A biomechanical and physiological investigation of atypical gaits used in badminton

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A BIOMECHANICAL AND PHYSIOLOGICAL INVESTIGATION OF 'ATYPICAL' GAITS USED IN BADMINTON

by

Gregor Kuntze

A Doctoral Thesis

Submitted in partial fulfilment of the requirements for the award of

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Abstract

This thesis is concerned with quantifying the biomechanics and physiological consequences of sport-specific movements in order to answer the question if atypical movements in badminton result in abnormally large demands that could be linked to the relatively high levels of injuries sustained.

An initial study of the movement repertoire used in competitive badminton established that sidestepping (SS), crossover stepping (XS) and lunging movements make an important contribution to the game. These movements are related, within the context of the game, and were viewed as a unit. In order to assess the potential injury risk posed by these atypical movements a series of experiments was performed to record the biomechanical as well as physiological demands of SS, XS and lunging for experienced, inexperienced, male and female badminton players. The first of these studies concerned the kinematics and kinetics of preferred speed SS and XS. This was followed up by an investigation of the electrical activity of 7 muscles of the leading and trailing limb and a comparative assessment of their metabolic demands was performed. The biomechanics of lunging were thereafter investigated, followed by a final investigation of the kinematics of atypical movement use in the competitive setting.

The results from these investigations indicate that lateral stepping tasks result in biomechanical demands that are within the range expected for running. An asymmetric contribution of the leading and trailing limb to the gait cycle was identified as well as a shift toward the use of proximal joints for force production. Furthermore, no significant difference in metabolic power between SS, XS and running was identified. Differences in the demands of different lunging movements were observed with implications for both injury prevention and performance enhancement. Overall it was observed that the data recorded in these investigations was in agreement with the competitive, real-life application.

Based on the findings in this research it can be concluded that lateral stepping movements in badminton do not appear to expose the participant to abnormally large biomechanical or physiological demands and other factors related to movement may be involved in the relatively high levels of injury sustained.
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Introduction

1.1 General Introduction

Badminton is one of the world’s most popular participation sports. It enjoys a strong national and international appeal particularly since the admission of the sport to the Olympic Games in 1992 and the resultant increase in public exposure, recognition and funding (Jørgensen and Hølmich, 1994, Nash, 1999, Fahlström et al., 2002). It is primarily an indoor sport and according to Badminton England there are an estimated 4 million players in the UK (~8% of the population) and who participate on a regular basis (Cooke, 1996). The game is appealing because it combines speed and agility with a strong focus on tactics, which means that the player constantly moves, covering all areas of the court and has to be in good physical condition (Jørgensen and Hølmich, 1994). Movement therefore makes up a fundamental aspect of the game.

Biomechanical analysis provides us with the tools to investigate the performance of the human body in motion. There is a substantial volume of literature related to the analysis of human locomotion with walking and running gaits receiving a lot of interest because they are used regularly in every-day situations and represent a popular recreational past-time (Putnam and Kozey, 1989). These gaits can be classed as typical gaits, since they are readily seen in every-day adult movement. Research has often focused on the assessment of the physiological and biomechanical requirements of these gaits and in relation to sport, with a view to assessing and reducing the risk of injury to the performer. This is in line with the statement by Bartlett who describes the aim of biomechanics in sports as “the reduction of injury and the improvement of performance” (Bartlett, 1999 p. 1).

Many sports including badminton however, not only rely on typical movements but utilise atypical gaits, those not seen in normal adult gait, in order to perform the sport-specific goals. The coaching literature for badminton and observations of competitive singles badminton games suggest that lateral gaits, hopping and quick stopping movements (lunging) contribute heavily to the movement repertoire of competitive
singles badminton. In the context of competitive sport it is therefore thought that atypical gaits perform an important function in the completion of sport-specific goals and contribute to a large degree to the movement repertoire of the athlete in many sports. However, there appears to be very little scientific information relating to the use of sport-specific, atypical movements and their application in badminton.

The following review of the literature will show badminton to be among the physiologically most demanding racket sports and the epidemiological data will show that the sport is related to a comparatively large percentage of lower limb overuse injuries. The specific causes for these injuries however are not known. The mechanical requirements of running and the associated injury potential for the performer have been liked to a number of biomechanical and physiological factors in the past (Cavanagh and Lafortune, 1980, Grimston and Zernicke, 1993, Nigg et al., 1995a, Egermann et al., 2003). It may be argued that the described energy demands and epidemiological data in badminton are at least in part related to the type and application of gait on the badminton court. Quantification of the physiological characteristics of common atypical gaits in badminton is therefore essential for an improved understanding of how these movements affect the athlete during a game and in training. This data may provide vital objective data for the improvement of movement and the prevention of injuries of the athlete.

1.2 Objectives of the Research

This research therefore, concerns the use of atypical gaits in badminton and the biomechanical and physiological consequences of the use of these sport-specific movements. Initially, a literature review of the current level of knowledge relating to the biomechanics, physiological demands and epidemiology of typical and a number of atypical movements were performed. Based on this review, a number of gaps in our knowledge relating to the physiological requirements of atypical gaits were identified. These gaps were closely related to the specific use of movement in badminton and acted as the basis for the further assessment in a series of experiments presented in Chapters 4 to 9. It was the objective of the research to quantify the mechanical loads imposed on the lower limbs of the athlete, identify the coordination of the
musculature during these movement strategies and to evaluate the physiological consequences resulting from the use of atypical gaits. These data will allow for a greater understanding of the contribution of atypical movements to the competitive singles game of badminton, the mechanical and physiological demands of sport-specific movements and will contribute to a more detailed picture of the cause and effect relationship of motion and injury in badminton. This information may help to improve movement coaching and provide opportunities for injury prevention. The investigation of the relationship between movement and these biomechanical and physiological demands was done using a number of steps, summarised below.

1.3 Contents of the Thesis

The research for this thesis can be divided into a number of related parts. The first part concerns the assessment of

Chapter 1 Introduction
The background, objectives and content of this research are explained.

Chapter 2 Literature Review
A review of the literature is performed. The contents of the literature review are presented in four main sections: assessment of gait kinematics and kinetics; muscular activity during gait; assessment of energy expenditure; musculoskeletal injury.

Chapter 3 Methods
The methods used in this thesis are described. This includes information on the equipment used and the validity of the equipment for the required measurements. Analysis techniques are discussed in the individual experimental chapters (Chapters 4–9).

Chapter 4 Video-based classification, quantification and comparative analysis of gait usage in badminton
The first of the experimental chapters deals with the use of locomotion in the competitive singles badminton setting. The aim of the investigation is to quantify gait usage and identify the contribution of atypical movements to the singles game.

Chapter 5 A biomechanical analysis of the stance phase of a standardised, sport-specific sidestep and crossover step task
Chapter 5 investigates the kinematics and kinetics of lateral sidestepping and crossover stepping. The aim of the investigation is to identify the differences in biomechanical variables between two levels of skill as well as male and female participants.

Chapter 6 Kinematic, kinetic and EMG analysis of selected lower limb muscles during lateral sidestepping and crossover stepping tasks
The biomechanics and muscular activity of controlled speed lateral sidestepping and crossover stepping are assessed. The primary aim of the investigation is to identify the activation times of a selection of superficial muscles of the leading and trailing limb. A further aim is to record gait kinematics and kinetics in order to relate muscle activations to the gait mechanics to describe the functional significance of muscle activations.

Chapter 7 A comparative analysis of the metabolic requirements of atypical gaits used in badminton
Chapter 7 investigates the physiological requirements of typical and atypical gaits used in badminton. The aim of the investigation is to identify the changes in physiological demands when using lateral sidestepping and crossover stepping movements compared to walking, running and skipping.

Chapter 8 A biomechanical analysis of common lunge tasks in badminton
The biomechanics of the dominant limb during lunging in badminton are assessed. Three lunge recovery methods are performed during simulated training exercises. Motion capture and reaction force recordings are performed to assess changes between the different lunge tasks.
Chapter 9 Comparative analysis of sidestepping, crossover stepping and lunging in badminton – In-game compared to the laboratory
A comparative analysis was performed to assess the validity of data on the timing of footfall events for sidestepping, crossover stepping and lunging in the lab compared to in-game.

Chapter 10 General overview and conclusion
The findings of the research are summarised in this chapter and recommendations for future research are offered.
Chapter 2

Literature Review

2.1 Introduction

2.1.1 Aim

A lot of research regarding the contribution of the musculoskeletal system to the performance of human gait can be found reported in the literature. This chapter reviews those studies to assess the current knowledge of gait physiology and biomechanics. The aim of the review is to find out what the biomechanical and physiological parameters for typical and atypical gaits are, to identify the proposed factors that predispose the athlete to injury and how these can be avoided. Based on the review, the areas where more information is required for a more complete understanding of the fundamental mechanisms of locomotion and specifically the application of atypical gaits in singles badminton are identified.

2.1.2 Structure

The literature review consists of the following three sections:

Section 2.2: Assessment of gait kinematics and kinetics
This section deals with the description of gait at the level of the movements of the limbs in time and space (kinematics). Furthermore, this section deals with the internal and external forces that the body creates and is exposed to during locomotion (kinetics), as well as the models used to define human locomotion and the musculoskeletal energy saving process. The chapter also deals with the topic of muscular activity in gait. The aim of this section is to present the current knowledge of kinematics, kinetics and EMG resulting from bipedal locomotion and the related topic of injury prevention.
Section 2.3: The metabolic energy requirements of gait
This section deals with the metabolic requirements of gait and the consequences of choosing different gait strategies and the specific demands of badminton. The aim of this section is to present the metabolic cost of human locomotion, the consequences of choosing alternative gait strategies and the energy demands of badminton.

Section 2.4: Injuries in badminton
This section presents a review of the mechanisms of injury in badminton. The aim of this review is to identify the common injury mechanisms and specify the risk of injury within the game of badminton.
2.2 Assessment of Gait Kinematics and Kinetics

2.2.1 The Gait Cycle

Gait analysis is performed with reference to the gait cycle. The gait cycle represents a single sequence of the function of one limb and can be divided into two periods, stance and swing (Perry, 1992) (see Figure 2.1). According to Perry (1992) the term stance is used to designate the entire period during which the foot is on the ground. It therefore lasts from initial contact (IC) to toe-off (TO). Swing on the other hand describes the period where the limb is not in contact with the ground (from TO - IC). The further subdivisions of the gait cycle, the phases of gait, are illustrated in Table 2.1.

![Figure 2.1 Divisions of the gait cycle. Clear bar represents the duration of stance. Shaded bar is the duration of swing. Limb segments show the onset of stance with initial contact, end of stance by roll-off of toes, and end of swing by foot contact again. Adapted from Perry (1992).](image)

![Table 2.1 Divisions of the gait cycle. Adapted from Perry (1992).](image)
The division of the gait cycle into periods, basic functional tasks and related phases is based on the work by the Rancho Los Amigos gait analysis committee in an attempt to develop a generic terminology (Perry, 1992). The gait cycle therefore starts with IC which initiates the stance phase. In walking this is usually done using the heel (heelstrike, HS), also referred to as a rearfoot strike pattern (RF) (Cavanagh and Lafortune, 1980). In sprinting however, or in the presence of pathological gait patterns, the entire foot or the forefoot can be used to initiate ground contact (Józsa and Kannus, 1997, and Perry, 1992). Loading response, following IC, lasts until contralateral TO and corresponds to the first period of double limb support (both limbs are in contact with the ground) of walking. Midstance ends when the centre of mass of the body is located directly over the weight bearing foot and is followed by terminal stance which ends at contralateral IC. Terminal stance lasts for about 35% of the gait cycle and during this phase the heel rises from the ground. The period between contralateral IC and TO of the limb under investigation is called preswing and therefore represents the second period of double support of the gait cycle. The first phase of swing is termed initial swing and lasts until maximum knee flexion. This is followed by midswing which ends when the leg is at a perpendicular angle to the ground. The final phase of swing is the terminal swing which ends at the second occurrence of IC.

The fundamental difference between the walking and running gait lies in the change from double limb support during stance (where both feet are on the ground at the same time) with no aerial phase in walking, to a single limb support (only one limb supports the body) as well as a distinct aerial phase in running (no limb is in contact with the ground). This is due to fact that in walking the duration of the stance phase is in excess of 50% of the gait cycle while in running TO occurs before 50% of the gait cycle are completed and the stance phase in sprinting can last for as little as 22% of the gait cycle (Mann and Hagy, 1980). In the absence of double support the running gait cycle can be said to consist of three phases during stance: footstrike (or IC), midsupport (or midstance) and takeoff (or TO) (Slocum and James, 1968).
2.2.2 Assessment of Gait Kinematics

Kinematics is the study of bodies in motion without regard to the causes of the motion (Robertson et al., 2004). It is concerned with the temporal characteristics of movement, position or location, displacement (describing what movement has occurred), velocity (how fast something has moved) and acceleration (the rate at which velocity has changed) (Whiting and Zernicke, 1998).

Study of gait kinematics is vital for an understanding of the actions of the limbs in human locomotion. It acts as a research tool in itself and is essential as an intermediate step for kinetic analysis. Kinematic analysis is applied in the field of assessing athletic performance, as well as the assessment of injury risk and the treatment of gait abnormalities.

The assessment of gait kinematics has found an important application in the field of sports assessment. Particularly the relationship between movement, injury and injury prevention has been of interest to sport related research. An example of this is a study by McClay et al. (1994a) who, in an attempt to elucidate the relationship of overuse injury in basketball used high speed film, video and optoelectronic systems for the recording of multiple, sport-specific movement tasks. Similarly, cutting movements, due to their relationship to non-contact injury of the Achilles tendon in sports such as basketball have been the focus of studies by McLean et al. (1999) and McLean et al. (2004b) who used a six camera passive marker motion analysis system. These studies were performed in order to investigate gender related knee kinematics and identify the causes for knee ligament injury risk in women as well as the effect of a defensive opponent on the biomechanics of sidestep cutting in the laboratory. Kinematic analysis has found application in a number of different sport scenarios including pole vaulting (Schade et al., 2004) and tennis (Blackwell and Cole, 1994, Knudson and Blackwell, 1997) and provides useful data for sports such as swimming where force measures cannot be obtained.

These investigations reflect only a small section of the application of kinematics and demonstrate only a few of the techniques that are used for the recording of kinematic data which are summarised in Winter (1990 and 1991). The advantages between
different systems lie in the ease of use and handling as well as cost (Winter, 1991). However, as mentioned above, the lack of information relating to the involvement of the processes which are the ultimate cause of motion means that kinematics are often performed not as a standalone tool but in combination with electromyographic data recordings (Murray et al., 1985, Wank et al., 1998) and force data (Neptune et al., 1999) to obtain a more global picture of gait processes. These will be discussed in the following sections.

2.2.3 Assessment of the Ground Reaction Force

In biomechanics forces are classed as external (acting between the body and the environment) and internal (acting between body parts) (Zatsiorsky, 2002). Newton's 3rd law of motion states that "the mutual actions of two bodies upon each other are always equal, and directed to contrary parts", that is, "to every action there is always opposed an equal reaction" (Newton, 1848a). This applies equally to the human body in motion, where the force the body exerts to the ground causes an opposite force which acts on the body. It is this external force that is referred to as the ground reaction force (GRF).

Measurement of this force, using for example a force plate, allows for an assessment of the acceleration patterns of the centre of gravity and forms an essential part of the description of the mechanics of gait (Munro et al., 1987). These patterns are a reflection of the factors that influence kinematic or kinetic parameters of the lower limbs. Furthermore, modifications of technique are associated with changes in the GRF which has important implications for injury prevention and performance improvement (Vaughan, 1989). The investigation of GRFs has therefore been of interest to a number of researchers because of the postulated relationship of respective impact forces and injury of the body, particularly overuse injuries of the musculoskeletal system (Cavanagh and Lafortune, 1980, Hreljac et al., 2000, James et al., 1978, Nigg et al., 1981, Nigg et al., 1995a).

The study of the external forces that act on the body during gait is therefore fundamental for an understanding and appreciation of the magnitudes and
characteristics of force the body is exposed to as well as the potential risks associated with performing this movement.

2.2.3.1 Forces in Walking and Running

**The Force Pattern**

A number of researchers have investigated the reaction forces of walking and running as well as the effect of a variety of parameters on the shape and magnitude of the force curve. The basic shape of the GRF curve during walking is presented in Figure 2.2. Clearly visible are the two distinct peaks in the vertical force during walking. These peaks are related to the acceptance of the weight of the body at IC, when the downward motion of the body is stopped and a corresponding force is exerted in the opposite direction, and the second is due to the pushing of the limb against the ground to generate propulsive force at TO, as described in Winter (1991).

![Figure 2.2](image)

Figure 2.2 Vertical and horizontal (anterioposterior) ground reaction force components for walking at a natural cadence. Force data is expressed in Newton’s per kilogram body mass (N/kg) and presented for 0-100% of the total stride. Error bars indicate standard deviation from the mean. The data was adapted from Winter (1991).

As mentioned in section 2.3.1, running does not consist of any periods of double limb support and as a consequence the vertical GRF curve does not display the characteristic double peak shape typical of walking (see Figure 2.4). Running is generally performed using a rearfoot (or heelstrike) pattern, touching the ground at IC with the heel first, followed by contact of the forefoot (midfoot and toes) with the ground (Keller et al., 1996). Nevertheless, forefoot strike patterns, are observable, particularly when running at faster speeds (Hawley, 2000, Józsa and Kannus, 1997, Keller et al., 1996). Running with a rearfoot pattern results in the characteristic vertical force curve. There are two peaks, a small initial peak, referred to as the
maximum impact force (Munro et al., 1987), followed by a larger and longer duration force peak, the thrust maximum (Munro et al., 1987). It is the initial impact peak which is regarded as particularly important since it occurs quickly. This is a passive force which means the body cannot react to and accommodate this high frequency impact and runners who experience impact peaks of larger magnitude are more likely to get injured. Nigg et al. (1981) propose that exposure to this force may result in soft tissue and bone injuries through microtrauma. Furthermore, repeated loading of the limb during impacts has been proposed as an injury mechanism of overuse injury of the foot (James et al., 1978, Cavanagh and Lafontune, 1980). This view appears to be supported by a more recent study by Hreljac et al. (2000) who analysed the reaction forces for 2 groups of 20 participants running at 4 ms$^{-1}$. One group contained participants who had never experienced an overuse injury in their running career and the other contained participants who experienced at least one injury that can be attributed to running. The finding of larger forces in the injured group led the author to conclude that the magnitude of the peak vertical impact force is the most important variable that distinguishes between runners with no injuries and those with injuries.

When adopting a forefoot running pattern, the occurrence of the initial impact force peak is reduced and disappears due to the function of the ankle in absorbing the impact shock. Cavanagh and Lafontune (1980) show that force patterns in midfoot
runners are very homogenous while rearfoot runners display more variability. Furthermore, the anteroposterior force component (Fy) in rearfoot strikers exhibits a gradually rising mean curve while midfoot strikers exhibit a double peaked braking phase. Munro et al. (1987) name the initial positive part of the biphasic curve braking since the GRF opposes the forward movement and the second phase propulsion as the force acts in the direction of travel.

**Peak Forces**

The peak forces developed during gait are dependent on technique, speed of locomotion and the environment. The GRFs during walking at different speeds have been investigated by a number of researchers. Winter (1991) (walking at a slow, natural and fast cadence) and Keller et al. (1996) (walking at 1.5 to 3.0ms$^{-1}$) show that the forces during walking increase with speed of locomotion. Peak forces of the vertical component are in the region of 1.1 and 1.5 times body weight (BW) for the vertical force component while forces in the anteroposterior direction were about 0.19BW and -0.22BW for braking and push-off components respectively.

GRFs in running have received considerable interest due to the popularity of running sports and the postulated relationship between repetitive high force exposure and running injuries (Cavanagh and LaFortune, 1980, Hreljac et al., 2000, James et al., 1978, Nigg et al., 1981). Vertical forces, for running at 4 to 4.5ms$^{-1}$, are about twice as large as those in walking (Cavanagh and LaFortune, 1980, Keller et al., 1996). Cavanagh and LaFortune (1980) report a peak vertical (Fz) impact force (for running at 4.5ms$^{-1}$) of approximately 2.2BW and thrust maximum force of about 2.8BW for rearfoot strikers. Midfoot strikers showed a characteristic absence of the initial peak following contact with a mean peak of about 2.7BW which was not significantly different from rearfoot strikers. Forces in the mediolateral component (Fx) were 0.35BW and 0.12BW for midfoot and rearfoot runners respectively. The mean peak to peak amplitude was therefore three times greater in midfoot compared to rearfoot strikers. In the anteroposterior force component (Fy) rearfoot strikers exhibited a positive peak of 0.43BW while midfoot strikers exhibited an average initial peak of 0.45BW after which the curve falls to or below zero before a second positive peak of almost equal magnitude.
In a follow-up study, on the effect of speed on GRF in rearfoot runners, Hamill et al. (1983) attribute increases in vertical and horizontal force due to increasing speed to increases in both stride length and stride frequency, where an increase in stride length places the centre of gravity farther behind the point of ground contact resulting in larger breaking force. Furthermore, Munro et al. (1987), when collecting GRF data for running at self-selected speeds ranging from 2.5 to 5.5ms\(^{-1}\), show a reduction of stance time due to increased speed. Decreasing the speed of heel-toe running from 6ms\(^{-1}\) to 3ms\(^{-1}\) therefore, leads to a decrease in vertical impact and thrust maximum force by 40% and 20% respectively (Nigg, 1988). According to Keller et al. (1996) vertical GRF increased linearly from 1.2 BW to about 2.5BW, regardless of gender, up to running speeds of 6.0 ms\(^{-1}\). Thereafter it remains relatively constant due to the adoption of a forward lean. Furthermore, slow running (jogging), compared to walking causes a significant increase in force, particularly in male participants. This is attributed to the adoption of a higher centre of gravity which increases the downward velocity of the head, arms and trunk and in turn leads to an increase in vertical GRF (Keller et al., 1996).

These findings show that the way a person moves used for locomotion plays an important role in defining force magnitude. Keller et al. (1996) contribute comparatively longer vertical force magnitudes with reduced peak values to forward leaning which lowers the centre of gravity and thereby reduces the downward velocity of the head, arms and trunk at impact. This finding is in line with observations by Bobbert et al. (1991) who observed that running while keeping the centre of the body low in a 'groucho' style produced GRFs which were ~25% lower than normal heel-toe running at the same speeds (3.6–4.2 ms\(^{-1}\)). Furthermore, as mentioned above, Cavanagh and Lafortune (1980) identifies considerable differences in GRF parameters between midfoot and rearfoot strikers.

Finally, the external environment (the surface) and the interface between the human being and the environment (the shoe) play important roles in the control of force magnitudes. McMahon and Greene (1979) developed a model for the investigation of the influence of track compliance on running. Their results indicated that running on hard surfaces can result in initial force spikes up to 5 times BW. Running on softer surfaces on the other hand reduced this spike and athletes reported a subjective
impression of increased running comfort. As mentioned above, it is speculated that the impact force is one of the causes of the many running overuse injuries, as well as the other sports injuries (Nigg, 1988) and a reduction of this force may be beneficial to the athlete. Furthermore, it was found that running surfaces with a specific stiffness can enhance the running speed of the athlete (McMahon and Greene, 1979). Similarly, Nigg (1988) shows that the impact force (the force at IC) is much higher for running on asphalt then on grass or sand. However, the thrust maximum of the ground reaction force remains about the same. Interestingly, Renström and Roux (1988) report that, in tennis, adolescents often sustain more overuse injuries on high friction surfaces (such as hard-courts) than on low friction surfaces (such as clay courts). These findings therefore also relate to shoe design where cushioning, support and friction are stated as important elements of consideration. In combination with the large distances a runner travels on a weekly basis the cumulative effect of force exposure may therefore be the predisposition to overuse injury, particularly when an abnormal running style is used (Cavanagh and Lafortune, 1980).

2.2.3.2 Forces in Atypical Movements

Human gait is not restricted to the use of walking and running but also consists of a number of other movements which expose the body to force. This is of particular relevance in sports where atypical movements, those that are not regularly seen in every day normal adult gait, may be important to the successful performance in the sport. Understanding the requirements of these movements is essential in the determination of the requirements of the sport and subsequent assessment and minimizing of risk factors.

Jumping/Landing

Jumping movements play an important part in many sports and in some respects running can be regarded as a series of repeated jumps and landings. Dufek and Bates (1990) considered the effects of height, distance and technique on the impact force during landings. Their findings are in line with those observed in running where vertical GRF increased with increased height and knee extension at landing while forefoot landings reduce vertical peak force compared to flatfooted landings. This led
them to the suggestion that it may be advantageous for individuals who perform landings repeatedly to adopt a forefoot landing pattern and increased knee flexion in order to reduce force. However, the authors do not make any reference to the other injury mechanisms. Wright et al. (2000), using a simulation model, show that the use of a plantarflexed ankle at ground contact increases the risk of ankle sprain injury.

The findings for GRFs by Dufek and Bates (1990) are supported by Ricard and Veatch (1994) who identified speed and jumping height as important contributors to increased first and second peak impact forces and peak loading rates in aerobic dance. Furthermore, Seegmiller and McCaw (2003) observed an increase in first and second peak vertical impact, particularly in female gymnasts, with increasing height of a vertical drop. However, in the investigation by Ricard and Veatch (1994), time to peak impact force was delayed, peak loading rate was reduced and the high frequency content of the vertical force was reduced, compared to running, which may indicate that the tested movements were less stressful. Since landings in aerobic dance are typically performed using a forefoot approach the stresses the limb is exposed to may be exerted on different anatomical structures, the forefoot, lower leg and knee, compared to typical running (Ricard and Veatch, 1994).

With respect to previous injury of the ankle GRFs in landings are particularly important. Caulfield and Garrett (2004) show that there are significant differences in lateral and anterior force patterns in single leg landings in participants with functional instability of the ankle joint. These force patterns are thought to predispose the participant to repeated injury since they result in significantly increased stress on the ankle joint structures during jump landing.

Jumping parameters therefore affect the force of impact. Jumping technique itself has also been shown to affect the resultant reaction force. Fukashiro et al. (1995) determined that hopping (4.17BW) produces larger vertical reaction forces than squat (2.77BW) or countermovement jumping (2.96BW). While Perttunen et al. (2000) report peak vertical and horizontal impact forces of 15.2BW and 7BW respectively for the step phase of the triple jump.
Cutting
Cutting, an evasive manoeuvre where the athlete suddenly changes from a forward run to move to the side at an angle, is a further movement that is regularly observed in a number of sports and has received attention because of the associated risks of injury to the cruciate ligaments of the knee (Besier et al., 2001b, Mclean et al., 1999). (Houck, 2003)

McLean et al. (2004b) show that sidestep cutting movements result in substantial mediolateral reaction forces, particularly when facing an opposing player. There is an initial peak force short duration peak of 0.98BW and 1.08BW for the non-defensive and plus defensive player situation respectively which is followed by a larger and longer duration peak of 1.04BW and 1.19BW for the non-defensive and plus defensive player situation respectively. (Cowley et al., 2006) furthermore show that differences in reaction forces during cutting movements exist between players of different sports.

Neptune et al (1999) and Dayakidis and Boudolos (2006) report reaction forces for both a cutting movement using a crossover step, where the trailing limb passes behind the leading limb at the point where the player changes direction from a forward run to a cut to the side at an angle, as well as a sidestep-shuffle where the direction of travel reverses at impact of the right limb. Dayakidis and Boudolos (2006) show that the shuffle resulted in the highest mediolateral forces (0.89 and 0.60BW for the first and second peak respectively), compared with 0.61 and 0.54 of BW in the v-cut. These inverting forces, combined with the increased rearfoot supination exhibited during these movements, may be potentially damaging and place the ankle joint at increased risk for sprain. Neptune et al. (1999) furthermore showed that healthy subjects exhibited higher vertical GRF in the crossover cut than in the shuffle which is in line with the findings by Dayakidis and Boudolos (2006) (2.97 and 2.42BW for the first and second peak in the crossover cut compared to 1.87 and 1.47BW in the lateral sidestep). Furthermore, regardless of the attempts to control and standardize technical execution of the maneuvers, high variability was noted between trials amongst participants.
It should be noted that these studies have considered the function of the right limb during changes in direction only. None of these studies have considered the action and contribution of either the right or left limbs to the performance of a lateral movement in one direction. There appears to be no information in the current literature addressing the topic of force exposure during continuous lateral movements.

2.2.4 Assessment of Joint Moments and Powers

Investigations of the moments and powers at the joints of the lower limbs are essential for an understanding of "the cause of the resulting movements", an understanding of "why and how aberrant movements occur" (Vaughan, 1996) and the contribution of different groups of muscles in generating and absorbing energy.

Since direct assessment of the moments and powers in the human is not possible, not least because of the ethical implications, an indirect approach, combining force and kinematic data using inverse dynamics calculations, is used. Inverse dynamics is the specialised branch of mechanics that bridges the areas of kinematics and kinetics. It is the process by which forces and moments of force are indirectly determined from the kinematics and inertial properties of, primarily, moving bodies (Bresler and Frankel, 1950, Robertson et al., 2004). The resultant net joint moments and powers reflect the actions of all the muscles, tendons and ligaments that act on a particular joint.

2.2.4.1 Moments and Powers in Walking and Running

Moment Patterns in Walking

Figure 2.4 displays example data for the extensor and flexor moments at the hip, knee and ankle joints. Winter (1984) describes the general shape and functional significance of the moment patterns in walking. While the moment pattern of the hip and knee tends to be variable there generally is an initial extensor moment at the hip, used for energy absorption during weight bearing, which is followed by a flexor moment which peaks in late stance and helps in changing the direction of movement of the posteriorly moving leg and generate energy for the swing phase.
At the \textbf{knee} there is an initial extensor moment in early stance as the knee flexes to absorb energy during weight bearing. This turns into a flexor moment during mid-stance and an extensor moment in late-stance as the leg extends and swings forward.

The moment pattern of the \textbf{ankle} is less variable than that of the knee or hip. At heel contact there is a small ankle dorsiflexor moment, which is followed by a much larger plantarflexor moment and a quick reduction of the moment to 0 at toe-off. The sole function of the ankle plantarflexors, therefore, is to generate the major energy burst required at push-off. The contribution of the plantarflexor moment appears to be particularly prominent in slow speeds (Riley et al., 2001).

It is thought that the activity of predominantly single joint muscles at the ankle, with the exception of gastrocnemius, is the cause for the smaller variations in moment patterns. In the hip and knee on the other hand there are a larger number of biarticular muscles (see Chapter 2.2.5.3) that contribute to the generation of force. Activation of the hamstrings for example causes extension of the hip and flexion of the knee while rectus femoris performs the opposite function. There is therefore an increased variability for the contribution of different muscles to the total generation of extensor or flexor moments at the hip and knee. These conclusions are supported by Vaughan (1996) who reports very similar moment patterns and similar variability for joint moments in children.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{image.png}
\caption{Joint moments at the hip, knee and ankle for walking at a normal cadence (Winter, 1991). Positive moment values refer to ankle plantarflexor and hip and knee extensor moments. Negative values refer to dorsiflexor and flexor moments. Units are 0-100\% stride and Newton meters per kg body mass (Nm/kg).}
\end{figure}
Moment Patterns in Running

The joint moment patterns during stance in level running are well documented, with dominant extensor, flexor and plantarflexor moments at the knee and ankle accompanied by extensor hip moments during the impact phase and flexor hip moments during the later push-off (Winter, 1983, Jacobs et al., 1993, Derrick et al., 1998, Swanson and Caldwell, 2000). Winter (1983) summarise the moments during the running gait. The total moment of force pattern of the lower limb during stance is primarily extensor and flexor during swing with considerably less variability in moment patterns during running compared to natural walking (Winter, 1983).

The hip moment pattern is very similar between walking and running with hip extensor activity in early stance and late swing. The hip flexors can be seen to dominate from mid-stance to mid-swing, decelerating the backward rotating thigh and reversing the thigh’s direction to drive it forward into swing. The extensor moment in late-swing on the other hand decelerates the thigh before heel contact (Winter, 1983).

At the knee joint, stance phase is dominated by a large extensor moment during which the is knee extensor muscles act eccentrically in the first half of stance and concentrically in the second half of stance. They therefore perform a vital function in absorbing the impact shock at IC. The contribution of the knee in shock absorption is however reduced when using a forefoot landing approach, due to the action of the ankle joint in absorbing the shock impulse (Mann and Hagy, 1980).

The ankle joint is completely dominated by a large plantarflexor moment during stance.

Power Pattern in Walking

The patterns of hip, knee and ankle joint power in walking are summarised by Winter (1984, 1990 and 1991) (see Figure 2.5). At the hip there are 3 major power phases during the gait cycle. The first phase at the hip is one of small positive power which is not always present and corresponds to the concentric activity of the hip extensor muscles during loading response. This is followed by a longer period of negative power during mid-stance where the flexor muscles of the hip act eccentrically. Finally
there is a further phase of positive power during pre- and initial swing where the hip flexors act concentrically. This represents the gait cycles second largest contributing power phase for the purpose of generating propulsive power.

At the knee there are a total of 4 power phases. The first phase is one of negative power during the loading response where the knee extensors act eccentrically. This is followed by positive power during mid-stance where the knee extensors act concentrically. After a period of relative inactivity between mid-stance and pre-swing there is a period of negative power with eccentric activity of the knee extensors. This is followed by a further period of negative power during terminal swing during which the knee flexors absorb power.

At the ankle joint there are 2 distinct power phases. Firstly a period of negative power during mid- and terminal stance during which the ankle flexors act eccentrically. This is followed by large positive power as the plantarflexors act concentrically. This indicates that the ankle is primarily involved in energy generation.

Power Patterns in Running
In running both hip flexors and extensors are responsible for an increase in power generation in running and sprinting (Winter, 1983). The hip extensors therefore generate power in the first half of stance while the hip flexors become maximally active in late stance in order to decelerate the backward rotating thigh and initiate
swing. According to Winter (1983) the hip did not display large power levels and the power pattern was inconsistent, which he attributed to the function of the hip flexors and extensors in driving the lower limbs as well as stabilising the upper body.

At the knee joint, the knee extensor muscles act eccentrically in the first half of stance, absorbing power, and concentrically in the second half of stance, generating power. The knee performs an important shock absorption function in rearfoot runners. During swing the muscles of the knee absorb power in order to control the movement of the leg and foot and the negative power peak in late swing indicates the function of the knee flexors in decelerating the leg and foot before IC.

The power pattern at the ankle joint is essential the same in running as in walking with 2 distinct phases of negative large positive power. According to Winter (1983) the ankle muscles generated 2.9 times as much energy as they absorbed over an entire stride, while the knee muscles absorbed 3.6 times as much energy as they generated. Furthermore, Johnson and Buckley (2001) show that in sprinting the timing in the generation of peak extensor power occurs in a proximal to distal sequence involving the hip and knee extensors and then ankle plantarflexors to aid in producing maximal peak power.

**Peak Moments and Powers**

One of the most important factors involved in the magnitude of the moments and powers at the joints of the lower limbs is the speed of locomotion. Peak joint moments in walking for hip and knee extension and ankle plantarflexion increase with increasing speed of locomotion (Winter, 1984, and Riley et al., 2001).

Speed also affects the mean powers at the joints of the lower limb. For walking, Wang et al. (1996) show that, while the mean powers for the ankle and hip increase with speed, the mean power at the knee decreases. As a result, the sum of mean power for all joints remained almost unchanged, at speeds of less than 2 ms⁻¹, but at faster speeds, the sum increased very quickly. The optimal walking speed that minimises the necessary power output was therefore concluded to be about 2ms⁻¹.
Riley et al. (2001) argue that ankle plantarflexor power has a limited role in propelling the body in late stance. In their view, adaptations to changing speed requirements occur primarily at the hip flexors and extensors, with the ankle acting primarily to supporting the upper body.

As in walking, speed of locomotion affects the joint powers in running. Belli et al. (2002) identified that peak joint power increased in every investigated joint with increasing running speed. The highest changes were observed in the hip joint. The results may suggest that the role of the ankle and knee extensors is to create high joint stiffness before and during the contact phase, while the hip extensors are the prime forward movers of the body with increasing running speed. In the contact phase, the active role of the hamstring muscles is essential for producing significantly increased hip power with increasing running speed.

A further potential contributor to differences in joint kinetics is gender. Kerrigan et al. (1998) reported that women exhibited significantly greater peak hip flexion and negative work during walking, but these differences were not evident during running. Ferber et al. (2003) also report no differences in sagittal plane hip and knee kinematics or kinetics between male and female recreational runners. In the frontal plane on the other hand, Ferber et al. (2003) report significantly greater peak hip adduction angle and hip frontal plane negative work while running in women compared to men. Furthermore, there was significantly more hip frontal plane negative work in women. Since peak joint moments were similar between genders the greater frontal plane negative work exhibited by the female runners was attributed to a greater hip adduction angular velocity. These data suggest that as a result of the greater hip adduction angle and velocity in women, greater eccentric demands were placed on the hip abductors compared to men.

2.2.4.2 Moments and Powers in Atypical Movements

Jumping

The mechanics of jumping have been investigated by a number of researchers. The magnitude of joint moments and powers in jumps are dependent on a number of
factors. Vanezis and Lees (2005) show that muscle strength and rate of strength development in all lower limb joints are the main factors that cause increases in joint moment and powers. Technique was seen as less important since differences were less noticeably between experiences and inexperienced jumpers. Performing a jump at speed, as is the case in the long jump or vertical jump in basketball, furthermore does not appear to influence joint kinetics. This study shows the importance of the ankle joint, particularly in the vertical jump, and the hip joint, particularly in the long jump, in the generation and absorption of energy (Stefanyshyn and Nigg, 1998). Interestingly, the variability in joint kinematics and kinetics appears to much lower in maximal jumping than walking or running (Yoon and Challis, 2005). Furthermore, findings by Jacobs et al. (1993) show that the net plantar flexion moment during running was 158% and 127% higher than the peak values reached in maximal jump and sprint push-offs respectively. This larger mechanical output in running was ascribed to the activity of the stretch-shortening cycle (described in Chapter 2.2.5.2) in running and differences in muscle stimulation between running and sprinting. Furthermore, gender appears to influence the landing strategy, with women choosing to land with a more erect posture than men, thereby maximising energy absorption by the proximal joints (Decker et al., 2003).

Cutting
Cutting movements are important in a number of sports, such as basketball and football, where they are used to evade an opposing player (Mclean et al., 1999). However, these movements are also associated with non-contact injury of the anterior cruciate ligament (ACL) (Besier et al., 2001b, Mclean et al., 1999). The observation that particularly female athletes are at risk of ACL injury (Arendt and Dick, 1995) has resulted in a number of studies to investigate the joint moments and powers that are acting on the knee joint. A study by Besier et al. (2001b) determined that the external flexion/extension loads on the knee joint were similar to running. However, varus/valgus (VV) and internal/external (IE) rotation moments on the knee during side and cross-step cutting manoeuvres were considerably larger. This was attributed to a task-specific combination of flexion, VV, and IE rotations at the knee joint and therefore increased ligament loading. Following the determination of the importance of VV and IE kinetic pattern to the loading of the knee ligaments, a number of contribution factors were determined. Preparation ahead of a cutting task was
demonstrated to be an important aspect, where unplanned cutting resulted in increased VV and IE rotation moment magnitude compared to planned cutting (Besier et al., 2001a). Gender was considered to be a contributing factor in ACL injury mechanisms, however only within individuals rather than as a general overall, with some female participants showing similar kinetic patterns to men (Sigward and Powers, 2006). Similarly Pollard et al. (2004) could not identify differences in moments during a randomly cued cutting movement between male and female soccer players. Therefore, while these findings demonstrate the importance of VV moments to knee joint loading during cutting tasks they also highlight the difficulties in determining the contributing factors that lead to higher non-contact ACL injury risk within females in the laboratory setting.

2.2.5 Energy Saving Mechanisms in Human Gait

2.2.5.1 Mechanical models of gait

The information presented above and the data following below display the multitude of contributing and interacting mechanisms that are involved in bipedal locomotion. However, despite the complexity of these interactions, it has been possible to describe the walking and running gaits through the use of two paradigms, based on mechanical energy. Walking and running effectively consists of phases of transfer between potential energy (PE, m* g*h) and kinetic energy (KE, \( \frac{1}{2} m v^2 \)) within each stride, where m is the mass of the body, g is the acceleration due to gravity, h is displacement of the body centre of mass (CoM), and v is the speed of the body centre of mass (Cavagna et al., 1976, Cavagna, 1978). Potential energy therefore is the energy used during the lifting and lowering of the body, while kinetic energy is due to the acceleration and deceleration of the body.

In walking the interaction of PE and KE is such that the two factors exchange out of phase, which means that the overall change in total mechanical energy (the sum of PE and KE) is relatively small. The model describing the relationship of PE and KE in walking is referred to as the pendulum or rolling egg paradigm (Cavagna et al., 1963, Ralston and Lukin, 1969, Cavagna et al., 1976). When looking at the gait kinematics
it can be seen that at IC the speed of the CoM of the body is at its fastest and the position of the CoM is at its lowest point. This is followed by midstance where the CoM is maximally vertically displaced and at its lowest speed. This means the potential gravitational energy at IC is translated into KE. In an ideal situation this would mean that no energy is needed for walking. However, the human body in motion does not represent a perfect system and deviations in the CoM during gait, resembling and inverted pendulum (Cavagna and Margaria, 1966), and other factors such as the speed of locomotion (Cavagna et al., 1976) and stride length (Minetti et al., 1995) influence the degree of energy return. As a consequence, the model accounts for up to 70% of the total energy changes taking place within a stride, leaving about 30% of the energy for locomotion to be supplied by muscles (Cavagna et al., 1977). This estimate appears high and a more current estimate for energy recovery is somewhat smaller with values up to 60% (Minetti et al., 1995). However, the model demonstrates the importance of the exchange between PE and KE in conserving energy during locomotion in walking.

In running the relationship between PE and KE is quite different from that in walking. Here, PE and KE are not offset but occur in phase (Cavagna et al., 1977). The CoM is at its highest vertical position and fastest speed at the same time. Therefore, there is no exchange between the two, as seen in walking, and the movement is far more dependent on the return of elastic energy through tendon recoil in order to conserve energy. The model for running is referred to as the bouncing ball or pogo-stick paradigm (Cavagna et al., 1971, Cavagna et al., 1964).

There appears to be some crossover between the models, with energy storage and return in walking, with contributions by the Achilles tendon (Fukunaga et al., 2001) and the elastic elements in the foot (Ker et al., 1987), however, most of the mechanical processes of the walking movement appear to be predominantly related to the pendulum model.
2.2.5.2 Elastic Energy Storage and Power Amplification

Work has been performed on skeletal muscles of a number of different animals using in-vitro work-loop experiments (Josephson, 1985). This type of experiment consists of stretching a muscle at a given rate and stimulating the muscle at a specified point in time in the stretch cycle to obtain muscular contraction. The results from these type of experiments show that skeletal muscle produce the largest power output at intermediate cycles of stretching and shortening with maximal power output of about 200 W/kg muscle (Askew and Marsh, 1997, Askew and Marsh, 1998, Bennett, 1985, James et al., 1995, James et al., 1996, James et al., 2004, Swoap et al., 1993). Figure 2.10 summarises the findings of maximum power output for a number of different investigations. The relationship between these cyclic activations and the production of muscular power is not as straight forward in gait, since stretching of the musculature immediately before the shortening of the same muscle results in enhanced power production of the muscle through a process termed “stretch activation” (Alexander, 2003b p.28).

![Figure 2.6 In-vitro muscle powers for a selection of skeletal muscles in the mouse, rabbit and reptiles. Data is presented for the mouse soleus and extensor digitorum longus (EDL) muscles, latissimus dorsi (LD) of the rabbit and reptilian skeletal muscle. All power values are expressed as Watts per kg muscle.](image)

Power amplification in movement is therefore thought to be due to a number of contributing factors. Stretch activation (or prestretch) appears to affect the musculature primarily through two processes. Firstly, potentiation of muscle fibres, described by Hill (1970), has been shown to cause increases in muscle force.
(Cavagna, 1978, Edman et al., 1978). This process is dependent on the speed of stretch (Edman et al., 1978), with larger force resulting from faster stretching, and the duration of the delay between stretch and consecutive contraction (Edman et al., 1978), with lower force with increasing duration. Secondly, the development of a higher active state before the beginning of concentric muscular contraction may contribute, by reducing the time delay between muscle activation and maximum force production, which may take 300-500 ms (Bobbert and Van Ingen Schenau, 1990, Jaric et al., 1985) and therefore increasing the time the muscle can effectively use to develop force. Efficient use of the stretch-shortening cycle therefore causes larger moments at the calf-muscles during running, compared to jumping or sprint push-offs (Jacobs et al., 1993)

As stated in the previous section on mechanical gait model, running depends on the return of elastic energy, stored within tendons, to increase the efficiency of the movement. These series elastic elements therefore return the energy stored during stretching during the shortening phase. There appears to be some controversy about the contribution of the elastic element to the maximum force production with suggestions energy stored in the series elastic element may contribute to power amplification through storage of energy at low rates and their subsequent release at high rates (Schenau et al., 1997). Similarly Bobbert (2001) contributed a higher power output at the ankles during jumping to the contribution of the series elastic element of the triceps surae. The contributory role of elastic energy to power amplification is however questioned after a countermovement (Schenau et al., 1997). However, it is recognised that the storage and reutilisation of elastic energy contributes to gait efficiency as well as other activities, such as throwing, jumping (Van Ingen Schenau et al., 1997).

