peak angular trunk rotation velocity (Chu, Sell and Lephart, 2010; Hellstrom, 2009). Furthermore, in the golf swing, the hips leading the downswing is also just one phase of a much more complex movement pattern (Robinson, 1994). Therefore, to gain a more complete understanding of golf swing mechanics it is important to consider other components of trunk rotation and the rotations of other body segments (Vena *et al.*, 2011a).

More comprehensive three-dimensional kinematic analyses of the proximal-to-distal sequence in the golf swing have been performed by assessing the sequence in which joint angular velocities (Zheng, 2008) and segment angular velocities (Cheetham *et al.*, 2008; Neal *et al.*, 2007; Tinmark *et al.*, 2010) achieve their maximums. Analyses of the proximal-to-distal sequence in terms of joint angular velocities can provide a description of the actual anatomical motion which occurs at each joint (Rodacki *et al.*, 2001). However, the use of analyses of joint angular velocities has been limited to an examination of the effect of playing standard on golf swing related injuries (Zheng *et al.*, 2008). It was reported that whilst professional players adopted a proximal-to-distal sequence, less skilled players accelerate the club towards the ball at the beginning of the downswing causing maximum distal joint angular velocities to be achieved much earlier (Zheng *et al.*, 2008).

Descriptions of the proximal-to-distal sequence in the golf swing have been most frequently made in terms of segment angular velocity components. This form of analysis has been termed the kinematic sequence by Cheetham *et al.* (2008). Although there appears to be agreement that a description of the kinematic sequence is the most appropriate way to analyse the proximal-to-distal sequence in the golf swing, a number of different techniques and definitions of the segment co-ordinate system have been used (Brown, Selbie and Wallace, 2013).

Neal *et al.* (2007) used four Polhemus sensors attached to the pelvis, upper torso, left arm and left hand to provide position and orientation data for a model of five body segments. The motion of the forearm was predicted on the basis of the motion of the arm and hand. Resultant angular velocities were calculated with respect to each segment's local coordinate system. This was defined along the principal axis and had its origin at the centre of the proximal joint. The proximal-to-distal sequence has also been analysed using only three body segments (Cheetham *et al.*, 2008; Tinmark *et al.*, 2010). Tinmark *et al.* (2010) used three Polhemus sensors to examine the motion of the pelvis, torso and hand. They calculated the resultant angular velocity for each segment with respect to the global coordinate system. Meanwhile, Cheetham *et al.* (2008), used four Polhemus sensors to analyse the sequential motion of three body segments (pelvis, thorax and upper arm) and a club segment. Using a similar method to Neal *et al.* (2007), Cheetham *et al.* (2008) described rotations in relation to each segment's anatomical coordinate system. However, they only considered one component of rotation for each segment. For the pelvis and thorax, angular velocity was represented by the component around the superior-inferior axes whilst, for the arm and club, the angular velocity component around a normal to the instantaneous swing plane was reported.

Regardless of analysis technique or segment configuration, these kinematic analyses unequivocally suggested that the magnitude of segment angular velocities increased sequentially from the most proximal to the most distal segments (Cheetham *et al.*, 2008; Neal *et al.*, 2007; Tinmark *et al.*, 2010). However, the proximal-to-distal sequence also requires peak segment angular velocity to be achieved in a proximal-to-distal order. These analyses of segment angular velocities provided far less conclusive evidence about the timing of peak segment angular velocity in the golf swing.

For professional (Cheetham *et al.*, 2008) and highly skilled male and female (Neal *et al.*, 2007) golfers it has been suggested that the timing of peak segment angular velocities followed a proximal-to-distal sequence. The proximal-to-distal sequence has also been identified in miss-timed shots when clubhead speed is dramatically reduced (Neal *et al.*, 2007). However, in these lower quality shots, the separation between the pelvis and upper body was much larger at the start of the downswing and pelvis angular velocity reached its peak much earlier in the downswing than in well-timed shots (Neal *et al.*, 2007). Research has also suggested that the existence of a proximal-to-distal sequence in the timing of peak segment angular velocity is dependent on playing standard (Cheetham *et al.*, 2008). Although, regardless of playing standard, peak pelvis angular velocity occurred first, for the amateur players, peak trunk angular velocity occurred after peak arms angular velocity (Cheetham *et al.*, 2008).

These studies (Cheetham *et al.*, 2008; Neal. *et al.*, 2007) appear to present similar findings which have practical relevance for less skilled players and support the existence of the proximal-to-distal sequence. However, the results of both studies were not supported by statistical analyses. Closer inspection of the results actually indicated that peak thorax and peak upper arm angular velocity occurred at a similar time (Table 2.1). Without support from statistical analysis, the results reported by Cheetham *et al.* (2008) also suggest that similar timing sequences were produced by professional and amateur golfers.

Table 2.1 Time (ms) of peak angular velocity prior to impact.

	Cheetham *et al.* (2008)				Neal *et al.* (2007)			
	Pros		Amateurs		Males		Females	
	Mean	SD	Mean	SD	Mean	SE	Mean	SE
Pelvis	87	19	78	38	113	2	116	2
Thorax	68	14	59	29	74	2	88	2
Arm	65	8	64	23	73	2	83	2
Forearm					53	3	19	2

Note- SD – Standard deviation, SE – standard error

Using a three segment model, Tinmark *et al.* (2010) established that a proximal-to-distal sequence in the timing of segment angular velocities was used by elite golfers in full golf swings with the emphasis on distance and in partial golf swings with the emphasis on accuracy. The existence of a proximal-to-distal sequence in partial shots is consistent with studies of skilled individuals in reaching movements (Yamasaki *et al.*, 2008), piano playing (Furuya and Kinoshita, 2007) and overarm throwing at different speeds (Hirahsima *et al.*, 2007). These studies all suggested that the proximal-to-distal sequence is required to control the accuracy of the movements. However, the findings presented by Tinmark *et al.* (2010) could be questioned as a three segment model was used which ignored the motion of the arms. Previous research (Cheetham *et al.*, 2008; Neal *et al.*, 2007) has suggested that peak angular velocity of the upper arm occurs at a similar time to peak angular velocity of the thorax. Therefore, due to its close proximity to the club, the most distal segment, the use of the hand segment is more likely to produce a proximal-to-distal sequence.

Similar to the studies which used angles projected onto the transverse plane, more complex kinematic analyses have suggested that the pelvis should initiate the downswing and precede the rotation of the thorax. When the rotation of three or more segments has been considered, research has unquestionably suggested that the magnitude of peak segment velocity followed a proximal-to-distal sequence. It has also been suggested that a proximal-to-distal sequence in the timing of peak segment angular velocities is evident in the swings of highly skilled players. However a number of methodological limitations, particularly a lack of statistical analysis and the use of only one additional arm segment have been identified which question the validity of these results.

Descriptions of the proximal-to-distal sequence in terms of the angular velocity of body segments relative to a reference frame imposed on that segment have frequently been used in the golf swing. However, it has been suggested that comprehensive descriptions of movement sequences require the translational and rotational components of movement to be calculated (Teu *et al.*, 2006). Despite the widespread use of the previously described methodologies and popularity of the kinematic sequence (Cheetham *et al.*, 2008), these conventional analysis methods make no reference to the translation of body segments.

The instantaneous screw axis (ISA) technique can decompose segment motion into a translational velocity in the ISA direction and a rotational velocity from a rotation in the ISA orientation (Ying and Kim, 2005). Therefore, the ISA technique can provide the best representation of segment motion in the golf swing (Vena *et al.*, 2011a). Furthermore, by intuitively choosing the location and orientation of a coordinate system more accurate expressions of angular velocity can be achieved (Vena., 2011a). This is especially important if the segment rotation does not occur about a fixed point. For example, in the golf swing, the rotation of the pelvis should occur about the right leg during the back swing and then about the left leg at impact for a right-handed golfer (Vena *et al.*, 2011a). Use of ISA technique would show this change in the rotation axis as it allows the axis of rotation to change position and orientation during movement (Woltring, 1994).

Vena *et al.* (2011a) validated the ISA technique for use in golf swing analyses as they suggested that the majority of hip, shoulder and arm displacement is the result of rotation about the ISA (Teu and Kim, 2006). Using a three segment model comprising of the pelvis, shoulders and left forearm, and the ISA method, inconclusive findings were reported about the existence of the proximal-to-distal sequence in the golf swing (Vena *et al.*, 2010b). For all five low-mid handicap golfers, the magnitude of segment angular velocities increased from the most proximal to the most distal segment.

However, a proximal-to-distal sequence in the timing of peak segment angular velocities was only evident for two of the golfers. The others adopted a subject specific pattern of peak segment angular velocities (Vena *et al.*, 2011a).

Despite the advantages previously associated with the ISA method, Vena *et al.*,(2011a) did not consider the translation of body segments. The limited use of this technique in golf swing analyses may also be explained by the difficulty associated with the physical interpretation of the results. The ISA method is not compatible for describing the clinical motions of flexion / extension, adduction-abduction and internal rotation-external rotation (Ying and Kim, 2002). The ISA technique is also heavily dependent on segment rotation magnitude (Duck, Ferreira, King, and Johnson, 2004). At instances of low angular velocity, it may not be possible for an instantaneous axis of rotation to be defined (Duck *et al.*, 2004).

In addition to the ISA technique another complex analysis method using a dual Euler angles algorithm (Teu *et al.*, 2006) has been used to analyse segmental sequencing in the golf swing. Although the joint coordinate system (Grood & Suntay, 1983) and Euler angle convention (Chao, 1980) are biomechanical standards for describing angular motion, Teu *et al.* (2006) suggested that the complete description of motion using these techniques requires segment translation to be calculated using an additional three-dimensional position vector. Therefore, as the dual Euler angle algorithm considers both linear and angular components of motion it can provide a more comprehensive representation of 3D motion. Furthermore, using this technique linear and angular components can be merged into dual numbers and treated as a single entity (Teu *et al.*, 2006).

Another advantage of the dual Euler angle method compared to the ISA technique is that it can provide intuitive physical interpretation (Ying and Kim, 2002). Therefore, this technique has been considered suitable for analysing the sequencing of segment motions in the golf swing (Teu *et al.*, 2006). Although, Teu *et al.* (2006) did not examine the proximal-to-distal sequence using this technique, they did suggest that longitudinal axis rotations, external rotation of the upper arm and supination of the forearm were important movements in the development of clubhead speed late in the downswing. Teu *et al.* (2006) also determined that torso rotation and left arm motion made the largest contribution to clubhead speed at impact.

2.5.1.1. Summary of kinematic analyses of sequencing in the golf swing

In summary, it is apparent that there is a lack of agreement amongst researchers regarding the most appropriate kinematic technique to analyse the proximal-to-distal sequence in the golf swing. Numerous techniques of varying complexity ranging from the calculation of segment rotation from the relative angle between two one-dimensional lines to the calculation of segment angular velocity from a non-stationary axis have been used. Regardless of technique, kinematic analyses have suggested that the magnitude of angular velocity increases sequentially from the proximal to the distal segments. The majority of these analyses have also suggested that for professional players the timing of peak angular velocity also followed a proximal-to-distal sequence. However, several methodological limitations have been identified which question the validity of these results.

2.5.2. Kinetic analyses of sequencing in the golf swing

A fundamental question that cannot be answered by kinematic analyses is why the proximal-to-distal sequence is apparent in the golf swing? The previously discussed kinematic analyses only provide a description of golf swing motion. To provide an explanation of segmental sequencing in the golf swing it is necessary to understand the underlying kinetics (Tinmark *et al.*, 2010; Neal *et al.*, 2007).

Few if any athletic disciplines have formed the basis of as many models and simulations as the golf swing. The most likely reason for this is the underlying complexity of the seemingly simple task of bringing a golf club in contact with a ball (Betzler, *et al.*, 2008). As a result of this complexity, every golf swing model is an approximation, with many variables being neglected that are judged or calculated to be of minor importance (Sprigings and Neal, 2000). The relative simplicity of many models may also be explained by the difficulty associated with understanding the influence of each motion-dependent interaction torque (Neal *et al.*, 2007). The use of overly complex models has also been cautioned against as the large number of intertwined independent variables can be extremely difficult to interpret (Sprigings and Neal, 2000).

The first attempt at understanding the underlying kinetics of the golf swing was made by Cochran and Stobbs (1968). To gain an insight into the basic golf swing mechanics Cochran and Stobbs (1968) modelled it using a double pendulum. Since then, various double pendulum models have been used to investigate the mechanics of the golf swing (Jorgensen, 1994; Pickering and Vickers, 1999; Milburn, 1982). In a double pendulum model, the upper lever is formed by the golfer's shoulders and arms and the lower lever represents the club. The two levers are connected by a hinge representing the golfer's hands and wrists. This hinge only works in a single plane in which the upper lever is swung about a fixed point which roughly corresponds to a point in the middle of the golfer's chest.

Using this simple double pendulum model Cochran and Stobbs (1968) reported that maximal clubhead velocity was achieved if the release of the distal segment followed that of the more proximal segment. Cochran and Stobbs (1968) also reported that, an increase in the forward acceleration of the upper lever would tend to increase the acceleration of the lower lever. Since this initial study, the main use of the double pendulum model in the golf swing has been to examine the timing of wrist release. It has been reported that, early uncocking of the wrist joint dramatically reduces the speed of the clubhead at impact (Jorgensen, 1994; Pickering and Vickers, 1999; Milburn, 1982). These studies have also provided support for the summation of speed principle and the proximal-to-distal sequence. It has been suggested that fast, early rotations of the torso segment enable the uncocking of wrist to be delayed and subsequently produce higher clubhead speeds (Sprigings and Neal, 2000).

The double pendulum model has the clear advantage of simplicity. However, its ability to reproduce the typical movements of the golf swing has been questioned as it cannot account for the interactions between the arm and shoulder (Turner & Hills, 1998) and the rotations of the torso (Betzler *et al.*, 2008; Sprigings and Neal, 2000). It has been suggested that a two-dimensional, three-segment model with a second hinge representing a simplified shoulder joint provides a reasonable compromise in the level of complexity required to examine the golf swing (Betzler *et al.*, 2008; Sprigings and Neal, 2000) - Figure 2.1.

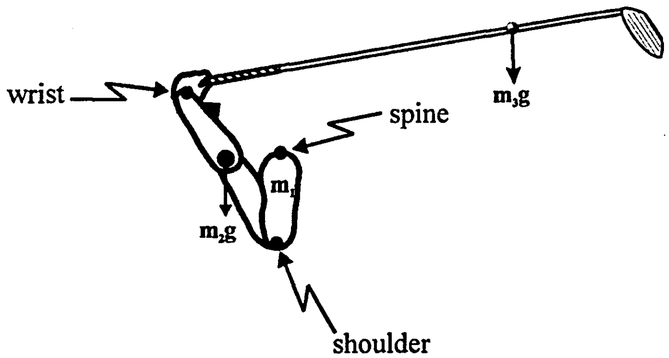


Figure 2.1 Three segment model comprising torso, left arm and club positioned at the top of the backswing, taken from (Sprigings and Neal, 2000).

Two-dimensional three segment models have provided further support for the benefits of a delayed wrist release in the golf swing. It has been suggested that, improvements of 1.6% (Sprigings and MacKenzie, 2002) and 9% (Sprigings and Neal, 2000) in clubhead speed can be achieved if an optimally timed muscular wrist torque is applied just before impact. Cross (2005) questioned whether a golfer is capable of applying a wrist torque late in the downswing. However, in support of the proximal-to-distal sequence, it has been suggested that, the natural release of the clubhead is a function of the angular acceleration of the segments proximal to the wrist joint and can be delayed by employing a coordinated sequence of torso and arm segment accelerations (Sprigings and Mackenzie, 2002). Two dimensional three segment models have also indicated that optimal performance is only achieved if the counter-clockwise rotation of the torso initiates the downswing (Sprigings and Neal, 2000). Furthermore, these models have suggested that in the downswing, activation of torque generators in a proximal-to-distal sequence produced the highest clubhead speed (Sprigings and Neal, 2000; Sprigings and MacKenzie, 2002). Simultaneous activation of the joint torque generators, reduced clubhead speed at impact (Sprigings and Neal, 2000) and activation of the arms prior to the torso also decreased clubhead speed at impact (Turner and Hills, 1998).

Although the use of three segment models significantly increased the ability to replicate the golf swing, the golf swing is fundamentally 3D, not planar (MacKenzie and Sprigings, 2009). Therefore, more realistic, yet more complex models have been developed that consider the movement of the body and club in 3D. Although these 3D models haven't focussed on the sequencing of body segments, it has been suggested that angular movements of body segments in the golf swing should adhere to the proximal-to-distal sequence (MacKenzie and Sprigings, 2009). Using a three segment, 3D model, MacKenzie and Sprigings (2009) also reported that the optimal activation of torque generators was in a proximal-to-distal sequence with the torso being activated first.

The previously described kinetic analyses have advanced the understanding of the proximal-to-distal sequence in the golf swing. However, unlike kinetic analyses of striking and throwing movements which provided an explanation of segment motion sequences, they have yet to provide a comprehensive understanding. A possible explanation is that the full golf swing is one of the most difficult sporting movements to model (MacKenzie and Sprigings, 2009). The many possible interactions make the task of determining how each motion-dependent torque influences the other segments' motion extremely challenging (Neal *et al.*, 2007). The ability of kinetic golf swing analyses to provide an explanation of the underling mechanics is also reduced by the fact that golfers generally grip the golf club with both hands. Consequently, a closed-loop is created which increases the complexity of any inverse or forward dynamics model as an assumption has to be made with regards to the load distribution between the two arms (Betzler *et al.*, 2008). This has led to the vast majority of models ignoring the right arm or combing both arms in one segment (Miura, 2001; Pickering and Vickers, 1999; Sprigings and Neal, 2000) and limited the ability of kinetic models to replicate the actual motion of the body in the golf swing.

Kinetic analyses of the proximal-to-distal sequence in the golf swing have also not considered the motion of the whole body. Even, the most complex 3D models have only considered three body segments (MacKenzie and Sprigings, 2009). These three segment models have not examined the role of the legs, or accounted for any translation of the torso during the downswing (Sprigings and Mackenzie, 2002). It has been suggested that peak clubhead speeds were increased when the central trunk hub moved towards the target (Jørgensen, 1994). Furthermore, an upward shift of the central rotation axis has also been associated with increase clubhead speed at impact (Miura, 2001). Therefore, for kinetic analyses to accurately represent the golf swing, more complex models which consider the translational motion of the trunk would need to be developed.

2.5.2.1. Summary of kinetic analyses of sequencing in the golf swing

In summary, the sophistication of kinetic golf swing models ranges from very simple to very complex. Regardless of which model has been used it is apparent that golf swing modelling has improved our understanding of the proximal-to-distal sequence. The wrist action has been of particular interest to researchers and it has been shown that clubhead velocity at impact can be increased by delaying wrist release. Furthermore, kinetic analyses have reported that optimal clubhead speeds can be achieved if the onset of torques generators follows a proximal-to-distal sequence. However, more complex 3D models which consider the translational movements of the trunk are required to provide comprehensive support for this notion.

2.6. Analyses of kinetic energy

In a system of multiple linked segments, the distal end speed is determined by the magnitude, distribution and transfer of KE (Ferdinands, 2011). Therefore, to achieve maximum distal end velocity in open kinetic chain movements, it is thought that the body must be manoeuvred in such a way as to transfer the most amount of energy from the proximal, to the more distal segments (Chu, Sell and Lephart, 2010). In the golf swing, skilled performance has also been linked with the concept of energy transfer (Hume; Keogh and Reid, 2005; Cheetham *et al.*, 2008; Chu, Sell and Lephart, 2010). It is thought that the successful transfer of energy from proximal to distal segments can lead to increased clubhead velocities being produced (Ferdinands, Kersting and Marshall, 2012; Kreighbaum and Barthels, 1985). However, the term energy transfer has regularly been used when analyses have actually examined the sequencing of segment angular velocity.

Energy is a well-known physical quantity that, despite its popularity in golf literature, is not well understood. Energy takes many forms including nuclear, electrical, and chemical but the form of interest to sports biomechanists is mechanical energy (Robertson *et al.*, 2005). Mechanical energy can appear as KE; the energy a mass possess because it is in motion and potential energy; the energy a mass possesses because of its position (Nigg, MacIntosh, & Mester, 2000). In other words, energy is the motion of particles or the potential to create motion (Robertson *et al.*, 2005).

All of the methods of determining the mechanical energy of a system are indirect as there is no direct way to measure the flow of energy into the system (Robertson *et al.* 2005). The transfer of energy in human movement can be analysed using two distinct methods; the segmental energy calculation and the joint energy calculation (Nigg, MacIntosh and Mester, 2000; Wells, 1988; Winter, 1987).

The joint energy approach was first used by Elftman (1939) to quantify joint power and energy for the ankle, knee and hip during walking. The joint energy approach utilizes the work-energy relationship to describe the sources of segmental energy (Winter, 1987). Using this approach, changes in mechanical energy can be computed by determining the work done on a body by resultant joint forces and resultant joint moments of force at segment endpoints (Robertson *et al.*, 2005; Wells, 1988). The output of this approach enables the collective effect of various body motions to be summarised and the transfer of energy to be studied (Winter, 1987).

Using the joint energy approach, Nigg, MacIntosh and Mester (2000) suggested that the total power flowing to or from a segment (P_{total}) can be calculated using the power due to the resultant joint force (P_{JF}) and that due to the resultant joint moment (P_M)

(Equation 2.1). Winter (1987) noted that power could enter a joint through reaction forces at segment endpoints and was a product of that segment's translational velocity (Equation 2.2). Furthermore, using the angular velocity of a given segment (ω,) Winter (1987) described the power at a joint for a given muscle moment (M) (Equation 2.3):

$$P_{total} = \Sigma(P_{JF} + P_M) \qquad \text{Equation 2.1}$$

$$P_{JF} = \mathrm{Fv} \qquad \text{Equation 2.2}$$

$$P_m = M\,(\omega_1 - \omega_2) \qquad \text{Equation 2.3}$$

Joint energetics and the sources of power have been examined in the golf swing using a three segment model (Sprigings and Mackenzie, 2002). Using a forward dynamics approach to optimise the model variables, Sprigings and Mackenzie (2002) calculated the power at each joint produced by the muscular torques and joint contact forces. The results indicated that the timing of work done followed a proximal-to-distal sequence and that muscular power increased distal segment energy at the expense of decreasing energy in the proximal segment. Although this model applied a high level of realism in the muscle models, they did not account for an onset of muscle torque during the backswing when golfers typically produce force to initiate the downswing (Cheetham *et al.*, 2001).

Perhaps the most comprehensive investigation into joint power in the golf swing was performed by Nesbit (2005). This analysis was performed using a variable 3D full body computer model of a human golfer coupled with a flexible golf club model (Nesbit, 2005). The model assumed that the load between both hands was distributed equally and that all joints were either ball and socket or hinge. The results supported the notion that a delayed wrist motion is extremely important in the generation of large clubhead velocities (Nesbit, 2005). However, rather emphasising the importance of retarding wrist torques as proposed in earlier simulation studies (Sprigings and Neal, 2000; Sprigings and Mackenzie, 2002), the overall coordination of body movements and path of the hands was suggested as the mechanisms that enabled the outward movement and uncocking of the wrists to be delayed (Nesbit, 2005). The ability to do work was also related to skill level as the better golfers were able to do more work rapidly through impact and subsequently generate increased clubhead velocities (Nesbit, 2005). In a later study, Nesbit and Serrano (2005) calculated the work done at each joint in the model using a joint power approach. The results indicated that the back (lumbar and thoracic) and hip joints generated up to 72.2% of the total work done in the swing (Nesbit and Serrano, 2005). Approximately 26% of the total work was contributed by the shoulders and arms. Furthermore, Nesbit and Serrano (2005) suggested that the generation of work and transfer to the club is predominantly a bottom-up, proximal-to-distal phenomenon where a segmental summation of work occurs as the swing progresses.

Although the joint energetic analysis has provided useful information regarding the proximal-to-distal sequence, the difficulties associated with the complex estimation of the net muscle moments and net joint reaction force have limited the use of this technique (Betzler *et al.*, 2008). The accuracy of joint power analyses also depends on a number of assumptions made in the biomechanical modelling process. This method relies on the assumption of spherical joints between segments and joint power measurements are also dependent on the estimation of muscle moments applied at the joint (Anderson, Wright and Stefanyshyn, 2006). Due to these associated difficulties, this technique has had limited use and researchers have tended towards the simpler segment energy approach, (Wells, 1988). Furthermore, it has been suggested that the most accurate way to assess total energy in analyses of human motion and in particular analyses of the golf swing is to use the segmental energy approach (Anderson, Wright and Stefanyshyn, 2006).

The measurement of segmental energetics provides a direct assessment of the quantity of energy a segment contains (Stefanyshyn, 1996). Analyses of segment energetics also enable the destinations of the created energy to be identified (Stefanyshyn, 1996). Wallace Fenn (1930) was the first to use this approach to determine the energetics of human body motion. Fenn (1930) determined the instantaneous mechanical energy of a segment (Equation 2.4) during walking and running by summing the translational KE (Equation 2.5) and rotational KE (Equation 2.6) with the potential energy (Equation 2.7). Total body energy was then calculated as the sum of the energies from each segment (Metzler, 2002).

$$Total\ \mathrm{KE} = Translational\ KE + Rotational\ KE + Potential\ Energy \quad \text{Equation 2.4}$$

$$Translational\ KE = \frac{1}{2}\mathrm{mv}^2 \quad \text{Equation 2.5}$$

$$Rotational\ KE = \frac{1}{2}I\omega^2 \quad \text{Equation 2.6}$$

$$Potential\ Energy = mgH \quad \text{Equation 2.7}$$

where:

m = mass of the segment

v = translational velocity of the segment centre of mass

I = moment of inertia of the segment

ω = angular velocity of the segment

g = acceleration due to gravity

H = height of the segment centre of mass

Analyses of segment KE can be extremely useful in movements when maximal speed of the distal segment is required (Slawinski *et al.*, 2010). The effectiveness of these open kinetic chain movements can be established by examining the magnitude and timing of peak segment KE (Slawinski *et al.*, 2010). Analyses of segment KE can also be used to give an indication of the flow of energy through a linked system (Wells, 1988). It may not be possible to actually measure the transfer of energy through a linked system as changes in segment energy could occur because of muscular activity (Metzler, 2002; Wells, 1988). However, if one segment loses energy and one gains energy, it is possible that the positive changes resulted from energy transfer (Williams and Cavanagh, 1983). Despite this limitation, it has been suggested that the most comprehensive quantification of segmental sequencing is in terms of segment KE (Stefanyshyn, 1996).

The analysis of segment KE has many advantages over conventional kinematic analyses of human movement. Unlike analyses of velocity, this analysis technique considers the inertial properties of segments and is independent of direction (Anderson, Wright and Stefanyshyn, 2006). Analyses of segment KE are also appropriate for examining movement sequences as it enables linear and angular components of movement to be considered separately and as a single entity (Bechar *et al.*, 2009). The KE of individual segments can also be summed to enable the movements of the whole body to be examined simultaneously (Bechard *et al.*, 2009). Furthermore, analyses of KE are related to the square of velocity which makes them significantly more sensitive to subtle changes in technique than analyses of linear or angular velocity (Bechard *et al.*, 2009).

These advantages of analyses of segment KE demonstrate that as a technique it provides a more comprehensive and flexible approach to the analysis of sports movements than traditional kinematic approaches. Therefore, since the pioneering works of Fenn (1930), there has been considerable interest in the analysis of segment KE during human movement. Segment KE has been examined during walking (Usherwood, 2005), in running gait to examine the transition between walking and running (Segers, *et al.*, 2007), in sub-maximal and maximal jumping to establish the energy benefit of the arm swing (Lees, Vanrenterghem and De Clercq, 2004) and in rowing to compare the magnitude of segment KE at racing and training paces (Bechard *et al.*, 2009).

Analyses of segment KE have also been used to examine the proximal-to-distal sequence in open kinetic chain movements such as the sprint start (Slawinski *et al.,* 2010) and cricket fast bowling (Ferdinands, Kersting and Marshall, 2012). Slawinski *et al.* (2010) provided support for the principle of optimal coordination of partial momenta (Van Gheluwe and Bebbelinck, 1985) in the sprint start. It was suggested that to increase the efficiency of this movement total KE should be maximised just before the blocks were cleared. More specifically it was indicated that this could be achieved through better synchronisation of the movements of the head, trunk and lower limbs with the movements of the upper limbs (Slawinski *et al.,* 2010). In contrast, Ferdinands, Kersting and Marshall (2012) provided support for the kinetic link principle (Kreighbaum and Barthels, 1985) in cricket fast bowling. Although there were some notable deviations from the proximal-to-distal sequence such the simultaneous peak in forearm and hand rotational kinetic energies the data provided good evidence that sequential timing of peak segment KE constitutes an important component of effective bowling performance. The early timing of thoracic linear KE and the sequentially later timings of the arm movements were strongly predictive of ball speed (Ferdinands, Kersting and Marshall, 2012).

2.6.1. Analyses of segment kinetic energy in the golf swing

The most appropriate quantification of segmental sequencing as defined by the kinetic link principle is in terms of KE (Ferdinands, 2011). As a result, the concept of energy transfer has been frequently discussed in scientific golf literature (Hume; Keogh and Reid, 2005; Cheetham *et al.*, 2008; Chu, Sell and Lephart, 2010). Despite this popularity and the many advantages associated with the analysis of segment KE, until recently there was no description of the transfer of KE in the golf swing.

The first study to examine the sequencing of segment KE in the golf swing used a kinematic golf model and kinematic data from 45 scratch golfers to calculate total KE of four linked segments as the sum of rotational and translational energies (Anderson, Wright and Stefanyshyn, 2006). The sequencing of segment KE has also been examined by calculating KE using a forward dynamics simulation model in which the muscle torques were driven by three-dimensional kinematic data captured from a single elite male golfer (Kenny, *et al.*, 2008). Regardless of the calculation technique it was reported that the magnitude of peak segment KE increased sequentially from proximal to distal segments (Anderson, Wright and Stefanyshyn, 2006; Kenny, *et al.*, 2008). This finding is in agreement with the results of the kinematic studies which suggested that the magnitude of peak angular velocity followed a proximal-to-distal sequence. However, unlike the timing of angular velocities in kinematic analyses which, in general, provided support for the summation of speed principle, the timing of peak segment KE did not follow the same sequential pattern. Instead, the results provided support for the principle of optimal coordination of partial momenta (Van Gheluwe and Hebblinck, 1985). The three human based segments (Lower Body, Upper Body and Arms) peaked simultaneously at approximately 80% of the downswing movement time whilst peak club segment KE occurred significantly later in the swing, just before impact.

The different timing sequences reported in the analyses of segment KE and segment angular velocities may be caused by analyses of segment KE considering the translational movements of body segments. It is possible that the translational movements of the lower body, trunk and arms segments which have been highlighted as important components of skilled golf swings (Beck *et al.,* 2013; Hume, Keogh and Reid, 2005; Nesbit and McGinnis, 2009) do not conform to the proximal-to-distal sequence. The different timing sequences may also be explained by examining the segments involved in each type of analysis. The analyses of segment KE were able to consider the motion of every body segment involved in the golf swing whilst the kinematic analyses considered the motion of much fewer segments. More specifically, the contribution of the legs and the right arm were considered in the KE analyses (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008) whilst the majority of kinematic analyses ignored these segments (Cheetham *et al.*, 2001; Teu *et al.*, 2006; Tinmark *et al.*, 2010; Vena *et al.*, 2011b). Furthermore, only Neal *et al.* (2007) considered the angular velocity of all the segments comprising the lead arm. Most other studies only considered the more distal segments such as the hand.

Although the sequencing of segment KE has been examined in the golf swing, a comprehensive understanding has yet to be formulated. Both of the previously described studies only considered the sequencing of total segment KE.
It has been suggested that comprehensive analyses of segment motion in the golf swing should consider both translational and rotational movements (Vena *et al.*, 2011a). Therefore, an examination of the linear and angular components of segment KE would take advantage of the benefits associated with the KE technique and provide a more detailed description of the complex movement patterns associated with the golf swing. This information could be used to educate golfers on the best approach to

manoeuvre the body and club during the swing, which will transfer the most amount of energy into the ball and maximise ball flight distance (Hellstrom, 2009).

The sequencing of segment KE has also only been examined for highly skilled golfers (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008). It has been suggested that highly skilled players produce similar timing profiles in the sequencing of segment angular velocity whilst amateur players are much more inconsistent in their timing (Cheetham *et al.*, 2008). Therefore, further analyses of the sequencing of segment KE are required to assess whether these findings are representative of a larger sample of golfers of varying skill level. Investigating the differences in swing techniques between players of different abilities would also enable a better understanding of how to improve golf performance to be formulated (Lindsay, Mantrop and Vandervoort, 2009).

Finally, the effect of club type on the sequencing of segment KE has yet to be comprehensively established. Only Kenny, *et al.* (2008) examined the effect of club type on the sequencing of segment KE however, only one player was considered in this analysis. Whilst it might be possible that their test golfer had a representative golf swing, this in highly unlikely due to the individuality of the golf swing motion (MacKenzie and Sprigings, 2009; Nesbit, 2005). Despite this limitation, it was suggested that the same swing should be used with both clubs. However a more detailed analysis of the results suggests that peak club and hip segment kinetic energies were different between the two club types (Table 2.2). The reduction in club KE would be expected based on the smaller radius of rotation with the 7 iron than the driver. However, the reason for the smaller hip segment energy with the 7 iron is not clear and warrants further investigation.

Table 2.2 Kinetic energy values for the driver and 7 iron (Kenny *et al.*, 2008).

Segment	Driver KE (J)	7 Iron KE (J)
Club	219	174
Arms	136	141
Torso	34	33
Hip	24	13

2.6.2. Summary of segment kinetic energy in the golf swing

The concept of energy transfer has frequently been discussed in scientific golf literature but a paucity of scientific research has examined this notion. Research has reported that the magnitude of peak segment KE increased sequentially from proximal to distal segments. It has also been indicated that for three human based body segments peak KE occurred simultaneously at approximately 80% of the downswing movement time whilst peak club segment KE occurred significantly later, just before impact. Although the research has presented comparable results only highly skilled players and the sequencing of total segment KE has been examined. The effect of club type has also not been analysed in detail. Therefore, further research is required to provide a more detailed exploration of the sequencing of segment KE in the golf swing.

2.7. Three-dimensional data capture

To enable the sequencing of segment KE in the golf swing to be analysed, kinematic descriptions of the golf swing were required. Kinematic analyses of human movement require the body to be considered as a system of rigid bodies linked at joints (Maletsky, Sun, and Morton, 2007). Kinematics can then be described by measuring the motion of these rigid bodies (Maletsky, Sun, and Morton, 2007). In any analysis of human movement accurate and reliable kinematic data are required (Richards *et al.*, 1999) and the choice of data collection system is therefore important.

Three-dimensional (3D) motion capture systems provide a means of quantifying whole body motion (Richards *et al.*, 1999). In golf swing analyses, two categories of instrumentation have commonly been used: electromagnetic and optoelectronic. Electromagnetic systems use electro-magnetic sensors and an electro-magnetic source box to determine the position and orientation of segments in laboratory space whilst optoelectronic systems use infra-red cameras and retro-reflective markers identify body segment position and orientation (Richards *et al.*, 1999).

Traditionally, much of the 3D data collection in scientific golf swing literature has been performed using optical tracking systems (Coleman and Anderson, 2007; Egret *et al.*, 2003; Healy *et al.*, 2011; Myers *et al.*, 2008; Zheng *et al.*, 2008) as they can provide extremely accurate kinematic data (Richards *et al.*, 1999). However, in recent years, as a result of new hardware design and error correction, new generations of electromagnetic tracking systems have shown impressive improvements with respect to accuracy and sensor size (Hummel *et al.*, 2005). It has been suggested that electromagnetic tracking systems have reached a level of sophistication where the positions and orientations of sensors can be determined with accuracies that have clinical relevance (Frantz *et al.*, 2003). Furthermore, electromagnetic tracking systems enable 3D position and orientation to be collected and analysed in real time (Mills *et al.*, 2007), they do not suffer from marker occlusion problems as the body is transparent to magnetic fields (Frantz *et al.*, 2003) and they are extremely portable (Ng *et al.* 2007). Therefore, the use of electromagnetic tracking systems in golf swing analysis has become more prevalent (Cheetham *et al.*, 2008; Neal *et al.*, 2007; Teu *et al.*, 2006; Tinmark *et al.*, 2010).

2.7.1. Electro-magnetic tracking systems

Generally, an electromagnetic tracking system comprises a small sensor, an electromagnetic transmitter and an electronic controller. The transmitter and sensor both contain three orthogonal coils; the coils in the transmitter generate three orthogonal magnetic fields that induce electrical currents in the sensor coils (Barratt *et al.*, 2001). The relative strengths of the induced currents are used to calculate precisely, the 3D position and orientation of the sensor in relation to the transmitter (Barratt *et al.*, 2001).

Two types of electromagnetic tracking system are commercially available. In principle, the tracking methods used by both systems are identical (Hummel *et al.*, 2005) however the systems differ in the manner in which they generate magnetic fields. Alternating current devices (Polhemus Inc., Colchester, VT, USA) generate continuously varying magnetic fields whilst, direct current devices (Ascension, Burlington, USA) use quasistatic magnetic fields. It has been suggested that the devices are capable of tracking body motion with a similar level of accuracy (Barratt *et al.*, 2001). However, it is possible that direct current devices are less susceptible to errors caused by the presence of metals within the magnetic field as measurements can be taken once the direct current magnetic field has stabilized (Milne *et al.*, 1996).

The accuracy of translational and rotational measurements is a major concern when describing the relative motion of the human body (Bull, Berkshire and Amis, 1998). As such, numerous analyses have been performed to determine the accuracy of various electromagnetic tracking systems (Bull, Berkshire and Amis, 1998; Mills *et al.*, 2007; Milne *et al.*, 1996; Richards *et al.*, 1999). It has been suggested that during a static trial, an electromagnetic tracking system (Polhemus), was capable of measuring the distance between two sensors (40 cm) with a root mean square (RMS) error of 3 mm and a maximum error of 6.6 mm (Richards *et al.*, 1999). More recently, it has been reported that an electromagnetic tracking system (Aurora, Northern Digital Inc., Waterloo, Canada) was capable of measuring the distance between two static sensors (50 cm) with a maximum translational error of just 0.5 mm (Schuler *et al.*, 2005). It has also been reported that, under static conditions, the Aurora electromagnetic tracking system had a maximum rotational error of 1° (Schuler *et al.*, 2005). These results suggest that during static trials electromagnetic tracking systems can collect kinematic data with acceptable accuracy. However, static trials are not reflective of motion capture in golf swing analyses.

The accuracy of positional and rotational movements made by the 'Flock of Birds' (Ascension, Burlington, USA) electromagnetic tracking system has been examined using a technique reflective of their use in kinematic analyses (Milne *et al.*, 1996). It was reported that, when operating within its optimal range, the Ascension system had a maximum positional error of 1.0 mm, a mean error of 1.8% of the translational step size for steps of 25 - 150 mm and a mean rotational error of 1.6% of the rotational increment for rotations of 1 - 20° (Milne *et al.*, 1996). It was concluded that the Ascension system was sensitive enough to determine positional and rotational changes of 0.25 mm and 0.1° respectively. Using a very similar technique, Bull, Berkshire and Amis, (1998) also reported that the Flock of Birds Ascension system was capable of measuring sensor translation with a mean error of 0.2% for step sizes of 25 - 200 mm.

A slightly larger mean rotational error of 2.0% was reported for a rotation range of 0.7 - 29.8° (Bull, Berkshire and Amis, 1998).

It has been suggested that for kinematic analyses, estimates of length are acceptable if errors of less than 3 mm are produced (Klein and DeHaven, 1995). Therefore, the previously described results suggest that electromagnetic tracking systems are capable of measuring sensor positions with the required accuracy to provide acceptable measures of length (An *et al.*, 1988; Barratt *et al.*, 2003; Bull, Berkshire and Amis 1998; Milne *et al.*, 1996). The relatively small errors provided for rotational measurements also suggest that electromagnetic tracking systems can measure sensor orientation with acceptable accuracy. However, optical tracking systems have traditionally been regarded as the gold standard for kinematic measures (Hasan, Jenkyn and Dunning 2007).

2.7.2. Optical tracking systems

The linear accuracy of optical tracking systems has been examined by measuring the distance between two markers attached to a reference bar (Everert *et al.*, 1999). Using two markers 10 cm apart, the accuracy of linear measurements made using the Northern Digital's 'Watsmart' optical tracking system was examined at nine locations in a capture volume (Haggard and Wing, 1990). For all nine locations, it was reported that the standard deviation of the length of the bar was between 2.1 and 3.4 mm and was independent of location (Haggard and Wing, 1990). Using two markers 50 cm apart it has also been reported that the mean estimate of length (49.87 ± 0.35 cm) produced by the Ariel Performance Analysis optical tracking system was accurate and independent of location (Klein and DeHaven, 1995). Furthermore, the rotational accuracy of the Ariel Performance Analysis optical tracking system was examined by attaching three markers to a goniometer, one at the axis and one at the end of each arm. This system achieved high levels of rotational accuracy as a mean error of 0.26 ± 0.21° was reported for angles (10 - 180° at 10° increments) created by these markers (Klein and DeHaven, 1995).

The previously described studies (Haggard and Wing, 1990; Klein and DeHaven, 1995) provided support for the accuracy of static linear and rotational measurements made using optical tracking systems. To provide further support for the use of optical tracking systems in kinematic analyses the linear and rotational accuracy of the Motion Analysis Expert Vision system was examined using markers moving unsystematically during gait and sit-to-reach motions (Vander Linden, Carlson and Hubbard, 1992). Although the known distance (180.00 mm) between markers was consistently underestimated in gait (177.1 mm) and sit-to-reach (176.8 mm) motions mean rotational errors of just 0.5° from 20 - 90° and 0.6° from 100 - 160° were reported (Vander Linden, Carlson and Hubbard, 1992).

The most comprehensive study of optical tracking system accuracy determined the accuracy of seven such systems (Richards *et al* 1999). When measuring the distance (50 cm) between two stationary markers, they reported that five of these systems produced a root mean square (RMS) error of less than 1.0 mm. Although the RMS error increased to 2.0 mm when the sensors were moving these systems retained sufficient accuracy of linear measurements (Richards *et al.*, 1999). The seven optical tracking systems were also capable of measuring the absolute angle formed by three markers within 1.5° of the actual angle (95.8°) and with a RMS error of 3.0°. Furthermore, the most accurate optical tracking system (Motion Analysis Corporation) was able to estimate this angle within 0.1° of the actual angle and with a RMS error of just 1.7°. More recently it has been suggested that the Optotrak 3020 (Northern Digital Inc., Waterloo, Canada) optical tracking system was able to measure translation and rotation with high accuracy (Maletsky, Sun and Morton, 2007). This system measured the change in position between two markers (±30 mm in 10 mm steps) with a mean error of just 0.03 mm and it also measured the change in rotation (30° in 10° increments) with a mean error of just 0.04° (Maletsky, Sun and Morton, 2007). In summary, the majority of studies suggested that optical tracking systems can measure marker translation and rotation with acceptable accuracy.

The results from the previously described studies suggested that similar levels of accuracy can be achieved using optical and electro-magnetic tracking systems.

Furthermore, in a study which examined the accuracy of both types of system, Richards *et al.*, (1999) concluded that the accuracy of position and orientation data acquired from both studies was comparable. Further support for the use of electromagnetic tracking systems in kinematic analyses has been provided (Hassan, Jenkyn and Dunning, 2007). In an analysis of upper arm and forearm kinematics performed using the Flock of Birds electromagnetic tracking system and Motion Analysis Corporation optical tracking system (Santa Rosa, CA, USA) it was suggested that both systems performed with acceptable accuracy (Hassan, Jenkyn and Dunning, 2007). The mean errors in the estimation of elbow flexion-extension and internal-external angles for the optical tracking systems were -0.92° and -1.28° compared to -1.09°and -1.23° respectively for the electromagnetic tracking systems (Hasan, Jenkyn and Dunning, 2007).

2.7.3. Practical considerations of data capture systems

When comparing electro-magnetic and optical tracking systems it has been suggested that electromagnetic tracking systems can produce position and orientation information faster than optical tracking systems as only minimal data processing is required (Evans *et al.*, 2012; Richards *et al.*, 1999). Furthermore, a major advantage of using electromagnetic tracking systems is that the 3D position and orientation of multiple sensors can be made in real time at up to 240 frames per second (Mills *et al.*, 2007; Ng *et al.*, 2007). This can be extremely useful in kinematic analyses of the golf swing as the quality of data can be assessed in real-time (Ng *et al.*, 2007). It also provides electromagnetic tracking systems with a distinct advantage when large sample sizes are required as data processing can take up to 30 minutes per trial when optical tracking systems are used (Evans *et al.*, 2012; Richards *et al.*, 1999).

Another advantage of using electromagnetic tracking systems in golf swing analyses is that the body is transparent to magnetic fields (Frantz *et al.*, 2003; Mills *et al.*, 2007). Unlike optical tracking systems which require an unobstructed optical path between the sensors and cameras to be maintained, electromagnetic tracking systems are not subject to the same marker occlusion problems (LaScalza, Arico and Hughes, 2003). In golf swing analyses, optical tracking systems can also experience problems caused by marker accumulation (Betzler *et al.*, 2006). Marker occulsion and marker accumulation can cause data to be lost and subsequently increase post-collection data processing time (Evans *et al.*, 2012). Although these issues can be addressed by using an appropriate number of cameras with proper spacing and orientation achieving this set-up takes significantly longer than the set-up of a electromagnetic tracking systems (Evans *et al.*, 2012; Hasan, Jenkyn and Dunning, 2007).

The increasing popularity of electromagnetic tracking systems in golf swing analyses may also be explained by their portability (Ng *et al.*, 2007). Unlike optical tracking systems which require fixed equipment and are therefore often confined to an indoor laboratory (Richards *et al.*, 1999; Betzler *et al.*, 2006) it is relatively easy to transport electromagnetic tracking systems to a range of testing venues (Schuler *et al.*, 2005). Furthermore, electromagnetic tracking systems have become a popular choice for 3D data capture due to their low cost (Barratt *et al.*, 2001; Schuler *et al.*, 2005; Mills *et al.*, 2007). Low cost optical tracking systems are available; however, the accuracy of these systems has been questioned. In static and dynamic tests of linear accuracy, a low cost optical tracking system (OptiTrack, Natural point, OR, USA) produced higher errors than a high cost system (Vicon System, Oxford, UK) (Thewlis *et al.*, 2011). It was also reported that measurements of lower body angular kinematics produced by these systems differed by up to 4.2°

Despite these advantages of electromagnetic tracking systems there are limitations concerning their use in kinematic data capture. For example, the accuracy of electromagnetic tracking systems can be heavily compromised by presence of metal objects within the capture volume (Barrett *et al.*, 2001; Mills *et al.*, 2007). This reduced accuracy is caused by the development of eddy currents in nearby metal objects, which subsequently produces a secondary magnetic field and causes the magnetic field emitted from the transmitter to be distorted (LaScalza, Arico and Hughes, 2003). This susceptibility to metal interference can limit the type of kinematic analyses for which electromagnetic tracking systems are suitable (Ng *et al.*, 2007). It has also been suggested that when electromagnetic tracking systems are used outside of their optimal range the decay of the magnetic field range produces increased positional and rotational errors (Bull, Berkshire and Amis, 1998; Day, Murdoch and Dumas, 2000; Meskers *et al.*, 1999; Milne *et al.*, 1996). Research has suggested that electromagnetic tracking systems can operate optimally when sensors are within 55 cm (Schuler *et al.*, 2005), 70cm (Day, Murdoch and Dumas, 2000) and 76 cm (Richards *et al.*, 1999) of the transmitter.

These limitations have the potential to limit the accuracy of kinematic golf swing analyses. However, it has been suggested that appropriate calibration protocols can improve the performance of these systems (Richards *et al.*, 1999). Calibration processes create a distortion map of the volume and enable errors associated with metallic interference and magnetic field decay to be reduced (Richards *et al.*, 1999). Furthermore, calibration procedures can enable accurate data to be collected in environments previously considered unsuitable for electromagnetic tracking systems (Day, Murdoch and Dumas, 2000).

Although errors associated with metallic interference and magnetic field decay can influence the accuracy of kinematic data collected using electromagnetic tracking systems, kinematic data collections with optical tracking systems are not without limitations. It has been suggested that the accuracy of these systems is decreased when rigid bodies are rotated away from the optical axis of the cameras (Klein and DeHaven, 1995; Scholz, 1992; States and Pappas, 2006). Vander Linden, Carlson and Hubbard (1992) suggested that larger errors (1.0°) were produced in the measurement of angles (110 - 160°) when markers were rotated by 45° compared to errors produced (0.5°) when the markers were rotated into the plane of the camera. This decreased accuracy may have been caused by distortion of the image of the spherical marker when they were rotated away from the optical axis of the camera (States and Pappas, 2006). Marker occulsion and marker accumulation can also cause data to be lost and subsequently increase post-collection data processing time (Evans *et al.*, 2012). Therefore, in movements that produce significant rotations such as the golf swing this limitation negatively affects the accuracy of kinematic data collections.

2.7.4. Summary of three-dimensional data capture systems

It is apparent both 3D kinematic data capture systems have a number of associated advantages and limitations that must be considered before they are used in golf swing analyses. Optical tracking systems are expensive, suffer from marker occlusion, accumulation and distortion and require considerable post data collection processing. Electromagnetic tracking systems can be susceptible to interference from ferrous metals and have a limited capture volume. However, if appropriate calibration protocols are followed, electromagnetic tracking systems and optical tracking systems can collect kinematic data of comparable accuracy suitable for measuring 3D position and orientation in the golf swing.

2.8. Summary of literature review

The literature review sought to provide an overview of the research and issues associated with the programme of study. The review highlighted that the mechanical and muscular rewards associated with the principles of the proximal-to-distal sequence can enable high speed at the distal end of a linked system to be produced. Due to these advantages, segmental sequencing has become an influential concept in the literature and has therefore been examined using kinematic, kinetic and energetic analyses. In general, kinematic and kinetic analyses have provided support for the existence of the proximal-to-distal sequence in striking and throwing movements. Kinematic analyses of varying complexity also identified that a proximal-to-distal sequence was evident in the swings of skilled golfers. Furthermore, kinetic golf swing analyses also suggested that maximum clubhead speed was generated if the onset of torque generators followed a proximal-to-distal sequence.

Although kinematic and kinetic analyses have provided insight into the sequencing of body segments in the golf swing the literature review suggested that distal end speed in the golf swing was determined by the magnitude and transfer of KE. Furthermore, it was identified that analyses of segment KE provide the most accurate way to assess human motion. Research has indicated that although the magnitude of segment KE increases in a proximal to distal order, peak segment KE for three human based segments occurred simultaneously just before impact. Based on this review and the close scrutiny of these analyses it has been suggested that further research is required to examine the effect of playing standard and club type and also to examine the translational and rotational components of segment KE.

Finally, the literature review assessed methodological considerations associated with the collection of accurate and reliable 3D position and orientation. The review suggested that if appropriate calibration protocols are followed, electromagnetic tracking systems and optical tracking systems can collect kinematic data of comparable accuracy suitable for measuring 3D position and orientation in the golf swing.

2.9. Thesis aim and hypotheses

The overall aim of this thesis was to address the research question; is a proximal-to-distal sequence of segment KE evident in the golf swing? It was hypothesised that the magnitude of peak total segment KE would conform to the kinetic link principle and increase sequentially from the most proximal to the most distal segment. In contrast, it was also hypothesised that the timing of peak total body segment KE would conform to the principle of optimal coordination of partial momenta as peak total body segment KE would occur simultaneously with peak total Club segment KE occurring later in the downswing just before impact. This thesis also sought to examine the translational and rotational components of total segment KE. It was hypothesised that both the magnitude and timing of these components of total KE would follow a proximal-to-distal sequence.

In order to provide a comprehensive answer the overall research question, a series of studies were designed with associated sub research questions. To establish if a sequence of segment kinetic is used by highly skilled golfers the first of these studies sought to answer the research question; does club type have an effect on the sequencing of segment KE in the golf swings performed by highly skilled players? The thesis also sought to examine if this sequence was bespoke to highly skilled players by answering the research question; does playing standard affect the sequencing of segment KE in the golf swing? Finally, it is possible that different swing strategies are used in the golf swing which might affect the sequencing of body segment movements. Therefore, the final study sought to answer the research question; does weight transfer style affect the sequencing of segment KE in the golf swing?

3. Chapter III - Methodology

3.1. Introduction

The review of previous literature in Chapter II highlighted the need for research to examine the sequencing of segment KE in the golf swing. This chapter presents the method that was used to undertake such an analysis. It also provides information on some of the methodological considerations associated with the measurement of segment KE in the golf swing such as the accuracy and reliability of body segment inertial parameter estimations and the effect of associated errors on measures of segment KE in the golf swing. The methods described were developed in conjunction with Golphysics Ltd.

This chapter is divided into four main sections. The first section provides a description of the geometric model used to collect subject specific body segment inertial parameters. The second section presents a detailed description of the geometric modelling approach used to represent the hand segment. The third section outlines the method used to collect kinematic data whilst the final section explains the technique that was used to calculate the magnitude and timing of segment KE in the golf swing.

3.2. Body segment inertial parameters

3.2.1. Introduction

The literature review (Chapter II) highlighted that a comprehensive understanding of the proximal-to-distal sequence in the golf swing has yet to be formulated. Due to the advantages associated with analyses of segment KE, it was also suggested that this form of analysis would provide the most appropriate quantification of body segment sequencing. However, few studies (Anderson, Wright and Stefanyshn, 2006; Anderson, 2007; Kenny *et al.*, 2008) have considered the sequencing of segment KE in the golf swing. A possible explanation for the paucity of research might be the additional methodological complexity associated with its measurement. The analysis of segment KE requires translational and rotational velocities to be calculated for each rigid body.

It also requires the inertial parameters mass, centre of mass location (COM) and moment of inertia (MOI) to be estimated for each rigid body.

Several methods have been used to calculate body segment inertial parameters (BSIP) which fall into three main categories; regression based techniques (Hindrichs, 1985; Yeadon and Mortlock, 1989), scanning techniques (Durkin and Dowling, 2006; Pearsall and Reid, 1996), and methods based on geometric modelling (Hanavan, 1964; Hatze, 1980; Jensen, 1978). Using simple input parameters such as height and body mass regression based techniques allow the rapid estimation of BSIP. However, the determined parameters are not fully customised to an individual's body morphology (Lariviere and Gagnon, 1999). As small changes in BSIP estimates can have a significant influence on kinetic measures the use of this technique has the potential to reduce the accuracy of measures of segment KE (Damavandi *et al.*, 2009).

Scanning techniques such as magnetic resonance imaging (MRI), gamma ray scanning, dual-energy X-ray absorptiometry (DEXA) and computed tomography (CT) enable the accurate BSIP measurements to be made directly on humans (Durkin and Dowling, 2003). However, because of inaccessibility, financial expense, the need for highly-trained operators and their time consuming nature, it is generally not practical for investigators to use these techniques in scientific studies (Damavandi, Farahpour and Allard, 2009).

Geometric models can provide accurate, subject specific BSIP estimates and they are often seen as a compromise between data collection time and accuracy (Wicke and Dumas, 2010). These models assume that body segments can be represented by simple shapes the dimensions of which can be obtained by taking anthropometric measurements on the participant. From this information, it is possible to obtain approximations of segment volume, then, by providing an estimate of segment density; it is possible to estimate inertial parameters.

