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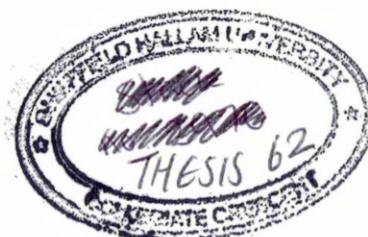
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**THE MEASUREMENT OF VARIABILITY IN
COORDINATION DURING LOCOMOTION**

Jonathan S. Wheat

A thesis submitted in partial fulfillment of the requirements of
Sheffield Hallam University
for the degree of Doctor of Philosophy

May 2005





Abstract

Traditionally in sport and exercise biomechanics, variability in movement has been regarded as system noise or error. However, the advent of dynamical systems theory has stimulated a radical reassessment of the concept of variability with it being regarded as functional in some circumstances. With the application of dynamical systems theory to studies in biomechanics, many methodological considerations have emerged. These include the effects of measurement error on observed variability, the suitability of techniques available for quantifying coordination variability and the efficacy of using a treadmill to simulate overground locomotion in terms of coordination variability. The overall purpose of this thesis was to address the issues related to the methodological considerations for studies of coordination variability. In Chapter III, the effects of measurement error on coordination variability were investigated. It was suggested that, of the two major components of measurement error – skin movement and instrument errors - skin movement errors are likely to have a greater effect on measures of coordination variability due to their larger magnitude. If skin movement errors were entirely random, rather than systematic, they would be extremely detrimental to the study of coordination variability, as their magnitude is often greater than the magnitude of variability in human movement. Preliminary analysis of data presented by Holden *et al.* (1997) revealed that errors due to skin movement were primarily systematic, which suggests that they are less problematic for investigations of variability. In Chapter IV, the suitability of techniques available for quantifying variability in coordination was assessed. The results of an analysis of supporting experimental data suggested that disparate answers to certain research questions might be obtained dependent on the technique used to quantify variability in coordination. It is clear from Chapter IV that comparisons between studies of coordination variability that used different quantification techniques should be made with caution. Researchers should be aware of the benefits and limitations of each technique and the choice of technique should be based on the research question of interest. A comparison of variability in coordination measured overground and on a treadmill was provided in Chapter V. Results indicated that overground running was associated with greater variability in coordination than treadmill running. Therefore, it is possible that performing studies on a motorised treadmill might mask differences in coordination variability between experimental groups. Potentially, the reduced coordination variability during treadmill running can be explained by the artificially constant speed of the treadmill belt externally driving the foot through the stance period. The ‘treadmill-on-demand’ (Minetti *et al.*, 2003) is an innovative type of feedback controlled treadmill on which the belt speed is not constant so the participant is not constrained to run at a constant speed. Therefore, in Chapter VI, variability in coordination measured during overground, conventional treadmill and treadmill-on-demand running was compared. Treadmill-on-demand running resembled overground running no better, in terms of variability in coordination, than conventional treadmill locomotion and it appears to be no more suited to studies of coordination variability than the conventional treadmill. Although some unanswered questions remain, this thesis has enhanced understanding of important methodological considerations for the study of coordination variability during running.

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CHAPTER I

1 Introduction

The mechanics of running have been studied for many years (McClay, 2000). Modern interest in the biomechanics of running was propagated by its growth as a recreational activity in the 1970s. Since then, biomechanists have investigated questions related to running mechanics with the aim of enhancing performance (e.g. Saunders *et al.*, 2004) or, more frequently, identifying biomechanical factors that cause overuse injury (e.g. Messier *et al.*, 1991; Lafortune *et al.*, 1994; McClay and Manal, 1997; Stergiou *et al.*, 1999; Hreljac *et al.*, 2000). Traditional approaches to the study of running mechanics have been greatly influenced by theories of the cognitive approach to movement control. Consequently, studies in biomechanics have tended to focus on identifying the invariant properties of human movement that are predicted by the information-processing models of cognitive psychology. Therefore, biomechanics researchers have consistently adopted the assumption that within- and between-participant variability is of little or no importance. Indeed, techniques for reducing and eliminating both within- and between-participants variability are used frequently (e.g. Sadeghi *et al.*, 2000; Hunter *et al.*, 2004; Schwartz *et al.*, 2004; Mullineaux *et al.*, 2004).

Another, more contemporary, approach to movement coordination and control is known as dynamical systems theory (Hamill *et al.*, 1999). Dynamical systems theory applies the mathematics of non-linear dynamics to human movement in an attempt to describe its functional properties (c.f. Beek *et al.*, 1995). This approach

challenges traditional views of movement variability which assume variability to be system noise or error that must be eliminated. Indeed, a major tenet of dynamical systems theory is that variability is functional. Hamill *et al.*, (1999) stated that a central message of the work in motor control from a dynamical systems perspective (e.g. Kelso, 1981, 1984; Schöner *et al.*, 1986) is that variability in movement is necessary for changes in the coordination of movement. Further, the idea that variability is functional is gaining recognition in a wide variety of disciplines including, for example, cardiac dynamics (e.g. Goldberger *et al.*, 1990) and brain pathology (e.g. Stam *et al.*, 1994).

Recently, various authors have recognised the notion that variability is functional and have adopted methods from the dynamical systems approach to address biomechanical research questions (e.g. James, 1996; Holt *et al.*, 1995; Hamill *et al.*, 1999; Heiderscheit *et al.*, 1999; James *et al.*, 2000; Stergiou *et al.*, 2001a,b; Heiderscheit *et al.*, 2002; Kurz and Stergiou, 2003). As a result of this work, several authors have postulated that another function of movement variability might be to attenuate impact shocks during activities in which performers are subjected to large forces e.g. running (James, 1996; Holt *et al.*, 1995; Hamill *et al.*, 1999; Heiderscheit *et al.*, 1999; James *et al.*, 2000; Heiderscheit *et al.*, 2002; James, 2004). These authors suggested that variability in movement might provide a broader distribution of stresses among different tissues, potentially reducing the cumulative load on internal structures of the body. Further, James (2004) recently formulated the ‘variability-overuse injury hypothesis’ – see section 2.5.1 – in support of which some experimental evidence exists. For example, James *et al.*

(2000) noted that, during a drop jump task, 'injury-prone' participants exhibited significantly lower variability for some joint kinetics parameters than a healthy control group. However, this was only apparent for selected dependent variables; in others, there were either no significant differences or increased variability was noted in the injury-prone group. Because of the potential functional roles of movement variability, it would appear that there is a need to re-assess the solely negative views of variability.

Traditionally, dependent variables in studies of running biomechanics have also tended to be discrete data from isolated joints (e.g. Paradisis and Cooke, 2001). However, the dynamical systems approach advocates that the coordination or coupling between joints of the lower extremity is important. Running is a complex motor skill that involves many degrees of freedom. To produce coordinated movement and master the myriad of interacting components in the human body, the runner must solve what Bernstein (1967) termed the 'degrees of freedom problem'. Dynamical systems theorists have proposed that, to aid in managing the degrees of freedom in the system, synergies emerge between the different interacting components. These synergies are known as coordinative structures which are an important facet of the dynamical systems approach and allow for a solution to the degrees of freedom problem (Turvey, 1990). Recently, many authors have recognised that analysing discrete variables from isolated joints does not effectively capture the complexity of the coordinated motions of components of the body. An excellent example of this during running is the coordinated actions of the subtalar and knee joints. Briefly, both knee flexion and subtalar eversion

promote internal rotation of the tibia. Conversely, subtalar inversion and knee extension promote external rotation of the tibia. Therefore, it has been suggested that a disruption to the coordination between the subtalar and knee joints during the stance phase of running might create torsional stresses on the tibia and abnormal loads on the knee joint (e.g. Bates *et al.*, 1979; Hamill *et al.*, 1992; McClay and Manal, 1997; Stergiou and Bates, 1997; Stergiou *et al.*, 1999; DeLeo *et al.*, 2004). With this in mind, investigating the actions of the subtalar and knee in isolation might omit important information about running injury mechanics.

Hamill *et al.* (1999) were amongst the first to use the dynamical systems approach to investigate overuse running injuries. These authors recognised the two important tenets of dynamical systems theory outlined previously in this section – the importance of movement variability and inter-segment coordination. Using a retrospective research design, they compared lower extremity coordination variability of participants with patellofemoral pain with a group of healthy, matched controls. Less variability was reported for the patellofemoral pain group than the control group (Hamill *et al.*, 1999). Potentially, these results provide further support for the hypothesised link between variability and overuse injury. Follow-up studies (Heiderschiet, 2000b; Heiderschiet *et al.*, 2002) cited similar results to Hamill *et al.* (1999). However, with the retrospective research designs used in these studies (Hamill *et al.*, 1999; Heiderschiet, 2000b; Heiderschiet *et al.*, 2002) it is impossible to determine whether the decreased variability was the cause or the effect of the patellofemoral pain. In addition to the possibility that lower variability caused the injury, it is just as plausible that the decreased variability seen in the injured

participants was the result of the pain (c.f. Hamill *et al.*, 1999; Heiderschiet, 2000b; Heiderschiet *et al.*, 2002). Hamill *et al.* (1999) suggested that the decreased variability seen in the patellofemoral pain group could have been a result of the participants constraining their movements within tight boundaries inside which the pain was reduced; Heiderscheit (2000b) presented preliminary findings that provide support for this notion. Heiderscheit (2000b) monitored variability in coordination after reduction in pain due to the application of patella taping. It was reported that coordination variability in the injured group increased to be close to that of the healthy group after reduction in pain.

The findings of Hamill, Heiderscheit and colleagues (Hamill *et al.*, 1999; Heiderschiet, 2000b; Heiderschiet *et al.*, 2002), together with the results presented by James and co-workers (James, 1996; James *et al.*, 2000), are important as they have demonstrated a potential relationship between coordination variability and overuse injury. As many authors have highlighted, more work is required to determine whether the decreased variability seen in injured participants is the cause or the effect of the injury. Specifically, work is required to confirm or refute the variability-overuse injury hypothesis presented by James (2004).