2.2.5.3 Biarticular Muscles

The skeletal musculature of the lower limbs can be broadly divided into mono- and biarticular muscle, based on the number of joints they affect. The distinguishing feature of biarticular (or two-joint) muscles is that they affect two different joints, such as rectus femoris which originates on the ilium (anterior inferior iliac spine and
iliole superior to the acetabulum) and inserts on the tibia (insertion at the base of the patella and via the patella ligament on the tibial tuberosity) (Agur and Lee, 1999). This means that contraction of a biarticular muscle, with only one head, non-selectively influences both joints equally, unless it is assisted by other muscles that act to stabilise one of the joints (Rasch, 1989).

Biarticular muscles are thought to contribute to the energy efficiency of human locomotion (Elftman, 1966). However, efficiency may be decreased if the required actions in a particular movement are not those that the two-joint muscle is capable of producing. This is known as Lombard's Paradox: The activity of a biarticular muscle when the required moment at one of the joints is in the opposite direction to that caused by the muscle (Gregor et al., 1985, Andrews, 1987, and Rasch, 1989). This has important implications on metabolic energy.

In locomotion the action of a biarticular muscle might be advantageous as it causes an energy exchange between segments, something a single uniarticular muscle cannot achieve (Zajac et al., 2002). Rasch (1989) shows that, in running, the transfer of energy between body segments leads to an increase in efficiency, over the action of one-joint muscles alone. In the second half of swing the hip and knee are both extending while the hamstrings are contracting. The hamstrings produce an extensor moment at the hip (positive work) and simultaneously a flexor moment at the knee (negative work). This dual role of the hamstrings means that the hamstrings absorb energy at the knee and generate energy at the hip which reduces the mechanical cost. As a consequence of these actions, the function of biarticular muscles have been referred to as 'energy straps', transferring energy between limb segments, with tendinous action, belt-like action, or pulley action because they cannot cause a full range of motion simultaneously at each joint on which they act (Rasch, 1989). However, for since a biarticular muscle contracts without being able to direct its force to a particular joint, one of the joints has to be fixed through the action of other muscle groups in order for the biarticular muscle to function as an 'energy strap' (Rasch, 1989).

Therefore, there are essentially two important contributions of biarticular muscles to energy transfer. Firstly, they are involved in power transfer from proximal (the hip) to
distal muscles (the knee) (Jacobs et al., 1996). This means that, for example at push-off, rectus femoris and gastrocnemius are able to transfer mechanical energy from the joints of the leg to the distal ones to help extend the distal joints, thereby increasing the power output at the distal joints, in line with the observations by Johnson and Buckley (2001) for the timing of power development at the lower limbs in sprinting. The work done by the proximal single-joint muscles is transferred to the distal joints and the power transfer aids in high power output at the distal joints (Prilutsky et al., 1996, Jacobs et al., 1996). Secondly, they are involved in power transfer from distal to proximal joints. This means that at IC and the first half of stance, the hamstrings are able to help dissipate the mechanical the mechanical energy of the body (Novacheck, 1998). These functions allow for energy savings since single joint muscles would have to actively lengthen (absorb power) at the same time as others produce power, while a two-joint muscles can transfer the power without wasting energy through unnecessary eccentric contractions (Kuo, 2001).

2.2.6 Muscular Activity during Gait

Electromyographic (EMG) data allows direct access to the activities of individual muscles and the physiological processes of muscle activations by recording the motor unit action potential produced at the activation of a motor unit (Winter, 1991).

The use of EMG data in the analysis of movement concerns the areas of muscle activation timings, EMG as an indicator of force production; and effects of fatigue (Bonato, 2001). The signal can be recorded using either surface electrodes, placed on the surface of the skin, or an indwelling electrode, which penetrates the muscle to record to activity. Therefore, surface electrode are non-invasive and limited in their usefulness to the assessment of superficial muscles. Indwelling electrodes on the other hand allow for the assessment of deep muscles.

There are a number of methods which are used in order to interpret the EMG signal. The raw data can be used (Mann and Hagy, 1980), ensemble averages can be created (Gottlieb and Agarwal, 1970) and normalisation processes applied (Eberhart et al., 1954), using a number of different methods for the normalising process (Burden et al.,
Complications in the interpretation of the EMG signal arises from the effects of cross-talk which may be difficult to resolve since they do not appear to be removable post-capture through filtering techniques (Dimitrova et al., 2002). Furthermore, the application of EMG to dynamic situations such as gait is complicated by a number of variable including subcutaneous fat, electrode location, and changes in electrode location due to changes in joint angles (Farina et al., 2001).

Again, the use of EMG data as a standalone application limits its usefulness compared to the use in combination with other measures such as joint kinetics. This allows for an assessment of the contribution of a particular muscle to a joint movement which of particular interest in cases of co-contraction and two-joint muscles (Gregor et al., 1985) discussed in Chapter 2.2.5.3. Furthermore, the use in combination with kinematic data during gait (Mann and Hagy, 1980, Murray et al., 1985, Neptune et al., 1999, Wank et al., 1998) allows for the assessment of the functional significance of the electrical activity of the muscle.

2.2.7 Discussion

The aim of this section was to present the current knowledge of kinematics and kinetics resulting from bipedal locomotion and the related topic of injury prevention. In doing so the fundamentals mechanics of human walking and running have been reviewed and a baseline for comparison of atypical gaits has been created.

Analysis of the body in time and space through various methods represents an imminently useful method for an external observer to assess a large number of aspects associated with locomotion. While this information contain very useful information about aspects of speed, accelerations and displacements, it contains no information on the processes, the mechanical forces that are exerted by the muscle or the force the body is exposed to as a result of motion.
The ground reaction forces in walking and running differ both in the shape of the force curve and the magnitude of force. However, a common set of variables have been shown to influence the magnitude of the reaction force, including body mass, speed of locomotion, technique and a number of external factors. These factors also apply to atypical gaits which have been shown to place distinct forces on the body. With regard to injury of the musculoskeletal system these forces have both positive and negative effects on the body. In one respect, force exposure can have strengthening effects in the skeletal system on the other hand repetitive force exposure has been linked to a number of overuse injuries of the lower limb. This is thought to be due, in particular, to the influence of high frequency force impulses. Lateral movements expose the limb to potentially damaging forces.

Joint moments and powers reflect the contribution of the muscular system to the generation of movement through the activity of the musculature at the joints. It is clear that hip, knee and ankle joints perform distinct and coordinated functions in walking and running. This coordinated activity, together with the specialised properties of biarticular muscles and the utilisation of the stretch-shortening properties of the muscular system allows for efficient and powerful locomotion.
2.3 The Metabolic Energy Requirements of Gait

Three questions that biomechanics is trying to answer are: how does one move at speed, at the lowest energetic cost and without excessively stressing any muscle groups? (Saziorski et al., 1987). Since "mechanical work does not represent the physiological work of the muscles" (Chen et al., 1997) it is necessary to use an alternative approach in order to determine the physiological requirements of locomotion. Respiratory gas analysis, in combination with other physiological measures, is a crucial tool for the assessment of the physiological consequences of exercise and the requirement of exercise on the body.

2.3.1 Metabolic cost of Walking and Running

The metabolic energy expenditure of gait can be expressed as the gross or net power, where the net power is the result of subtracting the energy requirements at rest from the gross metabolic requirements (Saziorski et al., 1987). A number of authors have shown that there is a linear relationship between the energy expenditure of locomotion (including walking, running and cycling) and body mass (Mahadeva et al., 1953, Malhotra et al., 1962, Adams, 1967, Wyndham et al., 1971, and Van Der Walt and Wyndham, 1973).

2.3.1.1 Effect of Speed of Locomotion

A substantial number of researchers have investigated and reviewed the metabolic energy requirements of human locomotion at different speeds (Margaria et al., 1963, Van Der Walt and Wyndham, 1973, Saziorski et al., 1987, Alexander, 1992, Saibene and Minetti, 2003). In walking there is a non-linear increase in metabolic power with values in the region of 2.17 to 12.6Wkg\(^{-1}\) for speeds of 0.3 to 2.36ms\(^{-1}\). The metabolic power of running on the other hand assumes a linear relationship with speed of locomotion with values of 9.5 to 24.8Wkg\(^{-1}\) at speeds of 2.17 to 6ms\(^{-1}\). Data from Kyröläinen et al. (2001) support these findings and the view that the causes for the
observed increase can be explained by the increase in the activity of the working muscles with increasing speed.

When viewed in relation to the distance travelled, walking and running can be seen to behave in a very distinct manner with increasing speed of locomotion. The economy of travel of walking assumes a U-shape with an optimal speed at about 1.08 to 1.75 ms\(^{-1}\) for men and about 1.0 ms\(^{-1}\) for women (Saziorski et al., 1987). Griffin et al. (2003) show that the cost of leg swing is relatively small and a predominantly passive process as suggested by the pendulum paradigm (Cavagna et al., 1963, Ralston and Lukin, 1969, Cavagna et al., 1976). The active muscle volume required to generate force on the ground and the rate of generating this force accounted for >85% of the increase in net metabolic rate across moderate speeds which means that the metabolic cost of walking is largely due to the cost of generating muscular force during stance (Griffin et al., 2003).

In contrast to walking, running displays a near constant metabolic cost, in relation to the distance travelled, for increasing running speeds. These observations are supported by Harris et al. (2003) who report no change in mean metabolic task cost, whereas mechanical task cost decreased significantly with running speed. They conclude that the consistency of cost to travel for a given distance appears to be caused by kinematic and kinetic alterations in running patterns with increasing running speed. Furthermore, Kram and Taylor (1990), in an investigation of oxygen consumption across animals, determined that the independence of energy cost of running and speed of locomotion can be explained primarily by the cost of supporting an animal's weight and the time course of generating this force. McNeill and Ker (1990) explain that the proportional relationship between the energy expenditure across animals stems from the time that the feet spend in contact with the ground and the mechanical advantages of the limb muscles. Chang and Kram (1999) contributed to these observations by investigated the role of generating horizontal forces in the generation of metabolic cost during running in humans. By applying an external aiding and impeding force to the hip while running, they observed that aiding and impeding forces caused a reduction and an increase in VO\(_2\) respectively. The generation of horizontal propulsive forces are therefore concluded to contribute more than one third of the total metabolic cost of normal running.
There is some argument about why humans switch between walking and running gaits. Investigations by Hreljac (1993) and Tseh et al. (2002) suggest that gait transition does not take place in order to minimize metabolic energy consumption since energetically optimal transition speeds were faster than preferred transition speeds and that the rating of perceived exhaustion for walking at preferred transition speeds was higher than for running.

Martin et al. (1993) show that the quantification of the causes for interindividual differences in walking and running economy is difficult and that there is a poor relationship between kinematic, kinetic and metabolic variables when attempting to identify possible relationships for walking and running. However, there are a number of factors that are recognised as influencing parameters in determining the metabolic cost of locomotion:

2.3.1.2 Effect of Technique, Skill and the External Environment

The technique used for locomotion and the level of skill of the performer have been identified as important contributors. In his book of the biomechanical foundations of endurance, Saziorski (1987) presents examples of differences in metabolic power of 18.9% to 42.3% between Russian national team cross-country skiers and members of lower classed performance teams. Similarly, oxygen consumption in swimming and speed-skating (at controlled speed) was lower for, so-called, champions of their sport, compared to athletes classes as three performance group below the champions (42.9% and 11.8% for gross metabolic power in swimming and speed-skating respectively). Similar findings were furthermore observed for running at 5ms⁻¹ where champion runners and beginners differed by 30.7%. Unfortunately, the original investigations by Michajlov were not available in a language other than Russian. However, the findings appear to be similar to the observations by Miyashita (1978) who observed differences in the vertical displacement of the CoM and stride length between good and poor runners. Good runners reduced the vertical displacement and used longer strides than poor runners. The overall effect therefore, was that good runners reduced the total work performed in a long distance run. These positive effects of skill are
support by Margaria et al. (1963), as well as Kyröläinen et al. (2001), who observed that runners who were more efficient than others were so across speeds. Again, running technique appears to be important but no exclusive biomechanical parameters were identified to explain the differences in running economy. In analysing the energetics of skipping, a gait primarily used by children, Minetti (1998) shows that the metabolic demands of gait are dependent on the vertical displacement of the body centre of mass, where skipping results in much higher metabolic demands than walking or running due to a much larger vertical displacement of the centre of mass. Furthermore, walking with bent knees has been shown to cause increases in the cost of walking (Waters, 1992).

In addition to skill and technique the external environment has an important impact on the metabolic demands of locomotion. Air resistance is an important factor, with the metabolic cost of overcoming wind resistance in track running contributing 8% of the total energy cost at 21.5 kmh⁻¹ and 16% for sprinting 100m in 10 seconds (Pugh, 1970). A number of other studies have focused on the differences in treadmill and track running. This is of particular importance due to the common use of treadmills in the assessment of human locomotion in the laboratory. Ralston (1960) and McMiken and Daniels (1976) report no significant differences in the energy expenditure of walking and running on a treadmill or track, while the external conditions are comparable, respectively. Maksud et al. (1971) report the same findings for running at 7mph⁻¹. However, at 10 and 12 mph⁻¹ oxygen uptake was higher in track running which was attributed to the effect of air resistance at the higher speeds and more vigorous arm movements used in track running. Running surface stiffness has been shown to affect running economy, where a 12.5 fold decrease in surface stiffness caused a 12% decrease in the participants' metabolic rate and a 29% increase in their leg stillness while their support mechanism remained relatively unchanged. Increased energy rebound from the surface therefore contributes to enhanced running economy (Kerdok et al., 2002). This has implications on treadmill running.
2.3.2 Energetics of badminton

The heart rates in a number of racquet sports are at the extreme end of the age related scale for very long periods of time, however this is especially true for high level badminton where "the highest heart rate values" are recorded (Lees 2003). Neurophysiological factors are suggested as causes for the high average and maximum heart rates with the release of stress hormones during competitive badminton contributing to the acceleration of heart rates above the level "provoked by the actual effort." (Cabello Manrique and Gonzalez-Badillo, 2003). However no proof is provided and the interaction between stress and physical activity seems to be true not only for badminton but most sports. A contribution of isometric muscle activity when the player is stationary, getting ready for the next movement, has also been proposed as a contributing factor for high heart rates in the game (Reilly, 1990). Monitoring the aerobic requirements of the game, however, is complicated by the complexity and temporal patterns of on-court movement. This means that training methods are often the focus point of research since they are more easily controllable.

Badminton-specific training, including practice and simulated competitive games display similar metabolic demands. Hughes (1994) show that practice games and simulated competitive games cause high heart rates (HR) (>80% max HR for >85% of the time; and >80% max for 96% of the playing time, respectively). These observations are supported by Dias et al. (1994), (Majumdar et al., 1997) and Cabello and Gonzalez-Badillo (2003). In regard to competitive match play however, there still seems to be a gap in our knowledge as to the important performance factors in badminton that, when optimised, lead to improved playing levels. In order to optimise effective training programs Cabello and Gonzalez-Badillo (2003).

2.3.3 Discussion

Walking and running are associated with distinct physiological requirements which are reflected in the shape of the metabolic power with increasing speed and the metabolic cost per unit distance travelled. Furthermore, deviations from these typical movements, without the assistance of man-made aids, often result in increased metabolic demands.
Badminton is recognised as a sport that induces extreme physiological responses in the athlete, particularly when performed at the advanced level. Since it is thought that the sport takes advantage of a number of atypical gaits, it may be assumed that these gaits are metabolically more demanding and their repeated use in the game may contribute to the high physiological demands of the sport. Intense workouts can interfere with coordination, which is very important in sports requiring high technical skill such as badminton (Majumdar et al., 1997). Therefore, if footwork can be regarded as a high technical skill as well then the high metabolic demands of the sport must be recognised as an interfering influence with this skill as well.
2.4 Biomechanical Principles of Injury

2.4.1 Injury of the Musculo-Skeletal System

According to Nigg, B.M. (1985) loading of the musculoskeletal system can have positive and negative effects on the body. The biopositive effects result in the strengthening of the muscular and skeletal tissues through moderate loading. The bionegative effects result in the weakening of the muscular and skeletal tissues through either insufficient loading, causing weakening of the muscles and bones, or exposure to excessive loads that cause long or short term injury of the tissues depending on the magnitude and rate of application of the load. Therefore, an injury occurs when the load applied to a tissue exceeds its failure tolerance. It is one of the aims of biomechanics research to identify the causes for injury thereby allowing for the development of preventive measures to reduce their occurrence, insure the long-term health of the participant and reduce the financial costs resulting from injury (Hawkins and Metheny, 2001).

Injuries of the musculoskeletal system can be classified as acute or overuse. According to Ardendt and Griffin (2000) an acute injury occurs due to a single episode of force which can be external, such as a blow to a limb, or internal, such as a non-contact rotational injury to a limb. These injuries are commonly associated with the "immediate onset of pain and dysfunction". An overuse injury on the other hand is characterised by the absence of a specific injury. It is thought to be a repetitive trauma resulting in macroscopic or microscopic damage to an area (Arendt and Griffin, 2000).

Both acute and overuse injuries are common in many sports. Acute tears of the anterior cruciate ligament in indoor ball games such as handball and basketball (Bencke et al., 2000, Simonsen et al., 2000, Mclean et al., 2004a) as well as a variety of other team and individual sports such as gymnastics, football, lacrosse and field hockey (Hutchinson and Ireland, 1995). Furthermore, overuse injuries such as tendonitis, sprains, strains and stress fractures are common in many sports (Nigg et al., 1995a, Keller et al., 1996, Kader et al., 2002, Kvist, 1994, Ulreich et al., 2002).
While the causes for acute injury are more readily definable, a stress in excess of the compensating capacity of a structure is applied, the causes for overuse injury appear to be multi-factorial and more difficult to define.

Overuse injuries are due to inadequate rest and recovery of a structure following the application of stress to this structure (Rolf, 1995, Subotnick, 1985). Whereas a structure would tend to increase in strength following the application of stress (Maganaris et al., 2004, Grimston and Zernicke, 1993, Fahlström et al., 2003), a reduction in recovery time and excessive use, as occurs for example when changing training schedules, may cause a weakening of the structure and ultimately to the development of an overuse injury (Kvist, 1994). This idea is supported by James et al. (1978) and Cavanagh and Lafortune (1980) who propose that high mileage and consequently repeated impacts may be the underlying cause of injury in running.

As yet the exact causes for overuse injury remain unknown, however, the main proposed injury factors are placed into three general categories: training, anatomical, and biomechanical factors (Hreljac, 2004). Training variables, as already mentioned, include running frequency and distance as well as duration and speed. However, it appears that there is no clear cut relationship between training method and injury occurrence (Hreljac, 2004) but rather a list of common training errors that may cause injury (Kvist 1994). Anatomical variables have been considered in search for injury causes but the literature appears to contain conflicting information with studies in support of anatomical causes and other studies finding no relationship (Hreljac, 2004). Further contributing factors to injury that are proposed in the literature include anatomical misalignment and muscular disproportions (Grace et al., 1984, Knapik et al., 1991, Fowler and Reilly, 1993, Mikkelsen, 1979), which may potentially alter the normal direction in which a tendon exerts force and thereby expose the athletes to an increased likelihood of overuse injury (Hess et al., 1989, Witvrouw et al., 2000, Kannus, 1992), insufficient strength and flexibility (Witvrouw et al., 2004), previous injury (Dauty et al., 2003), as well as age and gender (Kannus et al., 1989, Jarvinen, 1992).

As a consequence of the lack of a clear understanding of the cause of overuse injury based on training and anatomical variables, biomechanical factors have received a lot
of attention in attempting to define the boundary conditions for injury development. As stated previously, force exposure is deemed an essential aspect of injury development. Scott and Winter (1990) identify joint moments and forces as important factors in assessing injury risk, as stated in Chapter 2.2.4. Other biomechanical variables that may cause or contribute to overuse injury may be tissue properties (Hawkins and Metheny, 2001) such as the tensile strength of tendons (Kader et al., 2002), the response of tendons to repeated loading or disuse and the effects of tissue deformation such as heat damage to tendinous tissue (Maganaris et al., 2004). Since overuse injury of the lower limbs appears to be closely related to the forces and loads imposed on the limbs during locomotion it is vital to quantify the biomechanical demands imposed on the body. This provides an objective background for the determination of the cause and effect relationship between movement specific loads and the sport-specific demands on the body and furthermore aid in our understanding of the causes of injury in the sport.

2.4.2 Injuries in Badminton

Badminton is classed as a low risk sport since the risk of serious injury is comparatively low (see Table 2.2). However, in terms of the number of injuries per year or the number of injuries for a 1000 hour exposure time badminton can be seen to be similar to tennis, ice hockey and volleyball.

| Table 2.2 Sports injury incidence in comparable studies (Jørgensen and Holmich, 1994) |
|----------------------------------------|-----------------|-----------------|-----------------|
| Sport       | Injuries year⁻¹ | Incidence (injuries 1000h⁻¹) | Serious injuries 100 participants⁻¹ |
| Badminton  | 0.85            | 2.9              | 0.4             |
| Tennis      | 0.56            | 2.8              | 0.7             |
| Soccer      | 1.36            | 4.1              | 3.9             |
| Handball    | 1.11            | 8.3              | 2.9             |
| Ice Hockey  | 0.9             | 4.7              | -               |
| Volleyball  | 0.75            | 3.1              | 1.9             |

What sets badminton apart from other sports is the apparent difference in the nature of the injury. Caine et al. (1996) show that while other racket sports display a high percentage of acute trauma about 3/4 of injuries in badminton can be classed as overuse while the remaining 1/3 are classed as acute trauma (see Table 2.3). More
specifically, they show that the vast majority of injuries are located in the lower extremities (82.9%) which is far larger than any of the other racket sports.

Table 2.3 Injury onset by sport. Adapted from Caine et al. (1996).

<table>
<thead>
<tr>
<th>Injury Onset</th>
<th>Tennis</th>
<th>Badminton</th>
<th>Squash</th>
<th>Racquetball</th>
</tr>
</thead>
<tbody>
<tr>
<td>Overuse</td>
<td>30%</td>
<td>74%</td>
<td>20%</td>
<td>10%</td>
</tr>
<tr>
<td>Acute trauma</td>
<td>70%</td>
<td>26%</td>
<td>80%</td>
<td>90%</td>
</tr>
</tbody>
</table>

Table 2.4 Injury location by sport. Adapted from Caine et al. (1996).

<table>
<thead>
<tr>
<th>Injury location</th>
<th>Tennis</th>
<th>Badminton</th>
<th>Squash</th>
<th>Racquetball</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head/Neck</td>
<td>Not incl.</td>
<td>4.1%</td>
<td>18.8%</td>
<td>52.9%</td>
</tr>
<tr>
<td>Upper extremity</td>
<td>35%</td>
<td>11.1%</td>
<td>23%</td>
<td>12.1%</td>
</tr>
<tr>
<td>Trunk/Spine</td>
<td>45%</td>
<td>1.8%</td>
<td>10.1%</td>
<td>3.2%</td>
</tr>
<tr>
<td>Lower extremity</td>
<td>20%</td>
<td>82.9%</td>
<td>48.1%</td>
<td>31.8%</td>
</tr>
</tbody>
</table>

These findings are supported by a Danish study by Jørgensen and Winge (1987) who found that 74% of injuries were due to overuse, 12% due to strain injuries, 11% due to sprains, 1.5% due to fractures and 1.5% due to contusions. Furthermore, the authors report a shift in injury site from a roughly even distribution of injuries between the lower and upper limb in recreational players to predominantly lower limb injuries in the elite players (see Table 2.5). Particularly the foot and ankle in the elite and knee in recreational players are common injury sites.

The knee was furthermore identified as a major injury site in racket sports by Mohtadi and Poole (1996). They report that ligament, muscle and tendon injuries make up the majority of injuries in badminton (Table 2.6). Furthermore, achillodynia (inflammation in the peritenon or bursae) is quoted as the single most frequent injury, followed by tennis elbow, anterior knee pain (mainly patellofemoral pain syndrome and jumper's knee), plantar fasciitis and femoral muscle strains (Hensley and Paup, 1979, Jørgensen and Winge, 1987). This data adds to the reported trend toward overuse injuries in badminton with traumatic injuries a more common occurrence in other sports.

Høy et al. (1994) investigated the occurrence of badminton injuries at a casualty ward over a one year period. 100 badminton players were registered during this period constituting 5% of all sports injuries during this period. They found that 17% of badminton injuries were minor, 56% were moderate and 27% were severe, according to the Abbreviated Injury Scale (AIS) with 56% of the severe cases being in the over
25 years of age range. The most frequently diagnosed injuries were sprains (56%), fractures (5%), torn ankle ligaments (10%), ruptured AT (13%).

Traumatic injuries in badminton are sprains, strains, tears or breaks. The large percentage of sprain injuries and relatively high number of Achilles tendon ruptures stated by Hoy et al. (1994) confirms the ankle as one of the most likely sites for acute injury. Kaalund et al. (1989) state that Achilles tendon ruptures appear to be more prevalent in badminton based on their finding of 40% of ruptures occurring in badminton compared to 18% in soccer despite their estimation that more people participate in soccer. Furthermore, Kvist (1994) reports that of a total of 4000 Achilles tendon ruptures, 75% of the injuries were related to sports activities and particularly to sports involving abrupt repetitive jumping and sprinting movements. Similarly, Fahlström et al. (1998) found that male recreational players are at risk. With the majority of injuries (94%) occurring during the middle or end of game. These findings are supported by Fahlström et al. (2002) who found that 32% of young elite badminton players (16-34 years) had experienced disabling pain in the AT region during the previous 5 years and during midseason and 17% had an ongoing painful condition. Furthermore, 44% of 32 middle-aged badminton players reported pain in the middle of the Achilles tendon within the past 5 years which confirmed that AT pain seems to be relatively common problem for middle-aged competitive players. Maffulli (1999) suggests 3 Achilles tendon rupture mechanisms: Pushing off with the weight bearing forefoot while extending the knee (53% of ruptures), sudden, unexpected dorsiflexion of the ankle (17% of ruptures) and violent dorsiflexion of a plantarflexed foot (10% of ruptures).

<table>
<thead>
<tr>
<th>Table 2.5 The injury pattern in elite and recreational badminton (Jørgensen and Winge, 1987).</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elite players</td>
</tr>
<tr>
<td>Number of injuries</td>
</tr>
<tr>
<td>Foot</td>
</tr>
<tr>
<td>Ankle</td>
</tr>
<tr>
<td>Achilles tendon</td>
</tr>
<tr>
<td>Crus</td>
</tr>
<tr>
<td>Knee</td>
</tr>
<tr>
<td>Femur</td>
</tr>
<tr>
<td>Groin</td>
</tr>
<tr>
<td>Lower extremity total</td>
</tr>
<tr>
<td>Back</td>
</tr>
<tr>
<td>Thorax</td>
</tr>
</tbody>
</table>
Table 2.6 Injury type by sport (Mohtadi and Poole, 1996)

<table>
<thead>
<tr>
<th>Injury type</th>
<th>Tennis</th>
<th>Badminton</th>
<th>Squash</th>
<th>Racquetball</th>
</tr>
</thead>
<tbody>
<tr>
<td>Joint/Ligament/Sprain</td>
<td>64%</td>
<td>58.5%</td>
<td>20.3%</td>
<td>34%</td>
</tr>
<tr>
<td>Muscle</td>
<td>10%</td>
<td>19.8%</td>
<td>18.8%</td>
<td>Not included</td>
</tr>
<tr>
<td>Tendon</td>
<td>18%</td>
<td>8.8%</td>
<td>7.2%</td>
<td>1.3%</td>
</tr>
<tr>
<td>Skin</td>
<td>Not included</td>
<td>5.1%</td>
<td>36.2%</td>
<td>46.5%</td>
</tr>
<tr>
<td>Bone</td>
<td>Not included</td>
<td>5.1%</td>
<td>2.9%</td>
<td>7.6%</td>
</tr>
<tr>
<td>Eye</td>
<td>Not included</td>
<td>2.3%</td>
<td>-</td>
<td>5.7%</td>
</tr>
<tr>
<td>Inflammation</td>
<td>Not included</td>
<td>Not included</td>
<td>14.5%</td>
<td>1.9%</td>
</tr>
<tr>
<td>Nasal/Dental</td>
<td>Not included</td>
<td>Not included</td>
<td>-</td>
<td>3.1%</td>
</tr>
<tr>
<td>Other</td>
<td>8%</td>
<td>0.5%</td>
<td>-</td>
<td>Not included</td>
</tr>
</tbody>
</table>

Therefore, racket sports are associated with a variety of injury mechanisms. Musculoskeletal injuries appear to be mainly due to acute tissue overload such as acute musculotendinous injuries, acute ligamentous sprains, and overuse injuries involving the tendons. Musculotendinous injuries (e.g. "tennis leg") occur due to excessive eccentric overload. Kvist (1994) suggests that in competitive athletes the vast majority of overuse injuries are attributable to repetitive microtrauma over a period of time (months and years) while overloading, through for example vigorous training activities, is regarded as the main pathological stimulus (Kader et al., 2002) particularly in recreational athletes (Kvist, 1994). The cumulative effect of repetitive trauma in the sport is therefore singled out as the cause for the degenerative changes within tendons which could cause mechanical microtrauma. Jørgensen and Winge (1987) state that the combination of the sport-specific footwork and fast forward, backward and sideway movements together with repetitive stops and the use of forceful heel strikes and eccentric work result in repeated cycles of fast-changing high tension in the Achilles tendon which could produce microtrauma. This mechanism could also be the cause for patellofemoral pain syndrome where the fast-cycles of eccentric and concentric work of the quadriceps in knee flexion and rotation can create high retropatellar stress.
It therefore appears that there are a number of factors that influence injury frequency in badminton. Extrinsic factors include playing level, where elite players sustain more injuries than non-elite club or recreational players (0.92 and 0.7 injuries/season respectively) however, when looking at exposure time, elite players have a lower injury risk (2.8 vs. 3.1 injuries/1000h) (Jørgensen and Winge, 1987). A further factor is training and match play. It appears that there is an increased risk of injury for elite players during training than during matches (3.1 and 2.3 injuries / 1000h respectively). This is true for both men and women (3.1 : 2.8 injuries / 1000h and 3.1 : 2.1 injuries / 1000h respectively). Interestingly, recreational players who had trained less showed no difference in injury incidence. The exact reasons for these observations are however not known. The court surface may furthermore play an important role since badminton can be played on a variety of surface. However, there is no information available as to the influence this may have in badminton. Finally the badminton shoe and the shoe-surface interaction may contribute to overuse injuries due to the lack of heel support, low shock absorption and lack of heel elevation and the high friction and low shock absorption of the surface (Jørgensen and Hølmich, 1994).

Intrinsic injury factors include gender, physiology and pre-exercise warm-up. Male players appear to get injured more frequently than female players (0.9 and 0.78 injuries/season respectively) with men being at particular risk during matches (2.8 vs. 2.1 injuries/1000 match hours). It is thought that this may relate to a more intense and faster game adopted by male badminton players (Jørgensen and Winge, 1987). Furthermore, both pre-exercise warm up and stretching have been advocated to prevent injury but there is no evidence that lack of warm up or inflexibility are risk factors (Mohtadi and Poole, 1996). Physiological causes of tendon damage are related to the energy dissipated as heat within the tendon. During multiple stretch-shortening cycles the heat production may result in cumulative tendon thermal damage increasing the likelihood of tendon rupture (Maganaris et al., 2004). It is estimated that temperature during exercise may reach more than 42.5 degrees C, the thermal threshold for fibroblast viability.
2.5 Chapter Conclusion

The aim of this literature review was to assess the current knowledge of the physiology and biomechanics of typical and atypical gaits and to identify the proposed factors that predispose the athlete to injury and how these can be avoided. Based on the review, gaps in our knowledge relating to the fundamental mechanisms of atypical gaits, particularly with a view to their application in the singles badminton scenario are identified. With the reviewed information as a foundation and baseline it is now possible to assess the physiological requirements that result from the use of atypical movements in badminton.

While the kinematics and kinetics of walking and running gaits have been extensively studied, using a variety of methods, there is astonishingly little information relating to the mechanisms involved in atypical movements such as lateral stepping or lunging. In relation to lateral gaits, attention has been predominantly focused on the effects of cutting movements. There is also very little information available on the biomechanics of lunging and no information of the effects of lunge methods in the game of badminton. Furthermore, no information related to the use of movement in the context of the competitive game was found in the literature. This is surprising, considering the importance of movement, the high metabolic demands and the epidemiological evidence of a comparatively high occurrence of overuse injuries in badminton compared to a number of other sports. If the loads imposed on the body during movement can be assumed to be the fundamental cause of overuse and acute injuries in sports, then the high level of lower limb injury in badminton may be related to the loading characteristics of the movements used in the game and in training. Quantification of the movements used in the game and an assessment of the biomechanics of these movements appears essential in order to identify the contribution of movement to the game and the possible relationship between the observed biomechanical parameters and the occurrence of overuse as well as acute injuries. Moreover, the large metabolic demands of the sport may be due, in part, to a heavy reliance on the use of a number of atypical movements, particularly in the singles game. The repeated use of atypical movements may therefore be a key contributor to the high physiological demands of the sport.
The main aim of this research therefore is to quantify the biomechanical loads the musculoskeletal system of the lower limbs is exposed to during the performance of sport-specific lateral gaits and lunging. Furthermore, it is the aim to determine the contribution of the lower limb musculature in coordinating lateral movements and finally to describe the metabolic energy demands of the performance of lateral gaits which will aid in understanding the physiological requirements of the game of badminton on the athlete.

In order to achieve these aims, the following chapters (Chapter 4-9) present data for a number of investigations into the distribution of gait use, the kinematics, kinetics, muscular activity and metabolic demands of badminton-specific lateral-stepping and lunging movements, as well as the kinematics and kinetics of badminton-specific lunging are covered. The combined investigation of lateral stepping and lunging movements is essential, since lunges are thought to form an important link with lateral movements through the combined use of lateral stepping and lunge movement in-game. These data provide important objective data with the potential to reduce the risk of injury of the athlete and improve athletic performance.
Chapter 3

Methods

3.1 Chapter Introduction

Biomechanics "is the science that examines forces acting upon and within a biological structure and effects produced by such forces" (Nigg et al., 1999). According the review of the history of biomechanics in Nigg et al. (1999), the foundations of biomechanics can be traced back to the Antiquity (650 B.C. – 200 A.D.) where the Greeks and in particular the works by Thales, Pythagoras and Hippocrates developed the cornerstones of modern science. Due to changes in scientific focus, particularly the fascination of the Romans with civil and military engineering, the study of biological systems took a step back. Galen and his interest in understanding musculature and mechanisms of movement stand out as one of the major developments of the time. It was however not until the 1400s, after a generally bleak time for sciences during the middle-ages, that the study of the human being gathered pace again. Leonardo da Vinci (1452-1519) contributed significantly to the understanding of mechanics at the time and he furthermore studied not only the dead but also the living body in motion and the anatomy and workings of the muscles, bones and ligaments by combing aspects of art and science. There is some argument about the true father of biomechanics but the works of Galileo (1564-1642), who is regarded as the father of modern science (Einstein, 1954); Harvey (1578 - 1657), for laying the foundations of the study of the circulatory system; Borelli (1608 – 1679), for his work on describing the movement of animals including his works 'De Motu Animalium'; and Newton (1848b) stand out for their contributions to the foundation of the modern scientific approach and the development of the theory of mechanical analysis. However, it was the late 18th century that saw a real drive forward in the use of biomechanical analysis. Fuelled by a rise in sport and leisure activities, coupled with rapidly advancing technology, the interest in the biomechanical analysis of the
processes involved in locomotion increased leading to the development of an increasing number of methods for the study and assessment of human and animal gait.

A variety of different methods were used to answer the questions posed in this thesis. These include notational analysis of gait use within competitive badminton; the assessment of energetics of selected forms of locomotion; the recording of kinematic and ground reaction force data during gait; the calculation of kinetic data using inverse dynamics and the recording of electromyographic data during dynamic situations. The following section presents detailed information on the techniques used, the involved hardware and some of the data handling and analysis methods.

3.2 Notational Analysis

"Notational analysis refers to the recording of events for purposes of collating statistical details of performance. The performance of individual players can be outlined in minute detail, including the number of occurrences of each event, the actions and individual players concerned, and the location of these events on the pitch" (Reilly and Gilbourne, 2003). The technique was initiated in the 1970s by Downey who developed a complex analysis system for the analysis of tennis which, due to its complexity, has rarely been used in its full form, but the principles have proved transferable across racket sports (Hughes and Franks, 2004). Hughes (1998) defines the purposes of notational analysis as:

1. tactical evaluation
2. technical evaluation
3. analysis of movement
4. development of a database and modelling
5. educational use for both coaches and players

Notational analysis has found widespread application in a variety of sports for the assessment of the physical demands of the sport and the evaluation of tactical aspects of the game. In racket sports "the data collected are related to the position, action,
time and outcome of an event in the game; consequently, notational analysis is characterised by an extraordinarily large amount of data." (Lees, 2003)

In their review of research related directly to football Reilly and Gilbourne (2003) show how notational analysis has been used to assess match demand by recording total distance covered, intensity of activity and modes of locomotion. In football, notational analysis gives detailed information for individual players and information on differences between players, positions of play, differences in style of play and sport-specific demands over time as well as the assessment of risk of the game.

In racket sports the focus has been primarily on the evaluation of the technical and tactical aspects of the game. In tennis it has been used to assess the effects of new tennis balls on the singles game (O'donoghue and Liddle, 1998b) and the influence of court surface on the success of serve and net play by recording information on area of play (baseline/net) and shot selection and success rate (O'donoghue and Liddle, 1998a). Similarly, O'Donoghue and Ingram (2001) determined the influence of gender and court surface on elite tennis strategy by recording rally duration and shot characteristics. Hughes and Moore (1998) investigated the consequences of inefficient movement in terms of rally results and to provide a framework of the basic footwork movements of the tennis player while Taylor and Hughes (1998) used video footage to analyse competitive games of the top British under juniors to identify the distinguishing playing characteristics of British and other comparable players. Further applications of notational analysis include table tennis (Wilson and Barnes, 1998) and squash (Mcgarry et al., 1998).

In badminton notational analysis has been used by a number of researchers to investigate the physical requirements of the game and training as well as a large selection of game parameters affecting the technical and tactical aspects of the game.

- Pearce (2002) and Cabello and Gonzalez-Badillo (2003) combined notational analysis and physiological measures to evaluate the effect of different scoring systems on the athlete in terms of physical and tactical differences and to describe the general characteristics of badminton in order to determine the energy requirements and temporal structure of the game.
• Blomqvist et al. (1998) compared game understanding and game performance of participants between two age groups and genders to identify differences in game awareness and technical proficiency.

• Liddle and O'Donoghue (1998), O'Donoghue (2001), Pritchard et al. (2001) and Cabello Manrique and Gonzalez-Badillo (2003) established the general shot characteristics of the men’s and women’s events in order to described average rally length and rest periods and the interaction of these parameters and to identify variation in these parameters due to the influenced of the season (the year the competition is held), the rules used to play the game, the level of competition and the skill of the players.

The methods used for data collection are very much dependent on the data of interest and the level of detail. Methods therefore range from the simple recording of numbers or timings of events such as the number of shots played or length of a rally, which can be performed live, using a notepad or computer, to the more intricate analysis of tactical components, including the position of the player, shot selection and success and movement on-court, which tends to require post-match analysis using video recordings and is far more time consuming.

For the purpose of the investigation of gait use in competitive badminton singles games in this thesis notational analysis was performed using frame-by-frame analysis of digital video footage on a Microsoft Windows® based computer. The original badminton footage was either recorded live at competition venues, using a digital video camera (Panasonic NV-DS27), or was recorded from television broadcasts using a video cassette recorder. All video footage was in the phase alternating line (PAL) format with a frame rate of 25 frames per second. This footage was then digitised and saved to a PC as an uncompressed AVI file using Virtual Dub (freeware video editor by Avery Lee, http://www.virtualdub.org/index) or Power Director Pro (CyberLink Corp., http://www.cyberlink.com/), retaining the original temporal resolution and picture quality.

Notational analysis of the digitised video footage was performed following recommendations by Martin and Bateson (1986), using focal sampling techniques, the
observation of one participant for a specified period of time, for the description of gait bout frequencies and durations, where a bout is equivalent to the total duration of performing a distinct movement. Analysis of video recordings, identifying the numbers and duration of individual movements is time consuming but offers the greatest detail in identifying gait use. For the analysis, the video file under investigation was uploaded into Virtual Dub. The Virtual Dub interface allows the image to be enlarged, which ensured the best possible view of the gait events, and allows for swift frame-by-frame analysis of the footage by using the right and left cursor button for forward and backward progression through the individual frames. Furthermore the interface supplies a precise time and frame display which allowed for maximum temporal detail in identifying gait events and bout durations (see Figure 3.1).

Data analysis of the information on gait numbers and durations as well as information of durations of periods of activity and rest was performed in Microsoft Office Excel as outlined in Chapter 4.
Figure 3.1 VirtualDub video capture and playback window used for the sampling of rally and rest durations and the numbers and temporal characteristics of gaits used in the game. Figure (a) shows a typical XS/lunge combination and figure (b) shows a typical SS/lunge combination. The bar underneath the video footage acts as a visual display of the current frame in relation to the previous and following frames. The bottom right corner of each frame shows the frame digital readout of the frame counter and timer.
3.3 Energy Expenditure

Several approaches are currently used for the measurement of metabolic variables in exercise which are based on the methods developed for the Douglas bag system. The Douglas bag system was first developed by C.G. Douglas in preparation for the 1911 Anglo-American expedition of Pike’s Peak in Colorado, for the measurement of oxygen consumption (VO2) and carbon dioxide production (Macfarlane, 2001). In this system expired air is collected in an impermeable bag and gas analysis is performed after gas sampling using, at first, a chemical absorption method which was eventually replaced by faster electronic analysis methods. Further development of this system led to the creation of automated and semi-automated data recording metabolic carts (Macfarlane, 2001). The Douglas bag system has been in use for a significant period of time and is recognised for producing reliable and accurate data of very low temporal resolution.

To overcome the restrictions in temporal resolution imposed by the use of large mixing chambers, such as the Douglas bag, new systems have been developed using smaller mixing chambers (e.g. Metalyser II, Cortex, www.cortex-medical.de or VistaMX, Vacumed, www.vacumed.com) or a breath-by-breath approach (Metamax 3B and Metalyser 3B, Cortex or Quark b2, Cosmed, www.cosmed.it). In the mixing chamber method computer-driven gas sampling and analysis is performed at regular time intervals from a fixed or variably-sized chamber that is pervaded by expired air. However, while this increases the temporal resolution of a mixed gas concentration this method is a compromise between chamber volume and breath size, particularly during heavy exercise, and washout durations, which can lead to some data inaccuracy (Roecker et al., 2005).

The breath-by-breath analysis approach represents an attempt to solve the problem of obtaining high temporal resolution of the metabolic data and maintaining data accuracy over a large range of exercise intensities. With the aid of fast gas analysers breath-by-breath measurements can be taken in a virtual mixing chamber where respiratory volumes as well as gas concentrations are averaged from the
instantaneous gases between the onset and the end of each breathing cycle' (Roecker et al., 2005).

Further development of the technology and the reduction in size and weight of gas analysis and flow-rate measuring components and microprocessor technology thereafter led to the production of compact, lightweight and portable gas analysers such as the MetaMax 3B. The accuracy and reliability of this and a number of related breath-by-breath analysis systems has been assessed by a number of researchers:

- Meyer at al. (2001) compared the reliability of a portable mixing chamber system (MetaMax I) and a stationary breath-by-breath measurement system (Metalyzer 3B). The results from 23 participants performing three ramp tests on a cycle ergometer to exhaustion, exhausting participants within 10 to 12 minutes, indicated that both instruments represent reliable data for exercise testing.

- Larsson et al. (2004) investigated the validity of the MetaMax II portable metabolic measuring system against the Douglas bag technique. The MetaMax II system used a small mixing chamber with respiratory rates provided every 10 seconds. Their sample consisted of nine recreationally active male participants performing bicycle ergometry at 100W, ten well trained male participants at 200W and ten well trained male participants at 250W and maximal exercise intensity (volitional fatigue at a mean workload of 325W). The findings of the investigation led the authors to conclude that the MetaMax II system is a valid tool for metabolic gas measurements between 100 and at least 250W.