To provide BSIP estimates which take into account segment volume and shape geometric models of varying complexity have been proposed (Kingma *et al.*, 1996). An early model proposed by Hanavan (1964) used a single geometric shape and uniform density to represent each segment. Although relatively simple, this approach was not sensitive to segment contour variations (Wicke and Dumas, 2010) which resulted in the model being able to estimate body mass with an average estimation error of 9.7%. More complex models have since been proposed; Jensen (1978) introduced a photogrammetric method which divided segments into a series of elliptical zones two centimetres thick whilst Hatze (1980) proposed a model consisting of 17 segments subdivided into small mass elements of different geometric structures. These models were capable of providing accurate BSIP estimates with body mass estimation errors of only 2% and 3% respectively being reported. However, their complexity and large data collection times have limited their use in biomechanical research (Wicke, Dumas and Costigan, 2009).

More recently, Yeadon (1990) presented a geometric model comprising 40 geometric solids which has subsequently been used in numerous biomechanical analyses (e.g. Muri, Winter and Challis, 2008; Domire and Challis, 2007; Kong, 2010). Possible reasons for this popularity include a relatively low body mass estimation error (2.3%) and a reduced data collection time, the necessary measurements can be made in less than 30 minutes. Therefore, to enable the sequencing of segment KE to be analysed in this programme of research BSIP were estimated using a geometric model similar to that defined by Yeadon (1990). The geometric model used 28 geometric shapes to produce BSIP estimates for 17 rigid bodies. This reduction in geometric solids compared to the Yeadon (1990) model provided an appropriate balance between model complexity and data collection time.

3.2.2. Geometric representation

Geometric models (Hanavan, 1964; Jensen, 1978) have typically used elliptical solids (Figure 3.1) to model rigid bodies because it was mathematically convenient to do so. However, Yeadon (1990) suggested that stadium solids (Figure 3.1) provide a more accurate representation of the trunk and hand. A stadium may be defined as a rectangle of width (w) 2t and depth (d) 2r with an adjoining semi-circle of radius r at each end of its width (Figure 3.1). Therefore, following the recommendations of Yeadon (1990), in this programme of research, the arms, legs and feet were modelled using elliptical solids and the trunk, neck and hands were modelled using stadium solids. The cranium was also modelled as a semi-ellipsoid with base diameter d and height h.

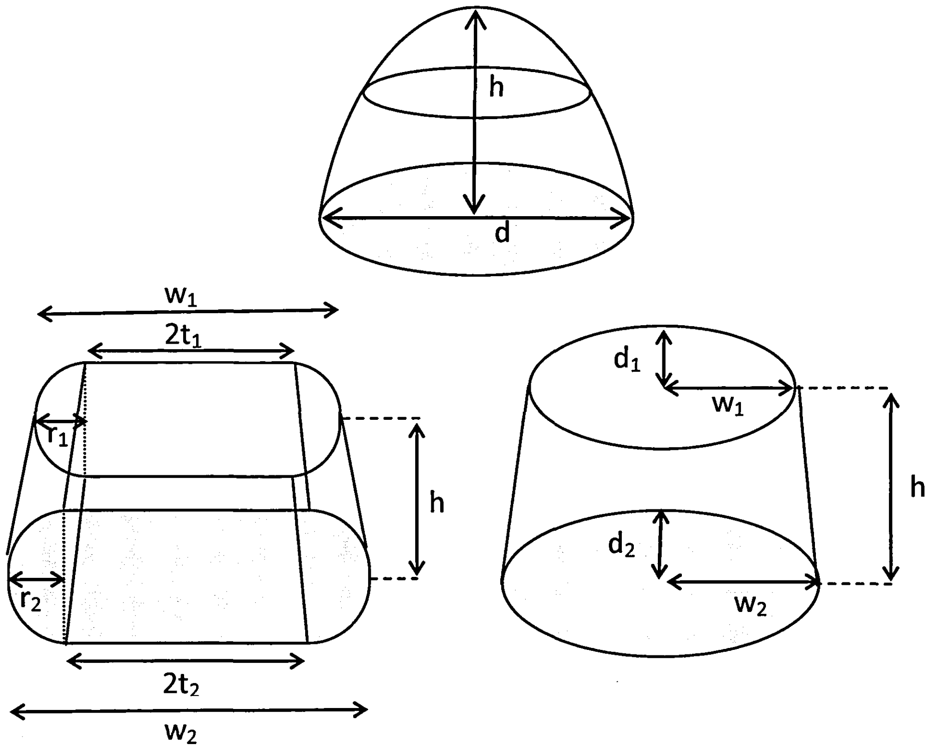


Figure 3.1 A semi-ellipsoid (top), a stadium solid (left) and an elliptical solid (right) used to model body segments (Yeadon, 1990).

Yeadon (1990) calculated the geometry and volume of geometric shapes by taking anthropometric measurements directly from participants using a tape measure. To form the stadium solids, Yeadon (1990) measured a perimeter and width at each boundary level as well as the height between boundary levels. The elliptical and semi-elliptical solids were formed using height and perimeter measurements, width measurements were not required. In this programme of research, a Polhemus electromagnetic tracking system was used to make the anthropometric measurements. The nature of this equipment dictated that width (w) and depth (d) values were used to estimate the perimeter of the stadium (Equation 3.1) and elliptical (Equation 3.2) solids. This technique has been reported to provide a successful alternative for obtaining BSIP estimates using the geometric modelling technique (Gittoes, Bezodis and Wilson, 2009).

$$Stadium\ solid\ perimeter = 4t + 2\pi r \qquad \text{Equation 3.1}$$

$$Elliptical\ solid\ perimeter = \pi \times \sqrt{(\left(\frac{1}{2}w\right)^2 + \left(\frac{1}{2}d\right)^2)} \qquad \text{Equation 3.2}$$

3.2.3. Segmentation

The geometric model segmented the body into geometric solids using planes perpendicular to the long axes of the rigid bodies at specified boundary levels (Figure 3.2, Figure 3.3, Figure 3.4 and Figure 3.5). Although the geometric model used in this programme of research was similar to that proposed by Yeadon (1990) some alterations were made. It has been suggested that the accuracy of trunk segment inertial parameter estimates are highly dependent on the number of shapes used in the modelling process (Erdmann, 1997). Therefore, to more effectively account for the contours of the trunk additional stadium solids were used to represent the upper and lower trunk segments (Figure 3.3). Furthermore, to maintain a balance between data collection time and model complexity, the geometric model represented the hands and feet using one geometric shape.

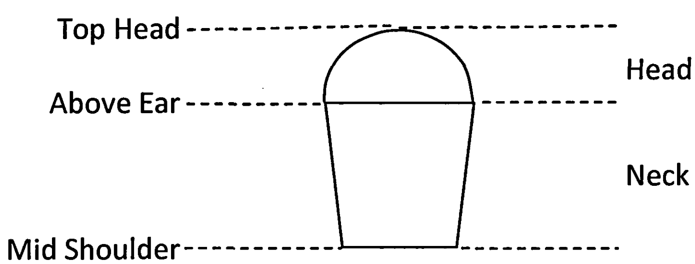


Figure 3.2 Head and neck segmentation panes.

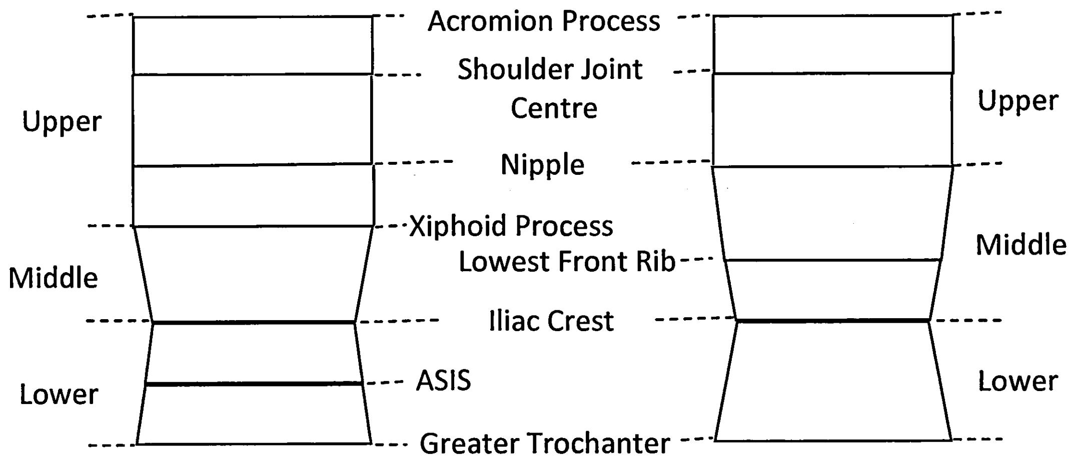


Figure 3.3 Trunk segmentation plane comparison: left – current geometric model, right – Yeadon (1990) model. ASIS – Anterior Superior Iliac Spine.

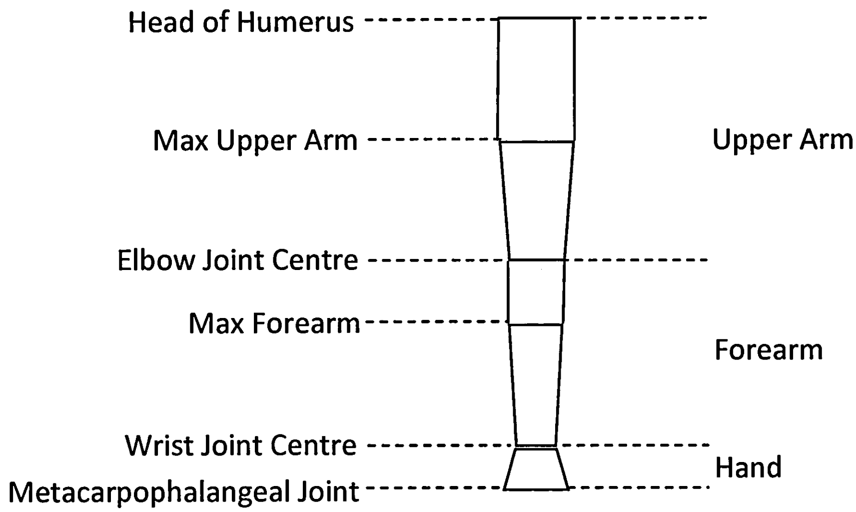


Figure 3.4 Arm and hand segmentation planes.

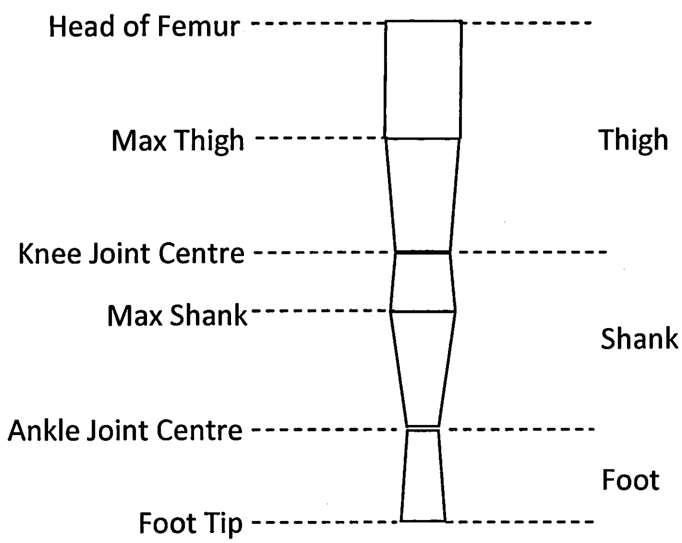


Figure 3.5 Leg and foot segmentation planes.

3.2.4. Anthropometric measurements

To enable anthropometric measurements to be made using the Polhemus electro-magnetic tracking system, a custom designed suit comprising a baselayer jacket with adjustable straps was used to attach twelve electro-magnetic sensors to the following locations: posteriorly to the upper trunk at the level of T3, posteriorly to the middle trunk at the level of T6 and posteriorly at the mid-point of each upper arm (Figure 3.6). Using adjustable straps, sensors were attached posteriorly to the shanks and thighs and laterally on the right side of the lower trunk at the mid-point between the anterior superior iliac spine and greater trochanter (Figure 3.6). Sensors were also attached to the back of each hand using modified golf gloves and to the right side of the head behind the ear using a cap (Figure 3.7). The position of 78 anatomical landmarks was then identified by palpation and recorded in relation to the relevant sensor using the Polhemus digital stylus (Figure 3.8), Polhemus electro-magnetic tracking system and custom written software.

Figure 3.6 Electromagnetic sensors attached using: left – baselayer jacket with adjustable straps, right – adjustable leg straps.

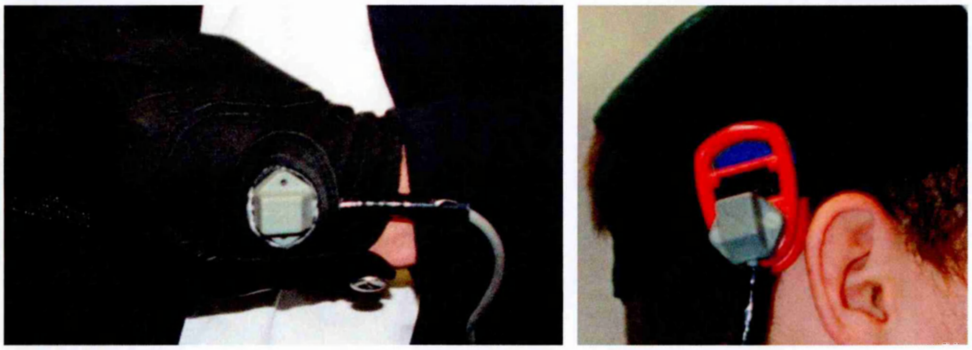


Figure 3.7 Electromagnetic sensors attached using: left – golf gloves, right – cap.

Figure 3.8 The Polhemus digital stylus.

To define the geometric model, the anatomical landmarks were identified on the right limbs with the participants in the anatomical position, standing upright with their arms by their sides, fist clenched and thumbs pointing forwards. The left and right limbs were assumed to be symmetrical. For each rigid body, width and depth measurements were made using the recorded position of anatomical landmarks at the specified segmentation planes (Table 3.1, Table 3.2 and Table 3.3). However, the proximal end of the neck at the acromion process and distal end of the foot at the foot tip were assumed to be circular. Therefore, the width and depth measurements at these levels were defined as the distances between the sternal notch and T2 and the medial and lateral toe respectively. The calculation of some anthropometric measurements also required virtual anatomical landmarks to be identified. These virtual landmarks were calculated as the mid-point between two previously identified anatomical landmarks (Table 3.4). Furthermore, segment height was calculated as the resultant distance between the joint centres at each end of the segment.

Table 3.1 Anatomical landmarks used to make anthropometric measurements at Head, Neck and Trunk segmentation planes.

Boundary level	AM	Anatomical Landmarks
Top head		Top Head
Temporal Mandibular	Width	Right Temporal Mandibular - Left Temporal Mandibula
	Depth	Bridge of Nose - Occipital Tuberosity
Acromion Process	Width	Right Acromion Process - Left Acromion Process
	Depth	Sternal Notch - T2
Shoulder Joint Centre (SJC)	Width	Virtual AL RSJC - Virtual AL LSJC
	Depth	Sternomanubrial Joint - T4
Nipple	Width	Right lateral aspect at Nipple level - Left lateral aspect at Nipple Level
	Depth	Virtual AL Nipple - T6
Xiphoid Process	Width	Right lateral aspect 6th rib - Left lateral aspect 6th rib
	Depth	Xiphoid Process - T8
Iliac Crest	Width	Right Iliac Crest - Left Iliac Crest
	Depth	Umbilicus - L4
ASIS	Width	Right Lateral ASIS - Left Lateral ASIS
	Depth	Lower Umbilicus - Virtual AL Posterior Superior Iliac Spine (PSIS)
Greater Trochanter	Width	Right Greater Trochanter - Left Greater Trochanter
	Depth	Right Mid-Groin Anterior - Right Mid-Groin Posterior

Table 3.2 Anatomical landmarks used to make anthropometric measurements at Arm and Hand segmentation planes.

Boundary level	AM	Anatomical Landmarks
Head of Humerus	Width	Anterior Shoulder - Posterior Shoulder
	Depth	Acromion Process - Axilla
Max Upper Arm	Width	Max Upper Arm Lateral - Max Upper Arm Medial
	Depth	Max Upper Arm Anterior - Max Upper Arm Posterior
Elbow Joint Centre	Width	Lateral Femoral Epicondyle - Medial Femoral Epicondyle
	Depth	Bicep Insertion - Olecranon
Max Forearm	Width	Max Forearm Lateral - Max Forearm Medial
	Depth	Max Forearm Anterior - Max Forearm Posterior
Wrist Joint Cente	Width	Radial Styloid - Ulna Styloid
	Depth	Mid-Wrist Extensor - Mid-Wrist Flexor
Metacarpophalangeal Joint	Width	2nd Metacarpal - 5th Metacarpal
	Depth	Posterior 3rd Metacarpal - Anterior 3rd Metacarpal

Table 3.3 Anatomical landmarks used to make anthropometric measurements at Leg and Foot segmentation planes.

Boundary level	AM	Anatomical Landmarks
Head of Femur	Width	(Right Greater Trochanter - Left Greater Trochanter)/2
	Depth	Right Mid-Groin Anterior - Left Mid-Groin Posterior
Max Thigh	Width	Max Thigh Lateral - Max Thigh Medial
	Depth	Max Thigh Anterior - Max Thigh Posterior
Knee Joint Centre	Width	Lateral Epicondyle – Medial Epicondyle
	Depth	Tibial Tuberosity - Popliteal Crease
Max Shank	Width	Max Shank Lateral - Max Shank Medial
	Depth	Max Shank Anterior - Max Shank Posterior
Ankle Joint Centre	Width	Lateral Malleolus - Medial Malleolus
	Depth	Posterior Fibula - Anterior Talus
Foot Tip	Width	Lateral Toe - Medial Toe

Table 3.4 Anatomical landmarks used to define the virtual anatomical landmarks.

Virtual AL	AL 1	A1 2
RSJC	Right Acromion Process	Right Axilla
LSJC	Left Acromion Process	Left Axilla
Nipple	Right Nipple	Left Nipple
PSIS	Right PSIS	Left PSIS

3.2.5. Joint centre locations

The locations of the majority of joint centres were estimated by calculating virtual landmarks at the mid-point between a medial and a lateral anatomical landmark (Table 3.5). This approach assumes that the two anatomical landmarks lie on the joint axis and that the anatomy of the joint is symmetrical. At most joints this assumption is adequate. However, for the shoulder and hip which define the proximal ends of the thighs and upper arms respectively it is not possible to estimate joint centres using this technique as easily palpatable anatomical landmarks which lie on the joint axis do not exist.

Several predictive methods have been presented in the literature to estimate the location of the hip joint centre (Bell, Pedersen and Brand, 1989; Davis *et al.*, 1991; Seidel *et al.*, 1995). The approaches defined by Bell, Pedersen and Brand (1989) and Davis *et al.* (1991) are commonly used in biomechanical analyses. However, it has been suggested that the equations defined by Bell, Pedersen and Brand (1989) are more accurate (Leardini *et al.*, 1999). Using this approach, the identification of easily identifiable landmarks on the lower trunk enabled a local coordinate system to be fixed in this segment using the technique described in section 3.4.1. The location of the hip joint centre was then estimated in the lower trunk-fixed frame using the following regression equations (Equation 3.3, Equation 3.4 and Equation 3.5) where x, y and z are the coordinates of the right hip joint centre and IAD is the distance between the right and left anterior superior iliac spines.

$$x = -0.19IAD \qquad \text{Equation 3.3}$$

$$y = -0.30IAD \qquad \text{Equation 3.4}$$

$$z = 0.36IAD \qquad \text{Equation 3.5}$$

Similar to the hip joint, easily identifiable anatomical landmarks are not available for the estimation of the shoulder joint centre. To estimate the location of the shoulder joint centre the International Society of Biomechanics recommends the use of functional methods such as those described by Stokdijk, Nagels and Rozing (2000) and Veeger (2000). However, these methods often require information about the movement of the scapula or the acromion which are not available using the proposed geometric model. The predictive equations presented by Meskers *et al.* (1998) were also deemed unsuitable as they require the movement of the scapula to be estimated. Other, simpler predictive approaches such as that proposed by Schmidt *et al.* (1999) which estimate the shoulder joint centre based solely on the location of the acromion processes have been adopted in Biomechanical research. Using this approach the shoulder joint centre was defined at a position 7 cm vertically below the acromion process in the global coordinate system.

Table 3.5 Anatomical landmarks (AL) used to calculate joint centre locations.

Joint Centre	Medial AL	Lateral AL
Ankle	Medial Malleolus	Lateral Malleolus
Knee	Medial Epicondyle	Lateral Epicondyle
Greater Trochanter	Left Greater Trochanter	Right Greater Troachanter
Iliac Crest	Left Iliac Crest	Right Iliac Crest
Xiphoid Process	Left Xiphoid Process	Right Xiphoid Process
Acromion Process	Left Acromion Process	Right Acromion Process
Temporal Mandibula	Left Temporal Mandibula	Right Temporal Mandibula
Elbow	Medial Femoral Epicondyle	Lateral Femoral Epicondyle
Wrist	Radial Styloid	Ulna Styloid

3.2.6. Inertial parameter calculation

Once the anthropometric measurements had been made, the BSIP; segment mass, centre of mass location and principal moments of inertia (Ixx, Iyy and Izz) were calculated using the equations defined by Yeadon (1990) and a uniform density function (Dempster, 1955) - Table 3.6. In accordance with the International Society of Biomechanics (ISB) guidelines, all local coordinate system were defined such that the x, y and z axes were predominantly sagittal, longitudinal and frontal respectively. Furthermore, the position of the centre of mass for each rigid body was calculated from its proximal end. More details on the definition of the local coordinate systems can be found in section 3.4.1.

Table 3.6 Rigid body density values (Dempster, 1955).

Segment	No. of Solids	Density (kgl^{-1})
Head	1	1.11
Neck	1	1.11
Upper Trunk Top	1	1.04
Upper Trunk	2	0.92
Middle Trunk	1	0.92
Lower Trunk	2	1.01
Upper Arm	2	1.07
Forearm	2	1.13
Hand	1	1.16
Thigh	2	1.05
Shank	2	1.09
Foot	1	1.10

3.2.7. The reliability and accuracy of body segment inertial parameter estimates made from the geometric modelling technique

A study investigating the repeatability of body segment inertial parameters estimated using the geometric model was conducted and is reported in Appendix I. As inertial parameters in this programme of research were estimated by one examiner, the primary purpose of this study was to examine the intra-examiner reliability of BSIP estimates. As the inertial parameters were also required to be made on different days a secondary purpose of this study was to examine the between-day reliability. The majority of BSIP were estimated with satisfactory intra-examiner and between-day reliability. It was concluded that if inertial parameter estimates were made by the same examiner, acceptable reliability was achieved using the geometric model and data collection technique outlined earlier in this section.

The accuracy with which geometric solids are capable of estimating the geometry and volume of human body segments was also assessed and is reported in Appendix II. This study suggested that the geometry and volume of human body segments can be accurately represented using the formation of simple geometric shapes proposed in the geometric model. Subsequently, using a uniform density function (Dempster, 1955) it was suggested that segment mass (Table A 2.2) and COM location could be estimated with mean absolute errors of 7.4 ± 1.1 % and 2.2 ± 1.3 % respectively. Furthermore, the results suggested that the three principal moments of inertia, I_{xx}, I_{yy} and I_{zz} could be estimated with mean absolute errors of 8.0 ± 4.1 %, 7.7 ± 4.7 % and 9.0 ± 5.1 % respectively (Table A 2.3). Therefore, the majority of limb inertial parameters could be accurately estimated using the geometric modelling technique.

3.3. The geometric hand model

3.3.1. Geometric hand model accuracy

In the study reported in Appendix II, the accuracy of hand inertial parameter estimates was not examined. To maintain a stable body position during the 3D scans the participants were required to hold a support frame. This meant it was not feasible to obtain gold standard volume estimations for the hand. However, errors in the estimation of hand inertial parameters, particularly hand mass and hand COM have the potential to significantly influence the magnitude of peak hand KE. Therefore, a study was conducted to examine the accuracy with which hand inertial parameters could be estimated using the geometric model outlined in section 3.2 – Appendix III. Compared to estimates made using a regression equation (Zatsiorsky, 2002), hand mass was consistently underestimated using the geometric model (Table A 3.1). These results suggested that a single stadium solid does not provide an accurate representation of the geometry or volume of the hand with the fist clenched. Therefore, improvements to the geometric modelling approach were required before it can be used to estimate hand inertial parameters in analyses of segment KE.

3.3.2. Geometric hand model development

In golf, the complexity of modelling the hand using geometric shapes is increased as the fingers wrap around the club grip. Therefore, popular geometric models such as those proposed by Hanavan (1968), Jensen (1978) and Yeadon (1990) lack validity for use in golf swing analyses as they model the hand with the fingers extended. Furthermore, it has been indicated that modelling the hand with the fist clenched using one geometric solid consistently underestimated the mass of the hand (Appendix III). Therefore, improvements to the geometric model were required which enabled accurate inertial parameters to be provided for the hand gripping a golf club.

It has been suggested that a hand gripping a dumbbell can be accurately modelled using a truncated cone to represent the base of the hand and a hollow cylinder to represent the fingers grasping a dumbbell (Challis and Kerwin, 1996) (Figure 3.9). However, Yeadon (1990) suggested that compared to a truncated cone the base of the hand can be more accurately represented using a stadium solid. The diameter of a golf grip is also smaller than that of a dumbbell which changes the shape created by the fingers. Therefore, it has been proposed that accurate estimates of hand inertial parameters can be produced by modelling the base of the hand using a stadium solid (Figure 3.1) and by modelling the fingers wrapped around a golf club using a segment of a hollow cylinder (Figure 3.10).

Hand

Figure 3.9 A truncated cone and hollow cylinder used to represent the hand gripping a dumbbell (Challis and Kerwin 1996).

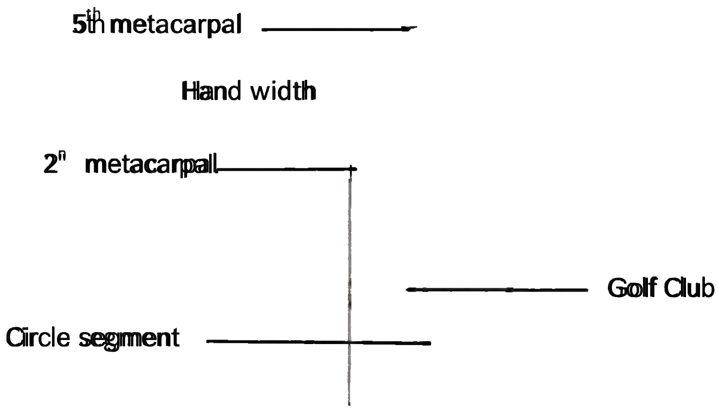


Figure 3.10 A segment of a hollow cylinder used to represent the fingers of the left hand grasping a golf club.

3.3.2.1. Hand mass

Hand mass was estimated by summing the volume of the stadium solid calculated using the technique described in section 3.2 with the volume of the hollow cylinder segment and multiplying this volume by a uniform density (Dempster, 1955) - Table 3.6. The volume of the hollow cylinder segment was calculated by multiplying its cross-sectional area (Figure 3.11) by the width of the hand; defined as the distance between the 2nd and the 5th metacarpals (Figure 3.10). The cross-sectional area of the hollow cylinder segment was determined by calculating the area of a large circle (Equation 3.6) with radius R (Equation 3.9) and subtracting the cross-sectional area of the golf club (Equation 3.7) and the segment of the circle not used to define the fingers (Equation 3.8). In these calculations, r was defined as the radius of the golf club grip, θ was the angle which defines the circle segment and the circle sagitta (Figure 3.11) was defined as the mean finger depth (Equation 3.10).

$$Area\ of\ the\ large\ circle = \pi\, R^2 \qquad \text{Equation 3.6}$$

$$Cross\ sectional\ area\ of\ the\ golf\ club = \pi\, r^2 \qquad \text{Equation 3.7}$$

$$Area\ of\ a\ segment = \frac{1}{2}\,(\theta - sin\theta)R^2 \qquad \text{Equation 3.8}$$

$$R = r + sagitta \qquad \text{Equation 3.9}$$

$$Circle\ sagitta = \frac{Hand\ width}{4} \qquad \text{Equation 3.10}$$

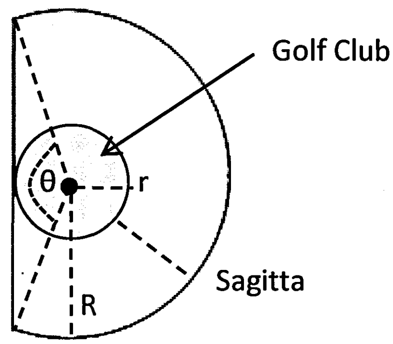


Figure 3.11 A cross-section of the hollow cylinder segment.

3.3.2.2. Centre of mass location

The hand, represented as a stadium solid and hollow cylinder segment, is symmetrical in the medio-lateral (x) and anterior posterior (z) axes. However, the combination of these geometric shapes is not symmetrical in the longitudinal (y) axis and as such, the COM is not located at the mid-point along the y axis. To locate the COM of the hand from its proximal end in the y direction (y_H) further calculations were required using the mass (m) and centre of mass location (y) of the stadium solid (SS) and hollow cylinder segment (HCS) (Figure 3.12) (Equation 3.11).

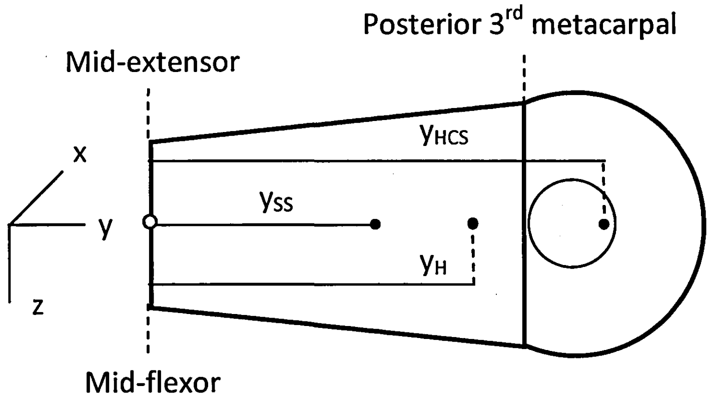


Figure 3.12 Hand COM location

$$y_H = \frac{m_{SS}y_{SS} + m_{HCS}y_{HCS}}{m_{SS} + m_{HCS}} \quad \text{Equation 3.11}$$

The mass and COM of the stadium solid and the mass of the hollow cylinder segment were calculated using previously defined techniques (Section 3.2). The location of the COM of the hollow cylinder segment (y_{HCS}) required further calculation (Equation 3.12) using the area (a) and COM location (y) of a large circle (LC) with radius R, a golf club (GC) with radius r and the segment of the large circle (LCS) which was not used to define the hollow cylinder segment. The COM location of the LC (y_{LC}) and GC (y_{GC}) from the proximal end of the hand were defined by summing the length of the SS and the radius of the GC. The COM location of the segment of the large circle (y_{LSC}) was calculated using Equation 3.13.

$$y_{HCS} = \frac{a_{LC}y_{LC} - a_{GC}y_{GC} - a_{LCS}y_{LCS}}{a_{LC} - a_{GC} - a_{LCS}} \quad \text{Equation 3.12}$$

$$y_{LCS} = \text{SS length} + \frac{4R\,sin^3\frac{\theta}{2}}{3(\theta - sin\theta)} \quad \text{Equation 3.13}$$

3.3.2.3. Moment of inertia

It was not possible to estimate hand moment of inertia using a hollow cylinder segment to represent the fingers. For calculations of moment of inertia, the fingers were modelled as a half cylinder with radius R. Hand MOI (Ixx, Iyy and Izz) were then calculated as the sum of the MOI of the SS and half cylinder. For the calculation of Ixx and Izz, the parallel axis theorem (Equation 3.14) was used to calculate the MOI of the SS and half cylinder about axes through the centre of mass of the hand. In these calculations d represented the perpendicular distance between the two parallel axes, I_{CM} was the MOI about the hand COM and I_{PA} was the MOI about the parallel axis.

$$I_{CM} = I_{PA} + md^2 \qquad \text{Equation 3.14}$$

The technique used to define the three components of MOI for the SS was described in section 3.2. The three components of MOI for the half cylinder were calculated using Equation 3.15 and Equation 3.16. These calculations required the mass (Equation 3.17) and the COM (Equation 3.18) of the half cylinder (HC) and the COM of the hand, modelled as a stadium solid and half cylinder (x_{H2}) to be calculated (Equation 3.19).

$$Ixx\ and\ Izz = \tfrac{1}{4}mr^2 + \tfrac{1}{3}ml^2 \qquad \text{Equation 3.15}$$

$$Iyy = \tfrac{1}{2}mr^2 \qquad \text{Equation 3.16}$$

$$m_{HC} = (\tfrac{1}{2}\pi R^2) \times hand\ width\ x\ hand\ density \qquad \text{Equation 3.17}$$

$$x_{HC} = \frac{4R}{3\pi} \qquad \text{Equation 3.18}$$

$$x_{H2} = \frac{m_{SS}x_{SS} + m_{HC}x_{HC}}{m_{SS}m_{HC}} \qquad \text{Equation 3.19}$$

3.3.2.4. Revised geometric hand model accuracy

The accuracy of hand mass estimates made using this geometric modelling technique was assessed in a study reported in Appendix IV. The results suggested that the stadium solid and segment of a hollow cylinder were capable of estimating hand mass with a mean error of only 1.57 ± 6.44 %. Furthermore, the results suggested that this model was capable of more accurately representing hand mass distribution than the geometric modelling technique proposed by Challis and Kerwin (1996). This accurate distribution would also enable more accurate estimates of hand COM to be calculated.

3.4. Kinematic data collection

If appropriate calibration protocols are followed, electromagnetic tracking systems and optical tracking systems can collect kinematic data of comparable accuracy suitable for measuring 3D position and orientation in the golf swing – see section 2.7. Although data of comparable accuracy can be collected electromagnetic tracking systems are not subject to marker occlusion, accumulation and distortion difficulties and it has been suggested that they are appropriate for large scale data collections due to their ability to provide 3D data in real time (Evans *et al.*, 2012; Richards *et al.*, 1999). Therefore, a sixteen-channel Polhemus Liberty electromagnetic tracking system (Polhemus, Inc., Colchester, VT, USA) sampling at 240 Hz was selected for kinematic data collection in this programme of study.

The electromagnetic transmitter (origin of the global coordinate system) was positioned approximately 0.4 m behind the golfer on a custom built stand with +x directed anteriorly, +y vertically upwards and +z directed away from and parallel to the target line. Twelve, six-degrees-of-freedom electromagnetic sensors were used to record 3D translational and rotational movements of the rigid bodies. These sensors were attached to the golfer at the previously identified landmarks using the custom designed suit and adjustable straps (Section 3.2.4).

The attachment of electromagnetic sensors using this technique has the potential to encumber golfers, make them feel uncomfortable and restrict their movements (Wright, 2008). Therefore, a study was conducted and reported in Appendix V which examined the effect of electromagnetic sensor attachment on swing and launch parameters in the golf swing. This study indicated that electromagnetic sensor attachment had no effect on outcome measures of ball flight and clubhead characteristics. Although lower clubhead (85.9 ± 4.1 vs 84.8 ± 4.3 mph) and ball speeds (100.2 ± 6.9 vs 98.3 ± 7.2 mph) were produced with the 5 and 9 irons respectively when sensors were attached to golfers the difference between these measures was smaller than that expected due to Trackman variability (Appendix VI). Therefore, it was concluded that electromagnetic sensor attachment did not have a meaningful effect on golf swing mechanics.

Polhemus sensors were not attached to secondary rigid bodies. Therefore, estimates of their position and orientation could not be made directly. Instead, the coordinates of proximal and distal joint centres together with a third anatomical landmark were recorded in the local coordinate systems of the primary rigid bodies directly proximal and directly distal to the secondary rigid body. The positions of the anatomical landmarks of the secondary rigid bodies were then reconstructed to enable their positions and orientations to be calculated. Reconstructing the position of rigid bodies in this manner has the advantage that fewer electromagnetic sensors are required to be attached to the participants.

The locations of the previously identified anatomical landmarks enabled each rigid body's local coordinate system to be defined such that the x, y and z axes were sagittal, longitudinal and frontal respectively (Section 3.4.1). To enable local coordinate systems to be defined and kinematic data to be collected for the left limbs, the locations of additional anatomical landmarks were identified and recorded (Table 3.7). The origin of the local coordinate system was translated from the proximal or distal end to the centre of mass of each rigid body and using the matrix method, the translation and rotation of each local coordinate system was calculated with regard to the global system in a manner consistent with the recommendations of the International Society of Biomechanics (Grood and Suntay, 1983; Wu and Cavanagh, 1995; Wu *et al.*, 2002). For all of the segments, angular rotations about the x-, y- and z-axes were defined as forward bend, axial rotation and lateral tilt respectively.

Table 3.7 Additional anatomical landmarks required to define the local coordinate systems for the left limbs.

Joint Centre	Anatomical Landmarks
Foot Tip	Medial Toe Tip
Ankle	Medial Malleolus
	Lateral Malleolus
Knee	Medial Epicondyle
	Lateral Epicondyle
Great Trochanter	Greater Trochanter
Elbow	Medial Femoral Epicondyle
	Lateral Femoral Epicondyle
Wrist	Radial Styloid
	Ulna Styloid
Hand	2nd Metacarpal
	5th Metacarpal

3.4.1. Local coordinate systems

Anatomical calibration trials were performed to enable the raw sensor data to be transformed into segment anatomical coordinate systems, required for the calculation of relevant kinematics. The identification of each local coordinate system required three non-collinear anatomical landmarks to be identified – the origin, endpoint and plane. The origin of each local coordinate system was coincident with the joint centre at its distal end, the endpoint was coincident with the joint centre at its proximal end and the plane was located on the same plane as the origin or endpoint (Table 3.8, Table 3.9 and Table 3.10).

Table 3.8 Anatomical landmarks used to identify the local coordinate systems of the lower body rigid bodies.

Rigid Body	Origin	Endpoint	Plane
Right Foot	Medial Toe Tip	Ankle JC	Medial Malleolus
Left Foot	Medial Toe Tip	Ankle JC	Lateral Malleolus
Right Lower Leg	Ankle JC	Knee JC	Medial Epicondyle
Left Lower Leg	Ankle JC	Knee JC	Lateral Epicondyle
Right Thigh	Knee JC	Hip JC	Medial Epicondyle
Left Thigh	Knee JC	Hip JC	Lateral Epicondyle

Table 3.9 Anatomical landmarks used to identify the local coordinate systems of the trunk, neck and head rigid bodies.

Rigid Body	Origin	Endpoint	Plane
Lower Trunk	Greater Trochanter JC	Iliac Crest JC	Left Iliac Crest
Middle Trunk	Iliac Crest JC	Xiphoid Process JC	Left Xiphoid Process
Upper Trunk	Xiphoid Process JC	Acromion Process JC	Left Acromion Process
Neck	Acromion Process JC	TM JC	
Head	TM JC	Top Head	Left TM

Table 3.10 Anatomical landmarks used to identify the local coordinate systems of the arms and hands.

Rigid Body	Origin	Endpoint	Plane
Right Upper Arm	Shoulder JC	Elbow JC	Medial Epicondyle
Left upper Arm	Shoulder JC	Elbow JC	Lateral Epicondyle
Right Forearm	Elbow JC	Wrist JC	Radial Styloid
Left Forearm	Elbow JC	Wrist JC	Ulna Styloid
Right Hand	Wrist JC	Posterior 3rd Metacarpal	2nd Metacarpal
Left Hand	Wrist JC	Posterior 3rd Metacarpal	5^{th} Metacarpal

Using these anatomical landmarks the following routine procedure was employed to fix the local coordinate system to each rigid body. The y-axis was defined first to be coincident with the long axis of each segment. The y-axis was defined as a unit vector coincident with the line connecting the distal and proximal joint centres – the origin and the endpoint. An intermediate vector **Q** coincident with a line connecting the proximal joint centre and the plane marker was then identified. A vector **X** was then defined as the cross product of vectors **Q** and **Y** and finally a vector **Z** was calculated as the cross product of vectors **X** and **Y**.

3.4.2. Protocol

Fifteen, good quality golf swings; five with a driver, 5-iron and 9-iron were performed by each golfer. Previous scientific golf swing research has analysed the repeatability of 3D pelvis and thorax kinematics (Eveans *et al.,* 2012), the relationship between biomechanical variables and driving performance (Chu, Sell and Lephart, 2010) and the 3D measurement of trunk rotations (Joyce, Burnett and Ball, 2010) using five swings per club. To establish the quality of golf swings, each shot was qualitatively rated on a ten point scale with a 1 representing a shot the player was completely unsatisfied with and 10 representing their interpretation of an ideal swing. Shots rated as less than seven were discounted and another shot was hit. When required, ball flight data from a radar tracking device (Trackman A/S, Denmark) positioned on a sliding rail behind the hitting mat and directly perpendicular with the target were also considered (Figure A 5.1). However, golfers were instructed to replicate their normal swing mechanics and not to focus on the outcome from the ball tracking device. To collect fifteen good quality swing trials, a maximum of 26 golf shots were performed by one of the participants.

A Ping (Ping, Phoenix, Arizona) G15 driver and Ping i15 irons with regular graphite shafts, standard lengths and standard lie angles were used. To control for the effect of fatigue and to ensure internal-validity, the clubs were presented in a randomly assigned order (Thomas, Nelson and Silverman, 2011). Sufficient time was also given for the golfer to perform their usual pre-game warm-up routine and adequate practice trials were allowed to ensure that the golfer was familiar with the clubs, the laboratory environment and the data collection protocol.

3.4.3. Mapping

A custom mapping approach was used to correct for distortions in the magnetic field of the tracking system caused by metal within the test volume. The mapping approach assumed that the transmitter's position was fixed and the surrounding metal does not move. As such, correction of the measurement errors associated with metallic interference and magnetic field decay was achieved by creating a distortion map of the capture volume and establishing the dependencies between the true receiver position/orientation and that reported by the tracking system.

3.4.4. Club inertial parameters

Club segment geometry and inertial parameters were based on measurements made by a non-contact laser scanner (Model Maker D100 non-contact laser scanner, Metris, Leuven, Belgium). The scanned clubhead was imported into Pro-Engineer (Parametric Technology Co., Waltham, MA, USA) and, given the known density of the steel clubhead and the scanned volume, the wall thickness of the clubhead was calculated assuming a constant thickness. This provided club specific mass, centre of mass and inertia parameters. The club segment was assumed to be a rigid body and position and orientation during swing trials were directly obtained from a sensor securely fixed to the shaft just below the grip (Figure 3.13).

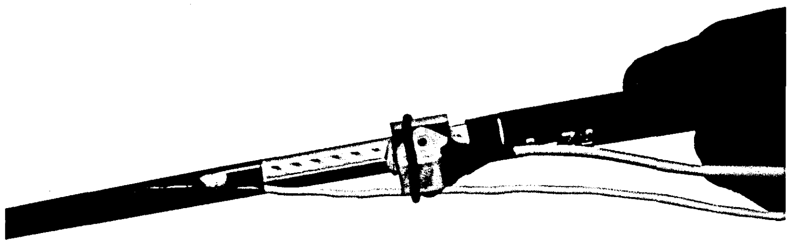


Figure 3.13 Golf club electromagnetic sensor attachment.

3.5. Kinetic energy calculations

KE was calculated for seventeen rigid bodies. Then, following the protocol adopted in the papers by Anderson, Wright and Stefanyshyn, (2006) and Kenny *et al.* (2008) KE was calculated for four grouped segments (Lower Body, Upper Body, Arms and Club) (Table 3.11).

Table 3.11 Rigid bodies defining each groped segment.

Segment	Rigid Bodies
Lower Body	Foot, Lower Leg, Thigh, Lower Trunk
Trunk	Middle Trunk, Upper Trunk, Neck, Head
Arms	Upper Arm, Forearm, Hand

Total segment and total rigid body KE were calculated as the sum of the translational and rotational kinetic energies (Equation 3.20). Translational rigid body KE was found using the mass of each rigid body ($m_{rigid\ body}$) and the resultant translational velocity vector of each rigid body's centre of mass ($\mathbf{V}_{rigid\ body}$) (Equation 3.21). Translational segment KE was found about the centre of mass of each segment as a whole (Equation 3.22). The mass of each segment was found as the sum of the masses of the rigid bodies comprising the segment (Equation 3.23) and the velocity of each segment centre of mass was calculated as the first time derivative of segment centroid position vector (Equation 3.24). The segment centre of mass position vector was a mass normalised sum of all positions of all the segment's rigid bodies (Equation 3.25).

$$Total\ KE = Translational\ KE + Rotational\ KE \qquad \text{Equation 3.20}$$

$$Translational\ Rigid\ Body\ KE = \frac{1}{2} m_{rigid\ body} \boldsymbol{V}_{rigid\ body}{}^2 \qquad \text{Equation 3.21}$$

$$Translational\ Segment\ KE = \frac{1}{2} m_{segment} \boldsymbol{V}_{cm}{}^2 \qquad \text{Equation 3.22}$$

$$m_{segment} = \sum_{i=1}^{n} m_i \qquad \text{Equation 3.23}$$

$$\boldsymbol{V}_{cm} = \frac{d\boldsymbol{S}_{cm}}{dt} \qquad \text{Equation 3.24}$$

$$\boldsymbol{S}_{cm} = \frac{\sum_{i=1}^{n} m_i \boldsymbol{S}_i}{m_{segment}} \qquad \text{Equation 3.25}$$

Rotational KE was calculated as the sum of the local rotational and remote rotational kinetic energies (Equation 3.26). Local rotational segment KE was calculated as the sum of the local rotational energy of the rigid bodies about their own centre of mass. Three local rotational kinetic energies were calculated for each rigid body using the angular velocity vector (ω) which was calculated in the local coordinate system and had x, y and z components and each rigid body's moment of inertia, Ixx, Iyy and Izz (I) (Equation 3.27). Total rigid body local rotational KE was then calculated as the sum of three local rotational kinetic energies.

$$Rotational\ KE = Local\ Rotational\ KE + Remote\ Rotational\ KE \qquad \text{Equation 3.26}$$

$$Local\ Rotational\ KE = \frac{1}{2} I\omega^2 \qquad \text{Equation 3.27}$$

Remote rotational segment KE was calculated as the sum of the remote rotational kinetic energies of the rigid bodies comprising each segment (Equation 3.28). For each rigid body, remote rotational segment KE was found using the mass of each rigid body and by calculating the tangential velocity of each rigid body about the segment centre of mass ($\mathbf{V}_{\Delta tan}$) (Equation 3.28). The tangential velocity component was calculated using the difference velocity vector ($\mathbf{V}_{\Delta}$) and the component of the difference velocity along the radius vector ($\mathbf{V}_{\Delta rad}$) (Equation 3.29). The difference velocity vector was calculated as the velocity of the rigid body minus the velocity of the segment centre of mass (Equation 3.30). Finally, to calculate the $\mathbf{V}_{\Delta rad}$, a radius vector r was calculated for each time interval from the segment centre of mass positon to the rigid body centre of mass position. The component of the difference velocity along the radius vector was then found as the projection of the velocity on the radius (Equation 3.31).

$$Remote\ Rotational\ Segment\ KE = \sum_{i=1}^{n} \frac{1}{2} m_i (V_{\Delta tan_i})^2 \qquad \text{Equation 3.28}$$

$$\boldsymbol{V}_{\Delta tan_i} = \boldsymbol{V}_{\Delta_i} - \boldsymbol{V}_{\Delta rad_i} \qquad \text{Equation 3.29}$$

$$\boldsymbol{V}_{\Delta_i} = \frac{dP_i}{dt} - \frac{dP_{cm}}{dt} \qquad \text{Equation 3.30}$$

$$\boldsymbol{V}_{\Delta rad_i} = \boldsymbol{r}_i\ (\boldsymbol{r}_i \cdot \boldsymbol{V}_{\Delta_i}) \qquad \text{Equation 3.31}$$

Using custom written Matlab scripts, the magnitude and timing of peak segment and peak rigid body total, translational, local rotational and remote rotation kinetic energies were calculated for the downswing phase of the golf swing. The downswing was defined as the time between the top of the backswing (TOB) and impact where TOB represented the point at which the club changed direction at the end of the backswing and impact represented the instance when the club makes contact with the ball. Impact was calculated as the time of a sudden increase in the output of an accelerometer attached at the end of the club shaft. The timing of peak KE was then calculated relative to the total downswing time with 0 representing the top of the backswing and 1 representing ball impact.

3.5.1. Effect of BSIP error on measures of segment kinetic energy

Given the use of analyses of segment KE in the golf swing, it is tempting to take the results at face value. However, the output of such analyses is only valid when the errors associated with its calculation are quantified (Sellers and Crompton, 2004).

Therefore, a study was carried out to examine the influence of BSIP errors on measures of segment KE – Appendix VII. This study suggested that the magnitude of peak segment KE was more sensitive to changes in inertial parameters than the timing of peak segment KE. The largest error of 5.29 % was reported for the estimation of peak Arms KE when the driver was used. Specifically, this study reported that the magnitudes of peak segment total Legs and Arms KE were dependent on the accuracy of estimates of thigh and upper arm geometry, volume and subsequently mass. Furthermore, to ensure that segment KE was estimated with the highest possible accuracy it was highlighted that care should be taken in the identification of anatomical landmarks which define these segments.

4. Chapter IV - The Reliability of Segment Kinetic Energy Measures in the Golf Swing

4.1 Introduction

In a system of multiple linked segments, such as that found in the golf swing, it has been suggested that optimal performance is achieved if a proximal-to-distal sequence of body segment movements is produced (Cochran and Stobbs, 1968; Putnam, 1993). As such, the sequencing of body segment movements has become an important theme in golf swing instruction and scientific research articles (Cheetham *et al.*, 2008; Neal *et al.*, 2008; Tinmark *et al.*, 2010; Vena *et al.* 2011a). It has been suggested that the analysis of segment KE is the most appropriate technique to examine the sequencing of body segments (Anderson, Wright and Stefanyshyn, 2006). As well as incorporating inertial parameters and being sensitive to subtle changes in technique (Bechard, 2009), distal segment speed in striking and throwing movements has frequently been linked with the magnitude, timing and transfer of segment KE (Ferdinands, 2011; Slawinski *et al.*, 2010).

The sequencing of segment KE in the golf swing has been examined in two main studies (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008). However, neither study reported the reliability associated with measures of segment KE. For the analysis of segment KE to be considered as a valid assessment tool in the golf swing, its reliability must be established (Robertson *et al.*, 2005). Reliability is an important measure as it provides an indication of the technical variation of a measurement protocol (de Vet *et al.*, 2006). Measures of absolute reliability also enable the variation in the output which reflects meaningful differences to be determined (Currell and Jeukendrup, 2008).

Measures of segment KE are sensitive to even the most subtle changes in technique (Bechard, 2009; Ferdinands, Kersting and Marshall, 2012). Therefore, repeated measurements of segment KE in the golf swing may show day-to-day differences due to slight changes in the golfer's technique (Drust *et al.*, 2005). Variations in measures of segment KE in the golf swing may also originate from more controllable sources such as the measurement instrument or data collection protocol (de Vet *et al.*, 2006). More specifically, in the calculation of segment KE errors may be caused by the collection of 3D translational and rotational kinematic data, the definition and computation of body segment axes and the estimation of inertial parameters.

In golf swing analyses, 3D kinematics have commonly been collected using two categories of instrumentation: electromagnetic and optoelectronic. In recent years, as a result of updated hardware design and error correction techniques, electromagnetic tracking systems can now collect 3D kinematic data of comparable accuracy to optoelectronic systems (Hummel *et al.*, 2005; Richards *et al.*, 1999). It has also been suggested that electromagnetic tracking systems are well suited to golf swing analyses (Neal *et al.*, 2008) and as such they have been used in several collections of golf swing kinematics (Cheetham *et al.*, 2008; Neal *et al.*, 2008; Teu *et al.*, 2006; Tinmark *et al.*, 2010). The primary advantage of electromagnetic tracking systems for golf swing analyse is that they are not subject to marker occlusion, accumulation and distortion difficulties which have been associated with optical tracking systems (Betzler *et al.*, 2006). They are also capable of providing real-time feedback while on the course, on the practice range or in the laboratory (Evans *et al.*, 2012). Finally, it has been suggested that they are capable of measuring 3D translation and rotation (An *et al.*, 1998; Evans *et al.*, 2012; Schuler *et al.*, 2005) and thorax and pelvis kinematics in the golf swing (Horan *et al.*, 2010) with acceptable reliability.

Although reliable kinematic data can be collected in the golf swing using electromagnetic tracking systems (Horan *et al.*, 2010), variability in the identification of anatomical landmarks has the potential to cause inconsistencies in the collected kinematics (Evans *et al.*, 2012; Ferber *et al.*, 2002; Horan *et al.*, 2010). Reliable identification of anatomical landmarks is important in repeated kinematic measures as it establishes the anatomical coordinate systems from which kinematic parameters are calculated. Slight changes in the position of these anatomical landmarks can produce cross-talk between the planes of motion, or an offset shift in the data (Evans *et al.*, 2012). Inconsistencies in the identification of anatomical landmarks were also suggested to be the cause of variations in repeated estimation of inertial parameters using the geometric modelling technique particularly for the middle trunk, shoulder and thigh segments (Section 3.2).

Variations in inertial parameter estimates caused by inconsistencies in the identification of anatomical landmarks only produced small effects on measures of segment KE (Section 3.6). However, the effect of inconsistent anatomical landmark identification in repeated kinematic data collections has yet to be examined. Furthermore, when combined with the biological variability, inherent measurement system variability and the variability associated with repeated inertial parameter estimates this has the potential to have a significant effect on the magnitude and timing of peak segment KE. Therefore, the aim of this study was to examine the reliability of measures of the magnitude and timing of peak segment KE in the golf swing. This assessment of the test-retest reliability will also enable subsequent studies to determine the meaningfulness of any differences in measures of segment KE (Atkinson and Nevill, 1998).

4.2 Methods

4.2.1 Participants

Seven male golfers (M ± SD; age 31 ± 12 years, stature 185.8 ± 4.7 cm, mass 85.0 ± 5.5 kg and handicap 9.3 ± 8.0 strokes, range +3 - 20 strokes) volunteered to take part in this study. At the time of testing the golfers were required to be playing or practising golf at least once a week. The participants were considered to have the heterogeneity needed to examine test-retest golf swing reliability (Bartlett and Frost, 2008; Evans *et al.,* 2012). The participant's characteristics were also representative of the players that were recruited for studies in Chapters V, VI and VII and of those likely to undergo 3D swing analysis (Evans *et al,.* 2012). That is, participants had golf handicaps ranging between +3 and 20 strokes, were aged between 22 and 55 years and were excluded from participating if they had an injury which prevented them from playing or practising. Between data collections the golfers were also asked not to make any technical alterations to their swing or change the frequency with which they were playing. Before data collection, ethics approval was granted by the Faculty of Health and Wellbeing Research Ethics Committee and each participant provided written informed consent.

4.2.2 Inertial parameters

Inertial parameters were estimated for the feet, lower legs, thighs, lower trunk, middle trunk, upper trunk, neck, head, upper arms and forearms using the geometric model described in section 3.2. Hand inertial parameters were estimated using a stadium solid and segment of a hollow cylinder (Section 3.5). The anthropometric measurements required to form the shapes of this geometric model were made using the Polhemus electromagnetic tracking system, electromagnetic sensors and custom written software (Section 3.2). The club segment geometry and inertial parameters were based on measurements made by a non-contact laser scanner (Section 3.4.4).

4.2.3 Data collection

Data were collected following the protocol described in section 3.4.2 which required participants to hit fifteen shots from an artificial mat into a net five meters away, five with a driver, 5 iron and 9 iron. After an initial data collection, golfers were asked to return to the laboratory approximately one week later to hit another fifteen golf shots. During this collection the golfers followed exactly the same protocol and the order in which the clubs were presented replicated the order used in the initial data collection.