The concept of variability in coordination being functional is contrary to the traditional association of variability with pathology. Any study of coordination variability is associated with many methodological considerations, some of which are different to those associated with studies that ignore variability. These methodological considerations are outlined briefly here; see section 2.6 for a

detailed review. An important consideration is the effect of measurement errors on coordination variability. The plethora of research papers that have discussed the effects of measurement error on traditional biomechanical parameters (e.g. Cappozzo *et al.*, 1995; Cappozzo *et al.*, 1996; Holden *et al.*, 1997; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b; Manal *et al.*, 2000) – e.g. segment translations and rotations – gives an indication of their perceived importance for any study of kinematics. However, very little attention has been given to the effects of measurement errors on the study of variability in coordination. Further, it is unclear which of the techniques available for the quantification of coordination variability - e.g. continuous relative phase, relative motion angles, Vector Coding and NoRMS - is most appropriate. Also, it is not clear whether coordination variability measured on a treadmill represents that which would be measured overground. Again, the suitability of using a treadmill to simulate overground locomotion in traditional studies of kinematics and kinetics has received much attention (e.g. Dal Monte *et al.*, 1974; Strathy *et al.*, 1983; Alton *et al.*, 1998; Wank *et al.*, 1998; Schache *et al.*, 2001). However, no studies have investigated the differences in coordination variability between overground and treadmill running.

1.1 Purpose of the thesis

The overall purpose of this thesis was to address the issues related to the methodological considerations for studies of coordination variability that were outlined briefly in the previous sub-section. To provide a clear direction for the thesis, the proposed relationship between variability and overuse injury (*c.f.* James,

2004) was taken as a context upon which to focus. Further, as both the coordination between body segments - see section 2.6.2 - and movement variability have recently been proposed as important in understanding the aetiology of injury this thesis specifically focused on the measurement of coordination variability during locomotion. Subsidiary aims of each Chapter were related to each of the methodological considerations outlined in the previous sub-section. Specifically, the purpose of Chapter III was to investigate the effects of measurement errors on the quantification of variability in coordination. The purpose of Chapter IV was to consider the suitability of different techniques available for the quantification of variability in coordination. The purpose of Chapter V was to compare the variability in coordination measured on a treadmill to that measured overground. Finally, the purpose of Chapter VI was to compare variability in coordination measured overground and on a conventional treadmill to that measured on an innovative treadmill known as the 'treadmill-on-demand'.

1.2 Structure of the thesis

To address the overall purpose of this thesis, it comprises six further Chapters. The structure of the thesis is as follows:

Chapter II provides a critical review of the literature relevant to the programme of research.

Chapter III examines the effects of measurement errors on the quantification of variability in coordination. The Chapter is divided into two sections. The first considers errors due to the measurement system and the second considers errors due to skin marker movement.

Chapter IV examines the suitability of techniques available for quantifying coordination and variability in coordination. With the inclusion of experimental data, it is illustrated that the choice of technique used to quantify coordination is important, as different answers to research questions can be obtained dependent on the choice of quantification technique.

Chapter V examines the suitability of using a treadmill to simulate overground locomotion in terms of variability in coordination.

Chapter VI provides a comparison of coordination variability measured overground and on a conventional treadmill with that measured on an innovative type of treadmill known as the ‘treadmill-on-demand’.

Chapter VII is a summary and discussion of the findings presented in the thesis. Subsequently, an account of the implications of the findings presented in this thesis for future studies of the variability-overuse injury hypothesis is given. Then, limitations of the research programme and directions for future research are discussed. Finally, an overall conclusion is provided.

CHAPTER II

2 Literature review

2.1 Introduction

This review will first introduce research into the analysis of variability in movement. Initially, traditional approaches to characterising variability in human movement will be outlined with a specific focus on how these approaches have been applied to locomotion. Subsequently, a more contemporary approach to the conceptualisation of variability will be outlined, specifically focusing on the dynamical systems theory of ecological psychology. An explanation of how some of the principles of the non-linear dynamics branch of mathematics can be applied to human movement will be given. The major tenets of dynamical systems theory will be explained in the context of some of the fundamental experiments in the area.

Then, studies in which the tools of dynamical systems theory have been applied to human locomotion will be highlighted. Also, the relationship between variability and health will be given particular attention and a hypothesis proposing a relationship between variability and overuse injury will be detailed. Finally, considerations for the study of coordination variability during locomotion in the context of this relationship between variability and overuse injury will be presented and reviewed.

2.2 Variability in movement

Variability is an inherent component within and between all biological systems (Newell and Corcos, 1993). Bernstein (1967) suggested that, for coordinated movement to arise, the numerous functional degrees of freedom of the body must be organised in time and sequence to form a functional movement pattern. The huge number of degrees of freedom in the body, with its 10^2 joints, 10^3 muscles, 10^3 cell types and 10^4 neurons and neural connections (Kelso, 1995), emphasises that some variability should be expected in all movements. It would seem impossible to generate identical movement patterns on different attempts at performing the same movement task.

The understanding and approach to variability have changed dramatically over recent decades. Different paradigms in motor control and movement coordination have approached this issue of variability in movement in contrasting ways. The traditional cognitive or information processing approach to motor control views within-participants variability as system noise or error that must be eliminated. However, ecological psychology presents the view, through the area of dynamical systems theory, that some degree of variability has a functional role in human movement.

2.2.1 Traditional views of variability: The cognitive approach

Traditionally, variability has been viewed as a detrimental component present in the human movement system. Ideas from the cognitive approach to psychology,

which were predicated on concepts such as the central pattern generators (von Holst, 1948), have had a major influence on this view. Generally, from this standpoint, it is suggested that any error in motor output represents error in the planning, execution and outcome of a task (James, 2004); these suggestions agree with concepts from theories employing information-processing models of motor control. One such theory, known as the Impulse-variability Theory (Schmidt *et al.*, 1979) suggested that the variability in the outcome of a movement is proportional to the variability in the impulse that created the movement. Schmidt *et al.* (1979) proposed three reasons for increased variability:

1. Error in the selection of the appropriate motor program for a given task (central command error).
2. Scaling errors in selecting the parameters needed for execution of the motor program (central error, peripheral error, or both).
3. Random noise in the system as the program is executed (peripheral error).

A common theme apparent in the reasons cited by Schmidt *et al.* (1979) is the view that the variability is random error that should be eliminated and that the variability has little or no deterministic component. The traditional, cognitive motor control literature is replete with studies citing similar explanations for variability in the human movement system. Another example is the extremely influential Schema Theory (e.g. Schmidt, 1975; Adams, 1986; Masson, 1990; Shea *et al.*, 1990). A schema is a set of abstract generalised motor programs that can be adapted for particular movement. Crussemeyer (1998) stated that movement control is

governed by these set programs and cited the same reasons for variability as presented by Schmidt *et al.* (1979). The theory suggests that movement is controlled by a set of programs, so each movement outcome should ideally be exactly the same as the last. Therefore, in schema theory, variability in the movement outcome is viewed as error or a hindrance. Variability is only viewed as constructive during practice where the schema is improved when practice conditions are diverse (Schmidt and Lee, 1998).

Biomechanics has been greatly influenced by the views and theories of the cognitive approach to movement control. Consequently, studies in biomechanics have tended to focus on identifying the invariant properties of human movement that are predicted by information-processing models in an attempt to identify models of performance to which all performers should aspire. An example of such an approach in biomechanics is the 'Elite Performer Template' approach used by Dapena (1984) which, essentially, takes the technique of top performers as ideal (Bartlett, 1999). Obviously, this approach assumes that within or between participants variability has little or no importance. Furthermore, there has been much attention paid to reducing the between- and within-participant variability to increase statistical power (e.g. Sadeghi *et al.*, 2000; Hunter *et al.*, 2004; Schwartz *et al.*, 2004; Mullineaux *et al.*, 2004). For example, Mullineaux *et al.* (2004) recently presented an offset-normalising technique to reduce variability and increase statistical power in biomechanical studies. Using this technique, each individual participant's scores for each dependent variable are offset-normalised using the mean value of the entire sample for that dependent variable. In addition to this,

multiplicative scatter correction was applied to the data, whereby the original data is normalised using a ratio of linear regression coefficients between the mean group and each participant. Both normalisation procedures resulted in a reduction of between-participants variability and an increase in statistical power (Mullineaux *et al.*, 2004).

2.2.1.1 Traditional views of variability during locomotion

Van Emmerik *et al.* (2000) reported that a common assumption in studies of locomotion is that increased variability in gait parameters such as stride length and stride frequency is associated with instability during locomotion. This has led to the general assumption that increased levels of variability during locomotion is indicative of pathology, instability and an increased risk of falling in some populations (Heiderscheit, 2000a). Examples of studies that have lead to these assumptions have usually reported results based on traditional gait parameters (e.g. Stride length, stride frequency, step length, step frequency, stride width). For example, Gabell and Nayak (1984) measured the step-to-step variability of stride time and double support time in walking; they reported that increased variability could predict an individual's risk of falling. This was confirmed by Hausdorff *et al.* (1997a) who stated that the stride, stance and swing duration were more variable in elderly fallers than in elderly non-fallers or young participants. The stride-to-stride variability in spatial gait parameters has also been the subject of study. Maki (1997) reported that an increase in the stride-to-stride variability in stride length increased the likelihood of elderly participants experiencing falls. More recently, other authors have reported similar findings. For example, Mbourou *et al.* (2003), in

an investigation of step length variability at gait initiation, reported that elderly fallers exhibited more than twice the variability of elderly non-fallers and young adults. Furthermore, Sheridan *et al.* (2003) reported that elderly adults with Alzheimer's disease exhibited greater stride time variability than elderly adults without Alzheimer's. However, Owings and Grabiner (2004) suggested that step width variability but not step length or step time variability discriminated effectively between the gait of healthy young and older adults.

In addition to being associated with the elderly and falling, increased variability in stride characteristics has also been seen among individuals with neuromuscular disease. For example, Hausdorff *et al.* (1998) reported increased stride-to-stride variability in stride time and double support time in patients with Parkinson's disease and patients with Huntington's disease when compared to a control group. Further, in a recent study, Hausdorff *et al.* (2003) again demonstrated increased variability in stride time in patients with Parkinson's disease.

However, as Heiderscheit (2000a) highlighted, depending on the gait parameter, it has been suggested that variability serves opposing roles. As suggested above, variability in basic stride characteristics – characteristics of the outcome of the movement - is generally associated with pathology and construed as detrimental. This is similar to work in the traditional approaches to understanding movement control in which outcome-based measures - for example, accuracy scores (e.g. Sherwood, 1988) and force production (e.g. Schmidt *et al.*, 1979) - have been used. However, evidence presented in sections 2.3, 2.4 and 2.5 suggests a

functional role for variability when considering the motor patterns that result in the outcome of the movement. The following section outlines a more contemporary approach to movement control, known as the dynamical systems approach, in which a functional role is attributed to variability.