- In their review of the current status of portable gas analysers Meyer et al. (2005) show that at least two of the tested devices, the MetaMax I/II and Cosmed K2/K4b² can be regarded as valid measuring tools in comparison to stationary metabolic carts. However they furthermore point out that it is important to consider a possible publication bias within studies performed to assess data accuracy of such systems as they are often manufacturer sponsored.
Carter and Jeukendrup (2002) assessed the validity and reliability of three commercially available breath-by-breath gas analysis systems by comparing them to the Douglas bag method. Their results, based on comparisons using a respiratory simulator as well as ten healthy participants using work-rates of 100 or 150W over 3 separate testing sessions indicated that two of the three tested systems were reliable and valid in comparison to the Douglas bag method. They also highlighted the fact that the third system produced highly variable data, which emphasises the need for accuracy testing of measurement equipment to assess and ensure data validity.

For the purpose of investigating the energy requirements of free moving gaits in this thesis, a portable breath-by-breath gas analysis system (MetaMax 3B, Cortex Biophysik GmbH, Leipzig, Germany) was used for the sampling and analysis of expired gases and gas volumes. The system uses a breath-by-breath approach meaning that “a particular physiologic value is determined for each of a subject’s single respiratory cycles” (Roecker et al., 2005). The system includes a lightweight portable system worn by the subject around the shoulders and chest and a receiver unit for wireless data transmission and recording by a PC (Figure 3.2). Hanging the system around the shoulders of the participant, thereby distributing the weight to the front and the back, has been shown to be superior to carrying the analyser on the back (Lloyd and Cooke, 2000). The portable unit weighs 2.7 kg and includes a silicon facemask with integrated flow-rate turbine and expired air sampling line, which connects to the battery-powered gas analysis and transmitter unit worn on the subjects’ chest. The turbine was calibrated regularly with a 3 l syringe using ambient air and a standard calibration gas mixture according to the manufacturers’ recommendations.
Shoulder Harness and Head Strapping

Gas Sampling and Drying Tube

Facemask with Flow Turbine

Cardiopulmonary Measurement System

Figure 3.2 MetaMax 3B mobile gas analysis system with important components identified. The system uses a harness to hold the cardiopulmonary measurement components in place. The silicone facemask comes in 3 different sizes and is equipped with a flow turbine to sample respiratory rates and volumes. Expired air is sampled from the mask by the measurement system via the gas sampling and drying tube.

The system was used in the wireless configuration with live data capture and direct data readout on-screen, which helped in identifying gas equilibrium and judgement of the necessary exercise duration. Use of the MetaMax gave the participant more freedom of movement than the traditional Douglas Bag system and allowed measurements to be taken quickly and directly throughout the free moving activities without the need for post exercise gas analysis. Participants therefore needed to perform the movement for a shorter period of time.

The accuracy of the MetaMax 3B system used for the analysis of energy consumption is this thesis was validated at the time of testing by Lucy Dorman PhD. A comparative analysis of oxygen consumption and heart rate during exercise was performed on 3 participants using the Douglas bag method and the MetaMax 3B gas analysis system. The three participants performed two 30 minute walking trials on the same day with a rest and recovery period in between exercise sessions. Oxygen consumption and heart rate were measured using the Douglas bag and a heart rate monitor (Polar belt and watch) in one session and the MetaMax and heart rate (Polar belt and integrated sampling via MetaMax) in the other. The participants sat at rest for at least 6 minutes.
prior to completing three 8 minute walking exercises at 3.5, 5 and 7.5 km/h. During the Douglas bag trials recordings were taken for at least 4 minutes with gas analysis being performed in the final 2 minutes of each stage. The findings, summarised in Figure 3.3 and Figure 3.4, showed close fits for the data collected using the two systems which suggests that the MetaMax 3B was indeed an accurate and reliable system for gas analysis.

Figure 3.3 Oxygen consumption data collected with Douglas bags and MetaMax at rest and over 3 different treadmill speeds.

Figure 3.4 Heart rate and oxygen consumption data collected with Douglas bags and MetaMax fitted with regression lines.
3.4 Anthropometry

Anthropometric measurements were performed using a set of bone callipers (Holtain Ltd., see Figure 3.5) and an anthropometric measuring tape (Harpenden Anthropometric Tape, Holtain Ltd.). The accuracy of the callipers was validated prior to measurements being taken using a 5cm long metal calibration bar. Following the instructions by Charnwood Dynamics (2003) for the hip, knee and ankle measures, the measurement criteria were as follows:

- **Leg length**: the distance between the anterior superior iliac spine (ASIS) and the lateral malleolus.
- **Knee width**: the distance between the lateral surfaces of the medial and lateral epicondyles of the femur.
- **Ankle width**: the distance between the lateral surfaces of the medial and lateral malleolus of the tibia and fibula respectively.
- **Hip width**: the distance between the lateral surfaces of the left and right ASIS.
- **Hip depth**: the distance between the anterior edge of the ASIS and posterior edge of the sacrum.

Figure 3.5 Anthropometric measuring tools used in this thesis. A Harpenden Anthropometer (a) was used for measuring hip width and depth and a Holtain Bicondylar Caliper (b) was used for the measurement of ankle and knee width.
3.5 Ground Reaction Force

A Kistler (Type 9286AA) mobile multi-component forceplate (Figure 3.6), integrated into a movable walkway, was used for the recording of the horizontal (Fx), lateral (Fy) and vertical (Fz) force components of ground reaction force (GRF) acting on the top plate of the force platform. The force applied to the top plate is distributed between the 4 piezoelectric three-component force sensors located in the corners of the force plate each of which measures force in the 3 dimensions. The natural frequency of the force plate is ~350Hz (Fx, Fy) and ~200Hz (Fz). An internal charge amplifier converts the electrical charge, which are proportional to the applied force, into analogue voltages. The top plate was covered with a hard carpet surface which was glued on using a strong carpet adhesive to create a stable top surface with high friction and low shock absorbance. The same carpet material was used for the forceplate and the walkway which ensured that the participant had a consistent surface to move on and disguised the location of the forceplate in the walkway.

![Figure 3.6 Kistler mobile multicomponent force plate Type 9286AA. Dimensions 600 x 400 x 35 mm.](image)

The forceplate was connected to an analogue 8 channel Kistler Control Unit (Type 5233A2) using a connecting cable (Type 1760A10). The Control Unit (Figure 3.7) acted as a power supply for and external control of the force plate and allowed the operator to set the required force range, reset the force plate and indicate force
overload. Furthermore it relayed the analogue data from the forceplate to the computer (using the connecting cable Type 1500B5), where the data was converted to digital signal (used as input for the CODAmotion software) using an on-board A/D converter on one of the CODA Mpx30 ISA cards.

![Figure 3.7 Kistler Control Unit Type 5233A](image)

Figure 3.7 Kistler Control Unit Type 5233A

![Figure 3.8 Digital force gauge used for the validation of the forceplate in the X and Y axes. The force gauge had a maximum accuracy of 0.05kg and was attached to a broad metal plate with a hook attached to the end of the plate. This hook was placed over the centre of the edge of the top plate and a force was applied by pulling the force gauge. The peak force displayed by the force gauge was used for comparison with the forceplate readings.](image)

Figure 3.8 Digital force gauge used for the validation of the forceplate in the X and Y axes. The force gauge had a maximum accuracy of 0.05kg and was attached to a broad metal plate with a hook attached to the end of the plate. This hook was placed over the centre of the edge of the top plate and a force was applied by pulling the force gauge. The peak force displayed by the force gauge was used for comparison with the forceplate readings.

The accuracy of the force plate was validated in all three axes. A digital force gauge (Chatillon Digital Force Gauge, DFIS 100, Ametek Inc., see Figure 3.8) and a
The weighing scale were used to compare expected with the recorded force for a series of forces within the range expected to be used in the experiment. The force gage and weighing scale were calibrated using the following methods:

The weighing scale was calibrated using five water volumes of 1 to 5 litres (measured using a 1 litre graduated measuring cylinder, where 1 litre is equivalent to 1kg weight). Throughout the calibration process the average of three measures was used to determine the weight. The data was entered into Microsoft Excel and the equation of the linear regression line for predicted and actual weight (Figure 3.9a) was used to calibrate the force gauge which was used to calibrate the X and Y axis of the force plate. For this nine weight measures ranging from 2.7 to 28.2kg were measured on the weighing scale and re-measured using the force gage. The offset was applied to the weight values from the weighing scale to obtain the calibrated predicted weight and was thereafter plotted against the weight measured using the force gage in order to obtain the offset value for the force gauge.

Figure 3.9 Measured versus predicted weights for a) the weighing scale and b) the force gauge. Weights are expressed in kilograms (kg). The resultant straight line equation in a) and b) was used to calibrate the measures of the weighing scale and force gauge respectively.

Calibration of the force plate along the X and Y axis was performed by pulling the top plate of the force plate, using the force gage, in both the positive and negative direction along the X and Y axis, thereby controlling the applied force. The maximum force reading from the force gauge and the force plate were used for comparison. The reading of the force gauge was calibrated using the equation from Figure 3.9b and the predicted and measured force was plotted for the X and Y axis (Figure 3.10a and b)
respectively). The data for this data range can be seen to be close to a perfect fit with a mean difference of 0.89% in the X-axis and 1.11% in the Y-axis for the predicted and recorded values. The observed discrepancy is thought to be in part due to the maximum accuracy of the force gauge of 0.05kg.

Figure 3.10 Measured versus predicted force for a) the X-axis and b) the Y-axis of the forceplate. Force is expressed in kg.

Calibration of the forceplate in the vertical axis was performed using a range of weights (8 weights ranging from 60 to 152kg) was measured on the weighing scale and this measure was compared to the mean measure of steady vertical GRF of the same weight. The mean data inaccuracy between the two measures was 0.47%. The forceplate was therefore deemed to produce sufficiently accurate data for the purpose of biomechanical investigations of human gait.
A consistent convention for the definition of the action of GRFs was used throughout based on the local coordinate system employed by the force plate. Positive and negative forces along the X-axis represent horizontal push-off and breaking forces respectively. Positive and negative forces along the Y-axis represent lateral sway to the right and left respectively for walking and running and posterior and anterior pushes of the foot during lateral movements and positive forces in the Z-axis represent vertical GRF.

Figure 3.11 Typical GRF trace for walking, including classification of the positive and negative force components. The functional significance of the positive and negative force components is affected by the direction of travel by the participant in relation to the forceplate and therefore Fy differs between walking / running and the lateral movements.
3.6 Kinematics

Real-time bilateral kinematic data was recorded using a two camera Cartesian Optoelectronic Dynamic Anthropometer (CODA, Charnwood Dynamics Ltd, UK). The CODA scanner units (CPX-1 and MPX-30; see Figure 3.12) have three infra-red sensors, with the associated processors being located in a host PC, that detect the location of a number of re-active light emitting diode (LED) markers through cross-correlation techniques. The quoted resolution is \( \sim 0.002^\circ \) when placed parallel to the walkway at a distance of 3m which is equivalent to a resolution of \( \sim 0.05\text{mm} \) for the X and Z axis and \( \sim 0.3\text{mm} \) for the Y axis. The MPX-30 scanner unit was mains powered and communicated with the PC (active hub) directly and acted as the master unit. Communication of the CPX-1 scanner unit was performed through a MiniHub that provides power and data connections for the CPX-1 scanner. The MiniHub furthermore routes a hardware synchronization signal between the two scanners used to synchronize multiple scanner units.

The marker diodes are connected to marker drive boxes (see Figure 3.13) that provide the marker with an intrinsic identity. This means that their position and therefore the position of the structure associated with the marker is instantly recognised without the need for post-processing of marker positions or trajectories. As a result both the driver box and diode must be visible to the cameras throughout. The system uses a variable recording frequency of 100Hz to 800Hz depending on the number of markers used. For the purpose of bilateral gait analysis, using 22 markers and 11 drive boxes, a sampling frequency of 200Hz was used. The markers were attached to a specialised
wand system, including a pelvic frame and thigh and leg wands, supplied by Charnwood Dynamics Ltd (see Figure 3.14). This method allowed for the calculation of internal joint centres for the hip, knee and ankle joints and their 3D internal rotations using CODAmotion software (Charnwood Dynamics Ltd. V6.68). The marker locations outlined in Figure 3.15 acted as the basis for the determination of 3D segments and joints, where each segment is defined by three separate points and additional subject-specific anthropometric data is used for a complete description of the limb segments and the respective joint centres as well as inverse dynamic calculations as outlined in Chapter 3.7.

Figure 3.14 Wand system used for gait analysis. The system consisted of one pelvic frame, two thigh and leg wands. The Pelvic frame was secured to the participant using the provided Velcro belt which itself was additionally secured using masking tape. The thigh and leg wands were secured to the limb using self adhesive strapping to assure the wands stayed in a fixed position throughout gait. Additional markers were required for the location of the knee and foot which were placed directly on the skin and surface of the shoe of the participant.
Marker Name | Marker Location
---|---
\'R. & L. Sac. Wand' | On opposite sides near the end of the sacral wand.
\'R. & L. PSIS' | On the sides of the pelvic frame, near the back.
\'R. & L. ASIS' | At the far front of the pelvic frame.
\'R. & L. Front. Wand' | Optional
\'R. & L. Post. Fem.' | At the posterior end of the femoral wand
\'R. & L. Ant. Fem.' | At the anterior end of the femoral wand
\'R. & L. Knee' | Directly on skin of lateral
\'R. & L. Post. Tib.' | At the posterior end of the tibial wand.
\'R. & L. Ant. Tib.' | At the anterior end of the tibial wand.
\'R. & L. Ankle' | Directly on skin of lateral malleolus.
\'R. & L. Heel' | On the surface of the lateral side of the calcaneus
\'R. & L. Toe' | On the surface of the lateral end of the 5th metatarsal

Figure 3.15 Marker identities and positions used for the purpose of bilateral acquisition of gait data. A total of 22 diodes were used where (a) and (b) represent the right and left limb respectively.

The orientation of the pelvic frame and thigh and leg wands was performed following factory recommendations:

- The tilt of the pelvic frame should reflect the plane of the ASIS and PSIS of the pelvis and the sacrum should lie just below the base of the sacral wand.
- The femoral wand should be rotated at the hinge until perpendicular to the knee axis. The femoral wand is strapped to the thigh just above the knee but below the major bulk of thigh muscle. Orientation must be perpendicular to the knee-joint axis.
- The tibial wand should be rotated at the hinge until perpendicular to the ankle axis. The bracing of the tibial wand is placed on the anterior bony surface of the tibia, below the knee. The wand orientation is perpendicular to the ankle-joint axis.
- The wands thereby define the orientations of the segment local embedded (transverse) axes, but do not correspond to any particular anatomical landmarks.
Assuming proper marker placement and wand orientation, these marker positions were used to derive the segments of the model (Figure 3.16). The dynamic representation of any segment is performed through vector reconstruction, using a minimum of three points. First the local co-ordinate frame (or 'Embedded Vector Basis' EVB), needed for the calculation of Euler Angles, is obtained, where the orientation of the pelvis EVB is obtained from the available marker set and the reference points are determined using local offsets. For the remaining limb segments the reference points are obtained by vector constructions followed by the derivation of the local EVBs from these.

Following is a summary of the methodology adopted for determining segment orientations and joint centres for the lower limbs as stated in the CODA user manual (Charnwood Dynamics Ltd., 2003).

**Pelvis**

- The orientations of the co-ordinate frame axes of the pelvis are determined before applying local offsets for the reference points. The sacral wand marker

![Figure 3.16 Representation of the segments of the lower limb model, derived from the recorded marker positions and local offset values. The complete pelvis and the right limb are presented and information for marker positions and reference points are given. The figure was taken from the CODA user manual (Charnwood Dynamics Ltd., 2003).]
is not used to determine the position of the sacrum; it merely contributes information about the pelvic plane.

- Bilateral compared to unilateral data acquisition allows for a robust platform for pelvis calculations since there may be as many as 6 markers representing the pelvis.
- The Gram-Schmidt process is used to generate a set of orthogonal axes to obtain the Sacrum reference point.
- The left and right ASIS reference points are offset anteriorly by the pelvic depth from the sacral reference point and medio-laterally by half of the pelvic width.
- Hip Joint Centres are obtained by adding local offsets to the mid-point between left and right ASIS reference points. The offsets are calculated as proportions of pelvic width in accordance with values allocated in the Patient Data input file in CODAmotion, based on the pelvis model described Bell et al. (1989).

**Thigh / Knee Joint**

- Thigh segment representation derives from a combination of data from the thigh wand markers and the knee and hip joint. The knee joint is modelled as a simple medio-lateral axis and as such is defined by the knee marker on the lateral aspect and a medially offset reference point on the medial aspect.
- The medial offset is 1 knee width from the lateral aspect in an effectively perpendicular direction to the plane defined by the VirtualHip and the two thigh wand markers. The VirtualHip point is laterally shifted from the Hip joint by approximately half a knee width as are the real thigh wand markers. The purpose of this adjustment to the thigh wand plane is to align it more accurately with the femur for an improved perpendicular medio-lateral knee axis.
- The KneeCentre reference point is simply the mid-point of the so-defined knee axis.
- The Thigh EVB is constructed around the principal femoral ‘z’ axis defined between HipJointCentre and KneeCentre. The thigh wand markers define the
local x axis following the Gram-Schmidt procedure applied with the principal axis already in place.

Leg and Ankle Joint

- This segment representation follows a similar procedure to that for the thigh and knee above. The principal axis of the tibia is defined between the KneeCentre (from above) and the AnkleCentre which is taken to be the midpoint of the ankle axis. The ankle axis is defined as the line extending medially from the Ankle marker with orientation perpendicular to the shank-lateral plane (as defined by the Tibial Wand markers and the line joining the Knee marker to the Ankle marker). The Tibial Wand local transverse projection will define the shank-local x axis and the ankle axis is taken to be perpendicular to that.

Foot Segment

- This is defined by three points: the first is the AnkleCentre, as defined above within the shank segment; the other two reference points, 'Heel' and 'Toe', are simply medially offset from the Heel and Toe markers by half an ankle-width, w, (Patient Data again). The principal axis of the foot is taken to be parallel to the line between Heel and Toe markers; this is the local x axis.

The segment rotations obtained from CODAmotion refer to the lower limb model and its own reference frame and are therefore independent from the direction of travel and consistent for all tested participants. The nomenclature used for joint rotations, used consistently in this thesis, is summarised in Table 3.1.

| Table 3.1 Nomenclature used for the definition of the three dimensional kinematic. |
|--------------------------------|--------------------------------|--------------------------------|
| Hip (Pelvis-Thigh)             | Knee (Thigh-Leg)               | Ankle (Leg-Foot)               |
| Hip X  +ve                     | Adduction                      | Knee Y +ve                     |
| -ve                             | Abduction                      | -ve Flexion                    |
| Hip Y  +ve                     | Flexion                        | Ankle Y +ve                    |
| -ve                             | Extension                      | -ve Dorsiflexion               |
|                                  | Extension                      | Plantarflexion                |

Cappozzo (1991) summarises the sources of error in the recording of kinematics. Errors in the reconstruction of marker trajectories can be systematic or random, where
the systematic error is due to optical distortion and other sources of error related to the instrumentation and random error is a result of the digitizing process, the transformation of positions into numerical values or related image processing equipment. The high frequency content of these error sources can make the unfiltered use of the raw signal impossible which means that low pass filtration may be necessary. As a consequence of this, the factory recommended filtration tools were used for the positional marker data. The system applies out of view interpolation, sub sample interpolation and a de-skew algorithm to prevent data loss when markers become obscured for short periods of time and to counteract marker detection artefacts. Furthermore the in-built low-pass filter, a running average algorithm, was used to filter out high frequency noise resulting from photo-detector current noise in the cameras, the AC component of the room lighting, the derivation of the depth measure from the triangulation process of the two outer cameras, as well as the mid-frequency noise that results from the movement of the markers on the skin or clothing. While this noise may not appear significant for the raw marker positional data, it may adversely affect the derived data, including joint angles and joint kinetics.

The accuracy of the two CODA scanner units (MPX-30 and CPX-1) for measuring angular data was validated by recording a number of markers over a range of distances and rotations. The cameras were positioned at opposite ends of the walkway at the distances indicated in Figure 3.17. A number of markers were attached to a wooden frame corresponding to angles of 20, 40, 60, 80 and 100 degrees. Measurements of the angles were taken at the 7 measurement points where each point was separated by a distance of 50cm. Furthermore, the frame was rotated by 20, 40 and 60 degrees at measurement point 3 to investigate the effect of rotation on the marker recognition and consequently accuracy of the angular data.
The recorded angular data was processed using manufacturer recommended out-of-view interpolation, sub-sample interpolation and marker de-skew. Furthermore, the marker data was low pass filtered at 10Hz as used for the analysis of the lateral movement tasks. Data analysis was performed using the root-mean-square (RMS) error of the residuals of the angular data. The results showed that increasing the distance between the markers and the cameras caused an increase in error (see Figure 3.18). This error was largest for the older camera (MPX-30) with a maximum error of 0.56° at the furthest distance, while the measurement error of the newer CPX-1 unit was considerably lower at 0.11° for the furthest separation. Similarly, rotation of the markers caused an increase in measurement error (Figure 3.19). Again this error was largest in the MPX-30 unit with a maximum error of 0.46° and 0.12° for the CPX-1 unit. Overall, the cameras were deemed sufficiently accurate for the recording of kinematics of human gait. It is important however to note that rotations exceeding 60° caused considerable marker loss. As a result camera positions were arranged to ensure optimal marker visibility with a parallel view of the markers during gait.

Figure 3.17 Set-up for marker recognition validation. Outline of the walkway (dashed line) and force plate as well as the motion capture cameras (MPX-30 and CPX-1) are given. 8 measurement point were used (0-7) each of which was separated by 0.5m.
3.7 Kinetics

Kinetic data was calculated from GRF and kinematic data using an inverse dynamics approach within the data capture and analysis software (CODAmotion) supplied by Charnwood Dynamics. For the calculation of inverse dynamics subject-specific information for the estimation of limb segment properties and joint centres was required in order to accurately calculate the relevant joint moments and powers. The required subject-specific data included participant age, weight, joint width and leg-segment data (segment length, mass-ratio, centre-of-mass position, radii of gyration). The segment data applied to the model for the calculation of joint kinetics in the
CODAmotion software was based on the body segment data reported by Jensen (1989) and took into account the change in segment parameters due to age.

For the calculation of inverse dynamics the lower body was considered as individual limb segments connected at the joints, where the limb segments were treated as free bodies. In order to simplify the inertial properties of the limbs each segment is modelled as a fixed, uniform distribution of mass around the longitudinal axis connecting the joint centres with the centre of mass located on this axis at a fixed proportion of the segment length from the proximal end. The coordinate system for each segment is defined as described above. Torque and angular velocity and accelerations are expressed with respect to the embedded co-ordinate system. Using a number of linear forces including weight due to gravity, ground and joint reaction force and the anthropometric data then allows for the calculation of inverse dynamics to determine joint moments and powers.

McCaw and DeVita (1995) quantified the effect of errors in the spatial alignment between the centre of pressure recorded from a force platform and the coordinates of the foot recorded from film on the resultant joint torques in the lower extremity during the stance phase of gait. Their findings of 7 and 14% changes in maximum joint torques due to +-0.5 and +-1.0cm shifts in location of the centre of pressure led them to conclude that the joint torques for gait, stated in the literature, should be considered to be approximations of the true values due to the potential errors in the spatial alignment of kinetic and kinematic data. This is in line with the findings by DeVita and Stribling (1991) who identified joint torque variation of up to 15% for maximum values when performing multiple repeats of data analysis.

3.8 Electromyography

According the De Luca (1997) the recording of surface electromyography (EMG) recording has three dominant uses in biomechanics: as an indicator of the initiation of muscular activation, its relationship to muscular force, and as an index of the processes of muscular fatigue. In this thesis the surface EMG signal was used as an
indicator of muscular activity in order to provide the timing and sequence of muscular activations of the leading and trailing limb during SS and XS gaits.

There are a number of issues connected to the recording and processing of EMG data. DeLuca (1997) summarises the considerations that an investigator should bear in mind in five cardinal questions:

1. Is the EMG signal detected and recorded with maximum fidelity?
   a. What are the configuration, dimension, and electrical characteristics of the electrode unit?

2. How should the EMG signal be analysed?
   a. How are the initiation and cessation times of the EMG signal measured?
   b. What are the preferred parameters for measuring the amplitude of the EMG signal?
   c. What are the preferred parameters for measuring the frequency spectrum?

3. Where does the EMG signal originate?
   a. Is there crosstalk? That is, does any of the detected signal originate from nearby muscles?
   b. Where is the electrode placed on the surface of the muscle in relation to its anatomical structures?
   c. How much fatty tissue is there between the electrode and the muscle surface?

4. Is the EMG signal sufficiently stationary for the intended analysis and interpretation?
   a. Does the muscle change length during the contraction?
   b. Is the activation pattern of the motor unit stable? That is, do some motor units alternate between the state of recruitment and derecruitment?

5. Where does the measured force originate?
   a. What is the state of the synergistic and antagonistic muscles associated with the task?
b. Are the motor control characteristics of the contraction stable for the intended interpretation? Is there any change in the relative force contribution among muscles during the contraction?

c. Is the force generated homogeneously throughout the muscle?

There are a number of factors that can influence the EMG signal. These include electrode configuration, the area and shape of the electrode detection surface, which influences the number of active motor units detected and the bandwidth of the differential electrode configuration, the location of the electrode on the muscle surface, which determines the amount of crosstalk that may be detected by the electrode and the orientation of the detection surfaces with respect to the muscle fibres, which affects the value of the measured conduction velocity of the action potentials and, consequently, the amplitude and frequency content of the signal. Furthermore, there are intrinsic factors due to the physiological, anatomical, and biomechanical characteristics of the muscle. These include the number of active motor units, which contributes to the amplitude of the detected signal, fibre type composition of the muscle, blood flow in the muscle, fibre diameter, the distance between the active fibres and the electrode detection surface and the amount of tissue between the surface of the muscle and the electrode.

Despite these and a number of other related factors and their interactions surface EMG can be used with confidence for the determination of muscle activations (on-off timings) provided there is no significant crosstalk from adjacent muscles and the electrode is placed on the surface of the muscle between theinnervation zone and the myotendinous junction. The likelihood of detecting crosstalk is high but can be reduced by accurate marker placement (De Luca and Merletti, 1988) and post data capture filtering appears unable to reduce cross-talk (Dimitrova et al., 2002).

An eight analogue channel system (Biometrics Ltd., Gwent, UK, www.biometricsltd.com) with two additional digital input channels was used for the recording electrical activity from seven muscles of the lower limb. The system uses dry reusable pre-amplified bipolar surface EMG electrodes (SX230) (Figure 3.20) that are attached to the body using double sided adhesive tape.
The electrodes are connected to a subject worn unit (Figure 3.22 b) which acts as a link between the participant and the stationary DataLink unit (Figure 3.22 a). The DataLink acts as a power supply for the subject unit. Real time data recording was performed using a desktop PC and the data recording and analysis software supplied by Biometrics (see Figure 3.23).
In addition to the EMG data analysis method described in Chapter 6.2.7 a number of alternative analysis methods were attempted. In order to automate the identification of activation times of the lower limb muscles of interest the time synchronised EMG data trace was used as input for signal processing in Matlab® (The MathWorks www.mathworks.com/). The datastream, saved as a text file, was imported and any DC offset was removed to ensure a baseline reading (when the muscle is inactive) equal to zero. This was followed by rectification of the EMG signal and low pass filtering of the rectified signal using a 5th order Butterworth filter with a 10Hz cut-off frequency. Thereafter a linear envelope was applied using the filt-filt command. The resultant trace was then used to detect muscle activation onset and offset times using the onset.m command.

Similarly when importing the data into Excel it was appended to automate or semi-automate the process of activation time identification. This was done by developing simple routines for the detection of muscle activation based on activation magnitudes in excess of a predefined magnitude e.g. 5% of the maximal activation (from baseline
to peak activation). The major problem associated with this and the method described above was caused by the frequent absence of a consistent baseline reading in a dynamic movement.

### 3.9 Foot Contact Identification

The identification of foot contact events (i.e. heelstrike (HS) and toe-off (TO)) is important for the definition of the swing and stance phase of the gait cycle and forms an important aspect in the description and definition of gait. The identification of foot contact events can be performed using a number of different methods including forceplates, contact switches, and kinematic data. The magnitude of vertical force, recorded by the forceplate, has been used by a number of authors in order to define foot contact events. Neptune et al. (1999) used a cut-off magnitude of force of >20N for HS and <20N for TO for cutting while Buczek and Cavanagh (1990) set a cut-off frequency of 44N for running.

An alternative to the forceplate is the use of footswitches which enables the recording of footfall events without the special restrictions of using a forceplate. Mills et al. (2007) compared the accuracy of the footswitch approach to the forceplate using a sliding window method. For the determination of HS the height of the sliding window was equal to the mean of the previous 40ms plus 3 standard deviations of the vertical GRF signal during the initial 100ms of the trial and the same approach, but moving in the opposite direction was used for the prediction of TO. The same technique was used for both the footswitch and forceplate data analysis, where the sliding window height for the footswitch data was defined as the mean of the previous 40ms plus 10% of the signal range. The techniques were found to result in very similar results with only small differences (<6ms or ~1% stance duration) in timings with different walking speeds.

Wall and Crosbie (1996) compared a field counting technique for video data with footswitches and force plate data. The timing of HS and TO events was taken to be equivalent to >2.5 and <2.5N for forceplate data (to assure equivalent force as necessary to switch footswitch). The footswitches were attached to light emitting
diodes (LED) and HS and TO events were assessed by advancing the video footage frame by frame and recording the point in time where the diode was activated and deactivated. Furthermore, the field counting technique was repeated without the visual LED signal to allow for unaided visual determination of footfall events. The findings indicated that the duration of the gait phases, derived from the footswitch data, was affected by the type of shoe the participant wore, where training shoes cause shorter duration of the gait phases since the heel switch closed later and open earlier compared to the times obtained from the forceplate. Similarly outdoor shoes showed some attenuation in the measurement of support time and the differences between the forceplate and footswitch data were least when the subjects were barefoot. This study furthermore demonstrated that the technique of visually determining the times of heel contact and toe-off from video recordings is highly reliable and shows minimal error. Ghoussayni et al. (2004) reported similar findings stating that there was good agreement in determining HS and TO timings between using a kinematic based algorithm, forceplate and visual inspection of markers.

As seen by the examples of Wall and Crosbie (1996) and Ghoussayni et al. (2004) footfall events can be determined using only kinematic data. In order to determine foot contacts from kinematic data it is necessary to specify the parameters that allow for the correct identification of footfall events. Hreljac and Marshall (2000) and Hreljac and Stergiou (2000) compared forceplate data (HS>10N, TO<10N) with kinematic algorithms based on the time of local maximum in the vertical component of acceleration of the heel markers and the horizontal component of acceleration of the toe marker for the determination of heel strike and time of toe-off respectively. Similarly, O’Connor et al. (2007) compared the validity of determining footfall events using a foot velocity algorithm based on the displacement time data for the heel and toe markers compared to measures from a forceplate (HC>10N, TO<5N). In all cases the use of kinematic algorithms was concluded to be an easy and reliable method for footfall determination. Moreover, Mickelborough et al. (2000) investigated the accuracy of kinematic data, the vertical displacement of the heel or toe and the marker velocity, compared to the forceplate (HS>10N, TO<2N) in determining footfall events and observed that 88-98% of all ratings were accurate to within 0.03s. Examples of the use of kinematic data for footfall determination include Alton et al. (1998) who used positional data from toe and lateral malleolus, Schache et al. (2001) who used

These findings show that a number of different analysis methods can be applied reliably in order to determine the timing of kinematic events. The fundamental determinant of the accuracy of the temporal patterns therefore must lie in the recording frequency used. Polk et al. (2005) states that “when using any motion analysis system, the beginning and end points of any time interval will always be recorded late because they must occur before they will first appear in the motion capture device.” This means that the limb under investigation will make contact with and leave the ground before the motion analysis system records these instances. These timing differences are random, since the recording and gait frequency are independent of each other and consequently the maximum error of the recorded time intervals will be equal to the duration of a single measurement point by the motion analysis system (i.e. ± d). A higher sampling frequency will therefore result in a smaller maximum error and a faster movement will result in underestimations unless the sampling frequency is increased accordingly.

Overall it can be concluded that in view of the comparisons summarised above there are a number of valid and reliable methods for determining footfall events. The use of force data represents a very accurate footfall recognition method however the resultant timings are very much dependent on the chosen cut-off level of force. The examples above show the variation in the chosen force levels between investigations and this has to remain a point of consideration when interpreting results. Furthermore, the use of footswitches simplifies footfall recognition by its accuracy is affected by the external factors (i.e. footwear). While the use of algorithms allows for automation of contact event identification and therefore saves time for the investigator, the method of manually inspecting video footage is no less accurate and simple to perform. For the purpose of the biomechanical analysis performed in this research a cut-off magnitude of 15N was chosen throughout for foot contact determination.
3.10 Force plate targeting

Challis (2001) investigated the influence of deliberately changing the normal gait pattern to make contact with the force plate. This modification of the gait pattern is called targeting. Seven experienced and regular male runner (height, 1.77 m ± 0.08 m; mass 72.4 kg ± 7.5 kg; age 23 to 32 years) took part in 3 different running protocols. The first objective was to perform a heel-toe run at 3.2 ms⁻¹ (± 5%) over a 16 meter long track where the starting position was adjusted to insure that the participant would strike the force plate without targeting it. In conditions 2 and 3 the starting position was shorted and elongated by 50cm respectively. This change in run up required the participant to actively target the force plate to ensure that the foot made contact.

The results of the investigation showed changes in peak impact forces and their timings when the participant targeted the force plate, peak active (at 45% contact time) forces however, were found to be invariant. Differences were found between short and long steps, but not between these and the normal condition. The overall outcome was that ground reaction forces in human running are consistently repeated regardless of force plate targeting and the differences in running kinematics and kinetics during the stance phase seem to occur early on and few differences occur during the later stance phases. Inadvertent targeting may therefore be acceptable depending on the parameters and phases of the contact phase of interest.

3.11 Summary

This section concerned the methods used for data gathering and processing for the methods used for the research performed in this thesis. All the methods used in this thesis have been validated. Furthermore, they have proved appropriate to the application and the methods are transferable to other applications.
Chapter 4

Video-based classification, quantification and comparative analysis of gait usage in badminton

4.1 Introduction

Badminton is an energetic and multifaceted sport with a strong national and international status. It combines relatively simple game play with small demands in terms of equipment and space and emphasises technique over strength alone. The popularity of the sport has been furthered since its admittance as an Olympic discipline in 1992 (Fahlström et al., 2002, Jørgensen and Holmich, 1994). Badminton however is not a simple sport but with increasing level of competition the game becomes physically more challenging (Docherty, 1982). The game relies heavily on tactics, which is reflected by the large variety of shot placement: drop shots to the opposite front and mid-court area; clears to the back court; smashes and drives to the mid- and rear court and net shots to the front court as well as a variety of other shot derivatives (Badminton England, 2005, Downey, 1993). This forces the player to cover and move to all areas of the court. As a consequence of these movement requirements the player adopts a tactical 'base' position from which movements to the other areas of the court are initiated (Badminton Association of England, 2005). This base, while being flexible to adapt to different game situations, is predominantly in the centre of the mid-court area and players are required to perform movements in any direction of the court at any moment within a rally. This is in contrast to for example tennis where O'Donoghue and Ingram (2001) show that baseline play, lateral movement in the rear court area, is the predominant game type in elite level tennis, particularly for female performers in the Australian, French and US Open.

The positioning of the player on the badminton court requires unique movement strategies to allow for efficient multidirectional progression, instead of predominantly
two directional sideways movements. Particularly high level of play requires efficient use of movement and as a consequence atypical gaits such as sidestepping and crossover stepping are incorporated into training (Badminton Association of England, 2005). The extent of use of these and other movements in competitive games however is not known.

A method that has been used to quantify the occurrence of movement or shot-making events in sports is notational analysis. Lees (2003) defines notational analysis as "the process of recording and analysing the movements made by players during play (...). The data collected are related to the position, action, time and outcome of an event in the game; (...)." Notational analysis results in significant amounts of data for analysis and in sports, has provided useful information for the assessment of game dynamics, demands and tactics.

So far the use of notational analysis in badminton has focused on the description of the temporal characteristics of the game, shot selection and success and failure rates. These studies have displayed interesting differences between male and female players Liddle and O'Donoghue (1998), while Hughes (1986) identified distinct differences between players of different levels of skill in the game of squash. While the investigations into rally durations, shot making and physiological demands are valuable in terms of planning a game, teaching game strategy and general training for the metabolic requirements of the game, they focus on the racket-shuttle interaction and contain no information on the selection of gait within a competitive game. Movement on the court however, is an essential part of the players' ability to play a shot. Efficient movement allows the player to intercept the moving shuttle sooner, which thereby allows for the selection of a larger variety of shot-making options. The larger the available shot repertoire, the more flexible the game strategy becomes, which gives the player an advantage in trying to win the shot. Hughes and Robert (2005) examined the effect of movement patterns in tennis, displaying the contribution of sidestepping to the movement routines. They state that the analysis of footwork and their patterns is important in order to determine what exactly happens in a game. However, no information relating to the application of movement in badminton is available.
The objective of this investigation is to quantify the range of movement employed by
the athlete in competitive badminton and therefore to identify the contribution of both
typical and atypical movements within rallies. Furthermore, based on the observations
of changes in the temporal dynamics of competitive singles badminton due to gender
and level of skill, both male and female players participating in national and
international players are analysed in order to verify the contribution of atypical
movement to the respective games. A higher contribution of atypical movements in
male and international athletes may help to explain the observation of a higher risk of
lower limb injury in these groups. It is therefore the aim of this investigation to use a
notational analysis approach in order to:

1. Identify and quantify the major forms of locomotion within the competitive
badminton match scenario and to identify differences in these parameters due
to differences in playing standards and gender within the context of male and
female national and international singles players.

2. Measure the number and duration of individual movements (bout numbers and
durations) to sample the temporal characteristics of these movements within
the game situation.

3. Investigate the temporal characteristics of match play, durations of activity
(rally) and rest, at the different standards of play in order to shed light on the
effect of skill and gender on these parameters.

The findings from this investigation will aid in our understanding of the movements
employed in competitive badminton and the nature of their application. This will shed
light on the differences in the movement repertoires used by male and female,
national and international competitive badminton players and aid in our interpretation
of the physical demands at the national and international level.
4.2 Methods

4.2.1 Participants

With approval by the local ethical committee, video recordings of 6 international male, 4 international female, 6 national male and 5 national female competitive singles games were recorded and analysed. Male and female national games were recorded, without interruption, for nationally ranked University student players (1st and 2nd team) during either BUSA (British University Sports Association) or County level tournaments. Verbal consent of the subjects was obtained prior to commencing with the recordings. These games were chosen to represent consistent national level of play. International level of play was investigated by examining televised footage of international players from World Championships, Olympic Games and international tournaments during the past 10 years. It was anticipated that the selection of only top level international competitions would minimize a possible influence of changes in player profiles within this time frame. The selection criteria for the international video footage were that the recording was continuous, not including shortened highlights footage, and consisted of complete sets. In some cases both players participating in the match were analysed, resulting in gait data for a total of 10 male international, 8 female international, 8 male national and 8 female national individuals (see Table 4.1 for a summary of the subject information).

<table>
<thead>
<tr>
<th>Table 4.1: Summary of subject and video information</th>
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<tr>
<td><strong>Games Analysed</strong></td>
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<tr>
<td>Male International</td>
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<td>Female International</td>
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<td>Male National</td>
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4.2.2 Video Analysis

Matches involving University students were taped using a digital video camera (Panasonic NV-DS27) while international footage was stored on VHS or DVD. All video footage was in the European encoding system PAL (Phase Alternating Line) with a resolution of 25 frames per second which helped to ensure consistent video and timeline matching during later analysis. Video footage of national and international games was transferred and saved to a Microsoft Windows® based PC as an uncompressed AVI file using Virtual Dub (freeware video editor by Avery Lee, http://www.virtualdub.org/index) or Power Director Pro (CyberLink Corp., http://www.cyberlink.com/). Only continuous recordings, not including shortened, highlights footage, was used in order to avoid the creation of bias towards activities during rallies within the data.

Notational analysis of the digitised video footage was performed using focal sampling techniques of the continuous recordings following recommendations by Martin and Bateson (1986) (see Chapter 3.2). Virtual Dub was used for frame-by-frame replay of the matches to allow for the highest possible detail in visual identification of gait events, bout durations and activity (rally) and rest periods (where a bout is equivalent to the total time used to perform a distinct movement). The resulting values for bout numbers, total duration a movement is performed for which can include several steps, and rally/rest durations were used as the basis for the subsequent statistical and descriptive analysis. Due to the nature of the games the duration of sampling varied according to game duration. A minimum of eight minutes of a set, and one set per participant, was examined for the analysis of bout durations and numbers, and one complete set per participant was analysed for the investigation of set characteristics (periods of activity and rest during a set). This minimum timeframe reflected the generally shorter set durations of the female national game and for the purpose of comparing male and female set characteristics similar durations were chosen. It was thought that this timeframe would correspond to a typical selection of activity during a set and typical gait use for the different performance levels within the game.
4.2.3 Gait classification

Gaits were identified using basic classification schemes where available and custom classification schemes where no predefined classification information could be found in the literature, such as sidestepping or crossover-stepping. The major classification schemes were:

Standing: No active progression of the body in any plane
Walking: Anterior or posterior progression of the body with a distinct double support phase of the body by the lower limbs
Running: Anterior or posterior progression of the body with single limb support and an identifiable aerial phase (no limb in contact with ground)
Lunging: A weight transfer to a bent leg (predominantly the dominant limb) with the other leg extended posteriorly
Sidestep (SS): Lateral progression of the body consisting of typically 4 phases per gait cycle depending on movement magnitude (see Figure 4.1)

Figure 4.1: Posterior view of a typical sidestepping gait. Red and blue limbs represent the dominant (D) and non-dominant limb (ND) respectively. The word impact is used loosely to denote contact of the limb with the ground as this can occur using the heel or the mid/forefoot.

Phases (numbered 1-4):

1. Wide stance, gait cycle initiated by dominant foot impact;
2. Non-dominant toe-off followed by movement of the non-dominant limb along the medio-lateral plane toward the dominant limb;
3. Dominant limb toe-off followed by brief aerial phase and movement of the dominant limb away from the midline;
4. Non-dominant foot impact and continued lateral swing of dominant limb;
1. Dominant limb toe-strike, completing the gait cycle.
Crossover step (XS): Similar to the sidestep with the non-dominant limb extending beyond the midline of the body, crossing over behind the weight bearing, dominant, limb (see Figure 4.2)

Figure 4.2: Posterior view of a typical crossover stepping gait. Red and blue limbs represent dominant and non-dominant limbs.

Phases (numbered 1-4):

1. Wide stance with dominant limb impact initiating the gait cycle;
2. Toe-off and adduction of the non-dominant limb beyond the midline of the body and beyond the weight bearing limb;
3. Dominant limb toe-off followed by non-dominant impact resulting in a brief aerial phase, the magnitude of which depends on the speed locomotion;
4. Dominant limb lateral swing until end of single support by the non-dominant limb, initiated by either dominant limb impact or non-dominant limb toe-off;
1. Dominant limb impact completes the gait cycle.

Periods where participants were not visible, for example during instant replays in public broadcast footage, were accounted for by registering these times as either out-of-view (OV) or out-of-view during periods while resting (OVR). OVR contributed occurred more frequently, due to instant replays in some international games, while OV periods during play was a rare occurrence. These interruptions were unavoidable as they were a direct consequence of the official broadcaster either interrupting periods of rest with instant replays of the previous rally or changing camera angles to focus on individual players. In order to avoid the occurrence of these out-of-view events the request was made to obtain uninterrupted raw broadcast footage from the BBC or record games at the 2005 Yonex All England international badminton championships. However both requests were denied and it was decided to proceed with the available televised footage.
4.2.4 Statistical Analysis

Statistical analysis of the effects of skill and gender on rally and rest durations as well as bout duration characteristics was performed in the statistical analysis tool SPSS®. ANOVA tests of variance were performed to identify the effects of skill and gender on rally/rest distribution and duration characteristics. Effects of skill and gender on gait characteristics were investigated using non-parametric Mann-Whitney tests. Statistical significance was taken as p<0.05 throughout.
4.3 Results

4.3.1 Match Characteristics

Total rally (activity during an unbroken sequence of shots) and rest (time spent between rallies) periods suggested a general 35 – 65 % distribution of rally and rest across all abilities under investigation (Figure 4.3 a). Gender was seen to have an effect at the international level only (p<0.05 and p=0.98 for international and national players respectively) with male players spending less time in rallies and more time resting than female players (Figure 4.3 b). Skill was only observed to have an effect for male players (p<0.05 for male international and national players and p=0.33 for female international and national players) with male international players spending a smaller percentage of the overall game in rallies and a higher percentage resting then national male players.