4.2.4 Segment kinetic energy

Translational and rotational velocity data were obtained for each rigid body using the Polhemus Liberty electromagnetic tracking system (Polhemus, Inc., Colchester, VT, USA) sampling at 240 Hz and the technique described in section 3.4. Then, for each golf swing, the magnitude and timing of peak KE was calculated for each rigid body, for three body segments (Lower, Upper Body and Arms), and for the Club segment using the technique described in section 3.5. The timing of peak segment KE was then normalised to total movement time (MT) in the downswing where 0 represents the top of backswing and 1 represents impact.

4.2.5 Statistical analysis

All data were analysed using SPSS (Version 19.0). The means of the five shots using each club in both data collections were used for statistical analysis. Initially, tests of normality (Shaprio-Wilk) were performed to ensure data sets were appropriate for parametric statistical tests. Then, following the recommendations of Atkinson and Nevill (1998), the relative and absolute reliability of the data were assessed using a variety of statistical tests. To compare the mean values produced using each club, paired sample t-tests were performed for each repeated measure of the magnitude and timing of peak segment and peak rigid body KE. Alpha was set at 0.05 and a Cohen's *d* effect size was calculated (Equation 4.1). These effect sizes were interpreted using the benchmarks presented by Cohen (1988) whereby 0.20 equates to a small effect, 0.50 equates to a medium effect and effects larger than 0.8 equate to large effects.

$$Effect\ size\ (d) = \frac{M1-M2}{SD} \qquad \text{Equation 4.1}$$

Intraclass correlation coefficients (ICCs) were calculated to establish relative reliability and enable agreement between data collections to be determined. Other statistical methods such as the Pearson correlation coefficient have frequently been used to determine relative reliability. However, more recently, its use has been discouraged for assessing test-retest reliability (Baumgartner, 2000; Weir *et al.,* 2005) as it cannot detect systematic error and it depends greatly on the range of values in the sample (Larsson *et al.*, 2003; Weir, 2005). Therefore, based on the recommendations of Shrout and Fleiss (1979) a two-way random measure ICC with absolute agreement (ICC 2.1) was chosen to establish test-retest reliability. Following the recommendations of Sleivert and Wenger (1994) the ICCs were interpreted as follows: good reliability: 0.8 - 1.00, acceptable reliability: 0.6 - 0.79, poor reliability: < 0.6. More conservative ICC scales have been used (de Vet *et al.* 2006) however, this ICC classification has been deemed appropriate for reliability analyses in health care studies (Munro, 1986) and it has been used in numerous reliability studies (de Vet *et al.* 2006).

To calculate absolute reliability and express measurement error in the original units of measurement the standard error of measurement (SEM) was calculated for each variable (Equation 4.2). The SEM was then used to calculate the minimum difference (MD) between measures of a variable for the difference in future analyses to be considered real (Weir, Therapy, & Moines, 2005) (Equation 4.3). The limits of agreement (LOA) described by Bland and Altman (1986) have come into vogue as a way of assessing absolute reliability (Atkinson and Nevil, 1998). However, this procedure was primarily developed for use in method comparison studies to examine the agreement between two different techniques of quantifying the same variable (Weir, 2003). Therefore, it has been suggested that the SEM and MD are more appropriate statistical methods to categorise relative and absolute reliability (Hopkins, 2008).

$$SEM = SD\left(\sqrt{1} - ICC\right) \qquad \text{Equation 4.2}$$

$$MD = SEM \times 1.96 \times \sqrt{2} \qquad \text{Equation 4.3}$$

4.3 Results

4.3.1 Magnitude of peak segment kinetic energy

In general, the magnitude of peak total segment KE was estimated with acceptable reliability. For the Upper Body, Arms and Club segments small effect sizes and good ICCs were reported for repeated measures of peak total KE (Table 4.1, Table 4.2 and Table 4.3). Regardless of club type, the majority of magnitudes of peak translational, local rotational and remote rotational Upper Body Arms and Club kinetic energies were also measured with acceptable reliability (Table 4.1, Table 4.2 and Table 4.3).

When the driver and 5 iron were used, acceptable reliability was achieved for the measurement of peak total Lower Body KE. However, when the 9 iron was used, significantly smaller magnitudes of peak total Lower Body KE ($t(6) = 2.50$, $p = 0.04$, $d = 0.39$) were reported in test 2 (Table 4.3). Despite a good ICC (0.945), the magnitude of peak translational Lower Body KE was also significantly smaller ($t(6) = 3.02$, $p = 0.02$, d

= 0.37) in test 2 when the 5 iron was used (Table 4.2). Although the majority of peak total, translational and rotational rigid body kinetic energies were measured with high reliability, questionable reliability was reported for the repeated measures of peak total thigh KE when the 5 iron and 9 iron were used. Significantly lower magnitudes of peak total thigh KE were reported in test 2 for the 5 iron (t(6) = 3.22, $p < 0.05$, $d = 0.29$) and 9 iron (t(6) = 2.82, $p < 0.05$, $d = 0.38$). Furthermore, significantly smaller peak translational thigh KE was reported in test 2 compared to test 1 when the 5 iron (t(6) = 2.05, $p > 0.05$, $d = 0.25$) and 9 iron (t(6) = 2.584, $p < 0.05$, $d = 0.51$) were used.

Table 4.1 Reliability of the magnitude of peak segment KE when the driver was used.

	Kinetic Energy (J) ± SD					
	Test 1	Test 2	*p*	*d*	ICC	SEM (MD)
Total						
Lower Body	23.9 ± 11.4	23.9 ± 11.9	0.99	0.00	0.99	0.8 (2.3)
Upper Body	30.7 ± 4.5	32.4 ± 4.2	0.20	0.19	0.84	1.8 (4.9)
Arms	87.2 ± 19.7	89.3 ± 21.0	0.45	0.10	0.97	3.3 (9.1)
Club	269.1 ± 36.8	262.8 ± 26.2	0.38	0.20	0.92	9.0 (24.8)
Translational						
Lower Body	10.8 ± 6.1	11.6 ± 6.3	0.25	0.12	0.98	0.8 (2.3)
Upper Body	14.4 ± 4.8	15.0 ± 4.5	0.48	0.13	0.95	1.0 (2.9)
Arms	48.5 ± 12.9	48.7 ± 12.9	0.88	0.02	0.98	1.8 (4.9)
Club	224.4 ± 29.2	228.6 ± 23.7	0.41	0.16	0.94	6.5 (17.9)
Local Rotational						
Lower Body	7.3 ± 2.1	7.4 ± 2.1	0.76	0.05	0.96	4.2 (1.2)
Upper Body	16.4 ± 2.7	17.7 ± 3.0	0.13	0.43	0.85	1.1 (3.1)
Arms	4.9 ± 12.	5.1 ± 1.0	0.27	0.20	0.95	0.3 (0.7)
Club	42.2 ± 4.8	43.2 ± 4.2	0.42	0.21	0.89	1.5 (4.3)
Remote Rotational						
Lower Body	8.8 ± 3.3	8.4 ± 3.5	0.12	0.12	0.99	0.3 (0.9)
Upper Body	2.4 ± 1.2	2.7 ± 1.3	0.10	0.24	0.96	0.2 (0.6)
Arms	37.7 ± 8.2	39.9 ± 9.6	0.13	0.25	0.96	1.9 (5.3)

Table 4.2 Reliability of the magnitude of peak segment KE when the 5 iron was used.

	Kinetic Energy (J) ± SD					
	Test 1	Test 2	*p*	*d*	ICC	SEM (MD)
Total						
Lower Body	20.4 ± 4.6	19.3 ± 4.8	0.11	0.25	0.96	1.0 (2.6)
Upper Body	29.5 ± 3.7	30.1 ± 4.7	0.50	0.15	0.93	1.2 (3.2)
Arms	83.7 ± 18.5	81.8 ± 19.5	0.11	0.10	0.99	1.6 (4.4)
Club	259.2 ± 30.7	255.4 ± 27.6	0.63	0.13	0.88	10.0 (27.6)
Translational						
Lower Body	7.4 ± 3.1	6.3 ± 2.6	0.02	0.51	0.95	0.6 (1.9)
Upper Body	12.8 ± 4.9	13.1 ± 5.4	0.40	0.07	0.99	0.5 (1.5)
Arms	47.6 ± 12.4	46.1 ± 12.6	0.09	0.12	0.99	1.1 (3.1)
Club	232.0 ± 35.1	223.6 ± 25.4	0.22	0.28	0.91	8.9 (24.6)
Local Rotational						
Lower Body	6.8 ±1.3	6.8 ± 0.9	0.99	0.00	0.82	0.5 (1.3)
Upper Body	16.6 ± 1.9	17.5 ± 3.4	0.52	0.31	0.75	1.3 (3.7)
Arms	4.5 ± 1.3	5.0 ± 1.2	0.24	0.40	0.98	0.2 (0.5)
Club	39.0 ± 6.4	38.0 ± 5.6	0.15	0.16	0.98	0.9 (2.5)
Remote Rotational						
Lower Body	8.6 ± 1.5	8.8 ± 1.1	0.71	0.11	0.85	0.5 (1.4)
Upper Body	1.9 ± 1.0	1.7 ± 0.9	0.43	0.23	0.85	0.4 (1.0)
Arms	35.1 ± 7.2	34.6 ± 8.4	0.56	0.07	0.98	1.0 (2.9)

Table 4.3 Reliability of the magnitude of peak segment KE when the 9 iron was used.

	Kinetic Energy (J) ± SD					
	Test 1	Test 2	*p*	*d*	ICC	SEM (MD)
Total						
Lower Body	20.5 ± 3.6	18.0 ± 4.8	0.04	0.59	0.90	1.2 (3.6)
Upper Body	26.8 ± 3.9	26.3 ± 5.8	0.82	0.10	0.72	2.5 (7.0)
Arms	79.3 ± 8.0	78.7 ± 11.2	0.83	0.03	0.97	1.6 (4.46)
Club	231.8 ± 36.1	223.8 ± 41.3	0.30	0.20	0.98	5.2 (14.4)
Translational						
Lower Body	6.7 ± 3.9	5.7 ± 3.0	0.06	0.31	0.95	0.7 (2.0)
Upper Body	12.1 ± 3.9	11.6 ± 3.7	0.71	0.14	0.94	1.0 (2.7)
Arms	45.4 ± 13.1	44.6 ± 13.0	0.58	0.06	0.98	1.8 (5.0)
Club	194.2 ± 32.1	190.7 ± 35.9	0.31	0.10	0.98	4.3 (11.9)
Local Rotational						
Lower Body	6.8 ± 1.7	6.1 ± 1.1	0.32	0.45	0.88	0.5 (1.3)
Upper Body	14.8 ± 1.8	15.3 ± 3.5	0.65	0.20	0.81	1.2 (3.2)
Arms	4.4 ± 1.3	4.4 ± 1.0	0.97	0.01	0.84	0.5 (1.3)
Club	33.8 ± 5.4	34.1 ± 7.2	0.82	0.04	0.94	1.5 (4.2)
Remote Rotational						
Lower Body	8.6 ± 1.2	7.8 ± 1.9	0.16	0.53	0.73	0.8 (2.2)
Upper Body	1.8 ± 0.8	1.6 ± 0.5	0.50	0.31	0.71	0.3 (0.9)
Arms	34.3 ± 7.7	34.3 (8.4)	0.13	0.00	0.93	2.2 (6.0)

4.3.2 Timing of peak segment kinetic energy

The timing of peak total segment KE was measured with high reliability. For the majority of segments, acceptable ICC values and similar mean times were reported for repeated measures of peak total KE (Table 4.4, Table 4.5 and Table 4.6). For the most segments, the timing of peak translational, local rotational and remote rotational KE was also estimated with high reliability when all three clubs were used. The timing of peak translational and local rotational Club KE was also estimated with high reliability (Table 4.4, Table 4.5 and Table 4.6).

Despite similar mean times being reported in test 1 and test 2, a medium effect size (d = 0.59) was reported for the timing of peak total Lower Body KE when the driver was used (Table 4.4). Although similar mean times were also reported in test 1 and test 2 for peak total Upper Body KE another medium effect size (d = 0.68) was reported when the 9 iron was used (Table 4.6). Medium effect sizes were also reported for the timing of peak local rotational and remote rotational Upper Body KE when the 9 iron and 5 iron were used (Table 4.5 and Table 4.6). Although the timing of peak total, translational and rotational kinetic energies was also measured with acceptable reliability for the majority of rigid bodies a medium effect size (*d* = 0.57) was also reported for the timing of peak translational upper trunk KE when the driver was used.

Table 4.4 Reliability of the timing of peak segment KE when the driver was used.

	Kinetic Energy (J) ± SD					
	Test 1	Test 2	*p*	*d*	ICC	SEM (MD)
Total						
Lower Body	0.755 ± 0.097	0.812 ± 0.094	0.21	0.59	0.91	0.029 (0.079)
Upper Body	0.692 ± 0.062	0.680 ± 0.128	0.84	0.12	0.83	0.023 (0.063)
Arms	0.718 ± 0.073	0.732 ± 0.051	0.57	0.22	0.95	0.014 (0.038)
Club	0.959 ± 0.064	0.978 ± 0.013	0.50	0.41	0.92	0.011 (0.030)
Translational						
Lower Body	0.813 ± 0.102	0.849 ± 0.085	0.33	0.39	0.95	0.021 (0.059)
Upper Body	0.730 ± 0.097	0.743 ± 0.102	0.16	0.39	0.91	0.031 (0.066)
Arms	0.693 ± 0.068	0.696 ± 0.044	0.96	0.04	0.87	0.020 (0.056)
Club	0.961 ± 0.064	0.979 ± 0.013)	0.50	0.38	0.94	0.010 (0.027)
Local Rotational						
Lower Body	0.693 ± 0.030	0.691 ± 0.074	0.62	0.17	0.76	0.038 (0.097)
Upper Body	0.652 ± 0.053	0.669 ± 0.097	0.58	0.22	0.94	0.018 (0.051)
Arms	0.763 ± 0.068	0.790 ± 0.052	0.20	0.16	0.84	0.022 (0.063)
Club	0.959 ± 0.064	0.973 ± 0.016	0.60	0.31	0.98	0.006 (0.016)
Remote Rotational						
Lower Body	0.710 ± 0.172	0.688 ± 0.116	0.62	0.11	0.95	0.045 (0.126)
Upper Body	0.828 ± 0.104	0.858 ± 0.151	0.51	0.23	0.78	0.060 (0.167)
Arms	0.820 ± 0.093	0.845 ± 0.084	0.25	0.28	0.89	0.029 (0.082)

Table 4.5 Reliability of the timing of peak segment KE when the 5 iron was used.

	Kinetic Energy (J) ± SD					
	Test 1	Test 2	*p*	*d*	ICC	SEM (MD)
Total						
Lower Body	0.742 ± 0.052	0.756 ± 0.088	0.73	0.20	0.84	0.028 (0.077)
Upper Body	0.720 ± 0.037	0.706 ± 0.072	0.52	0.26	0.71	0.029 (0.081)
Arms	0.725 ± 0.045	0.760 ± 0.096	0.23	0.46	0.72	0.038 (0.104)
Club	0.989 ± 0.006	0.982 ± 0.013	0.19	0.32	0.84	0.004 (0.011)
Translational						
Lower Body	0.735 ± 0.153	0.761 ± 0.173	0.47	0.16	0.92	0.047 (0.129)
Upper Body	0.710 ± 0.057	0.712 ± 0.059	0.98	0.04	0.79	0.026 (0.073)
Arms	0.687 ± 0.028	0.692 ± 0.027	0.47	0.18	0.76	0.013 (0.037)
Club	0.990 ± 0.007	0.982 ± 0.013	0.11	0.36	0.88	0.003 (0.009)
Local Rotational						
Lower Body	0.690 ± 0.080	0.726 ± 0.075	0.34	0.47	0.85	0.030 (0.083)
Upper Body	0.707 ± 0.019	0.703 ± 0.030	0.72	0.15	0.80	0.011 (0.031)
Arms	0.816 ± 0.091	0.896 ± 0.078	0.28	0.56	0.82	0.036 (0.099)
Club	0.988 ± 0.007	0.980 ± 0.012	0.11	0.35	0.95	0.002 (0.006)
Remote Rotational						
Lower Body	0.764 ± 0.115	0.752 ± 0.140	0.31	0.09	0.99	0.016 (0.043)
Upper Body	0.860 ± 0.158	0.760 ± 0.195	0.14	0.56	0.73	0.092 (0.256)
Arms	0.814 ± 0.073	0.848 ± 0.089	0.37	0.41	0.90	0.026 (0.072)

Table 4.6 Reliability of the timing of peak segment KE when the 9 iron was used.

	Kinetic Energy (J) ± SD					
	Test 1	Test 2	*p*	*d*	ICC	SEM (MD)
Total						
Lower Body	0.759 ± 0.075	0.758 ± 0.060	*0.98*	0.01	0.79	0.031 (0.086)
Upper Body	0.746 ± 0.040	0.719 ± 0.039	0.13	0.68	0.92	0.011 (0.031)
Arms	0.739 ± 0.058	0.745 ± 0.062	0.54	0.10	0.96	0.012 (0.033)
Club	0.988 ± 0.011	0.988 ± 0.008	0.98	0.01	0.91	0.003 (0.008)
Translational						
Lower Body	0.713 ± 0.131	0.732 ± 0.164	0.70	0.13	0.88	0.052 (0.143)
Upper Body	0.729 ± 0.099	0.757 ± 0.114	0.33	0.33	0.93	0.028 (0.079)
Arms	0.693 ± 0.032	0.706 ± 0.033	0.11	0.40	0.87	0.012 (0.033)
Club	0.989 ± 0.010	0.988 ± 0.008	0.99	0.05	0.92	0.003 (0.007)
Local Rotational						
Lower Body	0.751 ± 0.071	0.741 ± 0.071	0.96	0.14	0.80	0.032 (0.087)
Upper Body	0.723 ± 0.030	0.696 ± 0.040	0.28	0.75	0.71	0.019 (0.052)
Arms	0.897 ± 0.062	0.885 ± 0.104	0.62	0.15	0.89	0.028 (0.077)
Club	0.982 ± 0.021	0.986 ± 0.011	0.76	0.21	0.98	0.002 (0.006)
Remote Rotational						
Lower Body	0.788 ± 0.108	0.786 ± 0.105	0.63	0.01	0.99	0.011 (0.031)
Upper Body	0.832 ± 0.158	0.757 ± 0.157	0.09	0.48	0.80	0.070 (0.193)
Arms	0.857 ± 0.076	0.845 ± 0.099	0.34	0.13	0.96	0.018 (0.051)

4.4 Discussion

It has been suggested that analyses of segment KE are the most appropriate way to examine the proximal-to-distal sequence in the golf swing (Anderson, Wright and Stefanyshyn, 2006; Bechard *et al.*, 2009). The sequencing of segment KE has been examined in the golf swing in two scientific studies (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.,* 2008) however, neither study reported the reliability or associated validity of measures of the magnitude or timing of peak KE. Therefore, the present study evaluated the test-retest reliability of the magnitude and timing of peak segment KE measures in the golf swing. The assessment of absolute reliability also aimed to enable subsequent analyses to determine the meaningfulness of any differences in measures of segment KE.

4.4.1 Magnitude of peak segment kinetic energy

The magnitude of peak total, translational and rotational KE was measured with high reliability for the majority of segments. However, the magnitudes of peak total and translational Lower Body kinetic energies were measured with questionable reliability when the 9 iron and 5 iron were used respectively. Despite good ICCs, significantly lower magnitudes of peak total (9 iron) and peak translational (5 iron) Lower Body KE were reported in test 2 than in test 1. Closer examination of the results indicated that these reductions were caused by significant decreases in the magnitude of peak translational thigh KE produced in test 2 when these clubs were used.

When there are statistically significant differences between the means of repeated tests such as those reported for the magnitude of peak total and translation Lower Body KE, good ICCs can be reported if there is a large effect and the error term is small (Weir, 2005). Although only medium effect sizes were reported for these measures, the effect of test day was consistent as smaller peak total and translational Lower Body kinetic energies were generated in test 2 by 6 of the 7 golfers. The associated errors (SEM) could also be considered small as they were similar to those reported when the other clubs were used. In this scenario, it is possible that systematic variability in the data collection protocol can explain the significant differences (McGraw and Wong, 1996; Weir, 2005).

Variability in the magnitude of peak Lower Body KE was most likely caused by systematic differences in the identification of anatomical landmarks which define the thigh and the subsequent effect on the definition of the local coordinate systems and estimation of inertial parameters. This suggestion is consistent with kinematic studies of other motor tasks, where landmark identification errors were considered to be the key factor in decreased measurement repeatability (Ferber *et al.,* 2002; Mills et al., 2007; McGinley et al., 2009). Inconsistencies in the measurement of pelvis forward bend velocity in the golf swing have also been associated with variation in anatomical landmark identification between test–retest conditions (Evans *et al.,* 2012). As mean estimates of thigh mass and MOI were similar in the two tests (Table 4.7) Lower Body

KE variability was most likely caused by systematic differences in the anatomical landmarks which define the thigh co-ordinate system. Further research is required to confirm this assumption and identify the specific anatomical landmarks which cause variability to be produced in measures of peak Lower Body KE.

Table 4.7 Comparison of thigh inertial parameters estimated in test 1 and test 2.

	Mass (kg)		Ixx (kg/m^2)		Iyy (kg/m^2)		Izz (kg/m^2)	
Participant	Test 1	Test 2	Test 1	Test 2	Test 1	Test 2	Test 1	Test 2
1	12.04	12.62	0.1930	0.2065	0.0599	0.0655	0.1925	0.2027
2	9.71	10.46	0.1541	0.1816	0.0376	0.0415	0.1553	0.1832
3	10.98	10.37	0.1760	0.1537	0.0466	0.0454	0.1778	0.1531
4	11.09	10.57	0.1814	0.1710	0.0489	0.0430	0.1799	0.1718
5	10.27	10.48	0.1746	0.1771	0.0401	0.0418	0.1762	0.1782
6	11.05	11.08	0.2308	0.2419	0.0411	0.0424	0.2409	0.2416
7	11.40	10.80	0.1993	0.1837	0.0545	0.0497	0.1992	0.1842
Mean	10.93	10.91	0.1870	0.1879	0.0470	0.0470	0.1888	0.1878
SD	0.75	0.79	0.0241	0.0286	0.0081	0.0086	0.0271	0.0278

It has been suggested that golf swings of less skilled players are affected by movement variability during the latter stages of the downswing (Bradshaw *et al.*, 2009; Cheetham *et al.* 2008; Penner, 2003). Therefore, it is possible that the reduced magnitudes of peak Lower Body KE in test 2 were caused by changes in the technique used by the golfers (Drust *et al.,* 2005). However, it has been reported that participants of varying skill level (handicap range +2 – 14 strokes) are able to closely replicate their kinematics in multiple repeated tests (Evans *et al.,* 2012). Furthermore, as the magnitude of peak Lower body kinetic energies were consistently smaller in test 2 it is much more likely that systematic differences in the identification of the anatomical landmarks which define the thigh co-ordinate system were responsible.

4.4.2 Timing of peak segment kinetic energy

For the majority of segments and rigid bodies, the timing of peak segment KE was measured with high reliability as similar mean times, low effect sizes and good ICCs were reported. As changes in the timing of peak segment KE are primarily caused by changes in the measurement of translational and rotational velocities these results support the notion that electromagnetic tracking systems are capable of measuring 3D movements with acceptable reliability (An *et al.*, 1998; Evans *et al.*, 2012; Horan *et al.*, 2010). However, a medium effect size was reported for measures of the timing of peak total Lower Body KE when the driver was used. Medium effect sizes were also reported for the timing of peak total (9 iron), local rotational (9 iron) and remote rotational (5 iron) Upper Body KE. The medium effect sizes suggest that these measures were made with questionable reliability. However, other reliability indices suggested that acceptable reliability was achieved. Acceptable-good ICC were reported for these measures and the measures of absolute reliability (SEM and MD) reported for the timing were smaller than those reported for the same measures with the other clubs. Despite this support for the reliability of these measures, it is recommended that future studies consider the SEM and MD when interpreting the results of analyses of segment KE.

4.4.3 Limitations

Although support has been presented for the reliability of segment KE measures in the golf swing some limitations of this study should be noted. Recruitment for the study was limited by the strict inclusion criteria and requirement to attend the laboratory on two separate occasions. Therefore, the study analysed a limited sample of seven mixed ability subjects. Although this sample is reflective of golfers used in this programme of research it is likely that the measures of absolute reliability reported in this study may be conservative for a group of less skilled players. It has been suggested that golfers of varying skilled level were able to closely replicate kinematics in repeated tests (Evans *et al.,* 2012). However, it is unclear whether a group of less skilled players who produce more variable golf swings (Bradshaw *et al.,* 2009; Cheetham *et al.,* 2008) would also produce reliable measures of golf swing kinematics.

4.5 Conclusion

The magnitude and timing of peak total, translational and rotational KE was measured with high reliability for the majority of segments. The similar mean values, acceptable-good ICCs and low SEMs provided support for the examination of the proximal-to-distal sequence using analyses of segment KE. However, the magnitude of peak total and translational Lower Body KE was measured with questionable reliability when the 9 iron and 5 iron were used respectively. This variability was most likely associated with the repeated identification of the anatomical landmarks especially for the thigh. The implication of this finding is that when the magnitude of peak segment KE is compared at least part of the observed difference in segment KE may be attributable to sources of variability unrelated to the golf swing.

5. Chapter V - Sequencing of Segment Kinetic Energy in the Golf Swing

5.1. Introduction

The outcome of any golf swing, the flight of the ball, is extremely sensitive to small variations in the timing, tempo and rhythm of the golf swing (Jagacinski, Greenberg and Liao, 1997). Players and coaches often consider these factors to be subjective and kinaesthetic (Jagacinski, Greenberg and Liao, 1997). However, in many striking and throwing activities it has been suggested that a proximal-to-distal sequence of body segment motions enables greater distal segment velocities to be achieved (Morehouse and Cooper, 1950; Putnam, 1993; Hirashima *et al.* 2007; Nunome *et al.*, 2007).

The proximal-to-distal sequence has frequently been linked with the mechanical advantages of the kinetic link principle (Ferdinands, Kersting and Marshall, 2012; Kreighbaum and Barthels, 1985; Putnam, 1993) and the muscular advantages of the SSC (Finni et la., 2003; Lees, 2010; Nicol, Avela, and Komi, 2006; Welch *et al.*, 1995). These theoretical advantages provide an explanation for analyses of sequential body segment motions becoming an increasingly popular theme in biomechanical golf swing research (Mesiter *et al.*, 2012; Tinmark *et al.*, 2010; Vena *et al.*, 2011a).

In the golf swing literature, segmental sequencing has predominantly been examined in terms of the summation of speed principle using analyses of segment angular velocities (Neal *et al.*, 2007; Tinmark *et al.*, 2010). However, numerous techniques of varying complexity have also been used ranging from the calculation of segment rotation velocity from the relative angle between two one-dimensional lines (Myers, *et al.,* 2008) to the calculation of segment angular velocity from a non-stationary instantaneous screw axis (Vena *et al.*, 2011a). Regardless of technique, the majority of these analyses have suggested that, for skilled performers, the magnitude of peak angular velocity increases sequentially from the most proximal to the most distal segments (Cheetham *et al.*, 2008; Neal *et al.*, 2007; Tinmark *et al.*, 2010). Less conclusive evidence has been provided regarding the timing of peak segment angular velocity. Whilst it has been reported that timing conformed to a proximal-to-distal sequence (Neal *et al.*, 2007; Tinmark *et al.*, 2010), research has also suggested that the timing of peak angular velocities followed a subject specific sequence (Cheetham *et al.*, 2008; Vena *et al.*, 2011b).

Despite the large volume of research which has examined the sequencing of body segment angular velocities in the golf swing, there is still little agreement regarding the most appropriate analysis technique. The sequencing of segment KE is another analysis technique which is becoming increasingly popular in scientific studies as a means of examining the effectiveness of movement patterns (Bechard, 2009; Ferdinands, Kersting and Marshall, 2012; Slawinski *et al.*, 2010). Furthermore, Anderson, Wright and Stefanyshyn (2006) proposed that analyses of segment KE are the most appropriate to assess the sequencing of body segments in the golf swing. Unlike analyses of segment angular velocities, analyses of segment KE consider the inertial properties of body segments and are independent of direction (Bechard *et al.*, 2009). Segment KE is also calculated using the square of translational and rotational velocities which makes it much more sensitive to subtle changes in technique compared with analyses of velocity (Anderson, Wright and Stefanyshyn, 2006; Bechard *et al.*, 2009).

Skilled golf performance has frequently been linked with the concept of energy transfer (Hume; Keogh and Reid 2005; Cheetham *et al.*, 2008; Chu, Sell and Lephart 2010). It is thought that the successful transfer of energy from proximal to distal segments can lead to increased clubhead velocities being produced (Ferdinands, Kersting and Marshall, 2012; Kreighbaum and Barthels, 1985). However, the term energy transfer has sometimes been referred to when analyses of segment angular velocity sequencing have actually been performed.

The transfer of energy can be examined by analysing the energy changes of the body and its constituent segments or by computing the flow of energy using a joint by joint approach (Wells, 1988; Winter, 1987). Using the joint energy approach, Nesbit and Serrano, (2005) suggested that the generation of work is mostly a bottom-up phenomenon. The effectiveness of the joint energy approach is dependent on the accuracy of the biomechanical model and the complex estimation of the net joint forces and muscle moments (Betzler *et al.*, 2008). Due to the difficultly associated with these measurements, the joint energy approach has had limited use and researchers have tended towards the simpler segment energy approach (Wells, 1988).

Analyses of segment KE can be used to give an indication of mechanical energy flow through a linked system (Wells, 1988). If one segment loses energy and neighbouring segment gains energy, it is likely that this change resulted from energy transfer (Williams and Cavanagh 1983). However, it is not be possible to obtain a direct measure of mechanical energy flow as changes in segment energy could also occur because of muscular activity and segment interactions (Metzler, 2002; Wells, 1999). Despite this limitation, it has been suggested that the most comprehensive quantification of segmental sequencing is in terms of segment KE (Stefanyshyn, 1996).

Analyses of segment KE have suggested that the magnitude of peak total segment KE increased sequentially from proximal to distal segments (Anderson, 2007; Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008). These analyses also suggested that the timing of peak total segment KE did not follow the same sequential pattern. Instead, the results provided support for the principle of optimal coordination of partial momenta (Van Gheluwe & Hebblinck, 1985) as peak KE of three human based segments (lower body, upper body and arms) occurred simultaneously at approximately 80% of the downswing movement time. The Club was the only segment which demonstrated a delay in the timing as peak Club KE occurred significantly later just prior to impact (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008).

Although the sequencing of segment KE in the golf swing has been examined, a comprehensive understanding has yet to be formulated as the majority of studies have only discussed the sequencing of total segment KE (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008). Analyses of segment KE also enable the translational and rotational components of total KE to be considered separately and as a single entity (Bechard *et al.*, 2009). It has been suggested that an examination of the sequencing of the translational and rotational components of segment KE could provide a more detailed description of the complex movement patterns associated with the golf swing (Anderson, 2007). Analyses of these components can provide information regarding the origin of total KE and give a better indication of the movements that produce it (Bechard *et al.,* 2009). However, these components have only been considered with golf swing performed with a driver (Anderson, 2007).

The effect of club type on the sequencing of segment KE has yet to be comprehensively established. Kenny, *et al.* (2008) suggested that the same golf swing was used with a Driver and a 7 iron. However, this finding was based on the analysis of only one highly skilled player. A more detailed analysis of their results suggested that peak Club and Hip kinetic energies were lower when the 7 iron was used compared with the driver (Table 5.1). The reduction in Club KE would be expected based on the smaller radius of rotation with the 7 iron however, the reason for the reduced hip KE with the 7 iron is not clear.

Table 5.1 Kinetic energy values for the driver and 7 iron (Kenny *et al.,* 2008).

Segment	Driver KE (J)	7 Iron KE (J)
Club	219	174
Arms	136	141
Torso	34	33
Hip	24	13

A long held belief by many authors, professional golfers and golf teachers is that there is no ideal golf swing, but all golfers must have one, highly repeatable swing (Budney and Bellow, 1982; Richards *et al.*, 1985). It has been suggested that identical temporal characteristics of the upper limbs are evident in downswings with a driver, 5 iron and pitching wedge (Egret *et al.*, 2003; Neal, Abernethy and Moran, 1990). Further support for this belief was provided in an analysis of segment angular velocities as it was reported that a proximal-to-distal sequence is a common movement characteristic in golf swings with a driver, 5 iron and in sub-maximal golf swings with a pitching wedge (Tinmark *et al.*, 2010). The results of this analysis indicated that maximum angular velocity increased from the most proximal to the most distal segment in every shot condition and that the order in which these peaks occurred conformed to the summation of speed principle.

Despite this support for the similarity of golf swings made using different clubs many studies have also suggested that the physical characteristics of golf clubs may change the underlying movements of the golf swing (Budney and Bellow, 1982; Lindsey, Horton and Paley, 2002). The reduced shaft length of irons and wedges compared to the driver has been reported to shorten the length of the backswing and increase the inclination of the swing plane (Budney and Bellow, 1982). The shorter iron clubs have also been associated with an increase in the magnitude of forward flexion and left-side bending of the trunk (Lindsey, Horton and Paley, 2002). Other differences between golf swings with different clubs have been related to the additional distance requirement and clubhead speed required when longer clubs are used (Egret *et al.*, 2003). It has been reported that the speed of the arms (Nagao and Sawada, 1973), the magnitude of shoulder joint rotation (Egret *et al.*, 2003) and the rotation velocity of the proximal segments were all significantly larger when a driver was used compared to shorted irons and wedges.

These conflicting findings regarding the effect of club type clearly suggest that more research is needed to enhance the understanding of changes in golf club type and physical characteristics on the sequencing of body segment movements during the golf swing. Furthermore, due to the advantages associated with the analysis of segment KE it is hypothesised that using this technique would enable a better understanding to be determined. Therefore, the aim of this study was to examine the effect of club type on the sequencing of segment KE in the golf swings of highly skilled players.

5.1.1. Hypotheses

H_{NULL}: Club type will have no effect on the magnitude or timing of peak segment KE in the golf swing.

$H_{ALTERNATIVE\ 1}$: As the length of the club and distance requirement of the golf shots increase the magnitude of KE produced for each segment will also increase.

$H_{ALETERNATIVE\ 2}$: As the length of the club and distance requirement of the golf shots increases peak KE for each segment will occur later in the downswing.

5.2. Methods

5.2.1. Participants

Twenty one skilled male golfers (M ± SD; age 35 ± 12 years, stature 184.7 ± 4.1 cm, mass 86.7 ± 12.8 kg and handicap 3.4 ± 3.9 strokes, range +3 - 8 strokes) volunteered to take part in this study. It has been suggested that sample size planning is important in scientific studies as it can assure an adequate power to detect statistical significance (Muthen and Muthen, 2002. However, it is also a practical reality that sample size planning can be extremely difficult as suitable quantitative information such as the desired effect size and power is not always available prior to the study commencing (Venth, 2001). Therefore, following the recommendations of Venth (2001), sample sizes in all studies were reflective of those used in previous golf swing research (Neal *et al.,* 2007: Tinmark *et al.,* 2010).

For analysis purposes, professional golfers without a registered handicap were considered to have a handicap of 0 and plus handicaps (e.g. +3) were considered to be negative numbers. At the time of testing each golfer was required to be playing or practicing at least once a week. Furthermore, golfers were excluded if their registered handicap was greater than 8 or if they had an injury which was preventing them from regularly playing or practising golf. Ethics approval was obtained from the Faculty of Health and Wellbeing Research Ethics Committee and written informed consent was obtained from each golfer.

5.2.2. Inertial parameters

Inertial parameters for the feet, lower legs, thighs, lower trunk, middle trunk, upper trunk, neck, head, upper arms and forearms were estimated using the geometric model described in section 3.2. Hand inertial parameters were estimated using a stadium solid and segment of a hollow cylinder (Section 3.5). The anthropometric measurements required to form the shapes of the geometric model were made using the Polhemus electromagnetic tracking system, electromagnetic sensors and custom written software (Section 3.2). The club segment geometry and inertial parameters were based on measurements made by a non-contact laser scanner (Section 3.4.4).

5.2.3. Data collection

Data were collected following the protocol described in section 3.4.2 which required participants to hit fifteen shots from an artificial mat into a net five meters away, five with a driver, 5 iron and 9 iron.

5.2.4. Segment kinetic energy

Translational and rotational velocity data were obtained for each rigid body using the Polhemus Liberty electromagnetic tracking system (Polhemus, Inc., Colchester, VT, USA) sampling at 240 Hz and the technique described in section 3.4. Then, for each golf swing, the magnitude and timing of peak KE was calculated for each rigid body, for three body segments (Lower, Upper Body and Arms), and for the Club segment using the technique described in section 3.5. The timing of peak segment KE was then normalised to total movement time (MT) in the downswing where 0 represents the top of backswing and 1 represents impact.

5.2.5. Statistical analysis

A two-way repeated measures analysis of variance (ANOVA) was employed to determine the effect of club type and segment on the magnitude and timing of peak segment KE. Follow-up planned contrasts were performed to establish the origin of any significant differences. It has been suggested that by using univariate analyses, planned contrasts enable the pattern of responses for the independent variables of a factorial ANOVA to be identified (Schneiner and Gurevitch, 2001) without increasing the likelihood of a type II error (Vincent and Weir, 2012). Furthermore, planned contrasts have frequently been used in golf swing analysis to identify differences in body segment movements (Evans *et al.*, 2012; Horan *et al.*, 2010; Tinmark *et al.* 2010). In this study, the pre-defined planned contrasts were designed to identify the origin of any significant effects of club type and segment on the magnitude and timing of segment KE. All of these statistical analyses were conducted using SPSS (Version 19.0) and alpha was set at $p < 0.05$.

5.3. Results

5.3.1. Total segment kinetic energy

There was a significant interaction between club type and segment ($F(6,120) = 29.25$, $p < 0.01$) which indicated that club type had a significant effect on the magnitude of the difference between peak total Upper Body and Arms kinetic energies. This significant interaction was caused by the magnitude of peak Arms KE with the driver (99.2 ± 11.2 J) being significantly greater ($p < 0.05$) than that generated with the 9 iron (91.0 ± 16.8 J) - Figure 5.1. The interaction also indicated that the difference between peak total Arms and Club KE was greater when the Driver and 5 iron were used compared with the 9 iron. This effect was caused by the magnitude of peak total Club KE being significantly greater ($p = 0.01$) when the driver (267.8 ± 38.0 J) and 5 iron (261.3 ± 26.6 J) were used compared with the 9 iron (231.0 ± 24.3 J) - Figure 5.1.

There was a significant main effect of segment on the magnitude of peak total KE ($F(1.47, 29.35) = 844.10$, $p < 0.01$) which suggested that peak total KE increased sequentially from the Lower Body to the Club. There was also a significant main effect of club type ($F(1.57, 31.42) = 36.65$, $p < 0.01$) which indicated that regardless of segment, the magnitude of peak total KE increased from the 9 iron to the 5 iron ($F(1,20) = 51.53$, $p < 0.01$, $r = 0.85$) and from the 5 iron to the driver ($F(1, 20) = 9.00$, $p = 0.01$, $r = 0.56$).

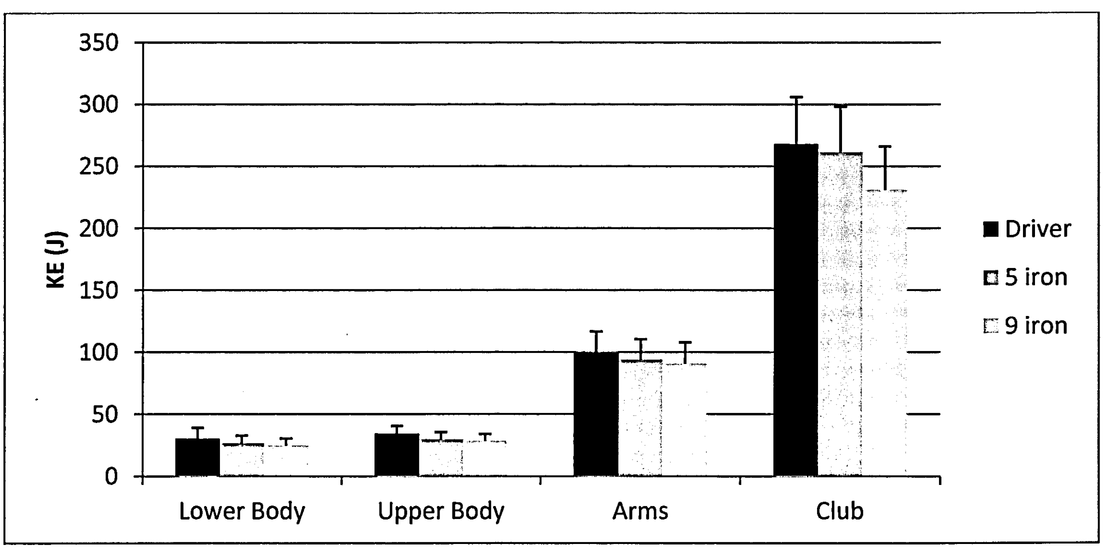


Figure 5.1 Peak total segment KE for all 3 clubs. Standard deviation is shown using vertical black bars.

There was no interaction (club type x segment) or main effect of club type for the timing of peak total segment KE. However, there was a significant main effect of segment on the timing of peak total KE ($F(1.89,37.87) = 46.32$, $p < 0.01$). The planned contrasts indicated that peak total Lower Body, Upper Body and Arms KE peaked simultaneously between 0.71 and 0.76 MT. However, with all three clubs, peak total Club KE occurred significantly later than peak total Arms KE ($F(1,21) = 342.68$, $p < 0.01$, $r = 0.97$) - Figure 5.2.

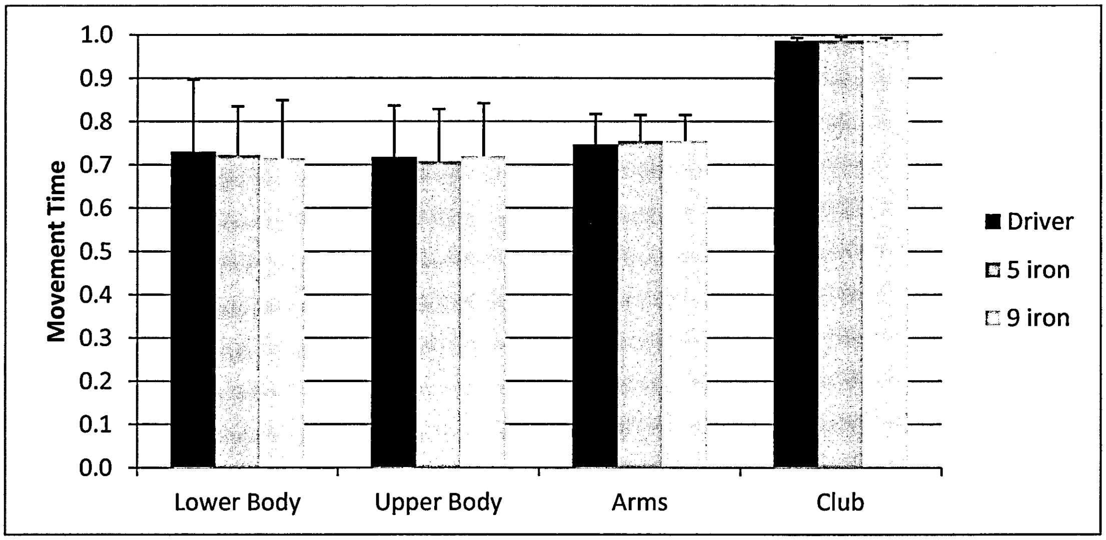


Figure 5.2 The timing of peak total segment KE.

For each segment the relative contribution of each component of KE was different (Figure 5.3). Peak Club total KE was composed of mainly translational KE whilst the Upper Body was largely composed of local rotational KE and total Lower Body and Arms KE were composed of a combination of two or three components of KE. The relative magnitude and timing of peak KE for each component was more closely examined in the following sections.

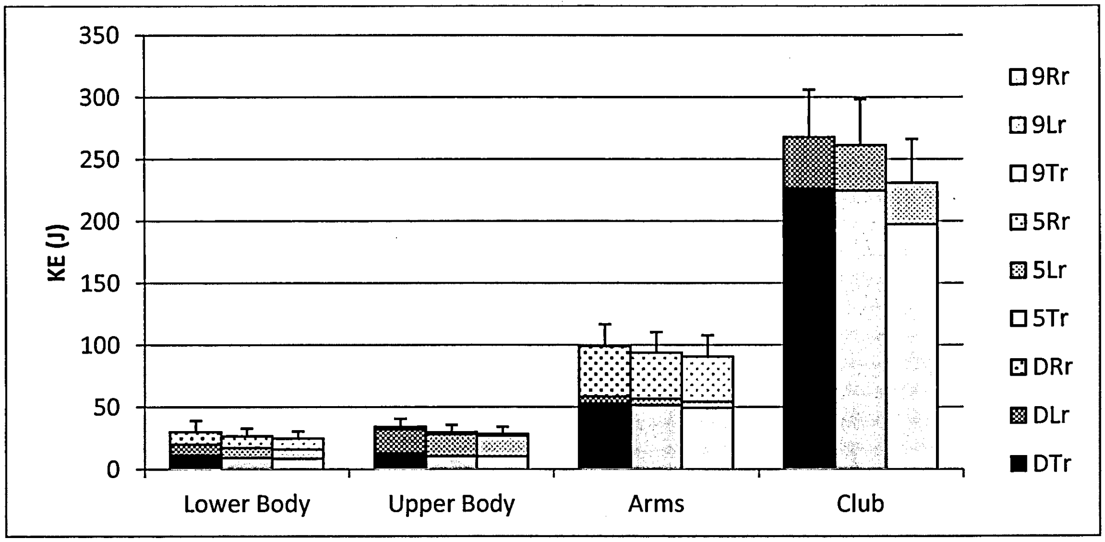


Figure 5.3 Total segment KE as the sum of its components.

5.3.2. Translational segment kinetic energy

There was a significant interaction (club type x segment) for the magnitude of peak translational segment KE (F(2.15,43.07) = 28.99, $p < 0.01$). This interaction suggested that the magnitude of peak translational Club KE was significantly greater ($p < 0.01$) when the driver (226.8 ± 32.0 J) and 5 iron (225.0 ± 30.0 J) were used compared with the 9 iron (198.3 ± 30.2 J) - Figure 5.4. The planned contrasts also indicated that the magnitude of peak translational Arms KE was significantly greater ($p = 0.05$) when the driver (57.2 ± 9.6 J) was used compared with the 9 iron (50.8 ± 9.2 J) - Figure 5.4.

The magnitude of peak translational segment KE increased in a proximal-to-distal manner from the Lower Body to the Club when all 3 clubs were used. However, the main effect of segment (F(1.29, 25.84) = 1020.35, $p < 0.01$) only indicated that peak translational KE increased significantly from the Upper Body to the Arms and from the Arms to the Club. The difference between peak translational Lower Body and Upper Body KE was not significant (f(1,20) = 4.23, $p = 0.53$, r = 0.42) (Figure 5.4). There was also a significant effect of club type (F (2, 40) = 29.11, $p < 0.01$) which suggested that peak translational segment KE was significantly larger when the 5 iron was used compared with the 9 iron (F(1,20 = 47.29, $p < 0.01$, r = 0.84).

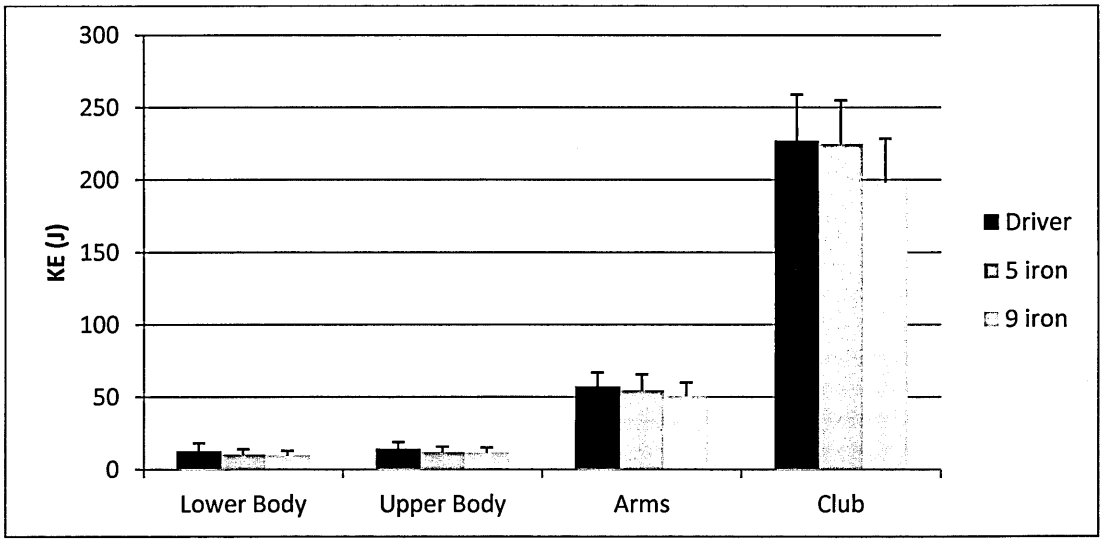


Figure 5.4 Peak translational segment KE.

There was a significant interaction (club type x segment) for the timing of peak translational segment KE (F(6,120) = 2.88, $p = 0.04$). This interaction was caused by peak Upper Body translational KE occurring significantly later ($p < 0.05$) with the driver (0.67 ± 0.17 MT) compared with the 5 iron (0.58 ± 0.16 MT) and 9 iron (0.60 ± 0.16 MT) - Figure 5.5. There was also a main effect of segment (F(3,60) = 32.54, $p < 0.01$) on the timing of peak translational KE which suggested that peak Lower Body and Upper Body translational KE occurred at the same time, whilst peak Upper Body, Arms and Club occurred in a proximal-to-distal order (Figure 5.5).

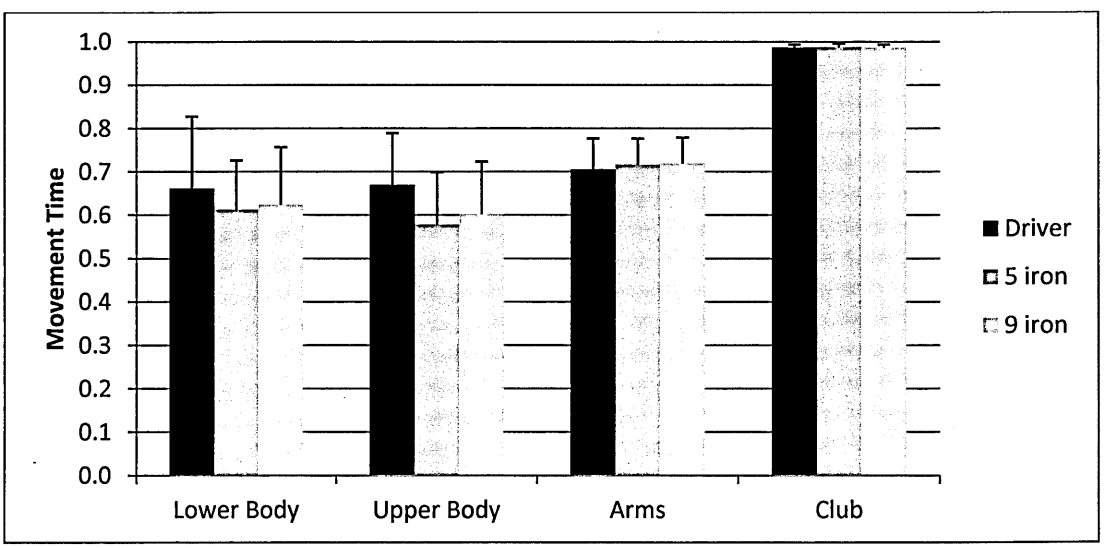


Figure 5.5 Timing of peak translational segment KE.

5.3.3. Local rotational segment kinetic energy

There was a significant interaction (club type x segment) for the magnitude of peak local rotational segment KE ($F(2.69,53.82) = 28.92$, $p < 0.01$). This indicated that peak local rotational Upper Body KE was significantly ($p < 0.05$) greater when the driver was used (20.9 ± 4.6 J) compared with the 5 iron (18.3 ± 4.7 J) (Figure 5.6). This interaction also indicated that the magnitude of peak local rotational Club KE increased from the 9 iron (33.3 ± 6.2 J) to the 5 iron (36.6 ± 7.3 J) and to the driver (41.1 ± 6.9 J) - $p < 0.05$.

The magnitude of peak local rotational KE did not follow a proximal-to-distal sequence as peak Upper Body KE was greater than peak local rotational Arms KE (Figure 5.6). However, there was a main effect of segment ($F(1.67,33.37) = 291.57$), $p < 0.01$) which suggested that the magnitude of peak local rotational KE increased from the Arms to the Lower Body, from the Lower Body to the Upper Body and from the Upper Body to the Club (Figure 5.6). There was also a main effect of club type ($F(2,42) = 45.50$, $p < 0.05$) which suggested that the magnitude of peak local rotational KE increased from the 9 iron to the 5 iron ($F(1,20) = 26.16$, $p < 0.01$, $r = 0.78$) and from the 5 iron to the driver ($F(1,20) = 33.89$, $p < 0.01$, $r = 0.79$).

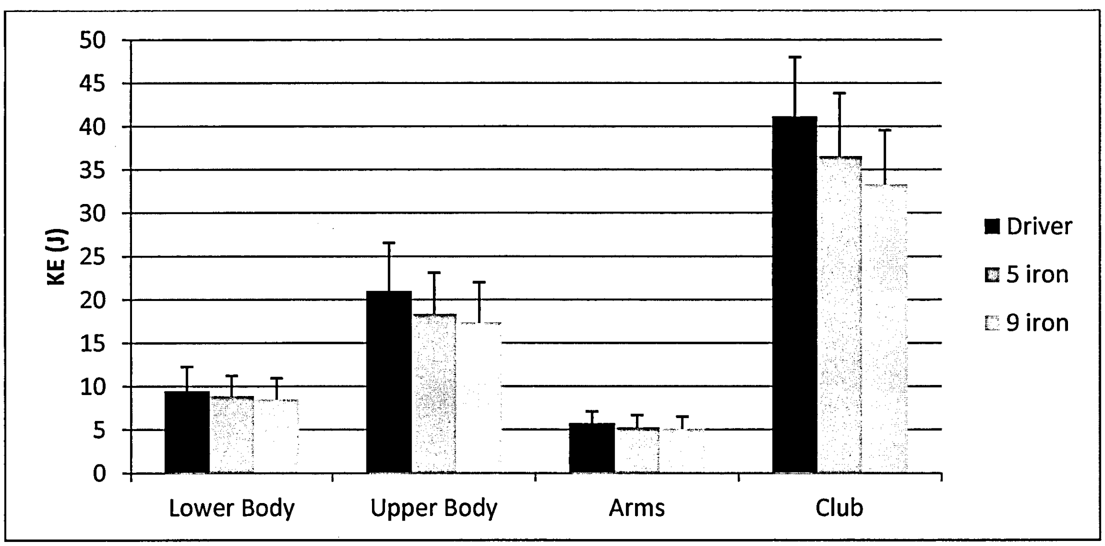


Figure 5.6 Peak local rotational segment KE.

There was a significant interaction (club type x segment) for the timing of peak local rotational KE ($F(2.85,57.08) = 4.00$, $p < 0.01$). As peak local rotational Upper Body KE occurred at approximately the same time with all clubs this interaction indicated that peak local rotational Arms KE occurred significantly earlier ($p < 0.05$) when the driver was used (0.82 ± 0.13 MT) compared with the 5 iron (0.88 ± 0.11 MT) and 9 iron (0.89 ± 0.11 MT) - Figure 5.7. The main effect of segment ($F(3,60) = 53.94$, $p < 0.01$) and planned contrasts also suggested that peak local rotational KE increased sequentially from the Upper Body to the Club.