2.3 Contemporary views of variability: The dynamical systems approach

Dynamical systems theory applies the mathematics of non-linear dynamics to human movement in an attempt to describe its functional properties (*c.f.* Beek *et al.*, 1995). In the movement sciences, Kugler *et al.* (1980) were among the first to introduce this approach in their seminal paper. This, and much early work, centred on the continuous movements of finger and wrist adduction-abduction. The major tenets of dynamical systems theory will be highlighted with reference to these early studies.

In Kelso's early experiments (1981, 1984), the task for the participants was to oscillate the index fingers of both hands back and forth (adduction-abduction), with the same frequency for each finger. Participants could stably and reproducibly perform two basic patterns, in-phase (homologous muscle groups contracting simultaneously) and anti-phase (homologous muscle groups contracting alternately). A pacing metronome was used to control the frequency of movement. Participants were required to perform one full cycle of movement with each finger for each beat of the metronome. They were also instructed not to resist any changes in the pattern of finger movement and to stay simply in the pattern that

was most comfortable. When participants, initially moving in the anti-phase mode, were instructed to increase cycling frequency, an involuntary and abrupt shift to the in-phase mode was observed at a critical frequency (Kelso and Schöner, 1988). These changes exhibit properties similar to those of non-equilibrium phase transitions observed in many physical systems. However, no such transition occurred when participants started in the in-phase mode. Thus, while people can produce two stable patterns at low frequencies, only one pattern remains stable as frequency increases beyond a critical value (Kelso, 1995). The coordination of the two fingers, and thereby the transition between the two modes, was monitored by calculating the phase relationship between the two fingers. Hamill *et al.* (2000), outlined ways in which the coordination between two bio-physical oscillators (the fingers in this case) can be monitored, including the specifics of how to calculate the phase relationship between two segments (relative phase) - for details of how relative phase is calculated see section 2.6.2.1.

In the above experiments, participants were drawn to either an in-phase or an anti-phase pattern of coordination. In dynamical systems theory, these stable patterns are known as attractors, or attractor states. Attractors are preferred patterns and represent stable areas of movement around which behaviour tends to occur when a system is allowed to perform in its preferred manner (Kelso, 1995). Furthermore, attractors identify preferred regions in a state space, which is made up of relevant system variables (van Emmerik and van Wegen, 2000). Dynamical systems theory emphasises the identification of variables that can help us to investigate the dynamics of the attractors in a given system (Stergiou *et al.*, 2001b). The

identification of a collective variable (also known as an order parameter) allows the collective state of the system to be classified. Collective variables are low-dimensional and functionally specific (Stergiou *et al.*, 2001b), meaning that the many degrees of freedom of the body can be described by one variable. These collective variables enable the identification of the coordinative state to which the system has been attracted at any one time. In the early Kelso experiments, the collective variable was defined as the relative phase (Φ) between the two fingers and the two emerging attractor states were in-phase ($\Phi = 0^\circ$), and anti-phase ($\Phi = \pm 180^\circ$).

Another important variable in dynamical systems theory is the control parameter. The scaling up or down of the control parameter leads the system through its potential states (Kelso, 1995). In the Kelso experiments, the control parameter was the movement frequency. Scalar changes in the control parameter, namely the increase in movement frequency, induced an abrupt change in the collective variable describing the system, from anti-phase ($\Phi = \pm 180^\circ$) to in-phase ($\Phi = 0^\circ$).

The abrupt changes in the patterns of movement for the fingers, induced by an increase in the control parameter, are accompanied by specific collective variable characteristics that are predicted by stochastic dynamic theory, formulated by Schöner *et al.* (1986). The major characteristic of the transitions in coordination was the presence of fluctuations when the participants were in the anti-phase mode just before the transition to the in-phase mode. Kelso (1981, 1984) measured, using standard deviation, the variability of the relative phase between

the two fingers as movement frequency was increased. It can be seen from Figure 2.1 that the variability in the results increased until the transition, before suddenly decreasing after the change had occurred to a low value similar to that of the in-phase mode.

Within the dynamical systems approach, this variability is felt to be essential in inducing a coordination change. The variability gives an indication of the stability of the movement. This can be seen with the finger experiments; as the movement reaches the transition frequency it becomes increasingly unstable, as indicated by the large standard deviation in the data. The flexibility in movement can be established through monitoring the variability (Hamill *et al.*, 1999). Without the variability in movement, the action being performed would not be flexible enough to adapt to changes in the environment, for example. Hamill *et al.* (1999) stated that the central message from Kelso's research is that variability is a necessary ingredient for coordination change.

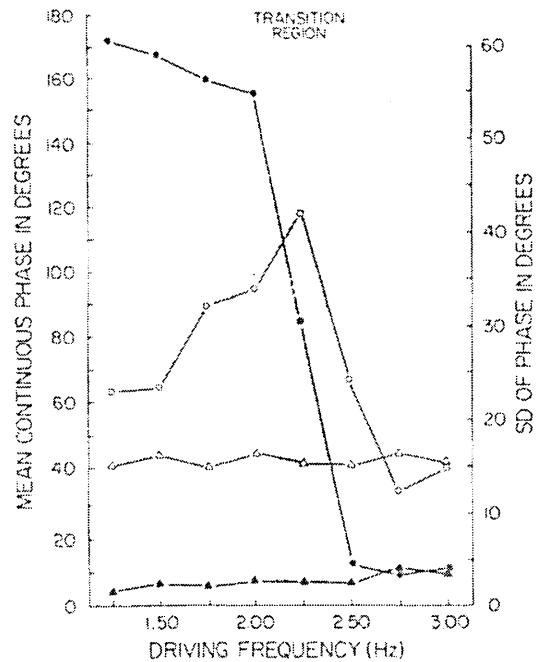


Figure 2.1: The average mean relative phase for the in-phase (closed triangles) and anti-phase (closed circles) modes of coordination and the average standard deviation (in-phase = open triangles, anti-phase = open circles) as a function of driving frequency for a set of 10 experimental runs. From Kelso and Schöner (1988).

2.3.1 The relationship between variability and health

The influences of dynamical systems theory and analysis techniques from areas such as non-linear dynamics have recently led to the association between some degree of variability and biological health. This is contrary to the earlier view that the dynamics of healthy biological systems are associated with orderedness and regularity (Goldberger and West, 1987). However, there are now many examples in the literature of increased variability being associated with biological health (e.g. Goldberger and West, 1987; Goldberger *et al.*, 1990; Gallez and Babloyantz, 1991; Stam *et al.*, 1994; van Emmerik *et al.*, 1999). These include examples from cardiac dynamics (e.g. Goldberger *et al.*, 1990), brain pathology (e.g. Stam *et al.*, 1994)

and other neurological disorders (e.g. van Emmerik *et al.*, 1999). Only the relationship between variability and healthy cardiac dynamics is discussed further here – see James (2004) for a review of relationship between variability and brain pathology and other neurological disorders.

As van Emmerik and van Wegen (2000) highlighted, a new perspective on abnormal cardiac dynamics has emerged from the studies of Goldberger and colleagues (Goldberger *et al.*, 1986; Goldberger and West, 1987; Goldberger *et al.*, 1990). Results of their studies indicated that healthy cardiac functioning might be more variable than first thought. Furthermore, it would appear that unhealthy cardiac functioning is characterised by more regular and stable cardiac dynamics. Also, Kleiger *et al.* (1987) reported that, following myocardial infarction, a decrease in heart rate variability was a strong predictor of mortality. This was reiterated by Tsuji *et al.* (1994) who demonstrated a significant link between heart rate variability and survival rate. In summary, healthy heart rhythms appear to be associated with greater system complexity¹ and often greater variability (Goldberger and West, 1987). These studies provide evidence that, contrary to traditional views, variability is synonymous with healthy biological states and this has had an impact on how the role of variability in physiological processes should be perceived (van Emmerik

¹ Complexity is inherently linked with other concepts in physics and biology including randomness and entropy which has produced several working definitions (*c.f.* Vaillancourt and Newell, 2002). However, the most commonly accepted definitions are driven by the operational consideration of the number of system elements and their functional interactions.

and van Wegen, 2000). Furthermore, variability might distinguish between healthy and diseased biological systems (James, 2004).

However, as Hamill *et al.* (2005) noted, this is not to say that all variability is beneficial. Rather, the studies cited in this section challenge the notion that all variability is detrimental (Hamill *et al.*, 2005). For example, the investigations of Goldberger and colleagues, amongst others, and the advent of dynamical systems theory, have led to a new view of variability in terms of biological health. In line with this, recently, the application of dynamical systems theory to the study of locomotion has received much appraisal. This research is reviewed in section 2.4 and, subsequently, research into the relationship between coordination variability and pathology during locomotion is reported in section 2.5.

2.4 Dynamical systems theory and locomotion

The methods outlined briefly in section 2.3 have been used to study the dynamics of both human and animal gait (e.g. Alexander and Jeyes, 1983; Clark and Phillips, 1993; van Emmerik and Wagenaar, 1996; Diedrich and Warren, 1995; Hamill *et al.*, 1999; Heiderscheit *et al.*, 1999; Li *et al.*, 1999; Stergiou *et al.*, 2001a; Stergiou *et al.*, 2001b). Diedrich and Warren (1995) studied the walk-to-run and run-to-walk transitions in human locomotion. They tested the assumption that the transition from walking to running is governed by the same dynamical laws first discovered in Kelso's early experiments. They used the discrete relative phase - see section 2.6.2.1.1 - between segments of the lower extremity as order parameters to describe the collective state of the system. Specifically, these were the relative

phase between the ankle plantar flexion-dorsiflexion angle and knee flexion-extension angle and the relative phase between the ankle plantar flexion-dorsiflexion angle and the hip flexion-extension angle. These variables were chosen because they were thought to be adequate to describe the collective state of the system, which in this experiment was the right lower extremity. The control parameter chosen for the experiment was speed of locomotion, which was manipulated with the use of a treadmill. The control parameter was increased or decreased, with the participants completing both walk-to-run and run-to-walk transitions.

The authors suggested that the walk-to-run transition exhibited four hallmarks of a non-equilibrium phase transition. The four specific characteristics present were:

1. A qualitative re-organisation in the phasing of the segments in the right lower limb at the transition.
2. This qualitative shift occurred in a sudden jump within a single stride cycle.
3. A hysteresis effect was present (the transition back to walking from running occurred at a slower speed than the original transition from walking to running).
4. Fluctuations in relative phase increased near the transition for walking and running (i.e. variability in the relative phasing increased near to the transition region).