Figure 4.3

a) Overall rally and rest distributions for male and female, international and national players. During the observed time period about 35% of the time was spent in rallies and about 65% resting. Error bars represent standard deviations of the sampled data points.

b) Significant effects of gender within international players and of skill within international and national male players on rally and rest distributions were observed. International male players spent a smaller fraction of the total game in rallies and a larger fraction of time resting then either female international or national male players.
Mean rally and rest durations (Table 4.2) lasted about 5-10 and 10-20 seconds respectively and the values were in line with those reported by Liddle and O’Donoghue (1998), Cabello (2003) and Pearce (2002). Significant effects of gender and skill on mean rest durations (p<0.05) and of skill on mean rally durations (p<0.05) were observed where international players were involved in both longer rally and rest periods than national players (see Figure 4.4) and male international players took longer rest compared to female international players. No significant differences in mean rest durations due to gender were observed for national players. These finding are in line with those by Liddle and O’Donoghue (1998) who reported longer rally and rest periods in male players however, no significant difference in rally times were observed in this study.

<table>
<thead>
<tr>
<th>Table 4.2: Mean rest and rally durations for the four groups tested.</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Rally Duration (sec)</strong></td>
</tr>
<tr>
<td>---------------------------</td>
</tr>
<tr>
<td>Mean</td>
</tr>
<tr>
<td>Male Experienced</td>
</tr>
<tr>
<td>Female Experienced</td>
</tr>
<tr>
<td>Male Inexperienced</td>
</tr>
<tr>
<td>Female Inexperienced</td>
</tr>
</tbody>
</table>

Figure 4.4 Effect of skill and gender on mean rally and rest durations. Skill was seen to have a significant effect on both rally and rest durations whereas gender was seen to have a significant influence on rally durations only at the international and not the national level.
Further breakdown of rally and rest durations into time clusters showed that their distribution differed between the four groups under investigation (Figure 4.5). Significant effects of both gender and skill were identified. Male participants were involved in significantly more short duration rallies, while women were involved in a higher percentage of medium duration rallies. Women were also involved in a significantly higher percentage of medium duration rest. International players were involved in significantly more long duration rallies (>10 sec) than national players and took longer rest between rallies.

![Figure 4.5: Graph displaying the breakdown of rally and rest durations (expressed as % total number of rallies and % total periods of rest) across the investigated groups of international and national student players, into sections of: <5, 5-10, 10-20 and >20 sec for rally durations and <10, 10-20, 20-30 and >30 sec for rest durations. A higher proportion of long duration rallies could be observed in the international players with predominantly short duration rallies in the group of national players. Furthermore, international players displayed rest times situated predominantly in the region of 10 to 20 seconds and above. Student players on the other hand showed a tendency for shorter duration rest periods, in the 0-10 and 10-20 second region with few examples of prolonged recovery.](image)

4.3.2 Gait Characteristics

The identifiable gaits in the badminton footage included standing, walking, running, jumping, lunging, hopping, skipping, sidestepping (SS) and crossover stepping (XS) as well as a variety of single step movements. Of these gaits eight classes were used for analysis: stand, walk, run, SS, XS, lunge, jump and scramble (including hopping and skipping as well as a variety of other movements that were deemed unclassifiable). The resultant data is summarised in Table 4.3.
The data showed high bout numbers of walking and standing. While walking and standing were not seen to be performed during rallies they were prominent during rest phases and their high numbers reflect the large percentage of rest periods in the game. The relatively low numbers of walking and standing in the international group are assumed to be due to the higher number of OVRs. From the observations of the video footage it was assumed that the player performed walking and standing during prolonged periods of OVR, as he/she returns to the service area and waits for the next rally to start. For the purpose of analysing movements during rallies, walking and standing as well as OVR and OV were subtracted from the bout number total to focus on the distribution of those gaits that are thought to be predominantly used within rallies (i.e. run, lunge, SS, XS, jump and scramble).

<table>
<thead>
<tr>
<th></th>
<th>International Mean %</th>
<th>SD</th>
<th>International Mean %</th>
<th>SD</th>
<th>National Mean %</th>
<th>SD</th>
<th>National Mean %</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>9.46</td>
<td>1.41</td>
<td>9.09</td>
<td>3.28</td>
<td>20.18</td>
<td>3.56</td>
<td>16.18</td>
<td>3.39</td>
</tr>
<tr>
<td>Stand</td>
<td>9.38</td>
<td>2.38</td>
<td>9.32</td>
<td>3.46</td>
<td>18.58</td>
<td>2.73</td>
<td>16.50</td>
<td>5.18</td>
</tr>
<tr>
<td>Walk</td>
<td>6.23</td>
<td>2.92</td>
<td>9.93</td>
<td>2.87</td>
<td>4.42</td>
<td>4.29</td>
<td>6.98</td>
<td>5.49</td>
</tr>
<tr>
<td>Run</td>
<td>14.66</td>
<td>3.39</td>
<td>12.32</td>
<td>4.34</td>
<td>8.52</td>
<td>3.10</td>
<td>9.92</td>
<td>3.70</td>
</tr>
<tr>
<td>Lunge</td>
<td>14.39</td>
<td>5.08</td>
<td>16.91</td>
<td>5.72</td>
<td>11.92</td>
<td>3.23</td>
<td>18.21</td>
<td>4.30</td>
</tr>
<tr>
<td>SS</td>
<td>6.64</td>
<td>1.44</td>
<td>6.34</td>
<td>2.78</td>
<td>2.96</td>
<td>2.52</td>
<td>7.11</td>
<td>4.88</td>
</tr>
<tr>
<td>XS</td>
<td>5.39</td>
<td>1.75</td>
<td>2.70</td>
<td>2.95</td>
<td>4.63</td>
<td>1.90</td>
<td>0.52</td>
<td>0.69</td>
</tr>
<tr>
<td>Scramble</td>
<td>30.18</td>
<td>3.20</td>
<td>28.60</td>
<td>7.86</td>
<td>27.71</td>
<td>6.91</td>
<td>24.45</td>
<td>7.85</td>
</tr>
<tr>
<td>OV</td>
<td>1.24</td>
<td>1.48</td>
<td>1.26</td>
<td>1.51</td>
<td>0.66</td>
<td>1.20</td>
<td>0.13</td>
<td>0.27</td>
</tr>
<tr>
<td>OVR</td>
<td>5.43</td>
<td>4.70</td>
<td>3.54</td>
<td>3.97</td>
<td>0.42</td>
<td>0.87</td>
<td>0.00</td>
<td>0.00</td>
</tr>
</tbody>
</table>

Mean bout numbers are expressed as a percentage of the total bout numbers of all recorded locomotor classes within the individual and averaged across participants within the group.
The bout numbers of gaits used in rallies (Table 4.4) showed the large contribution of scrambling movements to the overall use of gait. SS contributed about 22% to the overall use of gait within rallies, followed by lunging, running, XS and jumping. The bout numbers of the gaits used within rallies were affected by both gender and skill. Gender was seen to significantly affect SS, XS and jump numbers with female players performing more SS and XS movements and male players performing more jumps. Skill had a significant effect on lunge numbers with international players performing more lunges than national players (see Figure 4.6).

![Graphs](image)

Figure 4.6 Graphs representing the effect of gender and skill on bout numbers of gaits used in rallies, where rallies refers to total sum total of rallies performed within the sample period. Statistically significant findings (*) were observed for the effect of gender on SS, XS and jump numbers and of skill on lunge numbers within international players.
Variation in the use of gait was seen to be lower in the group of international players compared to the national players (Figure 4.7). The group of international males displayed a very consistent distribution of gait use. Female international players displayed similarly consistent use of movement with a slightly larger proportion of lateral movements. Jumping can be seen to be performed consistently more frequently in male participants while the use of scrambling movements were generally consistent in international players with slightly larger variation in national players.

![Figure 4.7 Visualisation of inter-individual variation in the use of running, sidestepping, crossover stepping, lunging, jumping and scrambling movements within international, national, male and female badminton players. Values expressed as % of total bout numbers.](image)

The bout durations of the investigated gaits were generally short with standing and walking displaying the longest times (see Table 4.5). The data show gait use in badminton to be dominated by short duration movements.
<table>
<thead>
<tr>
<th></th>
<th>Male International</th>
<th>Female International</th>
<th>Male National</th>
<th>Female National</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean</td>
<td>SD</td>
<td>Mean</td>
<td>SD</td>
</tr>
<tr>
<td>Stand</td>
<td>5.17</td>
<td>0.96</td>
<td>4.37</td>
<td>1.67</td>
</tr>
<tr>
<td>Walk</td>
<td>8.72</td>
<td>5.05</td>
<td>5.09</td>
<td>0.73</td>
</tr>
<tr>
<td>Run</td>
<td>1.31</td>
<td>0.13</td>
<td>1.44</td>
<td>0.28</td>
</tr>
<tr>
<td>SS</td>
<td>0.81</td>
<td>0.06</td>
<td>1.08</td>
<td>0.54</td>
</tr>
<tr>
<td>XS</td>
<td>1.02</td>
<td>0.12</td>
<td>1.26</td>
<td>0.48</td>
</tr>
<tr>
<td>Lunge</td>
<td>1.02</td>
<td>0.10</td>
<td>1.54</td>
<td>1.35</td>
</tr>
<tr>
<td>Jump</td>
<td>0.95</td>
<td>0.08</td>
<td>1.44</td>
<td>1.37</td>
</tr>
<tr>
<td>Scramble</td>
<td>0.87</td>
<td>0.13</td>
<td>1.88</td>
<td>1.24</td>
</tr>
</tbody>
</table>

Table 4.5 Mean Bout durations of standing, walking, running, SS, XS, lunging, jumping and scrambling movements.
4.4 Discussion

The objective of this investigation was to quantify the range of movement employed by male and female, national and international level athletes in competitive badminton and therefore to identify the contribution of both typical and atypical movements within these groups. The aims of this investigation therefore were to identify and quantify the major forms of locomotion within competitive badminton matches and to measure the number and duration of these movements (bout numbers and durations) to sample the temporal characteristics of in-game gait use. The temporal characteristics of activity and rest were measured to describe the temporal dynamics of activity in games.

Before discussing the results of the study in more detail some restrictions of the experiment need to be discussed at this point. As mentioned in the methods section the international footage chosen for investigation consisted of televised badminton footage. While this footage was not edited in its original length but represented live matches, it did include instant replays of previous rallies and changes in camera angles, focusing on individual players, during periods of time where the players were resting. While this has an undoubted effect on the distribution of walking and standing numbers and durations reported within the international game, it did not affect the timing of rally and rest durations or the activities during rallies which were uninterrupted. Since the attempts to secure uninterrupted footage of international competitions were unsuccessful this limitation to the study was unavoidable. A further limitation was the short duration of movements used in the game and the nature of gait use. Because movements were used in rapid succession identification of individual durations was not always possible. This was true particularly for run/lunge and SS/lunge or XS/lunge movements. This was exaggerated by the confinement of movement to a very restricted space. The knock-on effect of this was that many movements did not always consist of a complete gait cycle but instead consisted of only single steps (half a gait cycle), which led to the inclusion of the scramble category containing a variety of non-classifiable movements or those deemed to be non-essential to the study, as they occurred only sporadically, such as hopping, single steps in any direction, or skipping.
4.4.1 Rally / Rest: Temporal Characteristics

The results of the examination of the temporal characteristics of the four groups of international and national, male and female badminton players showed that the general distribution of activity and rest was similar between the groups with about 35% of the total match spent in rallies and 65% spent resting. There was evidence of the effects of both skill and gender on the performance of the game. Male international players appeared to spend a smaller fraction of time in rallies and more time resting than female international or male national players. In terms of the durations of rallies and rest a division due to skill becomes obvious. International players were seen to be involved in significantly longer rallies than their national counterparts and subsequently also used more time recovering from the previous rally.

This is in support of the findings by Docherty (1982) who reports increasing heart rates with increasing skill of the participant. Gender played an important role in the international game with male players being involved in slightly shorter rallies and significantly longer rest durations than female international players. This difference however was not seen to play a significant role in the national game. The data therefore substantiates findings by Liddle and O'Donoghue (1998) who reported significant effects of gender on rest durations, where male players take longer breaks between rallies. However in contrast to their observations of longer rallies in elite men's compared to women's singles, no significant differences in mean rally times due to gender were observed in this study for international or national players. The influence of higher skill levels, causing both longer rally and rest periods, had previously been reported by Hughes (1986) for recreational, county and international squash players and seems to apply equally to badminton. In squash technical ability results in longer rallies as the player is able to enforce their game plan. County and recreational players however are more erratic in their shot making and can therefore only implement more simplistic tactics which, coupled with their lower fitness levels, causes a shortened rally. In this study, international players were more likely to be involved in longer duration rallies than national players and male participants were more likely to play short duration rallies, while women were involved in a higher percentage of medium duration rallies and rest. At the same time, this increase of
activity lead to increased rest phases as the athlete recovered from the previous rally. It is however important to remember that rally and rest characteristics are very much dependent on the individual game and discrepancies in rally and rest timings have been identified even within the same level of play (Lees, 2003). Care must be taken in the definition of skill and there may well be large differences even within national or international level players. Furthermore, environmental factors such as temperature, humidity or court surface may play crucial roles in determining the speed of the player and shuttle and create knock-on effects on the duration characteristics of the game. While court surfaces at the international level have now become very consistent this is not the case at the national level where a variety of different surfaces, including rubber, foam, carpet, wood or concrete, can be used and the quality of the court may not be maintained adequately. While there is no information on the effects of court surface in badminton it is thought that changes in friction may well affect the temporal characteristics of the game as well as the applied movement repertoire to reduce the risk of slipping. Furthermore, the use of high friction surfaces in tennis has been related to an increased occurrence of overuse injury (Renström and Roux, 1988). This would appear to be relevant to the game of badminton where high friction indoor courts are often used.

4.4.2 Gait Characteristics

The considerable variety of movements identified in the video footage, suggest a great diversity of movement strategies during the competitive game situation. Scramble movements can be seen to contribute to a large extent to the game. Scrambles include a variety of movements that are not easily classified and mainly used to adjust the positioning of the body before a specific movement is initiated. These include, for example split steps, as seen in tennis (Chu and Rolley, 2001) and hopping to maintain dynamic rather than static balance (Dent and Hagelauer, 2001). Movements used to travel for longer distances involved the use of a large number of atypical movements, particularly lateral movements (SS and XS) and lunging, within a rally. The purpose of movement in badminton is to aid in the ultimate goal of positioning the body on court in such a way, as to allow the player to utilise their entire shot repertoire. While data on the direction of movement was not included in this study it should be noted
that SS movements were commonly used in conjunction with a lunge finish and XS was often used in an attempt to cross the court diagonally to the back in order to reach a high clearing shot to the back of the court and was regularly performed in conjunction with a jump. Therefore, a distinct characteristic of many of the gaits used within the game scenario was their inclusion in and combination with other movements, especially the jumping and lunging movements, which explains the short bout durations of the individual on-court movements.

Because of the restrictions in space and the structure of the shuttle cock, being made of feathers that slow down its speed, the game, despite the fast shuttle speeds that can be achieved, relies heavily on outmanoeuvring the opposition. Players of equivalent high standard will be able to return the shuttle unless forced into a disadvantageous position. The significantly higher number of lunges at the international compared to the national level fits in with this idea and the previously mentioned influence of technical ability on shot placement. The lunge is used as the primary method of stopping from lateral or anterior progression of the body within the game, in order to stop and recover quickly to the base position. It can be assumed that the higher numbers are a consequence of the ability of international players to move the opponent around the court. This is in line with observations in squash, where elite players were employed an ‘all-court’ game, based on more complex tactics, due to their higher skill levels and ability to successfully play a shot. County standard players on the other hand utilised a far simpler game plan (Hughes, 1986).

Within national players SS movements were seen more often in female players and overall SS were performed more regularly than XS. The exact reasons for the higher number of SS within female national players are unclear but the higher number of SS is a clear reflection of the more varied use of the movement within the game. While the XS was used predominantly for movement in the direction of the dominant limb, and particularly for lateral movements to the back and less frequently the front court, the SS was used more frequently as a lateral movement in all directions of the court. It can be argued that the female game is less reliant on power shots such as the jump smash but relies more on a strategic positional approach. This is supported by the longer duration of rallies and the lower number of jumps in the female game. Jumping was seen to be more prominent in male players within international and national
players. Jumping is a powerful movement used to intercept the shuttle early in flight, create steeper angles and increase shot power when smashing the shuttle. Jumping therefore is an aggressive move and the greater number of jumps in the male game is thought to reflect the importance of a pressure and attack game, particularly in male international players (Hong and Tong, 2000).

As described in the literature review, investigations into the nature and occurrence of injuries within the game (Fahlström et al., 1998, Fahlström et al., 2002, Høy et al., 1994, Jørgensen and Holmich, 1994, Jørgensen, 2002, Lees, 2003, Mohtadi and Poole, 1996) have linked badminton, in comparison to a number of other sports, to a high incidence of overuse injuries of the lower limbs in both professional and non-professional players with relatively few acute injuries. These however tend to be more severe than in other racket sports and tears of the Achilles tendon in older players and joint and ligament damage, predominantly of the lower limbs, are amongst the leading injury factors. The results from this study show that badminton players are in constant motion even when not playing a shot, when the player is either recovering from the previous shot or preparing to respond to the next shot, and use a variety of atypical movements throughout the game. The high occurrence of atypical movements identified in this study may therefore be a potential contributing factor to the occurrence of these injuries.

4.5 Conclusion

Understanding the influence of both skill and gender on the nature and application of the movement repertoire and the temporal dynamics of a match provides useful objective information for the athlete and the coach in order to prepare efficiently for the physical requirements of the national and international competitive game. Future studies may combine directional data with gait use to describe the movement patterns. Furthermore, investigations into other game modes, such as straight or mixed doubles, would supply essential data on the changes in demands between game modes and enable game specific training programmes. If combined with information on the success and failure rate, as proposed by Hughes (2005) for tennis, information on movement strategies would provide a more global view of the contribution of
movement within the game. This would result in very useful objective information, not only on the success of certain tactics but also the benefits of specific movement patterns and combinations. Information of this kind may enable a player to fine tune their game plan before and during a match.
A biomechanical analysis of the stance phase of a standardized sport-specific sidestep and crossover step task

5.1 Introduction

The information presented in Chapter 4 established that competitive badminton is characterised by the application of a variety of movements. Walking, running and lateral gaits are common, where lateral gaits can be viewed as being atypical modes of bipedal locomotion as they are not commonly seen in normal adult gait. However, within the sport scenario they perform an important function. The observations made in the notational analysis indicate that, in badminton, lateral gaits allow for movement of the body toward the shuttle, while keeping the body rotated toward to opposite court and minimizing the time to arrive at the shuttle. This has the further advantage of putting the upper body in a favourable position for the action of the racket arm during the stroke process (Tang et al., 1995, Badminton Association of England, 2005, Badminton England, 2005). The distinguishing feature of lateral sidestepping (SS) and crossover stepping (XS) gaits in badminton, therefore, is their linear application.

The literature review showed that there is an apparent lack in our knowledge related to the biomechanics of the leading and particularly the trailing limb during the performance of lateral movements. This is surprising considering the apparent importance of lateral movements to badminton and tennis (Hughes and Robert, 2005). The current badminton coaching literature recommends the use of these gaits for direct movement of the player to, for example, the front or rear corners of the court, as well as a means of moving back into the court from the front or rear court areas.
This application of the lateral movements distinguishes them from the previously studied sidestep and cross cutting manoeuvres, which have received a lot of interest due to their effects on the ankle and knee joints and their relevance to ligament injuries in sports such as basketball and in relation to female athletes in particular (McLean et al., 1999, McLean et al., 2004b, Pollard et al., 2004, Sigward and Powers, 2006). These cutting movements are essentially evasive manoeuvres following, for example, a walk or run. In badminton however, the player is rarely in a position where sudden directional changes are necessary. Thus the temporal movement dynamics in badminton tend to allow for planned linear motion, starting from a base position and moving along a single line of travel.

As mentioned above, there is a distinct gap in our knowledge relating to the bilateral biomechanics of linear SS and XS movements. No information relating to the biomechanics of linear SS or XS gaits were found. Vertical reaction forces of the leading limb (e.g. the right limb in a SS movement to the right, referred to as the dominant limb in Chapter 4) during a SS shuffle have been reported for the stance phase of the leading limb when changing direction from a right hand to left hand sidestep movement (Dayakidis and Boudolos, 2006, McClay et al., 1994b, Neptune et al., 1999). Furthermore, kinematic and electromyographic profiles of the leading limb for a similar movement have been reported (McClay et al., 1994a, Neptune et al., 1999). However, the contribution of the leading and trailing limb (referred to in Chapter 4 as the non-dominant limb), during a continuous linear SS or XS remain unknown. Importantly, in the context of badminton, there appears to be no information on the benefits of applying these movements on-court. Therefore it is largely up to the individual player to choose between SS or XS based on personal preference.

According to the epidemiological data for badminton injuries, presented in Chapter 2.4.2, as well as the injury-related interest in SS and XS cutting movements, it may be argued that the lateral movements used in badminton contribute, at least in part, to the lower limb injury patterns observed in badminton. Furthermore, based on the interest by a number of authors in the effect of gender on the biomechanics of the lower limbs during running, cutting movements and landings (McLean et al., 1999, Decker et al.,
2003, Ferber et al., 2003, Mclean et al., 2004b, and Pollard et al., 2004), and an increased risk of injury in male badminton players (Jørgensen and Winge, 1987), it was thought that differences in lower limb biomechanics of lateral SS and XS gaits between male and female participants may be observable. Similarly, as these lateral movements are not commonly used in everyday situations, it was considered that the increased exposure time to lateral movements by experienced badminton players may affect the kinematics and kinetics of these movements. It was therefore decided to investigate both the effect of skill and gender on the biomechanics of the SS and XS stance phase.

It was the objective of this investigation to quantify the magnitude of ground reaction forces and joint kinetics the lower limbs are exposed to during the performance of lateral SS and XS gaits. Large ground reaction forces and joint moments compared to running may be regarded as contributors to an increased risk of injury for the participant. Based on the literature it was thought that experienced participants may show adaptations to the movement technique which may aid in minimising potential injury risk. The purpose of this investigation was therefore, to describe the gait processes of a standardised lateral SS and XS manoeuvre in the lab to assess the demands of the movements on the athlete. The aims were to:

- Record ground reaction forces for the leading and trailing limb during the SS and XS to quantify force exposure and identify differences between the limbs and gaits.
- Calculate joint kinetics using an inverse dynamics approach to assess the moments and powers acting on the joints of the lower limbs during the SS and XS.
- Indirectly calculate Achilles tendon loading to evaluate the demands of the movements on this at-risk structure.
- Compare the data from this experiment with the available literature on walking, running and skipping to assess the similarities and differences between these movements.
- Identify the potential differences in reaction force and kinetics due to gender and ability level.
This information will provide valuable insight into the mechanisms involved in lateral movements, as applied to the sport of badminton, and will allow for a comparative assessment of the demands of these movements on the musculoskeletal system of the participant. Understanding these demands should provide useful information of the advantages of gait use in badminton and valuable information for the coaching of movement skills in the game.
5.2 Methods

5.2.1 Participants

With approval by the local ethical advisory committee 26 participants (8 male and 6 female University 1st and 2nd team badminton players (Squad); as well as 6 male and 6 female recreational badminton players (Rec) were recruited from the Loughborough Students' Badminton club (see Table 5.1). Written informed consent for all participants was obtained before testing and the participants were informed of the general aims and requirements of the study. Anthropometric data was collected at the beginning of the study for all participants for knee and ankle width and hip width and depth as described in Chapter 3.4. This subject-specific data was necessary for the inverse dynamics calculations of joint moments and powers as outlined in Chapter 3.7. Three dimensional lower limb kinematics and ground reaction force data were thereafter recorded for the participants during the performance of SS and XS lateral movements. Only participants with no history of orthopaedic injury to the lower extremity joints and some experience in performing SS and XS movements were included in this study.

Table 5.1 Summary of participant information.

<table>
<thead>
<tr>
<th>Experienced Participants</th>
<th>Experienced (squad)</th>
<th>Male (n=8)</th>
<th>Female (n=6)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Players were classed as experienced if they played considerably more than 6 hours of badminton per week and played for the University 1st or 2nd team. Level of expertise varied from national to international level.</td>
<td>Age (yrs)</td>
<td>21.13</td>
<td>2.64</td>
</tr>
<tr>
<td></td>
<td>Height (m)</td>
<td>1.78</td>
<td>0.07</td>
</tr>
<tr>
<td></td>
<td>Total Body Mass (kg)</td>
<td>73.15</td>
<td>11.83</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Inexperienced Participants</th>
<th>Inexperienced (rec)</th>
<th>Male (n=6)</th>
<th>Female (n=6)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Players were classed as inexperienced if they played no more than 6 hours of badminton per week and at no higher than average recreational (rec) level.</td>
<td>Age (yrs)</td>
<td>19.5</td>
<td>2.74</td>
</tr>
<tr>
<td></td>
<td>Height (m)</td>
<td>1.81</td>
<td>0.06</td>
</tr>
<tr>
<td></td>
<td>Total Body Mass (kg)</td>
<td>72.72</td>
<td>11.39</td>
</tr>
</tbody>
</table>

5.2.2 Experimental Design

The experiment was conducted in one session lasting a maximum of 1.5 hours. Upon filling in a general health form (Appendix A1.1), signing the informed consent form
(Appendix A1.2) and collecting the anthropometric measurements the participant was equipped with the lower limb wand system supplied by Charnwood Dynamics (described in detail in Chapter 3.6), which acted as a base unit for marker attachment. Participants were asked to wear the shoes they would normally wear when playing badminton and no attempt was made to control for shoe make. Thereafter, participants were asked to perform a regular 5 minute warm-up including a series of low and high intensity SS and XS. During data recording the gait tasks were performed on a raised wooden walkway (8.4 x 1.2m) with the cameras located at either ends of the walkway. For the purpose of this study, only a self-selected, preferred speed was investigated. The available lab-space and walkway length were restricted and preliminary testing indicated that faster speeds may result in an increased risk of tripping for the participant. Furthermore, there was not sufficient space to allow for the use of a mat at the end of the walkway that the participant could use to stop the movement safely. Such a method would, furthermore, have likely resulted in damage to the testing equipment. The preferred speed was determined during a familiarisation period where the participant performed a series of SS and XS movements on the walkway. The familiarisation period ended once both the participant and investigator were satisfied with the correct understanding and performance of the required tasks and the participant was asked to try and maintain the determined exercise intensity throughout the experiment. After determining the optimal starting position for the participant to assure foot contact with the force plate without targeting the force place, the subject was provided with a badminton racket (mass = 90 g). Participants were asked to keep their arms in front of their chest and maintain this position throughout the movement. This method provided a more natural recreation of the in-game situation and assured unrestricted view of the markers by the cameras throughout. Data sampling was performed until ten successful task repeats were collected for leading leg SS and XS (SS_L and XS_L) and trailing leg SS and XS (SS_T and XS_T) for a total of 40 successful gait repeats. A successful movement consisted of the participant hitting the middle of the forceplate with the foot of the leading or trailing leg during the second stride, following the process in (Neptune et al., 1999), with full marker visibility during the support phase.
5.2.3 Data Collection

Ground reaction forces (GRF) were sampled using a Kistler multi-component forceplate (Kistler 9286A, Kistler Instrumente, Winterthur, Switzerland) recording at a frequency of 200Hz as specified by CODA (see Chapter 3.5 and Chapter 3.6). The force plate was installed in the middle of the walkway and at the same level as the walking surface to prevent a potential tripping hazard. Both the walkway and the forceplate were covered in the same compact and thin carpet covering which was glued on top of the wooden and metal surfaces. This provided consistent, slip resistant grip for the participant and helped to conceal the location of the forceplate.

Three-dimensional bilateral kinematic data was recorded at 200Hz using 22 active infra-red emitting markers (11 per limb) defining a seven-segment rigid link model of the lower limb as described in chapter 3.6. The two cameras were positioned at opposite ends of the walkway to ensure optimal marker visibility during the stance phase of the lateral gaits.

5.2.4 Data Analysis

Ground reaction force, kinematic and kinetic data was processed in CODAmotion V6.68 (Charnwood Dynamics Ltd.). Kinematic and kinetic data calculations were performed within the supplied software package. Please refer to Chapter 3.6 and Chapter 3.7 for a detailed review of the methods used for the calculation of kinematic and kinetic data, via inverse dynamics calculations, respectively. Stance phase was defined as the period of time from contact of the foot of interest with the forceplate to lift-off of the same foot from the forceplate. This was taken to be equivalent to the vertical ground reaction force impulse at a cut-off magnitude of 15N (see Chapter 3.10). Positional data of the markers was filtered at cut-off frequencies of 10Hz for X, Y and Z coordinates (see Chapter 3.9). Further analysis of the GRF, kinematic and kinetic data was performed after exporting the information from CODAmotion to the spreadsheet application Excel© (Microsoft Corp.).

Positive and negative work at the joints was calculated by trapezoidal integration of the power curve where periods of power generation (positive) and power absorption
Achilles tendon (AT) loads were determined using moment arm data by Rugg et al. (1990). They determined AT moment arms for a moving centre of rotation of the ankle joint using magnetic resonance images for 10 adult male participants. In their investigation, ankle joint angular data was defined as the angle between the midline of the tibia, the ankle centre of rotation and the toe. To fit this definition of the ankle angle, a virtual angle was created in CODAmotion defining a two dimensional angle between the knee marker, the ankle and the toe marker. The data for a variety of ankle joint angles, published in Rugg et al. (1990), was used to derive a linear regression line for the estimation of the AT moment arms over a functional range of motion (100 to 150 degrees). Tendon load was taken as the quotient of the ankle flexor / extensor moment divided by the moment arm, where:

\[ MA = m \times \alpha_A + b \]

and

\[ TL = \frac{AM_{FE}}{MA} \]

Where \( MA \) = moment arm; \( m \) = gradient of regression line; \( \alpha_A \) = ankle joint angle; \( b \) = constant of regression line; \( TL \) = tendon load; \( AM_{FE} \) = flexion/extension joint moment at ankle.

Assuming this relationship to be valid mean peak tendon loads were determined for each participant and averaged across participants.

5.2.5 Statistics

Statistical analysis of the GRF and kinetic data was performed using SPSS. ANOVA tests of variance with repeated measures were performed in order to identify differences in GRF and joint kinetics within the leading or trailing limb during SS and XS (effect of gait) and within SS and XS gaits to identify differences between the leading and trailing limb (effect of limb). Furthermore, ANOVA tests of variance were used to identify differences in GRF and joint kinetics due to skill and gender. Statistical significance was taken as \( p<0.05 \) throughout.
5.3 Results

5.3.1 Footfall Mechanisms

It was anticipated that there may be some variability in footfall mechanisms at impact between participants, similar to those observed in running (Cavanagh and Lafortune, 1980, Perry, 1992, and Józsa and Kannus, 1997). Therefore, footfall mechanisms at impact were assessed by visual inspection of the kinematic data for the gait repeats of individual participants. Participants were classed as rearfoot (RF) or forefoot strikers (FF), if they initiated contact with the ground at impact with the heel first or with the mid-foot region respectively. The findings indicate that all participants used FF landings for trailing limb SS and XS. There was some variability in the leading limb during the SS and XS with 5 participants (4 Squad and 1 Rec player) performing RF landings for the SS and 9 participants (7 Squad and 2 Rec players) for XS with the remaining participants utilising a FF approach. This data indicates a change in landing technique for the leading limb within the group of squad members.

5.3.2 Ground Reaction Force

Typical ground reaction force (GRF) traces for the stance phase of the leading and trailing limb during the SS and XS gaits are presented in Figure 5.1. Vertical ground reaction force mostly lacked a high frequency impact peak in early stance, typically seen in heel-toe running due to the preference for midfoot striking in-line with forefoot running (Cavanagh and Lafortune, 1980). The largest force values were identified for the vertical and horizontal force parameters. Analysis was therefore performed on peak loading in the vertical and horizontal axes only.
Figure 5.1 Typical bilateral GRFs for one participant performing SS and XS gaits. Averaged GRF data of 10 repeats for horizontal (Fx), antero-posterior (Fy) and vertical (Fz) forces are presented. Error bars indicate the force range for the participant. Positive and negative force values represent:

<table>
<thead>
<tr>
<th>Fx</th>
<th>+ve=</th>
<th>Braking</th>
<th>-ve=</th>
<th>Push-off</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fy</td>
<td>+ve=</td>
<td>Anterior push</td>
<td>-ve=</td>
<td>Posterior push</td>
</tr>
<tr>
<td>Fz</td>
<td>+ve=</td>
<td>Vertical impact</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

5.3.2.1 Effect of Body Mass on Maximum GRF

Ground reaction forces are commonly normalised to the participants body mass in the literature, as a method for reducing inter-individual differences (Cavanagh and Lafortune, 1980, Hamill et al., 1983, and Keller et al., 1996). For this method of normalising to be applicable to lateral gaits, a linear relationship of vertical GRF and body mass must exist. The effect of body mass on vertical GRF in SS and XS was therefore investigated for a selection of participants performing SS and XS gaits within a defined speed range (where, SS_L = 2.45 ms\(^{-1}\) ± 10% (n=13); SS_T = 2.59 ms\(^{-1}\) ± 10% (n=13); XS_L = 2.68 ms\(^{-1}\) ± 10% (n=10) and XS_T = 2.23 ms\(^{-1}\) ± 10% (n=14). Linear relationships between body mass and vertical GRF (Fz) was observed for both the leading and trailing limb during SS and XS (see Figure 5.2). This relationship was used as the basis for reducing inter-individual variation through ratio normalisation using body mass, as recommended by Mullineaux et al. (2006). As a consequence of this finding all further analysis of GRFs was performed using ratio-normalised GRF expressed as body mass units (BMU).
5.3.2.2 Effect of Speed of Locomotion on Maximum GRF

Speed of locomotion has been shown previously to play an important role in determining GRF magnitude in walking (Keller et al., 1996) and running (Munro et al., 1987, Hamill et al., 1983), showing increased vertical and horizontal force due to faster speeds. This may play an important role in SS and XS, contributing to data variation between participants. This means that for the purpose of comparing between groups it may be necessary to normalise for the speed of locomotion. In order to determine the relationship between speed and reaction force in lateral gaits, the influence of speed on normalised vertical and horizontal GRF (expressed as BMU) was assessed using force averages at selected speed ranges. GRF repeats for each participant were selected on the basis of speed of locomotion (mean ± 5%) so that each force average is representative of a narrow speed range. The resultant data is representative of more than 60% of the total data for each participant for SS_L, SS_T, XS_L and XS_T.

The relationship between speed and GRF was assessed using linear regression analysis. When assessing the relationship of force and speed for all participants the findings indicate a significant linear decrease in peak vertical and horizontal impact force with speed of locomotion in SS_L (p<0.05). Furthermore, significant increases in peak horizontal push-off forces were observed for SS_L, SS_T and XS_L (p<0.05) (see Figure 5.3).
While the increase in horizontal push-off force due to increasing speed was expected, the decrease in horizontal braking force and decrease in peak vertical force only for SS_L was surprising and warranted further investigation. In addition to the speed of locomotion, the movement technique has been shown to influence GRF variability. Forward lean and 'groucho' style running, results in reduced GRFs due to a lowered centre of gravity (Bobbert et al., 1991, Keller et al., 1996). To investigate the effect of technique on SS_L, vertical GRF was plotted against vertical hip displacement. The results indicate a linear relationship between force and total vertical displacement (see Figure 5.4). No such relationship was observed for XS_L. The data indicate that the mean maximum hip displacement in the SS was 40% larger than in the XS (where SS = 19.3 ± 5.5cm and XS = 13.7cm ± 6.6cm total vertical displacement). Furthermore, the mean hip displacement was ~37% larger in recreational players compared to squad members during the SS, and ~22% larger during the XS, indicating an alteration in movement kinematics in the group of experienced players. The vertical force component therefore appears to be influenced by differences in vertical hip displacement, with larger forces resulting from larger vertical displacement of the hip. The causes for the reduction in horizontal braking force are unclear. No conclusive
relationship could be determined but it is thought that the finding is related to the preference of a forefoot landing approach. The more medially rotated limb, compared to the more laterally rotated limb during RF striking, may allow for greater knee flexion at impact, reducing the braking impulse in some participants. Overall, it can be concluded there does not appear to be a clear linear relationship between force and speed, as reported for typical gaits in the literature, with the exception of peak horizontal push-off force. As a consequence, speed will not be considered as a covariate in the statistical analysis of the effects of gender or skill on vertical or peak horizontal impact force, with the exception of horizontal push-off force.

![Graph](image.png)

Figure 5.4 A linear relationship was determined for the effect of increased vertical hip displacement on the vertical component of the GRF.

5.3.2.3 Effect of Gait and Limb on Maximum GRF

The overall findings (using force data for all participants) indicate a significant effect of both gait and limb. Gait displayed a very limited effect with slightly larger horizontal braking forces in the leading limb during the SS (0.42 ± 0.11 BMU and 0.35 ± 0.09 BMU for SSL and XSL respectively, p<0.05). No other differences due to gait were observed. Differences in GRF were more pronounced between the leading and trailing limbs within the SS and XS gaits. The findings for the effect of limb are summarised in Figure 5.5.
5.3.2.4 Effect of Gender and Skill on Maximum GRF

The results for the effects of gender on vertical and horizontal GRF indicate that, in SS₁, females generated slightly greater maximal horizontal braking forces in SS₁ (0.49 ± 0.09 BMU and 0.35 ± 0.08 BMU for female and male participants respectively, p<0.05). However, in view of the relationship of braking force and speed presented in Figure 5.3, these findings are likely related to differences in speed of locomotion, where male participants performed the movement at a faster average speed than female participants (2.56 ± 0.26 and 2.28 ± 0.21 ms⁻¹ respectively).

Within Squad and Rec participants small but significantly larger horizontal braking forces were observed in the experienced group for XS₁ only (0.39 ± 0.09BMU and 0.29 ± 0.07BMU for squad and recreational players respectively, p<0.05). Again, these findings are likely related to a slightly faster speed of locomotion in squad players compared to recreational players (2.5 ± 0.25 and 2.36 ± 0.29 ms⁻¹ respectively).

5.3.3 Joint Moments

Typical hip, knee and ankle joint moment traces for the stance phase of one participant are summarised in Figure 5.6. The mean moment of force pattern was determined by normalising the data for each gait repeat to 0-100% stance phase and scaling joint moments to the participant’s body mass. Peak joint moments were
determined for individual gait repeats and averaged across the total number of repeats by a participant for the leading and trailing limb during the SS and XS. These moment means acted as the basis for the investigation of the effects of gait, limb, skill and gender on peak moments in the following sections. It can be observed that the major moments occur during hip adduction/abduction, hip flexion/extension, knee flexion/extension, and ankle plantarflexion/dorsiflexion. As a consequence, the analysis will focus on the joint kinetics during these movements.
5.3.3.1 Effect of Gait and Limb on Peak Joint Moments

The results of the analysis of the effect of gait on the peak joint moments in the leading and trailing limb for all participants indicated a number of significantly differences. In the leading limb the SS resulted in slightly larger hip adductor
moments (-1.59 ± 0.54 Nmkg⁻¹ and -1.39 ± 0.51 Nmkg⁻¹ for SS and XS respectively, p<0.05). The XS in turn resulted in a significantly larger mean knee extensor moment (2.63 ± 0.44 Nmkg⁻¹ for XS_L and 1.59 ± 0.46 Nmkg⁻¹ for SS_L, p<0.05). In the trailing limb the SS resulted in a small but significantly larger hip adductor moment (-0.95 ± 0.47 Nmkg⁻¹ SS_T and -0.69 ± 0.34 Nmkg⁻¹ XS_T), a larger hip extensor moment (1.71 ± 0.72 Nmkg⁻¹ SS_T and 1.06 ± 0.48 Nmkg⁻¹ XS_T). Furthermore, the peak hip flexor moment was slightly larger in XS_T (-1.32 ± 0.55 Nmkg⁻¹ XS_T and -1.02 ± 0.64 Nmkg⁻¹ SS_T).

Figure 5.7 Peak joint moments of the leading and trailing limb during SS and XS gaits. Joint moments are expressed as Nmkg⁻¹. The data represents mean peak joint moments for all participants included in the study. * denotes statistically significant differences in moment values between the leading and trailing limbs, where p<0.05.
While the findings above show that there were few differences between the SS and XS, with largely small but significant differences, the differences within gaits due to limb were more pronounced. The findings of the statistical analysis of the effect of limb are summarised in Figure 5.7.

5.3.3.2 Effect of Gender and Skill on Peak Joint Moments

The results indicated a significant effect of skill for a number of joint moments. In SSL the peak hip extensor moment was larger in squad members (2.53 ± 0.77 Nm kg\(^{-1}\) Squad and 1.89 ± 0.73 Nm kg\(^{-1}\) Rec, \(p<0.05\)) while the ankle plantarflexor moment was larger in recreational players (2.01 ± 0.27 Nm kg\(^{-1}\) Rec and 1.7 ± 0.32 Nm kg\(^{-1}\) Squad, \(p<0.05\)). In XSL the peak hip abductor moment was slightly larger in squad than recreational players (1.06 ± 0.035 Nm kg\(^{-1}\) Squad and 0.81 ± 0.22 Nm kg\(^{-1}\) Rec, \(p<0.05\)). No significant effect of skill was observed in the trailing limb. Gender appeared to have a more marked effect on peak joint moments. Figure 5.8 summarises the findings of the effect of gender on peak joint moments for the stance phase of SS and XS gaits.
Figure 5.8 Summary of the findings related to the effect of gender on peak joint moments. Mean joint moment data (Nm kg⁻¹) for male and female players is presented for peak hip, knee and ankle joint moments in all three dimensions. * denotes significant differences in joint moments due to gender.

5.3.4 Joint Powers

For the purpose of visualising the joint powers at the hip, knee and ankle example data for one participant is presented. As for the joint moments, the average joint power pattern were determined by normalising the data for each gait repeat to 0-100% stance phase and scaling joint moments to the participant’s body mass (see Figure 5.9). The power curves are the product of the joint moment and angular velocity and the area under each the curve is the work done at the particular joint. Power generation was said to occur when the product was positive and power absorption when the product was negative. Two major power phases were present in all participants at the hip flexion/extension (FE), knee (FE) and ankle plantarflexion/dorsiflexion (PD). Furthermore, there was a distinct hip adduction/abduction (AA) power phase during SS and XS for both the leading and
trailing limbs. Peak joint powers were determined for individual gait repeats and averaged across the total number of repeats by a participant for the leading and trailing limb during the SS and XS.

Figure 5.9 Typical leading and trailing limb joint power averages of 10 gait repeats for one female club badminton player performing SS and XS gaits. Joint powers are presented in Watts per kg body mass and data is averaged over 100% stance phase. Internal/external (IE) moments are presented for visualisation purposes only.

5.3.4.1 Effect of gait and limb

Due to the apparently limited effect of skill and the generally larger joint moments exhibited by male participants it was decided to focus on the groups of male participants for an assessment of joint powers in the SS and XS. No differences in
Joint power were identified between the SS and XS for the leading limb. A number of significant differences between gaits were observed in the trailing limb. Positive power during hip extension was larger in SS compared to the XS (2.6 ± 1.4 Wkg\(^{-1}\) and 1.6 ± 0.9 Wkg\(^{-1}\) for SS\(_T\) and XS\(_T\) respectively). Power absorption during the hip extensor moment was furthermore larger in the SS than XS (-1.6 ± 0.8 Wkg\(^{-1}\) and -1.0 ± 0.4 Wkg\(^{-1}\) for SS\(_T\) and XS\(_T\) respectively). Furthermore, positive power during knee extension was larger in the XS (8.03 ± 3.06 Wkg\(^{-1}\) for XS\(_T\) and 6.29 ± 2.4 Wkg\(^{-1}\) for SS\(_T\), p<0.05).

As for the joint moment data, differences between limbs were more pronounced than those due to gait. The findings are summarised in Figure 5.10.

Figure 5.10 Effect of limb on joint powers during the SS and XS. Data is presented for the energy power generation and absorption phases (expressed in Watts per kg body mass). * indicates significant differences in joint powers between limbs.
5.3.4.2 Comparison of Joint Powers with In-vitro Literature

Several musculoskeletal optimising mechanisms are involved in running gaits which increase the ability of the musculature to generate force and enhance gait efficiency. Amplification of muscular power through stretch activation (Cavagna, 1978, Edman et al., 1978, Bobbert and Van Ingen Schenau, 1990, Jaric et al., 1985) and the return of elastic energy from the tendons of the lower limb (Jacobs et al., 1993, Van Ingen Schenau et al., 1997) have been shown to play an important role in movement. In order to identify the influence of amplification of muscular power through the processes described in Chapter 2.2.5.2 it was decided to assess muscular powers in relation to the total mass of the muscles involved in a particular joint movement. The movements of interest were hip adduction and abduction, knee extension as well as ankle plantarflexion. The mean power values at the joints for these movements were divided by the total mass of the muscles involved in generating the corresponding movements. The normalisation of muscular powers was based on the muscle mass reference data from Pierrynowski (1995). Body mass and height, of the model presented in the reference data, were a close match to the mean data for the group of squad players (Body mass: 70.8 kg and 72.9 kg; Height: 1.78 m and 1.79 m for the reference and mean data of the male participants respectively) and the reference data was therefore not rescaled.