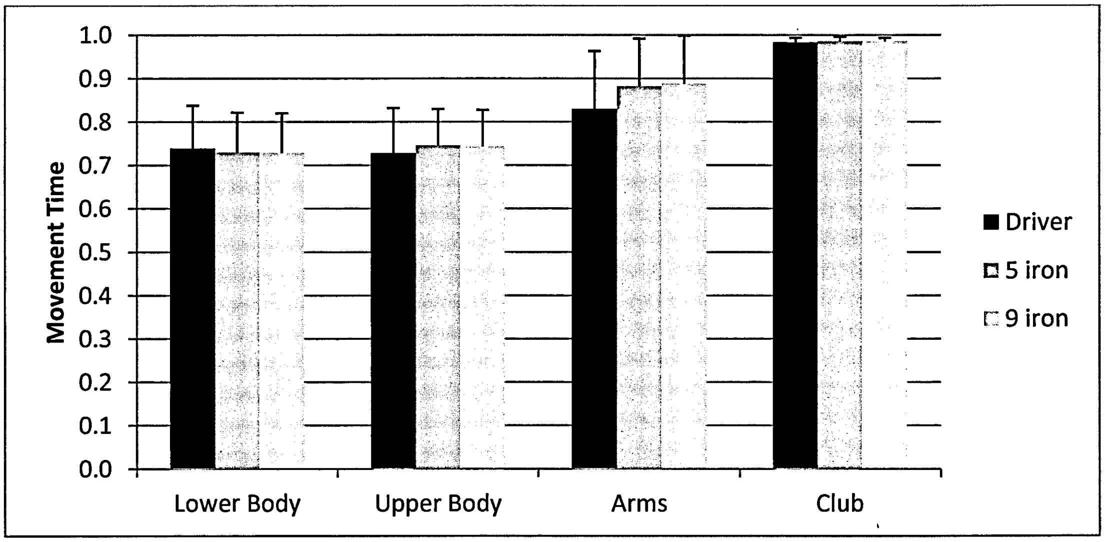


Figure 5.7 The timing of peak local rotational segment KE.

5.3.4. Remote rotational segment kinetic energy

As the Club segment was considered to be a rigid body it did not generate any remote rotational KE (Figure 5.8). There was a significant interaction between club type and segment ($F(2.11, 42.19) = 8.89$, $p < 0.01$). As peak Upper Body remote rotational KE was similar when the driver, 5 iron and 9 iron were used this interaction suggested that peak remote rotational Arms KE was significantly greater ($p < 0.05$) when the driver (42.8 ± 9.0 J) was used compared with the 5 iron (39.4 ± 8.0 J) and 9 iron (38.8 ± 8.4 J) - Figure 5.8.

There was a main effect of segment ($F(1.20,24.07) = 405.90$, $p < 0.01$) which indicated that the magnitude of peak remote rotational Arms KE was significantly larger than peak remote rotational Upper Body KE ($F(1,20) = 468.63$, $p < 0.01$, r = 0.99) which was also significantly smaller than peak remote rotational Lower Body KE ($F(1,20) = 168 .89$, $p < 0.01$, r = 0.95) (Figure 5.8). Furthermore, the significant main effect of club type

($F(2,40) = 12.61$, $p < 0.05$) suggested that the magnitude of peak remote rotational KE increased from the 5 iron to the driver ($F(1,20) = 11.53$, $p < 0.01$, $r = 0.60$) - Figure 5.8.

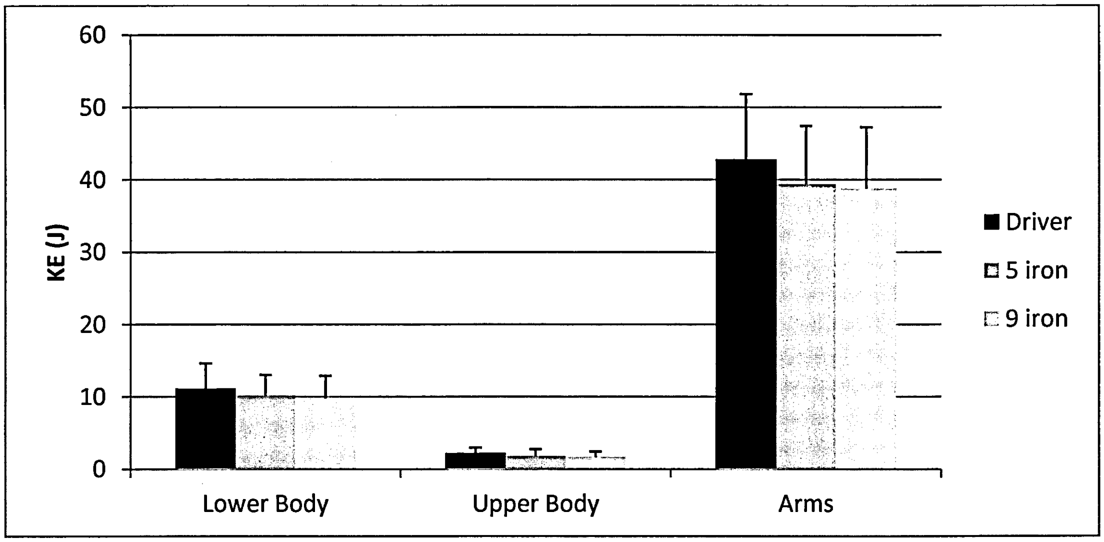


Figure 5.8 Peak remote rotational segment KE.

There was no interaction (club type x segment) or main effect of club type on the timing of peak remote rotational segment KE (Figure 5.9). However, there was a main effect of segment ($F(2,40) = 8.42$, $p < 0.05$) which suggested that peak remote rotational Upper Body KE occurred significantly earlier than peak remote rotational Arms KE ($F(2,20) = 9.72$, $p < 0.01$, $r = 0.57$). As the results suggested that peak Lower Body and Upper Body remote rotational KE occurred at the same time it is reasonable to assume that peak Lower Body remote rotational KE also occurred significantly earlier than peak remote rotational Arms KE.

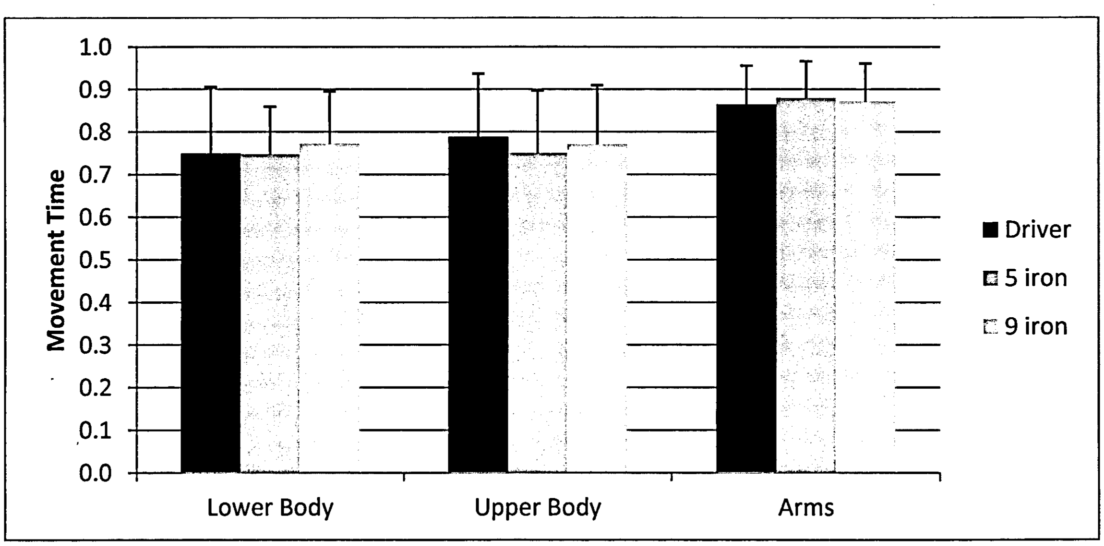


Figure 5.9 The timing of peak remote rotational segment KE.

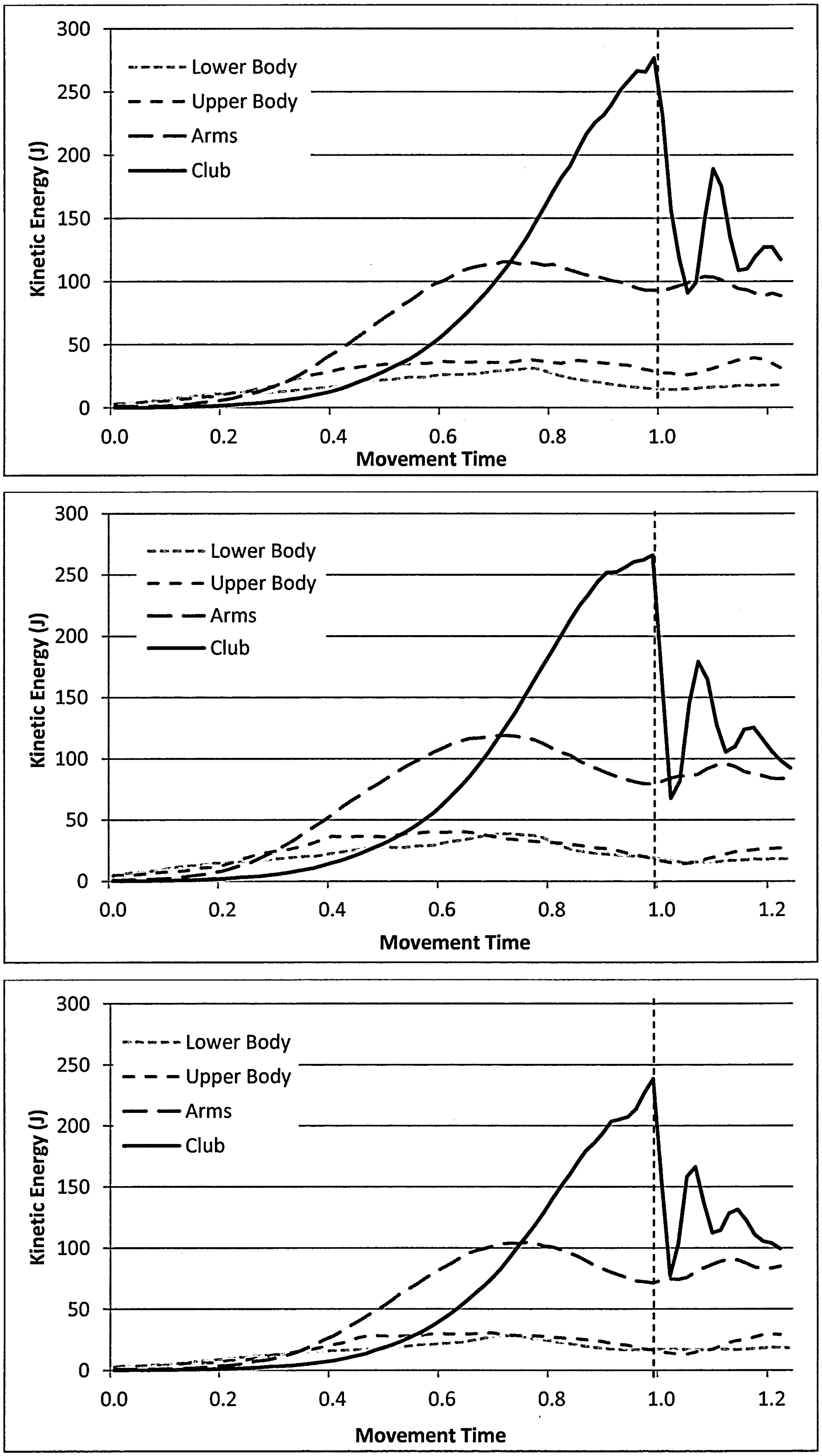


Figure 5.10 Total kinetic energy for player 5. Time is taken from the Top-of-Backswing to Mid-Follow-Through (impact is highlighted). Top: driver, Middle: 5 iron, Bottom: 9 iron.

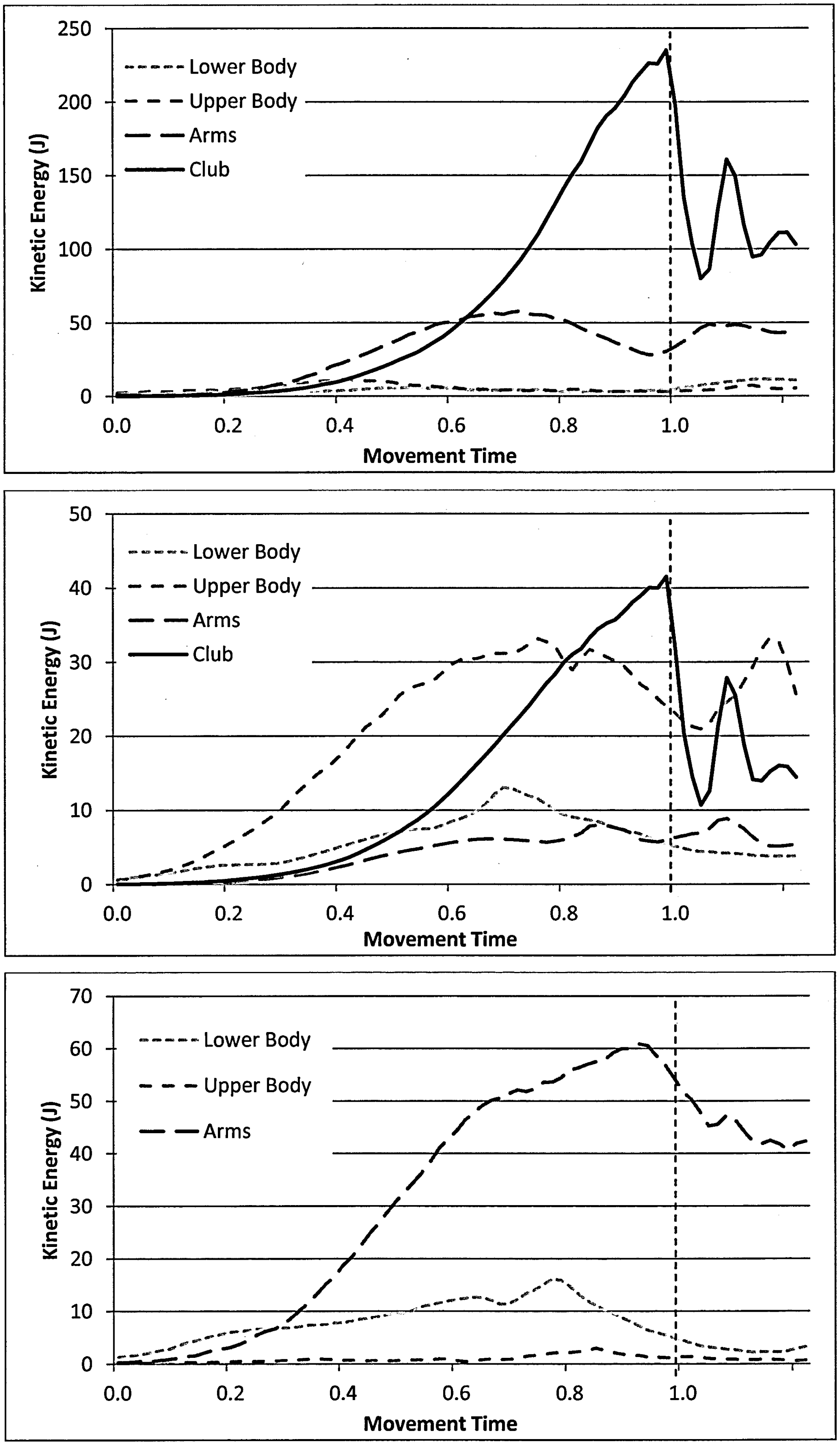


Figure 5.11 Kinetic energy components for player 5 with the driver. Time is taken from the Top-of-Backswing to Mid-Follow-Through (impact is highlighted). Top: Translational, Middle: Local Rotational, Bottom: Remote Rotational.

5.4. Discussion

The purpose of this study was to determine if, as in many other sports (Feltner and Dapena, 1989; Hirashima *et al.*, 2008; Mero *et al.*, 1994; Nunome *et al.*, 2006), a proximal-to-distal sequence of body segment movements underpins skilled golf swings. The existence of a proximal-to-distal sequence was examined by describing the magnitude and timing of peak segment KE. Using this analysis technique, this study also aimed to enhance understanding of the sequencing of body segment movements in the golf swing by examining the effect of club type and the sequencing of each component of segment KE.

The results of this study agree with those presented by Anderson, Wright and Stefanyshyn (2006) and Kenny *et al.* (2008) as, when the driver was used, the magnitude of peak total segment KE increased sequentially from the proximal to the distal segments. Additionally, this sequence was also evident when the 5 and 9 irons were used. Research has previously suggested that regardless of club type the magnitude of segment angular velocities increased in a proximal-to-distal order for highly skilled players (Tinmark *et al.*, 2010). However, the sequential increase in peak total segment KE for the 5 and 9 irons had not previously been identified. This sequential increase in the magnitude of peak total segment suggests that KE could be transferred through the linked system (Well, 1998). It also suggests that golfers might do additional work on distal segments at each joint. Nesbit and Serrano (2005) also suggested that in golf swings with a driver the lumbar spine, shoulders and wrists all do positive work in the downswing. It could be argued that accuracy, rather than distance is the desired outcome of golf swings with the 5 and 9 irons (Tinmark *et al.* 2010). Therefore, these results indicate that a sequential increase in the magnitude of total segment KE is a fundamental feature of skilled golf swings which focus on accuracy as well as those which focus on distance.

The analysis of the magnitude of peak total segment KE provided support for a proximal-to-distal sequence underpinning skilled golf performance. The magnitude of peak translational KE also increased sequentially from the Upper Body to the Club.

However, similar magnitudes of peak translational Lower Body and Upper Body KE were reported. A possible explanation for the absence of an increase in peak translational KE from the Lower Body to the Upper Body is that highly skilled players produce similar magnitudes of peak translational pelvis and upper torso velocities (Horan *et al.*, 2008). The lateral movements of these segments have been linked with skilled golf performance as they cause the golfer's weight to transfer in the direction of the shot during the downswing (Beak *et al.*, 2013) and subsequently enable increased clubhead speeds to be produced (Burden, Grimshaw and Wallace, 1998; Okuda *et al.*, 2010).

Despite similar magnitudes of peak translational Lower and Upper Body KE being produced, the magnitude of peak local rotational Upper Body KE was significantly greater than that of the Lower Body. As it has frequently been suggested that significantly larger magnitudes of peak upper trunk angular velocities are produced compared with the pelvis (Burden, Grimshaw and Wallace, 1998; Horan *et al.*, 2010; Myers *et al.*, 2008) this sequential pattern was expected. However, local rotational KE did not peak in a sequential manner as peak Upper Body KE was also significantly greater than peak local rotational Arms KE. This raises concerns about the appropriateness of analyses of segment angular velocity which have attempted to characterise segmental sequencing in the golf swing. The majority of these analyses have suggested that skilled golf swings are underpinned by proximal-to-distal sequences of segment angular velocities (Cheetham *et al.*, 2008; Neal *et al.*, 2007; Tinmark *et al.*, 2010). However, as translational movements accounted for approximately half of the total KE produced by each segment (Figure 5.4) the results of this study suggest that by only examining the rotational movements of body segments these studies did not account for a large proportion of golfer's movements.

As well as examining the sequencing of segment KE, the aim of this study was also to assess the effect of club type on the magnitude and timing of segment KE. Club type had a significant effect on the magnitude of peak total KE and each of its components as for each segment peak KE increased sequentially from the 9 iron to the driver. Previous analyses of segment KE (Kenny *et al.*, 2008) had reported that golf swings with a 7 iron were as 'forceful' as those with a driver. Despite this suggestion, a closer examination of the results reported by Kenny *et al.* (2008) suggested that higher total Hips and Club KE were produced with the driver (Table 5.1). Therefore, combined with the results of this study which also used more highly skilled golfers, these findings provide evidence that as the distance requirement of the golf shot increases, more energetic golf swings are produced.

An examination of the club type and segment interactions provided further insight into the differences between golf swings with different clubs. These interactions indicated that, compared with the driver and 5 iron, lower magnitudes of peak total Arms and Club KE were evident when the 9 iron was used. The result for the Club was expected due to the larger radius of rotation for the driver and 5 iron. The results for the Arms were also expected as it has been suggested that the maximum angular speed of the arms is higher when the driver and 5 iron are used compared with a pitching wedge (Tinmark *et al.*, 2010) and 9 iron (Nagao and Sawada, 1973). However, in this study, the larger magnitudes of peak total Arms KE generated when the driver and 5 iron were used were related to larger magnitudes of peak translational Arms KE. The club type and segment interactions also suggested that peak local rotational Upper Body KE was larger when the driver was used compared with the 5 iron. Therefore, these results indicate that golfers should attempt to increase the translational KE of the Arms segments and the local rotational KE of the Upper Body when the distance requirement of the golf shot increases.

With all three clubs, total segment KE did not peak in a proximal-to-distal order. Similar to the findings presented in previous examinations of segment KE, total body segment (LB, UB and Arms) KE peaked simultaneously at approximately 74% MT whilst total Club KE peaked just before impact (Anderson, 2007; Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008). If energy were being transferred sequentially between body segments, peak distal segment KE should have occurred later in the downswing than peak segment KE for their proximal neighbours. The only segment that showed a delay in peak total KE was the Club. This suggests that the wrist was the only joint to permit efficient energy transfer between neighbouring segments and supports the notion that the golfer may be effectively represented by a two segment or double pendulum model comprising of the body and the club (Anderson, 2007). However, the analysis of the timing of peak segment KE components provided a different perspective.

Although the components of peak Lower Body and Upper Body KE occurred at approximately the same time, it is possible that sequential translational and rotational body segment movements are characteristics of good golf swings. With all three clubs the timing of peak translational, local rotational and remote rotational KE occurred in a proximal-to-distal sequence from the Upper Body to the Club. A similar sequence of rotational body movements was reported by Tinmark *et al.*, (2010) who suggested that the upper torso, hand and club followed a sequential pattern of angular rotations. Zheng *et al.*, (2008) also suggested that the shoulder joint attained maximum angular speed prior to the wrist and Club. Therefore, although peak total segment KE conformed to the principle of optimal coordination of partial momenta (Van Ghelwe and Hebblinck, 1986) these results indicate that highly skilled golf swings are underpinned by sequential increases in translational and rotational components of segment KE from the Upper Body to the Club.

A sequential increase in segment KE components was not evident from the Lower Body to the Upper Body. Although this could provide support for the principle of optimal coordination of partial momenta, this simultaneous peaking of Lower Body and Upper Body KE could be explained by the Lower Body being composed of the legs and the lower trunk. It has been suggested that the legs and lower trunk perform different roles in the golf swing with the legs tending to become static support for the rotation of their distal counterpart (Nesbit and Serrano, 2005) which initiates the downswing (Cheetham *et al.*, 2001). Therefore, with this in mind, there is a need for further research to examine the sequencing of segment KE using a separated Lower Body segment. If the Lower Body was separated into a Legs and Lower Trunk (Pelvis) segment, although a sequence of segment KE magnitude would not be produced due to the lower mass of the Pelvis compared to the Legs it is possible that the timing of peak total KE for the Legs, Pelvis and Upper Body would occur in a proximal-to-distal sequence.

Despite peak Lower Body and Upper Body KE occurring simultaneously the interactions reported in this study suggested that peak translational Lower Body and Upper Body KE occurred significantly later when the driver was used compared with the 5 iron and 9 iron. The lateral movement of the hips and upper torso during the downswing has been suggested to be a mechanism used by skilled golfers to deliver maximum energy to the clubhead at impact (Beak *et al.*, 2013). Furthermore, the results of this study suggest that when a driver is used a strategy employed by highly skilled players to increase clubhead velocity may be to delay the lateral shift of the lower and upper trunk segments and subsequent peaks in translational Legs and Upper Body KE. To assess the validity of this finding, further research is required to compare the timing of highly skilled golfer's swings with those of less skilled players. Investigating other differences in sequencing of segment KE between players of different abilities would also enable a better understanding of how to improve golf performance to be formulated (Lindsay, Mantrop and Vandervoort 2009).

5.5. Conclusion

In conclusion, a proximal-to-distal sequence was evident in the magnitude of peak total segment KE when all three clubs were used. However, the magnitude of peak segment KE for each component did not occur in a proximal-to-distal order as the results highlighted the importance of producing translational Legs and local rotational Upper Body KE. Club type also had a significant effect on the magnitude of peak total KE as more energetic golf swings were produced with the driver. More specifically, skilled golfers produced larger magnitudes of translational Arms KE and local rotational Upper Body KE when the distance requirement of the shot increased.

With all three clubs, the timing of peak total KE did not occur in a proximal-to-distal sequence which suggested that the timing of peak total KE conformed to the principle of optimal coordination of partial momenta. However, the results suggested that this regardless of club type, highly skilled golf swings were underpinned by sequential translational and rotational movements from the Upper Body to the club. Finally, the results suggested that when the driver was used and the distance requirement of the shot increased, peak translational Lower Body and Upper Body KE occurred significantly later in the swings of highly skilled golfers.

6. Chapter VI - The Effect of Playing Standard on the Sequencing of Segment Kinetic Energy in the Golf Swing

6.1. Introduction

Golf is a sport enjoyed by people of all ages and skill. While simple in its purpose, the golf swing can be extremely difficult to master. It can appear effortless and graceful in the hands of an expert but awkward and jerky for the novice (Bradshaw *et al.*, 2008). The golf swing has been recognised as one of the most difficult biomechanical motions in sport to execute and master (Lai *et al.*, 2011). During the swing the golfer forms an open kinetic chain with the feet at the closed end, the clubhead at the open end and several body segments in between. The complexity of the golf swing is often associated with the necessity to coordinate these body segments to produce controlled clubhead speed and accurate ball contacts at the open end of the system (Cheetham *et al.*, 2008).

Chapter V explored the movement patterns used by highly skilled players to produce successful golf swings. Although an understanding of the sequencing of segment KE was developed, only highly skilled players were considered. When complex motor actions such as the golf swing are executed by players of lower ability it has been suggested that alterations are made in the sequencing of body segment motions (Kenny *et al.*, 2008). It has also been suggested subtle variations exist in the movement patterns exhibited by professional and amateur golfers (Horan *et al.*, 2010). As segment KE is calculated using the square of translational and rotational velocities this form of analysis is sensitive to subtle changes in technique (Bechard *et al.*, 2009). Therefore, using an analysis of segment KE may enable these subtle differences in technique between players of varying ability to be identified and subsequently highlight mechanisms which may allow golfers to improve their performance and assist coaches in developing more playing standard specific coaching programmes (Vaughn, 1979; Zheng *et al.*, 2008).

Analysis techniques based on segment and joint kinematics (Cheetham *et al.*, 2008; Zheng *et al.*, 2008), rotational speeds (Meister *et al.*, 2011; Myers *et al.*, 2008) and energy transfer through the principle of work (Nesbit and Serrano, 2006) have been used to examine the effect of playing standard on movement patterns in the golf swing. Using an analysis of segment kinematics, Cheetham *et al.*, (2008) suggested that, for professional and amateur golfers, the magnitude and timing of peak segment angular velocity of the linked system (pelvis, thorax arms and club) followed a proximal-to-distal sequence. Despite this suggestion, other research has indicated that numerous differences are apparent between the swings of high and low skilled golfers. For example, it has been suggested that less skilled golfers accelerate and release their club early in the downswing (Robinson, 1994) which causes peak clubhead angular velocity to occur before ball impact (Cooper and Mather, 1994). This swing deficiency exhibited by less skilled players has since been attributed to the earlier release of the arms in the downswing and subsequent early production of peak angular velocities of the arm segments (Zheng *et al.*, 2008).

Biomechanical golf swing analyses have identified numerous other differences between the movement patterns of skilled and less skilled players. This research has suggested that skilled golfers produce larger clubhead velocities (Cheetham *et al.*, 2008; Meister *et al.*, 2011; Nesbit and Serrano 2005; Zheng *et al.* 2008) as a consequence of producing faster (Cheetham *et al.*, 2008; Zheng *et al.*, 2008) and more consistent (Bradshaw *et al.*, 2010; Sanders and Owen, 1992) movements. Numerous studies have also reported that the magnitude of the separation between the upper torso and pelvis at the top of the backswing (x-factor) can differentiate between players of varying ability. It has been suggested that greater separation between the pelvis and upper torso is produced by professional golfers compared with amateurs (Cheetham *et al.*, 2000; Zheng *et al.*, 2008), by golfers who produce high ball velocities (Myers *et al.* 2008) and by golfers with greater driving distance (McLean, 1992). An increased x-factor has frequently been associated with the increased rotational velocity of these segments produced by skilled players during the downswing (Callaway *et al.*, 2012; Myers *et al.*, 2008). Furthermore, research has suggested that unlike amateur players, skilled golfers utilise sequential rotations of the pelvis and upper trunk (McTeigue *et al.*, 1994; Burden, Grimshaw and Wallace, 1998; Myers *et al.*, 2008; Robinson, 1994; Zheng *et al.*, 2008) and increased forward translational velocity of the upper torso (Sanders and Owen, 1992) to maximise the speed of the clubhead at impact.

Despite these differences being reported between the swings of high and low skilled golfers, a comprehensive understanding of the effect of playing standard on the movement patterns utilised in the golf swing has yet to be formulated. The majority of the previously discussed studies compared golfers of varying ability by calculating the segment angular velocities of the pelvis and upper torso segments when only the driver was used (Burden, Grimshaw and Wallace, 1998; Myers *et al.*, 2008). These studies determined the effect of playing standard by examining the angular velocity of body segments (Cheetham *et al.*, 2008) and joints (Zheng *et al.* 2008) and the translational velocity of the central hub (Sanders and Owen, 1992). However, they did not consider the combined translation and rotation of body segments, the inertial parameters of each segment or the movements of the whole body. Therefore, the advantages associated with the analysis of segment KE (Chapter V) combined with the use of multiple clubs could enable a thorough appreciation of the effect of playing standard on segmental movement sequences in the golf swing to be determined.

Analyses of energy have previously been used to examine the effect of playing standard on the sequencing of segmental motions in the golf swing. Nesbit and Serrano (2005) suggested that larger joint toques and more efficient energy transfer were produced by skilled players. However, Nesbit and Serrano (2005) only examined four golfers, one of each playing standard and therefore, their findings could be considered to be subject specific and may not reflect the movement patterns of a larger group of golfers. Other studies and Chapter V of this thesis, which considered the sequencing of segment KE, only analysed the swings of highly skilled golfers (Anderson, Wright and Stefanyshn, 2006; Kenny *et al.*, 2008). The results of Chapter V suggested that when the distance requirement of a golf shot is great and a driver is used numerous mechanisms such as: increasing the magnitudes of rotational Upper Body and translational Arms KE, delaying the peak in translational Lower Body and Upper Body KE and sequential timing of peak segment KE components from the Upper Body to the Club were utilised by highly skilled players. However, the mechanisms which differentiate players of varying ability have yet to be determined. Therefore, the purpose of this study was to examine the effect of playing standard on the sequencing of segment motions in the golf swing.

6.1.1. Hypotheses

H_{NULL}: Playing standard will have no effect on the magnitude or timing of peak segment KE in the golf swing.

$H_{ALTERNATIVE\ 1}$: Skilled golfers will produce significantly higher magnitudes of segment KE compared with less skilled golfers.

$H_{ALTERNATIVE\ 2}$: Lower Body and Upper Body translational kinetic energies will occur later in the downswing for highly skilled players.

$H_{ALTERNATIVE\ 3}$: Skilled golfers will exhibit sequential timing of peak segment KE components from the Upper Body to the Club whilst less skilled players will not.

6.2. Methods

6.2.1. Participants

Thirty three male golfers (M ± SD; age 36 ± 13 years, stature 184.6 ± 4.4 cm, mass 87.2 ± 11.3 kg and handicap 8.1 ± 6.5 strokes, range +3 - 20 strokes) volunteered to take part in this study. Playing standard was determined using each players registered handicap and the categories used in the CONGU unified handicapping system. The three CONGU categories (Table 6.1) each contained 11 players and, for categorisation and analysis purposes, professional golfers without a registered handicap were considered to have a handicap of 0 and plus handicaps (e.g. +3) were considered to be negative numbers. At the time of testing each golfer was required to be playing or practicing at least once a week. Furthermore, golfers were excluded if their registered handicap was greater than 20 or if they had an injury which was preventing them from regularly playing or practising golf. Ethics approval was obtained from the Faculty of Health and Wellbeing Research Ethics Committee and written informed consent was obtained from each golfer.

The mean attributes for each CONGU category were compared using multiple one-way ANOVAs ($\alpha = 0.05$) and Tukey's *post-hoc* multiple comparison test was used to assess which categories were significantly different from each other. The results suggested that the three CONGU categories were matched in terms of age, mass and stature.

Table 6.1 Participant details for each handicap category.

Category	Handicap Range (strokes)	Age (years)	Mass (kg)	Stature (cm)	Handicap (strokes)
1	0 – 5	34 ± 13	87.7 ± 16.7	185.1 ± 4.0	0.6 ± 1.3
2	6 – 12	41 ± 11	88.3 ± 7.6	185.2 ± 4.8	8.2 ± 2.0
3	13 - 20	37 ± 12	85.6 ± 8.0	183.6 ± 4.6	15.5 ± 2.7

6.2.2. Inertial parameters

Inertial parameters for the feet, lower legs, thighs, lower trunk, middle trunk, upper trunk, neck, head, upper arms and forearms were estimated using the geometric model described in section 3.2. Hand inertial parameters were estimated using the geometric modelling technique described in section 3.5. The anthropometric measurements required to form the shapes of the geometric model were made using the Polhemus electromagnetic tracking system, electromagnetic sensors and custom written software (Section 3.2). The club segment geometry and inertial parameters were based on measurements made by a non-contact laser scanner (Section 3.4.4).

6.2.3. Data collection

Data were collected following the protocol described in section 3.4.2 which required participants to hit fifteen shots from an artificial mat into a net five meters away, five with a driver, 5 iron and 9 iron.

6.2.4. Segment kinetic energy

Translational and rotational velocity data were obtained for each rigid body using the Polhemus Liberty electromagnetic tracking system (Polhemus, Inc., Colchester, VT, USA) sampling at 240 Hz and the technique described in section 3.4. Then, for each golf swing, the magnitude and timing of peak KE was calculated for each rigid body, for three body segments (Lower, Upper Body and Arms), and for the Club segment using the technique described in section 3.5. The timing of peak segment KE was then normalised to total movement time (MT) in the downswing where 0 represents the top of backswing and 1 represents impact.

6.2.5. Statistical analysis

All of the statistical analyses were conducted using SPSS (Version 19.0). Three-way mixed design analyses of variances (ANOVA) were employed to determine the effect of playing standard, club type and segment on the magnitude and timing of peak segment KE ($\alpha = 0.05$). Follow-up planned contrasts were performed to further examine significant main effects or significant interactions. By using univariate analyses, these planned contrasts enable the pattern of response for each independent variable of a factorial ANOVA to be identified (Schneiner and Gurevitch, 2001) without increasing the likelihood of a type II error (Vincent and Weir, 2012). Furthermore, Tukey's honest significant difference (HSD) was calculated to identify the specific differences that were responsible for the significant interactions (Equation 6.1) (Vincent and Weir, 2012). Tukey's HSD calculates the minimum difference that must be evident between the raw score means to declare a significant difference between two groups in an interaction:

$$HSD = q_{(k,df_E)}\sqrt{\frac{MS_E}{n}} \qquad \text{Equation 6.1}$$

where q is the value from the Studentized range distribution for k (number of groups) and df_E at a given confidence level (Vincent and Weir, 2012), MS_E is the mean square error from the main ANOVA analysis and n is the number of the groups.

6.3. Results

Comprehensive results which examine the effect of playing standard on the magnitude and timing of peak segment KE are presented in Table 6.2 and Table 6.3.

6.3.1. Total segment kinetic energy

The interaction between club type, segment and playing standard was non-significant (F (4.56, 68.45) = 1.27, p = 0.29, r = 0.14). However, there was a significant interaction between segment and playing standard (F(3.16,47.43) = 6.34, p < 0.01, r = 0.34) as category 1 (99.93 ± 19.86 J) golfers produced significantly higher magnitudes (p = 0.01) of peak total Arms KE than category 3 players (74.14 ± 8.16 J) (Figure 6.1a). The interaction also identified that category 1 golfers produced significantly higher magnitudes (p < 0.01) of peak total Club KE than category 2 and 3 golfers (Figure 6.1b).

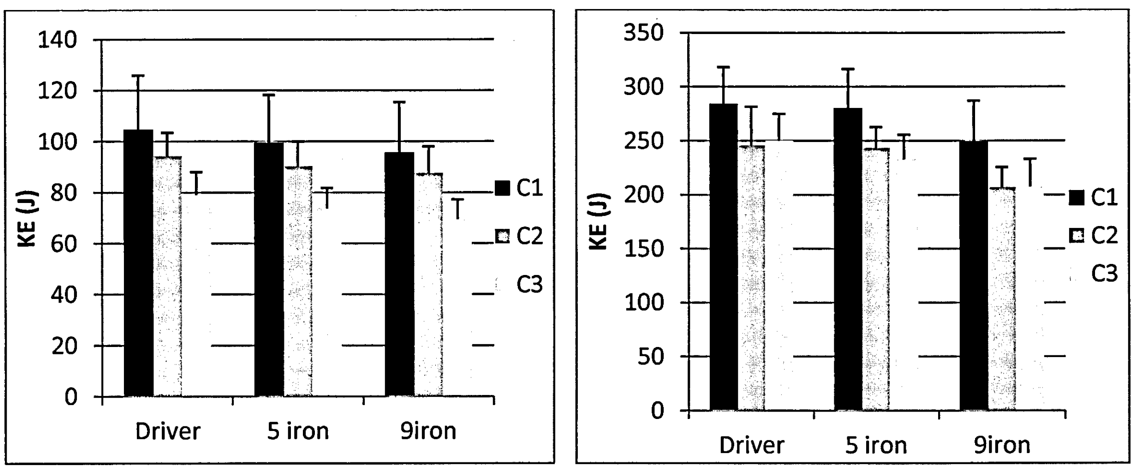


Figure 6.1 (a) left: Mean magnitude of peak total Arms KE. (b) right: Mean magnitude of peak total Club KE. Standard deviations are shown using vertical black bars.
C1 – Category 1 golfers, C2 – Category 2 golfers, C3 – Category 3 golfers.

There was a significant main effect of playing standard on the magnitude of peak total segment KE (F(2,30) = 10.71, p < 0.01, r = 0.51). The *post-hoc* tests indicated that regardless of club type category 1 players produced significantly higher magnitudes of peak total segment KE than category 2 and 3 players. Furthermore, there was a significant main effect of club type (F(1.62,48.62) = 74.27, p < 0.01, r = 0.78) which suggested that regardless of segment, the magnitude of peak total KE increased sequentially from the 9 iron to the driver. As the interaction between club type and playing standard was non-significant the results indicated that the magnitude of peak total KE increased sequentially from the 9 iron to the driver for all three playing standards (Table 6.2).

The interaction between club type, segment and playing standard was non-significant for the timing of peak total segment KE ($F(5.10, 76.45 = 0.47$, $p = 0.81$, $r = 0.08$). There was also no effect of playing standard on the timing of peak total segment KE ($F(2, 30) = 2.68$, $p = 0.09$, $r = 0.29$). However, there was a significant effect of segment ($F(2.27, 68.12) = 76.40$, $p < 0.01$, $r = 0.73$) which suggested that peak total Lower Body, Upper Body and Arms KE occurred simultaneously whilst peak total Club KE occurred significantly later ($F(1, 30) = 371.34$, $p < 0.01$, $r = 0.96$) (Table 6.3). As the interaction between segment and playing standard was non-significant ($F(4.09, 61.32) = 0.59$, $p = 0.67$, $r = 0.10$) this timing sequence was evident for all thee playing standards (Table 6.3).

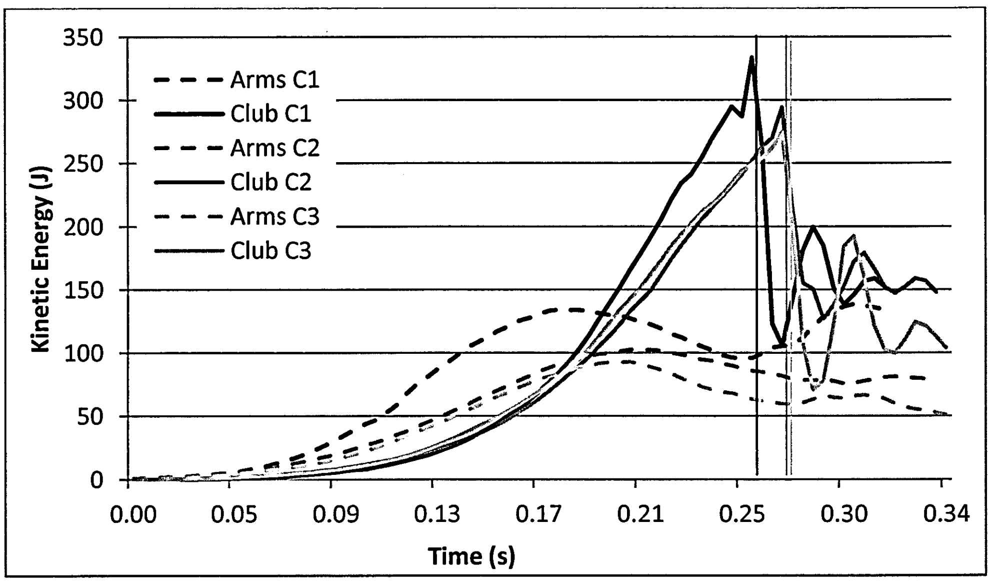
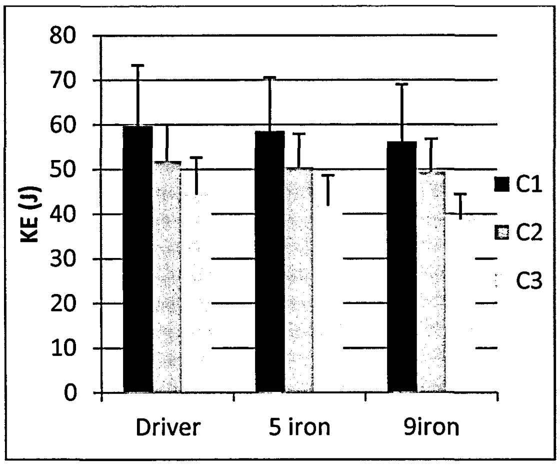


Figure 6.2 Total Arms and Club KE for a representative player from each playing standard using the driver from Top-of-Backswing to Mid-Follow-Through (impact is highlighted for each golfer).

6.3.2. Translational segment kinetic energy

There was a significant interaction between segment and playing standard ($F(2.83, 42.39) = 6.52$, $p < 0.01$, $r = 0.37$) which indicated that category 1 (58.12 ± 12.83 J) golfers produced significantly greater magnitudes ($p < 0.05$) of peak translational Arms KE than category 3 players (41.73 ± 6.95 J) - Figure 6.3a. The significant interaction was also related to the significantly higher ($p < 0.01$) magnitudes of peak translational Club KE produced by category 1 (231.17 ± 31.07 J) golfers than by category 2 (198.97 ± 20.70 J) and 3 (196.75 ± 21.44 J) golfers (Figure 6.3; Figure 6.5).

There was a significant main effect of playing standard on the magnitude of peak translational KE ($F(2,30) = 8.83$, $p < 0.01$, $r = 0.48$). The *post-hoc* tests indicated that category 1 players produced significantly larger magnitudes of peak translational segment KE than category 2 and 3 players. There was also a significant main effect of segment ($F(1.41,42.39) = 2227.6$, $p < 0.01$, $r = 0.99$) which suggested that regardless of playing standard the magnitude of peak translational KE increased sequentially from the most proximal to the most distal segment.



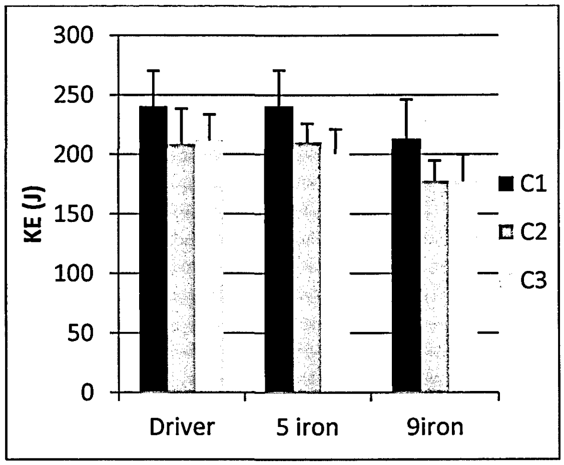


Figure 6.3 (a) left: Mean magnitude of peak translational Arms KE. (b) right: Mean magnitude of peak translational Club KE.

There was a significant interaction (segment x playing standard) for the timing of peak translational segment KE ($F(6, 90) = 3.78$, $p < 0.01$, $r = 0.30$). This interaction was related to peak translational Upper Body KE occurring significantly earlier ($p < 0.01$) in the downswing for category 1 (0.616 ± 0.083 MT) and 2 golfers (0.676 ± 0.108 MT) than for category 3 players (0.847 ± 0.098 MT) (Figure 6.4; Figure 6.5).

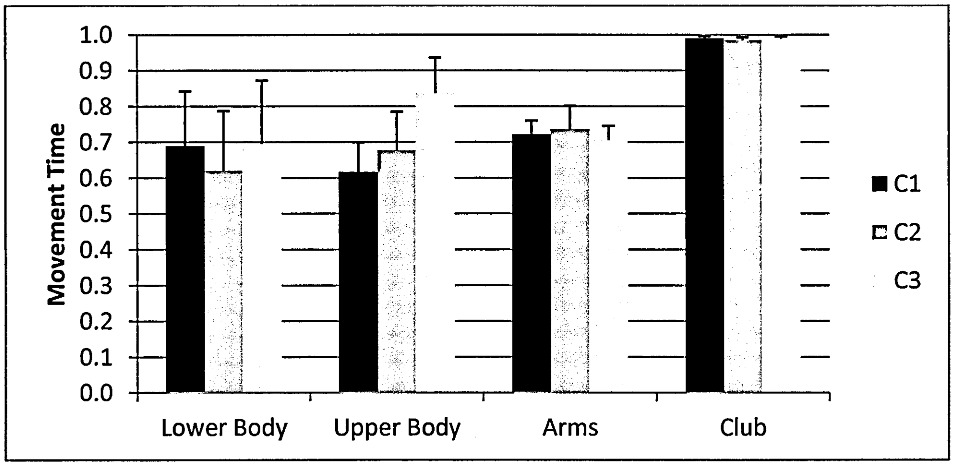


Figure 6.4 Timing of mean translational segment kinetic energies regardless of club type.

There were also significant main effects of segment ($F(3, 90) = 52.395$, $p < 0.01$, $r = 0.61$) and playing standard ($F(2,30) = 4.14$, $p = 0.04$) on the timing of peak translational KE. The main effect of segment indicated that peak translational Lower Body, Upper Body and Arms KE occurred at the same time whilst peak translational Club KE occurred significantly later in the downswing. The main effect of playing standard suggested that peak translational segment KE occurred earlier for category 1 players compared with category 3 players. When interpreted together, these statistical analyses indicate that different translational KE timing profiles were used by each category. For category 1 golfers peak Lower Body and Upper Body KE occurred at the same time followed sequentially by peak Arms and Club. For category 2 golfers mean peak translational Lower Body, Upper Body and Arms KE occurred at approximately the same time followed by peak translational Club KE. Finally, for category 3 players, mean peak translational Upper Body KE occurred significantly later in the downswing compared with mean peak translational Lower Body and Arms KE (Figure 6.4; Table 6.3).

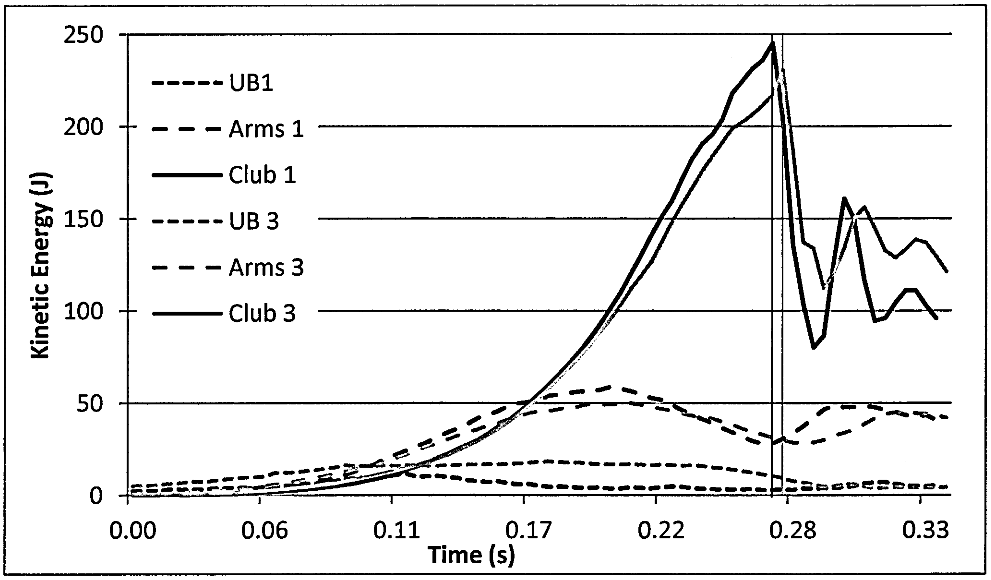


Figure 6.5 Translational UB, Arms and Club KE for a representative category 1 and category 3 player using the driver from Top-of-Backswing to Mid-Follow-Through (impact is highlighted for both players).

6.3.3. Local rotational segment kinetic energy

The interaction between segment and playing standard was significant (F(3.84, 57.56) = 3.29, p = 0.02, r = 0.23). This interaction indicated that category 1 golfers, regardless of club type produced significantly larger magnitudes ($p < 0.05$) of peak local rotational Upper Body KE (20.10 ± 5.8 J) compared with category 3 golfers (15.72 ± 4.00 J) (Figure 6.6a). This interaction was also related to category 1 golfers producing significantly larger magnitudes ($p < 0.01$) of peak local rotational Club KE than category 3 golfers (Figure 6.6b; Figure 6.8).

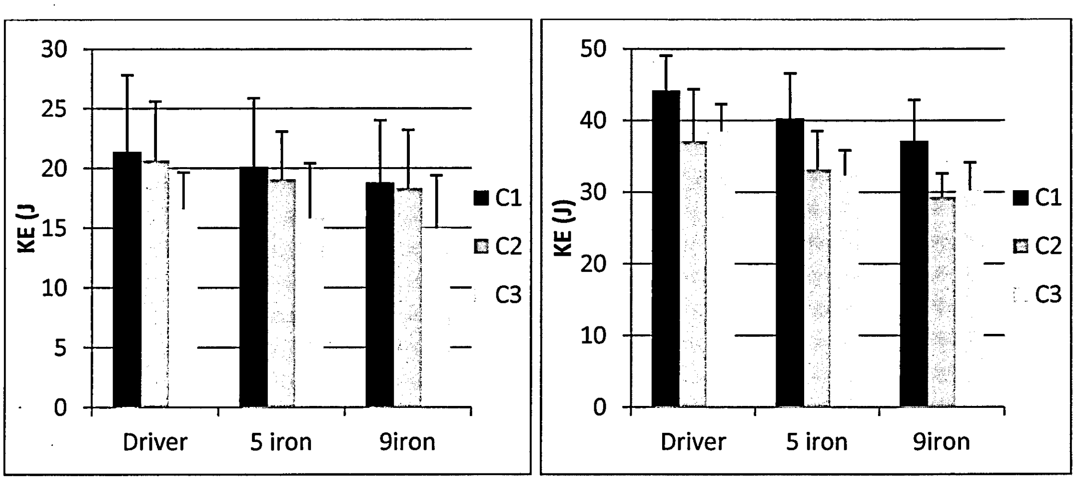


Figure 6.6 (a) left: Mean magnitude of peak local rotational Upper Body KE. (b) right: Mean magnitude of peak local rotational Club KE.

There was a significant main effect of playing standard on the magnitude of peak local rotational KE ($F(2,30) = 7.28$, $p < 0.01$, $r = 0.44$). This indicated that category 1 and 2 golfers produced significantly larger magnitudes of peak local rotational segment KE compared with category 3 golfers. Furthermore, there was also a significant main effect of segment ($F(1.92, 57.56) = 536.88$, $p < 0.01$, $r = 0.95$) which suggested that for all categories, a proximal-to-distal sequence was not evident. Instead, the magnitude of peak local rotational Upper Body KE was significantly larger than peak local rotational Arms KE ($F(1, 30) = 248.18$, $p < 0.01$, $r = 0.94$) (Table 6.2). Finally, there was a main effect of club ($F(1.58, 47.49) = 79.81$, $p < 0.01$, $r = 0.79$) which suggested that the magnitude of peak local rotational segment KE increased sequentially from the 9 iron to the driver. As the interaction between club and playing standard was non-significant the results indicate that this strategy was employed by all playing standards.

For the timing of peak local rotational segment KE there was no interaction between segment and playing standard ($F(4.09, 61.32) = 0.59$, $p = 0.67$, $r = 0.10$). However, peak Upper Body local rotational segment KE occurred earlier for category 1 golfers (0.714 ± 0.066 MT) than for category 2 (0.791 ± 0.092 MT) and 3 golfers (0.791 ± 0.072 MT) (Figure 6.7; Figure 6.8). There was also a main effect of segment on the timing of peak local rotational segment KE ($F(2.04, 61.32) = 71.98$, $p < 0.01$, $r = 0.74$). The planned contrasts suggested that for all playing standards peak local rotational Lower Body and Upper Body KE occurred at the same time whilst peak local rotational Upper Body, Arms and Club local rotational KE occurred in a proximal-to-distal sequence (Table 6.3).

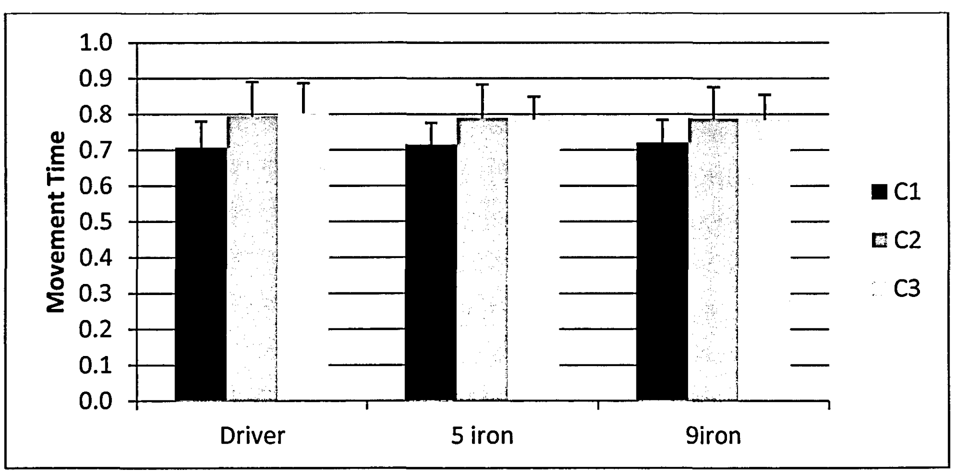


Figure 6.7 Timing of mean local rotational Upper Body KE.