They concluded that the results provided support for dynamical systems theory and that changes in the parameters of gait described in the study behave like non-equilibrium phase transitions (Diedrich and Warren, 1995). It would also seem that the results of this investigation provide evidence of an important role for movement variability in locomotion and, as in Kelso's finger experiments; it seems to be vital for coordination change. Recently, Kao *et al.* (2003) repeated the experiment of Diedrich and Warren (1995) but used the continuous method of relative phase calculation as opposed to the discrete method. Kao *et al.* (2003) reported findings that conflicted with those presented by Diedrich and Warren (1995). Specifically, although a significant change in intra-limb coordination occurred before and after the walk-to-run or run-to-walk transition, variability of continuous relative phase did not increase before either transition. Kao *et al.* (2003) concluded that there must be a mechanism forcing gait transitions other than changes in coordination variability. Further, they suggested that this was due to the fact that during locomotion, the motor system has to handle a dual-control problem – that of maintaining balanced upright posture and coordinating the limbs. This conclusion certainly has merit and is reinforced by Newell and McDonald (1994) who suggested that the variability of coordinated limb movements should be considered within the context of the postural stability requirements of the task. However, the efficacy of using continuous relative phase as a dependent measure in studies of gait is questionable and this subject warrants further work - see sections 2.6.2.1.2 and 4.4.

Li et al. (1999) used the tools of the dynamical systems approach to movement coordination to compare the gait patterns of walking and running at similar speeds. Continuous relative phase was calculated between the thigh and lower leg angles. The authors concluded that the patterns of relative phase were similar for the two forms of locomotion except during 20-40% (late stance) of the gait cycle. They concluded that the variability in relative phase was not significantly different between the two forms of locomotion. However, it is possible that the values for relative phase variability were unusually high in both modes of locomotion. Participants were forced to walk and run at the same speed ($2.24 \text{ m}\cdot\text{s}^{-1}$). This speed would appear to be both unusually fast for walking and unusually slow for running, meaning that the coordination in both modes of locomotion might have been unstable (high variability). Because of the issues of stability owing to the forced locomotion velocities, a measure of variability may not have been appropriate in this study. Also, *Li et al.* (1999) only used one average value of the standard deviation over the entire gait cycle, which may have obscured differences in variability at specific points in the cycle.

Stergiou et al. (2001a) used analytical techniques from dynamical systems theory to investigate intra-limb coordination over obstacles of differing heights. Their dependent variables included a 'frontal relative phase' and a 'sagittal relative phase'. They used a mean value of the continuous relative phase over the impact and active periods, and a measure of the mean standard deviation of relative phase, to give an indication of the variability in relative phase over the two periods. The authors reported that the involved segments became more in-phase in the

impact phase and reduced the independent actions of each segment in both periods of the stance phase. They speculated that this change in coordination may have occurred because of the increases in vertical ground reaction forces associated with an increase in obstacle height. However, there were no statistically significant increases in the variability of the relative phase measures. On visual inspection of the data presented in the paper, a tenuous relationship is apparent between an increase in obstacle height and relative phase variability in the sagittal plane. It is possible that the small sample size in this investigation ($n = 10$) limited the power of the study and the ability to detect differences between conditions.

Stergiou *et al.* (2001b) again studied the intra-limb coordination of the lower extremity while running over obstacles of various heights. However, on this occasion, they only considered the lower extremity in the sagittal plane and used a greater range of obstacle heights. Sagittal plane couplings between the foot and the leg and the leg and the thigh were monitored in the study. In this investigation the highest obstacle was 22.5% of each participant's stature as opposed to 15% in the study of Stergiou *et al.* (2001a). Stergiou *et al.* (2001b) defined their order parameters as the relative phase between both the thigh and shank segments and the shank and foot segments. As in the previous experiment, mean relative phase and the variability of the relative phase were averaged over two different phases and used as the dependent variables in the study. However, the phases were defined slightly differently as a pre-landing phase and a stance phase. The authors concluded that evidence of change did exist in the coupling between the foot and shank segments at the 15% obstacle height. The two segments changed to a more

out-of-phase pattern, which may be associated with increased plantar flexion at the ankle and more of a fore-foot strike pattern (Stergiou *et al.*, 2001b). The relative phase variability for the foot and shank coupling increased with obstacle height for both the pre-landing and stance phases. There were also increases in the variability of the relative phase between the shank and thigh segments as obstacle height was increased. Based on the tenets of dynamical systems theory, it might be expected that, if the 15% condition was a value of the control parameter associated with coordination change, variability in the order parameter would increase up to the 15% condition but reduce after the transition. This was not the case in this study but the authors suggest that this may have been due to individual responses of participants to changing obstacle heights. Differential responses of the participants could have 'masked out' any true patterns; the results underline the importance of individual variability (Stergiou *et al.*, 2001b). In all couplings and in all portions of the stride, variability in the relative phase measures was highest in the condition with the greatest obstacle height. It should be noted, however, that the methods employed by Stergiou *et al.* (2001a,b) to determine coordination and coordination variability are questionable – see Chapter IV.

The coordination between the thorax and pelvis has also received attention in the literature. Van Emmerik and Wagenaar (1996) investigated the effect of walking velocity on thorax/pelvis coordination. They used continuous relative phase to monitor the changes in coordination between the thorax and pelvis as the control parameter of walking speed was changed. The relative phase between the two segments changed from a more in-phase pattern at lower velocities to a more out-

of-phase pattern at higher velocities. The results led the authors to question the traditional distinctions between the two modes (walking and running) in human gait as walking at different velocities appears to elicit different coordination patterns between the pelvis and thorax. Trunk coordination was also investigated in a more recent investigation by La Fiandra *et al.* (2003) using relative phase techniques. They investigated the effect of load carriage on trunk coordination. The introduction of a backpack resulted in significantly decreased continuous relative phase between the thorax and pelvis. Similarly to van Emmerik and Wagenaar (1996), an increase in walking velocity resulted in an increased continuous relative phase between the pelvis and thorax regardless of the load carriage condition.

Lamoth and colleagues have recently undertaken studies investigating the coordination and coordination variability between the pelvis and thorax during gait (Lamoth *et al.*, 2002; Lamoth *et al.*, 2004). Lamoth *et al.* (2002) presented data suggesting that, as speed of locomotion increased, the pelvis and thorax moved from an in-phase to an out-of-phase pattern, and variability in coordination increased. In a more recent study, Lamoth *et al.* (2004) reported that neither the coordination nor the coordination variability between the pelvis and thorax changed as a result of pain or fear of pain during walking. Parenthetically, Lamoth and colleagues used an interesting measure to quantify coordination and coordination variability, which overcomes some of the limitations of continuous relative phase, called relative Fourier phase - see section 2.6.2.1.2.

Due to the possible relationship between the magnitude of the Q-angle and patellofemoral pain syndrome (e.g. Messier *et al.*, 1991), Heiderscheit *et al.* (1999) investigated lower extremity coordination during running, for two groups with either high or low Q-angles. The Q-angle was defined as the angle between the quadriceps vector and the infrapatellar tendon at the centre of the patella. They considered the relative phase variability of three different couples in the lower extremity. These included thigh flexion/extension with leg rotation, thigh adduction-abduction with leg rotation and leg rotation with calcaneal inversion-eversion. In an attempt to provide a more detailed analysis, the stance phase of the running gait cycle was divided into four intervals based on specific rearfoot events. No statistically significant differences were found between the high and low Q-angle groups. However, significant differences were reported between the stance intervals in all segment couplings. In each coupling, relative phase variability was highest in the initial portion of stance.

Heiderscheit *et al.* (1999) offered two possible explanations for this. First, they suggested that it provides an adaptive mechanism to potential external perturbations. They proposed that the increased variability found between coupled segments at heel strike indicates a flexible system. If there was a stable pattern at heel-strike the system might not be able to recover from a perturbation (such as uneven ground), resulting for example in the individual falling. Once the terrain is known the system can become less variable without compromising itself. The second explanation was that the variability at heel strike may be functionally necessary to attenuate the large impact shocks present at the beginning of the

stance phase. A pattern with little variability would result in the same anatomical structures, and the same regions of those structures, receiving repeated impact shocks. The variable pattern present in the investigation might allow for the impact shocks to be imparted to various structures, potentially reducing the cumulative load and the risk of injury.

2.5 Joint coordination variability and pathology

As suggested in section 2.2.1, a common assumption in studies of human locomotion is that increased variability in traditional gait parameters (e.g. stride length and stride frequency) is associated with instability, pathology and ageing. Furthermore Heiderscheit (2000a) indicated that the association of increased variability in stride characteristics with ageing and pathology suggests that movement variability is undesirable. However, all stride characteristic variables are outcome measures. Bernstein (1967) considered human movement to be a process of mastering the degrees in the body to produce a controllable system. The large number of degrees of freedom in the human body means that, to allow reduction of the number of degrees of freedom to a controllable level, coordinative structures must be formed². In view of such a large number of degrees of freedom in the human body, some degree of variability should be expected in the formation of these coordinative structures (Heiderscheit, 2000a). Also, it would appear that there are numerous ways in which to coordinate these degrees of freedom to

² A coordinative structure is a functional synergy between neurons, muscles and joints (Turvey, 1990)

produce the same task outcome (Bernstein, 1967; Newell and Corcos, 1993). Hamill *et al.* (2005) summarised this concept nicely in schematic form (Figure 2.2), highlighting that some degree of variability should be expected in movement patterns.