The findings of the comparison indicate high positive power outputs, particularly of the trailing limb. High values for the plantarflexors indicate an important contribution of the Achilles tendon as well as stretch induced muscular power amplification. Adductor muscles of the leading and abductor muscles of the trailing limb, furthermore, appear to produce power outputs toward the high end of the power range observed in-vitro, which are in the range of ~200 W/kg muscle (see Chapter 2.2.5.2). Knee extensor muscles of the trailing limb also appear to be generating large power indicating a possible contribution of stretch-induced power amplification. Generally speaking, power amplification appears to play an important role in the trailing limb with a reduced contribution in the leading limb.
Figure 5.11 Positive power values for the group of male squad members during the SS and XS. Data is presented for the leading and trailing limb during hip adduction (leading limb), hip abduction (trailing limb), hip and knee extension as well as ankle plantarflexion.

5.3.5 Positive and Negative Work

Two main extensor/flexor power phases were identified at the hip (HE1, HE2), knee (KE1, KE2) and ankle (AE1, AE2) joints as well as two abductor/adductor power phases at the hip joint (HA1, HA2) as shown in Figure 5.12 and summarised in Table 5.2.

Figure 5.12 Example data of the main positive and negative work done at the hip, knee and ankle joints for the leading and trailing limb of 1 participant.
Table 5.2 Summary of the findings for mean positive and negative work done at the hip, knee and ankle joints. Units of work shown are Joules per kg body mass (Jkg⁻¹).

<table>
<thead>
<tr>
<th></th>
<th>SSL Mean (Jkg⁻¹)</th>
<th>SSL Min</th>
<th>SSL Max</th>
<th>SST Mean (Jkg⁻¹)</th>
<th>SST Min</th>
<th>SST Max</th>
</tr>
</thead>
<tbody>
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<td>IIA1</td>
<td>0.21 ± 0.10</td>
<td>0.12</td>
<td>0.34</td>
<td>0.16 ± 0.06</td>
<td>0.09</td>
<td>0.28</td>
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<tr>
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<td>-0.01</td>
<td>-0.02 ± 0.02</td>
<td>-0.05</td>
<td>0.00</td>
</tr>
<tr>
<td>IIF1</td>
<td>-0.12 ± 0.10</td>
<td>-0.28</td>
<td>-0.03</td>
<td>-0.05 ± 0.05</td>
<td>-0.11</td>
<td>-0.01</td>
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<tr>
<td>IIF2</td>
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<td>0.07</td>
<td>0.36</td>
<td>0.18 ± 0.09</td>
<td>0.04</td>
<td>0.32</td>
</tr>
<tr>
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<td>-0.03</td>
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<td>-0.63</td>
<td>-0.02</td>
</tr>
<tr>
<td>KF2</td>
<td>0.20 ± 0.17</td>
<td>0.02</td>
<td>0.42</td>
<td>0.40 ± 0.18</td>
<td>0.14</td>
<td>0.57</td>
</tr>
<tr>
<td>AFL</td>
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<td>-0.03</td>
<td>-0.50 ± 0.16</td>
<td>-0.69</td>
<td>-0.23</td>
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<tr>
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<td>0.38</td>
<td>0.64 ± 0.22</td>
<td>0.49</td>
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<tr>
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<td>0.36</td>
</tr>
<tr>
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<td>-0.30</td>
<td>-0.03</td>
<td>-0.01 ± 0.01</td>
<td>-0.03</td>
<td>0.00</td>
</tr>
<tr>
<td>IIF1</td>
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<td>-0.49</td>
<td>0.00</td>
<td>-0.04 ± 0.05</td>
<td>-0.13</td>
<td>0.00</td>
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<tr>
<td>IIF2</td>
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<td>0.66</td>
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<td>-0.13 ± 0.17</td>
<td>-0.42</td>
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<tr>
<td>KF2</td>
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<td>0.01</td>
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<tr>
<td>AF2</td>
<td>0.30 ± 0.13</td>
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<td>0.49</td>
<td>0.53 ± 0.16</td>
<td>0.26</td>
<td>0.68</td>
</tr>
</tbody>
</table>

5.3.6 Achilles Tendon Load

The injury literature identifies the Achilles tendon as an at-risk structure in badminton (Kaalund et al., 1989, Hoy et al., 1994, Kvist, 1994, Fahlström et al., 2002). The estimated Achilles tendon loads were similar for the leading limb during the SS and XS and the trailing limb in the SS and XS. The load was generally larger in the trailing limb, which is thought to be due to the trailing limb accepting the body weight at impact, following on from the aerial phase. The findings are summarised in Figure 5.13. The loads were higher than those reported for squat or countermovement jumps (3.08 and 2.61 BMU respectively) but lower than the values quoted for hopping (5.21 BMU) (Fukashiro et al., 1995). Furthermore they were lower than the values reported for running (12.5 BMU) (Komi, 1990) but higher than those reported for walking (3.6 BMU) (Kader et al., 2002). Compared to in-vitro measures of Achilles tendon failure loads reported in the literature, about 4600 to 5600N or about 7 times body mass (Thermann et al., 1995), the predicted loads during SS and XS were therefore within the expected limits. It is generally recognised however, that comparisons to in-vitro
measures provide limited usefulness since peak loading can be seen to surpass the in-vitro measures readily during in-vivo use.

Figure 5.13 Mean peak tendon loads for the leading and trailing limb for all participants performing SS and XS gaits. Units are presented in relation to body mass (BMU). Values within the leading or trailing limb were similar with generally higher loads exerted on the AT of the trailing limb.
5.4 Discussion

It was the objective of this investigation to quantify the magnitude of ground reaction forces and joint kinetics the lower limbs are exposed to during the performance of lateral SS and XS gaits. Furthermore, differences in kinetics between four groups of male and female, experienced and inexperienced participants was performed to identify the effects of gender and playing ability. The purpose of this investigation was therefore, to describe the gait processes of a standardised lateral SS and XS manoeuvre in the lab to assess the demands of the movements on the athlete.

Before discussing the results of the study in more detail some restrictions of the experiment need to be considered. The investigation of gait kinematics and kinetics was restricted to the stance phase of the leading or trailing limb due to labspace restrictions. The cameras used for motion capture require placement in an environment largely free of infra-red radiation to prevent interference during the marker triangulation process. The available indoor space was restricted in size causing field-of-view issues restricting consistent data capture to the stance phase only, rather than the entire gait cycle. As a consequence reference to the entire gait cycle is made only for visualisation purposes. A further restriction of the experiment was the hardware restricted recording frequency of the forceplate. As explained in Chapter 3.6 recording of bilateral gait kinematics and kinetics requires in a data capture frequency of 200Hz. This may result in underestimation of the force values, however, no alternative method was available to record gait at higher data capture frequencies. Furthermore, due to the lab-space restrictions it was decided to restrict movement measures to a preferred speed. The use of a badminton racket during the experiment may have had an effect on the force recordings. However, this is thought to be minimal due to the light weight of the racket. The benefits of using a racket, simulating arm movement in a sport-specific fashion and creating a specific movement goal, were thought to outweigh the negative effects of racket use.

5.4.1.1 General Movement Processes
The most readily observable differences between the lateral gaits and typical walking and running are the direction of travel and the number of aerial phases (the period in the gait cycle where no limb is in contact with the ground). The body moves laterally, instead of an anterior direction of travel, and a single aerial phase is used in a gait cycle, instead of no aerial phase in walking and two aerial phases in running, as shown in Chapter 2.2.1. Figures 5.14 and 5.15 illustrate that the aerial phase occurs after toe-off of the leading limb, indicating its function in generating the force necessary for vertical displacement of the body during the aerial phase. The end of the aerial phase, in turn, is followed by two consecutive single limb support phases. When defining the start and end of the SS gait cycle (GC), based on the progression of the leading limb (Figure 5.14), leading limb swing phase starts with toe-off of the leading limb and ends in contact of the foot with the ground (initial contact, IC) (0 to ~64% GC). This initiates stance phase, which ends with toe-off of the leading limb (64 to 100% GC). Trailing limb stance phase is initiated by impact of the trailing limb at ~32% GC of the gait cycle and ends at toe-off of the same limb at ~66% GC. At this point trailing limb swing phase is initiated, which lasts until the consecutive impact.

Figure 5.14 Data for one representative participant, visualising the SS gait cycle from a posterior point of view. The stick figure and gait cycle timings (% GC) are presented for descriptive purposes only. Blue and red bars indicate the stance phase of the trailing and leading limb respectively.
The XS gait cycle (Figure 5.15) consists of essentially the same combination of leading and trailing limb sequences as the SS. The fundamental difference in the gait cycles however, is due to the movement of trailing leg, moving the trailing limb beyond the midline of the body. It appears that during the XS the duration of the aerial phase is reduced, with initial contact occurring earlier on in the GC, while the duration of the stance phase of the leading limb is increased (~55-100% GC).

![Typical Gait Cycle - Crossover stepping (Anterior to Posterior View)](image)

Figure 5.15 Data for one representative participant visualising the XS gait cycle from an anterior point of view. The stick figure and gait cycle timings (% GC) are presented for descriptive purposes only. Blue and red bars indicate the stance phase of the trailing and leading limb respectively.

5.4.1.2 Ground Reaction Force

Gender and skill appeared to have a limited effect on force parameters. Differences in speed of locomotion between groups, with faster mean speeds in male participants and Squad players, as well as differences in technique, such as larger vertical hip displacement, appeared to be the dominant factors. Interestingly, the data indicate that Squad players adopted a smaller mean maximum vertical displacement of the hip (37% and 22% lower than recreational players) and showed a higher number of rearfoot landings for the leading limb, particularly during the XS. While vertical hip displacement appeared to relate well to vertical force magnitude and therefore
explained some of the vertical force variability, no conclusive cause of the apparent reduction in horizontal braking forces in SS\textsubscript{L} could be determined. It is thought that these may be related to greater knee flexion at impact, similar to the findings for 'groucho' style running (Bobbert et al., 1991). Moreover, in line with the observations by Cavanagh and Lafortune (1980) for toe-running, the consistent use of a forefoot landing approach by the trailing limb resulted in the absence of a peak vertical impact force, which is an important characteristic of heel-toe running, related to overuse injury (Nigg, 1988). This was also true for the participants using a forefoot landing approach for the leading limb. However, a number of participants, particularly within the group of squad players, preferred a rearfoot strike pattern, resulting in a force pattern similar to that seen in heel-toe running. As a consequence, the potentially damaging effect of the high frequency content of the GRF appears to contribute to a lesser degree in lateral movements.

Differences in force parameters between the leading and trailing limb were more clearly observable (see Figure 5.5). These findings support the view of a distinct contribution of the leading and trailing limb to the SS and XS. The magnitude of the maximum vertical and horizontal force components was within the expected rage for walking and running (see Table 5.3). Maximum vertical force was generally below the force maxima reported for running, which is likely due to a generally lower speed of locomotion in the SS and XS, compared to the running data from the literature. Interestingly, horizontal braking forces for the leading limb and push-off forces for the trailing limb were generally comparable to those seen in faster running speeds. The underlying cause for the significantly larger vertical loading and horizontal push-off forces in the trailing limb, as well as the significantly larger horizontal braking forces in the leading limb (see Figure 5.5), can be attributed to the alignment of the respective limb at IC. While the leading limb is maximally abducted at ground contact the trailing limb is maximally adducted. The trailing limb therefore, makes contact with the ground while being located under the centre of the upper body and is able to immediately move the body laterally without being exposed to large braking forces. The leading limb on the other hand accepts the body weight while in abduction, placing the centre of gravity further behind the point of ground contact. In agreement with the observations by Hamill et al. (1983), this results in larger horizontal braking forces as the upper body begins to move over the adducting limb. These findings and
the significantly larger maximum vertical force in the trailing limb (see Figure 5.5),
provide strong evidence that the trailing limb acts in a shock absorbing and body
support function during stance, while the leading limb is involved in supporting the
body during stance and providing vertical lift for the following aerial phase and
creating the lateral momentum carried through by the trailing limb.

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5.4.1.3 Moments and Powers

In order to gain further insight into the role of the musculature of the leading and
trailing limbs in the investigated lateral gaits and verify the proposed contributions of
the limbs, it is necessary to quantify and compare the patterns and magnitudes of the
joint moments and powers. The joint moments, summarised in Figure 5.6, show that
the major contribution to the generation of force occurs due to flexion and extension
of the hip, knee and ankle joints as well as adduction and abduction at the hip joint.
Figure 5.16 shows the time plots of joint excursion angles, moments and powers for
the leading and trailing limb during the stance phase of the SS for a representative
male, experienced participant. Due to the similarities in SS and XS the following
discussion of the joint kinetics will focus on the data from the SS with reference to XS
where appropriate. Furthermore, since joint moments were generally larger in male participants, the data presented below is in reference to the group means for the male participants only.

Figure 5.16 Joint angles, moments and powers for the main contributors to propulsive force generation and absorption for the leading and trailing limb during SS. Joints are organised by column. As a visual aid the joint motion is shown in the first row with net joint moment and joint power in the second and third row. Positive joint powers refer to power generation and negative powers to power absorption. The results are representative of one participant and normalised by body mass (Nm/kg and Watts/kg).

Following initial contact, the **trailing limb** begins to abduct until toe-off creating a peak **hip** abductor moment of \(-1.5\) Nmkg\(^{-1}\) with peak joint power of \(-2.4\) Wkg\(^{-1}\) and positive work of \(0.16\) Jkg\(^{-1}\). At the same time, the hip of the trailing limb extends with a peak moment of \(-2\) Nmkg\(^{-1}\) and generating a peak power of \(-2.6\) Wkg\(^{-1}\). The work done at the hip joint is mostly positive producing \(0.18\) Jkg\(^{-1}\) in the second half of stance. The peak hip extensor moment was smaller in the XS than the SS, which is likely due to a more extended hip of the trailing limb in the XS at impact (\(~9\) degrees larger hip extension in XS\(_T\) compared to SS\(_T\) in the group of skilled badminton players) due to the progression of the limb beyond the midline of the body.
Furthermore, the trailing limb appeared to perform larger positive and negative work during hip extension in the SS movement suggesting a larger contribution of the extensor muscles of the trailing limb in the SS.

Similar to the hip, the knee of the trailing limb responds to loading by flexing and extending from mid-stance producing a peak extensor moment of \( \sim 3 \text{Nmkg}^{-1} \). The power pattern displays a strong biphasic shape with peak positive power of \( 6.3 \text{Wkg}^{-1} \) and negative power of \( 4.4 \text{Wkg}^{-1} \), absorbing energy during early stance. The work done at the knee joint consists of absorbing \( 0.26 \text{Jkg}^{-1} \) and generating \( 0.4 \text{Jkg}^{-1} \) during extension. The amount of positive power at the knee joint during extension was greater in the XS than the SS suggesting a larger contribution to force generation by the limb during the XS gait.

The ankle joint of the trailing limb is involved in the largest amount of negative and positive work. As previously mentioned the ankle joint dorsiflexes in early stance, absorbing \( 0.5 \text{Jkg}^{-1} \), with a corresponding peak negative power of \( 8.3 \text{Wkg}^{-1} \). This is followed by a large plantarflexor moment (\( \sim 2.2 \text{Nmkg}^{-1} \)) that coincides with peak ankle dorsiflexion at mid-stance. During plantarflexion from mid-stance to toe-off the ankle generates \( 0.64 \text{Jkg}^{-1} \) of positive work with a peak extensor power of \( 8.6 \text{Wkg}^{-1} \). The use of a forefoot landing approach, therefore, is an integral part for the absorption of energy generated at impact of the limb.

Trailing limb stance phase is followed immediately by leading limb stance phase. When the leading limb makes contact with the ground the limb immediately begins to adduct until toe-off. The peak hip adductor moment occurs in early stance (\( \sim 1.9 \text{Nmkg}^{-1} \)). The hip joint is dominated by positive adductor work with a peak joint power of \( \sim 2.7 \text{Wkg}^{-1} \) and \( 0.21 \text{Jkg}^{-1} \) of work performed. Peak hip flexion occurs in early stance with a corresponding peak hip extensor moment of \( \sim 2.7 \text{Nmkg}^{-1} \). The hip performs negative work of \( \sim 1.9 \text{Wkg}^{-1} \) in early stance and generates \( 2.4 \text{Wkg}^{-1} \) in mid- to late stance. Overall the hip generates about twice as much extensor work as it absorbs (\( 0.22 \text{Jkg}^{-1} \) and \( 0.12 \text{Jkg}^{-1} \) respectively) and generates a significantly larger hip extensor moments than the trailing limb. In line with the findings by Stefanyshyn and Nigg (1998) this can be regarded as evidence for the role of the leading limb as a major provider of propulsive force.
Similar to the trailing limb the knee and ankle joints of the leading limb flex and extend during early and late stance with peak flexion occurring in mid-stance. The leading limb generated a peak knee extensor moment of \( \sim 1.6 \text{Nmkg}^{-1} \). Furthermore, the extensor moment of the knee was larger in the XS than SS, which may be a reflection of a change in limb alignment due to the larger number of participants choosing a rearfoot strike pattern, placing increased emphasis on the contribution of the knee. The ankle joint created peak moments of \( \sim 1.9 \text{Nmkg}^{-1} \), with peak power of \( \sim 4 \text{Wkg}^{-1} \) and \( 5.7 \text{Wkg}^{-1} \) (negative and positive power respectively) and work values of \( \sim 0.23 \text{Jkg}^{-1} \) and \( 0.26 \text{Jkg}^{-1} \) (negative and positive work respectively).

Overall, the mean peak joint moments and powers, summarised in Figure 5.15 and Figure 5.16 respectively, with reference to the literature, are within the limits expected for walking and running at comparable speed ranges. The findings show that gender and skill make a relatively small contribution to joint moments in line with observations in running (Kerrigan et al., 1998, Ferber et al., 2003).

Figure 5.17 Comparison of joint moment data for the group of male participants with the literature. Joint moments are expressed in Newton meters per kg body mass. Speed of locomotion is expressed in meters per second.
Figure 5.18 Comparison of joint power data for the group of male participants with the literature. Joint powers are expressed in Watts per kg body mass. Speed of locomotion is expressed in meters per second.

The joint kinetic data appear to support the proposed functions of the leading and trailing limb. The basic gait kinematics and the reaction forces suggested a shock absorption function by the trailing limb and a propulsive force generation function by the leading limb. Figure 5.7 and 5.10, as well as the summaries of joint moments and powers above, display the distinct differences in kinetics between the limbs. The larger moments and negative power at the knee and ankle of the trailing limb show the importance of the limb in absorbing the shock of impact. This shock absorption function therefore, is similar to the function of the knee absorbing the shock of impact in walking and running (Winter, 1984, Winter, 1983). Furthermore, a larger hip extensor moment in the leading limb points toward a dominant contribution of the limb to the generation of propulsive force. This view concurs with findings for long jumping (Stefanyshyn and Nigg, 1998), where the hip joint performs an important function in generating energy. Furthermore, the ankle plantarflexor moment was much lower in the leading than the trailing limb, which agrees with findings by Jacobs et al. (1993) who show that the net plantar flexion moment during running is far higher than the peak values reached in maximal jump and sprint push-offs. Therefore, the lateral movements appeared less efficient at utilising the musculoskeletal energy enhancing mechanisms (Chapter 2.2.5) that are utilised for propulsive force generation in running.
The cause for the lower moments and powers in the leading limb is related to the gait cycle. The trailing limb accepts the body weight from maximal vertical displacement of the centre of mass, while the leading limb accepts the weight of the body at the lowest vertical displacement (see Figure 5.14 and 5.15). This mechanism raises a number of interesting issues with a view to gait efficiency. Figure 5.11 shows that, when normalising power for muscle mass, the knee extensors and particularly the ankle plantarflexors of the trailing limb are working at and above the maximum isotonic capacity (~200Wkg\(^{-1}\) muscle) determined for skeletal muscle in-vitro (see Figure 2.10). Power amplification of the muscle using potentiation (Cavagna, 1978, Edman et al., 1978) and the development of a higher active state (Bobbert and Van Ingen Schenau, 1990, Jaric et al., 1985), as reviewed in Chapter 2.2.5.2, therefore likely play an important part at the ankle. The magnitude and rate of the stretch impulse are key factors in amplifying muscular work output (Edman et al., 1978). The finding of larger power values in the trailing limb therefore, are not surprising since the total force absorbed by the limb at IC is larger, which is reflected by the larger vertical GRF for the trailing limb. This optimises the ability of the muscle to provide power and of the tendon to return stretch energy. When the leading limb makes contact with the ground, on the other hand, the hip is at the lowest vertical displacement, with the hip already near peak flexion angles. Therefore, the vertical force impulse is reduced. This reduces the potential of the leading limb to utilise the energy enhancing mechanisms and ultimately results in lower power outputs observed in the leading limb. As a consequence of the results stated above, it can be argued that both the SS and XS appear to consist of effectively two phases of firstly, lateral jumping and secondly two controlled periods of shock absorption.

5.4.1.4 Achilles Tendon

Achilles tendon (AT) injuries occur relatively frequently in badminton (Kaalund et al., 1989, Høy et al., 1994, Fahlström et al., 2002). The findings of the experiment show an important function of the AT, particularly in the trailing limb. This is confirmed by higher AT load in the trailing (~4.4BMU) than the leading limb (~3.8BMU) (Figure 5.13). AT loads were always below the tendon failure loads determine in-vitro, however they were higher than those reported for squat or countermovement jumps.
(3.08 and 2.61BMU respectively) walking (3.6BMU) (Kader et al., 2002), but lower than the values quoted for hopping (5.21BMU) (Fukashiro et al., 1995) or running (Komi, 1990). These findings display the importance of the ankle of the trailing limb in absorbing the impact.

Despite the multitude of factors that may be related to the occurrence of injury in sports, the data presented in this chapter point toward a number of specific contributing factors. The demands on the Achilles tendon do not appear to be excessive. However, other factors may still play a role. Since not all of the energy generated during tendon stretching is returned, but is dissipated as heat, repeated loading of the tendon may increase the internal temperature of the tendon structure to levels above the fibroblast viability threshold of 42.5°C (Wilson and Goodship, 1994). This causes the degeneration of the tendon cellular elements (fibroblasts) and could contribute to tendon rupture in badminton (Hoy et al., 1994, Kvist, 1994, Kaalund et al., 1989, Fahlström et al., 2002).

5.4.1.5 Conclusion

The findings showed that lateral SS and XS movements displayed very similar biomechanical demands which were within the range expected for walking and running gaits. Furthermore, the data provides evidence that the leading and trailing limbs perform distinct roles in the gait cycle. Unlike walking or running where the activity of the two limbs can be described as essentially mirroring each other, lateral movements require a transfer of momentum by the trailing limb and a lift by the leading limb. The lateral gaits can therefore be described as lateral jumping with two consecutive phase of controlled falling. While the findings do not identify excessively large physiological demands in the lateral movement tasks the asymmetric contribution of the leading and trailing limbs points toward a mechanically inefficient mode of transport.

Moreover, the findings support the view that experienced participants show adaptations to the movement technique. These differences may provide protection for the ankle from sprain injury by reducing the likelihood of excessive ankle pro-
supination. While differences in kinetic data due to skill or gender were few and generally low in magnitude, experienced participants performed lateral movement at generally faster speeds and displayed a preference for 'steady hip' movement and rearfoot footfall mechanisms, which appears to be related to the reduction of vertical reaction force.
Chapter 6

Kinematic, kinetic and EMG analysis of selected lower limb muscles during lateral sidestepping and crossover stepping tasks

6.1 Introduction

The results presented in the previous chapter indicated that lateral sidestepping (SS) and crossover stepping (XS) gaits place largely similar biomechanical demands on the participant. Importantly, there appeared to be an asymmetric contribution of the leading and trailing limb to the gait cycle. This resulted in comparatively large contribution of the hip and reduced contribution of the ankle muscles, compared to those seen in running, which points toward a comparatively inefficient gait process due to the reduced involvement of musculoskeletal energy saving mechanisms. However, this study contained no detailed information on the coordination and contribution of the muscles of the lower limb during the performance of these lateral gaits. Furthermore, due to the restrictions of the previous biomechanical investigation, there was no information relating to the kinematics of the entire gait cycle. It will be the focus of this chapter, to look into the relationship of muscular activity and the resulting joint biomechanics, since this information provides essential understanding of the activation and coordination of the musculature of the lower limbs and its relationship to the development of muscular force.

A method of determining the activity of the muscles directly is the recording of the electrical impulses resulting from the active muscle (contracting concentrically, eccentrically or isometrically) through the use of surface electromyography (sEMG).
Recording the electrical activity (action potentials) of muscles allows for direct access of the physiological processes involved in movement. However, as outlined in Chapter 3.8 there are a number of factors that affect the application and interpretation of the EMG signal. A common and relatively straightforward use of EMG is the assessment of muscular activity by recording the onset and offset times of the contracting muscle. De Luca (1997) explains that determining the activation times of muscles is not affected by the type of muscular contraction but only if any part of the muscle in the vicinity of the electrode is active. This information provides very useful and direct insight into the coordination of muscular activity. Furthermore, by relating the activity of the musculature, determined from EMG, to the kinematics and kinetics of the joints of the lower limb it is possible to determine muscle function and coordination as well as the relationship between muscular activity and the contribution and functional activity of the musculature to the movement.

A number of authors have used EMG recordings in dynamic movements in order to determine the coordination of muscular activity. Montgomery et al. (1994), and Dietz et al. (1979) investigated muscular coordination in running for a variety of different speeds using needle and surface EMG respectively, while Karamanidis et al. (2004) investigated the effect of speed and stride frequency on muscle coordination in running using surface EMG. Neptune et al. (1997) used sEMG for the assessment of muscle coordination in cycling at different pedal rates. Furthermore, Neptune et al. (1999) recorded muscle activations, force and kinematic data for the right limb during sidestepping and cutting tasks in order to identify muscle coordination and function with a view to ankle sprain injuries. However, to the best knowledge of the investigator, there have been no studies examining the activity of the muscles of the lower limb for both the leading and trailing limb during continuous lateral SS and XS gaits.

The objective of this investigation is to analyse the activation patterns of a selection of major superficial lower limb muscles of the leading and trailing limb during lateral gaits. This description of the timed activation patterns, in combination with kinematic and kinetic data, allows for the quantification of the function of the underlying anatomy in generating the asymmetric gait process. A second objective was to relate the ground reaction force and kinetic data recorded in this investigation to the data
presented in Chapter 5 and to a straight line run, in order to re-examine the effects of gait on stride length and reaction force at the controlled speed of locomotion. Therefore, the current investigation had the following aims:

- To record electromyographic data for seven muscles acting on the hip, knee and ankle joint of the leading and trailing limb in order to determine their activation times and establish a database of bilateral muscular coordination during sidestepping and crossover stepping gaits performed at a controlled speed.
- To record kinematic and force data in order to relate the electrical activity of the muscles to the resultant joint excursion angles for the complete gait cycle and to the joint kinetics during stance to describe the functional significance of the muscular activities during gait and particularly in the generation of propulsive force.
- To re-examine ground reaction force and joint kinetics for a controlled speed of locomotion to gain further information on lateral gaits.

It was hoped that this data would shed light on differences in muscle coordination strategies between the leading and trailing limb and due to the use of the SS and XS gait. Furthermore, it was hoped that this data would provide information on possible muscle deficiencies, which could be implicated in overuse injuries at the lower limbs.
6.2 Methods

6.2.1 Participants

Nine male squad players from the Loughborough University Students Badminton club, with no history of chronic lower limb injuries, volunteered for participation in the study (height = 177.7 ± 5.2 cm, weight = 73.21 ± 9.29 kg, age = 20.7 yr). Ethical approval was received by the local ethics committee and informed consent was obtained from the participants before data collection. Because of reasons of personal privacy and lack of a partitioned area it was decided to select male participants only.

6.2.2 Experimental Design

Data was collected during repeats of three movement tasks: a straight line run, lateral sidestepping (SS) and lateral crossover stepping (XS). 10 repeats of each task were performed on a raised wooden walkway (8.4 x 1.2m) at a controlled speed of 3 ± 0.3 ms\(^{-1}\). This speed was chosen to represent the top end of the preferred speed of locomotion for SS and XS by experienced male participants, based on the speed data from Chapter 5. This speed was furthermore chosen to allow for comparisons of the gathered data with the biomechanics literature where 3 ms\(^{-1}\) was frequently chosen to represent a slow running speed. Speed of locomotion was controlled by a set of wireless light gaits (IRD-T175, Brower Timing Systems) located at waist height around the centre of the forceplate. After the attachment of markers and surface electrodes for the recording of kinematics and EMG to the lower limbs, the light gates were aligned parallel to the walkway and separated by 1.5m. The movement task order was randomly assigned to control for fatigue. Sidestepping commenced with the participant in a slightly crouched, wide-stance position to the left of the forceplate. The participant then performed the movement to the right, hitting the force platform on the third stride with the leading limb, for leading limb EMG recordings, and the trailing limb, for trailing limb EMG recordings, and returned to the starting position to the left of the forceplate. Before recordings were taken, the participants were given as many practice runs as necessary to assure they felt comfortable with the movements, the process of data capture and the kinematic and EMG recording equipment attached
to their lower extremities. Furthermore, this time was used to ensure the participants successfully and reliably achieved the required speed of locomotion and consistently hit of the centre of the forceplate with their leading or trailing foot. In addition to the data capture of squad players the investigation was also performed on one recreational player, the investigator, to gain information on data variability for between day measurements. The experimental protocol was the same as that chosen for the experienced student players and care was taken to perform the movement as naturally as possible.

6.2.3 Ground Reaction Force

A Kistler (Type 9286A) mobile multi-component forceplate, integrated into the middle of the walkway, was used to record ground reaction forces (GRFs). The cut-off magnitude of force was set to 2.5kN in the Z and 1.25kN in the XY direction. GRF data was sampled at a hardware limited frequency of 200Hz. GRF onset and offset was determined from the vertical GRF with a cut-off value of 15N.

6.2.4 Kinematic Data

A two-camera, Cartesian Optoelectronic Dynamic Anthropometer (CODA, Charnwood Dynamics Ltd, UK) was used to record 3D bilateral kinematics for the lower limb using a specialised wand system supplied by Charnwood Dynamics Ltd (see Chapter 3.6). Marker positions were sampled at 200 Hz. The raw marker position data was filtered at 10 Hz following experience from the previous investigation (see Chapter 5) and recommendations by Winter (1990) and Neptune et al (1999). The cameras were placed on opposite ends of a walkway (8.5m x 1.2m), on which all movements were performed (see Figure 6.1). The placement of the cameras was influenced by the space restrictions of the room and offered the best compromise for optimal marker visibility at the point of contact of the foot with the forceplate and to allow for the greatest possible field of view to capture as much of the movement as possible.
6.2.5 Kinetic Data

Kinetic data was calculated from GRF and kinematic data using an inverse dynamics approach within the data capture and analysis software supplied by Charnwood Dynamics (CODA motion V6.68) (see Chapter 3.7). The resultant data on moments and powers was exported to Excel (Microsoft® Office Excel 2003, Microsoft Corporation) for further processing and analysis.

6.2.6 Surface Electromyography

An eight analogue channel system (Biometrics Ltd., Gwent, UK, www.biometricsltd.com) with two additional digital input channels was used for the simultaneous recording of muscular activities from seven muscles of the lower limb. (see Chapter 3.8 for a more detailed description of the system). Pre-amplified bipolar surface EMG electrodes (Biometrics Ltd, see Figure 6.2) were placed in line with the middle of the muscle belly of the medial gastrocnemius (GAS), tibialis anterior (TA), rectus femoris (RF), biceps femoris (BF), adductor longus (AL), gluteus medialis (Gmed) and gluteus maximus (Gmax). Electrode placement was performed following marker placement guidelines by Delagi.
and Perotto (1980) with consideration of the recommendations by De Luca (1997) and Hermens et al. (2000) (see Chapter 3.8). The marker placements and methods for marker placement identification are summarised in Figure 6.3 and Figure 6.4. Furthermore, a measure of skin impedance was taken, through one of the digital channels, by placing an electrode on an electrically neutral location by securing it with an elastic strap on the bony surface of the wrist (lateral styloid process).

![Muscle Electrode Placement Test Manoeuvre Pitfalls](image)

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Electrode Placement</th>
<th>Test Manoeuvre</th>
<th>Pitfalls</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gmax</td>
<td>Midway between the greater trochanter and the sacrum.</td>
<td>Hip extension with the knee flexed.</td>
<td>None.</td>
</tr>
<tr>
<td>Gmed</td>
<td>One inch distal to the midpoint of the iliac crest.</td>
<td>Abduction of the thigh.</td>
<td>Posterior placement of the electrode will record gluteus maximus; anterior placement will record tensor fascia lata.</td>
</tr>
<tr>
<td>AL</td>
<td>Palpate the tendon arising from the pubic tubercle. Attach the electrode four fingerbreadths distal to the pubic tubercle.</td>
<td>Adduction of the limb.</td>
<td>If the electrode is attached too medially it will be the gracilis; if attached too laterally it will be the sartorius.</td>
</tr>
<tr>
<td>RF</td>
<td>On the anterior aspect of the thigh, midway between the superior border of the patella and the anterior superior iliac spine.</td>
<td>Flexion of the hip with the knee extended.</td>
<td>Medial placement of the electrode will record vastus intermedius; lateral placement will record vastus lateralis; and distal and medial placement will record vastus medialis.</td>
</tr>
</tbody>
</table>

Figure 6.3 Electrode placement methodology for Gmax, Gmed, AL and RF. Pictures and information adapted from Delagi and Perotto (1980).
Prior to marker placement the skin was shaved and cleaned with isopropyl alcohol (Sterets Pre-injection swab, Seton Healthcare Group plc) to reduce impedance and the EMG signals were preamplified and sampled at a frequency of 1000 Hz. 10 trials were collected for each electrode set and each movement resulting in a total of 60 successful task repeats: leading limb sidestep (SSL), trailing sidestep (SSR), leading crossover step (XSL), trailing crossover step (XSR) and a run (RUN).

### 6.2.7 Data Synchronisation and Analysis

EMG and GRF/kinematic data were recorded on two separate Windows XP (Microsoft® Corporation) based computer systems. The kinematic data was used as an aid for identifying swing and stance phase durations. Therefore time synchronisation of the two data streams was necessary to relate EMG data to the gait cycle timings identified in the corresponding kinematic data stream. Hardware limitations, the lack and cost of a digital input box for CODA to allow for a digital synchronisation pulse to be sent to both systems simultaneously, an alternative method was developed. Synchronisation was achieved through the use of a pressure mat (RS Components Ltd. Stock No. 317-156; 29 x 15in) placed on the forceplate. The trigger mat was attached to the remaining free digital input channel of the Biometrics system. Before the start of a movement task, the trigger mat was activated.
by a physical tap. The force of the tap was registered by both the trigger mat and the underlying force plate and the tapping signal was recorded in both the EMG recording as a digital signal and as an analogue signal in the GRF trace, which itself is an integrated part of the kinematic recording. Synchronisation was achieved by identifying the timing of the first occurrence of significant vertical force (>15N) in the GRF trace and using this time point to align the GRF trace with the leading edge of the digital trigger, visible in the corresponding EMG trace within Excel. All further data processing was performed within a customised spreadsheet in Excel.

Following method in described in Winter (1990), De Luca (1997), Hermens et al. (2000) and Neptune et al. (1999) a number of automated and semi-automated data analysis techniques were attempted in order to identify muscle activation times (ATs) during the swing and stance phase of the gait cycle, with the aim of preparing a typical EMG activation template (see Chapter 3.8). The results of these methods were, however, not satisfactory due to the dynamic nature of the movement, which complicated the use of automated detection processes. Because of the limited number of participants in the study it was therefore decided to analyse the data manually by visually examining the raw data. In order to minimise the effect of individual gait variations, swing and stance phases of the gait cycle were analysed separately by normalising their duration to 0-100% of swing or stance phase. Information on swing and stance phase durations was obtained from visual inspection of the kinematic data within CODAmotion (V6.68). Swing phase was defined as the period of time from toe-off of the limb under investigation to consecutive ground contact of the same limb with the forceplate. Toe-off was identified from the kinematic data using both the visual representation of the foot (stick figure in CODAmotion) and the vertical and horizontal toe and heel marker displacement data. Mickelborough et al. (2000) showed that the determination of foot contact events using foot marker kinematics represents a valid and reliable method for normal adult gait (see Chapter 3.10 for a review). The suggested protocol however, could not be reproduced exactly due to the difference in limb kinematics in the lateral movements compared to a walk. In the SS and XS movements the horizontal movement of the limb at toe-off made an important contribution and there was some variation in the amount of vertical displacement of the foot. Therefore, identification of toe-off was based on both the vertical and horizontal displacement of the heel and toe to provide the highest level of detail when
determining when the foot left the ground. This method was initially practised with the foot hitting the force-plate to compare the visually determined values with those when defining toe-off as the point in time where vertical ground reaction force falls below 15N. Stance phase was defined as the period of time the limb spends in contact with the force plate which was taken to be equivalent to the GRF reading with a cut-off value of 15N.

The raw EMG data was rectified and ATs for swing and stance phases were analysed separately on a scale of 0-100% ± 50% to either side of the swing and stance phase durations to optimise data visibility. ATs were identified by visually inspecting the relevant EMG trace graphs in Microsoft© Excel by manually distinguishing between the activation beginning and end points of the 7 muscles, for the swing and stance phases of the gait cycle. Figure 6.5 displays example traces for all 7 muscles under investigation as well as indicators for the criteria used to distinguish between activation onsets and offsets. This process was repeated for 6 gait repeats of the SSₐ, SSₜ, XSₐ and XSₜ gaits for 9 student squad members.

![Figure 6.5 Traces represent compositions of swing and stance phases of the rectified raw EMG trace used for identification of muscular activity onset and offset values. Arrows are used to mark the time points where the muscle activity onset and offset was deemed to occur.](image-url)
6.2.8 Statistics

GRF and kinetic data for the three modes of transport was analysed using a repeated measures test of variance (ANOVA) in the statistical analysis tool SPSS. Statistical significance was taken as $p<0.05$. 
6.3 Results

6.3.1 Activation Times

Average activation times for the 7 muscles of the leading and trailing limb during the SS are presented in Figure 6.6. The data for the leading limb was largely in agreement with the data presented in Neptune et al. (1999). In the leading and trailing limb Gmax, BF and GAS exhibited distinct, single burst activity patterns with activity onset and offset occurring prior to the start of and before the end of stance phase respectively. The thigh adductor AL displayed a similar, single activity burst, in the leading limb however, this activity changed in the trailing limb where muscular onset occurred in early stance with activation offset shortly after toe-off. In the leading limb the thigh abductor Gmed was active until mid to late swing and from about mid stance. In the trailing limb on the other hand, onset occurred just prior to stance phase, with activity throughout stance and offset occurring at toe-off. The pattern of activity for the ankle dorsiflexor TA showed the greatest amount of variability across participants and gaits. Burst onset during stance phase was consistent for all participants in all gaits, occurring in the latter stages of stance phase. Offset timings and activity patterns during swing phase however, varied greatly between participants making an identification of offset timing not viable. The data indicated that, in the leading limb, offset tended to occur shortly before activity onset in stance phase and activation during swing phase was characterised by a mixture of either single burst activations, lasting throughout swing phase, or two distinct activations during swing phase. The trailing limb displayed similar characteristics with either single bursts of activity or two bursts of activity before onset in stance phase. The durations of activity varied greatly between participants and TA tended to be active through most of swing phase. In the leading limb RF was active from late swing with activity throughout stance and activity offset after toe-off. In the trailing limb in turn, activation offset of RF appeared to occur earlier than in the leading limb.
Sidestep Leading Limb

Sidestep Trailing Limb

Figure 6.6 Mean activation times of 7 lower limb muscles for 9 experienced student badminton players. Data is separated into swing and stance phases for the leading and trailing limb for the SS movement. Activation onset and offset times are presented as values in the range of 0-100% of either swing or stance phase. Additional reference data for the leading limb during the SS (broken line) was adapted from Neptune et al. (1999) for visualisation purposes. Note that the reference data is presented in relation to the 0-100% stance phase.

Activation times for leading and trailing limb XS are presented in Figure 6.7. The recorded activation times were similar to those during the SS scenarios. GAS and Gmax displayed distinct single activation bursts contained within the swing phase to stance phase period. AL and BF also displayed this activity pattern for the leading limb, however, activation onset occurred later, just prior to stance phase, and offset occurred later, shortly after toe-off, rather than in mid-stance for both AL and BF in the trailing limb. In both limbs, activity onset for TA occurred shortly before the end of stance phase. As seen in SS, TA displayed considerable variability for both leading and trailing limb XS. Furthermore the data indicated that two participants displayed no onset of TA in stance phase for XSₜ but a continuous activation. This activation was initiated just prior to stance phase with offset occurring just after toe-off. Activation onset of RF in the leading limb generally occurred in late swing phase and continued throughout stance phase with offset occurring in early swing. The activation pattern in the trailing limb differed from the leading limb with activation onset occurring in mid-swing and ending in late stance. The general pattern of activity of RF was mirrored by Gmed with onset in early stance and offset in mid to late swing.
6.3.2 Swing and Stance Phase Durations

Average swing and stance phase durations for the leading limb and trailing limb during the SS and XS gait are presented in (see Figure 6.8). No statistically significant differences in swing phase duration were observed due to gait (p=0.239 SS_L/XS_L, p=0.646 SS_T/XS_T) or between the leading or trailing limb (p=0.1 SS_L/SS_T, p=0.514 XS_L/XS_T). However, stance phase was seen to be affected by gait with significantly longer stance durations for leading limb XS (0.239 ± 0.019sec and 0.206 ± 0.033sec for XS_L and SS_L respectively) and trailing limb XS (0.275 ± 0.039sec and 0.236 ± 0.02sec for XS_T and SS_T respectively). Furthermore, there was a significant difference in stance duration between the leading and trailing limb during the SS where the trailing limb spent significantly more time in contact with the ground than the leading limb (0.236 ± 0.02sec and 0.206 ± 0.033sec for SS_T and SS_L respectively). The same trend was observed in the XS despite a lack of a significant difference (p=0.055). In comparison to the Run the data show that the swing duration of the leading and trailing limb was significantly lower in the SS and XS compared to the Run. Furthermore, the stance phase of the Run was significantly longer than SS_L, however, there were no significant differences in the stance phase of Run in comparison to SS_T (p= 0.134), XS_L (p= 0.077), or XS_T (p= 0.244).
6.3.3 Stride Length

Stride length was measured as the distance travelled by the leading or trailing limb during the swing phase. Six gait repeats for the leading and trailing limb were analysed for each participant to determine average stride length values. These values were averaged across participants to determine the mean stride length for the group (see Figure 6.9).

After adjusting the stride length values for the leg length of the individual participants it can be seen that the change from a SS to a XS caused a significant increase in the distance travelled by the trailing limb using the XS compared to the SS. There was no significant effect of gait in the leading limb \( (p=0.078 \text{ SS}_L/\text{XS}_L) \) and no significant difference between the leading and trailing limb during the SS or XS \( (p=0.061 \text{ and } p=0.324 \text{ for } \text{SS}_L/\text{SS}_T \text{ and } \text{XS}_L/\text{XS}_T \text{ respectively}) \). Changing from the SS to the XS therefore caused, on average, a 5.9% increase in leading limb stride length and a 6.8% increase in stride length of the trailing limb. Compared to the run stride lengths for both the leading and trailing limb in the SS and XS were significantly shorter than those seen in running.
6.3.4 Kinematic Data

Typical bilateral joint excursion angles for both SS and XS tasks during the swing and stance phase of the movement are presented in Figure 6.10. The following section summarises the joint excursion angle data for hip adduction/abduction, hip and knee flexion/extension as well as ankle dorsiflexion/plantarflexion with reference to the typical trace and the mean peak joint angle data from all participants. Joint angles are given positive and negative values in reference to the definition of joint angles given in Chapter 3.6 as visualised in Figure 6.10.

**Hip adduction/abduction**

Peak hip adduction in the leading limb SS occurred just after toe-off (-4.8 ± 7.3°). This was followed by abduction of the limb during the swing phase with peak abduction (-23.6 ± 7.4°) occurring just before impact of the limb with the ground. The trailing limb on the other hand was adducted until impact (mean peak = -10.9 ± 6.7°) and abducted during stance with peak abduction (-33.6 ± 4.6°) occurring at toe-off.

The same pattern of leading and trailing limb adduction and abduction was observed in the XS. However, the peak magnitude of adduction and abduction was affected by gait. There was a reduction in leading limb hip abduction (p=0.071) when using the XS (-18.2 ± 8.8°) as well as a significant reduction of peak abduction angle in the trailing limb (-28.4 ± 5.8° for XS_T). Furthermore, significantly larger hip adduction in both the leading and trailing limb was observed when using the XS (7.8 ± 5.9° and 1.2 ± 5.9° for XS_L and XS_T respectively). The total range of motion was therefore larger in the XS than the SS with an average increase of ~38% and 30% for the leading and trailing limb respectively.