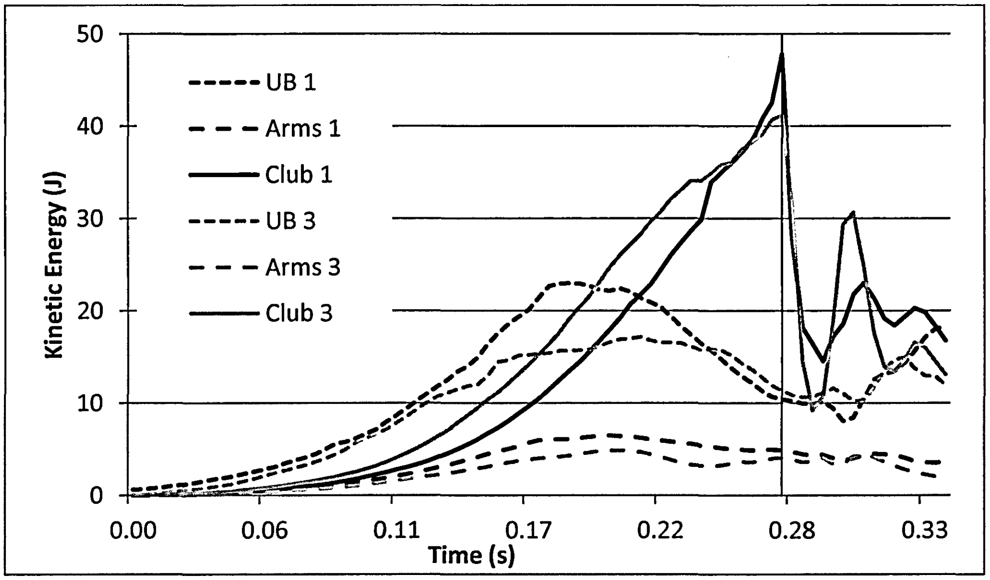


Figure 6.8 Local rotational UB, Arms and Club KE for a representative category 1 and category 3 player using the driver from Top-of-Backswing to Mid-Follow-Through (impact is highlighted for both players).

6.3.4. Remote rotational segment kinetic energy

The interaction between segment and playing standard was significant ($F(2.47,37.10) = 3.73$, $p = 0.03$, $r = 0.30$) which indicated that category 1 (41.33 ± 10.02 J) and category 2 golfers (41.06 ±6.58 J) produced significantly higher magnitudes ($p < 0.01$) of peak remote rotational Arms KE than category 3 players (32.78 ± 5.34 J) (Figure 6.9; Figure 6.10).

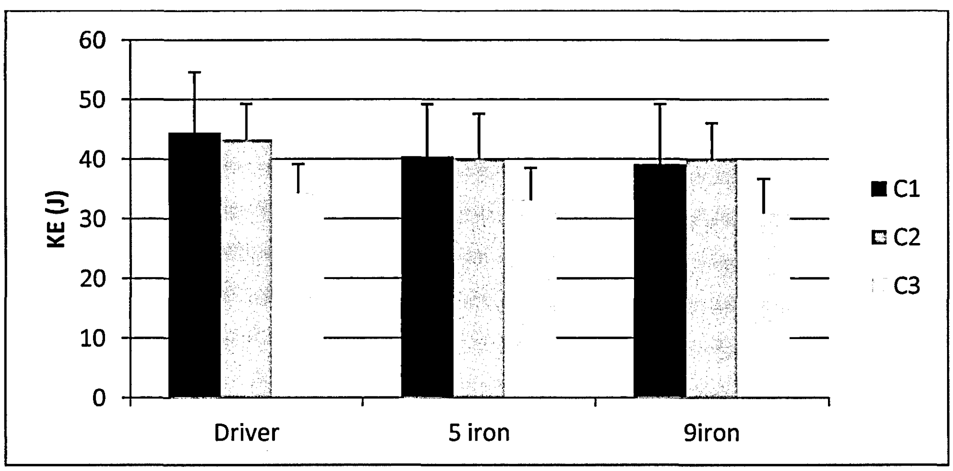


Figure 6.9 Mean magnitude of peak remote rotational Arms KE.

There was a significant main effect of playing standard on the magnitude of peak remote rotational KE ($F(2,30) = 5.49$, $p = 0.01$, $r = 0.39$). The *post-hoc* tests suggested that category 1 and 2 players produced significantly higher magnitudes of peak remote rotational segment KE than category 3 players. Furthermore, there was a significant main effect of segment (F(1.24, 37.10) = 717.00, $p < 0.01$, r = 0.98) which suggested that, for all categories the magnitude of peak Arms remote rotational segment KE was significantly greater than that of the Lower Body and Upper Body (Table 6.2). Finally, there was a main effect of club (F(2,60) = 22.81, $p < 0.01$, r = 0.53) which suggested that regardless of playing standard significantly larger magnitudes of peak remote rotational KE were generated with the driver compared with the 5 iron (F(1,30) = 16.36, $p < 0.01$, r = 0.60) and with the 5 iron compared with the 9 iron (F(1,30) = 6.88, $p = 0.01$, r = 0.43) (Table 6.2).

There was no effect of playing standard on the timing of peak remote rotational segment KE (F(2, 30) = 0.78, $p = 0.46$, r = 0.16). However, there was a significant main effect of segment (F(2,60) = 12.61, $p < 0.01$, r = 0.42) which indicated that peak remote rotational Lower Body and Upper Body KE occurred at the same time whilst peak Arms KE occurred significantly later in the downswing (F(1, 30) = 10.44, $p < 0.01$, r = 0.51). As the interaction between segment and playing standard was not significant the results suggest that this main effect was evident for all three categories (Figure 6.10; Table 6.3).

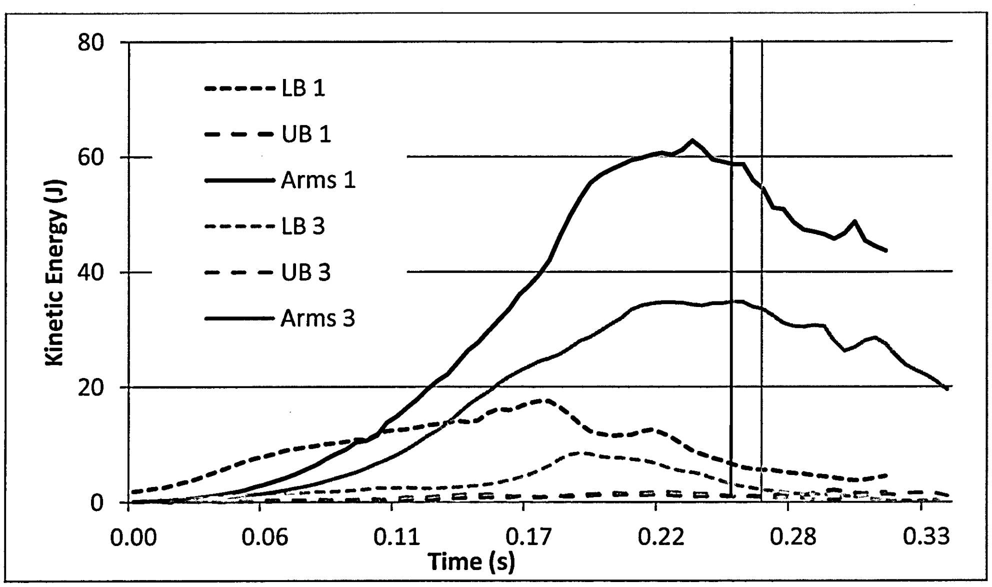


Figure 6.10 Remote rotational LB, UB and Arms KE for a representative category 1 and category 3 player using the Driver from Top-of-Backswing to Mid-Follow-Through (impact is highlighted for each player).

Table 6.2 Mean peak Total, Translational, Local Rotational and Remote Rotational KE for ea

	Category 1			Category 2			
	Driver	5 iron	9 iron	Driver	5 iron	9 iron	D
Total KE (J)							
Lower Body	31.59 ± 7.14	29.02 ± 6.23	26.38 ± 5.15	29.16 ± 9.80	25.63 ± 5.00	24.89 ± 5.44	21.
Upper Body	34.47 ± 5.87	30.77 ± 5.27	29.37 ± 4.90	33.83 ± 6.85	29.76 ± 5.99	28.46 ± 6.63	28.
Arms	104.68 ± 21.18	99.42 ± 18.78	95.70 ± 19.64	94.23 ± 9.24	90.25 ± 9.65	87.63 ± 10.42	79.
Club	284.11 ± 34.09	280.35 ± 36.13	249.17 ± 37.80	245.88 ± 35.67	243.53 ± 19.29	207.04 ± 18.19	250.
Translational KE (J)							
Lower Body	13.07 ± 6.32	10.89 ± 4.40	9.39 ± 3.44	11.77 ± 4.75	9.64 ± 2.19	9.69 ± 3.01	9.2
Upper Body	13.50 ± 3.84	11.87 ± 3.11	11.63 ± 3.06	14.47 ± 5.32	11.44 ± 4.50	11.05 ± 3.88	12.
Arms	59.68 ± 13.64	58.51 ± 12.06	56.18 ± 12.79	51.90 ± 8.14	50.50 ± 7.49	49.45 ± 7.37	44.
Club	240.19 ± 29.80	240.37 ± 30.12	212.95 ± 33.28	208.69 ± 29.49	210.43 ± 15.55	177.79 ± 17.04	211.
Local Rotational KE (J)							
Lower Body	10.10 ± 2.98	9.84 ± 2.59	9.39 ± 2.88	9.52 ± 3.02	8.76 ± 2.29	8.60 ± 2.42	7.9
Upper Body	21.38 ± 6.44	20.12 ± 5.75	18.81 ± 5.20	20.68 ± 4.93	19.10 ± 3.98	18.35 ± 4.86	16.
Arms	6.15 ± 1.66	5.74 ± 1.63	5.43 ± 1.69	5.73 ± 1.39	5.21 ± 1.44	5.17 ± 1.16	4.8
Club	44.17 ± 4.87	40.24 ± 6.36	37.11 ± 5.75	37.10 ± 7.25	33.17 ± 5.34	29.33 ± 3.26	38.
Remote Rotational KE (J)							
Lower Body	11.42 ± 4.12	10.37 ± 3.40	9.91 ± 3.80	10.80 ± 2.82	10.18 ± 2.00	9.74 ± 2.19	8.2
Upper Body	2.18 ± 0.90	1.93 ± 1.03	1.60 ± 0.81	2.22 ± 0.62	1.76 ± 0.79	1.78 ± 0.75	2.0
Arms	44.45 ± 10.14	40.41 ± 8.82	39.12 ± 10.10	43.30 ± 5.99	40.10 ± 7.49	39.78 ± 6.24	34.

Table 6.3 The mean time of peak Total, Translational, Local Rotational and Remote Rotational KE f

	Category 1			Category 2			
	Driver	5 iron	9 iron	Driver	5 iron	9 iron	D
l KE (J)							
er Body	0.764 ± 0.104	0.723 ± 0.115	0.731 ± 0.105	0.729 ± 0.168	0.730 ± 0.126	0.731 ± 0.115	0.79
er Body	0.698 ± 0.085	0.685 ± 0.072	0.698 ± 0.088	0.774 ± 0.129	0.773 ± 0.127	0.780 ± 0.119	0.78
s	0.746 ± 0.040	0.751 ± 0.035	0.756 ± 0.032	0.785 ± 0.074	0.776 ± 0.082	0.785 ± 0.070	0.75
	0.989 ± 0.007	0.988 ± 0.008	0.989 ± 0.005	0.984 ± 0.006	0.986 ± 0.009	0.986 ± 0.007	0.98
lational KE (J)							
er Body	0.712 ± 0.148	0.664 ± 0.154	0.689 ± 0.158	0.651 ± 0.191	0.602 ± 0.170	0.610 ± 0.137	0.75
er Body	0.665 ± 0.102	0.568 ± 0.076	0.615 ± 0.072	0.760 ± 0.110	0.641 ± 0.106	0.628 ± 0.108	0.82
s	0.714 ± 0.032	0.722 ± 0.045	0.726 ± 0.039	0.733 ± 0.057	0.734 ± 0.074	0.741 ± 0.064	0.70
	0.991 ± 0.005	0.988 ± 0.008	0.990 ± 0.005	0.984 ± 0.006	0.986 ± 0.010	0.986 ± 0.007	0.98
l Rotational KE (J)							
er Body	0.745 ± 0.086	0.730 ± 0.085	0.720 ± 0.072	0.767 ± 0.086	0.737 ± 0.086	0.761 ± 0.094	0.77
er Body	0.706 ± 0.073	0.714 ± 0.061	0.720 ± 0.063	0.794 ± 0.096	0.790 ± 0.093	0.788 ± 0.088	0.80
s	0.845 ± 0.128	0.868 ± 0.128	0.873 ± 0.123	0.866 ± 0.121	0.878 ± 0.124	0.924 ± 0.108	0.88
	0.986 ± 0.08	0.985 ± 0.011	0.987 ± 0.007	0.981 ± 0.011	0.984 ± 0.009	0.986 ± 0.008	0.98
ote Rotational KE (J)							
er Body	0.753 ± 0.107	0.730 ± 0.111	0.749 ± 0.135	0.781 ± 0.146	0.800 ± 0.083	0.806 ± 0.098	0.70
er Body	0.789 ± 0.159	0.759 ± 0.119	0.772 ± 0.136	0.812 ± 0.147	0.782 ± 0.170	0.802 ± 0.127	0.87
s	0.853 ± 0.078	0.865 ± 0.069	0.857 ± 0.070	0.911 ± 0.087	0.875 ± 0.096	0.913 ± 0.090	0.89

6.4. Discussion

The aim of this study was to examine the effect of playing standard on the sequencing of segmental motions in golf swings performed with a driver, 5 iron and 9 iron. It was hypothesised that all golfers would produce a proximal-to-distal sequence in the magnitude of total segment KE and that skilled golfers would produce larger magnitudes of segment KE than less skilled players. In accordance with the results of Chapter V, it was also predicted that skilled golfers would exhibit sequential timing of peak translational, local rotational and remote rotational segment KE from the Upper Body to the Club whilst less skilled players would not. Finally, it was expected that peak Lower Body and Upper Body translational kinetic energies would occur later in the downswing for highly skilled players.

The sequential increase in the magnitude of peak total segment KE exhibited by skilled golfers in Chapter V and by elite players in other examinations of segment KE (Anderson, 2007; Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008) was also evident in golf swings performed by category 2 and 3 golfers. Regardless of playing standard and club type, the magnitude of peak total KE increased sequentially from the Lower Body to the Club. The distinct total segment KE timing profile previously exhibited by skilled golfers (Chapter V; Anderson, 2007; Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008) was also evident for category 2 and 3 golfers. For all categories, peak total Lower body, Upper body and Arms KE occurred simultaneously whilst peak total Club KE occurred significantly later, just before impact. Therefore, the results of this study suggest that differences in playing ability between category 1, 2 and 3 golfers are not associated with variations in the sequencing of the magnitude or timing of peak total segment KE.

6.4.1. Magnitude of segment kinetic energy

Although the magnitude of peak total segment KE increased sequentially from the Lower Body to the Club for all playing abilities, there were some significant effects of playing standard on peak total segment KE. The results provided support for the initial research hypothesis ($H_{\text{ALTERNATIVE 1}}$) as, regardless of club type, category 1 players generated significantly larger peak total KE than category 2 and 3 players. These increases in segment KE produced by category 1 golfers were associated with the generation of significantly larger magnitudes of peak total Arms and Club KE. The finding for the Club segment was related to increased magnitudes of peak translational and local rotational Club KE and as the same club was used by all players provides further support for the notion that skilled players produce larger clubhead velocities than less skilled players (Cheetham *et al.*, 2008; Meister *et al.*, 2011; Nesbit and Serrano, 2005; Zheng *et al.*, 2008).

One factor that might be responsible for the faster golf swings and increased magnitude of peak total Arms KE generated by skilled golfers is the translational KE of the Arms segment. Category 1 golfers produced significantly larger peak translational Arms KE than category 3 golfers (Figure 6.3a). The importance of translational arms movements in the downswing have previously been highlighted as it has been reported that the translational velocities of the arms (Hume, Keogh and Reid, 2005) and linear work done by the arms (Nesbit and McGinnis, 2009) are extremely important in the acceleration of the club during the latter part of the downswing. Furthermore, in other open kinetic chains movements such as cricket bowling, higher peak translational Arms KE has been associated with increased speed of the distal segment at release (Ferdinands *et al.*, 2011). Therefore, translational movements of the rigid bodies which define the Arms segment appear to be extremely important in the generation of translational Arms KE and production of high clubhead speeds in the golf swing.

The results of this study provided further support for the importance of arm movements in the generation of fast, skilled golf swings. Category 1 and 2 players produced significantly larger peak remote rotational Arms KE than category 3 players. Remote rotational KE describe the energy of a segment rotating around its own centre of mass. The generation of remote rotational Arms KE has previously been identified as a source of large magnitudes of peak translational Club KE as the peaking of remote rotational Arms KE coincided with the rapid increase in translational Club KE (Anderson, 2007). Therefore, it is apparent that the generation of remote rotational Arms KE is important in the swings of skilled players and can differentiate between the swings of golfers of varying ability. However, the mechanisms which underpin this component of

Arms KE are less well defined. The magnitude of remote rotational Arms KE has been associated with the velocity of wrist 'uncocking' late in the downswing (Anderson, 2007). It has also been suggested that skilled golfers produce larger elbow (Nesbit and Serrano, 2005; Zheng *et al.* 2008) and right wrist (Chu, Sell and Lephart, 2010; Zheng *et al.*, 2008) extension velocities during the downswing compared with high handicap golfers. Therefore, it is possible that the larger magnitudes of peak remote rotational KE produced by category 1 and 2 players were associated with large extension velocities of the elbow and uncocking of the wrist. However, further research is needed to confirm this notion.

Another factor which might be responsible for the increased speed associated with the swings of skilled golfers is the generation of local rotational Upper Body KE. Category 1 and 2 golfers produced significantly higher peak local rotational Upper Body KE than category 3 golfers. Axial rotations of the upper trunk have previously been related to skilled golf performance as it has been reported that higher maximum rotational speeds of the thorax are produced by professional golfers than by less skilled amateurs (Cheetham *et al.*, 2008), by golfers who produce high ball velocities (Myers *et al.*, 2008) and by golfers with greater driving distance (McLean, 1992). Larger x-factors have also been associated with fast, skilled golf swings (Cheetham *et al.*, 2001; Chu, Sell and Lephart, 2010; Zheng *et al.*, 2008). Although the x-factor does not directly relate to movements during the downswing it has been suggested that it provides the foundations for increased axial rotation velocities of the upper torso (Lai *et al.*, 2011; Myers *et al.*, 2008).

The relationship between clubhead speed and the axial rotation velocity of the upper torso appears to be well defined. However, research has also suggested that unlike unskilled golfers who can only regulate the axial rotation of the upper trunk, skilled golfers are able to control the global rotations of the upper trunk in the downswing (Horan and Kavanagh, 2012; Yahara *et al.*, 2008). The swings of highly skilled golfers have been associated with increased forward and lateral tilt velocities of the upper trunk (Chu, Sell and Lephart, 2010) as these movements contribute to an increased overall angular speed of the thorax (Horan and Kavanagh, 2012) and larger clubhead velocities (Chu, Sell and Lephart, 2010). Combined with the results presented in this study, these studies suggest that rather than relating to axial rotations, the overall angular speed of the upper torso and subsequent generation of large Upper Body local rotational KE are the mechanisms which underpin skilful golf swings.

The importance of the association between local rotational Upper Body KE and skill level provides further support for the use of analyses of segment KE. They suggest that golf coaches and researchers must not solely rely on axial rotation-based analyses if they are to fully appreciate the coordination required of the upper trunk. Furthermore, the results of this study have highlighted the importance of the translational and rotational movements of the rigid bodies which define the Arms segment in the generation of translational and remote rotational Arms KE. Again, this findings supports the use of analyses of segment KE to examine the sequencing of segmental motions in the golf swing as, unlike the frequently used analyses of segment angular velocity (Cheetham *et al.*, 2008; Neal *et al.*, 2008; Tinmark *et al.*, 2010) this form of analysis considers both translational and 3D rotational movements.

6.4.2. Timing of segment kinetic energy

On initial observation, the timing of peak total segment KE and its components was similar for all playing standards. For all categories, peak total KE for the body segments occurred simultaneously whilst peak total Club KE occurred significantly later in the downswing just before impact. Despite this support for the principle of optimal coordination of partial momenta (Van Gheluwe and Hebbelinck, 1985) and in contrast with $H_{ALTERNATIVE\ 3}$ the results also suggested that peak local and remote rotational KE occurred in a proximal-to-distal sequence from the Upper Body to the most distal segments. For both professional and amateur golfers, Cheetham *et al.* (2008) suggested that peak angular velocities for the torso and lead arm occurred at the same time during the downswing. It is likely that these contrasting results reflect the ability of analyses of segment KE to account for more subtle changes in technique and to consider the rotations of rigid bodies in3D. As a result of these sequential motions of the Upper Body and Arms it is possible that golfers of all abilities were able to take advantage of the benefits associated with the SSC (Komi, 2000). It is likely that these earlier, rapid movements of the Upper Body caused the Arms to lag behind stretching the shoulder and trunk musculature between these segments. This pre-stretching would have enhanced the potential of these muscles to produce powerful concentric contractions (Van Ingren and Schenau, 1984).

Sequential timing of peak translational KE from the Upper Body to the Club was also evident for category 1 players. However, this sequence was not apparent for category 2 and 3 players. In contrast with $H_{ALTERNATIVE2}$, for these less skilled players peak Upper Body translational KE occurred later in the downswing. This would appear logical as later forward and upward translations of the upper trunk have previously been associated with golf swings of less skilled players (Horan and Kavanagh, 2012). Furthermore, an early peak in maximum translational thoracic KE has been associated with high ball release velocities in cricket fast bowling (Ferdinands, Kersting and Marshall, 2012). As a result of the late peak in Upper Body translational KE identified in this study, a sequential pattern of peak translational KE was not evident from the Upper Body to the Arms for these less skilled players. Therefore, it is possible that category 2 and 3 players did not transfer KE from the Upper Body to the Arms as efficiently as category 1 players. The significantly later peak in translational movements of the Upper Body may also have prevented these players from taking full advantage of the SSC. Due to these factors, it is possible that the earlier peak in Upper Body translational KE is responsible for the higher peak translational Arms and subsequently higher peak translational Club KE produced by category 1 players.

As well as emphasising the importance of sequential timing, particularly of peak translational KE distally from the Upper Body, the results suggested that, to produce fast and accurate golf swings, peak local rotational Upper Body KE should occur early in the downswing. Regardless of club type, category 1 golfers maximised local rotational Upper Body KE earlier in the downswing than category 2 and 3 golfers. Although the differences in timing were non-significant, the mean difference (0.077 MT) was larger than the minimal meaningful differences (0.031 – 0.052 MT) identified in Chapter 4. This early peak exhibited by skilled players would potentially have enabled them to take advantage of the SSC and provided more time for the KE possessed by the Upper Body to be transferred to the Arms and Club in time for impact (Chu, Sell and Lephart, 2010). Analyses of segment angular velocity have presented conflicting results regarding the effect playing standard on the timing of peak thorax angular velocity. Horan and Kavanagh (2012) suggested that skilled players maximise the axial rotations and lateral bending of the upper torso early in the downswing whilst Cheetham *et al.,* (2008) demonstrated that no timing differences were evident between professional and amateur players. These results provide further evidence that analyses of segment KE can more effectively account for changes in technique than more traditional analyses of segment angular velocity. It would also appear that an early peak in local rotational Upper Body KE is a mechanism which allows skilled golfers to generate faster golf swings.

6.5. Conclusion

The findings generally supported the notion that skilled golfers produce faster golf swings than less skilled players. Compared with category 2 and 3 golfers, highly skilled, category 1 players generated larger peak total Arms and Club KE. The results suggested that larger translational and remote rotational Arms KE and local rotational Upper Body KE may contribute to the larger total Club KE produced by skilled players. These results emphasised the benefits of examining the golf swing using analyses of segment KE as they highlighted the importance of considering the translational movements and 3D rotations of golfers. The results also highlighted the importance of a sequence of translational KE from the Upper Body to the Club and earlier peak in local rotational Upper Body KE. Both of these movement strategies were only utilised by category 1 golfers and it was suggested that they enabled these golfers to generate increased total Club KE via the kinetic link principle and stretch shortening cycle.

7. Chapter VII - The Effect of Weight Transfer Style on the Sequencing of Segment Kinetic Energy in the Golf Swing

7.1. Introduction

A style or movement strategy has been defined as the performance of a skill in a different way to achieve the same aim (Bates, 1996). It has been suggested that overlooking different styles or movement strategies could lead to important performance components being missed in biomechanical or statistical analyses (Bates, 1996; Lees, 2010). Examples of obviously different styles in sport are the double-handed versus the single-handed backhand in tennis, the hang versus the bicycle kick technique for the flight phase of the long jump, and the slide versus the rotational technique in the shot put (Lees, 2010).

In golf, three different strategies have been identified for gripping the club: the interlocking grip, the overlapping grip, and the baseball grip (Lythgoe, 2011). It has also been suggested that, two kinematic swing styles, the modern and the classic golf swings exist (Gluck *et al.*, 2008). The modern golf swing emphasised a large shoulder turn with a restricted hip turn which facilitated increased head speeds through the utilisation of the stretch shortening cycle whilst the classic golf swing emphasised a shorter backswing which facilitated a more consistent ball contact (Gluck *et al.*, 2008). Perhaps less obvious using kinematic analysis is the existence of multiple weight transfer styles.

In golf literature, 'weight transfer' has frequently been used to describe movement of the resultant ground reaction force vector, typically in the direction of the shot (Richards *et al.*, 1985, Koslow, 1994; Robinson, 1994, Wallace *et al.*, 1994; Ball and Best, 2007a). A long held belief in the golf swing literature is that an ideal weight transfer sequence exists. According to many scientific (Ball and Best, 2007a; Burden, Grimshaw and Wallace, 1998; Hume, Keogh and Reid, 2005; Okuda, Gribble and Armstrong, 2010) and coaching (Haney, 1999; Leadbetter, 1995) texts, this sequence starts with the weight evenly balanced between the feet at address (start of backswing). The weight then moves towards the back foot during backswing before moving towards the front foot just before the start of downswing. This forward movement becomes more rapid in the early phase of the downswing and continues through to the front foot at ball contact and at follow-through.

Golf coaching literature has frequently stressed the importance of producing this weight transfer pattern (Grant, Bann and Lynch, 1996; Leadbetter, 1995; Norman, 1995), as it was thought that transferring weight onto the front foot at impact was critical for the production of successful golf swings (Chu, Sell and Lephart, 2010; Healy *et al.*, 2011; Koslow, 1994; Miura, 2001; Okuda *et al.*, 2002; Wallace, Graham and Bleakley, 1990). Scientific studies have also linked this traditional sequence and the transference of weight exclusively in the intended direction of ball flight with the generation of increased clubhead velocities (Burden, Grimshaw and Wallace, 1998; Hume, Keogh and Reid, 2005).

It has been suggested that golfers do not always conform to the ideal weight transfer pattern. However, traditionally, these weight transfer patterns have been referred to as errors which adversely affect the quality of golf swings. For example, Koslow (1994) referred to a reverse pivot defined as the movement of weight from the front foot towards the back foot during the downswing as an error as it was associated with reduced clubhead speeds. Burden, Grimsahw and Wallace, (1998) also suggested that six out of eight sub-10 handicap golfers moved their centre of mass backwards immediately before impact and Neal (1998) subjectively defined a left-to-right and a rotational swing style. The left-to-right style produced greater CPy movement (parallel to the ball-target line) and less CPx movement (perpendicular to the ball-target line) compared with the rotational style (Neal, 1998). As low handicap golfers were examined, these styles might have been considered to be valid techniques. However, this was not stated by the researchers and nor was the criterion for each style identified.

In the only other peer reviewed study to define different swing styles, Ball and Best (2007a) identified Front Foot and Reverse weight transfer patterns by measuring weight distribution in the direction of the shot (CPy %) at eight different swing events. Both swing styles demonstrated similar patterns of weight distribution from takeaway (TA) to early downswing (ED) which conformed to the ideal weight transfer style. The

Front Foot players continued to follow this pattern with CPy % moving towards the front foot at impact (BC) (CPY % = 81 ± 11 %) and remaining close to the front foot at mid-follow-through (MF) (CPY % = 80 ± 11 %) - Figure 7.1. However, for the Reverse players, CPy% moved towards the back foot during the late downswing, was positioned at mid-point between the feet at BC (CPy % = 53 ± 12 %) and moved closer to the back foot at MF (CPy % = 41 ± 13 %) (Figure 7.1).

Traditional coaching literature would consider the Reverse weight transfer style (Ball and Best, 2007a) to be technical errors associated with falling away at impact (Leadbetter, 1990). However, Ball and Best (2007a) provided support for the validity of the Reverse strategy as they reported that both groups contained highly skilled golfers. Furthermore, no differences were observed between the two groups in terms of handicap or clubhead velocity at ball contact. Support for the Reverse weight transfer style can also be provided by examining the translation and rotation of the trunk during the downswing. It has been suggested that, regardless of playing ability, golfers produce lateral flexion and translation of the upper body away from the target during the latter portion of the downswing (Egret *et al.*, 2003, Horan, 2010; Lindsay, Horton and Paley, 2002; McTeigue *et al.*, 1994; Sanders and Owen, 1992; Okuda *et al.*, 2010). Since a golfer's weight transfer has been closely related to the movement of the trunk it has been suggested that these movements could be responsible for the reverse in CPy during the downswing (Beak *et al.*, 2013; Okuda *et al.*, 2010).

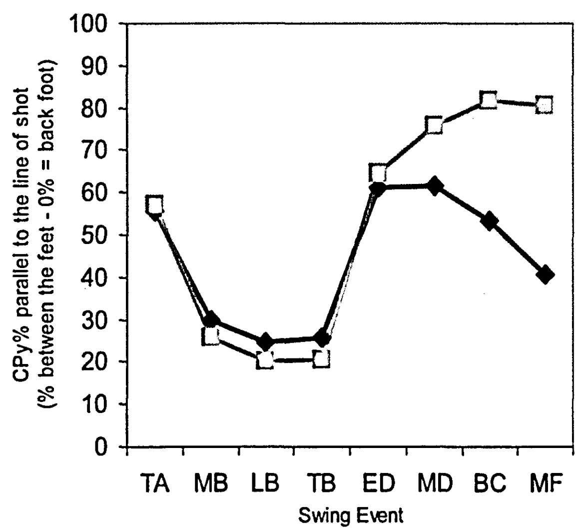


Figure 7.1 CPy% positions at each swing event for the front foot and reverse foot players. Taken from Ball and Best (2007a).

The importance of identifying different styles or movement strategies when performing analyses of a skill or technique has been emphasised (Bates, 1996). Performing group based analyses between different movement styles or strategies also enables group based statistical analyses to be performed that many researchers advocate (Reboussin and Morgan, 1996). As such, Ball and Best (2007b) examined the relationship between weight transfer style and clubhead speed. For the Front Foot style a greater CPy% range was associated with clubhead velocity at ball contract whilst for the Reverse style, positioning weight nearer to mid- stance in late backswing and, a more rapid rate of weight transfer towards the back foot at ball contact, were related to larger clubhead velocities. It was also speculated that the proximal-to-distal sequence was an underlying mechanism for both weight transfer styles as large CPy% range and forward velocities of weight transfer developed system energy, which could have been transferred to the clubhead and ball (Ball and Best, 2007b). The discussion of the proximal-to-distal sequence was limited by the absence of kinematic data (Ball and Best, 2011; Hellstrom, 2009). Therefore, combining kinetic data with an analysis of segment KE could provide further insight into the sequencing of body segment motions.

The analysis of segment KE has been reported to be the most appropriate technique to analyse the proximal-to-distal sequence in the golf swing (Anderson, Wright and Stefanyshyn, 2006). Chapter V also identified that analyses of segment KE could be used to identify differences in golf swings performed using three different clubs. Larger translational Arms and local rotational Upper Body KE and later peaks in translational Lower Body and Upper Body KE were identified when the distance requirement of the shot increased and the driver was used. Furthermore, Chapter VI identified that compared with less skilled golfers, highly skilled players produced larger translational and remote rotational Arms KE and larger local rotational Upper Body KE. The results also highlighted the importance of a sequence of translational KE from the Upper Body to the Club and earlier peak in local rotational Upper Body KE. It has been speculated that Reverse players are required to flex laterally away from the target during the downswing to enable their clubheads to approach the ball at an optimal angle (Hellstroom, 2009). However, although kinetic similarities such as CPy range and maximum Cpy have been associated with clubhead speed for both groups with each

Club, the mechanisms which underpin their golf swings have yet to be determined. Therefore, the aim of this study was to examine the differences in movement sequencing between the Front Foot and Reverse movement strategies when different clubs were used.

7.1.1. Hypotheses

H_{NULL}: Weight transfer style will have no effect on the magnitude or timing of peak segment KE in the golf swing.

$H_{ALTERNATIVE\ 1}$: The Front Foot and Reverse weight transfer styles will be generated by players when all three clubs are used.

$H_{ALTERNATIVE\ 2}$: Front Foot players will produce larger magnitudes of total Upper Body KE compared with Reverse strategy golfers.

$H_{ALTERNATIVE\ 3}$: Peak Upper Body KE will occur earlier for the Reverse players compared with the Front Foot players.

7.2. Methods

7.2.1. Participants

Thirty six male golfers (M ± SD; age 32 ± 12 years, stature 185.8 ± 4.9 cm, mass 84.9 ± 11.4 kg and handicap 8.2 ± 6.8 strokes, range +3 - 20 strokes) volunteered to take part in this study. Playing standard was determined using each players registered handicap and for analysis purposes, professional golfers without a registered handicap were considered to have a handicap of 0 and plus handicaps (e.g. +3) were considered to be negative numbers. Ethics approval was obtained from the Faculty of Health and Wellbeing Research Ethics Committee and written informed consent was obtained from each golfer.

7.2.2. Inertial parameters

Inertial parameters for the feet, lower legs, thighs, lower trunk, middle trunk, upper trunk, neck, head, upper arms and forearms were estimated using the geometric model described in section 3.2. Hand inertial parameters were estimated using the geometric modelling technique described in section 3.5. The anthropometric measurements required to form the shapes of the geometric model were made using the Polhemus electromagnetic tracking system, electromagnetic sensors and custom written software (Section 3.2). The club segment geometry and inertial parameters were based on measurements made by a non-contact laser scanner (Section 3.4.4).

7.2.3. Data collection

Data were collected following the protocol described in section 3.4.2 which required participants to hit fifteen shots from an artificial mat into a net five meters away, five with a driver, 5 iron and 9 iron.

7.2.4. Weight transfer style

Golf swings were performed with each foot on a separate force platform (Ball and Best, 2007a) sampling at 1000 Hz. The y axis was orientated medial-lateral in the direction of shot, the x axis anterior-superior and z in the vertical direction. Force data from the two plates was combined to calculate centre of pressure displacement parallel to the line of shot (CP*y*):

$$CPy = \frac{(Fz1 \times CPy1)+(Fz2 \times CPy2)}{Fz1+Fz2} \qquad \text{Equation 7.1}$$

where:

Fz1 = vertical force from force plate 1

Fz2 = vertical force from force plate 2

CPy1 = centre of pressure parallel to line of shot from force plate 1

CPy 2 = centre of pressure parallel to the line of shot calculated from force plate 2 + distance between the centre of force plate 1 and force plate 2

Following the technique outlined by Ball and Best (2007a) CPy was then normalised to foot position as address and expressed as a percentage (CPy%) of the distance between the back foot (0%) and the front foot (100%) using the following equation:

$$CPy\% = \frac{CPy}{FD} \times 100 \qquad \text{Equation 7.2}$$

where:

FD = Distance between the feet at address

To enable weight transfer style to be classified for each player, CPy% was determined at eight events during each swing (Figure 7.2). The eight swing events for each trial were calculated using kinematic data in the YZ plane and the criteria outline by Ball and Best (2007a). Weight transfer style was determined using the data provided (Table 7.1) and criteria defined by Ball and Best (2007a): that there was no difference between CPy% at TA, MB, LB, TB and ED but the Front Foot players positioned CPy% nearer the front foot at MD, BC, and MF and produced a larger maximum, smaller minimum, and greater range of CPy%. In contrast, the Reverse group employed a smaller range of weight transfer and a "reversing" strategy near ball contact, such that weight was positioned near mid-stance at BC and continued towards the back foot to MF.

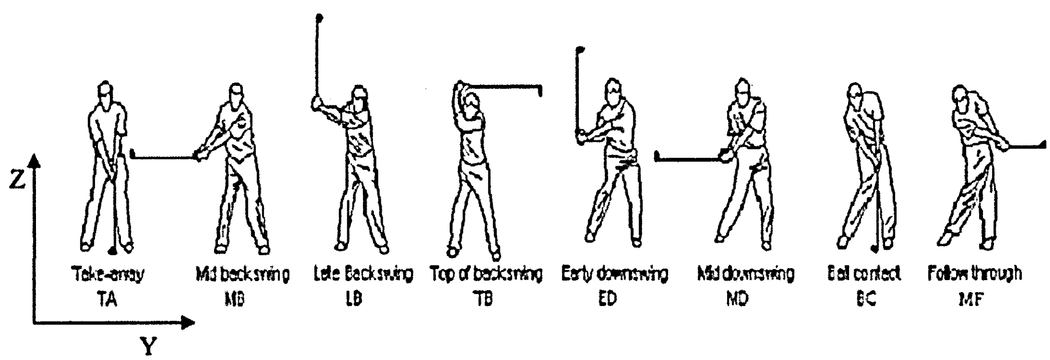


Figure 7.2 Golf swing events (taken from Ball and Best, 2007a).

Table 7.1 Comparison between Front Foot and Reverse groups when the driver was used (Ball and Best, 2007a).

	CPy%	
	Front Foot	Reverse
MD	76 ± 5	62 ± 10
BC	81 ± 11	53 ± 12
MF	80 ± 11	41 ± 13
Maximum	87 ± 9	69 ± 9
Minimum	12 ± 7	18 ± 8
Range	75 ± 11	51 ± 12

7.2.5. Segment kinetic energy

Translational and rotational velocity data were obtained for each rigid body using the Polhemus Liberty electromagnetic tracking system (Polhemus, Inc., Colchester, VT, USA) sampling at 240 Hz and the technique described in section 3.4. Then, for each golf swing, the magnitude and timing of peak KE was calculated for each rigid body, for three body segments (Lower, Upper Body and Arms), and for the Club segment using the technique described in section 3.5. The timing of peak segment KE was then normalised to total movement time (MT) in the downswing where 0 represents the top of backswing and 1 represents impact.

7.2.6. Statistical analysis

7.2.6.1. Weight transfer style classification

All of the statistical analyses were conducted using SPSS (Version 19.0). Following the approach adopted by Ball and Best (2007a), for each Club, CPy% for the Front Foot and Reverse foot groups at the eight swing events was compared using one-way analyses of variance (ANOVA) - $\alpha = 0.05$. Using the one-way ANOVA the two weight transfer groups were also compared to determine differences in CPy% maximum and CPy% minimum and CPy% range:

$$CPy\%\ Range = CPy\%\ Maximum - CPy\%\ Minimum \quad \text{Equation 7.3}$$

7.2.6.2. The effect of weight transfer style

A one-way ANOVA ($\alpha = 0.05$) was used to determine differences between the mean handicap, age, height and body mass of the two weight transfer style groups. Following this, three-way mixed design analysis of variances (ANOVA) were employed to determine the effect of weight transfer style, club type and segment on the magnitude and timing of peak segment KE and its components. Follow-up planned contrasts were performed to establish the origin of any significant main effects or significant interactions. By using univariate analyses, these planned contrasts enable the pattern of response for each independent variable of a factorial ANOVA to be identified (Schneiner and Gurevitch, 2001) without increasing the likelihood of a type II error (Vincent and Weir, 2012). Furthermore, Tukey's honest significant difference (HSD) was calculated to identify the specific differences that were responsible for the significant interactions (Equation 7.4) (Vincent and Weir, 2012). Tukey's HSD calculates the minimum difference that must be evident between the raw score means to declare a significant difference between two groups in an interaction:

$$HSD = q_{(k,df_E)}\sqrt{\frac{MS_E}{n}}$$ Equation 7.4

where, q is a value from the Studensized range distribution for k (number of groups) and df_E at a given confidence level (Vincent and Weir 2012), MS_E is the mean square error from the main ANOVA analysis and n is the size of the groups.

7.3. Results

7.3.1. Weight transfer style

Using the criteria outlined in section 7.2.4, three players were identified who exhibited different weight transfer strategies with different clubs and were therefore excluded from the analysis. For example, player 407 adopted the Reverse strategy with the driver and the Front Foot strategy with the 5 and 9 irons (Figure 7.3).

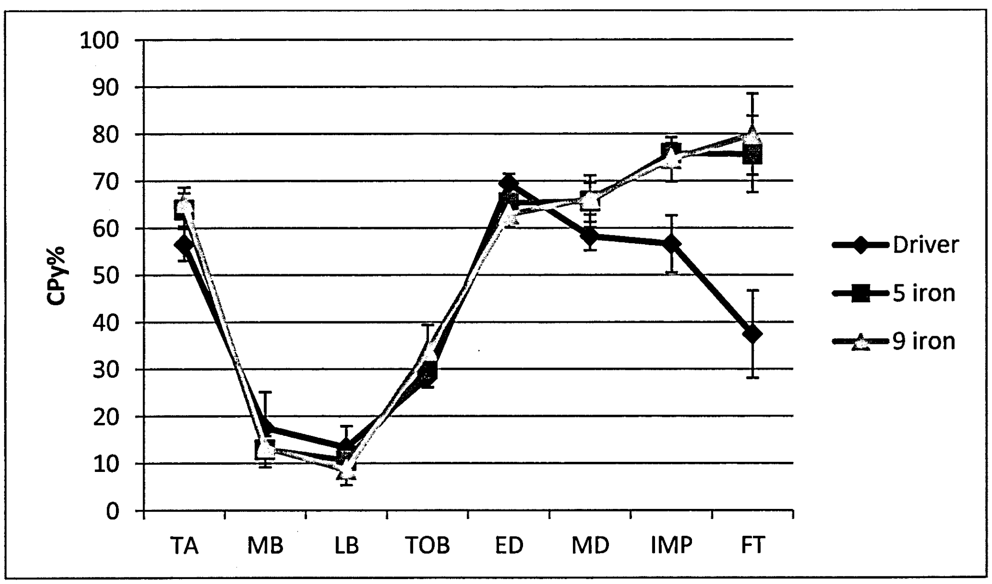


Figure 7.3 CPy% at swing events for player 407.

Using the same criteria, nine players were identified who adopted the Reverse strategy with all clubs. To ensure the sample size was the same for each group, nine Front Foot players were then randomly selected from the remaining twenty-four golfers (Table 7.2). CPy% traces for a representative Front Foot and Reverse player are displayed in Figure 7.4 and Figure 7.5 respectively.

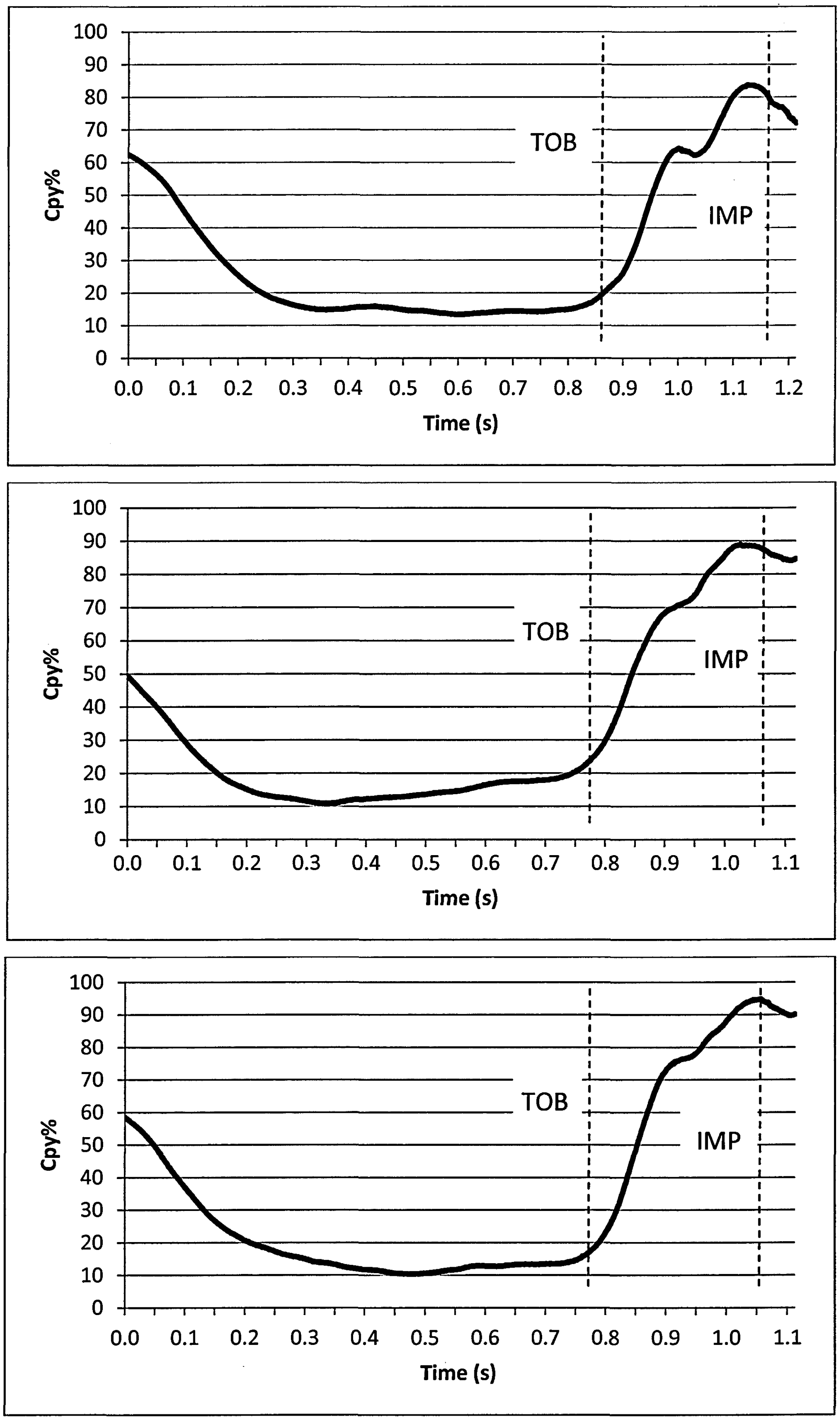


Figure 7.4 CPy% for a representative Front Foot player from Take-away to Mid-Follow-Through
Top: Driver, Middle: 5 iron, Bottom: 9 iron.

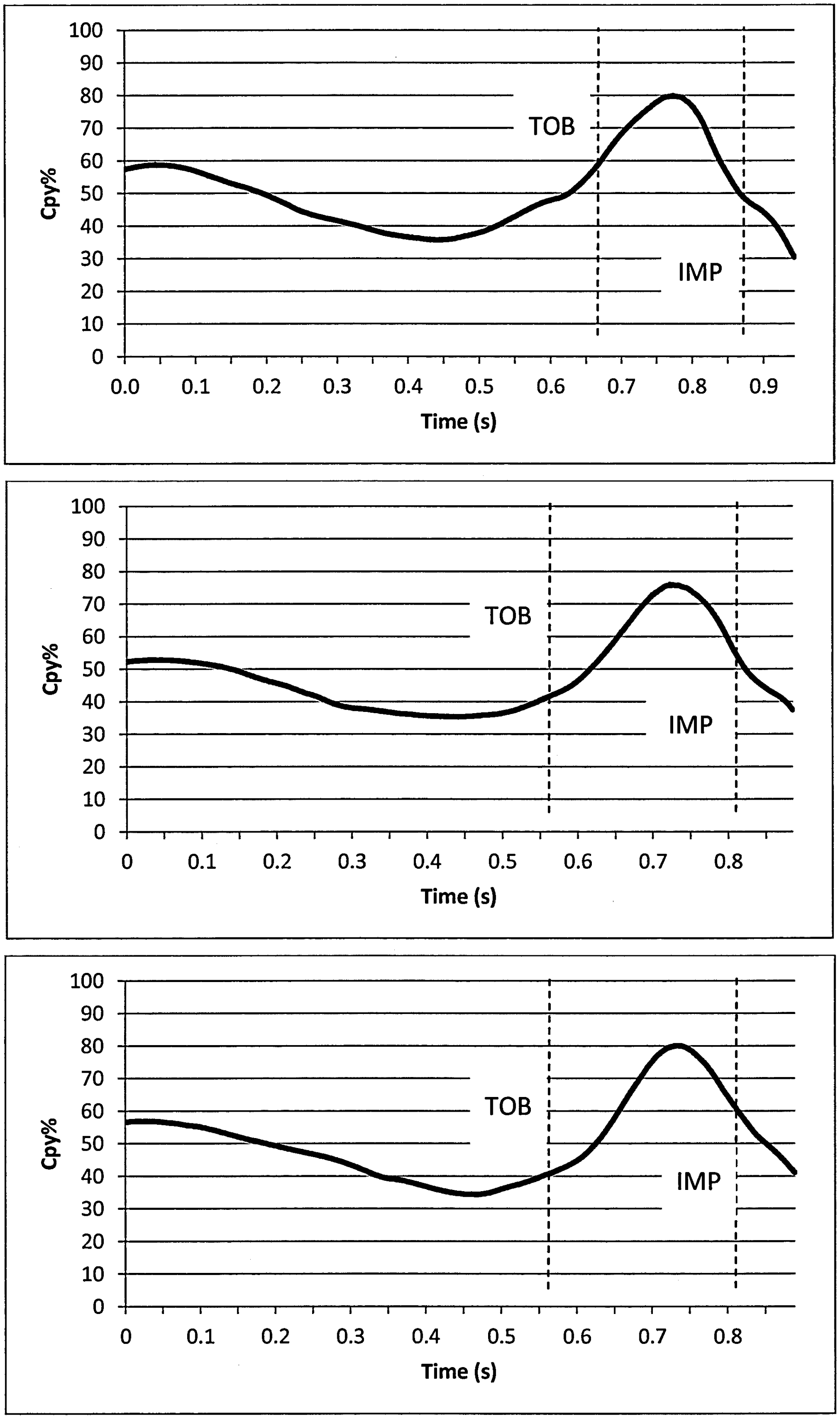


Figure 7.5 CPy% for a representative Reverse player from Take-Away to Mid-Follow-Through
Top: Driver, Middle: 5 iron, Bottom: 9 iron.

One-way ANOVA indicated that the mean age, mass, stature and handicap for the two groups were not statistically different at $p < 0.05$. The effect sizes were also small between the two groups for each variable (effect sizes: large $\eta^2 \geq 0.14$; medium $\eta^2 \geq 0.08$; small $\eta^2 \geq 0.02$; Cohen, 1988) – Table 7.2.

Table 7.2 Group category details.

WTS	Age (years)	Mass (kg)	Stature (cm)	Handicap (strokes)
Front	33 ± 12	84.1 ± 10.7	185.6 ± 6.2	8.2 ± 5.6
Reverse	30 ± 12	85.6 ± 12.7	185.9 ± 3.7	7.4 ± 9.3

Mean (± standard deviation) CPy% for the Front Foot and Reverse strategy players at each swing event with each club are shown in Figure 7.6. With all clubs, Cpy% for both groups followed a similar pattern during the backswing until the start of the downswing. However, from there CPy% for the Front Foot group continued forwards and was positioned towards the front foot at mid-downswing, ball contact and mid-follow-through whilst the Reverse strategy players demonstrated a reversal in CPy% movement. For these players CPy% moved closer to the back foot at ball contact (49 ± 13%) and continued towards the back foot at mid-follow-through (45 ± 13%) - Figure 7.6.

The one-way ANOVA indicated that, with all clubs the CPy% exhibited by both groups was similar at take away, mid-backswing, late backswing, top of backswing and early downswing. However, large effect sizes were produced at the top of backswing when all three clubs were used (Table 7.3, Table 7.4, Table 7.5) which suggested that, for the Front Foot group, CPy% was closer to the back foot at this swing event. With all three clubs, CPy% was significantly different for the Front Foot and Reverse stategy players at mid-downswing, impact and mid-follow-through (Table 7.3, Table 7.4, Table 7.5). In addition, the one-way ANOVA indicated that, with all clubs, maximum CPy% and CPy% range were also significantly different. However, similar minimum CPy% was produced reported for the two groups when all club were used (Table 7.3, Table 7.4, Table 7.5).

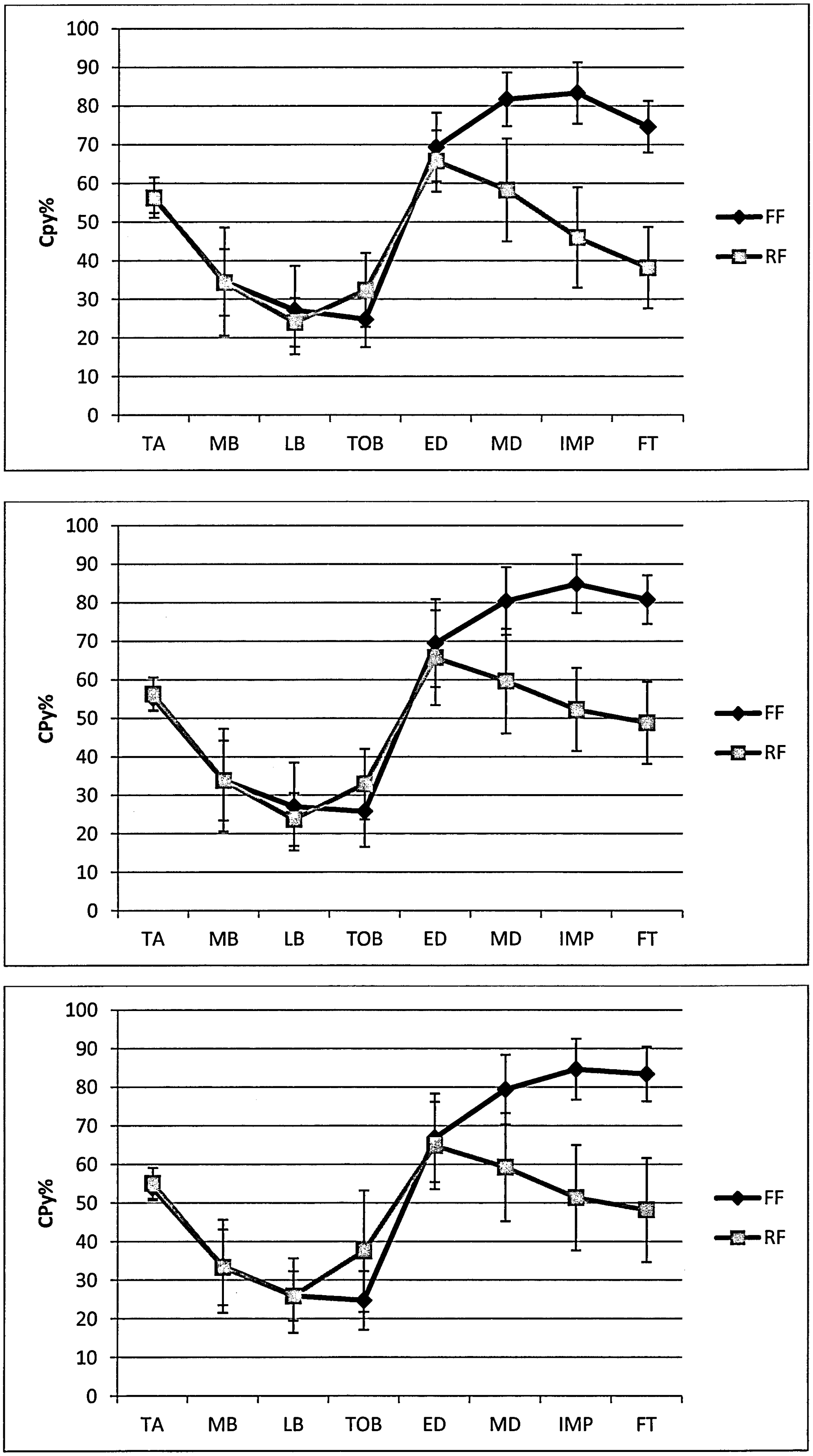


Figure 7.6 Mean CPy% at swing events for the Front Foot and Reverse groups.
Top: Driver, Middle: 5 iron, Bottom: 9 iron.