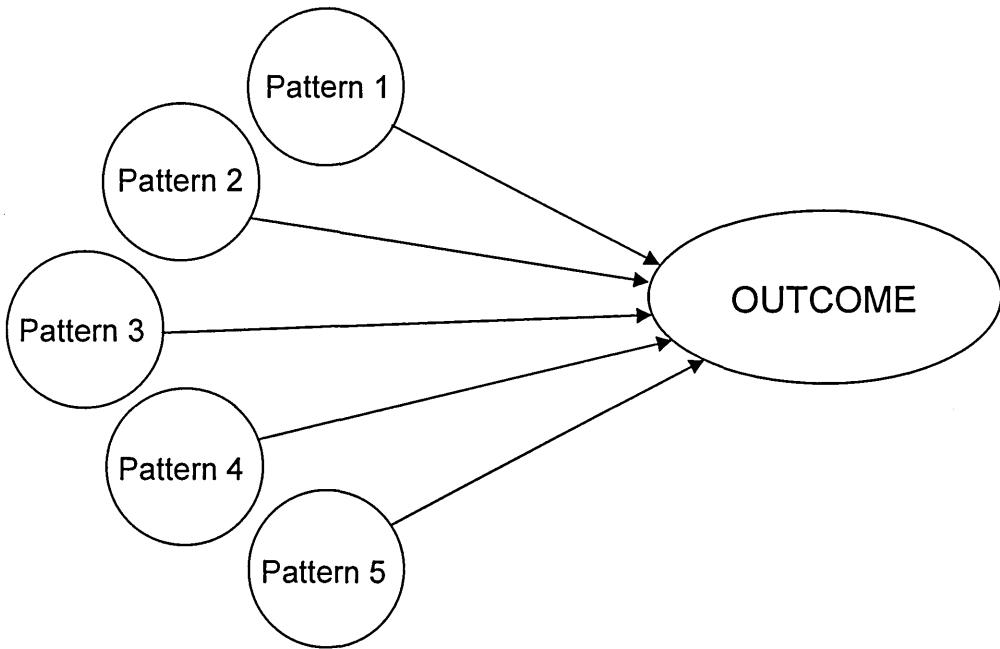


Figure 2.2: A schematic representation of the concept that numerous patterns of movement can produce identical outcomes. Adapted from Hamill *et al.* (2005).

Arutyunyan *et al.* (1969) demonstrated this phenomenon in their frequently cited study. During a pistol shooting task, in which the task criterion was accuracy, expert marksmen exhibited low variability in accuracy scores but accomplished this with a high degree of within-participant variability in upper extremity joint coordination patterns. Conversely, novice marksman exhibited large variability in the outcome of the task with lower variability in joint coordination patterns. It is, therefore, apparent that assessing variability from task outcome or joint

coordination pattern perspectives provides distinctly opposing views. Also, by limiting an investigation to the outcome of a task (i.e. stride characteristics), the variability in the motor patterns used to produce the outcome is ignored (Heiderscheit, 2000a).

Recent theoretical and empirical constructs emerging from the dynamical systems perspective on motor control have strongly challenged the concept that all variability is pathological. Several studies have investigated the relationship between joint coordination variability and patellofemoral pain using a selection of the techniques described in Chapter IV. In a preliminary study, Heiderscheit *et al.* (1998), tried to ascertain the importance of intersegmental coordination variability during running. They found differences in relative phase variability between healthy and patellofemoral pain groups, with the patellofemoral pain participants exhibiting decreased coordination variability. Heiderscheit *et al.* (1998) offered the same explanations for the presence of the variability as Heiderscheit *et al.* (1999). Briefly, they suggested that the variability provides an adaptive mechanism to potential external perturbations and to attenuate large impact forces. In addition to this, they also suggested that whether the decreased relative phase variability in the patellofemoral pain group was the cause or the result of patellofemoral pain needs to be determined through a prospective study.

Similarly to Heiderscheit *et al.* (1998, 1999), Hamill *et al.* (1999) investigated the coordination of the lower extremity during running using the tools from the dynamical systems approach to movement coordination and control. They stated

(p. 298) that: 'It is clear that the actions of the lower extremity are coupled and it is most likely that perturbations to the system can result in injury, particularly the knee'. They then suggested that the coupling relationship had not been clarified using traditional spatial constructs. Hamill *et al.* (1999) examined the coupling of the lower extremity segments in a comparison between participants symptomatic of, and participants free from, patellofemoral pain, along with a comparison between participants with high and low Q-angles. The stance phase of the running gait cycle was divided into four different portions, defined by key events in subtalar joint eversion-inversion. Hamill *et al.* (1999) looked at the same joint couplings as Heiderscheit *et al.* (1999) with the addition of femoral rotation-tibial rotation. The variability in these relative phase variables was also calculated over the portions of stance and included in the analysis. In the first study, comparing high and low Q-angle groups, no statistically significant differences were found in relative phase and relative phase variability. However, there was evidence of systematic differences in relative phase and relative phase variability over specific portions of stance. As in the study by Heiderscheit *et al.* (1999), for all couplings the highest variability in relative phase was in the initial period of the stance phase.

In the second part of the study (Hamill, *et al.*, 1999), participants with and without patellofemoral pain were compared. The variability in the relative phase couplings was less in the symptomatic patellofemoral pain group than in the asymptomatic healthy group. These differences were particularly apparent in the transitions from stance to swing and from swing to stance. Therefore, lower relative phase

variability appears to be an indicator of a non-healthy state (Hamill *et al.*, 1999). Hamill *et al.* (1999, p. 306) concluded that: 'the lack of relative phase variability indicated that segment actions were repeatable within a very narrow range and enabled these individuals to accomplish this task within a minimum of pain'. Had the participants moved out of this range, they might have experienced pain owing to the patellofemoral injury. The authors infer that the presence of low relative phase variability in the lower extremity can indicate the presence of an injury. Hamill *et al.* (1999) also suggested, similarly to Heiderscheit *et al.* (1999), that the patellofemoral pain group exhibiting less relative phase variability might produce an additional overuse injury risk that might compound the original problem. This is caused by a constant stress on the same tissues of the lower extremity. The authors then suggested that the healthy individuals, for whom there were multiple combinations of coupling patterns, would represent an optimal solution because no soft tissue would be repeatedly stressed.

More recently, Heiderscheit *et al.* (2002) conducted an investigation to determine whether individuals with patellofemoral pain displayed a reduction in joint coordination variability compared to non-injured participants. Similar to Hamill *et al.* (1999), they reported decreased intra-limb joint coordination variability in individuals with patellofemoral pain. Heiderscheit *et al.* (2002) again hypothesised that the decreased joint coordination variability in the patellofemoral pain group was a result of the pain they experienced. However, no experimental manipulation was conducted by either Hamill *et al.* (1999) or Heiderscheit *et al.* (2002) to support this common hypothesis. Heiderscheit (2000b) included such a manipulation in an

investigation to determine the effect of a reduction in the pain associated with patellofemoral pain syndrome on joint coordination variability. To assess the relationship between pain and joint coordination variability, the analysis of Heiderscheit *et al.* (2002) was repeated after treatment of the patellofemoral pain. Specifically, the patients were asked to repeat the collection procedures after the application of tape, using the patellar taping technique of McConnell (1986). With an average pain reduction of 60% after the application of the tape, the joint coordination variability increased, which provides support for the hypotheses of both Hamill *et al.* (1999) and Heiderscheit *et al.* (2002). However, the patellar taping technique used in the study of Heiderscheit (2000b) might have affected the formation of coordinative structures in the lower extremity, raising the possibility that the changes in coordination variability might have resulted from an artefact of the tape rather than a reduction in pain.

The studies reviewed in this section indicate that there is a link between the magnitude of coordination variability and overuse injury, but whether the decreased variability is the cause or the result of the injury is not yet known (Hamill *et al.*, 1999). However, a hypothesis to, potentially, explain the link between variability and overuse injury was presented recently by James (2004).

2.5.1 The variability and overuse injury hypothesis

Although many examples of a potential link between variability and health have been reported – see sections 2.3.1 and 2.4 - no direct association has been made between variability and musculoskeletal injury. James (2004) recently presented a

hypothesis that proposed a link between variability and overuse injury. Specifically, James (2004) referred to a link between variability in musculoskeletal loading – measured using internal joint moment calculations – and overuse injury. Although James (2004) explicitly suggested a link between musculoskeletal loading variability and overuse injury, it is reasonable to assume that the ideas can be applied to variability in kinematics and they will be reviewed in light of this.

James (2004) cited anecdotal evidence suggesting that variability might play a positive role in preventing overuse injuries. Variability might provide a broader distribution of stresses among different tissues or alter the stress magnitude, direction, rate and frequency within the same tissue (James, 2004). This anecdotal evidence included the fact that coaches and athletes commonly use techniques like cross-training or periodisation to increase both mechanical and physiological stress variability to reduce cumulative load and prevent overtraining. For example, experienced runners often introduce variability by rotating shoes or varying training surfaces in order to prevent chronic running injuries. Additionally, orthopaedists commonly prescribe different shoes and orthoses to change the kinematics and the distribution of forces among lower extremity structures (James *et al.*, 2000). It is possible that, under circumstances where repetitive loading could cause overuse injuries, inherent movement variability might minimise cumulative load by providing an internal protective mechanism that alters the loading characteristics (James *et al.*, 2000).

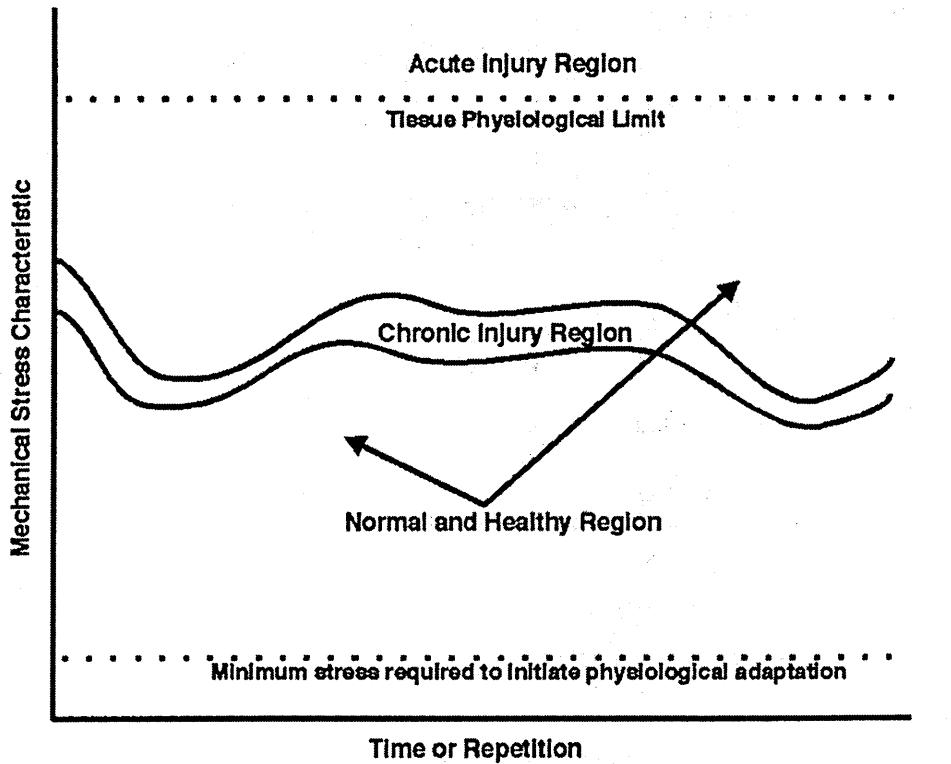


Figure 2.3: Variability model for musculoskeletal health and injury - from James (2004)