**Hip flexion/extension**

Hip flexion of the leading limb during the SS was largest in early swing (53.4 ± 8.2°). Thereafter the hip extended with peak hip extension occurring in late swing (41.6 ± 11.5°) followed by hip flexion and extension from early to late stance. The trailing
limb displayed a similar flexion/extension pattern as the leading limb but with a larger total range of motion (~12° for the leading limb and ~24° in the trailing limb).

The same hip flexion/extension pattern was observed in the XS for the leading and trailing limb with a similar difference in their respective total range of motion (~15° and 22° for the leading and trailing limb respectively). There was no significant difference in peak hip flexion/extension of the leading limb between the SS and XS (p=0.91 and p=0.32 for leading limb flexion and extension respectively). There was however a significant effect of gait on both peak flexion and extension of the trailing limb causing a reduction in peak hip flexion (35.6 ± 14.0°) and an increase in peak extension (13.9 ± 5.6°) when using the XS.

Knee flexion/extension
At the knee mean peak flexion values of 68.3 ± 8.9° and 66.1 ± 8.3° were recorded for the leading and trailing limb during the SS. Maximum knee flexion for both the leading and trailing limb occurred in mid-stance, as the limb accepts the weight of the body. The knee then extended from mid-stance with peak knee extension (40.8 ± 9.6° and 31.3 ± 6.3° for the leading and trailing limb respectively) after toe-off of the trailing limb and prior to impact for the leading limb. As seen for hip flexion, the total range of motion was, on average, larger in the trailing limb (~35°) than the leading limb (~27°).

A very similar pattern of knee flexion and extension was observed in the XS. There were no significant differences in knee flexion values for the leading (p=0.83) and trailing limb (p=0.129) during the SS or XS and there was no significant difference in knee extension angles in the leading limb (p=0.613). However, there was a significant decrease in trailing limb knee extension when using the XS (25.9 ± 6.9°). While the range of motion of the leading limb was similar in the SS and XS there was an average-increase of ~28% in the total range of motion of the knee of the trailing limb when using the XS.

Ankle dorsiflexion/plantarflexion
Ankle dorsiflexion and plantarflexion patterns for the leading and trailing limb were similar for the SS and XS. Peak dorsiflexion (17.7 ± 4.3° and 25.6 ± 3.3° for SS and
SSₜ respectively) occurred in mid-stance as the limb accepts the weight of the body. Thereafter, the ankle systematically plantarflexed, with peak plantarflexion occurring just after toe-off. There were no significant differences in dorsiflexion of the leading (p=0.307) or trailing limb (p=0.09) or in plantarflexion of the trailing limb (p=0.825) when changing from the SS to the XS. However, there was a significant difference in leading limb plantarflexion between the SS and XS, which can likely be explained by a preference for a heel-strike pattern at impact by some participants as seen in Chapter 5.
Figure 6.10 Lower extremity joint excursion angle data of a typical subject for the leading and trailing limb during the SS and XS movement tasks. Units are in degrees. The horizontal time scale is normalised to percent total gait cycle, from toe-off to consecutive toe-off of the leading limb. Dashed vertical lines indicate the timing of trailing limb impact (IT) and toe-off (TT) and the solid vertical line indicates leading limb impact (IL). Positive angles indicate hip adduction and flexion, knee flexion and ankle dorsiflexion.

6.3.5 Ground Reaction Force

Typical bilateral GRFs for one participant performing the SS, XS and running gaits are visualised Figure 6.11.
Figure 6.11 Bilateral GRF data averages for a typical subject performing SS, XS and running (dashed line) gaits. The vertical scale is normalised to body mass units (BMU, where $1 = 1 \times$ body mass) and the horizontal scale is normalised to stance phase duration from impact to toe-off.

The findings of the statistical analysis of GRF for the three modes of transport are summarised in Figure 6.12. The results indicate a significant effect of gait on peak vertical and horizontal push-off forces of the leading and trailing limb but not for horizontal breaking forces at the leading ($p=0.082$) or trailing limb ($p=0.516$). Further breakdown showed that there were no significant differences between SS$_L$ and XS$_L$ ($p=0.87$ for horizontal push-off and $p=0.627$ for peak vertical force). However, the
Run resulted in significantly larger horizontal push-off and vertical forces than either leading limb SS or XS. Furthermore, in the trailing limb the SS resulted in significantly larger peak horizontal push-off force than the XS or Run with the XS also resulting in significantly larger horizontal push-off forces compared to the Run. Peak vertical force in the trailing limb was significantly different between all three gaits with the Run resulting in the largest vertical force, followed by the SS and the XS.

The data furthermore indicates that peak horizontal push-off and peak vertical force in the SS differed significantly between the leading and trailing limb with larger horizontal push-off forces in the trailing limb (0.56 ± 0.04 BMU and 0.12 ± 0.07 BMU for SS_T and SS_L respectively) as well as larger vertical forces in the trailing limb (2.19 ± 0.27 BMU and 1.92 ± 0.2 BMU for SS_T and SS_L respectively). No significant trend was identified for the peak horizontal braking data for the SS (p=0.093). In the XS the trailing limb displayed significantly larger horizontal push-off forces (0.45 ± 0.09 BMU and 0.17 ± 0.07 BMU for the trailing and leading limb respectively), however, no significant trends between the leading and trailing limbs were identified for peak horizontal braking (p=0.375) or peak vertical force (p=0.406).

![Peak Horizontal Braking](image1)
![Peak Horizontal Push-off](image2)
![Peak Vertical Force](image3)

Figure 6.12 GRF averages for the 9 squad members is presented for the leading and trailing limb SS and XS as well as right limb Run. Data is presented for a) the leading limb and b) the trailing limb. * indicates statistically significant differences in vertical and horizontal GRF parameters.
Overall the data agreed with the data collected in the previous investigation (see Figure 6.13). Irrespective of the variation in speed of locomotion in the previous study and the relatively faster speed used in the current study, vertical GRF does not appear to be greatly influenced, displaying similar peak values and data variability. Horizontal braking force of the trailing limb was larger in the current investigation, which is thought to be a reflection of the increased speed of locomotion as the trailing limb makes contact with the ground first after the aerial phase. Furthermore, mean push-off forces by the leading limb were larger in the current investigation, in-line with an increased contribution of the leading limb during force generation at the faster speed of locomotion.

![Graph showing GRF data comparison](image)

**Figure 6.13** Mean vertical and horizontal GRF data from the current investigation and the data presented in Chapter 5 (all participants). Force data is expressed in relation to body mass (BMU).

### 6.3.6 Joint Moments

The joint moments in this investigation were in agreement with the data for male participants presented in Chapter 5 (see Figure 6.14). A significant effect of gait on joint moments was observed for the peak knee extensor moment of the leading limb ($1.92 \pm 0.047 \text{Nmkg}^{-1}$ and $2.33 \pm 0.42 \text{Nmkg}^{-1}$ for SS and XS respectively) and the peak hip extensor moment in the trailing limb ($2.62 \pm 1.18 \text{Nmkg}^{-1}$ and $1.37 \pm 0.68 \text{Nmkg}^{-1}$ for SS and XS respectively) (see Figure 6.15).
In line with the findings in Chapter 5 a number of significant differences were observed between the leading and trailing limb during the SS and the XS. The results are summarised in Figure 6.16.

![Figure 6.14 Peak joint moment data for hip adduction/abduction; and hip, knee and ankle flexion/extension from the current investigation and the group of experienced male players used in the investigation in Chapter 6 is presented. The data can be seen to compare favourably between the two studies.](image)

![Figure 6.15 Summary of the differences in peak joint moments between the SS and XS within the leading and trailing limbs. * indicates significant differences between the SS and XS. Joint moment values are expressed as Nmkg⁻¹.](image)
Figure 6.16 Summary of the differences in peak joint moments between the leading and trailing limbs during the SS and XS. * indicates significant differences between the leading and trailing limb. Joint moment values are expressed as Nm/kg.

Figure 6.17 Joint moments from the current investigation compared to the literature. Joint moments are expressed as Newton meters per kg body mass. The horizontal scale is speed, expressed in meters per second.

The variation in joint moments at the hip, knee and ankle during stance was furthermore investigated for one participant. The moment averages for 8 repeats of the SS and XS for the leading and trailing limb were obtained and their durations normalised to 0-100% stance phase. The coefficient of variation (CV) for each of the moment averages was calculated, where CV = mean σMj / | mean Mj | x 100%. The results are summarised in Table 6.1. The data indicate low variation at the ankle and knee, in line with the findings by Winter (1983). The highest variability was observed in the hip moments, particularly in trailing limb flexor / extensor moments. Lower, flexor / extensor moment variability of the leading limb appears to indicate a more
consistent function compared to the trailing limb whose function appears to be more
variable.

| Table 6.1 Coefficient of variance for stance phase moments of the hip, knee and ankle for one|
| participant performing 8 repeats of each movement. |  |
| SSL | SST | XSL | XST |
| Hip Add/Abd | 30% | 92% | 76% | 25% |
| Hip Flex/Ext | 40% | 103% | 37% | 72% |
| Knee Flex/Ext | 45% | 26% | 26% | 21% |
| Ankle Plant/Dorsi | 26% | 16% | 12% | 9% |

6.3.7 Joint Powers

A significant effect of gait on joint powers was observed for peak negative power the
knee of the leading limb (-4.89 ± 2.07W kg\(^{-1}\) and -3.47 ± 1.93W kg\(^{-1}\) for SS\(_L\) and XS\(_L\) respectively) (see Figure 6.18). Furthermore, positive and negative work at the hip,
during flexion/extension) was significantly larger during the SS at the trailing limb.
The results are summarised in Figure 6.18.

![Figure 6.18 Summary of the differences in peak joint powers bet\(^{\prime}\)een the SS and XS within the leading
and trailing limbs. * indicates significant differences between the SS and XS. Joint power values are
expressed as W kg\(^{-1}\).

Significant differences were observed between limbs. Positive power at the knee and
positive and negative at the ankle was significantly larger in the trailing limb during
the SS. Furthermore, positive power at the knee and negative power at the ankle was
larger in the trailing limb during the XS. The findings are summarised in Figure 6.19.
Figure 6.19 Summary of the differences in peak joint power between the leading and trailing limbs during the SS and XS. * indicates significant differences between the leading and trailing limb. Joint power values are expressed as Wkg⁻¹.

Figure 6.20 Positive power values, adjusted for the respective muscle group, during the SS and XS. Data is presented for the leading and trailing limb during hip adduction (leading limb), hip abduction (trailing limb), hip and knee extension as well as ankle plantarflexion. Data is expressed as Watts per kg muscle mass.
6.4 Discussion

The objective of this investigation was to analyse the activation patterns of a selection of major superficial lower limb muscles of the leading and trailing limb during lateral gaits. This description of the timed activation patterns allows for the quantification of the function of the underlying anatomy in generating the asymmetric gait process. A second objective was to relate the data recorded in this investigation to the data presented in Chapter 5 and to a straight line run, in order to re-examine the effects of gait on stride length and reaction force at the controlled speed of locomotion. The primary aim was to establish a database of bilateral muscle activation strategies for a selection of the major muscles acting around the joints of the lower limbs. EMG data and joint kinematics and kinetics were combined in order to identify the functional significance of the muscular activations in terms of the joint moments and powers generated during gait. This data was used to identify differences in activation strategies and joint kinematics and kinetics between the leading and trailing limb as well as the effect of gait within the limbs.

Before discussing the significance of the findings of this investigation it is important to address a number of limiting factors. Muscle activity was explored only for a selection of the major superficial muscles of the lower limbs. This means that the contribution of some superficial muscles and all deep muscles to the movement remains undefined in this study. Furthermore, the methodology applied to the determination of muscular activation times is an inherent source for discussion. In this study onset/offset timings were determined by manual inspection of the integrated raw EMG data trace during the standardised swing and stance phase durations. A number of different methods are used to automate the process of identifying muscle excitation burst onset and offset timings. Neptune et al. (1999) used automated waveform analysis software while Ferguson et al. (2004) used a activation level cut-off point of 10% above and below the resting level to determine activation onset and offset and Lephart et al. (2005) used the EMG voltage to determine onset where the cutoff value was chosen as the mean voltage plus 3 standard deviations of the resting trials. In the absence of the necessary software and the unsuccessful attempt of implementing similar selection criteria in either Excel or Matlab it was decided that
manual inspection of the traces would represent an adequate method for determining activation times, based on essentially the same selection criteria as used by the implemented search algorithms. Furthermore, crosstalk from surrounding muscles represents an important contributor to data analysis problems since it can lead to increased noise in the signal as a result of the interfering signals of other, nearby muscles as discussed in Chapter 3.8.

6.4.1 General Gait Processes

Both of the investigated lateral gaits displayed some distinct and expected differences, which were largely in agreement with the data presented in the previous chapter. The duration of the swing phase of the leading and trailing limb was similar between both movements, as visualised in Figure 6.8. The stance phase of the lateral movement however, was significantly longer during the XS for both the leading and trailing limb, reflecting the longer total length of the XS movement as well as the larger distance covered during the XS. This is reflected in the larger total range of motion, as indicated by the range of adduction and abduction and the stride length of the leading and trailing limb (see Figure 6.10), which is larger when using the XS, resulting in a mean increase in stride length of 5.9% for the leading and 6.8% for the trailing limb. Furthermore, the data indicates that the trailing limb spends significantly more time in contact with the ground than the leading limb in the SS, which to a lesser extent, is also true for the XS. This may be regarded as further proof of the propulsive action of the leading limb, due to a shorter, explosive extensor action of the limb rather than a longer duration shock absorbing function.

The GRF data between the current and the previous investigation were also largely in agreement (see Figure 6.13 and Figure 6.14 respectively). Interestingly, at this controlled speed, there were no significant differences between limbs in horizontal braking forces, unlike the findings presented previously. The results indicate that differences between lateral gaits are prominent in the trailing limb and not the leading limb, with SS\textsubscript{T} being exposed to significantly higher mean peak horizontal push-off and peak vertical force, reflecting its important shock absorption and function in the transition to the contact phase of the leading limb. Compared to the Run, these data
show that the lateral movements utilise a much more even distribution of stance and swing durations, which results in shorter stride lengths. The force data, in turn, shows that while braking forces between lateral movements and running are comparable, vertical forces were larger in running. It is evident that the leading limb is not capable of generating the same direction of force during the lateral gaits as in running (large vertical and horizontal push-off). This agrees with the findings in the previous investigation, showing that the lateral movement is far more dependent on a jump movement with controlled lateral falling.

Joint moments between the current and the previous investigation were very similar (see Figure 6.14). In-line with findings by Winter (1983) the variation in moment patterns at the leading and trailing limb, during the stance phase of the SS and XS, was largest at the hip and lowest at the ankle, indicating a very consistent contribution of the ankle to the gaits (see Table 6.1). As seen in the previous investigation, the moment data indicate a larger contribution of the knee extensors of the leading limb during the XS, while the hip extensors contributed to a larger degree at the trailing limb during the SS. As seen in Chapter 5, the trailing limb displayed a larger knee extensor moment, while the leading limb displayed a larger hip extensor moment, reflecting the proposed work of the trailing limb as a shock absorber and of vertical force generator function of the leading limb. In comparison to the literature (Figure 6.17) these data support the view of a different contribution of the hip, knee and ankle joint to the generation of force, with the proximal joints contributing comparatively more to force generation, in comparison to walking or running. This distribution is in line with the findings by Thorpe et al. (1998) and Stefanyshyn and Nigg (1998) for jumping. Therefore, joint powers were again shown to be generally larger in the trailing limb (see Figure 6.18), with clear indications of musculoskeletal power amplification of positive joint power in the trailing limb as well as a reduced contribution in the leading limb (see Figure 6.19).

In order to further investigate the consequences of this movement strategy, muscle function was assessed by relating the electrical activity of the muscle to the motion, moment and power of the joint affected by the muscle. The discussion of the results was conducted by focusing on the activity of the muscles involved in a particular joint motion during the SS with reference to differences due to the use of the XS when
appropriate. Figure 6.21 summarises the relevant data for the leading limb sidestep and Figure 6.22 summarises the data for the trailing limb sidestep. Overall, the results for SSL were found to be largely in line with those by Neptune et al. (1999). When discussing muscle activations it is important to remember that there is a time delay between the activation of a muscle and the development of force by the muscle which, according to Winters and Stark (1988), has a duration of 20 to 100 ms (~7-34% of the swing and ~8-39% stance phase).

6.4.2 Hip adduction/abduction

AL was active from about mid-swing to mid-stance, coinciding with peak hip abduction. The data indicates a strong adductor moment and positive power in early stance, which suggests that AL decelerates the abducting limb from mid-stance and initiates limb adduction in early stance. It is thought that this contribution of the muscle for propulsive force production may increase the likelihood of strain injury by exposing the muscle to larger than normal force. It is possible that this may be exaggerated when using a larger jump in combination with the lateral step, however no definitive can be provided at this point. It is thought that AL then acts isometrically to stabilise the hip as the upper body moves over the adducting limb. From about mid-stance Gmed becomes activated coinciding with the generation of a weaker abductor moment and negative power indicating that Gmed isometrically stabilised the hip joint together with AL in mid-stance and was involved in abducting the limb from early to mid-swing where muscle activity appeared to peak. These findings are in contrast to those by Neptune et al. (1999) who state Gmed activation ~20% prior to impact and offset shortly before toe-off, in their investigation of the right limb during the change in direction from a right- to left-hand lateral sidestep. Hip adduction/abduction angles in the current investigation were larger than those reported by Neptune et al. and instead of purely stabilising the hip isometrically, Gmed appeared to prevent excessive adduction of the thigh in stance and actively abduct the limb from toe-off to mid-swing. It is therefore suggested that it acted to move the swinging limb laterally and stabilise the pelvis on the weight bearing leg from early to mid-stance.
In the trailing limb AL was activated in early stance rather than mid-swing and active until early swing. It was therefore active from peak adduction to peak abduction and is thought to provide isometric hip stabilisation until mid-stance and eccentrically protects the hip from excessive abduction in late stance. Similarly, Gmed was activated just prior to stance rather than mid-stance and was active until early swing thereby decelerating the limb before stance. The co-activation of AL and Gmed throughout stance indicates that both of these muscles were involved in stabilising the hip joint in stance as the body moves over the limb, preventing the limb from excessive abduction in late stance. The kinetic data supports the previous findings indicating that the leading and trailing limb perform distinct roles in these gaits where the leading limb creates a significantly larger adductor moment and the trailing limb creates a significantly larger abductor moment in both the SS and XS. Furthermore, the timing of muscular activations in the leading limb, with smaller durations of co-contraction, indicates a contribution of AL to force generation in early stance, rather than a purely stabilising function.

The XS appeared to cause an earlier activation of AL in the trailing limb before stance, which may be linked to the greater adduction of the trailing limb before stance. Overall however, the function of both AL and Gmed in the leading and trailing limb closely reflected that described for the SS above. Interestingly there were no significant differences in the kinetic parameters between the SS and XS for the leading and trailing limbs. However, as mentioned above, the total range of motion of the leading and trailing limb was larger when using the XS with significantly larger hip adduction in both the leading and trailing limb and a significant reduction in peak hip abduction in XSt.

Compared to the average EMG profiles for walking and running, in Gazendam and Hof (2007), these data show that the activity of Gmed in the trailing limb relates well to the function in walking and running, while the leading limb displays a distinct activation strategy. Direct comparison of AL is not possible since the authors investigated the function of adductor magnus in walking and running. It appears however, that the activation strategy of the adductor muscles in the lateral tasks is unique to the movement.
6.4.3 Hip flexion/extension

The data for Gmax was largely in agreement with the data by Neptune et al. (1999). In both the leading and trailing limb Gmax can be seen to be activated before stance with activation ending in late stance. Gmax appears to extend the hip from ground contact as the leading limb is maximally abducted and the trailing limb is maximally adducted. During stance there is a strong hip extensor moment as the hip extensor muscles absorb power in the first half of stance and provide positive power as they act concentrically from mid-stance to extend the hip. BF is a further muscle with the potential to contribute to hip extension. BF was activated in mid-swing with peak activity at impact, coinciding with peak hip and knee extension. This suggests that the primary role of BF is that of a stabiliser of the hip, preventing excessive hip flexion and of the knee, preventing excessive extension and transferring power from the knee to the hip to assist Gmax in extending the hip during stance. Furthermore, RF, like BF, is another two-joint muscle involved in both hip flexion and knee extension. However, it is unlikely to act as a hip flexor since its’ peak activity coincides with peak hip extension.

In the trailing limb BF appeared to be activated later in swing in both the SS and XS, which is likely due to the adoption of a more flexed hip position at impact compared to a more extended hip used by the leading limb. The finding of a significantly larger initial flexor moment in the trailing limb, which is present in both the SS and XS, is thought to be related to the function of the trailing limb which is in fact the first limb to make contact with the ground following the aerial phase. The major force absorption is therefore performed by the trailing limb. Furthermore, the larger hip extensor moment in the leading limb, which is significantly larger in the XS reinforces the idea that it is the leading limb that acts predominantly in creating propulsive force, while the larger range of motion of the trailing limb is not thought to relate to a greater contribution in terms of propulsive power, but is related to the greater extension of the trailing limb prior to ground contact of the leading limb.

In the trailing limb Gmax appeared to become activated earlier (mid-swing) and BF appeared to be active for a longer time in the trailing limb (until early stance). The causes for these differences are thought to be related to the significant reduction in
mean peak hip flexion angle and increased hip extension in the trailing limb. The hip of the trailing limb therefore remained more extended throughout stance, which explains the significantly reduced hip extensor moment as well as reduced positive and negative power in the trailing limb compared to the SS where the hip is involved in larger hip flexion and in more absorption and generation of power at the hip joint.

Compared to the activation patterns in walking and running from the literature, Gmax of the leading limb behaved in a similar manner to that seen in walking, while it behaved in a similar manner to that seen in running at the trailing limb. However, in the lateral movement Gmax did not display the initial activation peak observable in running. Furthermore, the activity of BF was very similar between the leading and trailing limb, showing a distinct contribution to systematically extend the hip through the coordinated activity with Gmax.

6.4.4 Knee flexion/extension

As mentioned above the role of BF appears to be that of a hip extensor rather than a knee flexor. However, the activation of BF from mid-swing with peak activity during early stance would suggest that it also acts to stabilise the knee joint and prevent hyper extension of the knee before impact.

RF activity started in late swing and finished in early swing. There was some variability in activation shapes with some participants showing a continuous trace of activity with others showing two distinct activation periods at impact and in late stance. The activation of RF therefore, coincides with the start of knee flexion at weight acceptance. Until mid-stance, RF appears to eccentrically stabilise the knee absorbing the load of the body during stance, as indicated by the strong extensor moment and the large negative power peak. From mid-stance RF extends the knee concentrically resulting in strong positive power at the knee joint. This data supports the findings by Neptune et al. (1999) indicating that RF appears to be transferring power from the hip extensors to the knee joint.
In the trailing limb RF appeared to be active for a shorter period of time, starting before stance and ending in late stance rather than after toe-off. This is thought to be further indication of the fact that the trailing limb is not the main source of propulsive force generation but that RF is primarily involved in absorbing the force of impact resulting from the aerial phase. This absorption of the impact force also appears to be the primary cause for the significantly larger mean peak knee extensor moment and positive power in the trailing limb during the SS and XS due to the increased range of motion and significantly larger peak knee extension angles of the knee of the trailing limb compared to the leading limb.

Furthermore, RF appeared to become activated earlier in swing in trailing limb XS compared to the SS reflecting an earlier involvement of the muscle in extending the lower limb while it moves beyond the midline of the body and preparing the limb for contact with the ground. While the function of RF appears to be in line with the function proposed by Montgomery et al. (1994) and Jacobs et al. (1993) in running, the comparatively longer activity until early stance in the leading limb during the lateral movements may be seen as evidence of a larger contribution of RF to knee extension, similar to the function of RF in jumping (Jones and Caldwell, 2003). Considering the generally late onset of the second peak of RF activity in a number of participants, prior to toe-off, and the dual role of this biarticular muscle, this may also indicate that RF is involved in hip flexion of the leading limb in early swing. This appears to be a reasonable assumption since the antagonistic hip extensor Gmax is not active at this stage in the gait cycle. The lack of a stabilising function of Gmax, as described by (Rasch, 1989), would allow RF to act as a hip flexor, creating increased ground clearance, while Gmed abducts the limb.

6.4.5 Ankle flexion/extension

The activity of GAS was virtually identical in all gaits. Activation occurred before impact and stopped in late stance. GAS therefore prevented excessive dorsiflexion in mid-swing and plantarflexed the foot at impact, absorbing the impact and thereby reducing joint loading. However, as mentioned in Chapter 6 the leading and trailing limb contributed differently and while the trailing limb consistently used the front-
and mid-foot to support the bodyweight at impact and during stance, the rearfoot (heelstrike) was used frequently by the leading limb, resulting in a laterally rotated and dorsiflexed alignment of the foot at impact. Five of the nine participants adopted the heelstrike for the SS and all participants used the heelstrike for the XS. This is thought to be attributable to training to prevent the mechanisms of ankle instability at impact as reported by Wright et al. (2000) where excessive plantarflexion and supination were seen to increase ankle sprain injury occurrence. In the leading limb therefore, the role of GAS in plantarflexing the ankle before and at impact to absorb the impact or reduce joint loading is dependent on the individual participant.

In the trailing limb, GAS plantarflexed the foot before impact and controlled dorsiflexion of the loaded ankle joint. The muscle is thought to act isometrically, absorbing impact forces and reducing joint loading, and concentrically as suggested by Lichtwark and Wilson (2006), Jacobs et al. (1993) and Fukunaga et al. (2001) thereby optimising the muscle tendon apparatus, storing elastic energy during weight acceptance and contributing to force generation by working together with RF, Gmax and BF. The data furthermore indicates that the peak negative power at the ankle joint of the trailing limb was significantly larger than that of the leading limb again supporting the previously stated idea that the trailing limb is involved primarily in energy absorption following the aerial phase.

TA displayed the greatest amount of variation in all gaits with a tendency to be active for large periods of swing and stance with consistent onset in late stance for all lateral movement scenarios. TA is thought to prevent excessive plantarflexion in the push-off phase of stance to ensure sufficient clearance between the front foot and the ground. Furthermore, in the leading limb it can be argued that TA contributes to dorsiflexion as the foot prepares for heel-strike at impact and acts to stabilise the ankle joint in stance. In the trailing limb it was thought that TA acts to prevent misalignment of the foot at impact by stopping the foot from excessive plantarflexion and pre-stretching GAS in preparation for its activity in stance. In both the leading and trailing limb therefore, the activity of GAS closely matches that seen in running. This can be said also for TA, however, TA does not appear to display the activation burst before toe-off, which is a consistent characteristic of the lateral gaits.
Figure 6.21 Summary of the kinematic, EMG and kinetic data for the leading limb during the SS. Joint angles are expressed in degrees and joint moments and powers are expressed in relation to body mass (Nm kg⁻¹ and W kg⁻¹ respectively). The rectified EMG trace, as used for the determination of excitation onset and offset timings, is presented. Reference data of averaged EMG profiles for walking at 1.25 ms⁻¹ (broken lines) and running at 3 ms⁻¹ (solid lines), adapted from (Gazendam and Hof, 2007), are presented for comparison. Comparative data for adductors is in reference to adductor magnus rather than longus as used in this investigation.
Figure 6.22 Summary of the kinematic, EMG and kinetic data for the trailing limb during the SS. Joint angles are expressed in degrees and joint moments and powers are expressed in relation to body mass (Nm kg\(^{-1}\) and W kg\(^{-1}\) respectively). The rectified EMG trace, as used for the determination of excitation onset and offset timings, is presented. Reference data of averaged EMG profiles for walking at 1.25 m s\(^{-1}\) (broken lines) and running at 3 m s\(^{-1}\) (solid lines), adapted from (Gazendam and Hof, 2007), are presented for comparison. Comparative data for adductors is in reference to adductor magnus rather than longus as used in this investigation.
6.5 Conclusion

Through the recording of kinematic, kinetic and EMG data it was possible to identify the co-ordination and contribution of the major superficial musculature of the lower limbs to the performance of the two selected lateral movement tasks. The data indicates a number of differences and similarities in activations at the leading and trailing limb and in comparison to walking and running. Hip adductor and abductor activation strategies in the trailing limb appeared very similar to those in running, indicating a stabilising function, while their coordinated action in the leading limb appeared unique to the movement and indicates an early contribution to adductor force generation of AL. It is thought that this may expose to muscle to higher than normal stress, particularly when performing larger jumping movements in combination with the sidestep. While largely supporting the findings of muscle activations for hip and knee flexion and extension and particularly ankle plantarflexion and dorsiflexion, in running and those presented by Neptune et al. (1999) for lateral sidestepping, some deviations in patterns, particularly at the leading limb were observed. The findings for muscular activations and moment patterns therefore indicate a coordinated sequence that optimises extensor force production.
Chapter 7

A comparative analysis of the metabolic requirements of atypical gaits used in badminton

7.1 Introduction

The game of badminton poses distinct physiological requirements on the athlete and high-level badminton along with squash is recognised for recording "the highest heart rate values" (Lees, 2003) amongst racket sports. During a game the athlete performs all movements on a restricted playing field of relatively small proportions (~35m² for each half of a singles court (Badminton England, 2006) compared to ~98m² for a tennis court (Elliott Courts, 2007). The video data analysed for the notational analysis, performed in Chapter 4, supports the view that the athlete is constantly moving up, down and from side to side of the court in an attempt to play and return the shuttle and outmanoeuvre the opposition. To achieve this goal the athlete uses specific movement strategies, which include the use of typical gaits, including walking (Walk) and running (Run), but also atypical gaits including lateral sidestepping (SS) and crossover stepping (XS) gaits as well as skipping (Skip), hopping, lunging and jumping. These atypical gaits are short in duration, used repetitively throughout a match and have been shown to make up a considerable proportion of the gait repertoire employed by the athlete during rallies. The previous chapters have investigated the effect of using badminton-specific lateral gaits on the biomechanics of the lower limbs. This data has resulted in strong indications of a mechanically inefficient mode of transport due to the asymmetric contribution of the leading and trailing limbs to the gait cycle of the lateral gaits and an apparent reduction in the efficiency of the use of elastic elements. However, this data does not contain any information about the metabolic demands of these movements. It will be the focus of this chapter to determine the metabolic demands of a selection of typical and atypical
gaits used in badminton, in order to determine the demands of these movements and the possible contribution of atypical gait use to the high physiological demands of badminton.

The energy requirements of badminton-specific training and game play have been investigated by a number of researchers, including Hughes (1994), Dias (1994), Majumdar et al. (1997), Lees (2003), and Cabello and Gonzalez-Badillo (2003) who focused predominantly on training at the elite level, in an attempt to quantify the demands of the sport and improve training techniques (summarised in Chapter 2.4.2). These studies have shown that simulated competition and training exercises using a shuttle and without a shuttle caused very high heart rate responses and considerable stress of the muscular and cardiovascular system (Majumdar et al., 1997). Heart rates during practice games were greater than 80% of the measured maximum heart rate for over 85% of the playing time (Hughes, 1994), while an average heart rate of 85% of the maximum heart rate is reported for Indian badminton players (Majumdar et al., 1997). Reilly (1990) states that isometric contractions during the ready position as well as short recovery periods, compared to tennis, may contribute to the high physiological demands. Furthermore, neurophysiological factors, such as stress hormones, are suggested as causes for the high average and maximum heart rates, which, based on the findings of comparatively low blood lactate concentrations, appear too be "above that provoked by the actual effort" (Cabello Manrique and Gonzalez-Badillo, 2003). However, no definitive causes for the high physiological demands of the sport have been identified. Despite the knowledge gained on the demands of the sport in training and game play scenarios, there is limited information available on the possible contribution of atypical gaits used in the game on the metabolic demands of the sport.

The review of the literature has shown that there are distinct differences in the energy expenditure in walking and running (see Chapter 2.4.1) and that deviation from the norm result in larger metabolic demands. Therefore, walking at an optimal speed results in much lower metabolic demands than running (Margaria et al., 1963, Margaria, 1976, Ralston, 1960, Pugh, 1970), while larger knee flexion increases the metabolic demands (Waters, 1992). Therefore, the form of locomotion chosen, affects the physiological demands. This would suggest that the form of transport chosen in
badminton may contribute to the total demand of the sport. Lateral movements are frequently and repeatedly used within the game for optimised body movement. The specific movement demands of the sport, the inclusion of specific gaits, such as SS and XS, and their respective physiological demands may therefore determine the basic demands of the sport. Furthermore, according to the summary of the literature by Sazioński et al. (1987) and data presented in Miyashita (1978), experienced athletes display reductions in metabolic and mechanical demands during the performance of a variety of different forms of locomotion, compared to less experienced athletes. This was also seen in Chapter 5, where experienced participants displayed generally reduced vertical hip displacement during the SS and XS. This may be argued to be a reflection of the effect of increased exposure time to a movement and a resulting adaptation of technique which may cause a reduction in metabolic demands. Similar mechanisms may also be observable for atypical, badminton-specific gaits.

It is hypothesised that sidestepping and crossover-stepping gaits result in significantly higher metabolic demands than typical gaits. The aims of this investigation were to examine the energy demands of a selection of gaits used in badminton through the use of indirect gas analysis in order to compare and contrast the metabolic demands. This information is vital for the purpose of determining the contribution of atypical gaits to the metabolic demands of badminton. Furthermore, it was the aim to examine differences in movement dynamics and perceived movement demands between typical and atypical gaits. A final goal of the study is to examine the influence of skill on the metabolic demands of the SS and XS within experienced than inexperienced participants which may indicate adaptations of the athlete to the demands of the sport-specific locomotor tasks.
7.2 Methods

7.2.1 Participants

With approval by the local ethics committee nine experienced (7 male and 2 female) and six inexperienced (3 male and 3 female) badminton players participated in the study (see Table 7.1 for a summary of the criteria used for classifying experienced and inexperienced players). All participants were recruited from the Loughborough Students’ Badminton club at Loughborough University and informed of the general aims and requirements of the experiment before participating. Participants were asked to complete a general health questionnaire (Appendix A1.1) and no participants with existing bone or physiological impairments were included in the study. Furthermore, only regularly active participants, familiar with SS and XS movements, were recruited as it was thought likely that non-active individuals may experience difficulty in performing the exercise protocol. Anthropometric measurements were taken for height, weight and leg length (from the greater trochanter to the lateral malleolus). The data were required for the data analysis described below. The data are summarised in Table 7.1.

| Table 7.1: Explanation of the classification of skill used in this investigation and summary of subject information |
|---------------------------------|----------------|---------|---------|
| Experienced Participants       |                |         |
| Players were classed as experienced if they played considerably more than 6 hours of badminton per week and played for the University 1st or 2nd team. Level of expertise varied from national to international level. | Age (yrs) | 20.33   | 1.32    |
|                                | Height (m)     | 1.79    | 0.08    |
|                                | Total Body Mass (kg) | 73.57   | 11.43   |
|                                | Leg Length (m) | 0.92    | 0.06    |
| Inexperienced Participants     |                |         |
| Players were classed as inexperienced if they played no more than 6 hours of badminton per week and at no higher than average recreational level. | Age (yrs) | 21.5    | 3.15    |
|                                | Height (m)     | 1.76    | 0.1     |
|                                | Total Body Mass (kg) | 77.48   | 11.54   |
|                                | Leg Length (m) | 0.95    | 0.07    |

7.2.2 Data Collection

A portable breath-by-breath gas analysis system (MetaMax 3B, Cortex Biophysik GmbH, Germany) was used for the sampling and analysis of expired gases and gas
volumes (see Chapter 3.3 for a detailed description). The system includes a lightweight portable gas sampling unit worn by the subject around the shoulders and chest and a receiver unit for wireless data transmission and relay to a PC for data analysis and processing. The portable unit weighs 2.7 kg and includes a silicone facemask with integrated flow-rate turbine and expired air sampling line, which connects to the battery-powered gas analysis and transmitter unit worn on the participants' chest. The turbine was calibrated regularly with a 3 l syringe using ambient air and a standard calibration gas mixture following the manufacturers' instructions.

The system was used in the wireless configuration with live data capture and direct, on-screen data readout. This set-up aided in identifying gas equilibrium and judgement of the necessary exercise and rest durations. Use of the MetaMax system allowed for comparatively unrestricted movement for the participant in contrast to a traditional Douglas Bag system which would require carrying an air sampling bag and allowed measurements to be taken quickly and directly throughout the free moving activities without the need for post exercise gas analysis.

7.2.3 Experimental Design

7.2.3.1 Pilot Study

A pilot study was performed using one recreationally active male badminton player (age = 24 years, body mass = 92 kg) performing walking (Walk), running (Run), skipping (Skip), lateral side-stepping (SS) and crossover stepping gaits (XS) at a range of speeds on a treadmill. All lateral movements were performed in the direction of the dominant limb where dominance was deduced from left or right hand racket arm preference. This method of determining limb dominance was used for all other participants in the main study as it is the racket arm that determines body orientation and movement on the badminton court. The observations from Chapter 4 show that particularly the XS movements are performed nearly exclusively in the direction of the dominant limb. Metabolic energy requirements were determined via the breath-by-breath gas analyser (MetaMax 3B) throughout. The pilot showed that performing SS
and especially XS gaits on a treadmill was not a viable option for the main study. The available treadmills did not have belts with a wide enough running surface to allow for sufficient space for the feet to perform the movement without the risk of stepping onto the stationary sides of the treadmill. Stepping beyond the boundaries of the belt could have resulted in a trip and fall of the participant. A harness system was considered but, due to the possible effect of a harness on the performance of the gaits, it was decided to perform the main study free moving, on a marked indoor oval track (16.5m x 7.1m). This had the disadvantage that speed of locomotion could not be controlled but the advantage that participants could move with a great degree of freedom and choose their most comfortable speed. Furthermore, this reduced the risk of tripping and allowed for the performance of continuous locomotion.

7.2.3.2 Main Study

The participants for the main study were equipped with the portable gas analyser and asked to sit down during the measurement of their resting metabolic rate (RMR). Thereafter, the participants were given a badminton racket to hold in front of their body during the exercise (mass = 95g) and were instructed to walk at a self-selected, preferred speed, followed by a run and a skipping gait at their preferred speed. This was followed by a slow and medium effort SS (SSₜ and SSₘ), a slow and medium effort XS (XSₜ and XSₘ) and finally a fast effort SS (SSₖ) (see Table 7.2 for the exercise plan). The use of a badminton racket may have influenced energy expenditure however the effect is thought to be minimal due to the low weight of the racket. The advantage of using a racket was in recreating a more badminton-specific environment by restricting the assistance of the arms. It was anticipated that in a normal setting arm-swing may contribute to the performance of lateral movements. Ortega et al. (2007) report reduced metabolic costs when using arm-swing in gait, compared to a fixed arm position. The video footage of competitive badminton matches, analysed in Chapter 4 however suggests that in badminton the arms remain relatively steady and focused on guiding the racket to the shuttle. The participants were instructed that the medium effort should be equivalent to their preferred speed for the particular gait and slow and fast speeds should correspond to slower and faster speeds than their preferred speed of locomotion for the particular gait. The SS and XS
gaits consisted of lateral movements as described in Chapter 4 of this thesis. The XS could not be performed at a fast speed since it became unstable at high speeds. Furthermore, fast speed SS was physically very demanding and showed high physiological responses. Therefore, in order to sample gait characteristics at high speeds, the fast SS was performed for a short period of time only, which means that no steady state metabolic data was available for this movement. Participants were encouraged to maintain equal speed during gait repeats by verbally relaying their lap times as a reference.

<table>
<thead>
<tr>
<th>Exercise Protocol</th>
<th>Order</th>
<th>Exercise</th>
<th>Speed</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walk</td>
<td>1</td>
<td>Medium</td>
<td></td>
</tr>
<tr>
<td>Run</td>
<td>2</td>
<td>Medium</td>
<td></td>
</tr>
<tr>
<td>Skip</td>
<td>3</td>
<td>Medium</td>
<td></td>
</tr>
<tr>
<td>SS</td>
<td>4</td>
<td>Slow</td>
<td></td>
</tr>
<tr>
<td>SS</td>
<td>5</td>
<td>Medium</td>
<td></td>
</tr>
<tr>
<td>XS</td>
<td>6</td>
<td>Slow</td>
<td></td>
</tr>
<tr>
<td>XS</td>
<td>7</td>
<td>Medium</td>
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</tr>
<tr>
<td>SS</td>
<td>8</td>
<td>Fast</td>
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</tr>
</tbody>
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Each gait in the protocol was performed for 3-5 minutes, until gas equilibrium was reached (steady state) and automated interval gas measurements were continued for another 15 - 30 seconds for steady state readings. Steady state was taken as the point where traces for both heart rate and the concentration of carbon dioxide in the exhaled air reached a plateau and maintained consistent levels. These data traces were displayed live throughout the exercises and inspected to determine when steady state occurred. After completion of gas sampling for a particular gait the participant was then asked to stop and rest and complete a post-exercise evaluation sheet to sample the perceived gait difficulty and risk of injury (Appendix A1.3). This feedback was gathered with a view to quantifying the effect of atypical compared to typical gait use for the experienced and inexperienced participants.

For the duration of the exercises step numbers and lap times were manually recorded. This data was used for estimations of mean speed of locomotion, stride length and distance travelled. Between exercise trials participants were encouraged to take prolonged rest and stretch the muscles of the lower limb to counteract muscular discomfort. Four participants were not able to complete the entire protocol due to
either equipment malfunctions or excessive heart rate responses to SS and XS at higher speeds. Furthermore, strapping of the lower left leg had to be supplied to counteract shin splints, which was a sporadically occurring condition for one participant. After this intervention the participant was able to perform the exercise without discomfort.

7.2.4 Data Analysis

One continuous trace of liberated energy (kcal) for the duration of the experiment was recorded in the gas analysis software provided with the MetaMax 3B gas analyser (MetaSoft). Raw energy expenditure data was then imported into the spreadsheet application Excel (see Figure 7.1) and analysed in the following fashion:

- In order to distinguish between the different activities, virtual markers were used to mark the end points of activities.
- To calculate metabolic power, the data points for the final 10 to 15 seconds of the liberated energy trance for each gait were plotted against time. A regression line was plotted for each set of data points.
- The slope of the line was taken to be equivalent to energy expenditure for a particular gait at metabolic equilibrium. The units were thereafter converted from Kcal sec\(^{-1}\) to J sec\(^{-1}\) (where 1 Kcal = 4186.8 J and 1 J sec\(^{-1}\) = 1 Watt (W)). Energy expenditure is therefore equivalent to metabolic power (P) and will be referred to as such.

![Liberated Energy and Heart Rate vs Time](image)

Figure 7.1 Visualisation of the continuous trace recorded for one participant performing the pilot exercise protocol on a treadmill. The dips visible in the heart rate trace are caused by the short period of rest taken during the changeover from one exercise to the next. Vertical lines represent the virtual markers used to identify exercise end-points.
In order to identify the relationship between metabolic power and distance covered, metabolic power was divided by the speed of locomotion (ms⁻¹). The resultant value (cost of transport, CoT) expressed in Joules per meter (Jm⁻¹), reflects of the metabolic cost of travelling a set distance. Furthermore, for the purpose of comparing CoT between groups a distinction was made between the gross and net value. The gross value includes the basic metabolic rate at rest. In order to calculate the net metabolic rate, the resting value, determined while sitting on a chair, was subtracted from the gross value. The net value therefore represents the change in response to exercise only. Net metabolic P was therefore derived by subtracting P at rest from P during activity (gross P) and the same principle was applied to CoT (Saziorski, 1987). Therefore,

\[ P_{\text{Gross}} - P_{\text{Rest}} = P_{\text{Net}} \]

Based on observations in the literature and findings in this study, metabolic values were furthermore normalised for body mass in order to reduce inter-individual variability and allow for comparisons between groups. The resultant values were therefore expressed as net Pkg⁻¹ and net CoTkg⁻¹.

7.2.5 Statistics

In order to verify the influence of body mass on gross CoT and the influence of speed on gross P correlation analysis, using Pearson's correlations, was performed in the statistical analysis software SPSS. Significance was taken as p<0.05.

Statistical analysis of the effect of gait on net CoTkg⁻¹ and net Pkg⁻¹ at preferred speed was performed using an ANOVA analysis of variance with repeated measures in the statistical analysis tool SPSS. Therefore, comparisons were performed for walking, running, skipping, SSs, SSm, XSs and XSm. This approach did not take into account the effect of a specified speed but allowed for a general comparison of the effect of gait at a preferred magnitude.
The effect of skill on the metabolic cost of the lateral movement tasks was performed for the preferred medium speeds, using an ANOVA test of variance including speed as a covariate. The level of statistical significance was taken as $p<0.05$ throughout.
7.3 Results

A summary of the results of the change in gross CoT for the investigated gaits is presented in Figure 7.2. CoT for SS and XS was generally high, particularly at low speeds.