Table 7.3 Comparison of Front Foot and Reverse groups with the driver.

	CPy%		ANOVA		
	Front Foot	Reverse	F	*p*	η^2
TA	56 ± 4	56 ± 5	0.00	0.960	0.00
MD	35 ± 14	34 ± 9	0.00	0.967	0.00
LD	27 ± 11	24 ± 6	0.47	0.501	0.03
TOB	25 ± 7	35 ± 10	6.31	0.053	0.28
ED	69 ± 9	66 ± 8	0.73	0.405	0.04
MD	82 ± 7	58 ± 13	19.60	0.000	0.55
BC	83 ± 8	46 ± 13	48.00	0.000	0.75
MF	75 ± 7	38 ± 11	68.30	0.000	0.81
Maximum	86 ± 8	71 ± 7	14.96	0.001	0.48
Minimum	18 ± 7	21 ± 5	0.65	0.433	0.04
Range	69 ± 10	50 ± 6	21.99	0.000	0.58

Table 7.4 Comparison of Front Foot and Reverse groups with the 5 iron.

	CPy%		ANOVA		
	Front Foot	Reverse	F	*p*	η^2
TA	55 ± 3	56 ± 4	0.30	0.591	0.02
MD	34 ± 13	34 ± 10	0.00	0.991	0.00
LD	27 ± 11	24 ± 7	0.51	0.485	0.03
TOB	26 ± 9	33 ± 9	2.40	0.141	0.15
ED	69 ± 11	66 ± 12	0.40	0.535	0.02
MD	80 ± 9	60 ± 14	13.17	0.002	0.45
BC	85 ± 8	52 ± 11	49.06	0.000	0.75
MF	81 ± 6	49 ± 11	53.46	0.000	0.77
Maximum	87 ± 8	73 ± 9	11.21	0.004	0.41
Minimum	19 ± 8	21 ± 6	0.19	0.672	0.01
Range	68 ± 10	52 ± 8	12.89	0.002	0.45

Table 7.5 Comparison of Front Foot and Reverse groups with the 9 iron.

	CPy%		ANOVA		
	Front Foot	Reverse	F	*p*	η^2
TA	54 ± 3	55 ± 4	0.50	0.492	0.03
MD	34 ± 12	33 v 10	0.00	0.959	0.00
LD	26 ± 10	26 ± 6	0.00	0.981	0.00
TOB	25 ± 8	37 ± 15	4.28	0.055	0.21
ED	67 ± 11	65 ± 11	0.12	0.734	0.01
MD	79 ± 9	59 ± 14	11.68	0.004	0.42
BC	85 ± 8	51 ± 14	35.64	0.000	0.69
MF	83 ± 7	48 ± 13	42.86	0.000	0.83
Maximum	87 ± 8	71 ± 9	14.20	0.002	0.47
Minimum	19 ± 7	22 ± 6	0.57	0.461	0.03
Range	68 ± 10	49 ± 9	15.03	0.001	0.48

7.3.2. Segment kinetic energy

7.3.2.1. Total segment kinetic energy

The interaction between weight transfer style, club type and segment was non-significant (F (2.28, 36.54) = 4.84, p = 0.10, r = 0.34). However, the medium effect size suggested that, when the driver was used, it is possible that the Reverse players produced larger peak Arms KE compared with the Front Foot players and compared to when the 5 and 9 irons were used (Figure 7.7 and Figure 7.9). The interaction between weight transfer style and segment (F (1.30, 20.74) = 0.88, p = 0.46, r = 0.20) and the main effect of weight transfer style (F (1, 16) = 0.04, p = 0.84, r = 0.05) were non-significant. Therefore, the significant main effect of segment (F(1.30, 20.74) = 861.55, p < 0.01, r = 0.99) suggested that for both weight transfer styles the magnitude of segment KE increased sequentially from the Lower Body to the Club (Table 7.6). Furthermore, the main effect of club type (F(2, 32) = 64.96, p < 0.01, r = 0.82) suggested that, for both weight transfer styles and all segments, the magnitude of peak total segment KE increased from the 9 iron to the driver (Table 7.6).

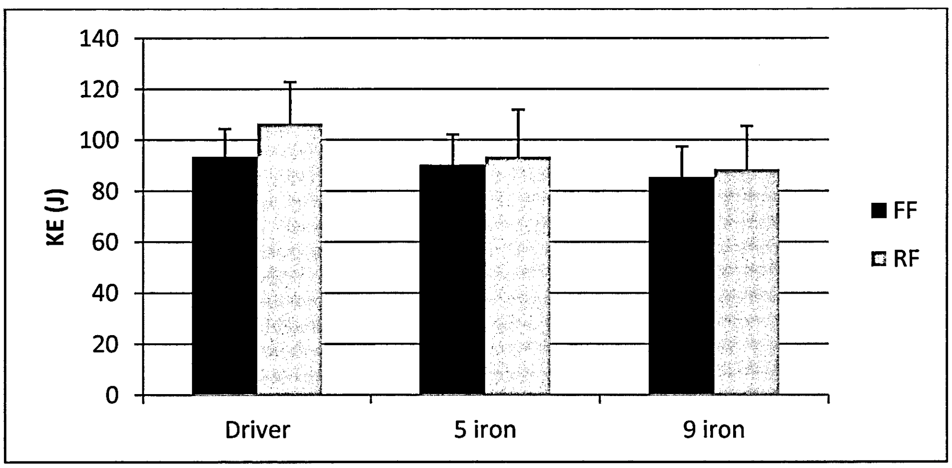


Figure 7.7 The magnitude of peak Arms KE for both groups with each club type.

There was a significant interaction between weight transfer style and segment (F(3, 48) = 5.60, p = 0.04, r = 0.32) for the timing of peak total KE which was related to peak total Lower Body KE occurring significantly earlier (p < 0.05) in the downswing for Reverse strategy golfers (0.703 ± 0.084 MT) than for Front Foot golfers (0.794 ± 0.092 MT) with all clubs (Figure 7.8). There was also a main effect of segment (F(3, 48) = 70.61, p < 0.01, r = 0.77) which indicated that peak total Lower Body, Upper Body and

Arms KE occurred simultaneously whilst peak total Club KE occurred significantly later in the downswing, just before impact. As the interaction between weight transfer style, segment and club type was non-significant (F(2.44, 39.07) = 0.08, p = 0.95, r = 0.05) the results suggested that this simultaneous peak of total body segment KE was apparent for both weight transfer styles when all clubs were used (Table 7.7).

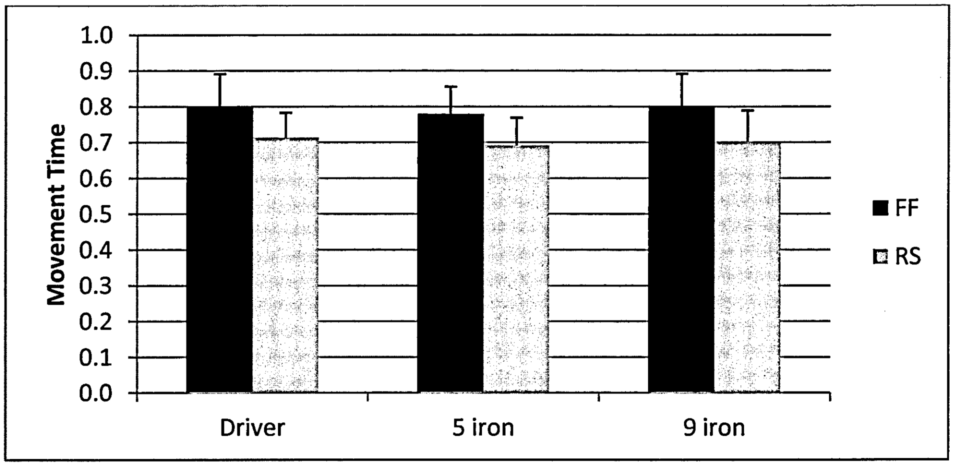


Figure 7.8 Timing of mean total Lower Body KE for both groups with each club type.

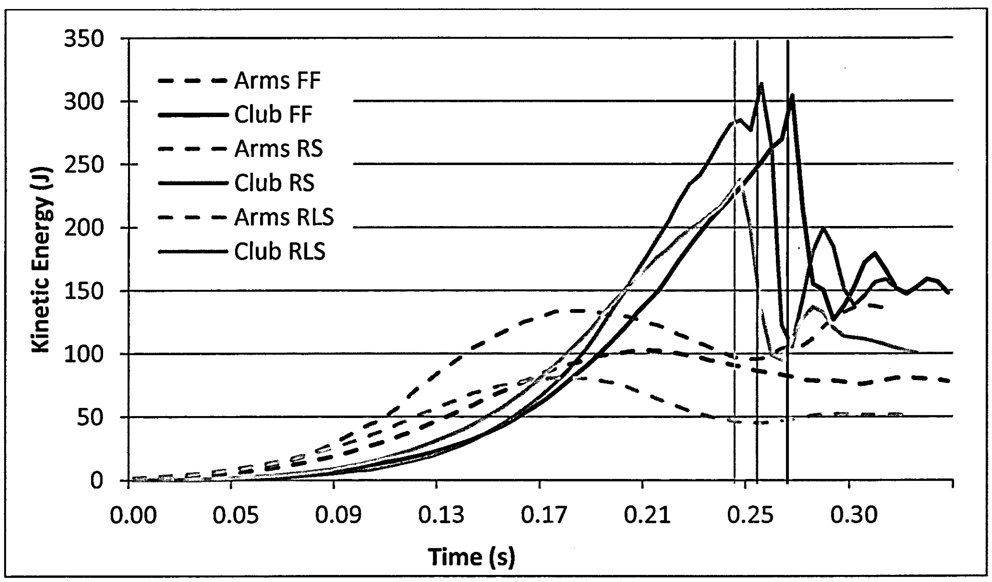


Figure 7.9 Total Arms and Club KE for representative Front Foot, skilled Reverse (RS) and less skilled Reverse (RLS) golfers using the driver. Time is taken from the Top-of-Backswing to Mid-Follow-Through (impact is highlighted for each player).

7.3.2.2. Translational segment kinetic energy

The interaction between weight transfer style, club and segment was non-significant (F (2.27, 36.57) = 20.49, p = 0.14, r = 0.23). The interaction between weight transfer style and segment (F (1.16, 18.59) = 0.34, p = 0.36, r = 0.18) and the main effect of weight transfer style (F (1, 16) = 0.01, $p = 0.98$, r = 0.01) were also not significant. Therefore, as there was a significant main effect of segment (F(1.16, 18.59) = 835.56, $p < 0.01$, r = 0.99) the results suggested that regardless of weight transfer style, the magnitude of peak translational KE increased sequentially from the most proximal to the most distal segment (Table 7.6). There was also a main effect of club type (F(2, 32) = 51.02, $p < 0.01$, r = 0.78) which suggested that regardless of segment and weight transfer style the magnitude of peak total segment KE increased from the 9 iron to the driver.

There was a significant main effect of segment on the timing of peak translational segment KE (F(3, 48) = 38.54, $p < 0.01$, r = 0.67) . Combined with the non-significant interaction between weight transfer style and segment (F(3, 48) = 0.89, $p = 0.46$, r = 0.14) this indicated that for both weight transfer styles, peak translational Lower Body, Upper Body and Arms KE occurred at the same time whilst peak translational Club KE occurred significantly later in the downswing (Table 7.7). Despite these similarities, there was a significant interaction between segment and club type (F(2.45, 39.28) = 2.68, $p = 0.02$, r = 0.25). This interaction was related to peak translational Lower Body and Upper Body KE occurring significantly ($p < 0.05$) later with the driver compared with the 5 iron and 9 iron (Figure 7.10a). Further examination of the results suggested that the later translational KE peaks produced with the driver were related to peak Lower Body and Upper Body KE occurring later in the downswing for Front Foot players (Figure 7.10b and Figure 7.11).

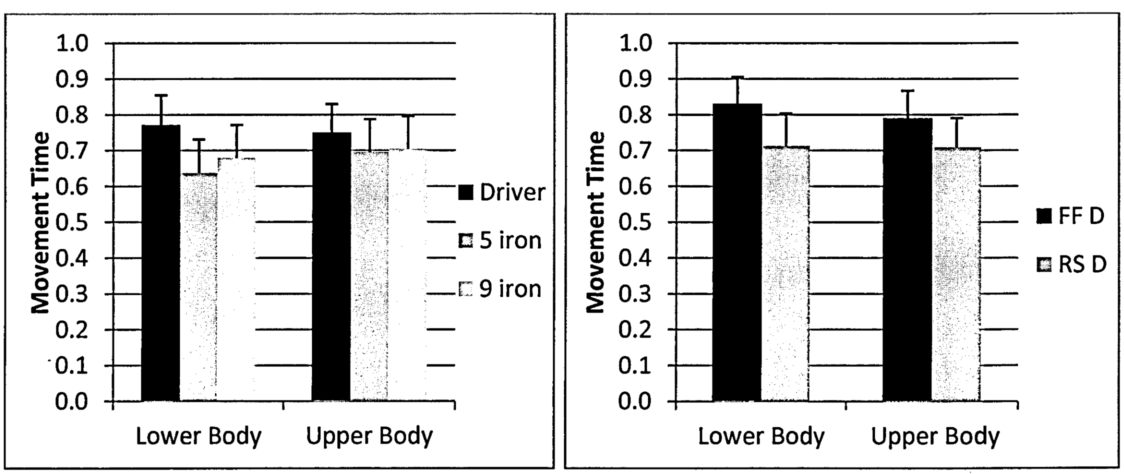


Figure 7.10 (a) left: Timing of mean peak translational Lower Body and Upper Body KE with all three clubs. (b) right: Timing of peak translational Lower Body and Upper Body KE for both groups with the driver.

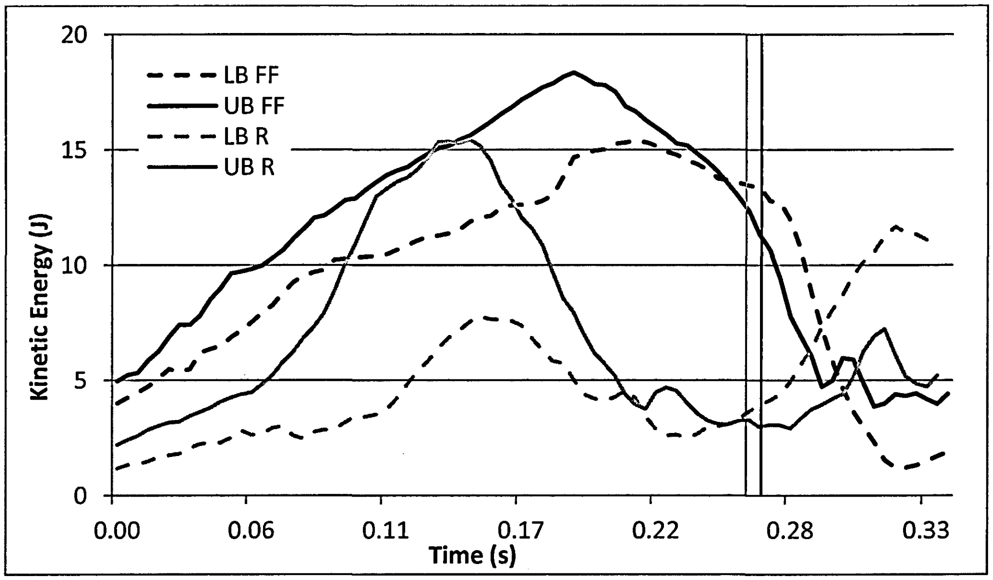


Figure 7.11 Translational Lower Body and Upper Body KE for a representative Front Foot and Reverse golfer using the driver. Time is taken from the Top-of-Backswing to Mid-Follow-Through (impact is highlighted for each player).

7.3.2.3. Local rotational segment kinetic energy

The interaction between weight transfer style, club and segment was non-significant (F (2.90, 46.34) = 0.45, p = 0.71, r = 0.10). However, there was a significant main effect of segment (F(2.09, 33.49) = 313.97, p < 0.01, r = 0.95) which suggested that regardless of weight transfer style and club type peak local rotational Upper Body KE was significantly larger than peak local rotational Lower Body and Arms KE. Furthermore, this main effect of segment indicated that peak local rotational Club KE was also significantly larger than peak local rotational Upper Body KE. As the interactions between weight transfer style and segment (F (2.09, 33.49) = 0.60, p = 0.62, r = 0.13) and weight transfer style and club type (F (1.39, 22.26) = 0.61, p = 0.50, r = 0.16) and the main effect of weight transfer style (F (1, 16) = 0.01, p = 0.98, r = 0.01) were not significant, the results suggested that this pattern was evident for both weight transfer styles when all three clubs were used.

There was a significant interaction between weight transfer style and segment (F(2.01, 32.14) = 3.65, p = 0.04, r = 0.32) for the timing of peak local rotational KE which related to peak local rotational Lower Body KE occurring significantly later (p < 0.05) for the Front Foot players (0.786 ± 0.076 MT) compared with the Reverse strategy players (0.695 ± 0.059 MT) - Figure 7.12 and Figure 7.14. This significant interaction between weight transfer style and segment was also related to peak local rotational Arms KE occurring significantly later ($p < 0.05$) for the Front Foot players (0.921 ± 0.077 MT) compared with the Reverse strategy players (0.822 ± 0.087 MT) - Figure 7.13 and Figure 7.14. Furthermore, the significant main effect of weight transfer style ($F(1, 16) = 19.73$, $p < 0.01$, $r = 0.74$) indicated that in general, peak local rotational segment KE occurred later for the Front Foot players than for the Reverse Strategy players . There was also a main effect of segment ($F(2.01, 32.14) = 70.20$, $p < 0.01$, $r = 0.83$) which was related to peak local rotational Lower Body and Upper Body KE occurring at the same time and peak local rotational Upper Body, Arms and Club local rotational KE occurring in a proximal-to-distal sequence for both weight transfer styles (Table 7.7).

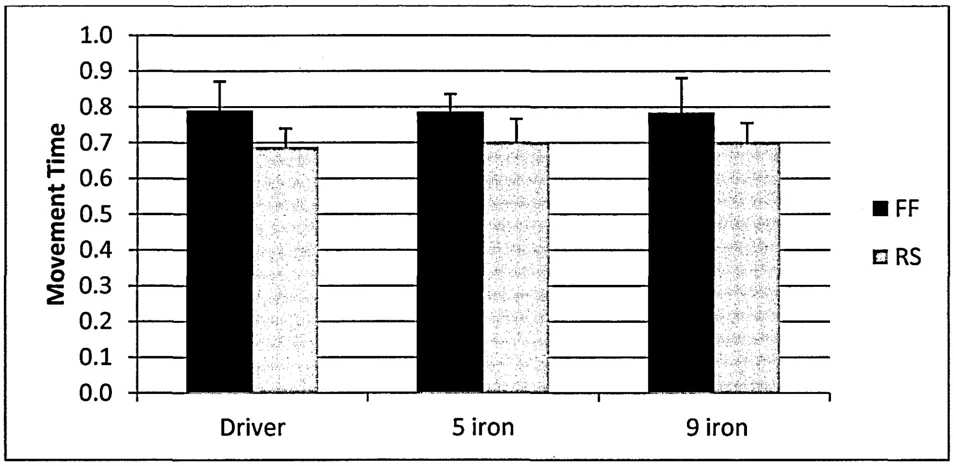


Figure 7.12 Timing of mean peak local rotational Lower Body KE for both groups with each club type.

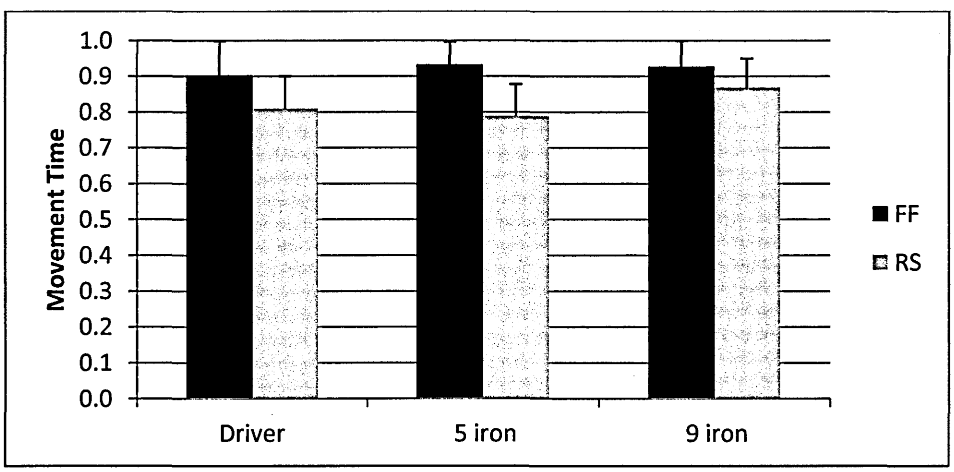


Figure 7.13 Timing of mean peak local rotational Arms KE for both groups with each club type.

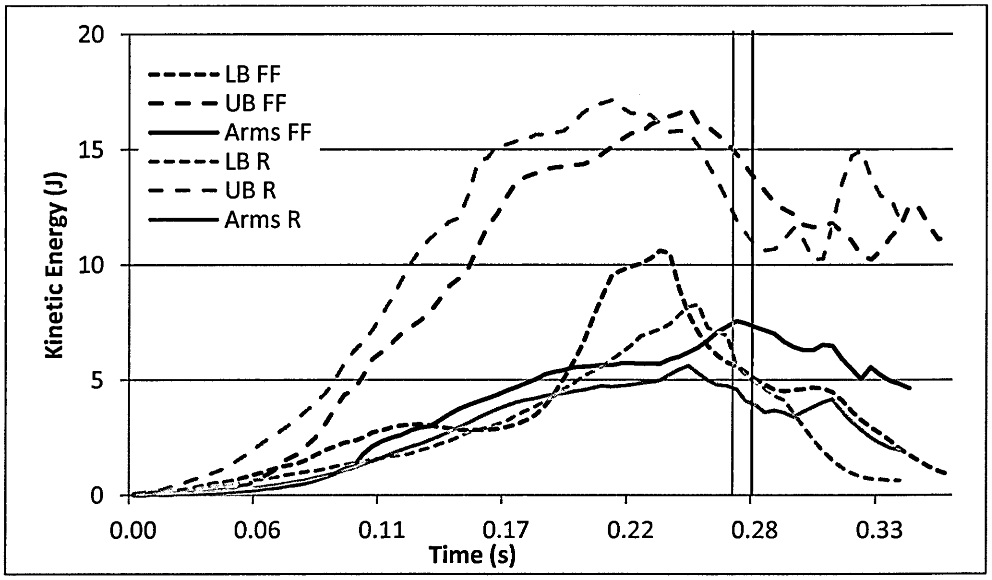


Figure 7.14 Local rotational Lower Body, Upper Body and Arms KE for a representative Front Foot and Reverse golfer using the driver. Time is taken from the Top-of-Backswing to Mid-Follow-Through (impact is highlighted for each player).

7.3.2.4. Remote rotational segment kinetic energy

The interaction between weight transfer style, club and segment was non-significant (F (2.11, 33.72) = 2.08, p = 0.14, r = 0.24). However, there was a significant main effect of segment (F(1.20, 19.21) = 323.60, p < 0.01, r = 0.97) which suggested that regardless of weight transfer style and club type peak remote rotational Upper Body KE was significantly smaller than peak remote rotational Lower Body and Arms KE. As the interaction between weight transfer style and club type was also not significant (F (2, 32) = 0.51, p = 0.59, r = 0.13) the results indicated that this pattern was also evident with all three clubs.

There was a significant interaction between weight transfer style and segment for the timing of peak remote rotational KE (F(2, 32) = 2.18, p = 0.01, r = 0.25). This interaction was related to peak remote rotational Arms KE occurring significantly later (p = 0.05) for the Front Foot players (0.932 ± 0.055 MT) compared with the Reverse strategy players (0.824 ± 0.068 MT) - Figure 7.15 and Figure 7.16. There were also main effects of segment (F(2, 32) = 13.10, p < 0.01, r = 0.54) and weight transfer style (F(1, 16) = 4.77, p = 0.04, r = 0.48). The main effect of segment suggested that for both weight transfer styles peak remote rotational Lower Body and Upper Body KE occurred at the same time whilst peak remote rotational Upper Body and Arms KE occurred in a proximal-to-distal sequence (Table 7.7). The main effect of weight transfer style indicated that, in general peak remote rotational KE occurred significantly later for the Front Foot players (Table 7.7).

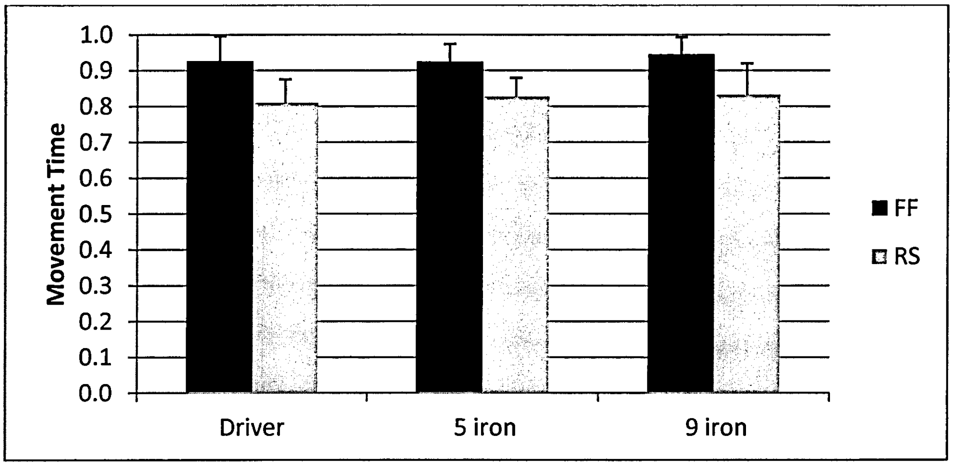


Figure 7.15 Timing of mean peak remote rotational Arms KE for both groups with each club type.

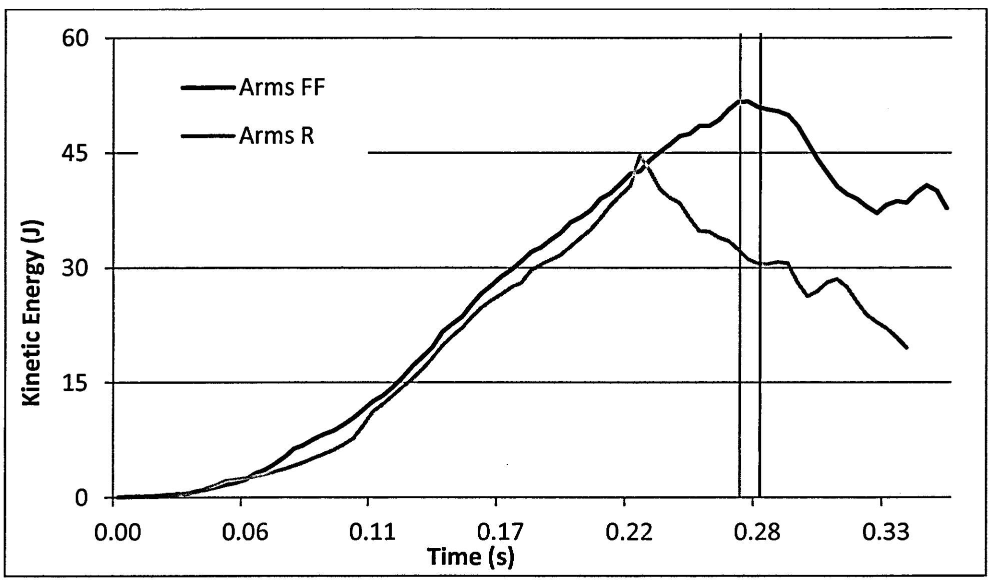


Figure 7.16 Remote rotational Arms KE for a representative Front Foot and Reverse golfer using the driver. Time is taken from the Top-of-Backswing to Mid-Follow-Through (impact is highlighted for each player).

Table 7.6 Mean peak Total, Translational, Local Rotational and Remote Rotational KE for both weight transfer st

	Front Foot			Reverse S
	Driver	5 iron	9 iron	Driver	5 irc
Total KE (J)					
Lower Body	33.70 ± 8.91	27.69 ± 7.02	26.04 ± 5.78	26.03 ± 11.28	22.64 ±
Upper Body	34.32 ± 6.82	29.22 ± 7.14	26.95 ± 7.64	31.53 ± 8.44	26.77 ±
Arms	93.42 ± 10.87	90.07 ± 11.93	85.35 ± 12.00	106.49 ± 16.32	93.39 ±
Club	271.50 ± 32.64	259.28 ± 29.99	225.60 ± 33.40	276.56 ± 35.94	259.12 ±
Translational KE (J)					
Lower Body	14.02 ± 4.12	10.52 ± 3.01	9.46 ± 2.93	9.92 ± 7.06	6.39 ± 1
Upper Body	14.43 ± 3.63	11.12 ± 4.29	11.07 ± 4.26	10.56 ± 3.53	9.53 ±
Arms	52.70 ± 8.80	50.82 ± 8.51	48.14 ± 7.50	57.58 ± 14.58	53.21 ±
Club	320.56 ± 28.06	223.78 ± 25.65	193.30 ± 29.53	232.40 ± 34.49	222.43 ±
Local Rotational KE (J)					
Lower Body	10.45 ± 2.94	9.53 ± 2.35	8.87 ± 2.68	8.78 ± 3.55	8.34 ±
Upper Body	20.17 ± 5.48	18.50 ± 4.92	16.78 ± 4.59	19.72 ± 6.88	17.89 ±
Arms	5.95 ± 1.37	5.51 ± 1.38	5.29 ± 1.04	6.19 ± 1.92	5.64 ±
Club	41.08 ± 5.10	35.59 ± 5.54	32.35 ± 4.38	42.13 ± 4.69	36.99 ±
Remote Rotational KE (J)					
Lower Body	12.26 ± 3.39	10.56 ± 3.34	10.04 ± 3.14	10.31 ± 4.76	9.59 ±
Upper Body	2.38 ± 0.62	1.87 ± 1.10	1.75 ± 0.98	2.29 ± 1.16	1.56 ±
Arms	41.71 ± 7.09	37.93 ± 7.05	36.84 ± 5.78	46.16 ± 11.21	41.58 ±

Table 7.7 The timing of mean peak Total, Translational, Local Rotational and Remote Rotational KE for both weight tra

	Front Foot			Reverse St	
	Driver	5 iron	9 iron	Driver	5 iror
Total KE (J)					
Lower Body	0.801 ± 0.091	0.779 ± 0.076	0.802 ± 0.089	0.714 ± 0.069	0.692 ± 0
Upper Body	0.775 ± 0.028	0.754 ± 0.077	0.763 ± 0.074	0.774 ± 0.067	0.757 ± 0
Arms	0.771 ± 0.039	0.781 ± 0.057	0.778 ± 0.035	0.744 ± 0.105	0.759 ± 0
Club	0.987 ± 0.005	0.988 ± 0.007	0.990 ± 0.004	0.986 ± 0.008	0.989 ± 0
Translational KE (J)					
Lower Body	0.830 ± 0.076	0.639 ± 0.094	0.688 ± 0.092	0.713 ± 0.091	0.637 ± 0
Upper Body	0.790 ± 0.077	0.697 ± 0.091	0.707 ± 0.092	0.709 ± 0.082	0.695 ± 0
Arms	0.731 ± 0.038	0.744 ± 0.043	0.735 ± 0.044	0.689 ± 0.057	0.714 ± 0
Club	0.987 ± 0.05	0.988 ± 0.007	0.990 ± 0.004	0.987 ± 0.006	0.990 ± 0
Local Rotational KE (J)					
Lower Body	0.790 ± 0.082	0.785 ± 0.051	0.784 ± 0.097	0.688 ± 0.052	0.699 ± 0
Upper Body	0.777 ± 0.045	0.763 ± 0.064	0.765 ± 0.058	0.761 ± 0.074	0.747 ± 0
Arms	0.903 ± 0.096	0.933 ± 0.063	0.927 ± 0.071	0.809 ± 0.091	0.789 ± 0
Club	0.984 ± 0.012	0.986 ± 0.006	0.990 ± 0.003	0.983 ± 0.009	0.986 ± 0
Remote Rotational KE (J)					
Lower Body	0.780 ± 0.070	0.776 ± 0.121	0.816 ± 0.104	0.682 ± 0.115	0.717 ± 0
Upper Body	0.800 ± 0.088	0.770 ± 0.126	0.760 ± 0.153	0.811 ± 0.154	0.766 ± 0
Arms	0.926 ± 0.069	0.924 ± 0.049	0.944 ± 0.048	0.810 ± 0.065	0.827 ± 0

7.4. Discussion

The aim of this study was to examine the effect of weight transfer style on the sequencing of segmental movements in golf swings performed with a driver, 5 iron and 9 iron. It was hypothesised that, in general, both weight transfer style groups would exhibit similar movement characteristics. However, as the movements of the Upper Body are closely related to the movement of a golfer's centre of pressure (Beak *et al.*, 2013; Okuda *et al.*, 2010), it was also hypothesised that compared with Reverse players, Front Foot players would produce larger magnitudes of Upper Body KE which occurred later in the downswing.

7.4.1. Weight transfer style

Two distinct weight transfer styles, the Front Foot and Reverse have been identified when the driver (Ball and Best, 2007a), 3 iron and 7 iron are used (Ball and Best, 2011). The Front Foot golfers exhibit a weight transfer pattern recommended in the coaching literature (Leadbetter, 1995) whilst, reverse golfers produce a backward movement during the downswing, such that the weight is positioned near mid-stance at ball contact and continues towards the back foot during the follow-through. It has been suggested that approximately 30% of all golfers produce the Reverse strategy (Ball and Best, 2007a). A similar ratio was identified in this study as 12 of the 36 recruited players were classified as Reverse strategy players when the driver was used. This study also presented novel findings as nine Reverse strategy players were identified when the 5 and 9 irons were used (Table 7.4 and Table 7.5). The existence of different weight transfer styles was expected for the 5 iron as the distance requirement is greater than that of a 7 iron for which the Reverse strategy has previously been identified (Ball and Best, 2011). It was also anticipated for the 9 iron as, although the distance requirement is less than that of the 7 iron, Chapter V indicated that similar swing mechanics were generated with the 5 iron and 9 iron.

The validity of both weight transfer styles has been supported in previous studies as it has been suggested that players adopting both styles had similar mean handicaps and produced comparable clubhead velocities (Ball and Best, 2007a; Ball and Best, 2011).

In this study, there were also no statistical differences between the mean handicaps of players in both groups. Both groups also included highly skilled golfers which provides further support for the validity of both groups and suggests that with all three clubs neither style is a technical error. In this study, 33 of the 36 golfers (92%) analysed adopted the same weight transfer style when all three clubs were used. This also provides support for Ball and Best (2011) who also reported that 44 of 46 golfers (96 %) adopted the same style with three different clubs. However, in both studies a slightly smaller proportion of players generated the Reverse strategy when the irons were used. This would appear logical as, positioning CPy% further back when the longer driver club is used is a strategy adopted by golfers to enable a more stable position to be adopted (Cooper *et al.*, 1974: Jenkins, 2008).

7.4.2. Magnitude of segment kinetic energy

As the magnitude of peak total segment KE increased sequentially from the Lower Body to the club it is possible that the swings of both groups conformed with the principle of the proximal-to-distal sequence. Despite the two weight transfer groups exhibiting different CPy% patters during the downswing, both groups produced similar magnitudes of peak total segment KE and its components. As it has previously been reported that similar clubhead velocities are produced by both weight transfer style groups (Ball and Best, 2007a; Ball and Best, 2011) the similar magnitudes of peak total segment KE were expected for the Club segment when all three clubs were used. However, it was not anticipated that similar magnitudes of peak segment KE would be generated for the three body based segments.

As the Front Foot group produced a significantly larger CPy% range (Ball and Best, 2007a; Ball and Best, 2011) and the movements of the Upper Body are primarily responsible for weight transfer (Beak *et al.*, 2013; Okuda *et al.*, 2010), it was hypothesised ($H_{ALTERANTIVE\ 2}$) that the Front Foot group would produce larger peak total and translational Upper Body KE. However, similar magnitudes of peak total and translational Upper Body KE were reported for both groups. Therefore, the results suggest that forward translational movement of the Upper Body is not responsible for the differences in CPy% movements. Research has also stressed that, rather than positioning the Upper Body over the front foot at ball contact, it is more important to position the central hub over the back leg at the top of backswing as this enables the necessary lateral translation and forward velocity of the central hub to be produced during the downswing (Horan *et al.*, 2010; Jenkins, 2008). Therefore, the results of this study suggest that regardless of weight transfer style, golfers should focus on positioning their weight over the back foot at top of backswing rather than positioning their weight over the front foot at impact.

The majority of results indicated that similar magnitudes of peak segment KE were produced by both groups of golfers. However, when total KE was examined, a medium effect size was produced for the interaction between weight transfer style, segment and club type. This was related to larger peak total Arms KE being produced by the Reverse golfers when the driver was used (Figure 7.7). Although not significant, the difference between peak total Arms KE when the driver was used (13.07 J) was larger than the minimal difference (9.10 J) required for the result to be meaningful (Table 4.1). The medium effect size also suggests that, had a larger sample size been examined, it is likely that a significant interaction would have reported for this effect (Hopkins *et al.*, 2008). Furthermore, the results suggested that the five skilled Reverse strategy players produced considerably larger peak total Arms KE (120.20 ± 13.15 J) than the four less skilled Reverse players (85.86 ± 7.50 J). These highly skilled players also produced larger total Club KE (293.59 ± 38.52 J) than the lower skilled players (255.27 ± 19.43 J). Although fast arm movements have been associated with fast, skilled swings for players of all abilities (Chapter VI; Chu, Sell and Lephart, 2011, Hume, Keogh and Reid, 2005: Zheng *et al.*, 2008) the results of this study suggest that the generation of large total Arms KE is important when Reverse players are required to produce high clubhead velocities.

7.4.3. Timing of segment kinetic energy

For both groups, peak total segment KE appeared to conform to the principle of optimal coordination of partial momenta (Van Gheluwe and Hebbelinck, 1985) as peak total body segment KE occurred simultaneously whilst peak Club KE occurred significantly later in the downswing, just before impact. This finding supports the results presented in the literature (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008) and combined with the results of Chapter VI, suggests that regardless of playing standard and weight transfer strategy, the timing of peak total segment KE follows this pattern. When the components of total segment KE have been examined, timing differences have been identified between skilled and less skilled players (Chapter VI) and between the two weight transfer style groups. It has also been suggested that golf swings are underpinned by sequential increases in rotational components of segment KE from the Upper Body to the Club. However, it would appear that the different timing sequences and sequential movements have no effect on the timing of peak total segment KE which always conforms with the principle of optimal coordination of partial momenta (Kreighbaum and Barthels, 1985).

In a similar manner to peak total KE, for both groups, peak translational KE appeared to conform to the principle of optimal coordination of partial momenta (Van Gheluwe and Hebbelinck, 1985). This pattern was also evident in Chapter VI for the category 2 and 3 players. However, for the category 1 players, peak translational KE occurred in a proximal-to-distal sequence from the Upper Body to the Club. Therefore, it is likely that neither swing style group generated the pattern exhibited by skilled golfers as both contained less skilled, category 2 and 3 players. Furthermore, these results suggest that regardless of swing style, skilled golfers should aim to produce a proximal-to-distal sequence of translational KE from the Upper Body to Club. This sequence may enable players to generate increased total Club KE via the kinetic link principle (Kreighbaum and Barthels, 1985) and stretch shortening cycle (Komi, 2000).

For both groups, peak local and remote rotational Lower Body and Upper Body KE occurred simultaneously. Both components of segment KE also followed a proximal-to-distal sequence (Putnam, 1993) from the Upper Body to the most distal segment. Similar results were also reported in Chapter VI as peak local and remote rotational KE followed a sequential pattern for players of all abilities. Therefore, the results suggest that the sequential movements of the Upper Body and Arms cause musculature of the shoulder and trunk to be stretched and enable players of all abilities and from both groups to generate clubhead speed by taking take advantage of the stretch shortening cycle (Chu, Sell and Lephart, 2010; Komi, 2000).

Despite these similarities between the two weight transfer styles, the results indicated that different timing strategies were used. In general, these differences related to peak segment KE occurring later in the downswing for the Front Foot group. More specifically, peak total, translational (driver only) and local rotational Lower Body, peak translational (driver only) Upper Body and peak local rotational and remote rotational Arms KE all occurred significantly later for the Front Foot players compared with the Reverse players. Analysis of the results suggested that, for all club types, these differences occurred between the early (FF = 0.646 ± 0.074 MT; RS = 0.637 ± 0.067 MT) and late (FF = 0.856 ± 0.031 MT; RS = 0.844 ± 0.032 MT) downswing events when CPy% for the Front Foot players continued forwards whilst for the Reverse group CPy% moved towards the back foot.

The early peak in total Lower Body KE produced by the Reverse players provides evidence that different movement strategies are used by the two weight transfer groups. This early peak in total Lower Body KE when the driver was used was related to peak translational Lower Body KE occurring earlier for the Reverse players. However, it is probable that the consistent early peak in total Lower Body KE was related to peak local rotational Lower Body KE occurring earlier in the downswing for the Reverse players. As large axial rotations are produced by the pelvis early in the downswing (Horan *et al.*, 2010; Myers, *et al.*, 2008) it is likely that the earlier peak local rotational Lower Body KE was related to the timing of this rotation. This early peak in local rotational KE may have enabled the Reverse players to take advantage of the stretch shortening cycle (Komi, 2000) and allowed their Lower Body to become static support for the more distal segments earlier in the downswing (Nesbit and Serrano, 2005).

When the driver was used, peak translational Lower Body and Upper Body KE occurred just after early downswing for the Reverse players, significantly earlier than for the

Front Foot players. It has been suggested that during the early downswing the swing hub translates almost exclusively in the direction of the target (Burden, Grimshaw and Wallace, 1998) and that this movement produces large clubhead velocities (Jorgensen, 1994; Miura, 2001). Therefore, the early peak in translational KE when the driver was used would appear to be related to the lateral movement of the swing hub. Furthermore, these differences in the timing of peak translational Lower Body and Upper Body KE were not evident between the two groups when the 5 iron and 9 iron were used. Therefore, the results suggest that this movement and the early peak in translational Lower Body and Upper Body KE would appear to be crucial when Reverse players are required to produce larger clubhead velocities with the driver.

Regardless of club type, peak local rotational and remote rotational Arms KE occurred significantly earlier in the downswing for the Reverse players. For this group, the large forward motion of the centre of mass early in the downswing may have been a result of the forward motion of the Arms (Burden, Grimshaw and Wallace, 1998). This early peak in Arms KE may also have allowed more time for Reverse players to transfer KE from the body segments to the Club. For the Front Foot players, the later peaks in local and remote rotational Arms KE were potentially related to the generally later peaks in KE for the other both segments. For this group, the later peaks in Arms KE may have been required to enable the KE generated by the more proximal segments to be transferred to the Club and to allow these golfers to take advantage of the proximal-to-distal sequence (Putnam, 1993) and stretch shortening cycle (Komi, 2000; Van Ingren and Schenau, 1984).

7.5. Conclusion

For each segment, similar magnitudes of peak total KE and its components were produced by the Front Foot and Reverse golfers. Regardless of weight transfer style, the magnitude of peak total KE also increased sequentially from the Lower Body to the Club. Despite these similarities this study highlighted the importance of identifying weight transfer styles within the golf swing as different movement strategies were also used by the Front Foot and Reverse players. More specifically, the generation of large total Arms KE was essential when Reverse players were required to produce high clubhead velocities. Furthermore, peak translational, local and remote rotational Lower Body KE occurred significantly earlier for the reverse players. The rotational components of peak total Arms KE also occurred significantly earlier for the Reverse golfers which may have also have allowed more time for KE to be transferred from the body segments to the Club.

8. Chapter VIII – Summary and Discussion

8.1. Introduction

The proximal-to-distal sequence has been associated with both mechanical and muscular rewards which enable high speed at the distal end of a linked system to be produced. As such, most striking and throwing movements which attempt to maximise distal end speed are characterised by sequential motions of body segments. Since Cochran and Stobbs (1968) also suggested that the most effective golf swings adopt a proximal-to-distal sequence, the sequencing of body segments has become an important theme in golf swing instruction and research. It is believed that sequential movements of body segments can ensure that clubhead velocity increases throughout the downswing and achieves its maximum at impact.

Segmental sequencing in the golf swing has predominantly been examined in terms of the summation of speed principle using analyses of segment angular velocities. However, it has been proposed that analyses of segment KE are more appropriate as they consider the inertial properties of segments, enable linear and angular components of movement to be considered separately and as a single entity and are extremely sensitive to subtle changes in technique. Although the sequencing of segment KE in the golf swing has been examined, research has predominantly examined the sequencing of total segment KE. Despite the potential for analyses of translational and rotational components of segment KE to provide more detailed descriptions, these components have only been analysed for golf swings performed with a driver. Therefore, the overall purpose of this thesis was to examine the sequencing of segment KE in the golf swing.

This chapter provides an overall discussion and summary of this thesis. Firstly, a brief summary of Chapters III, IV, V, VI and VII is provided. Secondly, the implications of findings in the thesis are discussed in the context of current and future research regarding the sequencing of body segment movements in the golf swing. Thirdly, the limitations of the thesis are acknowledged before, a thesis conclusion is presented.

8.1. Chapter summaries

8.1.1. Chapter III

Chapter III presented the method that was used to undertake analyses of segment KE in the golf swing. This chapter also sought to examine methodological considerations associated with the measurement of segment KE. This chapter was divided into four main sections. Initially, Chapter III described the geometric model - adapted from Yeadon (1990) - that was used to provide estimations of subject specific BSIP. Chapter III also provided a detailed description of the geometric model used to estimate hand inertial parameters. This model represented the base of the hand using a stadium solid and the fingers wrapped around a golf club as a segment of a hollow cylinder. To help support the use of this model, supplementary studies (Appendix I and Appendix II) assessed the reliability and accuracy of the BSIP estimates made using this modelling approach. It was suggested that inertial parameter estimates made by the same examiner could achieve acceptable reliability. These studies also indicated that the majority of limb and trunk inertial parameters could be accurately estimated using the described geometric model. Finally, it was suggested that the stadium solid and segment of a hollow cylinder were capable of estimating hand mass with a mean error of only 1.57 ± 6.44% (Appendix IV).

Chapter III then outlined the method that was used to collect kinematic data using a Polhemus electromagnetic tracking system. As the attachment of electro-magnetic sensors to body segments has the potential to encumber golfers and restrict their movements (Wright, 2008) the effect of electromagnetic sensor attachment on swing and launch parameters was examined (Appendix V). Using ball flight and clubhead characteristics from a Trackman Pro launch monitor, this study concluded that electromagnetic sensor attachment did not have a meaningful effect on golf swing mechanics. The final section of Chapter III explained the technique that was used to calculate the magnitude and timing of segment KE in the golf swing for the four grouped segments. The effect of BSIP estimation errors on measures of segment KE in the golf swing was then examined (Appendix VII). This study suggested that the magnitude of peak segment KE was much more sensitive to segment inertial parameter estimation errors than the timing of peak segment KE. Furthermore, it was indicated that the magnitude of peak total Arms KE was most sensitive to errors produced in the estimation of inertial parameters.

8.1.2. Chapter IV

Chapter IV sought to examine the reliability of measures of the magnitude and timing of peak segment KE in the golf swing. Although it has been suggested that the analysis of segment KE is the most appropriate technique to examine the sequencing of body segments (Anderson, Wright and Stefanyshyn, 2006) the reliability associated with its measurement has not been reported. Repeated measurements of segment KE in the golf swing may show day-to-day differences due to changes in the golfer's technique (Drust *et al.*, 2005). Variations in measures of segment KE in the golf swing may also originate from the collection of 3D translational and rotational kinematic data, the definition and computation of body segment axes and from the estimation of inertial parameters.

The magnitude and timing of peak total, translational and rotational KE were measured with high reliability for the majority of segments. The similar mean values, acceptable-good ICCs and low SEM provided support for the examination of the proximal-to-distal sequence using analyses of segment KE. However, the magnitude of peak translational and subsequently total Lower Body KE was measured with questionable reliability when the 5 and 9 irons were used. This suggested that, at least part of the observed difference in future examinations of total Lower Body KE may be attributable to sources of variability unrelated to the swing. The results of Chapter IV therefore, suggested that future studies should consider the SEM and MD reported in this study when interpreting the results of analyses of segment KE.

8.1.3. Chapter V

Previous research has predominantly discussed the sequencing of total segment KE in the golf swing (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.*, 2008). Furthermore, despite this technique allowing the translational the rotational components of segment KE to be examined separately these components have only been considered with golf swings performed with a driver (Anderson, 2007). Therefore, Chapter V aimed to examine the effect of club type on the sequencing of peak total, translational and rotational segment KE in the golf swing.

Regardless of club type, the magnitude of peak total segment KE increased sequentially from the Lower Body to the club. However, the magnitude of peak segment KE for each component did not occur in a proximal-to-distal order as the results highlighted the importance of producing translational Legs and local rotational Upper Body KE. The results of Chapter V also indicated that club type had a significant effect on the golf swings of highly skilled players. Specifically, the results indicated that skilled golfers produced larger peak translational Arms KE and peak local rotational Upper Body KE when the distance requirement of the shot increased.

With all three clubs, the timing of peak total KE conformed to the principle of optimal coordination of partial momenta as peak total body segment (Lower Body, Upper Body and Arms) KE peaked simultaneously at approximately 74% MT whilst total club KE peaked just before impact. However, regardless of club type, sequential timing was evident for the translational and rotational components of segment KE from the Upper Body to the Club. Finally, the results suggested that when the driver was used and the distance requirement of the shot increased, peak translational Lower Body and Upper Body KE occurred significantly later in the swings of highly skilled golfers.

8.1.4. Chapter VI

Although Chapter V developed an understanding of the sequencing of segment KE with different clubs, only highly skilled players were considered. Therefore, Chapter VI sought to examine the effect of playing standard on the sequencing of movements in the golf swing. Regardless of playing standard and club type, the magnitude of peak total KE increased sequentially from the Lower Body to the Club. However, the findings generally supported the notion that skilled golfers produce faster golf swings than less skilled players. The highly skilled golfer's generated larger peak total Arms and Club KE compared with category 2 and 3 golfers. The results suggested that larger translational and remote rotational Arms KE and local rotational Upper Body KE may contribute to the larger total Club KE produced by skilled players. These results also emphasised the benefits of examining the golf swing using analyses of segment KE as they highlighted the importance of considering translational movements and 3D rotations.

For all playing standards, peak total KE for the body segments occurred simultaneously whilst peak total Club KE occurred significantly later in the downswing just before impact. Despite these similarities, significant differences were identified between the timing of body segment movements in the golf swing. The results highlighted the importance of a sequence of translational KE from the Upper Body to the Club and of an early peak in local rotational Upper Body KE as these movement strategies were only used by category 1 golfers. It was suggested that these movement strategies enabled skilled golfers to generate increased total Club KE via the kinetic link principle and stretch shortening cycle.

8.1.5. Chapter VII

By measuring weight distribution in the direction of the shot (CPy %) Ball and Best (2007a) identified Front Foot and Reverse weight transfer patterns. For both weight transfer styles it was speculated that golf swings were underpinned by the proximal-to-distal sequence as large forward movements of CPy % were proposed to develop system energy, which was transferred to the clubhead at impact (Ball and Best, 2007b). This discussion was limited by the absence of kinematic data (Ball and Best, 2011; Hellstrom, 2009). Therefore, Chapter VII sought to examine the effect of weight transfer style on the sequencing of segment KE.

For both weight transfer styles the magnitude of peak total KE increased sequentially from the Lower Body to the Club. For each segment, similar magnitudes of peak total KE and its components were also produced by the Front Foot and Reverse golfers. Despite these similarities this study highlighted the importance of identifying weight transfer styles within the golf swing as different movement strategies were also used by the Front Foot and Reverse players. More specifically, the generation of large total Arms KE was essential when Reverse players were required to produce high clubhead velocities.

For both weight transfer styles, peak total body segment KE occurred simultaneously whilst peak Club KE occurred significantly later in the downswing, just before impact. However, peak segment KE generally occurred later in the downswing for the Front Foot group. More specifically, peak total, translational and local rotational Lower Body and peak translational Upper Body and peak local rotational and remote rotational Arms KE all occurred significantly later for the Front Foot players compared with the Reverse players. Chapter VII suggested that the different timing strategies exhibited for the Arms segments may have allowed more time for Reverse players to transfer KE from the body segments to the Club

8.2. Practical implications of findings

The findings of the thesis have significant implications in furthering knowledge with regards to the sequencing of segment KE in the golf swing. The results of Chapters V, VI and VII reported similar patterns of total segment KE sequencing. Regardless of club type, playing standard and weight transfer style, the magnitude of total segment KE increased sequentially from the Lower Body to the Club. Furthermore, peak total body segment KE occurred simultaneously, significantly earlier than peak total Club which occurred just before impact. Although these chapters also identified differences in the magnitude and timing of the components of total segment KE, these differences had no effect on the sequencing of peak total segment KE which always adopted these distinctive patterns.

Although the timing of peak total segment KE conformed to the principle of optimal coordination of partial momenta (Kreighbaum and Barthels, 1985), Chapters V, VI and VII indicated that, regardless of club type, playing standard and weight transfer style, golf swings are underpinned by sequential timing of the rotational components of total segment KE from the Upper Body to the club. These results have significant practical implications for golf coaches as they indicate that all golfers were able to take advantage of the benefits associated with the SSC (Komi, 2000). It is likely that these earlier, rapid movements of the Upper Body caused the Arms to lag behind which stretched the shoulder and trunk musculature between these segments and produced more powerful concentric contractions.

Despite these similarities, Chapter V provided an insight into the effect of club type on the sequencing of segment KE in the golf swing. An understanding of these results has practical implications for golf coaches and biomechanists who work with skilled players. The results highlighted that, for skilled golfers, as the distance requirement of the shot increased, more energetic swings were produced. Compared with golf swings performed with the 5 and 9 irons, larger magnitudes of peak total Arms and Club KE were evident when the driver was used. These increases were associated with larger peak translational Arms KE and peak local rotational Upper Body KE. Therefore, these results indicate that golfers and golf coaches should attempt to increase the translational KE of the Arms and the local rotational KE of the Upper Body when the distance requirement of the golf shot increases. Furthermore, the results indicated that, for skilled golfers, peak translational Lower Body and Upper Body KE should occur significantly later in the downswing when the driver is used compared with the 5 and 9 irons.

The findings of Chapter VI have significant implications for understanding of the differences between golf swings of skilled and less skilled golfers. They also provided support for future golf swing analyses examining translational movements. Skilled, category 1 golfers produced significantly larger peak translational Arms KE than less skilled category 3 golfers. Furthermore, Chapter VI suggested that an early peak in translational Upper Body KE enabled highly skilled players to exhibit sequential timing of peak translational KE from the Upper Body to the Club. The sequential pattern in translational KE may have enabled highly skilled players to take advantage of the SSC. It is also possible that the earlier peak in Upper Body translational KE is responsible for the higher peak translational Arms and subsequently higher peak translational Club KE produced by category 1 players. These observations raise concerns about the current reliance on analyses of segment angular velocity and serve to highlight the importance of exploring the translational and rotational movements of golfers. These findings might encourage a greater focus on segment KE when the sequencing of body segment movements is examined in the golf swing and in other open kinetic chain activities.