A hypothesis relating variability and overuse injury was first presented by James (1996); a schematic representation of the proposed relationship is given in Figure 2.3. The x-axis represents time and the y-axis denotes a ‘generalised characteristic of loading’ – James (1996) used stress magnitude in the current example. The two dotted lines at the top and bottom of the figure represent boundaries of healthy physiological adaptation. The lower boundary denotes the threshold below which no physiological adaptation would be elicited as a result of the stress. The upper boundary denotes the physiological threshold of the tissue; stress values above this threshold would result in tissue failure and this region is designated the acute injury region, which represents traumatic single-event-loading injuries (James,

2004). James (2004) conceded that, in reality, this threshold fluctuates with the integrity of the tissue and thus is dynamic across time. Further, the area between the upper and lower boundaries is denoted as the normal and healthy region. Performances that fall within this region elicit stresses that are of sufficient magnitude to promote physiological adaptation without causing traumatic injury. An additional feature of the diagram is the undulating area in the graph denoted as the chronic injury region. As James (2004) explained, variability in movements would result in a broad distribution of stresses in the region bounded by the upper and lower dotted lines. Occasionally, performances in the overuse injury region should be expected. However, long-term repeated performances within this region are postulated to contribute to overuse injury (James, 2004). James (2004) also suggested that the size of the chronic injury region is susceptible to change. Factors that affect the size of the region include neuromuscular factors, tissue integrity, physiological adaptation and general training variables.

James (2004) also presented a more direct depiction of the hypothetical relationship between variability and injury (see Figure 2.4). The x-axis depicts the relevant variability characteristic, stress variability in this example, and the y-axis represents the likelihood of overuse injury occurring. As variability increases, the likelihood of overuse injury decreases. Conversely, the likelihood of injury is greatest at small magnitudes of variability. It should be noted, however, that as variability is increased the chance of a single traumatic event that might cause injury is increased (James, 2004).

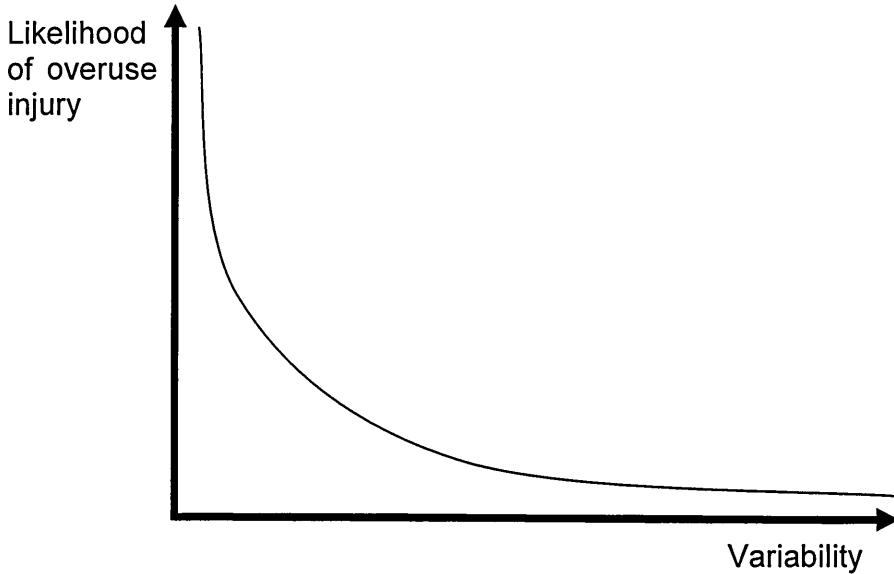


Figure 2.4: Hypothetical relationship between variability and the likelihood of overuse injury - adapted from James (2004)

Figure 2.4, however, fails to depict the influence of other important factors related to overuse injury. If the hypothesis were applied to locomotion, which is the focus of this thesis, an additional dimension could be applied to Figure 2.4. It is generally accepted in the literature that the nature of running injuries is multi-factorial and diverse (Hreljac *et al.*, 2000). These factors have been grouped into intrinsic – including anthropometric and biomechanical factors – and extrinsic factors – such as equipment and frequency, intensity and duration of training. Therefore, if the variability-injury hypothesis proposed initially by James (1996) and refined by James (2004) were applied to running injuries, the relationship might be better depicted by the graph in Figure 2.5, which represents the influence of other factors on the occurrence of injury.

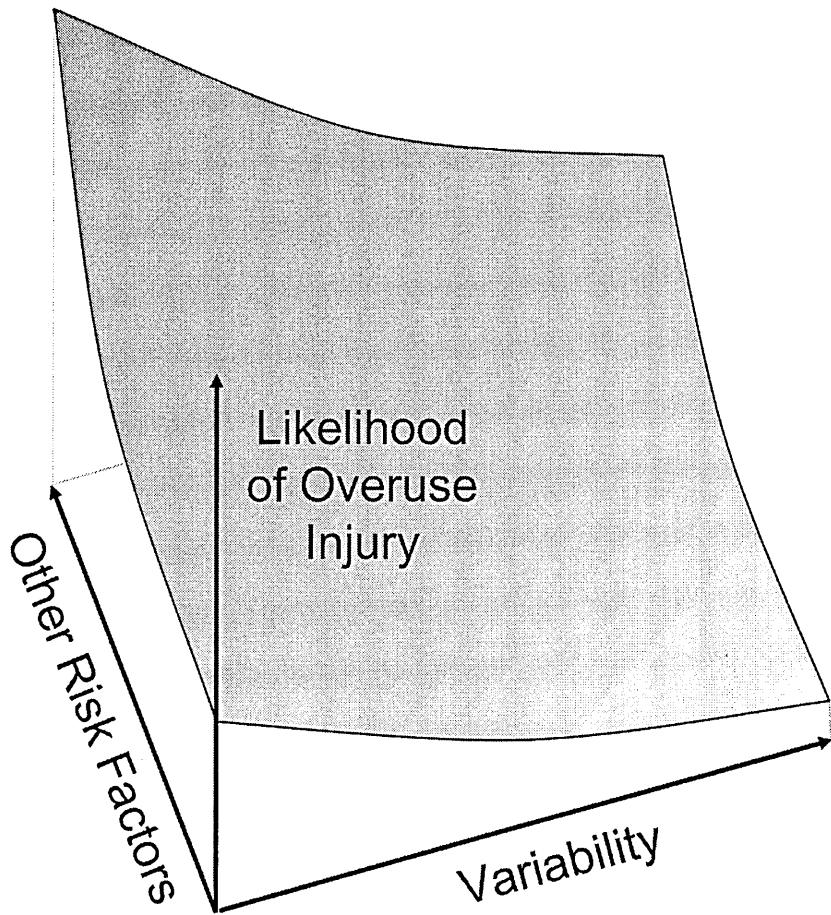


Figure 2.5: Hypothetical relationship between variability and the likelihood of overuse injury with a representation of the influence of other risk factors associated with overuse injury

James (2004) concedes that, at present, the variability-overuse injury hypothesis is speculative. However, experimental evidence that was presented by James *et al.* (2000) has given some early support. These authors examined differences in ankle, knee and hip joint moment variability during jump landings between injury-prone participants and a control group. Significant differences between the groups were seen in each of the loading parameters – time-to-peak-moment, impulse and peak moment. However, the direction of the differences was not always consistent

between dependent variables analysed (James *et al.*, 2000). For example, variability in the time to peak ankle joint moment at the lowest height was greater for the healthy control group whereas variability in the peak ankle joint moment was greater in the injury-prone group. Furthermore, the final dependent variable, joint moment impulse variability, exhibited no significant differences between the healthy or injury-prone groups at any of the joints. A potential problem with the analysis of James *et al.* (2000), however, is that they focused on discrete dependent variables. Many authors (e.g. Hamill *et al.*, 1999) have recently questioned the efficacy of this focus in light of the theoretical considerations that have emerged from the dynamical systems approach to movement control. Therefore, the effect of kinetic and kinematic variability during landing tasks warrants further appraisal. This is confirmed by James (2004) who suggested that further studies are required to confirm or refute the variability and overuse injury hypothesis.

Moreover, there is early empirical support for the hypothesis in locomotion – see section 2.5. Briefly, joint coordination variability has been shown to be reduced in participants with patellofemoral pain when compared to healthy controls (Hamill *et al.*, 1999; Heiderscheit, 2000b; Heiderscheit *et al.*, 2002). However, again, whether the reduced variability was the cause or effect of the injury is unclear. Interestingly, Heiderscheit *et al.* (2000b) reported that variability in the patellofemoral pain participants increased to normal levels when pain was relieved artificially through the application of patellar taping. This provides support for the hypothesis of Hamill *et al.* (1999) who stated that the decreased variability was the result of pain.

Specifically, patellofemoral pain participants constrained movement to be within pain-free boundaries which had the effect of reducing coordination variability (Hamill *et al.*, 1999). However, artificially reducing pain through the application of patella taping might have interfered with typical lower extremity coordination and further studies, monitoring participants over the rehabilitation period, are required to confirm or refute the findings of Heiderscheit (2000b). In conclusion, carefully conducted studies that use a longitudinal research design are required to test the variability and overuse injury hypothesis (James, 2004). It is only this type of research design that can be used to determine whether the decreased variability seen recently in participants with pathology is the cause or the effect of the injury.

So far, this review of literature has highlighted that the traditional view of variability as noise or error that should be eliminated is not necessarily correct. Many different functions of variability have been proposed. These include, 1) providing the system with the flexibility for coordination change in adapting to changing organismic, environmental or task constraints and potential perturbations (e.g. Kelso, 1981, 1984; Diedrich and Warren, 1995, Davids *et al.*, 2003), 2) providing the system with the ability to make slight changes to joint configurations in order to maintain consistency in motor output - compensatory variability (e.g. Arutyunyan *et al.*, 1968; Kudo *et al.*, 2000; Button *et al.*, 2003; Ko *et al.*, 2003) and 3) reduce the cumulative load on tissues during movement (e.g. James, 1996; Hamill *et al.*, 1999; Heiderscheit, 2000; James *et al.*, 2000; Heiderscheit *et al.*, 2002; James, 2004). Only the third proposed function will be considered in detail in this thesis. Many methodological issues have to be considered when assessing variability in

this context. Specifically, methodological considerations for studies³ of coordination variability during locomotion and its relationship with injury - similar to those presented by Hamill *et al.* (1999) and Heiderscheit *et al.* (2002) - are investigated in this thesis.