![Figure 7.2 Data summary for all participants of gross CoT (adjusted for body mass) for Walk, Run, Skip, SS and XS over a range of speeds. The units used are Joules/kilograms/meter for the CoT and meters/sec for speed of locomotion.](image)

7.3.1 Effect of body mass

A significant linear relationship (p<0.05) was identified for walking, running and SS gaits when investigating the effect of body mass on gross P at selected speed ranges (see Figure 7.3). The number of participants and selected speed ranges represent a compromise between a narrow speed range and maximised participant numbers. These findings were along the same lines as findings by previous researchers (Mahadeva et al., 1953, Malhotra et al., 1962, Wyndham et al., 1971, Van Der Walt and Wyndham, 1973) who report linear increases for P with body mass for walking and running. No definite significant relationship however was observed for XS and
skipping (p= 0.08 and 0.25 respectively) despite XS approaching significant levels. It can be argued that, due to the similarity between XS and SS and the low p-value for XS, it is very likely that with an increased sample size a similar finding would be observed. The linear relationship of body mass and gross P shows that body mass had an effect on the energetics of not only typical but also atypical gaits. Further investigation of the effect of gait on the CoT and P was therefore performed on mass adjusted values by dividing P and CoT values by the participants’ body mass.

Figure 7.3 Graph displaying the change in gross power (in Watts) with increasing body mass at the indicated speed ranges. Raw data points and linear regression lines are presented for the change in gross P with increasing body mass. Significant linear relationships between body mass and gross power were observed for walking, running and SS gaits with XS approaching significance. * indicates reference data for walking and running adapted from van der Walt and Wyndham (1973).

7.3.2 Cost of Transport

Collectively, net CoT values (adjusted for body mass) for both walking and running were seen to behave in line with the data from the literature (see Figure 7.4).
At preferred speed net CoT kg$^{-1}$ showed significantly lower values for walking compared to all other gaits. Furthermore, significant differences were observed for run/skip, run/SSs, run/SSm, run/XSs, skip/SSs, skip/XSs (see Figure 7.5). No significant differences were detected between skip and SSm or XSm (p=0.71 and p=0.768 respectively). Furthermore, there were no significant differences between SSs and XSs or SSm and XSm (p= 0.829 and p= 0.608 respectively). Run/XSm displayed low p value (p= 0.063) indicating a similar relationship as that of Run/SSm.

Figure 7.5 Summary of the results of the statistical analysis of net metabolic cost of transport per kg body mass at preferred speed of locomotion. $\{, a$ and $b$ indicate statistically significant differences between gaits, where $a= $ significantly lower net CoT for Walk compared to all other gaits; $b=$ significantly lower net CoT for Skip compared to SSs and XSs; and $\{$ indicates differences between the remaining gaits.
7.3.3 Metabolic Power

Figure 7.6. Net metabolic power for the investigated gaits in relation to speed. Solid lines represent linear regression lines for the data on run, SS and XS. Data is normalised for participant’s body mass (Watts per kg).

Net metabolic power for the investigated gaits is summarised in Figure 7.7. Comparison of net $P_{kg^{-1}}$ at preferred speeds showed significantly lower values for walking in comparison to all other test gaits (see Figure 7.7). No significant differences were observed for run/skip ($p = 0.379$), run/SSm ($p = 0.865$), run/XSm ($p = 0.464$), SSs/XSs ($p = 0.249$), SSm/XSm ($p = 0.336$) or skip/SSm ($p = 0.426$) and skip/XSm ($p = 0.247$).

Figure 7.7 Summary of the results of the statistical analysis of net metabolic power per kg body mass at preferred speed of locomotion. a, b, indicates statistically significant differences between gaits, where a = significantly lower net $P$ for Walk compared to all other gaits; b = significantly higher net $P$ for Skip compared to SSs and XSs; and { indicates differences between the remaining gaits.
7.3.4 Stride Length

Stride length (SL) was derived by dividing the distance travelled for one lap of the track by the mean number of strides taken per lap during the sample period. The resultant SL data for walking and running was in line with the data from the literature (Figure 7.8b). As expected, stride length was lowest in SS followed by XS. Both walking and running at preferred speed displayed higher SL values than SS and XS at the comparable speed ranges while skipping at preferred speed displayed the highest SL values.

For the purpose of statistical analysis of the effect of skill on SL for SS or XS, the SL of each individual was divided by their leg length (LL) to obtain a SL/LL ratio to compensate for possible effects of variations in LL. An ANOVA test of variance was performed for the analysis of the effect of skill on the SL/LL ratio, including speed as a covariate to compensate for the effect of speed on SL variations. The results showed no significant effect of skill on SL for the experienced and inexperienced players for SS, SSm, SSf, XS, or XSm (p= 0.864, 0.367, 0.547 and 0.335 respectively). In all cases a significant effect of speed in SL/LL was observed (p<0.05 for SS, SSm, SSf, XS, and XSm).

![Graph a)](image1)
![Graph b)](image2)
![Graph c)](image3)

Figure 7.8 Effect of speed of locomotion on the stride length (SL). Graph (a) summarises the raw data. Graph (b) comparison the data from this experiment to data on walking and running adapted from Minetti and Alexander (1997). Graph (c) summarises the data for SL scaled for individual leg length (LL).
7.3.5 Participant Feedback

The participant feedback on perceived difficulty and perceived risk of injury, summarised in Figure 5.11, shows a distinct change in perception to gait use and speed of locomotion. Walking, running and skipping at preferred speeds were regarded as generally low in difficulty and low risk. SS was perceived as more difficult with a clear increase in both difficulty and risk of injury with increasing speed of locomotion. XS, in relation to the magnitude of the movement, was perceived as the most difficult task in both difficulty and the risk of injury. XS at higher speeds was perceived to be particularly risky which is reflected in the exclusion of high speed XS from the study.

![Figure 7.9: Feedback scores on a) perceived difficulty and b) perceived risk of injury when performing the movement task.](image)

7.3.6 Effect of skill on CoT and P

Skill appears to affect the net CoT for SS and XS Figure 7.10. The results from the ANOVA analysis indicated a significant effect of skill on net CoT between SSm and XSm, which remained significant when including speed as a covariate. CoT for inexperienced participants appears to be lower than that of experienced participants. At the medium speed of locomotion experienced participants appear to use 37% more energy during the SS and about 8% more energy in the XS. When examining the data for individual participants (Figure 7.12 b), two participants stand out due to their particularly low CoT. These participants display comparatively low values for all investigated gaits. Without the influence of these participants, differences in CoT between the groups are reduced to 21% for the SS and equivalent values for the XS.
Figure 7.10 (a) Net CoT means for experienced and inexperienced participants for SS and XS at self-selected speeds. (b) Net P means for experienced and inexperienced participants for SS and XS at self-selected speeds. (c) Mean speed of locomotion (meters per second) for experienced and inexperienced participants. Units for net cost of transport are Joules per kg body mass per meter (J/kgm) and net P is expressed in Watts per kg body mass.
7.4 Discussion

The underlying hypothesis of this investigation was that sidestepping and crossover-stepping gaits result in significantly higher metabolic demands than typical gaits thereby contributing to the high physiological demands of the sport. The aims were to identify the metabolic demands of a selection of sport-specific gaits in order to quantify the differences between gaits and the effect of level of skill. This information may indicate the advantages and disadvantages of gaits use within the context of badminton and lead to a better understanding of the demands of the sport.

Before discussing the findings in detail, some of the issues of measurement of metabolic energy expenditure in free moving exercises and the methodology employed in this study should be mentioned at this point. The findings of the pilot study highlighted the risks to the participants and equipment created by the width restrictions of the available treadmill belts, particularly for the performance of the XS task. Therefore, the experiment was performed free moving on an indoor oval track. This method avoided the adverse effect of treadmill artefacts or weather conditions on the energetic demands of the gaits as highlighted by Ralston (1960) and Pugh (1970) respectively. Furthermore, it allowed for continuous motion and the maintenance of the selected speed throughout the exercise without repeatedly stopping and starting the movement, which would have occurred on a straight line track. Constant acceleration and deceleration would have prohibited a true comparison of gaits due to the lack of a consistent level of effort by the participants. The oval track therefore presented the best possible compromise at the time in terms of repeatability and controllability of the required gait tasks. Nevertheless maintaining speed on an oval track is affected by the corners of the track and the overall speed of locomotion, used for the purpose of data analysis, is equivalent to the best possible speed estimate for the average speed of locomotion during the sample period.

The findings from this study support the observations that different gaits evoke very different physiological responses with changes in speed of locomotion. While the metabolic demands of self-selected walking and running were in agreement with the literature (see Figure 7.4 and Figure 7.6), there appears to be a marked difference in
the response of the body to lateral movement tasks. Net CoT for SS and XS at slow and medium speeds was significantly higher than that of walking or running and only with increasing speed of locomotion did CoT for SS and XS decrease sufficiently to appear metabolically similar to relatively slow running (Figure 7.4). Whereas the relationship between net CoT and speed of locomotion assumes the form of a quadratic equation for walking, with an intermediate metabolically optimal speed (Hreljac, 1993, Margaria et al., 1963, Cavagna and Kaneko, 1977, Saibene and Minetti, 2003) and remains more or less constant over large speed ranges for running (Cavagna et al., 1976, Kram and Taylor, 1990, Margaria et al., 1963), net CoT for SS and XS appeared to decrease linearly over the performed range of speeds (see Figure 7.4). Interestingly, the findings for metabolic P showed SS and XS to be similar to running in both the shape of the increase in net P with speed of locomotion and the magnitude of net P between speeds of about 2 and 2.5 ms\(^{-1}\) (Figure 7.6). Over comparable speed ranges, net P for SS and XS was slightly larger than for running. However no significant differences were detected for preferred speed SS, XS and running (Figure 7.7). This indicates that the metabolic differences between the SS, XS and running are likely related to the differences in speed rather than the result of larger metabolic power demands of the lateral movements.

There are a number of possible contributors to the discrepancies and similarities in the metabolic demands between the tested lateral movements and walking and running. As seen in the current investigation, as well as in Chapter 6, SS and XS displayed generally lower stride-length values compared to walking, running or skipping (Figure 7.8), due to the restriction of limb excursion by the midline of the body. In the XS, stride-length was slightly larger due to the increased travel of the trailing limb beyond the midline of the body. Particularly at low speeds SS and XS stride-lengths were smaller than in walking and the increase in stride-length with speed occurred rather slower than in walking or running. This means that for a given distance, the stride-frequency for SS and XS was higher compared to walking or running. It may therefore be argued that the increased stride frequency, in turn, results in more frequent muscular activations, compared to the typical gaits. Considering the co-contraction of the adductor and abductor muscles during the SS and XS, observed in Chapter 6, this may be an important factor in explaining the high CoT at slow speeds, since co-contraction of antagonistic muscles, with one muscle contracting
eccentrically, is considered to be mechanically and metabolically inefficient (Kuo, 2001, Ortega and Farley, 2005).

As discussed in Chapter 2.2.5, there are a number of energy saving mechanisms that reduce the metabolic demands of walking and running gaits, as well as a number of mechanical models that describe the mechanisms of walking and running in an attempt to explain their respective physiological demands. Walking has been described by the pendulum or rolling egg paradigm (Cavagna et al., 1963, Ralston and Lukin, 1969), while a bouncing ball or pogo-stick paradigm has been used to describe running (Cavagna et al., 1971, Cavagna et al., 1964). While there appears to be some crossover between these models, with contributions of elastic energy storage by the Achilles tendon and the foot in walking (Fukunaga et al., 2001, Ker et al., 1987), these models define the importance of the exchange of potential energy, used to displace the body centre of mass, and kinetic energy, which is due to the acceleration and deceleration of the body, in walking, while running is far more dependent on elastic energy storage and return of stored energy through tendon recoil. Lateral SS and XS gaits appear to be combining elements of the respective models (see Figure 7.11). The data indicates that the training limb accepts the weight of the body at initial contact from maximum horizontal velocity and vertical displacement of the hip. This is followed by the changeover between the trailing and leading limb, which accepts the weight of the body at the lowest vertical hip displacement and lowest horizontal hip velocity. This is followed by extension of the leading limb during stance, resulting in synchronous hip vertical displacement and horizontal hip velocity increase.
Figure 7.11 (a) Simplified model of sidestepping. The dashed curve represents the vertical displacement of the hip (rectangle). The solid line represents the training (T) and leading limb (L). The calf muscle and hip and knee muscles are indicated for the trailing and leading limb respectively to visualise their distinct contribution at the limb. Large and small Three phases can be identified: (1) impact of the trailing limb, followed by controlled impact absorption; (2) changeover between the trailing and leading limbs; followed by (3) extensor force production by the leading limb and vertical hip displacement. (b) Example data for hip vertical displacement and horizontal velocity during a SS and XS.

This mechanism in turn may help to explain the observable decrease in CoT with increasing speed of locomotion and the relatively low cost of transport at medium speeds. Griffin et al. (2003) argue that in walking the metabolic cost is largely due to the cost of generating muscular force during stance and in running the metabolic cost appears to be explained primarily by the cost of supporting the body weight and the time course of generating this force (Kram and Taylor, 1990) and it appears to be the generation of horizontal propulsive forces that contributes the majority of the total metabolic cost of normal running (Chang and Kram, 1999). If these processes can be assumed to have relevance also for lateral stepping, they offer a number of possible explanations for the relatively low CoT and metabolic power at medium speeds. The force data in Chapters 6 shows that compared to running, the vertical force at the leading and trailing limb, as well as the horizontal push-off force at the leading limb during lateral stepping was significantly lower than during running. Furthermore, the
findings for joint powers in Chapters 5 and 6 show the comparatively low mechanical power output at the leading limb as well as the importance of the hip in generating extensor moments. Therefore, the lateral movements appear to utilise smaller propulsive forces, which, according to the theories stated above may reduce the demands of the movement.

Furthermore, at medium speeds, the gait process appears to be optimised for generating vertical lift to allow for sufficient time for medial swing of the trailing limb. The trailing limb then acts in a shock absorbing and stabilising function rather than force generating function, while the leading limb abducts to contact the ground and generates the extensor forces required for vertical lift. The kinematic and kinetic data in the previous investigations have shown the importance of the forefoot strike pattern and the consequent utilisation of the Achilles tendon at the trailing limb. Elastic energy return from the tendon would therefore benefit the conservation of energy during the movement due to the enhanced efficiency through of elastic energy return as seen in running (Van Ingen Schenau et al., 1997). Furthermore, Lichtwark and Wilson (2006), Jacobs et al. (1993) and Fukunaga et al. (2001) suggest an isometric or quasi-isometric function of the calf muscles in human walking and running. This type of muscle activity is regarded as less metabolically demanding (Alexander, 2002) and may therefore contribute to reducing the metabolic cost of the lateral gait. This distribution of work, with a more or less passive trailing limb, utilising the Achilles tendon for energy return, and an active leading limb may therefore explain the relatively low CoT at preferred medium speeds. However, the data for stride length (Figure 7.8) also indicates the distinct disadvantage of the movement at fast speeds. Stride length is limited due to the limited range of motion at the hip in the mediolateral direction. Therefore, in order to generate larger stride length, an increase in the duration of the aerial phase is likely required. This in turn, would require larger extensor force at the leading limb. The consequence of this would be an increased physiological demand, which is supported by the large physiological requirements observed for the fast SS as well as the high perceived difficulty and injury risk ratings from the participant feedback (Fig. 7.9).

Skill appeared to have an opposite effect on the metabolic demands than was first anticipated. In contrast to the observations reported by Saziorski et al. (1987)
inexperienced participants displayed a lower net CoT and net P compared to experienced participants for the lateral SS and to a far lower degree for the XS (Figure 7.10). This is contrary to the observations for joint powers in Chapter 5, where no distinct differences in mechanical demands between skill levels were observed, but speed of locomotion appeared to be the main mediator of differences. The magnitude of the difference in metabolic parameters was however greatly reduced when discounting the influence of two participants with consistently low CoT values in all tested gaits. Nevertheless, the higher net CoT and net P values for experienced participants may be a reflection of an adaptation of movement technique as see in Chapter 5 where experienced participants performed the SS with smaller vertical hip displacement. Based on the findings by Ortega and Farley (2005), who observed an increased cost of transport for walking when using minimal vertical displacement of the centre of mass, it may be argued that a similar mechanism contributed to the higher metabolic cost in the group of experienced participants. However in the absence of kinematic data no proof for this idea can be provided.

7.5 Conclusion and Future Work

The findings reported above only partially support the hypothesis that sidestepping and crossover-stepping gaits result in significantly higher metabolic demands than typical gaits. At equivalent speeds, the lateral gaits are far more demanding than walking and displayed a higher net CoT than running, but did not display significantly larger net metabolic P compared to running, indicating a speed related effect on net CoT. The specific movement dynamics of the lateral gaits, as well as an effective utilisation of elastic energy appear to result in a metabolically relatively efficient form of transport. Higher levels of skill did also not lead to a reduction in metabolic demands. Indeed, a larger net CoT and net P were observed in experienced participants, which may be due to adaptations in movement technique.

High physiological responses are a recognised part of badminton (see Chapter 2.3.2) however their causes are unclear. While the findings of this investigation clearly show the differences in metabolic demands when adopting atypical movement strategies they do not give a clear indication of the relationship between the choice of movement
and the high heart rates observed in the game and in training. However, the application of SS and XS in the game differs from the continuous, steady state readings recorded in the lab. It therefore can be argued that it is the stop-start acceleration and braking during the short duration SS and XS in the game that may contribute to the high physiological demands of the sport.

Because of the apparent isolation of propulsive activity to a single limb at faster speeds and the high perceived difficulty and risk of injury ratings of lateral movement it would be recommended to include lateral stepping routines and specific strengthening exercises of the adductors, abductors of the thigh and plantarflexors of the ankle into the training routine to ensure sufficient strengthening of the specific muscle groups and the fine tuning of muscular coordination in the performance of these tasks. Future research should concentrate on re-examining the relationship of speed of locomotion on the metabolic demands of lateral movements. Use of a treadmill with a large running surface and high tensile strength may prove useful in specifying speed of locomotion and allow for a more precise estimate of the influence of skill or biomechanical variables between subjects. This information would provide very useful information for the examination of the metabolic differences between lateral movements and running at equal speeds. Furthermore, the previous investigations into the metabolic demands were heart rate dependent. Gas analysis represents a more reliable method of determining energy expenditure and should be performed to confirm the proposed demands of the sport.
Chapter 8

A biomechanical analysis of common lunge tasks in badminton

8.1 Introduction

The video footage of competitive badminton games in Chapter 4 showed the importance of atypical gaits and identified relatively high occurrences of lateral movements as well as stopping movements. Lunges were observed to be used as single movements as well as in combination with lateral gaits and can therefore be regarded as an integral part of the lateral movement, with in the context of badminton. Lunging forms an integral part of the movement repertoire of a competitive badminton player and has been identified as a critical aspect of sports such as squash, badminton and fencing (Cronin et al., 2003). Furthermore, the findings from the previous chapter indicate that the large physiological demands of badminton may be related to the starting and stopping of movements in game. As such lunging is a very interesting candidate to the demands of both starting and stopping, since the sport-specific lunge task consists of a weight acceptance (braking) and recovery (accelerating) phase. Considering the apparent importance of the movement it is surprising to find relatively little information available on this advanced movement skill in the literature. It is the purpose of this chapter to address this apparent lack of knowledge and identify the demands of lunge stops within the context of badminton-specific locomotion.

Movement in badminton is ultimately affected by the available space in which the game is played. The game is characterised by the repeated use of short bursts of activity during which the body is propelled toward the shuttlecock through the activity of the lower limbs. These short bursts of activity are the result of the player having to
react to the shot played by the opposition, playing their own shot and swiftly returning to the base (located toward the centre of the court) in order to avoid being outmanoeuvred by the opposition. This sport-specific movement requirement leads to the frequently coached start / stop / recover cycle (Badminton Association of England, 2005). This three part scheme essentially teaches the player to move toward the shuttle (start) by using for example the side- or crossover gait, reducing the momentum of the body (stop) and returning to the base position (recover) in preparation for the next shot. The lunge therefore forms a vital part of all movement and the SS and XS in particular. The high numbers of lunges recorded in the notational analysis of badminton games in Chapter 4 support this idea showing that the second phase, the stop, particularly when travelling toward the front or rear of the court (often using the SS or XS), commonly results in a lunge. It therefore is an important means of rapidly stopping the body from progressing in the direction of travel and furthermore acts a base for swiftly returning to the starting position. Interestingly, despite the apparent importance of the lunge task there is little information available about the kinematics and kinetics of the lunge, particularly when applied to badminton. Lees and Hurley (1994) recorded the forces during a typical badminton lunge for a number of experienced and inexperienced participants while a number of researchers have investigated the kinematics and kinetics of the fencing lunge (Adrian and Klinger, 1976, Gebhardt, 1981). Furthermore, Cronin et al. (2003) determined the strength qualities that act as important predictors to lunge performance and a number of authors investigated lunging as a rehabilitation exercise as it places less strain of the anterior cruciate ligament compared to open kinetic chain exercises such as quadriceps muscle isometric exercises (Stuart et al., 1996, and Heijne et al., 2004). However, to the best knowledge of the investigator there have been no studies investigating different lunge techniques in badminton.

The notational analysis performed in Chapter 4 suggests that there are two distinct lunging techniques that are used in the game (Figure 8.1). The first is a traditional lunge that will be referred to as the kick lunge. Here the weight bearing limb (the dominant limb) accepts the weight of the body during the weight acceptance phase of the lunge. The dominant limb bends to about 90 degrees at the knee with the upper body remaining upright and the non-weight bearing limb (the non-dominant limb) extended posteriorly. During the recovery phase of the lunge it is the dominant limb
that performs the majority of work required for recovery by extending the limb from the maximally flexed position while the non-dominant limb remains in a posterior orientation. The second method will be referred to as the step-in lunge. The weight acceptance phase in this lunge task is identical to the kick lunge. During recovery however, the non-weight bearing limb is pulled medially, toward the weight bearing limb to assist in the recovery from peak flexion. This method was regularly observed in deep lunges. It is thought that this method reduced the loading of the weight-bearing limb during recovery and therefore is advantageous for reducing fatigue. Furthermore, a third technique has been advocated recently, incorporating a hop between the weight acceptance and recovery phases of the lunge. To the best knowledge of the investigator there have been no studies investigating the effects of different lunge techniques on the biomechanics of the lower limbs.

Figure 8.1 Examples of a) the kick lunge and b) the step-in lunge. Three phases of the lunge are shown starting with initial contact, followed by weight acceptance and finally recovery.

The hypothesis of the current investigation is that the use of a step-in lunge technique reduces the loading of the dominant limb during lunge recovery compared to the kick lunge. A major goal of the study was to establish whether changes in movement technique would result in enhanced task performance. The aim of the investigation therefore was to measure lower limb kinematics and kinetics during three badminton-specific lunging tasks in order to identify differences in these parameters between the different lunge methods. It is thought that the use of the hop lunge technique may increase the ability of the dominant limb to generate force during the recovery phase.
by exploiting stretch activation (see Chapter 2.2.5.2). A further aim was to investigate the effect of these lunges on the duration of task completion and sample participant feedback on a selection of performance criteria. A quick and mechanically efficient method of recovery may benefit the athlete by increase the time available to react to the following shot by reducing the total time of the lunge task.
8.2 Methods

8.2.1 Participants

With approval by the Loughborough University ethical committee nine male 1st and 2nd team players from the Loughborough University Students badminton squad (age 20 ± 2.12 yrs; height 178.61 ± 6.4 cm; weight 70.58 ± 7.39 kg) were recruited and performed three standardised lunging tasks on a simulated badminton court in the laboratory. The participants were all actively participating in competitions with at least six years experience in the sport. The participants signed an informed consent form and were informed of the general requirements of the experiment. Upon signing the informed consent form the participants were asked to perform a general 5 minute warm-up to prepare the lower limbs for the lunging tasks. After this warm-up the participants were equipped with the wand system used for the capture of kinematic data (see Chapter 3.6 for a detailed description) and asked to perform a number of lunges of all three lunge methods, within a simulated badminton court, for familiarisation with the required tasks (see Figure 8.2).

Figure 8.2 Example to a left limb dominant participant during the loading phase of a kick lunge task during the warm-up period.
8.2.2 Experimental Design

The experiment was started once the participant completed the task familiarisation period and both the investigator and the participant were satisfied with the performance of the lunge task. The kinematics and GRF of the stance phase of the three lunge tasks (kick, hop and step-in) and the time required to complete the entire movement task, the stance-phase and the recovery phase were recorded for all participants. The kick lunge consisted of extension of the dominant limb (determined from racket arm preference) in front of the body followed by weight acceptance by the dominant limb with a posteriorly extended non-dominant limb (see Figure 8.3). The dominant limb acted as the main source of force generation during the extension phase to recover to the start. The step-in lunge consisted of the same dominant limb contact mechanism as the kick lunge however the non-dominant limb was pulled in toward the dominant limb during stance, resulting in a narrower stance than the kick lunge at recovery. Furthermore, after consultation with a registered badminton coach (Andrew Statham PhD, Badminton England, Coach Part 2 with 10 years coaching experience), a novel lunge method, the hop lunge, was included in the analysis. This task consisted of weight acceptance by the dominant limb with a posteriorly extended non-dominant limb. The dominant limb performed a hop after weight acceptance which was followed by a secondary ground contact phase and push-off from the forceplate to initiate recovery. Furthermore, the non-dominant limb was slightly adducted during the hop phase creating a wide crouched stance during the second hop phase. Confirmation of the validity of the tested lunge methods, their use in the game and their inclusion into badminton coaching was received from a number of experienced coaches (Mike Adams, Badminton England High Performance Coach and former National Coach of the Mauritius Badminton Team; Ray Learney, former National Junior Development Coach and Head Coach of the National Deaf Badminton Team; and Andrew Statham, see above).
All task repeats were performed on a simulated indoor badminton court in the lab to aid the participants understanding of the task and ensure successful completion of the lunge task in the lab situation (see Figure 8.4). The court tramlines were accurately measured and marked out using masking tape and a badminton net was placed at the front to the court to ensure accurate shuttle feeding and as a visual aid for the participant.
Figure 8.4 Layout of the simulated badminton court. The white lines (masking tape) represent the tramlines of the badminton court, with the two lines on the left defining the outside of the court and the single line to the right (with overlying black arrow) represents the central line of the court. The start position was 3.96m away from the net, along the central line of the court. A lunge start and finish line were marked out using duct tape (indicated by the red lines crossing the walkway). The start was adjusted for each individual and was set at 1.5 times leg length. The location of the force plate is marked out by the yellow rectangle.

All lunge tasks were initiated from a start position (about 4 meters away from the net) representing the middle of the back court (the base position) from which the lunge would typically be started. The position of the start was chosen in consultation with the registered badminton coach to represent the typical location of the base from which lunges are initiated on-court. Participants were instructed to stand at the start position facing the net with their shoulders in line with the net. From this position the participant was instructed to move along a raised walkway (start phase) and perform either a kick, hop or step-in lunge (stop phase) in response to a shuttle feed exercise (net feed), where the investigator manually fed the shuttle to the player by throwing it over the net with the shuttle landing in the front court area and the player returns the shuttle with the racket, and return to the start position (recovery phase). Shuttle feeding was practiced prior to the investigation under the observation of the badminton coach until both the coach and investigator were satisfied with the consistency and reliability of the feeding technique. This method was used for all three lunge tasks. A lunge distance of 1.5 times leg length (measured as the vertical distance between the anterior superior iliac spine and the ground) was chosen as the results of a pilot study, using two participants, and feedback from the coach which
suggested that this distance was equivalent to the largest distance from which the participant was able to successfully recover using a kick lunge which was thought to be the most demanding lunge task. Furthermore, it assured that the magnitude of the task was equivalent for all participants. Duct-tape was placed on the edges of the walkway to mark the lunge start and finish points as a guide for the participant during familiarisation, and the investigator to determine successful lunge task completion.

During the investigation the participants were instructed to perform the lunge task at their maximum effort while maintaining good lunging form to recreate a competitive environment. A successful task repeat consisted of correct foot placement at the lunge start line, contact of the dominant limb in the centre of the force-plate, contact with the shuttlecock and recovery to the starting position. In order to compensate for possible influence of fatigue the order of the lunge tasks were randomised for the participants. Due to a lack of a sufficiently large sample size of female participants at the time, only male participants with a long period of lunging experience were recruited to ensure good lunging form and sufficient muscular strength in order to assure the most consistent sample possible.

8.2.3 Data Collection

A Kistler (Type 9286A, Kistler, Switzerland) mobile multicomponent force plate, integrated into a raised, 1.2m wide wooden walkway, was used to record ground reaction forces (GRFs) at 200Hz. The cut-off magnitude of force was set to 2.5kN in the Z and 1.25kN in the XY direction. The XY force had to be increased from the normal high force setting (250N) used in the factory CODA setting to 1.25kN due to the excessive horizontal shear forces.

A two-camera, Cartesian Optoelectronic Dynamic Anthropometer (CODA, Charnwood Dynamics Ltd, UK) was used to record 3D bilateral kinematics for the lower limb using a specialised wand system supplied by Charnwood Dynamics Ltd. Please refer to Chapter 3.6 for detailed information on kinematic data capture. The 3D marker positions were low-pass filtered at 20 Hz. Limb segment orientations were
reconstructed using the supplied gait analysis software by Charnwood Dynamics and
kinematic and kinetic data was exported to Microsoft Excel for further data analysis.

Participant feedback was sampled for a number of self perceived performance criteria
(preference of lunge task, difficulty, level of fatigue and risk of injury of the lunge
tasks) using a five point scale for each element. Furthermore, written subject feedback
for each sampled performance criterion was collected to obtain subjective evidence
for the findings of the biomechanical investigation.

8.2.4 Data Analysis

Ground reaction force, kinematic and kinetic data were processed in CODAmotion
V6.68 (Charnwood Dynamics Ltd.). Stance phase was defined as the period of time
from contact with to lift-off from the forceplate by the dominant limb. Contact and
lift-off were taken to be equivalent to the vertical GRF impulse at a cut-off magnitude
of 15N (CODA factory setting).

The speed of approach and timing for the completion of the entire task, the stance-
phase and the recovery phase were determined from kinematic and force data. The
total duration of the tasks was defined as the time taken for the hip (defined by the
horizontal displacement of the marker located at the posterior superior iliac spine) to
travel from a pre-defined location (2.25m away from the far edge of the forceplate) to
the forceplate and return to this set location. Approach speed was measured as the
time taken for the hip to travel from the set location to the beginning of the GRF
(>15N) while stance duration was defined as the time from initial contact of the
dominant limb with the ground (>15N) to final push-off from the forceplate (<15N).
Furthermore, the recovery duration was defined as the time from final forceplate
contact (<15N) to arrival of the hip at the set location.

Ground reaction force was analysed for seven task repeats per participant while five
trials per participant were chosen for the analysis of the kinetic data. The smaller
number of tasks repeats for the analysis of kinetic data was due to technical issues
with the motion capture equipment at the time which resulted in a loss of marker data
in a number of participants. This issue did however not affect the GRF data. Because of the loss of marker data it was therefore decided to reduce the number of trials for all participants to ensure a sufficiently large and consistent sample size across participants.

Total positive and negative work at the joints of the lower limb during stance was calculated by trapezoidal integration of the respective power traces. By determining the area under the positive or negative phase of the power curve it was possible to assess the differences in mechanical work between the different lunge methods. Mean work values for the leading and trailing limb during SS and XS were determined for the group of male squad players for the purpose of statistical analysis.

8.2.5 Statistics

Statistical analysis of the influence of lunge method on the kinematic and kinetic parameters was performed using a repeated measures analysis of variance (ANOVA) in SPSS. Comparisons of kinematic or kinetic values for individual lunge methods were performed when appropriate to identify the contributors to a significant trend. Data analysis was performed for the mean peak values for GRF, joint moments and joint powers as well as mean total positive and negative work values. The average of these data points for each individual formed the basis of the statistical analysis where the level of significance was taken as $p<0.05$. 

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8.3 Results

8.3.1 Speed and Durations

There was no significant difference in approach speed between the different lunge methods. Speed was measured as the distance travelled by the hip marker (posterior superior iliac spine) from the 2.25m start point to initial contact with the force plate divided by the time taken to travel this distance (repeated measures ANOVA p= 0.266). The average speeds were 2.69 ± 0.29ms⁻¹ for the kick, 2.65 ± 0.32ms⁻¹ for the hop and 2.74 ± 0.35ms⁻¹ for the step-in lunge task. The total time taken to complete the lunge and recovery was slightly longer on average for the hop lunge, however, there was no statistically significant difference in total durations between the three lunge methods (p= 0.139). A significant effect of lunge method on stance phase was observed where the stance phase of the hop lunge was significantly longer that the kick or step-in lunges. No significant difference was detected for the kick and step-in lunges (p= 0.385). Furthermore, a significant effect was detected for the duration of the recovery phase with a significantly shorter time for recovery for the hop lunge compared to the step-in. No significant effects on recovery duration were detected for kick/hop (p= 0.122) or kick/step-in (p= 0.144) lunge methods.

Figure 8.5 Durations for the total duration, length of the stance phase and length of the recovery phase are presented. Durations are expressed in seconds. * indicates significant differences in the durations between the kick, hop or step in lunges.
8.3.2 Ground Reaction Force

Mean ground reaction force curves of the vertical, horizontal and transverse planes during kick, hop and step-in lunge methods for a representative participant are presented in Figure 8.6. The general shape of the horizontal and vertical curves is in accordance with the data by Lees and Hurley (1994). Following the classification used by Lees and Hurley five phases can be clearly identified in all lunge scenarios: Phase a, initial impact peak (heel strike transient); phase b, secondary impact peak (impact loading); phase c, amortization; phase d, weight acceptance (loading); and phase e, drive-off. Phase a, occurs due to contact of the heel with the ground. As the foot plantarflexes there is a force reduction ending in the secondary impact peak as the forefoot contact the ground. The knee continues to flex and the hip is lowered resulting in phase c. This phase ends at about the point where the knee is vertically aligned with the heel. Following this, the hip approaches its lowest point and the knee flexes further to about 90 degrees resulting in force peak d, the resultant force of weight acceptance. The forward momentum of the body is completely stopped and the direction of travel is reversed in the final phase of the ground contact phase (phase e).
Figure 8.6 Vertical (Fz), horizontal (Fx) and transverse (Fy) GRF in three badminton lunge tasks. Data represents the mean GRF for one participant performing kick, hop and step-in lunge methods. 5 force phases were identified (a,b,c,d,e). GRF is presented in Newtons (N). Error bars are used to visualise the magnitude of data variability between trials within the individual. The horizontal (Fx) and vertical (Fz) GRF curves display the most dominant force response while little force is produced in the transverse plane.

The peak values at phase a, b, c, d and e were generally in excess of the values quoted by Lees and Hurley (1994) for University level players (see Figure 8.7). Statistical analysis of the relationship of GRF phase and lunge method (findings summarised in Figure 8.7) was performed for horizontal and vertical force values. A significant effect of lunge method on vertical GRF was identified for the loading and drive-off phases. The results indicate that for both force phases the hop lunge resulted in significantly larger average vertical forces compared to the kick and step-in lunges.
No significant differences were observed for the kick and step-in lunges (p=0.38 for loading and p=0.41 for drive-off). The findings furthermore suggest a significant effect of lunge method on horizontal forces at drive-off. The hop lunge appeared to result in significantly larger forces than the kick and step-in lunges and significantly lower forces were produced by the step-in lunge compared to the kick lunge.

![Graphs showing vertical and horizontal ground reaction forces for kick, hop, and step-in methods, compared to literature values.](image)

Figure 8.7 Summary of the findings of the relationship of GRF and lunge method. indicates significant differences between lunge pairs. The findings from this investigation were generally larger than those quotes in the literature (Lees and Hurley, 1994).

8.3.3 Kinetics

The majority of the movement occurs in the sagittal plane and therefore the focus was placed on the flexor/extensor moments, powers and work done at the hip, knee and ankle joints.
8.3.3.1 Moments

The average peak hip, knee and ankle moments were analysed for the kick (MK), the first and second peak moment during the hop (MH1 and MH2 respectively) and the step-in lunge (MS) (see Figure 8.8). The results indicated no significant effect of lunge method on peak extensor moments at the hip (p=0.947). A significant effect was observed at the knee joint where the hop lunge caused a significantly larger extensor moment at MH1 compared to the kick or step-in lunges. No significant differences at the knee were observed for kick/hop(MH2) or kick/step-in (p=0.296 and p=0.501 respectively), step-in/hop(MH2) (p=0.476) or hop(MH1)/hop(MH2) (p=0.439). Furthermore, the influence of lunge method approached significant levels
at the ankle joint (p=0.065) where the ankle extensor moment at MH2 of the hop lunge was significantly larger than either the kick or step-in lunge ankle moments. No other significant relationships were observed at the ankle joint (Kick/hop(MH1) p=0.307; kick/step/in p=0.839; hop(MH1)/hop(MH2) p=0.149 and hop(MH1)/step-in p=0.353).

Figure 8.9 Summary of the findings for the peak hip, knee and ankle moments produced using the kick, hop and step-in lunges. † indicates significant differences between lunge pairs.

8.3.3.2 Joint Powers

The results indicate a significant effect of lunge method on peak joint positive power averages at the hip, knee and ankle joints. Peak hip power for both the step-in and PH1 of the hop lunge were significantly smaller than the kick lunge with significantly smaller values for PH1 than the step-in lunge. Furthermore, PH2 was found to be significantly larger than PH1 and no significant differences in PH2 or the kick lunge were observed (p=0.573). Peak knee power was largest at PH2 with significantly larger average peak power at PH2 than for the kick or step-in lunges. There was no significant difference in peak knee power for the kick and step-in lunges (p=0.253). At the ankle joint PH2 of the hop lunge was significantly larger than the peak power resulting from the kick or step-in lunges however there was no significant difference in peak ankle powers between the kick and step-in lunges (p=0.462) and significant difference in peak ankle power between hop(PH1)/step-in (p=0.469) or hop(PH1)/kick (p=0.085).
Figure 8.10 Summary of the findings for peak hip, knee and ankle power produced during kick, hop and step-in lunges. † denotes significant differences in peak power between lunge pairs.

8.3.4 Positive and Negative Work

The results from the statistical analysis (repeated measures ANOVA) are summarised in Figure 8.11. No significant trend for total negative work done at the hip (p = 0.317). Significant trends in total negative work were identified at the knee and ankle joints, where the knee and ankle performed significantly more negative work during the hop than the kick or step-in lunge. No significant differences were identified at the knee or ankle joint between the kick and step-in lunges (p = 0.918 at the knee and p = 0.286 at the ankle joint).

Significant trends for positive work done were identified at the hip, knee and ankle joints. At the hip joint the kick and hop lunges resulted in significantly more positive work than the step-in lung but no difference was detected for the kick and hop lunges (p = 0.083). At the knee joint the hop lunge resulted in significantly larger positive work than the kick or step-in lunges with larger, but not significant, differences in mean positive work values for the kick compared to the step-in lunge (p = 0.056). Similarly, the hop lunge resulted in significantly larger positive work during the hop lunge compared to the kick or step-in lunges but no significant difference was identified for the kick and step-in lunges (p = 0.449).
8.3.5 Participant Feedback

At this lunge magnitude participants generally appeared to prefer the kick lunge method over both the step-in and hop lunge with the step-in being the least preferred method. On average the kick lunge was perceived as the least difficult lunge method, however it did score highly in both the self perceived level of fatigue and injury risk criteria with the step-in lunge achieving the lowest score for fatigue and the hop lunge achieving the lowest score for injury risk.
Figure 8.12 Combined post exercise feedback averages on four selected performance criteria: preference of movement task, level of difficulty, level of fatigue and injury risk.
8.4 Discussion

The hypothesis of the current investigation was that the use of a step-in lunge technique reduces the loading of the dominant limb during lunge recovery. The aim was to establish whether changes in movement technique would result in enhanced task performance. A further aim was to investigate the effect of these tasks on subjective participant feedback for a selection of performance criteria. It was anticipated that the data from this investigation provides useful information on the effects of lunging within badminton, shed light on the physical demands of different lunge methods and in turn offer an opportunity to identify strategies to reduce muscular loading during this task.

Before discussing the results further it is important to discuss the limitations of the study. Lunging is an important skill of the game however lunge performance is affected by differences in technique and preference by the individual. Furthermore, parameters including body mass, leg length and flexibility have been demonstrated to affect lunge performance (Cronin et al., 2001, and 2003). Only skilled male badminton players with experience in performing the lunge tasks were selected as it was expected that these players would represent the most consistent sample possible for the assessment of lunge technique. Despite this an effect of individual variations in technique as well as the mentioned physical differences cannot be discounted as possible contributors to data variation between participants. Particularly the hop lunge represented a novel technique to most participants. Therefore the required adaptation to the technique was kept to a minimum, incorporating the hop phase into a basic lunging technique, which reduced the technical demands and was perceived as an easy adjustment by all participants.

The GRF values identified in this study were generally in excess of those quoted by Lees and Hurley (1994) for typical forces in a badminton lunge movement performed by experienced players. The cause for this discrepancy may lie in a difference in movement magnitude between the studies or an underestimation of peak force due to the lower sampling frequency employed in their study (100Hz). However no details of lunge distance or timings for lunge task completion, or any further information on
data recording or handling were available which makes cross-comparisons difficult. As expected there was no difference in the early impact forces (heel-strike transient, impact loading and amortization) between the tested lunge methods, since initial contact with the ground was made using the same heel strike method by all participants in all lunge scenarios (see Figure 8.7). The finding of a larger peak vertical force at loading (phase d) for the hop lunge can likely be explained by the involvement of the knee joint in immediately providing additional extensor force needed in order to lift the leg off the ground for the hop phase prior to the secondary ground contact. Furthermore, during the secondary contact phase the participant was able to apply larger vertical and horizontal forces to the ground at drive-off (phase e). These finding are in line with observations by Fukashiro et al. (1995) for hopping and goes some way to explain the faster recovery phase of the hop lunge described above, since the application of larger horizontal and vertical forces at the secondary ground contact of the hop lunge would allow the participant to generate more speed at drive-off and therefore recover more quickly. Despite the involvement of the non-dominant limb in the step-in lunge, aiding the dominant limb at drive-off, there was no significant reduction of vertical force at drive-off when compared to the kick lunge. However, a significant reduction in horizontal force was observed (see Figure 8.7). This indicates a shift in the action of the dominant limb, maintaining an important function in raising the body from the flexed position, but contributing to a lesser degree to horizontal push-off and the return to the base position.

In order to further assess the functional significance of the contribution of the joints of the dominant limb to the stance phase of the three lunge methods it is necessary to examine the moments, powers and work done at the joints. This will be done with reference to the kick lunge first, followed by a summary of the changes in these parameters due to the hop and step-in lunge recovery methods.

Figure 8.8, summarising the joint angles, moments and powers, visualises that after heel-strike the ankle joint plantarflexes as the forefoot (midfoot and toes) makes contact with the ground, with peak plantarflexion in early and late stance. After initial plantarflexion the ankle dorsiflexes as the upper body moves over the ankle joint resulting in peak dorsiflexion in mid-stance. The start of ankle dorsiflexion initiated the onset of the ankle plantarflexor moment, which lasted until late stance and had an
average peak value of $1.25 \pm 0.15\text{Nmkg}^{-1}$. The ankle joint furthermore generated positive power from mid-stance, with peak power $(1.89 \pm 0.67\text{Wkg}^{-1})$ occurring in late stance. The total negative work performed by the ankle until mid-stance was equal to $0.42 \pm 0.06\ \text{Jkg}^{-1}$, which was followed by positive work of $0.24 \pm 0.07\text{Jkg}^{-1}$ from mid-until late stance.

The knee flexes from heelstrike until maximum flexion of about $90^\circ$ in mid-stance and extends from mid-stance through to the end of stance phase. The knee displayed a strong extensor moment with an average peak of $2.92 \pm 0.30\text{Nmkg}^{-1}$. This moment started in very early stance as the knee extensors (quadriceps muscles) act eccentrically to decelerate the body until mid-stance, thereby preventing the knee from flexing beyond $90^\circ$. The knee extensors then act concentrically, extending the knee from mid-stance until the end of stance, creating a peak extensor power of $9.94 \pm 2.22\text{Wkg}^{-1}$ in late stance. The negative work done at the ankle joint in the first half of stance was equivalent to $2.15 \pm 0.31\text{Jkg}^{-1}$ followed by $1.42 \pm 0.32\text{Jkg}^{-1}$ of positive work in the second half. The knee therefore acts as the primary source of energy absorption and generation for the lunge. This supports the finding by Stuart et al. (1996) who identified a larger contribution of the quadriceps muscles to the performance of a lunge compared to other closed kinetic chain exercises, exercises where the foot remains in contact with the ground (e.g. squats or leg presses).

The hip mirrored the activity of the knee joint, flexing until mid-stance and producing a peak extensor moment of $2.49 \pm 0.6\text{Nmkg}^{-1}$. The peak positive power of $7.71 \pm 3.54\ \text{Wkg}^{-1}$ was smaller than the peak power at the knee, which is also true for the total negative ($1.57 \pm 0.55\text{Jkg}^{-1}$) and positive work done at the joint ($1.12 \pm 0.48\text{Jkg}^{-1}$). The hip therefore contributed to a slightly lesser extent to energy absorption during breaking and generation during recovery from the lunge.