The results of Chapter VI also have practical implications for future analyses of trunk rotations in the golf swing as they suggest that peak local rotational Upper Body KE should occur early in the downswing. Regardless of club type, category 1 golfers maximised local rotational Upper Body KE earlier in the downswing than category 2 and 3 golfers. This early peak would potentially have enabled them to take advantage of the SSC and provided more time for the KE possessed by the Upper Body to be transferred to the Arms and Club in time for impact. Analyses of segment angular velocity have presented conflicting results regarding the effect playing standard on the timing of peak thorax angular velocity. It is likely that these contrasting results reflect the ability of analyses of segment KE to account for more subtle changes in technique and to consider the rotations of rigid bodies in 3D. Therefore, they suggest that for coaches and researchers to fully appreciate the coordination of trunk segment, they must not solely rely on axial rotation-based analyses.

Chapter VII highlighted the importance of identifying weight transfer strategy before analyses of the golf swing are performed by golf coaches or golf biomechanics. The results indicated that larger peak total Arms KE are produced by the Reverse golfers when the driver was used. Furthermore, the results of this study suggested that the generation of large total Arms KE is important when Reverse players are required to produce high clubhead velocities. Chapter VII also identified that different timing strategies were used by the Front Foot and Reverse strategy golfers. In general, these differences related to peak segment KE occurring later in the downswing for the Front Foot group. More specifically, the results suggested that Reverse strategy golfers should be encouraged to generate peak total Arms KE earlier in the downswing.

In summary, this thesis provides an exploration of the sequencing of segment KE in the golf swing. Findings have highlighted some important practical applications for future analyses of the golf swing. This thesis suggests that future analyses should give consideration to the translational movements of body segments and rotations around the frontal and sagittal axes. Furthermore, before golf swings analyses are undertaken, the results suggest that that weight transfer style screening should be performed.

8.3. Future directions

The results presented in this thesis provide a basis for future research to use analyses of segment KE. The results of Chapters V, VI and VII developed an understanding of the sequencing of segment KE in the golf swing. They also recommended that, future analyses of body segment sequencing are performed using analyses of segment KE as it is subtle to changes in technique and examines the translational movements and three-dimensional rotations of body segments. As well as making this recommendation, other areas for future research were identified.

In this thesis, KE was examined for three grouped body segments and a fourth club segment. Previous examinations of segment KE in the golf swing (Anderson, Wright and Stefanyshyn, 2006: Kenny et al. 2008) had used these three grouped body segments as it was proposed that they could characterise movements of the body. This configuration provided insight into the sequencing of body movements. However, the results of this thesis suggested that future research could consider the sequencing of individual rigid body KE. For example, the simultaneous peak in Lower Body and Upper Body KE was explained by legs and lower trunk rigid bodies performing different roles in the golf swing. Therefore, it is possible that if the Lower Body was separated into legs and lower trunk segments the timing of peak total KE would occur in a proximal-to-distal sequence. Furthermore, the use of an analysis of rigid body KE could provide further insight into the nature of the sequence in translational segment KE which was produced exclusively by highly skilled golfers.

The results of Chapter VI highlighted the importance of arm movements in fast, skilled golf swings. More specifically, it was identified that the generation of large remote rotational Arms KE can differentiate between players of varying ability. The practical relevance of this finding was limited by interpretation of the complex calculation of this component of KE. It is possible that remote rotational Arms KE is associated with large extension velocities of the elbow and wrist. However, further research is needed to confirm this notion. Further investigation of this component of KE would allow firm conclusions to be drawn regarding the differences in Arm movements in golf swings of varying quality.

Chapter VII highlighted that golfers who adopted different weight transfer styles produced different magnitudes and timing of peak segment KE. More specifically, Reverse players produced larger magnitudes of peak total Arms KE and in general, peak segment KE occurred earlier in the downswing for these golfers. These results served to highlight that weight transfer strategies should be consider before the sequencing of body segment movements is examined. The results also provided support for future research to examine within style differences in the sequencing of segment KE. For example, the results suggested that the five skilled Reverse strategy players produced considerably larger peak total Arms KE (120.20 ± 13.15 J) than the four less skilled Reverse players (85.86 ± 7.50 J). The results suggested that Reverse strategy golfers should be encouraged to generate peak total Arms KE earlier in the downswing. However, further research which assesses a larger sample of Reverse strategy players of varying ability is required to confirm this notion.

The results of this thesis also provided support for the use of geometric models in biomechanical research. It was suggested that suitably accurate and reliable inertial parameters estimates could be obtained for the majority rigid bodies using this method. However, analyses which use geometric models will always be constrained by the errors associated with their measurement. To improve on the inherent inaccuracies associated with the geometric modelling technique, attention is turning to inertial parameter measurements using 3D scanning approaches. Conventional 3D scanning systems and full body scanners would appear to offer the best level of accuracy. However, their use in biomechanics is inhibited by their expense, complexity and time taken to produce a scans. More recently, the Microsoft Kinect games controller and associated software developer tools provides the potential for quick, low cost 3D scanning. Research has suggested that compared with geometric models, considerable improvements in the accuracy of segment inertial parameter estimates can be made (Clarkson et al. 2012). Use of such a system could improve the accuracy of future researcher. However, a full system that is able to produce fast and accurate estimates of the full segment inertia parameters does not yet exist.

8.4. Limitations

There are various limitations which could have influenced the results of this thesis. Chapter V which examined the sequencing of segment KE in skilled golf swings analysed players with a handicap of +3 – 8 strokes. Players of a similar ability have previously been considered skilled as they produce consistent swing mechanics (Burden, Grimshaw and Wallace, 1998; Egret *et al.,* 2004; Evans *et al.,* 2008). However, it has also been suggested that highly skilled players should have a handicap of less than 4 strokes (Myers *et al.*, 2008). Although only four players had a handicap of greater than 4 strokes caution should be exercised when drawing conclusions from this chapter regarding the sequencing of body segment movements for highly skilled players.

Another limitation of this thesis relates to the data analysis technique used in Chapters IV, V, VI and VII. In each analysis the sequencing of body segment movements was quantified by examining the magnitude and timing of peak segment KE. Analyses of the magnitude and timing of peak segment angular velocities (Cheetham, *et al.,* 2007; Neal *et al.,* 2007; Tinmark *et al.,* 2010) and peak segment KE (Anderson, Wright and Stefanyshyn, 2006; Kenny *et al.,* 2008) have previously been used to quantify the sequencing of body segment movements in the golf swing. However, other analysis techniques could have been used. For example, in signal processing, cross-correlation is a well-established approach for comparing signals and is currently used in many fields including audio-signal processing and image processing (Wren et al., 2006). Cross correlation is a measure of similarity of two waveforms as a function of a time-lag applied to one of them (Stergiou, 2004). This process involves two entire curves so that information between multiple peak values can be assessed (Stergiou 2004). However, it is possible that the use of cross correlation has been limited as it does not identify specific differences between two signals. Using cross correlations, it is also only possible to examine two curves (Wren et al., 2006). This would have made the analysis of four segments extremely difficult to interpret.

Finally, the accuracy of the geometric modelling technique was examined using only three, male participants (Appendix II). As participants with three different body morphologies were examined it was anticipated that the results of this study can be generalised to a wider population. However, caution should be exercised when drawing conclusions regarding the accuracy of the geometric model as the results may be specific to the individual participants. Furthermore, in Appendix I and II only male participants were analysed. This was sufficient this programme of research. However, given the increasing number of female golfers (Reis and Correia, 2013) and different movement patterns exhibited by female golfers (Horan *et al.,* 2010) future analyses of the sequencing of segment KE may wish to focus on female participants. This would require the reliability and accuracy of the adapted Yeadon (1990) geometric models to be established for female participants.

8.5. Conclusion

The purpose of this thesis was to examine the sequencing of segment KE in the golf swing. Initially, Chapter III and Chapter IV examined the methodological considerations associated with the measurement of segment KE. Three further studies were then reported which examined the sequencing of segment KE in the golf swing. Key findings within each study are outlined in section 8.1. Regardless of club type, playing standard or weight transfer style the magnitude of peak total segment KE increased sequentially from the Lower Body to the Club. Furthermore, peak body segment KE occurred simultaneously in the downswing significantly early than peak total Club KE which occurred just before impact. The research programme also highlighted that club type, playing standard and weight transfer style can have significant effects on the magnitude and timing of peak segment KE. In particular, these differences were related to the magnitude and timing of the translation and rotational components of total segment KE. As such, it was recommended that, future examinations of the golf swing should use analyses of segment KE as it is sensitive to subtle changes in technique and considers the three-dimensional translational and rotational movements of body segments. It is also anticipated that the results presented in this thesis will encourage golf coaches and biomechanists to consider the different weight transfer styles that exist within the golf swing.

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APPENDICES

A1. Appendix I - Reliability of Body Segment Inertial Parameter Estimates Made From the Geometric Modelling Technique

A1.1. Introduction

Reliability is an important measure as it provides an indication of the variation in a measurement protocol (Currell and Jeukendrup, 2008). Furthermore, a measurement technique cannot be valid if it is not adequately reliable (Nevill and Atkinson, 1998). As small changes in BSIP have been shown to influence kinetic measures, especially when movements involve large accelerations (Damavandi *et al.*, 2009), the reliable estimation of BSIP is essential for biomechanical analyses (Yeadon and Mortlock, 1989). In comparison with accuracy, the reliability of BSIP estimates is relatively easy to assess. However, the reliability of BSIP estimates made from the geometric modelling technique has been the focus of few scientific studies.

To represent golfers using the geometric model described in section 3.2 required the location of 78 anatomical landmarks to be identified by palpation. Therefore, inconsistent palpation has the potential to significantly reduce the reliability of BSIP estimates. This is especially likely for the trunk segments as anatomical landmarks which define these segments are notoriously difficult to locate (Lariviere and Gagnon, 1999). The complexity of trunk segment inertial parameter estimates using the geometric modelling technique is further increased due to its tendency to change shape as a result of breathing (Wicke and Dumas, 2010). For example, when the lungs contain an additional one litre of air, the volume of the torso is increased by one litre. For a 70 kg participant, this would produce an increase in total body mass of approximately 1.5% (Yeadon, 1990).

In this programme of research BSIP were estimated by one examiner. Due to the time consuming nature of the data collections BSIP were also required to be estimated on different days. This required the repeated set-up of the electromagnetic tracking system and the electromagnetic sensors to be accurately placed on the anatomical landmarks defined in section 3.2.4. Therefore, the primary purpose of this study was to examine the intra-examiner and between-day reliability of BSIP estimates. The experience, training and skill level of examiners all have the potential to influence the reliability of palpation and subsequently the reliability of the estimated BSIP (Huijbregts, 2002). Therefore, to assess the generalisability of the proposed geometric model, inter-examiner reliability was also examined.

The Yeadon (1990) model was validated by reporting a total body mass estimation error. For three participants, two male and one female (M ± SD, 61.7 ± 2.3 kg), Yeadon (1990) reported that the maximum error in total body mass estimation was 2.3%. It was also suggested that, based on this analysis the Yeadon (1990) model was sufficiently accurate to estimate individualised segment inertial parameters in biomechanical research. Earlier geometric models, Jensen (1978) and Hatze (1980) were also validated by reporting maximum total body mass estimation errors of 1.7 and 0.5 % respectively. Therefore, an additional aim of this study was to consider the accuracy of the adapted Yeadon (1990) model to estimate total body mass.

A1.2. Methods

A1.2.1. Participants and examiners

Eight males were recruited to participate in this study (M ± SD, age 27.4 ± 3.6 years, stature 1.81 ± 0.08 m and mass of 79.5 ± 12.7 kg). Before the data collection, ethics approval was granted by the Faculty of Health and Wellbeing Research Ethics Committee and each participant provided written informed consent.

Two examiners identified the 78 anatomical landmarks required to estimate BSIP using the adapted Yeadon (1990) model. Inertial parameters in this programme of research were estimated by examiner 1 however, both examiners had received the same training with respect to the identification of the anatomical landmarks. This training included discussion about the techniques used to palpate certain landmarks as this has been reported to improve inter-examiner reliability (O'Haire and Gibbons, 2000). Both examiners were also extremely familiar with the data collection protocol.

A1.2.2. Data collection

To enable inter-examiner, intra-examiner and days factor reliability to be examined, the 78 anatomical landmarks were identified eight times, twice by examiner 1 and twice by examiner 2: on two consecutive days. On each day, body mass and stature were measured. Then, the anatomical landmarks were identified in the same session lasting approximately an hour. The order in which the examiners performed the data collections was randomly assigned and the examiners were given no feedback on the accuracy or repeatability of previous data collections. Furthermore, to ensure that data collections were independent, at least 20 minutes elapsed between data collections by the same examiner. The BSIP, segment mass, segment centre of mass and segment moments of inertia (Ixx, Iyy and Izz) were then estimated using the technique described in section 3.2.

A1.2.3. Statistical analysis

The data were analysed using SPSS (Version 19.0). To ensure that each dependent variable was normally distributed the Shapiro-Wilk statistic was examined (Field, 2005). Repeated measures factorial analysis of variance (ANOVA) were then performed for each dependent variable to determine if significant differences existed between estimates made by the same examiner (intra-examiner), different examiners (inter-examiner) and on different days (between-day). For each test alpha was set at 0.05 and questionable reliability was determined by a significant difference between BSIP estimates.

Relative reliability was quantified using the intraclass correlation coefficient (ICC). Based on the recommendations of Shrout and Fleiss (1979) inter-examiner and between-day reliability were measured using a two way random effects model with single measures reliability (ICC (2,1)) and intra-examiner reliability was assessed using a two- way mixed effects model with single measure reliability (ICC (3, 1)). Acceptable reliability was determined by an ICC value greater than 0.50 (Shrout and Fleiss, 1979). More conservative ICC scales have been used (de Vet *et al.*, 2006) however, this value has been deemed appropriate for reliability analyses in health care studies (Munro,

1986) and it has been used in numerous reliability studies (de Vet *et al.*, 2006). Absolute reliability was assessed by calculating the standard error of measurement (SEM) (Equation A 1.1). This allowed the measurement error to be expressed in the original units of measurement (Weir, 2005).

$$SEM = SD\left(\sqrt{1} - ICC\right) \qquad \text{Equation A 1.1}$$

For each data collection, total body mass (BM) was estimated by summing the masses of each individual body segment. The accuracy of these estimates was then assessed by calculating a percentage error using the measurements of actual body mass taken at the start of each testing day (Equation A 1.2).

$$BM\ estimation\ error\ (\%) = \left(\frac{Estimated\ BM - Actual\ BM}{Actual\ BM}\right) \times 100 \qquad \text{Equation A 1.2}$$

A1.3. Results

The majority of BSIP were estimated with acceptable reliability. The results suggested that the majority of inertial parameters were estimated with acceptable intra-examiner reliability. However, both examiners estimated foot mass and foot Iyy with questionable reliability. Furthermore, low ICC values suggested that examiner 2 estimated forearm inertial parameters, mass, Ixx and Iyy with questionable reliability. Although not presented in Table A 1.1, a number of COM estimates (thigh, shoulders, upper arm, hand, forearm and neck) were made with questionable reliability by both examiners as small ICC values were reported.

The results suggested that there was little effect of testing day on the reliability of BSIP estimates (Table A 1.2). No questionable reliability was reported for estimates of segment mass and Ixx. Only thigh and shoulders COM and foot Iyy and Izz estimates were made with questionable day's factor reliability.

Table A 1.1 Questionable intra-examiner reliability for estimates of segment mass and mome

RB	Ex	Session 1	Session 2	Session 3	Session 4	p \
Mass (kg)						
FA	2	1.10 ± 0.22	1.13 ± 0.20	1.20 ± 0.34	1.10 ± 0.38	0
Foot	1	1.30 ± 0.16	1.28 ± 0.12	1.22 ± 0.15	1.20 ± 0.10	0
	2	1.24 ± 0.20	1.22 ± 0.11	1.06 ± 0.23	1.16 ± 0.15	0
I-xx (kg/m^2)						
FA	2	0.0065 ± 0.0015	0.0059 ± 0.0011	0.0073 ± 0.0020	0.0067 ± 0.0038	0
I-yy (kg/m^2)						
FA	2	*0.0007 ± 0.0003*	*0.0008 ± 0.0004*	*0.0006 ± 0.0003*	*0.0007 ± 0.0004*	0
Foot	2	0.0016 ± 0.0003	0.0015 ± 0.0002	0.0014 ± 0.0003	0.0013 ± 0.0002	0
	2	0.0016 ± 0.0004	0.0015 ± 0.0002	0.0012 ± 0.0004	0.0014 ± 0.0003	0

RB – Rigid Body, Ex – Examiner, FA – Forearm

Table A 1.2 Questionable between-day reliability for estimates of inertial paramete

Rigid Body	Day 1	Day 2	p value	ICC	SI
COM (cm)					
Thigh	26.29 ± 0.85	26.23 ± 0.86	0.91	0.23	1.
Shoulders	1.99 ± 0.13	2.02 ± 0.31	0.76	0.15	0.
Iyy (kg/m^2)					
Foot	0.0016 ± 0.0002	0.0013 ± 0.0002	0.01	0.58	0.0
Izz (kg/m^2)					
Foot	0.0032 ± 0.0007	0.0029 ± 0.0006	0.01	0.82	0.0

The results indicated that inter-examiner reliability was most frequently questionable (Table A 1.3). Significant differences were identified in the estimation of thigh, middle trunk and shoulder segment masses and for numerous segment centres of masses (Table A 1.3). Questionable inter-examiner reliability was also reported for the estimation of middle trunk and shoulder moments of inertia and for estimates of thigh and foot Ixx and Izz (Table A 1.3).

Table A 1.3 Questionable inter-examiner reliability for estimates of inertial parameters.

Rigid Body	Examiner 1	Examiner 2	p value	ICC	SEM
Mass (kg)					
Thigh	11.07 ± 1.77	11.56 ± 2.15	0.04	0.94	0.49
Middle Trunk	8.73 ± 2.68	7.26 ± 2.13	0.01	0.76	1.24
Shoulder	2.40 ± 0.43	2.83 ±0.62	0.01	0.63	0.37
COM (cm)					
Thigh	25.91 ± 0.93	26.60 ± 1.18	0.04	0.61	0.98
Middle Trunk	7.77 ± 1.46	6.50 ± 1.01	0.01	0.49	1.07
Upper Trunk	9.05 ± 0.78	9.58 ± 1.02	0.03	0.70	0.55
Shoulder	1.87 ± 0.15	2.15 ± 0.24	0.00	0.33	0.28
Upper Arm	11.43 ± 0.44	10.69 ± 0.54	0.01	0.21	0.67
Foot	8.79 ± 0.64	7.81 ± 0.64	0.00	0.35	0.73
Neck	10.25 ± 0.33	10.75 ± 0.29	0.01	0.04	2.79
Ixx (kg/m^2)					
Thigh	0.1790 ± 0.0363	0.2010 ± 0.0508	0.02	0.81	0.0210
Middle Trunk	0.0501 ± 0.0276	0.0364 ± 0.0190	0.02	0.75	0.0124
Shoulder	0.0059 ± 0.0019	0.0069 ± 0.0027	0.04	0.82	0.0010
Foot	0.0034 ± 0.007	0.0028 ± 0.0007	0.00	0.60	0.0005
Iyy (kg/m^2)					
Middle Trunk	0.0840 ± 0.0411	0.0694 ± 0.0344	0.01	0.88	0.0128
Shoulder	0.0321 ± 0.0078	0.0382 ± 0.0117	0.02	0.73	0.0055
Izz (kg/m^2)					
Thigh	0.1780 ± 0.0361	0.1987 ± 0.0492	0.02	0.81	0.0202
Middle Trunk	0.0746 ± 0.0349	0.0573 ± 0.0281	0.01	0.79	0.0329
Shoulder	0.0251 ± 0.0064	0.0299 ± 0.0087	0.02	0.72	0.0046
Foot	0.0034 ± 0.0007	0.0027 ± 0.0007	0.00	0.57	0.0005

A1.3.1. Body mass estimation accuracy

For all of the participants, total body mass was estimated with a mean absolute error of 1.96 ± 1.23 % (Table A 1.4). The maximum error, reported for an estimate of total body mass (-7.49%) was produced when examiner 2 palpated participant 8. Further examination of these results suggested that 56.3 % of estimation errors were under the 2.3 % error threshold used to validate the Yeadon (1990) model. A further 28.1 % of the estimation errors were between 2.3 % and 5 % whilst only 15.6 % of estimate errors were over 5 %.

Table A 1.4 Total body mass estimation accuracy.

	Actual Mass (kg)		Estimated Mass (kg)		Error (%)		
Participant	Day1	Day2	Day1	Day2	Day1	Day2	Mean Abs
1	61.3	61.6	60.5 ± 0.4	59.4 ± 1.4	-1.35	-3.53	2.44
2	67.6	67.9	68.2 ± 0.7	66.1 ± 0.9	0.92	-2.65	1.79
3	73.8	72.9	74.1 ± 0.6	73.2 ± 2.7	0.37	0.48	0.43
4	96.4	98.5	91.1 ± 0.9	96.7 ± 2.9	-5.50	-1.83	3.66
5	75.5	74.2	73.1 ± 2.2	74.0 ± 1.2	-3.18	-0.24	1.71
6	96.6	96.2	97.2 ± 1.5	96.1 ± 1.4	0.65	-0.10	0.38
7	79.3	76.9	80.1 ± 2.1	78.8 ± 2.7	0.95	2.47	1.71
8	85.4	85.6	81.6 ± 2.1	83.4 ± 1.2	-4.51	-2.57	3.54

Mean Abs - Mean absolute error

A1.4. Discussion

The aim of this study was to assess the reliability with which BSIP could be estimated using the geometric model and data collection technique outlined in section 3.2. The results suggested that the majority of BSIP were estimated with acceptable intra-examiner, between-day and inter-examiner reliability. Questionable between-day reliability was only reported for estimates of thigh and shoulder COM and for estimates of foot Iyy and Izz (Table A 1.1). Questionable intra-examiner and inter-examiner reliability was also suggested for numerous estimates of foot inertial parameters (Table A 1.1 and Table A 1.3). A possible cause of this reduced reliability for foot inertial parameter estimates is that participants wore golf shoes in the data collection. As a result, anatomical landmarks were made indirectly on the surface of the shoe which made accurate and repeatable palpation extremely difficult. Despite this reduced reliability for the foot segment, the practical importance of this finding is likely to be negligible. As a result of small foot segment mass, moment of inertia and peak translational and rotational velocities it is probable that small translational and rotational foot KE will be produced during the golf swing.

Middle trunk, shoulder and the majority of thigh inertial parameters were estimated with questionable inter-examiner reliability (Table A 1.3). Although the ICC values reported for these estimates suggested that they were estimated with acceptable reliability (ICC > 0.5), the two examiners produced significantly different inertial parameter estimates. It has been suggested that if significant differences are reported when the error term is small and only small differences between means actually exist, systematic variations in the data collection technique can explain the variability (Weir, 2005). For middle trunk, shoulder and thigh segments, the differences between the inertial parameter estimates made by both examiners and the error term (SEM) could be considered small as they fall within the reported standard deviations. As a result, it is likely that systematic differences in the palpation techniques used by the two examiners produced the variability in these inertial parameter estimates. Therefore, for multiple examiners to use the data collection technique and geometric model, further standardisation of the palpation techniques would be required. It has been reported that this could be achieved through more detailed discussions regarding the reference points which can be used to identify anatomical landmarks (Huijbregts, 2002; Holgrem and Waling, 2008).

Several ICC values under 0.5 were reported for the estimation of COM suggesting that it was regularly estimated with questionable intra-examiner and inter-examiner (Table A 1.1 and Table A 1.3) reliability. However, Weir (2005) suggested that it is possible for low ICC values to be reported when the difference between mean values and within participant variability are low. For the majority of COM estimates the difference between mean values could be considered small as it was within the SEM. It has also been suggested that the COM is the most accurately estimated inertial parameter (Outram *et al.*, 2011). As a result, the variability between repeated COM estimates was small. These factors could explain the low ICC values reported for COM estimation especially for intra-examiner COM estimation where it was expected that variability would be further reduced (Huijbregts, 2002).

The geometric model proposed by Yeadon (1990) was validated by reporting a maximum body mass estimation error of 2.3%. Using the geometric model outlined in section 2.3 a lower mean total body mass estimation error was obtained (-1.23 ± 2.91%). This mean error which is similar to the mean error of 1.98% reported for a much more complex geometric model (Jensen, 1978) suggests that the adapted Yeadon (1990) model which uses less geometric shapes to represent the body is capable of accurately estimating body mass. However, a maximum body mass estimation error of -7.49% was reported. Therefore, before this model can be used with confidence the accuracy of individual segment inertial parameter estimates needs to be established.

A1.5. Conclusion

The majority of BSIP were estimated with satisfactory intra-examiner and between-day reliability. Therefore, as all body segment inertial parameters estimates in this programme of research were undertaken by the same examiner the geometric model and data collection technique proposed in section 3.2 have sufficient reliability for use in analyses of segment KE. Questionable inter-examiner reliability was reported for middle trunk, shoulder and thigh inertial parameter estimates. However, it has been suggested that by standardising the reference points used to identify anatomical landmarks, the reliability of these estimates could be improved.

A2. Appendix II - Accuracy of Body Segment Inertial Parameter Estimates Made From the Geometric Modelling Technique

A2.1. Introduction

The accuracy of biomechanical analyses and importantly analyses of segment KE can depend on the extent to which the approximation of body segments represents the true anatomical structure (Sheets, Corazza and Andriacchi, 2010). Attempts to determine the accuracy and validity of BSIP estimates made using geometric models have generally been limited to the analysis of whole body mass estimation (e.g. Gittoes, Bezodis and Wilson, 2009; Hatze, 1980; Jensen, 1978; Yeadon, 1990). However, the true accuracy of geometric models is dependent on the ability of the geometric shapes to estimate individual segment inertial parameters (Sheets, Corazza and Andriacchi, 2010).

Detailed analyses of the accuracy of individual segment inertial parameter estimates have been performed. However, these analyses have studied only trunk segments and have been limited to analyses of centre of mass and moment of inertia in the frontal plane (Lariviere and Gagnon, 1999; Wicke and Dumas, 2010). Despite these methodological limitations, it was suggested that the accuracy of segment inertial parameter estimates made using geometric models is highly dependent on the accuracy of segment volume estimation (Wicke and Dumas, 2010). Applying a uniform density had only a small, secondary, influence on the accuracy of BSIP estimates (Wicke and Dumas, 2010).

The accuracy of segment inertial parameter estimates is dependent on the accuracy of segment volume estimation and consequently the accuracy with which anthropometric measurements can be made. O'Haire and Gibbons (2000) suggested that there is an inherent error associated with anthropometric measurements which has an undesirable influence on the accuracy with which BSIP can be estimated. Therefore, to determine the absolute accuracy with which the geometric shapes of the model proposed in section 3.2 are capable of estimating the volume of body segments required segment inertial parameters to be estimated without this inherent error. The analysis of the absolute accuracy of segment volume estimation also required gold standard segment volume to be defined for each segment.

It has been suggested that 3D laser scanners can make gold standard measurements of trunk segment volume (Sheets, Corazza and Andriacchi, 2010). 3D laser scanners can also estimate the volume and COM of the legs with errors of less than 1% (Norton, Donaldson and Dekker, 2002). Furthermore, using a uniform density function, 3D laser scanners can provide accurate inertial parameter estimates for all body segments in three dimensions (Sheets, Corazza and Andriacchi, 2010). Therefore, the main aim of this study was to determine the accuracy with which geometric shapes are capable of estimating the geometry of human body segments and subsequently, by assuming uniform density, estimate BSIP.

A2.2. Methods

A2.2.1. Participants

Three male participants with different body morphologies were recruited to take part in this study (Table A 2.1). Body morphology and body fat percentage were assessed using the Heath and Carter (1967) and Jackson and Pollock (1985) methods respectively. Before the data collection, ethics approval was granted by Sheffield Hallam University's ethics board and each participant provided written informed consent.

Table A 2.1 Participant details.

Participant	1	2	3
Age (years)	21	23	19
Stature (cm)	193.0	184.4	187.8
Mass (kg)	137.3	92.0	79.9
% Body Fat	22.8	13.6	11.3
Endomorphy	6	3	2
Mesomorphy	8	8	3
Ectomorpy	0.5	1.5	4

A2.2.2. Data collection

For each participant, data collection took place on the same day and lasted approximately 2 hours. Initially, head, arm, leg and trunk segments were scanned using a Model Maker D100 non-contact laser scanner (Metris, Leuven, Belgium). To prevent the irregular reflectance properties of body hair producing gaps in the data set participants wore lycra tights, a long sleeve lycra top and a lycra hat (Figure A 2.1). Retro-reflective markers (10 mm) were then placed on anatomical landmarks to enable the scanned segments to be segmented in accordance with the geometric model. Data cleaning and surface reconstruction were performed to improve the quality of the dataset and enable gold standard segment shape and volume to be calculated. Segment inertial parameters were then calculated in Pro Engineer using the equations defined by Yeadon (1990) and a uniform density (Dempster, 1955) (Table 3.6).

Figure A 2.1 Participants, marker placement and lycra clothing; left - participant 1, middle - participant 2, right - participant 3.

To examine the absolute accuracy of BSIP estimates made using the geometric model the required anthropometric measurements were estimated using a technique which negated anthropometric measurement error. Using Pro-Engineer, measures of segment width, depth and height were made for each participant directly from the laser scanned segments. Segment inertial parameters were then estimated using the equations defined by Yeadon (1990) and the same uniform density function (Dempster,

1955) used to calculate the gold standard measures. For both sets of inertial parameters each segments local coordinate system was defined such that the x, y and z axes were sagittal, longitudinal and frontal respectively.

A2.2.3. Data analysis

The accuracy of the geometric model was examined by comparing the segment inertial parameter estimates made using Pro-Engineer to the segment inertial parameters estimated using the gold standard volume and shape measurements from the body scanning protocol (Equation A 2.1).

$$Error\ (\%) = \frac{Geometric\ Model\ BSIP - Gold\ Standard\ BSIP}{Gold\ Standard\ BSIP} \times 100 \qquad \text{Equation A 2.1}$$

A2.3. Results

The three participants were classified as a mesomorph-endomorph (ME), a balanced mesomorph (BM) and balanced ectomorph (BE) (Table A 2.1) respectively. For all segments, the mean absolute segment mass estimation error was 7.4 ± 1.1 %. Upper trunk mass was estimated with the highest accuracy for all three participants whilst upper arm mass was estimated with the lowest accuracy (Table A 2.2). Mean absolute estimation errors of 1.6 ± 0.4 % and 18.8 ± 9.3 % were produced for the upper trunk and upper arm segments respectively.

Table A 2.2 Errors in segment mass estimation using the geometric model.

Participant	1			2			3		
	Mass (kg)		Error	Mass (kg)		Error	Mass (kg)		Error
	GS	PE	(%)	GS	PE	(%)	GS	PE	(%)
Forearm	2.47	2.58	4.3	1.58	1.64	3.8	1.44	1.54	6.9
Upper Arm	4.43	4.87	10.0	2.72	3.21	17.9	2.45	3.15	28.5
Shank	7.22	7.85	8.7	4.37	4.77	9.2	4.22	4.29	1.7
Thigh	20.12	17.96	-10.7	15.34	14.57	-5.1	11.05	11.65	5.4
Lower Trunk	14.81	15.34	3.5	9.19	9.76	6.2	10.46	10.22	-2.3
Middle Trunk	21.36	22.79	6.7	15.58	16.55	6.3	14.59	16.60	13.8
Upper Trunk	22.16	21.73	-2.0	12.64	12.48	-1.3	10.55	10.38	-1.6

Centre of mass (COM) was the most accurately estimated inertial parameter. For all three participants, the mean absolute COM estimation error was 2.2 ± 1.3 % and a maximum error of only 6.7 % was obtained for the estimation of participant 3's middle trunk. Moments of inertia were estimated with the lowest accuracy (Table A 2.3). Similar mean absolute errors were obtained in the estimation of Ixx (8.0 ± 4.1 %), Iyy (7.7 ± 4.7 %) and Izz (9.0 ± 5.1 %). However, a maximum error of -21.1 % was obtained for the estimation of thigh Iyy for participant 1.

Table A 2.3 Absolute errors in the estimation of moment of inertia for all three participants.

	Ixx			Iyy			Izz		
Participant	1	2	3	1	2	3	1	2	3
Forearm	4.0	10.6	6.3	5.5	2.0	12.4	5.3	13.7	5.7
Upper Arm	-0.3	-3.5	15.9	18.9	11.7	-4.1	1.8	0.1	14.5
Shank	18.4	19.3	8.9	7.9	11.1	-2.0	19.1	19.1	8.9
Thigh	-12.1	-4.7	4.2	-21.1	-15.9	4.0	-13.0	-4.3	4.7
Lower Trunk	6.6	9.4	-4.2	5.2	3.3	-2.0	6.6	16.7	-5.8
Middle Trunk	10.7	3.3	-3.4	9.6	10.6	-9.7	10.4	17.6	10.5
Upper Trunk	-9.2	-6.4	-7.1	-7.2	-10.0	-8.4	-0.6	8.1	2.6

A2.4. Discussion

The aim of this study was to determine the accuracy with which segment inertial parameters could be estimated using geometric shapes. The analysis of body segment inertial parameter estimation reliability (Appendix I) indicated that the geometric model was capable of accurately estimating total body mass. However, it is possible for accurate estimates of total body mass to be produced by a cancellation of positive and negative errors generated in the estimation of individual segment masses. Therefore, to provide a comprehensive analysis of the accuracy of the geometric modelling technique the accuracy of individual segment inertial parameters estimates was assessed (Sheets, Corazza and Andriacchi, 2010).

The majority of limb segment inertial parameters were estimated with high accuracy. Challis (1999) also reported that limb segment inertial parameters can be determined with high precision when estimated using the geometric modelling technique. Despite this relative accuracy of limb inertial parameter estimates, upper arm inertial parameters were estimated with much lower accuracy. This reduced accuracy may be caused by the geometric model segmentation plane not being parallel to that of the 3D scanned segment at the shoulder (Figure A 2.2). For the other limb segments which were estimated with higher accuracy, the segmentation planes of the geometric model and body scans were parallel (Figure A 2.3).

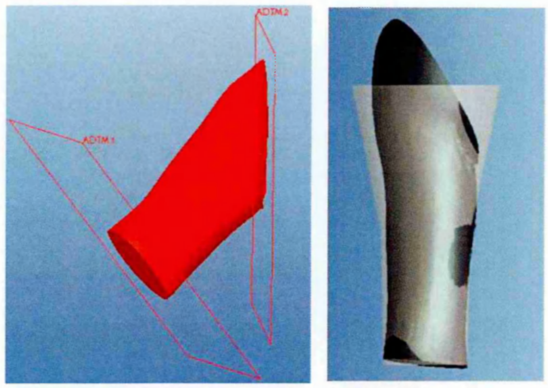

Figure A 2.2 Participant 1's upper arm segmentation. Left: Proximal and distal segmentation planes, Right: Transparent elliptical solid superimposed onto the dark grey scanned segment.

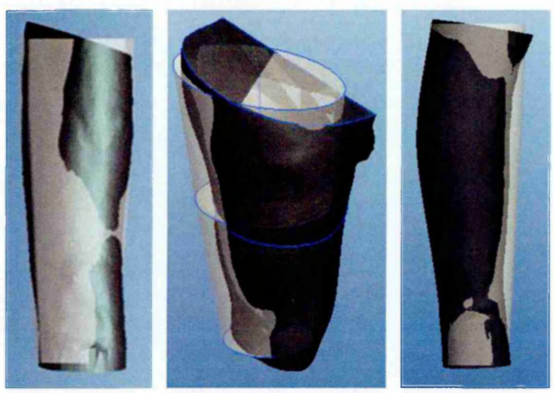

Figure A 2.3 Limb segment transparent elliptical solid superimposed onto the dark grey scanned segment. Left: Forearm, Middle: Thigh, Right: Shank.

It has been suggested that the awkward shape of trunk segments would cause inertial parameters to be estimated with lower precision compared with the estimation of limb segment inertial parameters (Challis, 1999). This reduced precision has been related to the difficulty associated with modelling trunk segments using simple shapes (Wicke and Dumas, 2010). However, in this study, trunk inertial parameters were estimated with similar accuracy to that achieved in the estimation of limb inertial parameters. In fact, the mass of the upper trunk was estimated with a mean absolute error of just 1.6%. It is possible that this accuracy was caused by the removal of anthropometric measurement error from the data collection technique. Due to the high proportion of trunk segment fat and the associated difficulty in the identification of trunk segment anatomical landmarks (Huijbregts, 2002; Wicke and Dumas 2010) it is possible that the inclusion of measurement error would have the largest effect on the accuracy of trunk inertial parameter estimates. However, it has also been suggested that the accuracy of trunk inertial parameter estimates was highly dependent on the number of shapes used in the modelling process (Erdmann, 1997).

When compared with other trunk segments, middle trunk inertial parameters were estimated with the lowest accuracy. As the middle trunk was represented by only one stadium solid this provides further support for the suggestion that using more geometric shapes in the modelling process increases the accuracy of body segment shape estimation (Erdmann, 1997). However, the geometry of the scanned middle trunk segments was relatively closely matched by that of the stadium solids (Figure A 2.4). When combined with the accuracy achieved in the estimation of middle trunk segment inertial parameter estimates this suggests that the adapted Yeadon (1990) model used an appropriate number of stadium solids to model each trunk segment and account for their various contours.

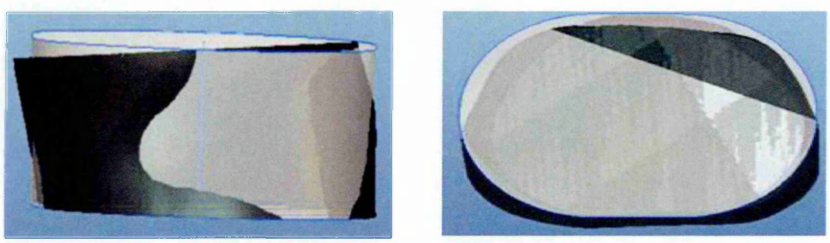

Figure A 2.4 Stadium solid created using technique 2 superimposed onto scanned image of participant 3's middle trunk.

A2.5. Conclusion

The geometry of human body segments can be accurately represented using simple geometric shapes. Subsequently, using a uniform density function, it is possible that the majority of limb segment inertial parameters would be accurately estimated using the geometric modelling technique. Assuming an appropriate number of stadium solids are used in the modelling process the geometry and volume of trunk segments can also be accurately modelled using the geometric modelling technique. However, further work is required to establish the effect of these inertial parameter estimation errors on the magnitude and timing of peak segment KE in the golf swing. Particular attention should be given to the effect of upper arm inertial parameter estimation errors.

A3. Appendix III - The Accuracy of Hand Mass Estimation Using the Geometric Modelling Technique

A3.1. Introduction

In the study reported in Appendix II, the accuracy of hand inertial parameter estimates was not examined. To maintain a stable body position during the 3D scans the participants were required to hold a support frame which meant it was not feasible to obtain scans of the hand segment. Errors in the estimation of hand inertial parameters, particularly hand mass and hand COM have the potential to significantly influence the magnitude of peak hand KE. For example, a 10 % error in the estimation of hand mass would produce a 10 % error in the estimation of peak linear hand KE. Therefore, it was important to examine the accuracy with which hand inertial parameters could be estimated using the geometric model outlined in section 3.2.

A3.2. Methods

An indication of hand inertial parameter estimation accuracy was calculated by comparing hand inertial parameters estimates made for the eight participants used in the study reported in Appendix I (M ± SD, age 27.4 ± 3.6 years, height 1.81 ± 0.08 m and mass of 79.5 ± 12.7 kg) with estimates of hand inertial parameters made using regression equations. Numerous regression equations have been formulated which enable hand inertial parameters to be estimated. Some are based on simple input parameters such as stature and body mass (Clauser 1978; Durkin and Dowling 2003; Zatsiorsky 2002) whilst other, more complex equations require further anthropometric measurements such as segment length (Hindrichs 1990; McConville 1980; Yeadon and Morlock 1989). In the study reported in Appendix I, measures of hand length were not established as anthropometric measurements were made with the fist clenched. This prevented complex regression equations from being utilised to establish the accuracy of hand inertial parameter estimates. It also prevented comparisons of hand COM and MOI being made as regression equations typically estimate these parameters with the fingers extended.

Vladimir Zatsiorsky developed numerous regression equations based on measurements performed on living people using the gamma radiation method. Some of these equations were based on detailed anthropometric measurements (Zatsiorsky and Seluyanov, 1983) whilst others were based on simple input parameters, body mass (BM) and stature (Zatsiorsky, 2002). It has been suggested that when using simple input parameters, the equations proposed by Zatsiorsky (2002) provide the best, most complete estimates of segment inertial parameters (Durkin & Dowling, 2003). Therefore, the regression equation proposed by Zatsiorsky (2002) (Equation A 3.1) was used to provide an indication of the accuracy of hand mass estimates made using the adapted Yeadon (1990) model.

$$Hand\ Mass\ (kg) = -0.1165 + (0.003BM) + (0.00175Stature) \qquad \text{Equation A 3.1}$$

The hand mass estimates made using the geometric hand model were then compared to those made using the regression equations (Equation A 3.2).

$$Error\ (\%) = \frac{Estimated\ Hand\ Mass - Gold\ Standrd\ Hand\ Mass}{Gold\ Standard\ Hand\ Mass} \times 100 \qquad \text{Equation A 3.2}$$

A3.3. Results and discussion

Compared with hand mass estimates made using the regression equation, hand mass was consistently underestimated using the geometric model (Table A 3.1). A mean hand mass of 0.29 ± 0.04 kg was reported using the geometric model which was -40.48 ± -5.45 % lower than the mean hand mass estimated using the regression equation proposed by Zatsiorsky (2002) (0.49 ± 0.06 kg).

The geometric model (Section 3.2) used one stadium solid to represent the hand. The results of this exploratory analysis suggest that this approach does not provide an accurate representation of the geometry or volume of the hand with the fist clenched. As such, using this model in analyses of segment KE would produce significant errors in the estimation of hand KE. Therefore, significant improvements must be made to this geometric modelling approach before it can be used to estimate hand inertial parameters in analyses of segment KE.

Table A 3.1 Hand mass estimates made using the geometric modelling technique and the regression equation proposed by Zatsiorsky (2002).

	Estimated Hand Mass (kg)		
Participant	Regression	Geometric Modelling	Error (%)
1	0.40	0.21	-46.57
2	0.44	0.28	-36.63
3	0.47	0.30	-36.62
4	0.57	0.35	-39.21
5	0.47	0.32	-31.72
6	0.55	0.30	-46.13
7	0.49	0.29	-40.94
8	0.51	0.28	-46.00
Mean	0.49 ± 0.06	0.29 ± 0.04	-40.48 ± 5.45

A4. Appendix IV - The Accuracy of Hand Mass Estimation Using the Revised Geometric Modelling Technique

A4.1. Introduction

The geometric model initially proposed in this programme of research (Section 3.2) used one stadium solid to represent the hand. However, it was subsequently suggested (Appendix III) that modelling the hand using this approach did not provide accurate estimates of hand mass. As such, a geometric modelling technique has been proposed (Section 3.3.2) which represented the base of the hand using a stadium solid and the fingers wrapped around a golf club using a segment of a hollow cylinder (Figure 3.10). Before this model could be used with certainty in analyses of segment KE its accuracy needed to be established. Therefore, the aim of this study was to determine the accuracy with which the revised geometric model was capable of estimating the mass of the hand segment.

A4.2. Methods

A4.2.1. Participants

Ten male participants (M ± SD; age 79 ± 6 years, height 185.3 ± 6.6 cm, mass 80.8 ± 9.5 kg) were recruited for this study. To ensure that the selected participants had a range of hand masses a qualitative assessment of hand size was initially performed. Ethics approval was obtained from the Faculty of Health and Wellbeing Research Ethics Committee and written informed consent was obtained from each participant.

A4.2.2. Gold standard hand mass estimates

Gold standard hand mass was calculated for each participant using a water immersion technique described by Hughes (2005). This technique, based on the Archimedes principle, assumes that an object partially immersed in a container of water placed on a balance will be buoyed up by the displaced water, and this force will be manifested as an increase in the weight recorded by the balance. Assuming that water has a density of 1 g cm^3 this increase in weight will be equivalent to the addition of an amount of water equal in volume to the part of the object immersed.

To enable hand mass to be calculated using this technique, a line was drawn around the right wrist between the ulna styloid, the mid-flexor tendon, the radial styloid and the mid-extensor tendon. The hand was then immersed in a container of water placed on an electronic balance up to this line with care being taken not to touch the sides of the container. In this manner, hand volume was established five times by measuring the change in weight when the hand was immersed. Between subsequent immersions, the hand was thoroughly dried with a paper towel. Gold standard estimates of hand mass were then made by combining this volume measurement with a uniform density function (Dempster 1955).

A4.2.3. Hand mass estimation

The hand was modelled using the technique described in section 3.3.2 which comprised of a stadium solid (Figure 3.1) and a segment of a hollow cylinder (Figure 3.10). To enable the required anthropometric measurements to be calculated using the position of anatomical landmarks, an electromagnetic sensor was attached to the back of the right hand using a golf glove (Figure 3.7). The location of 8 anatomical landmarks was then identified by palpation and recorded using the Polhemus digital stylus, the Polhemus electromagnetic tracking system and custom written software. The volume of hand was then calculated and combined with a uniform density (Dempster, 1955) function to estimate the mass of the hand.

A4.2.4. Data analysis

The accuracy of hand mass estimates made using the geometric hand model was examined by comparing these estimates with the gold standard estimates of hand mass made using the water immersion technique (Equation A 4.1).

$$Error\ (\%) = \frac{Estimated\ Hand\ Mass - Gold\ Standrd\ Hand\ Mass}{Gold\ Standard\ Hand\ Mass} \times 100 \qquad \text{Equation A 4.1}$$

A4.3. Results

Using the revised geometric modelling technique, hand mass was estimated with good accuracy for all ten participants (Table A 4.1). A minimum error of -0.82 % was reported for the estimation of participant 9's hand mass whilst a maximum error of only 10.43 % was reported for participant 5. As well as a low mean error of 1.57 ± 0.04 % being reported a relatively low mean absolute error of 4.97 ± 4.07% was also reported.

Table A 4.1 Hand mass estimation accuracy.

	Hand Mass (kg)		
Participant	Immersion	Geometric Model	Error (%)
1	0.51 ± 0.003	0.51	1.15
2	0.49 ± 0.001	0.53	9.15
3	0.46 ± 0.002	0.50	9.85
4	0.54 ± 0.004	0.55	2.16
5	0.45 ± 0.006	0.50	10.43
6	0.46 ± 0.006	0.45	-1.26
7	0.46 ± 0.000	0.46	-1.10
8	0.57 ± 0.006	0.54	-5.54
9	0.48 ± 0.006	0.48	-0.82
10	0.62 ± 0.006	0.57	-8.28
Mean ± SD	0.50 ± 0.060	0.51 ± 0.04	1.57 ± 6.44

A4.4. Discussion

The aim of this study was to determine the accuracy with which hand mass could be estimated using a stadium solid and a hollow segment of a cylinder. It was hypothesised that this model would represent the hand grasping a golf club with higher accuracy than a single stadium solid. The results suggest that this model was capable of accurately estimating hand mass as a mean error of only 1.57 ± 6.44 % was reported. This mean error was comparable with the lowest mean error reported for estimates of mass for all of the segments examined in the study reported in Appendix II. In that study, upper trunk mass was estimated with the highest accuracy as a mean error of only -1.6 ± 0.4 % was reported.

Using a water immersion technique, Challis and Kerwin (1996) suggested that hand volume and subsequently hand mass could be accurately estimated using a truncated cone and a hollow cylinder. It was suggested that the volume of the forearm, modelled as a series of 13 truncated cones and hand could be estimated with an error of only 0.15%. Although this error is smaller than that reported in this study, Challis and Kerwin (1996) only examined one participant. Yeadon (1990) also suggested that the base of the hand could be more accurately represented using a stadium solid than a truncated cone. As hand mass was slightly overestimated in this study using a stadium solid and a hollow cylinder segment it is logical that using a full hollow cylinder to model the fingers (Challis and Kerwin, 1996) would further increase the estimated hand mass and reduce the accuracy of the model. These results also suggest that the model used in this study more accurately represented the distribution of hand mass than the model proposed by Challis and Kerwin (1996). As a result, this would generate more accurate estimates of hand COM.

A4.5. Conclusion

The results of this study suggest that sufficiently accurate estimates of hand mass can be produced by modelling the hand using a stadium solid and a hollow cylinder segment for it to be used in analyses of segment KE. The results also suggest that representing the hand using this model provides an accurate description of mass distribution which would enable accurate estimates of hand COM to be generated.

A5. Appendix V - The Effect of Electromagnetic Sensor Attachment on Swing and Launch Parameters in the Golf Swing

A5.1. Introduction

The analysis of segment KE in the golf swing requires the golfer's kinematics to be described (Winter, 2005). Traditionally, much of the 3D data collection in scientific golf swing literature has been performed using optical tracking systems (Coleman and Anderson, 2007; Egret et al., 2003; Healy et al., 2011; Myers et al., 2008; Zheng et al., 2008) as they have been considered to be the gold standard tracking system (Richards et al., 1999). However, in recent years, as a result of new hardware design and error correction, new generations of electromagnetic tracking systems have shown impressive improvements with respect to accuracy and sensor size (Hummel et al., 2005). It has been suggested that electromagnetic tracking systems have reached a level of sophistication where the positions and orientations of sensors can be determined with accuracies comparable to that of optical tracking systems (Frantz et al., 2003; Hassan, Jenkyn and Dunning, 2007; Mills *et al,*. 2007). Therefore, as electromagnetic tracking systems are portable and enable 3D kinematic data to be collected in real time (Richards *et al.*, 1999) the use of electromagnetic tracking systems in golf swing analysis has become more prevalent (Cheetham *et al.*, 2008; Neal *et al.*, 2007; Teu *et al.*, 2006; Tinmark *et al.*, 2010).

The measurement of 3D translation and rotation using an electromagnetic tracking system requires electromagnetic sensors to be attached to body segments. For whole body kinematic descriptions such as that required to perform analyses of the sequencing of segment KE in this programme of research twelve electromagnetic sensors are required to be attached to the golfers. To provide a description of the translation and rotation of the clubhead a further sensor is required to be attached to the club shaft. Furthermore, these electro-magnetic sensors were required to be linked to a host computer using a series of wires. The attachment of electro-magnetic sensors to body segments and use of wires to link them to a host computer has the potential to encumber golfers, make them feel uncomfortable and restrict their movements (Wright, 2008). These restrictions have been related to the physical characteristics of the sensors and the technique used to attach the sensors to body segments (Manal *et al.*, 2000).

For valid analyses of the golf swing to be performed it is important that golfers have complete freedom of movement and are capable of replicating their normal technique (Bull, Berkshire and Amis, 1998). However, the effect of electro-magnetic sensor attachment on swing characteristics and quality of shots performed by the golfers has yet to be assessed. It has been suggested that the attachment of retro-reflective markers used with passive optical motion capture systems had no effect on estimates of tibial rotation during walking (Capozzo *et al.*, 1996; Manal *et al.*, 2000). However, it has also been suggested that the attachment of lightweight, wireless markers used with optoelectronic tracking systems is less restrictive and cumbersome than the attachment of electro-magnetic sensors (Aminian and Najafi, 2004). Furthermore, the attachment of electromagnetic sensors to baseball players caused their movements during pitching and batting to be constrained (Lampinski *et al.*, 2009). Therefore, before electro-magnetic tracking systems can be used with confidence in kinematic golf swing data capture the effect of sensor attachment on swing characteristics must be established.

Changes in ball flight and clubhead characteristics can reflect changes in golf swing mechanics (Lamb, 2012). Therefore, by using a commercial launch monitor (Trackman A/S, Vedback, Denmark) to measure ball flight and clubhead characteristics the aim of this study was to establish the effect of electro-magnetic sensor attachment on golf swing mechanics. Before these measurements of ball flight and clubhead characteristics could be used with confidence in this context, it was important to examine the reliability of swing and launch parameters measured using the Trackman Pro (Appendix VI).

A5.2. Methods

A5.2.1. Participants

Fifteen males golfers (M ± SD; age 33 ± 14 years, stature 182.5 ± 3.0 cm, mass 85.6 ± 9.8 kg and handicap 9.7 ± 7.7 strokes, range +2 - 20 strokes) were recruited to take part in this study. At the time of testing the golfers were required to be playing or practicing at least once a week. Golfers were excluded if their handicap was greater than twenty or if they had an injury which was preventing them from regularly playing or practising golf. Ethics approval was obtained from the Faculty of Health and Wellbeing Research Ethics Committee and before the data collection, written informed consent was obtained from each golfer.

A5.2.2. Data collection

Golfers were given time to complete their usual pre-game warm-up routine. Then, they were randomly assigned to one of two testing orders. In both orders there were two conditions: condition one in which golfers wore their usual golf attire and condition two in which electro-magnetic sensors were attached to the golfers. In order one they completed condition one followed by condition two and in order two they completed condition two followed by condition one. In condition two, a custom designed suit comprising a baselayer jacket, adjustable straps (Figure 3.6), golf gloves and a cap (Figure 3.7) was used to attach electro-magnetic sensors to the locations described in section 3.2.4.

In both conditions each golfer hit fifteen good quality shots from an artificial mat into a net 5 meters away (Figure A 5.1), five with a driver, five with a 5 iron and five with a 9 iron. Each shot was qualitatively rated on a ten point scale with a 1 representing a shot the player was completely unsatisfied with and 10 representing their interpretation of the perfect swing. Shots rated as less than seven were discounted and another shot was hits. A Ping G15 driver and Ping i15 irons with a regular graphite shaft, standard length and standard lie angle were used. Furthermore, to control for the effect of fatigue and to ensure internal-validity these clubs were presented in a randomly assigned order (Thomas, Nelson and Silverman, 2011). Sufficient practice trials were allowed to ensure that golfers were familiar with the clubs, the laboratory environment and the data collection protocol. Before data were collected in condition two, golfers were also given time to become familiar with the attachment of the electromagnetic sensors.

A5.2.3. Measurement device and experimental set-up

To provide an assessment of golf swing performance ball flight and clubhead characteristics were measured using a Trackman Pro (Trackman A/S, Vedback, Denmark) launch monitor. The launch monitor was set-up in accordance with the manufacturer's instructions and was positioned in line with the target, directly behind the hitting mat (Figure A 5.1). To ensure that measurements of ball flight characteristics were consistent and accurate Titleist Pro VI golf balls complete with a reflective marker were used for every shot.

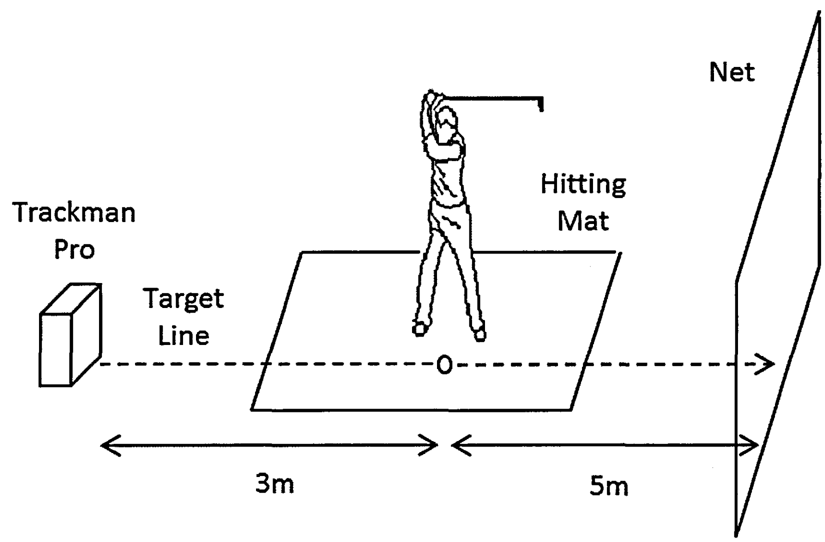

Figure A 5.1 Data collection set-up. Golfer taken from Ball and Best (2007a).