2.6 Considerations for the study of coordination variability during locomotion

When investigating the relationship between coordination variability and overuse injury, many different methodological considerations emerge. Such considerations include: selection of appropriate statistics for quantifying the magnitude of variability; quantifying the methodological errors that are related to data collection techniques used in studies of locomotion variability; the choice of technique used to quantify coordination variability; the efficacy of using a treadmill to simulate overground locomotion. The purpose of this section of this chapter is to review each of the considerations outlined above.

2.6.1 Measures of variability

Many methods exist for representing variability in a given variable. As James (2004) highlighted, variability can be quantified using traditional or more contemporary, nontraditional, techniques. Generally, nontraditional techniques are those that use methods from the area of non-linear dynamics in an attempt to study

³ Similar to those presented by Hamill *et al.* (1999) and Heiderscheit *et al.* (2002)

the structure of variability. Various authors in the movement control domain have advocated that studies investigate the structure rather than solely the magnitude of variability (*c.f.* Newell and Corcos, 1993). There is evidence to suggest that a deterministic structure exists in some aspects of human variability (Slifkin and Newell, 1998; Riley and Turvey, 2002). Slifkin and Newell (1998), amongst others, suggested that traditional techniques for quantifying variability, such as the standard deviation or coefficient of variation, might mask the important deterministic structure in variability and more complex non-linear techniques, such as approximate entropy and Lyapponov exponent, should be used. However, as stated in section 2.5.1, this thesis is concerned with quantifying variability in coordination during locomotion in a manner that would be relevant to the overuse injury and variability hypothesis. It would appear that the magnitude of any variability in coordination would have greatest affect on overuse injury and, therefore, only traditional linear techniques for quantifying variability will be discussed further⁴.

Traditionally, measurement of variability has been based on a single statistic (Crussemeyer, 1998). The variability in measurement has often been quantified using one or more of five descriptive statistics: range, interquartile range, variance,

⁴ See Stergiou (2004) for an overview of some non-linear methods. Detailed reviews of various non-linear methods are also available. For example, see Pincus (1991) for approximate entropy, Rosenstein *et al.* (1993) for Lyapponov exponent, Riley *et al.* (1999) for Recurrence Quantification analysis and Daffertshofer *et al.* (2004) for Principal Component Analysis.

coefficient of variation and, most commonly, standard deviation (Crussemeyer, 1998).

The range is a measure of the greatest value in a data set minus the least value. A large range suggests that there is a large spread of data and implies that variability is large. Conversely, a small range suggests a small spread and implies that the data are more concentrated about the median score. An extension of this is the interquartile range which represents the central grouping of a set that contains the middle 50% of the data (Coolican, 1999). If the interquartile range is large, the data are widely dispersed, indicating greater variability and vice versa. An advantage of the interquartile range is that, unlike some of the measures to be reviewed subsequently, it is not adversely affected by outliers i.e. extreme performances.

A measure of the variability, or spread, of the data is known as the standard deviation (Thomas and Nelson, 2004) and is calculated using Equation 2.1:

$$SD = \frac{\sqrt{\sum (x_m - x_i)^2}}{(n - 1)} \quad [\text{Equation 2.1}]$$

Where SD is the standard deviation, x_m is the mean of all values of x , n is the number of data points and x_i is the i th value of x . The square of the standard deviation is known as variance (Thomas and Nelson, 2004). Therefore, variance and standard deviation are calculated in a very similar manner. However, one important practical difference is that the units for variance are 'squared' (James,

2004). Hence, standard deviation often permits an easier interpretation of variability as its units are the same as the original data and the mean value.

However, Crussemeyer (1998) suggested that the standard deviation alone is sometimes not adequate and it must be reported in relation to the mean of the data. For example, sometimes, a direct non-adjusted comparison of the variability might not allow the researcher to ascertain whether differences or similarities in variability were the result of changes in the inherent variability in movement or due to changes in the magnitude of the mean. Therefore, in some instances, the coefficient of variation may be a better measure of variability as it is the quotient of these statistics (Newell and Corcos, 1993). The coefficient of variation is calculated using the following equation:

$$CV = \left(\frac{SD}{\bar{x}} \right) \times 100 \quad [\text{Equation 2.2}]$$

Where CV is the coefficient of variation, SD is the standard deviation and \bar{x} is the mean of the data set. Standard deviation gives a measure of absolute variability, whereas the coefficient of variation gives a measure of relative variability and the choice of technique should reflect this depending on the focus of any investigation. Further, Mullineaux (2000) suggested that normalising data to the mean can be useful when the means are similar in size, otherwise it can be misleading and should not be used. Mullineaux (2000) also reported some alternatives to standard deviation and the coefficient of variation in quantifying variability. These included:

root mean square difference, 95% confidence intervals and percentage root mean square difference.

More recently, a further technique to quantify variability, known as the spanning set, was presented by Kurz and Stergiou (2003). This technique is similar to the standard deviation as the time histories of the magnitude of the standard deviation are used in the calculation. What follows is a brief explanation of the steps to calculate the spanning set - see Kurz and Stergiou (2003) or Kurz and Stergiou (2004) for a more detailed description. First, the mean ensemble curve in addition to the mean ensemble plus one standard deviation and the mean ensemble minus one standard deviation are calculated for a sample of trials. Second, polynomial regression equations of an appropriate order are then fitted to the mean plus one standard deviation and mean minus one standard deviation curves. The coefficients of the resulting polynomial equations are then used to describe a vector space that defines the spanning set. The magnitude of the spanning set – the measure of variability – is then calculated as the norm of the difference between the two vectors of the respective spanning sets – the upper and lower standard deviation. The larger the norm of the difference between the two vectors in the spanning set, the greater the span between the two standard deviation curves (Kurz and Stergiou, 2003). A greater span between the two standard deviation curves indicates more variability (Kurz and Stergiou, 2003). Kurz and Stergiou (2003) investigated the effects of barefoot and shod running on kinematic variability using the coefficient of variation, standard deviation and the magnitude of the spanning set. Despite there being qualitatively higher variability in the

barefoot condition that could be seen in the graphs presented by Kurz and Stergiou (2003), only the spanning set revealed a significant difference between conditions. This led Kurz and Stergiou (2003) to conclude that the magnitude of the spanning set was more sensitive to detecting variability than the coefficient of variation and standard deviation. They also suggested that changes in variability may have gone undetected in previous investigations that used either of these measures owing to their lack of sensitivity. However, it should be noted that Kurz and Stergiou (2003) used a measure of standard deviation that was averaged across the entire stance period. This has previously been suggested to decrease the sensitivity of the standard deviation measure as the changing functional demand over the stance phase dictates that magnitudes of variability might change over its duration (Heiderscheit *et al.*, 2002). Therefore, periods of high and low variability over different periods of the stance phase might be nullified and mask any possible differences between conditions. Therefore, if Kurz and Stergiou (2003) had analysed variability over different periods of the stance phase, as advocated by Heiderscheit *et al.* (2002), differences in variability might have been observed between footwear conditions using the standard deviation measure.

In summary, different measures are available for quantifying the magnitude of variability in data sets. Details of the calculation of many of these techniques have been presented in this section. Selection of an appropriate technique should be dictated by the nature of the research question. Finally, inspection of the literature suggests that the standard deviation is most often used to quantify the magnitude

of variability during locomotion. All of the techniques used to quantify coordination variability in this thesis are based on the standard deviation.

2.6.2 Techniques for quantifying coordination and coordination variability

The literature reviewed in sections 2.3, 2.4 and 2.5 highlighted recent advances in the understanding of variability in the movement system. This thesis is focused on the study of movement variability in the context of the variability-overuse injury hypothesis (James, 2004 – see section 2.5.1). The variability-overuse injury hypothesis might be especially relevant in repetitive activities, such as running. A further recent development in the area of the biomechanics of running injuries is the notion that the coupling between body segments, or the patterns of coordination between joints, gives an important insight into running injury mechanisms, rather than solely the actions of each of the joints in isolation (e.g. Bates *et al.*, 1979; McClay and Manal, 1997; Hamill *et al.*, 1999; Stergiou *et al.*, 2001a,b; DeLeo *et al.*, 2004). For example, a proposed mechanism of injury in running is an asynchrony, or timing discrepancy, between subtalar and knee joint actions throughout the running stride (Bates *et al.*, 1978). Both subtalar pronation and knee flexion promote internal rotation of the tibia. Conversely, subtalar supination and knee extension promote external rotation of the tibia. The transitions of knee joint flexion to extension and subtalar pronation to supination occur at approximately the same time in the stance phase (Stergiou *et al.*, 1999). However, a possible antagonistic relationship might be present between the two joints through the tibia, if the transitions between subtalar pronation to supination and knee flexion to extension are asynchronous. If pronation continues after knee

extension has begun, the tibia will still be internally rotated at the distal end, but an external rotating moment will act at the proximal end of the tibia, creating tibial torsional stresses and abnormal loads on the knee joint (Stergiou, *et al.*, 1999).

Many studies examining the relationship between subtalar pronation-supination and knee flexion-extension have used discrete points in the movement cycle, such as time to maximum pronation angle and time to maximum knee flexion angle, in an attempt to determine whether asynchrony exists between the two joint actions (Hamill *et al.*, 1992; von Woensel and Cavanagh, 1992; McClay and Manal, 1997; Stergiou and Bates, 1997; Stergiou *et al.*, 1999). However, a limitation of these studies is that both knee flexion/extension and subtalar pronation/supination are continuous phenomena; which is not accounted for by this type of analysis. A major criticism of such discrete point analysis is that, in some instances, a value for maximum pronation is unclear (De Wit and De Clercq, 2000). Although the knee joint exhibits a fast transition from knee flexion to extension (De Wit and De Clercq, 2000), and a unimodal curve always exists for knee flexion-extension (Stergiou *et al.*, 1999), this is not always the case for subtalar pronation-supination. In some cases bimodal curves are present for subtalar pronation/supination (Stergiou and Bates, 1997; Stergiou *et al.*, 1999; De Wit and De Clercq, 2000), meaning two values for the timing of maximum pronation during any one stance phase. This presence of two maximum values for pronation makes the analysis of coordination between the subtalar and knee joints using discrete point analysis extremely difficult and it may give a poor representation of the true relationship. Von Woensel and Cavanagh (1992) proposed that rearfoot velocity should be used to determine

a threshold value, dividing the rearfoot curve into two phases. This technique was adopted by De Wit and De Clercq (2000) in a comparison of barefoot and shod running. However, this procedure still does not fully acknowledge the continuous nature of the movements at the subtalar and knee joints, suggesting a limitation to the area of work and a possible avenue for future study. This line of research suggests that the coordination of lower extremity joint actions is of high importance.