Differences in the joint kinetics between the lunge tasks are observable at all joints (see Figure 8.9). At the ankle there was no significant difference between the kick and step-in lunges, however, the second contact phase of the hop lunge was significantly greater than during the other two methods. The cause of this increase in ankle moment is likely due to the use of the forefoot at second ground contact, which allowed for a larger contribution of the ankle joint through ankle plantarflexion. This idea is further
substantiated by the finding of significantly larger ankle positive power (Figure 8.10) and work (Figure 8.11) in the second contact phase of the hop lunge compared to the kick and step-in lunges. These findings are in line with the current view of the effect of an enhanced stretch-shortening cycle and the contribution of elastic energy on the increase of work performed by the musculature (summarised in Chapter 2.3.4.4). By using the hop lunge, the task is effectively transformed into a hop movement in the second part of the lunge. Fukashiro et al. (1995) show the superior contribution of the calf muscles and the Achilles tendon in hopping exercises, compared to other jumping movements, where the Achilles tendon is estimated to store 34% of elastic energy. The tendons, and the Achilles tendon in particular, are recognised for their important function in returning stored elastic energy (Biewener, 1997, Van Ingen Schenau et al., 1997, Zatsiorsky, 1997, and Josephson, 1999). Furthermore, Alexander (2003a) shows that not only the Achilles tendon but also the foot return energy by stretching and recoil (35% and 17% respectively during running). In addition the stretching of the muscle itself causes a potentiation of the contractile machinery, which increases with the speed of stretch, and therefore enhances the ability of the muscle to do work (Schenau et al., 1997). By changing the orientation of the foot during stance in the hop lunge, it is therefore possible to enhance force production by taking advantage of elastic energy stored in the foot and Achilles tendon as well as the increased ability of Gastrocnemius and Soleus to produce work.

At the knee the initial contact phase of the hop lunge resulted in a significantly larger joint moment than for the other tested lunge tasks (see Figure 8.9). As mentioned for the ground reaction forces it is thought that the increased knee moment is due to the generation of additional force at the knee to allow for the hop, which immediately follows weight acceptance. While the first contact phase of the hop lunge resulted in significantly smaller positive joint powers than any other positive power phase, the secondary contact phase of the hop lunge enabled the quadriceps muscles to generate significantly more concentric power than either the kick or step-in lunges (see Figure 8.10). This is further reflected in the larger positive work done at the knee joint during the hop lunge (see Figure 8.11). It is interesting to note that the positive work done at the knee was larger during the kick that the step-in lunge and the total negative work performed during the hop was significantly larger than during the step-in lunge. These findings support the hypothesis that the step-in lunge may indeed be beneficial for
reducing muscular fatigue of the quadriceps muscles by reducing concentric work. The hop lunge method on the other hand may increase muscular fatigue through increased eccentric and concentric activity. However, as described above, the increased power production may be related to an enhancement of the stretch-shortening mechanism and therefore indicate enhanced mechanical efficiency.

While lunge method did not significantly affect hip moments (Figure 8.9) it did affect joint powers (Figure 8.10) and work done at the hip (Figure 8.11). The peak hip extensor power at the kick and second contact phase of the hop lunge were significantly larger than that produced during the step-in. There was however no difference between the kick and hop lunge. Interestingly there was no significant difference in negative work performed at the hip between lunge methods. However the step-in lunge was involved in significantly less positive work at the hip than either the kick or step-in lunges. This finding is in line with the work done at the knee joint and substantiates a reduction in positive work done by using the step-in task. In relationship to running this data emphasises the great amount of negative work performed during the weight acceptance phase of lunging (Figure 8.11). In light of the relationship between eccentric contraction and injury (Lieber and Friden, 2002), the repeated use, as seen in Chapter 4, may therefore be regarded as an important contributor to overuse injury in the sport.

The participant feedback indicates that there is no straight forward relationship between preference for a lunge task and the perceived level of fatigue or injury risk (see Figure 8.12). While the kick lunge was identified as the preferred lunge method on average, with a low difficulty score, it scored highly in both the self perceived fatigue and injury risk scales. The sampled written feedback by the participants furthermore indicated that despite a general recognition of the relatively large demands of the technique on the musculature of the knee participants generally preferred the use of the kick lunge at this magnitude of lunging. This was thought to likely be a reflection of the effect of coaching and the regular use of the technique by most of the participants in this sample. Interestingly, despite a general consensus of a reduction of fatigue when using the step-in lunge, there was no substantial reduction in perceived injury risk. The step-in lunge in fact received the lowest preference score of all lunge tasks which, based on the participant feedback, appears to be due to the
narrow stance used for recovery. This created a feeling of the recovery phase being "too slow" and "not dynamic enough" prohibiting the player to react quickly to a shot played close to net. The hop lunge, despite being a more novel technique for a number of participants, indicated by the higher difficulty rating, received a favourable preference score. Furthermore, it received a low average fatigue score and the lowest score of all tasks for injury risk, supporting the idea of a beneficial result from the contribution of an enhanced stretch-shortening method and elastic energy contribution.

Speedy execution of the total lunge task (start/stop/recover) is furthermore important because it enables the player to play the required shot and return to the base position with sufficient time to respond to the next shot. The lack of a significant difference in the total duration of the three lunge tasks (Figure 8.5) is therefore important because it indicates that use of the different methods does not negatively influence the time required to return to the base. This appears to be true despite the significantly longer total stance duration of the hop lunge (including the first and second forceplate contact). The reason for this is due to the quicker recovery phase when using the hop lunge which reduced the overall time taken to return to the base. This finding furthermore suggests that the speed at recovery for the hop lunge is quicker, which would be advantageous for the player in a situation where carrying the momentum gained from recovery needs to be carried through beyond the base position in order to, for example, respond to a clear played to the back of the court. Using the hop lunge instead of the kick or step-in lunges, therefore, might well result in the player arriving at the shuttle faster in a situation where movement beyond the base position is necessary.

8.5 Conclusion and Future Directions

Based on the findings of this investigation the hypothesis that the use of a step-in lunge technique reduces the loading of the dominant limb during lunge recovery can be accepted. Furthermore, the participant feedback supports the view that this movement is likely beneficial for reducing fatigue but was not perceived as advantageous for preventing injury by this group of participants. A possible
explanation for this observation may be the lunge magnitude which may not have been large enough to represent a situation where the step-in would commonly performed in-game. In terms of overuse injury, the data present further support for the large eccentric demands on the musculature of the lower limb during all lunge tasks. However further adjustment of the weight acceptance phase of the lunge may present an opportunity for reducing eccentric muscular work.

The hop lunge increased the ability of the dominant to generate force by optimising muscular force generation. Furthermore, the orientation of the lower limbs at the beginning of the lunge return is of great importance and the narrow stance adopted at the step-in lunge was, in this situation, not seen as favourable. The hop lunge displayed some potentially very useful characteristics such as the ability of the ankle and hip joints to contribute to a larger extent to positive work and the adoption of a wider stance that enabled the participants to generate more speed and therefore reduce the time taken for recovery, which has important implications when moving beyond the base position.

Based on these observations, it may be recommended to adopt the novel hop-lunge recovery technique, since this method, despite recording larger joint moments and powers, appears to optimise the musculotendinous system, allowing for larger force development.
9.1 Introduction

In the course of this thesis a number of studies were performed to investigate the biomechanics of lateral movements as well as lunging. As with the majority of biomechanical studies these were performed in the laboratory in order to create an environment, in which it is possible to study a number of parameters reliably, control certain performance factors (e.g. speed of locomotion) and ensure maximum safety for the participant. The equipment used for gait analysis such as forceplates or kinematic recording equipment is often bulky and requires a dedicated workspace. Forceplates for example, need to be either integrated into a walkway or placed directly into the floor of the lab. The cameras used to record the movement have to be positioned accurately to ensure optimal visibility of limb excursions during the events of interest. Particularly the CODA scanner units used for the recording of kinematics in this thesis are very sensitive to infra-red light radiation, used for communication of the scanner and marker units, which effectively restricts their functionality to the lab environment with specialised lighting.

Health and safety of the participants is furthermore of primary importance when performing a study and any equipment that is attached to the participant in order to record the movement represents a possible risk factor. In this thesis, this included the use of wands for the attachment of markers for motion capture and electrodes for the recording of surface EMG. The use of this equipment therefore favours the
performance of the gait in the laboratory, where the risk factors of the external environment to the participant can be largely reduced. Here accurate measurements can be taken while focusing on minimising the risk posed by the use of equipment attached directly to the body or placed in close proximity to the participant.

However, the lab environment and the control of movement parameters, while helpful in simplifying data gathering and assuring participant safety, may themselves act as variables that affect the validity of the gathered data compared to the real-life application of the task of interest. An example of this problem is the treadmill which provides an easy solution for controlling the speed of locomotion of the participant, minimizing the required space to perform the exercise and allowing for easy recording of a number of parameters. However, a number of researchers have reported significant differences in kinematic parameters when walking or running on a treadmill. Alton et al. (1998), Strathy et al. (1983) and Murray et al. (1985) describe a higher cadence when walking on a treadmill. Alton et al. (1998) furthermore describes reduced stance time while Strathy et al. (1983), Murray et al. (1985) and Wall and Charteris (1981) describe changes in stride length due to floor or treadmill walking. Furthermore, the hip range of motion has been reported to be affected by treadmill compared to floor walking (Alton et al., 1998, Vogt et al., 2002) and running (Schache et al., 2001). Ground reaction forces during mid-stance appeared to be significantly affected by treadmill walking (White et al., 1998) while compression and tension strain rates of the tibia were reported to be 48-285% higher in overground compared to treadmill running (Milgrom et al., 2003). Nigg et al. (1995b) related the observed differences in running kinematics to a number of contributing factors including treadmill and running shoe used while Wank et al. (1998) state that participants favoured a running style that provided more stability when using the treadmill. In contrast to this Riley et al. (2007) describe only small differences that appeared to reduce after a familiarisation period, in line with reports by Matsas et al. (2000). Care must therefore be taken when interpreting the results from lab based investigations and comparison to the real-life application of the investigated task provides an important measure to validate the data.

This issue applies directly to the data presented in the previous chapters on the biomechanics of sidestepping, crossover stepping and lunging. The biomechanics of
these movements were investigated in a controlled environment in order to obtain reliable and repeatable, steady-state conditions for the comparison of groups of participants. However, this data does not reveal information on their use within the competitive match scenario which is an important aspect of not only badminton competition but also training.

The objective of the investigation is to test whether or not differences in ground contact or swing times of the leading and trailing limb during sidestepping, crossover-stepping and lunging tasks in the laboratory and competitive match setting exist. Such differences might influence the relationship between motion and the risk of injury. In order to address this objective, data on the timing of footfall events in competitive badminton singles games was gathered and compared with data gathered in the laboratory environment. This cross-validation is hoped to shed light on the application of SS, XS and lunge movements in competitive training games and assess the validity of the gathered data to the real-life application.
9.2 Methods

9.2.1 Experimental Design

Digital video footage of competitive male singles badminton games for 7 University student squad players was captured using a digital camcorder (Panasonic NV-DS27) at 25 frames per second (fps). The participants in this investigation were identical to those that participated in the investigation of SS and XS (Chapter 7) and lunging biomechanics (Chapter 8). Following data capture, the video footage was transferred to a Windows XP based PC as an interlaced digital video (DV) file. The first seven SS, XS and lunge movements of a single set of the game were identified and isolated by cutting them from the main video footage using the video editing software VirtualDub© (v1.6.16 by Avery Lee www.virtualdub.org). The selection criteria are summarised in the data analysis section below. In order to accurately identify footfall events, the video footage was further processed using VirtualDub and AVIsynth© (v2.5 by Ben Rudiak-Gould et al. http://avisynth.org/ mediawiki/Main_Page), following instructions by Donald Graft (http://neuron2.net/bob.html) to remove video interlacing and double the temporal resolution of the video footage from 25 to 50 fps.

Interlacing of video footage is a process performed in order to improve the picture quality of the footage without using additional bandwidth. Each frame of the interlaced footage consists of an odd and even field, effectively containing half the information of the complete image each, which are displayed in short succession in order to create the illusion of one fluid image. De-interlacing of the interlaced DV footage consists of splitting the two fields that comprise the individual frames and increasing the size of each image through interpolation (see Figure 9.1 for an example of interlaced and de-interlaced video footage).

Field splitting was performed using AVIsynth as a preprocessor and involved creating a file called "clip.avs" containing the following command lines:

\begin{verbatim}
clip=AVISource("f:\capture.avi")
\end{verbatim}
clip.SeparateFields

where AVISource() contains the path to the input video clip.

In case of reversed field dominance (indicated by skipping of the footage instead of smooth movement at playback), the command clip.SeparateFields was replaced by:

clip.ComplementParity.SeparateFields

Figure 9.1 Example of before and after results of the application of the SmartBob filter showing the original interlaced frame of the 25fps and the resultant de-interlaced frame at 50fps.

The resultant “clip.avs” file was opened in VirtualDub resulting in the display of the input clip with its fields changed into half-height frames (see Figure 9.2). Finally, the SmartBob (v1.1 beta 2) filter was loaded in virtual dub which produces a doubled frame rate output file where each output frame is created from a single input field by interpolating data using bicubic interpolation.
9.2.2 Data Analysis

The timings of the stance and swing phase of the leading and trailing limb during the SS and XS were recorded (see Figure 9.3 and Figure 9.4) and entered in the spreadsheet application Excel© for further processing. The selection criteria for both the SS and XS were that they consisted of a complete gait cycle and consequently a distinct swing and stance phase of the leading and trailing limb. There are a variety of different approaches in the literature for the identification of swing and stance phases. Please refer to Chapter 3.10 of this thesis for a detailed review. The method adopted here was based on the video data at 50Hz. The identification of stance and swing phase using video data has been shown to be a reliable method (Wall and Crosbie, 1996, and Ghoussayni et al., 2004). Toe-off was taken as the first point in time where the foot of the lower limb under investigation can be seen to be completely separated from the ground. Impact was taken as the first instance of clear contact of the foot with the ground. The first 7 SS and XS movements of one set for the 7 participants under investigation were analysed by recording the timing of foot impact and toe-off events. Furthermore, the first 7 repeats of the movement in the laboratory setting (Chapter 6) were selected for each participant to allow for comparison of the in-game and laboratory setting. Here stance phase was defined as the time between the onset and offset of vertical ground reaction force (cut-off magnitude of force = 15N) and
swing phase was defined as the time between toe-off and force onset where toe-off was identified from the graphical representation of the participant in CODAmotion and the positional data of the markers of the foot.

Figure 9.3 Example of SS used in-game consisting of 1) leading limb impact, 2) trailing limb toe-off, 3) leading limb toe-off, 4) trailing limb impact, 5) leading limb impact and 6) trailing limb toe-off.

Figure 9.4 Example of XS used in-game consisting of 1) leading limb impact, 2) trailing limb toe-off, 3) leading limb toe-off, 4) trailing limb impact, 5) leading limb impact and 6) trailing limb toe-off.

The duration of the swing and stance of the dominant limb during in-game lunge was analysed by recording the timing of swing and stance onset and offset using the methodology described above. Only the dominant limb, determined by which arm is used to hold the racket, was investigated since it is the dominant limb that players are taught to use for weight acceptance in the lunge to increases the reach of the racket (Badminton Association of England, 2005). A total of 7 lunges per participant were
analysed for the group of student badminton players (see Figure 9.5). Furthermore, 7 repeats of the lunge movement were analysed for each participant from the data gathered in the lab following the swing and stance phase definitions described for the SS above.

![Figure 9.5 Example of in-game use of the lunge consisting of 1) dominant limb toe-off, 2) impact and 3) secondary toe-off.](image)

9.2.3 Statistics

Durations of leading and trailing limb swing and stance phases during the SS and XS and dominant limb during lunge swing and stance phases for the laboratory and in-game settings were analysed using repeated measures analysis of variance (ANOVA) in the statistical analysis tool SPSS. Statistical significance was taken to be equivalent to $p<0.05$. 
9.3 Results

9.3.1 SS and XS

The results indicated a significantly longer mean stance duration for SSL in-game than in the lab (0.26 ± 0.02sec and 0.22 ± 0.03sec respectively). Furthermore, mean swing phase durations of SST were significantly longer in the lab than in-game (swing= 0.3 ± 0.03sec and 0.25 ± 0.04sec respectively). In the XS both the swing and stance phase appeared to be significantly longer in-game than in the laboratory (swing= 0.42 ± 0.03sec (In-Game) and 0.31 ± 0.06sec (Lab); stance= 0.3 ± 0.02sec (In-Game) and 0.24 ± 0.02sec (Lab)). No significant effects on durations were observed for SSL during swing (p= 0.662), SST during stance (p=0.892), XST during swing (p= 0.287) or XST during stance (p= 0.291).

Visualisation of the data distributions for the respective stance and swing phases (see Figure 9.6 and Figure 9.10 respectively) indicates that stance durations were very similar between all settings and groups of participants. In the case of the stance phase of SSL it was determined that ~67% of all durations in the game setting fell within the time range observed for the same participants in the laboratory. For the swing phase of SST ~63% fell within the laboratory time range. In the case of the swing and stance phase of XSL about 50% of the in-game duration values fell within the range observed in the laboratory. The data show that the lab experiments appear to be generally in line with the temporal dynamics of in-game use. However, the trend toward faster stance and swing durations in XSL suggest that the values obtained in the laboratory may be toward the higher end of the scale compared to the application in the game.
Figure 9.6 Breakdown of stance phase data distributions for the leading and trailing limb during the SS and XS. Raw duration data is presented for the 7 task repeats sampled for the group of 7 participants in the game (In-Game) and laboratory (Lab Ch. 7) settings. Furthermore, mean durations for the participants involved in the study in chapter 6 are presented for comparison.

Figure 9.7 Breakdown of the raw swing duration data for the groups of 7 participants. For the purpose of comparison, 7 sampled repeats per participant are presented for the game (In-Game) and laboratory (Lab Ch. 7) setting.

9.3.2 Lunge

The results indicated that the mean duration of swing of the dominant limb for the group of participants was significantly higher in the lab (0.486 ± 0.039sec) than in-game (0.384 ± 0.032sec), however, no significant difference in mean stance duration between the two settings (0.609 ± 0.047sec (lab) and 0.661 ± 0.053sec (in-game)) were detected (p=0.199).

The application of the lunge can be argued to be dependent on both player preference and the ability of the opposing player. “Good” opposing players are able to force the athlete to move by successfully placing the shuttle in tactically favourable areas of the court, while “weak” players are not able to exert the same degree of control resulting in less movement required by the athlete during a game. Because of the interaction of personal preference and the game-specific, opposition-dependent, requirements it was decided to furthermore investigate the effect of the changing from the lab
environment to the in-game setting within individual players. The results (summarised in Figure 9.8) support the overall findings summarised above where 6 of the 7 participants displayed significantly shorter swing duration in-game compared to the lab and only 3 of the 7 participants displayed significantly longer stance durations in game.

![Swing Duration Dominant Limb](image)

![Stance Duration Dominant Limb](image)

Figure 9.8 Summary of the mean swing and stance durations of the dominant limb during lunges in the lab and in-game. 6 of the 7 participants displayed significantly shorter swing durations in-game with participant 7 displaying no significant difference between the two settings (p=0.213). Participants 2, 4, and 6 displayed significantly longer stance durations in-game with no significant effect in participants 1, 3, 5 and 7 (p=0.366, 0.217, 0.809 and 0.111 respectively).

9.3.3 SS/Lunge and XS/Lunge Combinations

A further application of the SS and XS was in combination with a lunge (see Figure 9.9). The use of the SS/Lunge or XS/Lunge was seen to be very much dependent on the individual participant and not all players used equal numbers of these movements. Within the one-set timeframe 4 participants used at least 5 repeats of the SS/Lunge movement and 2 participants used at least 5 repeats of the XS/Lunge movement. These were used to assess the impact of the combination of these movements on both the trailing limb during the approach phase of the lunge (swing and stance duration of
the trailing limb) and the leading limb during the lunge phase (swing and stance duration of the leading limb).

![Figure 9.9 Example of a) SS/Lunge combination; b) XS/Lunge combination.](image)

The period of time the trailing limb spent swinging appeared to be lower in the SS and XS compared to the lab and ‘normal’ in-game use (Figure 9.10). This may indicate a reduced stride length when applying the SS and XS in combination with a lunge compared to using the movement on its own.

![Figure 9.10 Data is presented for trailing limb swing and stance phase durations for 4 participants performing 5 repeats of the SS/Lunge movement compared to the Lab and In-Game setting. Durations are expressed in seconds.](image)
Figure 9.11 Data is presented for the swing and stance duration of the leading limb for 4 participants performing 5 repeats of the SS/Lunge movement (a-d) and 2 participants performing the XS/Lunge (a & b) compared to the Lab and In-Game setting. Durations are expressed in seconds.
9.4 Discussion

The aim of this investigation was to compare the timings of a number of footfall events of sidestepping, crossover stepping and lunging in competitive badminton scenarios with the data recorded in the laboratory environment. These data provide essential information on the application of these movements in-game and will be used to validate the observations made in the previous studies relating to the real-life setting. Differences between the settings may indicate a change in the relationship between motion and force observed in the previous investigation.

The results of this investigation indicate a number of differences between the in-game and lab settings. For the SS and XS, there was good agreement in the durations of the swing phase for SSL and XST, as well as for the stance phases of SST and XST. The stance phases of SSL and XSL however, appeared to be longer in-game, while the swing phase of SST was longer in the lab setting and XSL-swing was longer in-game. Despite the significance of these findings however, data ranges of the in-game and lab settings compared favourably (see Figures 9.6 and 9.7 for stance and swing times respectively). In the case of SSL-stance and SST-swing ~67% and ~63% of in-game durations, respectively, fell within the range observed in the lab. The results for XSL-swing and stance on the other hand were larger, with ~50% agreement between the in-game and laboratory settings. Therefore, it appears that the data range seen in the lab-setting somewhat overestimates the support phase of the leading limb, indicating that leading limb forces in the lab may be toward the higher end of the scale seen in-game. Furthermore, it appears that the swing phase of SS tends to be slightly shorter in-game, while the lab data tends to reflect the faster range of swing durations for XSL.

These findings point toward a number of contributing factors that can affect the duration of the swing and stance phase. It is recognised that, in walking and running, stance time decreases with increasing speed of locomotion, while the time for swinging the limb remains relatively constant (Enoka, 2001, Shumway-Cook and Woollacott, 2006). Therefore, the longer leading limb stance durations may be caused by the use of a slower speed in-game compared to the lab. This is supported by the finding of larger differences in the XS than the SS between the in-game and lab.
setting. The XS appears to be regularly used for movement to the back of the court, in response to a high, clearing shot. Therefore, the player has more time to respond to the shot and uses long slow strides to move to the back of the court. This, in turn, suggests a lower speed of locomotion which may contribute to the longer stance duration of the leading limb. The SS on the other hand is used for movement in all directions and may be performed at generally faster speeds than the XS.

Stance phase times of the trailing limb, however, were equivalent between the lab and in-game settings. This points toward further contributing factors, other than speed alone. As the player constantly changes direction in the game, he/she is repeatedly accelerating and stopping. Since lateral movements are performed in the direction of the leading limb it can be argued that in-game it is often involved in the acceleration phase of the movement, rather than a speed maintenance function. The leading limb would therefore be in contact with the ground for longer in-game than the lab, as force is applied to the ground to accelerate the body, in-line with the observations of this investigation. The trailing limb on the other hand, always performs a speed maintenance function and both swing and stance durations in the SS and XS therefore do not differ vastly between settings. Furthermore, game-specific factors may influence the recorded timings. Reaching for a shot may increase stance duration as the focus switches from speed to balancing the upper body and guiding the racket toward the shuttle to allow for maximum shot accuracy. Stride length may also be different between the settings. Shorter steps in the SS may contribute to the trend toward shorter SS\textsubscript{T} swing phase, while longer strides in the XS may contribute to the finding of longer swing durations for the leading limb in-game.

The findings furthermore contribute to the data gathered in Chapter 4, adding further prove that lunges are not only performed using a step and lunge approach but also the combination of a SS or XS with a lunge finish. The SS and XS are therefore performed as contained movements, as well as in combination with the lunge (see Figure 9.9). These combination movements were used by all participants. However, not all participants used these movements equally in the game, meaning that of the seven participants investigated in this study, only four participants used at least five SS/lunge combinations and only three participants used at least five XS/lunge combinations during the sample period. The data for the trailing limb swing and
stance durations in the SS/lunge and XS/lunge combinations, compared to the lab and in-game timings of the SS and XS (see Figure 9.10), indicate that the length of the trailing limb swing phase was generally shorter when using the SS and XS/lunge combinations. This is thought to indicate that the speed of the trailing limb was faster in this application of the trailing limb and consequently the data recorded in the lab may underestimate the resultant forces at trailing limb impact. This would support the rationale behind using the lunge, which amongst other aims, is to arrive at the shuttle quickly, to intercept it early in flight and therefore increase the shot-making options. There appeared to be no noteworthy change in the duration of the stance phase between any of the lunge scenarios, with particularly the XS stance time shifted toward the lower end of the scale observed in the lab. An alternative cause for the low swing duration of the trailing limb may be that step length was smaller compared to the SS and XS in-game and the lab. This could be due to a more stationary start in the SS/XS + lunge use, compared to the lab environment where the movement was performed at speed, and a fairly narrow stance before the start of swing.

The findings for weight bearing limb during the weight bearing phase of the lunge (the dominant limb and leading limb for the step-lunge and SS/XS + lunge tasks respectively) indicate that the swing phase was shorter in-game than in the lab (see Figure 9.8 and Figure 9.11 for step-lunge and SS/XS + lunge respectively). Furthermore, despite some variation within individual players, the stance phase of the lunge was not significantly affected by the lab or in-game setting and there did not appear to be marked differences in stance duration within players using the SS/lunge or XS/lunge approach. The shorter swing phase of the SS and XS lunge combinations was expected, since the range of motion of the limb is restricted, compared to the range of motion of the dominant limb during the step-lunge task. In this context, the shorter swing times of the dominant limb, during the step-lunge in-game (Figure 9.8) was somewhat surprising. However, it may be an indication that despite the effort taken to recreate a competitive environment in the lab, the match scenario favoured a faster approach phase for the lunge. It may however be a reflection of the fact that the initiation of the lunge from a number of different locations on the court and from a standing or moving start, means that the total lunge distance is never the same. Therefore, stride length is likely to be very variable in-game, with large strides for long lunges and shorter strides for short lunges. While there is a difference in swing
time of the dominant limb, the data is in line with the intended set-up of the lunge in the lab, representing a controlled and relatively large lunge movement.

Finally, an interesting observation of this investigation relates to the mechanisms of injury in badminton. The review of sports injuries in Chapter 2.6.2 showed that ankle sprain injuries are common in badminton and tennis (Table 2.6). However, in light of the protective mechanisms employed in lunging, landing with the heel first, followed by the rest of the foot and alignment of the knee, in line with the plantarflexor movement of the ankle, the relationship between lunging and sprain injury was unclear. Findings for a number of participants, with example data presented in Figure 9.12, showed that when using the lunge in the back of the court appeared to pose a particular risk for ankle sprain injury. The rear court lunge requires the participant to move back into the mid- and forecourt quickly. The example data shows that in the attempt the move back into the court, the upper body is rotated toward the trailing limb, in an effort to face the opposite court. This rotation is translated into medial rotation of the ankle, which ultimately results in excessive ankle supination, as the participant applies extensor force at the knee during the lunge recovery phase.

![Figure 9.12 Sequence of events involved in a near-injury situation.](image)

9.5 Conclusion

In view of the findings of this investigation it is apparent that differences in ground contact and swing times of the leading and trailing limb during sidestepping, crossover-stepping and lunging tasks in the laboratory and competitive match setting exist. The presented data show the variability of gait use in badminton. There are a
number of factors that influence the application and magnitude of movement in-game. These include the relative position of the player and the shuttlecock, as well as the speed and direction of travel of the shuttle. These and other factors discussed above, influence the speed of movement as well as the chosen stride length.

Overall however, the data from the lab-based experiments on lateral stepping and lunging are in-line with the observations in-game. There appears to be a general trend toward faster speeds for the lateral tasks in the lab, as well as a trend toward a larger lunge magnitude. This however, is in line with the intention of the experimental set-up, where a relatively large lunge was tested to allow for a lunge magnitude that would allow the participant to select either one of the three lunge movements.

It can therefore be concluded that the data obtained in the laboratory setting reflects a valid representation of the demands in-game and the conclusions drawn on the risk of injury associated with atypical movement are validated. Judging by the variability of movement in the match scenario, future research should focus on the assessment of a selection of specific movement tasks. These include the contribution of the trailing limb during the stance phase of the SS and XS lunge, as well as the forces in the leading limb, during the breaking phase of a lateral movement.
Chapter 10

General Overview and Conclusions

10.1 Introduction

The underlying objective of the research conducted in this thesis was to investigate the biomechanical and physiological demands of atypical gaits used in badminton. The gathered biomechanical data is essential in advancing our knowledge of the cause and effect relationship of motion and force exposure in badminton and may hold essential information related to the risk of injury in the sport. This chapter summarises the conclusions of the experimental research. The findings of each investigation (Chapter 4-9) are summarised individually and in the sequence used in the main body of this thesis, along with the related aims and conclusions drawn from the research. Therefore, the conclusions of the notational analysis (Chapter 4) will be presented first, followed by the biomechanical investigations into SS and XS (Chapter 5). Chapter 6 supports the findings, presented in Chapter 5 and expands on the contribution of the musculature, while Chapter 7 investigates the metabolic demands of the movements. The biomechanics of lunging are discussed in Chapter 8, while Chapter 9 validates the findings in Chapter 5, 6 and 8. This chapter will end with suggestions for future research.

10.2 Summary of Conclusions

10.2.1 Video-based classification, quantification and comparative analysis of gait usage in badminton (Chapter 4)

The findings of the notational analysis revealed a very varied use of movement in both national and international badminton matches including sidestepping (SS) and
crossover-stepping (XS), as well as lunging, hopping and jumping. This demonstrates the importance of atypical movement to the competitive match scenario. Differences in the use of movement between male and female participants and the duration characteristics of rally and rest periods were furthermore observed. These included shorter rally durations and a larger number of jumps in the male game as well as longer rally durations and a larger number of sidesteps (SS) within the female game. Skill level was reflected in the increased rally and rest durations in international players as well as higher number of lunges. Differences in the playing style between genders, with more aggressive tactics being adopted by the male players, and larger technical and tactical ability by the international player may explain the higher rate of injury of the male player.

While this study allowed for the quantification of the movements used within the game it also highlighted the difficulty in attempting such a quantification. Badminton is characterised by the short duration and variability of movement and as such movements can be used in combination in order to achieve the sport-specific goal. This complicates the temporal description of the movement and may lead to a simplified view of movement use. Furthermore, this study does not contain information on the relationship between movement and outcome of the rally which may be of specific interest for the purpose of coaching badminton at the national and international level. However, in view of the aim of this study in relation to the overall objective of the thesis it provides original and very detailed data on the use of movement in-game.

10.2.2 A biomechanical analysis of the stance phase of a standardized sport-specific sidestep and crossover step task (Chapter 5)

The joint moments and powers of sidestepping and crossover stepping in the laboratory setting were within the range expected for typical gaits and differences in kinetic data due to skill or gender were few and generally low in magnitude. The data indicated an asymmetric contribution of the leading and trailing limb to the lateral gaits. This verified the view that the limbs perform distinct roles in the gait cycle. Therefore, instead of the right and left limb essentially performing the same role, as in
walking and running, the trailing and leading limb in lateral movements perform a shock absorption and vertical force generation function respectively. Together with an observed shift in joint moments toward the more proximal hip joint, this led to the conclusion that the lateral gaits can essentially be described as lateral jumping with two consecutive phases of controlled falling.

10.2.3 Kinematic, kinetic and EMG analysis of selected lower limb muscles during lateral sidestepping and crossover stepping tasks

(Chapter 6)

The data reiterated the relative importance of the hip in generating extensor moments at the leading limb, as well as the reduced capacity of the muscles of the leading limb to generate muscular power. The XS appears recommendable due to the increased stride length and 'steady hip' motion. However, it is a more committed movement than the SS and in a setting when quick adjustments are necessary use of the XS may be detrimental to the task. The hip adductors and abductors perform a unique function in the locomotor process, compared to walking or running, which may increase the risk of injury of this muscle group. However, without epidemiological data this suspicion cannot be confirmed.

A limitation of laboratory based experimentation of the kinematics, kinetics and muscular activations during gait is the need to create a controlled environment in which the movement is performed. While all possible attempts were made to recreate a natural movement task, the performance of the movement may differ from that in-game. Furthermore, the measurement equipment used is subject to discussion as kinematics, kinetic and EMG measurements in a dynamic situation are affected by a number of variables. Marker movements, motion artefacts, signal interference and crosstalk are a number of factors that were carefully considered and efforts were made to reduce the potential affect of these factors on the collected data. Alternatively, modelling techniques could have been employed to assess the biomechanics of lateral gaits. However, these techniques are still novel at this point and commonly require kinematic and/or EMG data in order to predict joint loading or muscular activations.
using inverse dynamics. Despite the limiting factors it is a strength of this investigation that kinematic, kinetic and EMG data was collected simultaneously during these highly dynamic movements since only this combination of information allows for a true understanding of musculoskeletal coordination of the movement process. Furthermore, in time, the data from these studies may well be used as input for such modelling techniques and allow for greater insight into the demands of these movements on the joints of the lower limb, as well as in sport-specific settings without the constraints of the lab-space.

10.2.4 A comparative analysis of the metabolic requirements of atypical gaits used in badminton (Chapter 7)

A marked difference in the cost of transport was observed at slow speeds, while the medium speed resulted in metabolic power values which were equivalent for SS and XS and running in both the shape of the increase in power, with no significant differences at similar speeds. Therefore metabolic power for SS and XS was slightly, but not significantly, larger than for running. The data therefore showed that deviation from the preferred medium speed of locomotion resulted in either large cost of transport, at slow speeds, or excessive physiological demands when travelling at fast speeds.

As for the previous experiments the use of a laboratory setting, as well as the recording equipment itself should be considered in reviewing this experiment. While the accuracy of the gas analyser was validated the equipment required the continuous performance of the movement in order to quantify the metabolic demand. This use of the movement is very different from that in-game and therefore presents a limitation of the study. However, in order to avoid this issue, alternative assessment strategies, such as computational approaches to model and predict the energetics of motion would be necessary. These modelling approaches contain a number of uncertainties on the methods of evaluating muscular activity and the related energy demands in dynamic gait. Furthermore, since these alternative techniques were not available at the time the methodology used in this study was deemed to be the best compromise in assessing the metabolic demands of atypical gaits.
10.2.5 A biomechanical analysis of common lunge tasks in badminton

(Chapter 8)

The data showed that the step-in lunge resulted in lower positive power at the hip which indicates a reduction in the demands of recovery. However the participant feedback gives strong indication that this type of recovery would be used by this group of players at the defined lunge distance. Furthermore, within the context of the use of the movement the findings showed that the hop-lunge resulted in the highest moments and reaction forces indicating an enhancement in muscular contribution in this task. Within the context of injury and injury prevention, the low scores for perceived risk of injury and comparatively low fatigue score can be viewed as positive indicators of a beneficial effect of changing the technique for lunge recovery. With a view to enhancing the performance of the athlete within the game this data indicates that adaptation of a modified lunging technique may indeed be recommended for.

Assessing the performance of a specific skill is subject to performance variables that differ between participants. In this study it was attempted to avoid some of these variables by selecting experienced athletes who are accustomed to the performance of the required tasks. This means that the resulting data is reflective of experienced participants only and the observations may not be applicable to other groups of players. However, in assessing the effect of different techniques of the same task this limitation of the scope of the study was deemed necessary in order to focus on the effect within this representative group. As mentioned above, an alternative would be the use of modelling techniques for the assessment of the loads of different techniques. Such approaches have been performed in the past (Lees and Hurley, 1994) and additional studies using computational methodologies would certainly be beneficial in order to gain further and more detailed information on the biomechanical demands.
10.2.6 Comparative analysis of sidestepping, crossover stepping and lunging in badminton – In-game compared to the laboratory

(Chapter 9)

Differences in the recorded temporal parameters for SS, XS and lunging in the laboratory and in-game were detected. However, it was concluded that the data obtained in the laboratory based experiments presents a valid reflection of the use of movement in-game and therefore the conclusion drawn on injury risk factors within the game are validated.

This study helped in visualising the complexity of movement use in badminton. The highly dynamic nature of motion in badminton means that studying specific, isolated movements in the laboratory somewhat oversimplifies the assessment. As mentioned above, this is a necessary requirement of people based studies as the recording equipment does not readily allow for direct in-game assessment. While the data recorded was deemed to be a valid reflection of the demands of the sport, it is hoped that the limitation of biomechanical assessment will be reduced in the future. Direct assessment in-game would allow for larger scope in evaluating the demands of a range of movements and particularly extreme conditions of motion.

10.3 Summary

In summary, the research reported in this thesis has shown that atypical movements make a significant contribution to the movement repertoire of the competitive badminton player and can be regarded as an integral part of the game of the male and female, as well as the national and international player. However there does not seem to be a clear relationship between the use of lateral sidestepping and crossover-stepping gaits and the high physiological demands of the sport and the relatively high levels of lower limb injuries sustained. Despite the indications that lateral movements utilise mechanically less efficient gait processes they were a surprisingly efficient form of locomotion at medium preferred speed. However, greater metabolic costs for a given distance greatly increased at slow speeds and caused excessive physiological
demands at faster speeds. The asymmetric contribution of the leading and trailing limb and the use of proximal muscle groups by the leading limb led to the conclusion that lateral stepping can essentially be described as lateral jumping with two phases of controlled falling, which is supported by the combined electromyographic, kinematic and kinetic data. The trailing limb therefore performs a stabilising function while the leading limb generates vertical lift. Interestingly, experienced badminton players showed an adaptation of the movement kinematics using smaller vertical displacement of the hip during sidestepping and crossover-stepping and adopted a heelstrike for the leading limb which may be beneficial for preventing excessive ankle pronation or supination thereby creating higher joint stability. Furthermore, this asymmetric use may contribute to imbalances of the hip adductors/abductors and hip extensors. This may potentially alter the normal direction in which a tendon exerts force and thereby expose the athletes to an increased likelihood of overuse injury (Hess et al., 1989, Witvrouw et al., 2000, Kannus, 1992). However, other factors such as previous injury, which appears to have a lasting effect on hamstring and quadriceps muscle moment ratios (Dauty et al., 2003), or adaptive hypertrophy, such as those reported by Mikkelsen (1979) may be the underlying cause for muscular imbalances. The influence of the asymmetric contribution of the limbs in lateral gaits on the development of muscular imbalances therefore remains speculative. In light of these findings, both movements may be recommended for use in the game. From a practical point of view, the sidestep is the less complex movement and was generally regarded as the easier movement to perform. Furthermore, execution of a sidestep takes less time than the XS which is a more committed movement. Use of the sidestep may therefore be advantageous when changes in direction are required, while the greater stride length of the crossover allows the player to move further, with a steady upper body.

Lunging regularly forms an important role in combination with a lateral step by stopping the forward progression of the body and initiating the return movement. The movement displayed large eccentric muscle demands during the weight acceptance phase which are related to overuse injury of the lower limb. This seems particularly important in relation to the frequent use of the lunge, particularly within the higher skill groups. An interesting outcome of the investigation of lunge tasks was the evidence that altering established movement protocols can result in enhancement of
musculoskeletal functioning, increase gait efficiency and decrease the perceived injury risk.

It may therefore be speculated that the high physiological responses in badminton and the relatively high occurrence of lower limb injury are related to other factors. While sidestepping and crossover-stepping at the magnitudes tested in the laboratory were inconspicuous in their biomechanical and physiological demands their application in the game is likely affected by the short durations of activity that are a characteristic part of the sport. Therefore, it may be the demands of starting and stopping rather than the continuation of the movement that cause larger stress on the musculoskeletal system of the lower limb. Furthermore, as seen with lunging, movement combinations, such as sidestepping with a large lateral jump, are a common occurrence in badminton. Lower limb injuries may therefore be the consequence of less frequent movements that result in higher impact or larger muscular demands. Furthermore, the dynamics of the game and the use of atypical movements may have a combined effect. As seen in the example of rear-court lunging, the need to initiate different movements quickly can result in injury processes due to the misalignment of the limb during the force production or indeed during weight acceptance phase. The following section summarises some of the recommendations for the future examination of the relationship of movement and injury in the context of badminton.

10.4 Suggestions for Future Research

A logical continuation of the biomechanical research into badminton-specific locomotion would be the investigation of the relationship of speed of locomotion on the metabolic demands of lateral movements. Use of a treadmill with a large running surface and high tensile strength may prove useful in specifying speed of locomotion and allow for a more precise estimate of the influence of skill or biomechanical variables between subjects. This information would provide very useful information for the examination of the metabolic differences between lateral movements and running at equivalent speeds.
The data presented in this thesis show relatively low physiological demands of lateral movements. However it may be argued that it is the use of constant starting and stopping movements that contribute to the large physiological demands. Furthermore, it may be argued that it is the use of combinations of movements which result in increased risk of injury. There are a number of other sport-specific movement tasks which appear to be of particular interest for future research. These include: lateral stepping in combination with lateral jumping; rear-court lunging and the risks associated with rotational forces during weight acceptance as well as recovery (as shown in Figure 9.12); changes in lunging technique with the adoption of a more dynamic weight acceptance phase, involving a two-footed landing as well as recovery; and an evaluation of the forces acting on the trailing limb during SS + lunge and XS + lunge combination movements during lunges of different magnitude.

Assessment of the effect of adjustments to lunging technique may allow for improved performance of the athlete as well as reduced risk of injury by reducing the dynamics of weight acceptance. Furthermore, additional information on the effects of lunge task modifications, in terms of energy requirements and biomechanics, may provide useful information for coaches and players on the benefits of the badminton-specific lunge task. Future research should examine the effect of changes of lunge technique in relation to lunging at different magnitudes and different levels of fatigue. Evaluation of the long-term effects of lunge technique may provide valuable information on the consequences and benefits of different lunging and lunge recovery techniques.
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A1.1 General Health Questionnaire

HEALTH SCREEN QUESTIONNAIRE FOR STUDY VOLUNTEERS

- As a volunteer participating in a research study, it is important that you are currently in good health and have had no significant medical problems in the past. This is (i) to ensure your own continuing well-being and (ii) to avoid the possibility of individual health issues confounding study outcomes.
- If you have a blood-borne virus, or think that you may have one, please do not take part in this research [include for projects involving invasive procedures].

Please complete this brief questionnaire to confirm your fitness to participate:

1. **At present**, do you have any health problem for which you are:
   - (a) on medication, prescribed or otherwise
   - (b) attending your general practitioner
   - (c) on a hospital waiting list

2. **In the past two years**, have you had any illness which required you to:
   - (a) consult your GP
   - (b) attend a hospital outpatient department
   - (c) be admitted to hospital

3. **Have you ever** had any of the following:
   - (a) Convulsions/epilepsy
   - (b) Asthma
   - (c) Eczema
   - (d) Diabetes
   - (e) A blood disorder
   - (f) Head injury
   - (g) Digestive problems
   - (h) Heart problems
   - (i) Problems with bones or joints
   - (j) Disturbance of balance/coordination
   - (k) Numbness in hands or feet
   - (l) Disturbance of vision
   - (m) Ear / hearing problems
   - (n) Thyroid problems
   - (o) Kidney or liver problems
   - (p) Allergy to nuts

4. Has any, otherwise healthy, member of your family under the age of 35 died suddenly during or soon after exercise? Yes No

If YES to any question, please describe briefly if you wish (eg to confirm problem was/is short-lived, insignificant or well controlled.)

5. **Additional questions for female participants**
   - (a) are your periods normal/regular?
   - (b) are you on "the pill"?
   - (c) could you be pregnant?
   - (d) are you taking hormone replacement therapy (HRT?)

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A1.2 Informed Consent Form

Insert Name of Research Proposal

INFORMED CONSENT FORM
(to be completed after Participant Information Sheet has been read)

The purpose and details of this study have been explained to me. I understand that this study is designed to further scientific knowledge and that all procedures have been approved by the Loughborough University Ethical Advisory Committee.

I have read and understood the information sheet and this consent form.

I have had an opportunity to ask questions about my participation.

I understand that I am under no obligation to take part in the study.

I understand that I have the right to withdraw from this study at any stage for any reason, and that I will not be required to explain my reasons for withdrawing.

I understand that all the information I provide will be treated in strict confidence.

I agree to participate in this study.

Your name

Your signature

Signature of investigator

Date
### Difficulty Rating:

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### Perceived Injury Risk:

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Comments:
Participant Feedback Form: Lungeing and Lunge Recovery

Participant: ________________________________

Movement Performed: __________________________

Preference of Performing the Movement:

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Self Perceived Level of Difficulty:

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Self Perceived Risk of Injury to the Lower Limb:

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