The ball flight characteristics examined were initial ball velocity, spin rate, carry distance and side carry. Clubhead characteristics; clubhead velocity, horizontal swing plane angle, attack angle and face angle were also examined. Horizontal swing plane was defined as the direction of the clubhead in relation to the target line at impact where positive angles represented an in-to-out swing plane and negative angles represented an out-to-in swing plane (Figure A 5.2).

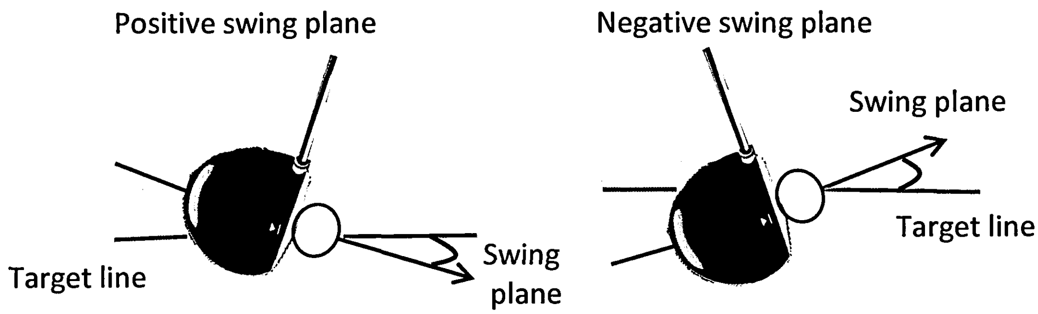

Figure A 5.2 Horizontal swing plane.

Attack angle was defined as the vertical angle with which the clubhead was moving at impact. A positive angle was produced when the golfers were hitting up and a negative angle was produced when the golfers were hitting down (Figure A 5.3).

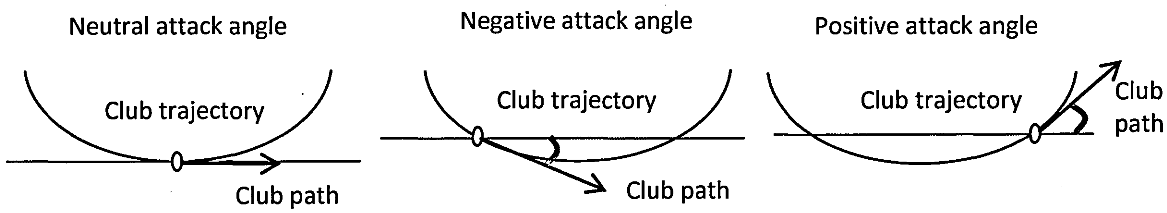

Figure A 5.3 Attack angle.

Finally, face angle was calculated using the relative angle between the club face and the target line (θ) (Equation A 5.1). A negative face angle was produced when the club face was closed and a positive face angle was produced when the club face was open (Figure A 5.4).

$$Face\ Angle\ (°) = 90 - \theta$$

Equation A 5.1

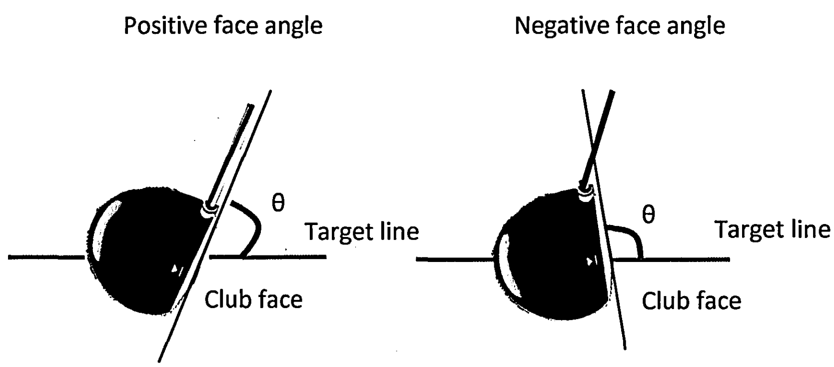

Figure A 5.4 Face Angle.

A5.2.4. Statistical analysis

The mean of each variable for the five shots with each club in both conditions was used for statistical analyses which were completed using SPSS for windows (Version 19.0). Tests of normality (Shapiro-Wilk) were undertaken to ensure that each variable was appropriate for parametric statistical tests. To test for significant differences between conditions, separate paired sample t-tests were performed for each dependent variable with all three clubs. When multiple comparisons like this are performed, it has been common practice to control the type I error rate by altering alpha. The Bonferroni adjustment has frequently been used for this purpose (Cheetham *et al.*, 2008; Tinmark *et al.*, 2010) and in golf swing analyses an alpha level of 0.01 has often been used (Ball and Best, 2007a; Healy *et al.*, 2011; Zheng *et al.*, 2008). However, the Bonferroni correction has received substantial criticism as it is often considered to be overly conservative (Feise, 2002; Perneger, 1998; Savitz and Olshan, 1995). Therefore, due to the nature of this study and the need to control the type II error rate the alpha level was set at 0.05 and Cohen's effect size was calculated (Equation A 5.2) where M_1 was the mean for condition 1, M_2 was the mean for condition 2 and SD was the standard deviation for both conditions.

$$Effect\ size\ (d) = \frac{M1-M2}{SD} \qquad \text{Equation A 5.2}$$

A5.3. Results

For all three clubs, no significant differences were produced for measures of carry, side carry and spin rate (Table A 5.1). Similar measures of horizontal swing plane, face angle and attack angle were also produced in both test conditions for all three clubs (Table A 5.2). Additionally, in both conditions, comparable measures of clubhead speed were produced with the driver and 9 iron (Table A 5.2). Despite these similarities, electromagnetic sensor attachment caused some significant differences to be produced in measures of ball flight and clubhead characteristics. With the 5 iron, significantly lower clubhead speeds were produced when electromagnetic sensors were attached to the golfers ($t(14) = 2.89$, $p < 0.05$, $d = 0.25$) (Table A 5.2). Furthermore, significantly lower initial ball speed was measured with the 9 iron when the electromagnetic sensors were attached to the golfers (t(14) = 2.53, $p < 0.05$, d = 0.28) (Table A 5.1).

Table A 5.1 Ball flight characteristics.

Parameter	Club	Golf Clothing	Sensors Attached	*p*	*d*
Ball Speed (mph)	D	142.1 ± 10.1	142.0 ± 8.3	0.97	0.00
	5	119.2 ± 6.8	118.0 ± 7.4	0.07	0.17
	9	100.2 ± 6.9	98.3 ± 7.2	0.02*	0.28
Carry (yd)	D	217.2 ± 21.7	218.6 ± 20.3	0.51	0.07
	5	169.4 ± 14.0	166.6 ± 15.2	0.16	0.19
	9	131.3 ± 13.1	129.0 ± 12.9	0.05	0.18
Side Carry (yd)	D	-3.4 ± 13.1	-2.6 ± 13.8	0.81	0.06
	5	1.2 ± 6.9	4.0 ± 8.0	0.17	0.26
	9	-3.4 ± 4.3	-2.0 ± 4.7	0.19	0.31
Spin Rate (rpm)	D	3371.8 ± 416.8	3453.1 ± 562.5	0.75	0.16
	5	4865.8 ± 597.7	4734.4 ± 907.4	0.12	0.17
	9	7247.0 ± 870.2	7219.6 ± 1156.5	0.22	0.03

Note. * indicates significant difference (p < 0.05) between means.

Table A 5.2 Clubhead characteristics.

Parameter	Club	Golf Clothing	Sensors Attached	*p*	*d*
Clubhead Speed (mph)	D	97.8 ± 5.8	97.8 ± 5.3	0.89	0.01
	5	85.9 ± 4.1	84.8 ± 4.3	0.01*	0.25
	9	78.3 ± 3.2	77.4 ± 4.1	0.17	0.22
H Swing Plane (°)	D	1.21 ± 5.62	1.03 ± 5.61	0.73	0.03
	5	-2.48 ± 4.33	-1.54 ± 4.76	0.06	0.21
	9	-0.84 ± 4.44	-0.70 ± 5.38	0.78	0.03
Face Angle (°)	D	0.89 ± 2.66	1.02 ± 2.85	0.75	0.05
	5	0.28 ± 2.17	1.01 ± 2.72	0.12	0.29
	9	-1.02 ± 2.30	-0.37 ± 2.49	0.22	0.27
Attack Angle (°)	D	-1.80 ± 2.25	-1.86 ± 2.26	0.76	0.16
	5	-3.85 ± 1.97	-3.86 ± 2.00	0.99	0.17
	9	-4.63 ± 3.02	-5.06 ± 2.68	0.46	0.27

Note. * indicates significant difference (p < 0.05) between means.

A5.4. Discussion

The aim of this study was to establish the effect of electro-magnetic sensor attachment on golf swing mechanics. Due to the physical characteristics of electro-magnetic sensors and use of wires to contact them to a host computer, the attachment of electromagnetic sensors to golfers has the potential to restrict normal golf swing mechanics and affect the quality of shots (Manal *et al.*, 2000; Wright, 2008). Despite these concerns, the majority of results suggested that the attachment of electromagnetic sensors had no effect on measures of ball flight and clubhead characteristics. Similar mean values and low effect sizes were reported for measures of carry, side carry, spin rate, horizontal swing plane, face angle and attack angle (Table A 5.1 and Table A 5.2). Although these results are specific to the sensor attachment technique and the specially designed suit used in this study it is anticipated that they can be generalised to the attachment of electromagnetic sensors using alternative techniques.

Although the majority of results suggested that electromagnetic sensor attachment had no effect on the quality of golf shots, significant differences were identified between conditions for measures of ball speed when the 9 iron was used (Table A 5.1) and clubhead speed when the 5 iron was used (Table A 5.2). These results imply that electromagnetic sensor attachment may negatively affect the quality and speed of golf swings with the iron clubs. However, when multiple comparisons like this are made, the *p* value should not be interpreted in isolation as this can cause type I errors to occur (Feise, 2002).

When the results of the study which examined the test-retest reliability of the Trackman Pro (Appendix VI) were also considered, the results of this study indicate that electromagnetic sensor attachment did not have a meaningful effect on the speed and quality of golf shots. The magnitude of the changes in ball speed with the 9 iron (1.9 mph) and clubhead speed with the 5 iron (1.1 mph) reported in this study were smaller than the minimal differences identified for these measures in Appendix VI (Table A 6.3 and Table A 6.4). As the minimal difference represents the required variance between mean values for an effect to be considered real (Weir, 2005), the results suggest meaningful differences were not produced for these variables. Furthermore, the differences between mean ball speed and clubhead speed were within the limits of agreement reported for these measures in Appendix VI (Table A 6.3 and Table A 6.4). When combined with the low effect sizes reported for 9 iron ball speed (d = 0.28) and 5 iron clubhead speed (d = 0.25) these results suggest that, despite being statistically significant, the differences reported in ball speed and clubhead speed are of little practical relevance.

The results presented in this study suggest that electromagnetic sensor attachment does not affect clubhead or ball flight characteristics. As changes in these impact and launch characteristics are reflective of changes in golf swing kinematics (Lamb, 2012), it was hypothesised that electromagnetic sensor attachment does not affect normal golf swing mechanics. However, this hypothesis should not be accepted without consideration of the fact that a variety of motor patterns can emerge under different task constraints to achieve stable movement outcomes (David *et al.*, 2003; Stergiou, 2004). For example, it has been demonstrated in pistol shooting that, under changeable task demands high levels of variability in the shoulder and elbow joints complemented each other to allow the pistol to maintain a stable position (Scholz, Schoner and Latash, 2000). Therefore, it is possible that stable outcome measures of ball flight and clubhead characteristics could have been produced using different swing mechanics. A more comprehensive understanding of the effect of electro-magnetic sensor attachment on golf swing mechanics could have been achieved by comparing kinematic data produced under the two testing conditions. However, collecting accurate kinematic data in both conditions would have been extremely difficult as condition one required golfers to wear their normal golf attire and the electromagnetic sensors may have obscured the retro-reflective markers in condition 2.

A5.5. Conclusion

This study reported that the attachment of electromagnetic sensors to golfers has no effect on outcome measures of ball flight and clubhead characteristics. Although lower clubhead speeds and ball speeds were produced with the two iron clubs when sensors were attached to golfers the difference between measures was smaller than that expected due to Trackman variability. Taking effect sizes and limits of agreement into account, the results suggest that electro-magnetic sensor attachment did not have a meaningful effect on golf swing mechanics and provide further support for the use of electromagnetic tracking systems in kinematic golf swing analyses.

A6. Appendix VI - Golf Shot Measurement: The Reliability of a Commercial Launch Monitor

A6.1. Introduction

The effectiveness of any golf shot is dictated by the distance and direction (Hellstrom, 2009). Although external factors such as wind, air density and friction of the landing surface affect the outcome, the internal components controlled by the player predominantly determine the quality of each shot (Sweeney *et al.*, 2009). Hellstrom, (2009) considered clubhead speed, centeredness of contact, clubhead path, club face angle and angle of approach to be critical parameters which affect ball flight. Therefore, the ability to repeatedly measure these variables as well as ball flight characteristics such as initial ball velocity, direction and ball spin is extremely important in attempting to examine the quality of golf shots (Lamb, 2012).

Launch monitors are most commonly used by golf coaches and club fitters to produce quantitative measurements of ball flight (Sweeney *et al.*, 2009). They have also been used to provide data that form the basis of scientific golf studies (Betzler *et al.*, 2012; Healy *et al.*, 2011; Kenny, Wallace and Otto, 2008; Lamb, 2012; Myers *et al.*, 2008) as they are capable of predicting the outcome of shots in laboratory environments where the ball is not able to reach its endpoint (Sweeney *et al.*, 2012). Several ball flight monitors such as the Vector Pro (Accusport, Winston-Salem, USA), the FlightScope X-Series (EDH, Ltd., South Africa), the PureLaunch (Zelocity, USA) and the Trackman Pro (Trackman A/S, Vedback, Denmark) are commercially available. Despite their widespread use, limited data has been published regarding their accuracy, validity and reliability.

Sweeney *et al.* (2009) assessed the accuracy of commercial tracking systems. Sweeney *et al.* (2009) compared initial ball flight parameters collected using a Vector Pro launch monitor to ball flight parameters derived from 3D data collected using a Vicon optical motion analysis system (Oxford Metrics, Oxford, UK). The launch monitor data compared favourably with the benchmark 3D system for all launch parameters analysed (launch angle, side angle and ball velocity). Furthermore, the mean error of each variable was small in comparison to the standard deviation of the 3D data (Table A 6.1).

The Trackman Pro is the golf industry preferred swing and ball flight analysis system. It has also frequently been used in scientific golf studies (Betzler *et al.*, 2012; Neal *et al.*, 2007; Robertson *et al.*, 2012). This device utilises two integrated Doppler-radar transceivers to provide real time swing and ball flight data. It has been suggested that this device is capable of estimating the landing position of the ball with an accuracy of 1 foot at 100 yards (30.5 cm at 91.4 m) and measuring clubhead speed with an accuracy of ±1 mph (Tuxen, 2009). Despite this data reported by Trackman suggesting that an appropriate level of accuracy can be achieved, there remains insufficient data regarding the reliability and therefore validity of the Trackman Pro.

Table A 6.1 Mean correlation, mean error, max error and 3D standard deviations (SD) for the comparison of ball flight data captured using 3D analysis and a commercial launch monitor (Sweeney *et al.* 2009).

Variable	Correlation	Mean Error	Max Error	SD of 3D data
Launch Angle	0.96	0.5 ± 0.6°	3.2°	2.4°
Side Angle	0.93	1.1 ± 0.9°	4.6°	3.9°
Ball Velocity	0.95	1.0 ± 1.0 m/s	4.8 m/s	14.3 m/s

To support the use of commercial launch monitors in golf coaching, club fitting and scientific study it is important to establish the uncertainty in their measurements (Mullineaux, Bartlett and Bennett, 2001). It has been suggested that an assessment of reliability can provide an indication of the variation of a measurement protocol (Currell and Jeukendrup, 2008). An assessment of Trackman reliability would also enable subsequent studies to determine, with confidence if actual differences in ball flight or clubhead characteristics exist (Mullineaux, Bartlett and Bennett, 2001). Therefore, the aim of this study was to assess the test-retest reliability of the Trackman Pro launch monitor.

A6.2. Methods

A6.2.1. Participants

Eight male golfers (M ± SD; age 29 ± 12 years, height 182.7 ± 3.4 cm, mass 86.5 ± 8.2 kg and handicap 9.9 ± 7.9 strokes, range 0 - 20 strokes) were recruited for this study. Participants were excluded if their handicap was greater than 20 or if they had an injury which was preventing them from regularly playing or practising golf. Ethics approval was obtained from the Faculty of Health and Wellbeing Research Ethics Committee and before the data collection, written informed consent was obtained from each golfer.

A6.2.2. Data collection

Golfers were required to hit fifteen shots, five with a driver, five with a 5 iron and five with a 9 iron into a net approximately 5 meters away. Each shot was qualitatively rated on a ten point scale with a 1 representing a shot the player was completely unsatisfied with and 10 representing their interpretation of the perfect swing. Golfers used a Ping G15 driver and Ping i15 irons with a regular graphite shaft, standard length and standard lie angle which were provided by the researcher. To control for confounding order effects the clubs were presented in a randomly assigned order (Thomas, Nelson and Silverman, 2011).

Before data collection, golfers were given time to perform their usual pre-game warm-up routine. Sufficient practice trials were also allowed to ensure that the golfers were familiar with the clubs, the laboratory environment and the data collection protocol. After the initial data collection golfers were asked to return to the laboratory to hit another fifteen golf shots. In this collection the golfers performed the same warm-up, followed exactly the same protocol and the order in which the clubs were presented replicated the order used in the initial data collection.

A6.2.3. Measurement device and experimental set-up

For each golf shot, ball flight and swing parameters were measured using a Trackman Pro (Trackman A/S, Vedback, Denmark) launch monitor. The launch monitor was set-up in accordance with the manufacturer's instructions and was positioned in line with the target, directly behind the hitting mat (Figure A 5.1). To ensure that measurements of ball flight characteristics were consistent and accurate, Titleist Pro VI golf balls with a reflective marker provided by Trackman attached were used for every shot.

The ball flight parameters examined were; initial ball velocity, spin rate, carry distance and side carry. Clubhead parameters clubhead velocity, horizontal swing plane, attack angle and face angle were also examined. Comprehensive definitions of these variables can be found in section A5.2.3.

A6.2.4. Statistical analysis

All data were analysed using SPSS for windows (Version 19.0). The mean of the five shots using each club in both data collections was used for statistical analysis. Tests of normality were performed to ensure each data set was appropriate for parametric statistical tests. Reliability was then assessed using a variety of statistical tests. Paired sample t-tests were performed for each variable to compare the mean values produced using each club in the two data collection sessions. Intraclass correlation coefficients (ICCs) were also calculated to enable agreement between means produced for each variable in both data collections to be determined. Based on the recommendations of Shrout and Fleiss (1979) an ICC (2.1) was chosen to establish test-retest reliability.

To express measurement error in the original units of measurement for each variable the standard error of measurement (SEM) was calculated (Equation A 1.1). The SEM was then used to calculate the minimum difference (MD) needed between measures of a variable for the difference to be considered real (Weir, Therapy, & Moines, 2005) (Equation 4.1). The test-retest reliability of the Trackman Pro was also assessed using Bland and Altman's 95% limits of agreement (Bland and Altman 1986).

A6.3. Results

Examination of the Shapiro-Wilk tests indicated that all of the dependent variables were normally distributed and appropriate for parametric statistical analyses. The majority of dependent variables were measured without any meaningful or statistically significant bias when the driver was used (Table A 6.2). With the exception of spin rate ($t(7) = -3.77$, $p = 0.01$, d = 0.31), there were no significant differences between test-retest measures when the driver was used. All of the ICC (2.1) values were also greater than 0.7, a value previously used in reliability studies to indicate acceptable reliability (De Vet *et al.*, 2006; Munro, 1986).

Table A 6.2 Trackman Pro launch monitor reliability measures for the driver.

Parameter	Test	Retest	*P*	ICC	SEM (MD)	LOA
Ball Speed (mph)	145.2 ± 7.1	144.9 ± 6.6	0.86	0.91	2.0 (5.6)	0.3 ± 8.3
Carry (yd)	224.8 ± 11.3	222.3 ± 12.6	0.14	0.96	2.4 (6.8)	2.6 ± 8.7
Side Carry (yd)	-4.9 ± 13.1	3.7 ± 10.6	0.66	0.90	3.5 (9.8)	-1.2 ± 7.4
Spin Rate (rpm)	3403 ± 501	3562 ± 513	0.01*	0.96	97 (269)	-159 ± 235
Club Speed (mph)	100.2 ± 2.4	100.5 ± 2.8	0.60	0.94	0.6 (1.7)	-0.2 ± 2.6
H Swing Plane (°)	1.5 ± 6.2	1.8 ± 6.7	0.36	1.00	0.4 (1.1)	-0.3 ± 1.6
Face Angle (°)	0.9 ± 3.2	0.7 ± 3.6	0.72	0.97	0.6 (1.7)	0.2 ± 2.6
Attack Angle (°)	-1.8 ± 3.0	-2.7 ± 2.8	0.14	0.91	0.8 (2.3)	0.9 ± 3.0

* denotes significant difference

In a similar manner to the driver results, the majority of variables were measured with acceptable test retest reliability when the 5 iron was used (Table A 6.3). All of the ICC (2.1) values were again greater than 0.7 and with the exception of ball speed ($t\ (7) = 3.90$, $p = 0.01$, d = 0.36), there were also no significant differences between the test and retest measures. There were no significant differences between the test and retest measures for any of the variables when the 9 iron was used. The results of the ICC (2.1), SEM and LOA also suggest that acceptable test-retest reliability was achieved when measuring 9 iron ball flight and swing characteristics (Table A 6.4).

Table A 6.3 Trackman Pro launch monitor reliability measures for the 5 iron.

Parameter	Test (Mean ± SD)	Retest (Mean ± SD)	*P*	ICC	SEM (MD)	LOA
Ball Speed (mph)	119.9 ± 5.9	117.2 ± 6.6	0.01*	0.93	1.6 (4.5)	2.7 ± 3.9
Carry (yd)	169.8 ± 11.9	167.4 ± 11.1	0.11	0.97	2.0 (5.6)	2.4 ± 7.1
Side Carry (yd)	1.5 ± 4.2	2.4 ± 7.2	0.76	0.74	2.1 (5.8)	-1.0 ± 16.8
Spin Rate (rpm)	4826 ± 417	4864 ± 880	0.87	0.88	211 (585)	37 ± 1184
Club Speed (mph)	87.0 ± 2.4	85.7 ± 3.0	0.12	0.78	1.2 (3.2)	1.3 ± 4.1
H Swing Plane (°)	-1.3 ± 4.6	-0.6 ± 4.4	0.30	0.96	0.9 (2.5)	-0.7 ± 3.5
Face Angle (°)	0.6 ± 2.1	0.9 ± 2.2	0.60	0.82	0.8 (2.3)	-0.3 ± 3.4
Attack Angle (°)	-3.3 ± 2.2	-4.0 ± 1.6	0.22	0.81	0.8 (2.2)	0.7 ± 3.0

*denotes significant difference

Table A 6.4 Trackman Pro launch monitor reliability measures for the 9 iron.

Parameter	Test (Mean ± SD)	Retest (Mean ± SD)	*P*	ICC	SEM (MD)	LOA
Ball speed (mph)	101.4 ± 5.9	98.6 ± 6.7	0.07	0.87	2.2 (6.0)	2.8 ± 7.2
Carry (yd)	134.2 ± 10.8	131.6 ± 10.7	0.27	0.91	3.1 (8.5)	2.5 ± 11.8
Side Carry (yd)	-3.4 ± 4.8	-0.4 ± 4.0	0.09	0.71	2.1 (5.7)	-3.1 ± 8.5
Spin Rate (rpm)	7482 ± 606	7273 ± 788	0.17	0.91	206 (572)	210 ± 754
Club Speed (mph)	79.0 ± 3.0	78.9 ± 2.3	0.77	0.82	1.0 (2.9)	-0.2 ± 4.3
H Swing Plane (°)	0.0 ± 4.7	1.5 ± 3.9	0.14	0.94	1.0 (2.8)	-1.4 ± 3.2
Face Angle (°)	-1.1 ± 2.9	0.7 ± 2.2	0.06	0.78	1.1 (3.0)	-1.9 ± 4.3
Attack Angle (°)	-5.6 ± 2.7	-4.1 ± 1.7	0.12	0.73	1.0 (2.8)	-1.5 ± 4.9

A6.4. Discussion

The main aim of this study was to assess the test-retest reliability of the Trackman Pro launch monitor. The majority of results indicated that, regardless of club type, the Trackman Pro was able to measure ball flight and clubhead characteristics with acceptable test-retest reliability. Despite these encouraging results, a small number of statistical tests questioned the reliability of the Trackman Pro. More specifically, the paired samples *t*-test indicated that significant differences were evident between the test-retest measures of driver spin rate and 5 iron ball velocities.

Although questionable reliability was suggested by the results of the *t*-test, this statistic should not be considered in isolation (Feise, 2002). A high ICC value (0.96) and a medium effect size (d = 0.36) were also reported for measures of spin rate with the driver which suggest that acceptable reliability may have been achieved. Furthermore, the difference between the test and retest (-159 rpm) was lower than the minimal difference (268 rpm) required for the difference to be considered real. The suggestion that spin rate was measured with acceptable reliability is reinforced when the results are considered from a practical perspective.

As spin rate affects the aerodynamics of the launched ball, there is an optimal amount of spin which will maximise shot distance (Penner, 2003). If the ball doesn't have enough spin it will not be able to maintain its flight however, too much spin will cause the trajectory to be too steep which significantly reduces carry distance (Lamb, 2012). Tuxen (2010) suggested that for a golf ball hit an initial velocity (150 mph) similar to that exhibited in this study, an increase of 1000 rpm translates to a 15 – 22 yard (13.7 – 20.1 m) decrease in carry distance when spin rates between 2000 and 4000 rpm are produced (Table A 6.5). The SEM (96.9 rpm), 95% LOA (234.7 rpm) and difference between the test-retest values (-159.4 rpm) reported in this study for driver spin rate (Table A 6.2) would all translate to much smaller changes in carry distance.

Table A 6.5 The effect of spin rate on the carry distance when using a driver (Tuxen 2010).

Ball Speed (mph)	Launch Angle (°)	Spin Rate (rpm)	Carry (yd)
150	13	2000	261
150	13	3000	246
150	13	4000	224

The majority of statistical tests suggested that 5 iron ball velocities were measured with acceptable reliability (Table A 6.3). However, the repeated measures *t*-test indicated that a significant difference was evident between test-retest measures. Despite this finding, the difference between the test-retest measures (2.7 mph) reported in this study was similar to the mean error (2.4 mph) reported by Sweeney *et al.* (2012) in the measurement of driver ball velocity. Furthermore, only a medium effect size (d = 0.36) was reported and the difference in 5 iron ball velocity did not have an effect on carry distance, a measure which is dependent on the magnitude of ball velocity (Kenny, Wallace and Otto, 2008).

The results suggest that the Trackman Pro has sufficient reliability to be used for golf coaching, club fitting tool and scientific golf study. In addition, the SEM provides an indication of the expected measurement error for each dependent variable. For each variable, knowledge of this measurement error as well as the MD will enable practitioners to determine, whether a difference in repeated measurements is practically meaningful. As well as the LOA, these values will be particularly useful in scientific studies which use ball flight or swing characteristics as dependant variables.

A6.5. Conclusion

The results of this study suggest that commercial launch monitors in particular the Trackman Pro can produce measures of clubhead and ball flight characteristics with acceptable reliability. The findings support the use of this launch monitor in golf coaching, club fitting and scientific study. They also provide an indication of the measurement error to expect for each variable which will enable practitioners to determine if meaningful differences occur between repeated ball flight and clubhead measurements.

A7. Appendix VII - The Effect of Inertial Parameter Estimation Error on the Magnitude and Timing of Peak Segment Kinetic Energy in the Golf Swing

A7.1. Introduction

In golf, players often seek to maximise the distance and control the direction of shots. To achieve these goals, it has been suggested that golfers should coordinate the movement of multiple body segments (Putnam, 1993; Vena *et al.*, 2011a). As a result, many golf coaches and biomechanists have examined the sequencing of body segment movements in the golf swing. Frequently this has been analysed by calculating the magnitude and timing of peak segment angular velocity (Cheetham *et al.*, 2008; Neal *et al.*, 2007, Tinmark *et al.*, 2010). However, such analyses have yet to provide conclusive evidence regarding the existence of the proximal-to-distal sequence in the golf swing.

It has been suggested that the analysis of segment KE is the most appropriate when examining the sequencing of body segment movements in the golf swing (Anderson, Wright and Stefanyshyn, 2006). As well as accounting for body segment inertial parameters (Anderson, Wright and Stefanyshyn, 2006), analyses of segment KE are much more sensitive to subtle changes in technique than analyses of angular velocity (Bechard, 2009). As a result of these advantages, analyses of segment KE have been used to examine segmental sequencing in cricket fast bowling (Ferdinands, Kersting and Marshall, 2012), tennis serving (Lopez and Navarro, 2009) and in sub-maximal and maximal jumping (Lees *et al.*, 2004).

Given the widespread use of analyses of segment KE, it is tempting to take the results at face value. However, the output of such analyses is only valid when the errors associated with its calculation are quantified (Sellers and Crompton, 2004). Furthermore, establishing the inherent variability in the output of segment KE analyses can determine the certainty with which statements can be made about the results of subsequent analyses (Sellers and Crompton, 2004). Such an analysis also enables data collection errors to be identified and controlled in an attempt to reduce the magnitude of output error (Mullineaux and Bartlett, 1997).

The calculation of segment KE requires the measurement of kinematic data and the estimation of BSIP. In this programme of research, kinematic data were captured using a Polhemus electromagnetic tracking system whilst BSIP were estimated using the geometric model described in section 3.2. Using an electromagnetic tracking system, kinematic data can be accurately and reliably measured (Milne *et al.*, 1996; Richards *et al.*, 1999; Schuler *et al.*, 2005; Thewlis *et al.*, 2007). However, larger errors have been associated with the estimation of BSIP using the geometric modelling technique (Outram, Domone and Wheat, 2011). Therefore, before analyses of segment KE can be used with confidence, the influence of such errors on measures of segment KE must be quantified.

A simple way to assess the influence of BSIP estimation error is to perform multiple calculations of segment KE using different sets of inertial parameters. Sensitivity analyses such as this are frequently referred to in biomechanical literature but are rarely actually performed (Sellers and Crompton, 2004). To run this type of analysis, the input parameters should be manipulated by amounts relating to the accuracy with which they can be measured (Mullineaux and Bartlett, 1997). This enables input parameters that have the largest effect on the results to be identified and improvements to be made in the data collection technique (Pearsall and Costigan, 1999). Therefore, the aim of this analysis was to determine the extent with which errors in the estimation of BSIP effect measures of segment KE in the golf swing.

A7.2. Methods

A7.2.1. Participant

One golfer (age, 23 years, mass, 92.0 kg, stature, 184.4 cm and handicap, 13 strokes) was recruited to take part in this study. This golfer was chosen as a set of inertial parameters had previously been calculated using the gold standard volume and shape measurements from the body scanning protocol (Appendix II: participant 2). Before the data collection, ethics approval was granted by the Faculty of Health and Wellbeing Research Ethics Committee and the participant provided informed consent.

A7.2.2. Inertial parameters

KE was calculated using two sets of inertial parameters (Table A 7.1, Table A 7.2 and Table A 7.3). The first set of inertial parameters was estimated using the geometric model (Section 3.2) and the revised hand model (Section 3.3.2). For the feet, lower legs, thighs, neck, head, upper arms, forearms and hands the required anthropometric measurements were determined from the positions of 56 anatomical landmarks using the technique described in section 3.4. However, the required anthropometric measurements for the lower, middle and upper trunk were not available using this technique. For these segments, anthropometric measurements were determined using Pro-Engineer and the technique described in section A2.2.2 which negated anthropometric measurement errors.

The second, more accurate set of inertial parameters were estimated by combining the gold standard volume and shape measurements from the body scanning protocol with a uniform density function (Dempster, 1955 - section A2.2.2. Using this technique, inertial parameters were estimated for the lower legs, thighs, lower, middle and upper trunk, upper arms and forearms. Hand mass was estimated using a water immersion technique (Appendix IV) whilst hand COM and moments of inertia were calculated by altering the estimated hand inertial parameters by the mean error in the estimation of COM and MOI for the other rigid bodies.

The foot, neck and head were not considered in this analysis as pilot research suggested that due to low translational and rotational velocities these rigid bodies generated negligible magnitudes of peak translational and rotational segment KE. For these rigid bodies, it was assumed that inertial parameter manipulation would have little effect on the magnitude or timing of peak KE. However, to enable grouped segment KE to be calculated, their inertial parameters were estimated using the technique described for the first set of parameters.

Table A 7.1 Inertial parameter estimates for the rigid bodies of the Lower Body.

	Set 1	Set 2	Error (%)
Lower Leg			
Mass (kg)	4.76	4.37	8.92
COM (cm)	25.56	25.04	2.09
Ixx (kg/m^2)	0.0709	0.0596	18.88
Iyy (kg/m^2)	0.0082	0.0075	9.06
Izz (kg/m^2)	0.0704	0.0588	19.67
Thigh			
Mass (kg)	14.02	15.34	-8.66
COM (cm)	30.16	28.97	4.12
Ixx (kg/m^2)	0.2741	0.3072	-10.76
Iyy (kg/m^2)	0.0698	0.0863	-19.17
Izz (kg/m^2)	0.2746	0.2897	-5.21
Lower Trunk			
Mass (kg)	9.19	9.76	6.23
COM (cm)	6.48	6.50	0.32
Ixx (kg/m^2)	0.1068	0.1168	9.36
Iyy (kg/m^2)	0.0849	0.0877	3.32
Izz (kg/m^2)	0.0498	0.0581	16.70

Table A 7.2 Inertial parameter estimates for the rigid bodies of the Upper Body.

	Set 1	Set 2	Error (%)
Middle Trunk			
Mass (kg)	16.55	15.58	6.26
COM (cm)	12.37	11.86	4.31
Ixx (kg/m^2)	0.2113	0.2045	3.28
Iyy (kg/m^2)	0.1951	0.1764	10.58
Izz (kg/m^2)	0.1695	0.1441	17.62
Upper Trunk			
Mass (kg)	12.48	12.64	-1.26
COM (cm)	7.22	7.49	-3.68
Ixx (kg/m^2)	0.1931	0.2063	-6.41
Iyy (kg/m^2)	0.1560	0.1733	-9.98
Izz (kg/m^2)	0.0870	0.0804	8.10

Table A 7.3 Inertial parameter estimates for the rigid bodies of the Arms.

	Set 1	Set 2	Error (%)
Upper Arm			
Mass (kg)	3.21	2.72	17.91
COM (cm)	9.68	9.70	-0.19
Ixx (kg/m^2)	0.0179	0.0204	-12.41
Iyy (kg/m^2)	0.0048	0.0043	11.74
Izz (kg/m^2)	0.0173	0.0193	-10.11
Forearm			
Mass (kg)	1.64	1.58	3.85
COM (cm)	11.13	10.78	3.32
Ixx (kg/m^2)	0.0095	0.0086	10.61
Iyy (kg/m^2)	0.0015	0.0015	2.03
Izz (kg/m^2)	0.0094	0.0083	13.73
Hand			
Mass (kg)	0.51	0.51	1.15
COM (cm)	7.00	7.01	-0.11
Ixx (kg/m^2)	0.0008	0.0007	1.79
Iyy (kg/m^2)	0.0002	0.0002	-2.37
Izz (kg/m^2)	0.0010	0.0009	8.64

A7.2.3. Data collection protocol

Fifteen golf swings; five with a driver, 5-iron and 9-iron were performed. Each shot was qualitatively rated on a ten point scale with a 1 representing a shot the player was completely unsatisfied with and 10 representing their interpretation of the perfect swing. Shots rated as less than seven were discounted and another shot was hit. A Ping G15 driver and Ping i15 irons with regular graphite shafts, standard lengths and standard lie angles were used. Sufficient time was given for the golfer to perform their usual pre-game warm-up routine and adequate practice trials were allowed to ensure that the golfer was familiar with the clubs, the laboratory environment and the data collection protocol.

A7.2.4. Kinematic data collection

Kinematic data were collected using a sixteen-channel Polhemus Liberty electromagnetic tracking system (Polhemus, Inc., Colchester, VT, USA) sampling at 240 Hz and the technique described in section 3.4.2

A7.2.5. Kinetic energy calculations

Using both sets of inertial parameters segment KE was calculated using the technique described in section 3.4.4. As in this section and the papers by Anderson, Wright and Stefanyshyn, (2006) and Kenny *et al.* (2008) KE was calculated for four grouped segments (Lower Body, Upper Body, Arms and Club) (Table 3.11). To examine the effect of specific errors in BSIP estimation, rigid body KE was also calculated for each swing.

A7.2.6. Sensitivity analysis

Errors in the estimation of the magnitude timing of peak segment KE were calculated using Equation A 7.1. Estimated segment KE represented measures of segment KE calculated using inertial parameters derived from the geometric modelling technique. Golf standard (GS) segment KE represented measures of KE calculated using inertial parameters derived by combining the gold standard volume and shape measurements from the body scanning protocol with a uniform density function. These errors were calculated for individual rigid bodies and for the three human based grouped segments (Lower Body, Upper Body and Arms).

$$Error\ (\%) = \left(\frac{(Estimated\ Segment\ KE - GS\ Segment\ KE)}{GS\ Segment\ KE}\right) \times 100 \qquad \text{Equation A 7.1}$$

A7.3. Results

With all three clubs, inertial parameter manipulation had a small effect on the magnitudes of peak total segment KE (Table A 7.4, Table A 7.5 and Table A 7.6). Inertial parameter manipulation had the smallest effect on the magnitude of peak total Upper Body KE. For all three clubs, the difference between peak total Upper Body KE was less than 0.4 J and the maximum error was only 1.30 % (Table A 7.5). Small differences of only -1.1 - -1.6 J were also produced in the estimation of peak total Lower Body KE which caused relatively small errors of only -3.56 - -4.04 %. Errors in inertial parameter estimation had the largest effect on the magnitude of peak total Arms KE with the largest difference (5.8 J) being reported when the driver was used.

Inertial parameter manipulation had a small effect on the magnitude of peak Upper Body and Arms translational kinetic energies. Larger errors were reported for measures of peak Lower Body translational KE but the maximum difference remained under 0.8 J. Furthermore, errors in the estimation of inertial parameters had only a small effect on the magnitude of peak Upper Body local and remote rotational kinetic energies (Table A 7.4, Table A 7.5 and Table A 7.6). Larger errors were reported for these components of KE for the Lower Body and Arms segments. Errors of 5.92 – 6.53 % were reported for the measures of peak remote rotational Arms KE. However, maximum differences of only 0.8 J were reported for peak local rotational Lower Body KE.

Table A 7.4 Effect of BSIP variation on the magnitude and timing of peak segment KE when the driver was used.

	Magnitude			Timing		
	Set 1 (J)	Set 2 (J)	Error (%)	Set 1	Set 2	Error (%)
Total KE						
LB	43.4 ± 5.8	45.0 ± 5.9	-3.56	0.831 ± 0.032	0.831 ± 0.032	0.00
UB	35.8 ± 3.7	35.4 ± 3.6	1.07	0.775 ± 0.043	0.775 ± 0.043	0.00
Arms	115.7 ± 10.5	109.9 ± 10.1	5.29	0.807 ± 0.019	0.807 ± 0.019	0.00
Translational KE						
LB	17.8 ± 3.2	18.5 ± 3.3	-3.85	0.828 ± 0.062	0.828 ± 0.062	0.00
UB	14.3 ± 1.3	14.0 ± 1.3	2.61	0.755 ± 0.043	0.766 ± 0.022	-1.49
Arms	55.8 ± 4.7	54.6 ± 4.7	2.30	0.790 ± 0.017	0.790 ± 0.017	0.00
Local Rotational KE						
LB	16.1 ± 1.4	16.9 ± 1.5	-4.67	0.863 ± 0.014	0.863 ± 0.014	0.00
UB	24.4 ± 0.6	24.8 ± 0.6	-1.23	0.744 ± 0.017	0.744 ± 0.017	0.00
Arms	7.7 ± 0.3	7.3 ± 0.3	5.34	0.876 ± 0.100	0.879 ± 0.097	-0.34
Remote Rotational KE						
LB	15.8 ± 1.5	16.8 ± 1.6	-5.86	0.816 ± 0.023	0.816 ± 0.023	0.00
UB	3.0 ± 0.3	3.0 ± 0.3	1.58	0.930 ± 0.089	0.930 ± 0.089	0.00
Arms	56.6 ± 3.0	53.1 ± 2.9	6.53	0.957 ± 0.097	0.953 ± 0.096	0.31

Nb: LB – Lower Body, UB – Upper

Table A 7.5 Effect of BSIP variation on the magnitude and timing of peak segment KE when the 5 iron was used.

	Magnitude			Timing		
	Set 1	Set 2	Error (%)	Set 1	Set 2	Error (%)
Total						
LB	31.9 ± 2.0	33.2 ± 2.1	-4.04	0.792 ± 0.108	0.792 ± 0.108	0.00
UB	29.7 ± 0.9	29.3 ± 0.8	1.30	0.715 ± 0.067	0.715 ± 0.067	0.00
Arms	98.7 ± 4.8	93.6 ± 4.5	5.40	0.839 ± 0.091	0.839 ± 0.091	0.00
Translational						
LB	11.2 ± 0.9	11.7 ± 1.0	-4.42	0.671 ± 0.091	0.697 ± 0.069	-3.74
UB	11.7 ± 0.6	11.3 ± 0.5	3.42	0.664 ± 0.047	0.664 ± 0.047	0.00
Arms	51.2 ± 3.3	49.9 ± 3.2	2.68	0.775 ± 0.023	0.775 ± 0.023	0.00
Local Rotational						
LB	12.8 ± 1.0	13.3 ± 1.1	-3.99	0.656 ± 0.010	0.653 ± 0.011	0.48
UB	21.4 ± 0.8	21.6 ± 0.8	-1.12	0.737 ± 0.004	0.737 ± 0.005	0.00
Arms	7.8 ± 4.1	7.4 ± 0.4	5.96	0.985 ± 0.000	0.985 ± 0.001	0.00
Remote Rotational						
LB	13.1 ± 0.9	13.9 ± 0.9	-5.96	0.798 ± 0.111	0.798 ± 0.112	0.00
UB	1.7 ± 0.2	1.7 ± 0.2	0.95	0.847 ± 0.072	0.847 ± 0.073	0.00
Arms	49.8 ± 2.7	46.8 ± 2.4	6.28	0.985 ± 0.010	0.985 ± 0.011	0.00

Table A 7.6 Effect of BSIP variation on the magnitude and timing of peak segment KE when 9 iron was used.

	Magnitude			Timing		
	Set 1	Set 2	Error (%)	Set 1	Set 2	Error (%)
Total						
LB	27.8 ± 2.7	28.9 ± 2.9	-3.94	0.857 ± 0.016	0.857 ± 0.016	0.00
UB	26.6 ± 0.9	26.3 ± 0.9	1.16	0.748 ± 0.023	0.748 ± 0.023	0.00
Arms	92.9 ± 3.5	88.3 ± 3.3	5.13	0.802 ± 0.033	0.797 ± 0.025	0.71
Translational						
LB	8.6 ± 0.9	9.0 ± 0.9	-4.66	0.785 ± 0.108	0.787 ± 0.110	-0.36
UB	10.7 ± 0.9	10.4 ± 0.9	2.96	0.759 ± 0.102	0.759 ± 0.102	0.00
Arms	49.5 ± 2.5	48.2 ± 2.3	2.79	0.771 ± 0.015	0.771 ± 0.015	0.00
Local Rotational						
LB	13.1 ± 0.6	13.6 ± 0.6	-4.08	0.631 ± 0.030	0.631 ± 0.030	0.00
UB	17.9 ± 0.5	18.1 ± 0.6	-0.92	0.733 ± 0.017	0.733 ± 0.017	0.00
Arms	6.4 ± 0.6	6.1 ± 0.6	5.75	0.896 ± 0.122	0.896 ± 0.122	0.00
Remote Rotational						
LB	12.3 ± 1.1	13.0 ± 1.2	-5.35	0.888 ± 0.033	0.874 ± 0.012	1.61
UB	1.7 ± 0.2	1.7 ± 0.2	1.11	0.877 ± 0.023	0.877 ± 0.023	0.00
Arms	44.8 ± 2.4	42.3 ± 2.2	5.92	0.980 ± 0.008	0.980 ± 0.008	0.00

The causes of errors in the magnitude of peak segment KE were examined by assessing the effect of inertial parameter errors on measures of peak rigid body KE. The underestimation of peak total Lower Body KE and its components were caused by the underestimation of the translational and rotational components of peak thigh KE. For all clubs, the magnitudes of peak translational (-8.17 – -9.70% error), local rotational (-8.19 – -10.23% error) and remote rotational (-8.42 – -8.46% error) thigh kinetic energies were underestimated using the estimated inertial parameters (Figure A 7.1). The analysis of peak rigid body kinetic energies also suggested that the overestimation of peak total Arms KE was caused by the overestimation of peak translational and remote rotational upper arm kinetic energies (Figure A 7.2). Although peak translational and remote rotational forearm and hand inertial parameters were also overestimated, the magnitude of these differences combined (translational = 0.7 ± 0.1 J, remote rotational = 0.9 ± 0.1 J) were much smaller than those produced for the upper arm (translational = 2.6 ± 0.3 J, remote rotational = 2.9 ± 0.3 J).

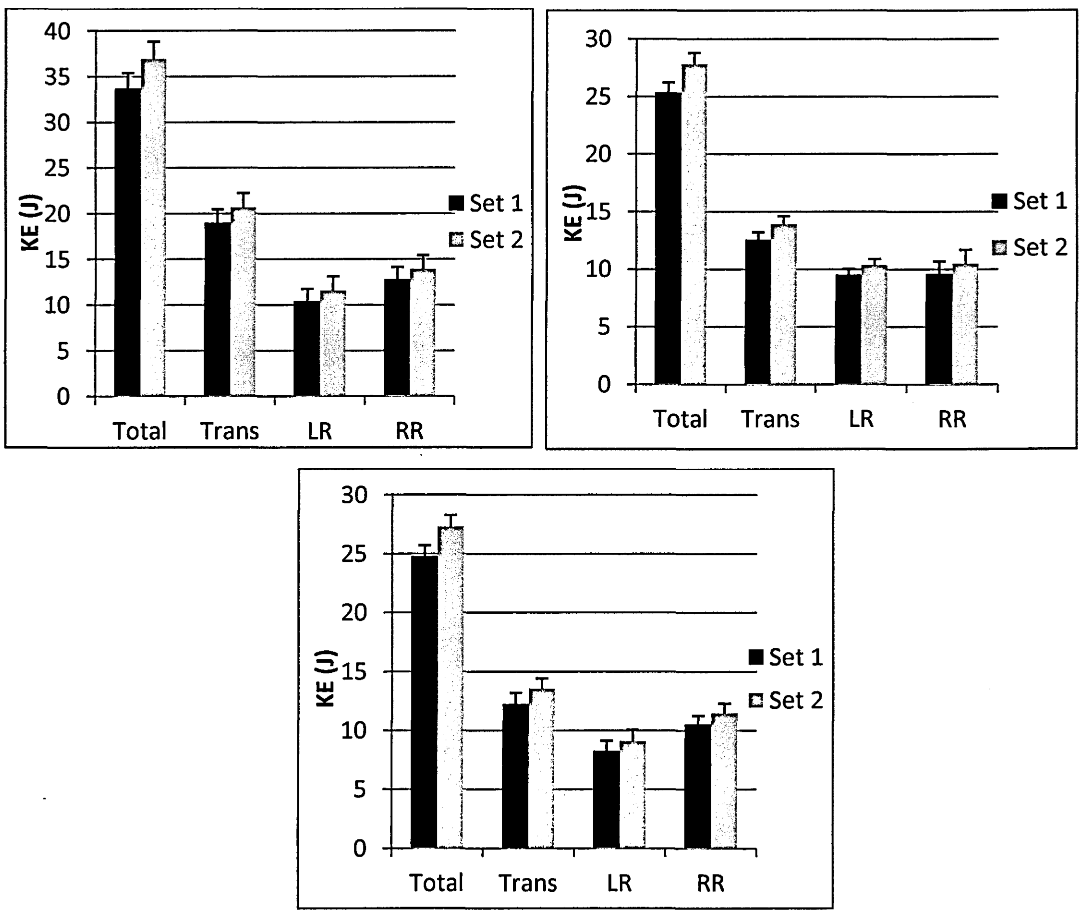

Figure A 7.1 The magnitude of peak total, translational (Trans), local rotational (LR) and remote rotational (RR) thigh KE calculated using both sets of inertial parameters: top left – driver, top right – 5 iron, bottom – 9 iron.

Compared with the magnitude of peak segment KE, inertial parameter error had a much smaller effect on the timing of peak segment KE. When all three clubs were used, inertial parameter manipulation had no effect on the timing of peak total Lower Body and Upper Body energies. There was also no effect of inertial parameter alteration on the timing of peak total Arms KE when the driver and 5 iron were used and only a small effect (0.71 %) when the 9 iron was used (Table A 7.6). The largest error (-3.74 %) was reported for the timing of peak translational Lower Body KE. However, inertial parameter manipulation generally had no effect on the timing of peak segment KE components for the majority of measures (Table A 7.4, Table A 7.5 and Table A 7.6) and only a small effect on others.

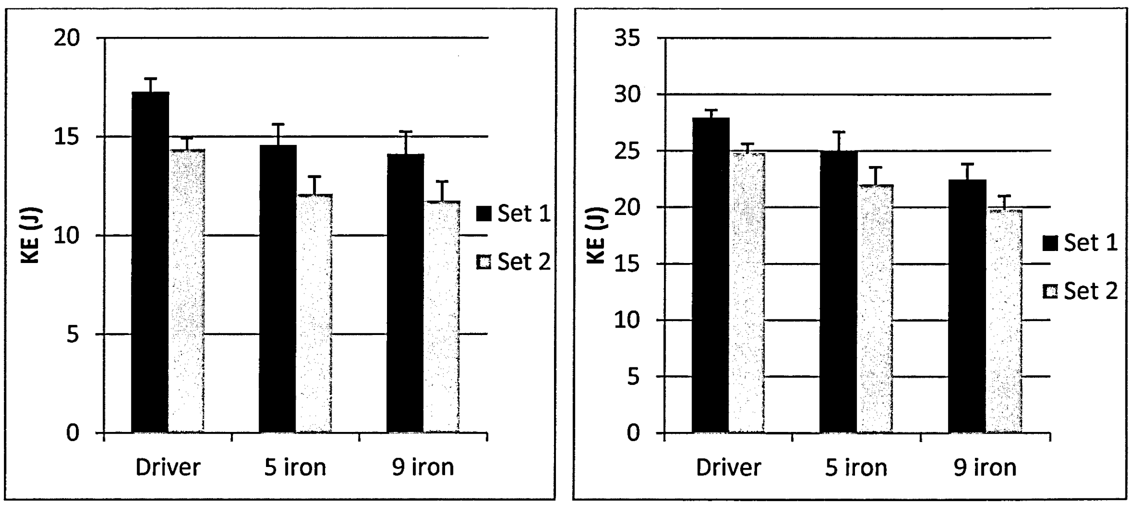

Figure A 7.2 The magnitude peak upper arm KE calculated using both sets of inertial parameters: left – translational KE, right – remote rotational KE. peak total, linear and angular middle trunk, upper trunk, upper arm, forearm and hand KE's.

A7.4. Discussion

The aim of this study was to determine the effect of BSIP estimation error on the magnitude and timing of peak segment KE in the golf swing. The BSIP estimation errors and the results of this study are specific to the golfer analysed. However, as the BSIP estimation errors were similar to those reported for other participants in Appendix II it is possible to generalise the results to golfers with different body morphologies. The results suggested that the magnitude of peak segment KE was much more sensitive to segment inertial parameter estimation errors than the timing of peak segment KE. Regardless of club type, inertial parameter manipulation had only a very small effect on the timing of segment KE. Although some relatively small errors (-3.74 – 0.31 %) were evident for approximately 22% of timing measures in general, the results suggest that future calculations of the timing of peak segment KE can be made with a good level of certainty (Mullineaux and Bartlett, 1997).

Inertial parameter manipulation had a larger effect on the magnitude of peak segment KE. The largest effect was reported for the magnitude peak total Arms KE. The analysis of peak rigid body KE suggested that this overestimation was predominantly caused by the errors associated with the estimation of upper arm inertial parameters. Furthermore, the overestimation of peak total upper arm KE was predominantly caused by the overestimation of upper arm mass and the consequent overestimation of the translational and remote rotational components of peak upper arm KE (Figure A 7.2). In comparison, the errors associated with the estimation of upper arm moments of inertia had a small effect on the magnitude of peak local rotational Arms KE. Therefore, improving the accuracy of upper arm geometry, volume and subsequently mass estimations would significantly improve the accuracy of the estimation of the magnitude of peak Arms KE.

Using the estimated inertial parameters the magnitude of peak total Legs KE and its translational and rotational components was underestimated. Although the Legs segment was composed by four rigid bodies, the accuracy of thigh inertial parameter estimates was predominantly responsible for this inaccuracy. Furthermore, it would appear that the magnitude of peak thigh KE was more reliant on accurate estimations of thigh mass. Peak total thigh KE contained a larger contribution from translational and remote rotational KE, the components derived from thigh mass. Local rotational thigh KE made a much smaller contribution to peak total thigh KE. Subsequently accurate estimates of thigh geometry, volume and mass would appear to be important in future analyses of segment KE.

In comparison to the magnitude of peak Legs and Arms KE, inertial parameter manipulation caused smaller differences to be produced in the magnitude of peak Upper Body KE. As the errors associated with anatomical landmark identification and anthropometric measurements were negated for these segments, these smaller errors for the Upper Body were expected. The accuracy of other palpation dependent analyses such as joint kinematic analyses (Della Croce *et al.*, 2005; Evans *et al.*, 2012) and the magnitude of joint angle estimations (Moriguchi *et al.*, 2009) has been reported to be dependent on the precision of palpation. The identification of anatomical landmarks which define the Upper Body and specifically the middle and upper trunk is also difficult as these segments have high proportions of fat tissue (O'Haire and Gibbons, 2000) and the landmarks are often not easily identifiable points (Della Croce *et al.*, 1999). Therefore, it is anticipated that the errors produced for the estimation of peak middle and upper trunk KE are smaller than the errors that will be produced in future KE analyses. However, this would not necessarily increase the errors produced in the estimation of peak Upper Body KE as similar errors were produced in the overestimation and underestimation peak middle (8.1 ± 0.1 J) and upper (-6.7 ± 0.2 J) trunk kinetic energies respectively.

A7.4.1. Practical application

The results suggest that errors in the estimation of inertial parameters using the geometric modelling technique cause relatively small errors to be produced in the estimation of the magnitude and timing of peak segment KE. However, care should be taken in the identification of anatomical landmarks to ensure larger errors than those used in this study are not produced. In particular care should be taken in the identification of landmarks which define the thigh and the upper arm as it has been indicated that these segments are particularly difficult to model (Appendix II) and the errors associated with mass estimation for these segments appear to be responsible for the inaccuracies in the estimation of peak Legs and Arms kinetic energies.

A7.5. Conclusion

In conclusion, it has been suggested that the magnitude of peak segment KE is more sensitive to changes in inertial parameters than the timing of peak segment KE. In particular, the magnitude of peak segment KE was dependent on the accuracy of estimates of thigh and upper arm geometry, volume and subsequently mass.

Therefore, to ensure that segment KE is estimated with the highest possible accuracy care should be taken in the identification of anatomical landmarks which define these segments.