Recently, many techniques have been presented in the literature that address the issue of providing a measure of the coordination between body segments, whether they be discrete measures, which have the limitation of only providing one measure of coordination per cycle as outlined above, or continuous measures. Examples of discrete measures include conjugant cross correlations (e.g. Amblard *et al.*, 1994) and discrete relative phase (e.g. Diedrich and Warren, 1995; van Emmerik and Wagenaar, 1996; LaFiandra *et al.*, 2003). In addition to these discrete methods, continuous measures including continuous relative phase (e.g. Hamill *et al.*, 1999; Heiderscheit *et al.*, 1999; Post *et al.*, 2000; van Uden *et al.*, 2003), relative motion angles (Heiderscheit *et al.*, 2002; Ferber *et al.*, 2005; Pollard *et al.*, 2005) and Vector Coding (Tepavac and Field-Fote, 2001; Field-Fote and Tepavac, 2002), have been presented. If the coordination between body segments is quantified over several cycles the variability in coordination can be determined. This variability in coordination would appear extremely important in light of the proposed running injury mechanism outlined previously in this sub-section and the variability-overuse injury hypothesis (James, 2004) - see section 2.5.1. However, each of the techniques to measure coordination and, arguably more importantly,

coordination variability have benefits and limitations that require consideration. For example, continuous relative phase gives a comprehensive indication of coordination and coordination variability at each data point in the cycle but should only be implemented on data that are sinusoidal and of a one-to-one frequency ratio (Peters *et al.*, 2003). In the following sections, a review of the techniques available for quantifying coordination and coordination variability is provided. Details of their calculation are provided in addition to a discussion of their relative merits and limitations.

2.6.2.1 Relative phase

Both the discrete relative phase (DRP) and continuous relative phase (CRP) techniques are based on the assumptions that the two oscillating segments under scrutiny are of a one-to-one frequency ratio and that they exhibit a sinusoidal time history (Hamill *et al.*, 2000). Clearly, segmental motions in gait and sports techniques do not always meet these assumptions and problems can arise when using relative phase to quantify the coordination between body segments in such activities. Care should, therefore, be taken when interpreting relative phase data in relation to inter-segment coordination, especially if these assumptions are violated. However, alternative techniques are available, such as the Hilbert transform (Rosenblum and Kurths, 1998) and relative Fourier phase (e.g. Lamothe *et al.*, 2002 - see section 2.6.2.1.2), to transform the data and to ensure that they satisfy the necessary assumptions.

2.6.2.1.1 Discrete relative phase

DRP is a point estimate approach that illustrates the latency of an event in a segment's motion with respect to another segment's motion (Kelso, 1995). When the relative timing of events between two separate segments is important, as in the investigation of the relationship between subtalar inversion-eversion and knee flexion/extension (e.g. Lafourte *et al.*, 1994; McClay and Manal, 1997; Stergiou and Bates, 1997; Nawoczenski *et al.*, 1998; Stergiou *et al.*, 1999), DRP may be an important variable to consider. DRP has already been used, for example, to measure the phase difference between thoracic and pelvic rotations during treadmill walking (e.g. Lamoth *et al.*, 2002), determine the effect of load carriage on trunk coordination (La Fiandra *et al.*, 2003) and examine the walk-run transition in human bipedal locomotion (Diedrich and Warren, 1995). DRP can be calculated using the following equation:

$$\Phi = \frac{t_{\max\phi 1(j)} - t_{\max\phi 2(j)}}{t_{\max\phi 1(j+1)} - t_{\max\phi 1(j)}} \times 360^\circ \quad [\text{Equation 2.3}]$$

Where t is time, $\max\phi 1$ is the maximum rotation of segment 1, $\max\phi 2$ is the maximum rotation of segment 2 and ϕ is the phase difference during the cycle j .

Hamill *et al.* (2000) suggested that, when considering the example of knee flexion/extension and subtalar inversion/eversion above, in which it might be appropriate to study the stance phase in isolation, foot strike could be used as the initial point. Furthermore, these authors suggested the length of the stance phase

could represent the cycle time. Therefore, in this example, $\max\varphi 1$ would be maximum subtalar eversion, $\max\varphi 2$ would be maximum knee flexion and the time (the denominator) would simply be stance time.

Since DRP is calculated in the range $0^\circ \leq \varphi \leq 360^\circ$ and there is redundancy in angles (e.g. 0° and 360° are equivalent), it is classed as a circular variable. Therefore, to avoid phase wrapping (Burgess-Limerick *et al.*, 1991; Lamothe *et al.*, 2002), average DRP over several cycles, along with the variability of coordination, should be calculated using circular statistics - see Appendix C for an overview of circular statistics.

DRP has the advantage that no further manipulation of the data is required, other than that which would normally be carried out in the calculation of joint angles (Hamill *et al.*, 2000). However, problems might arise if the data do not meet the assumptions outlined in the previous section of sinusoidal time histories and one-to-one frequency ratio. There would certainly be a problem if definite peak values could not be ascertained or respective peak values changed from cycle-to-cycle. In other words, angular displacement patterns that contain multiple maxima and minima might present a problem for the calculation of DRP if the magnitudes of the peaks change from cycle-to-cycle, making the selection of a peak in each cycle that corresponds correctly to other cycles difficult. This phenomenon would be most detrimental to the calculation of coordination variability, as erroneously high variability in coordination might be calculated simply because the magnitudes of the separate peaks are changing between each cycle. This is not a problem in the

examples from the literature on coordination cited earlier in this section. It might become a problem if DRP was used to analyse other sports techniques or more variable joint motions with more peaks and troughs in the angular displacement time series. This might introduce ambiguity into the definition of a peak. Another obvious disadvantage is that DRP provides only one measurement per movement cycle.

2.6.2.1.2 Continuous relative phase

CRP provides an indication of the phase relation between two oscillating segments at each sampled data point throughout the movement cycle. Both CRP and CRP variability have been used in studies examining running injuries (Hamill *et al.*, 1999; Heiderscheit *et al.*, 1999), the coordination of finger oscillations (Kelso, 1981, 1984), the coordination of thorax and pelvis rotations (van Emmerik and Wagenaar, 1996; Lamothe *et al.*, 2002; LaFiandra *et al.*, 2003), coordination patterns when walking and running (Li *et al.*, 1999), the analysis of one-legged hopping (van Uden *et al.*, 2003) and intra-limb coordination following obstacle clearance (Stergiou *et al.*, 2001a; Stergiou *et al.*, 2001b). The CRP between two oscillating segments at any given instant is defined as the difference between the respective phase angles of each segment. Hamill *et al.* (2000) recently highlighted that, before CRP can be calculated, the segment displacements and velocities need to be normalised to eliminate the effects of amplitude differences in the range of motion of each segment - this issue will be discussed later in this section. Also, the displacement and velocity data should be interpolated to a fixed number of data points to allow the calculation of ensemble averages and variability. These data are

then used to construct a phase plane portrait - normalised angular velocity on the ordinate axis against normalised angular displacement on the abscissa axis - for each segment; a typical phase plane portrait is illustrated in Figure 2.6. Phase plane portraits provide a graphical representation of all possible states of the segment (Clark, 1995) as the behaviour of a dynamical system may be captured by a variable and its first derivative with respect to time (Rosen, 1970).

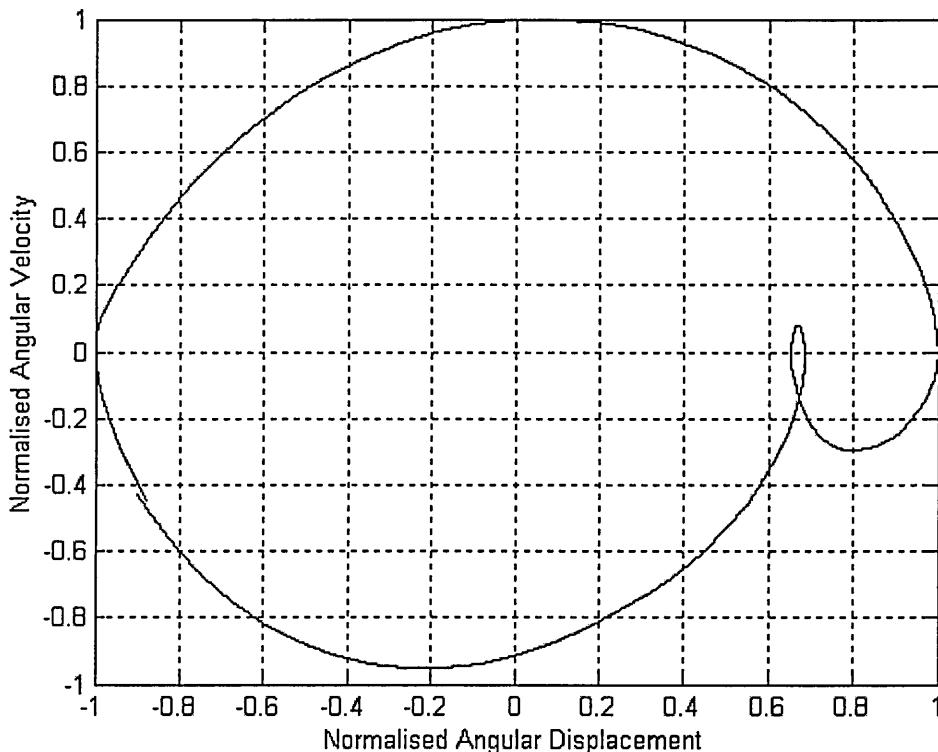


Figure 2.6: An example of a phase-plane portrait (Hip Flexion/Extension)

The Cartesian coordinates of each individual data point on the phase plane are then converted to Polar coordinates. The phase angle component is given by the following equation:

$$\varphi(t) = \tan^{-1} \left(\frac{\dot{\theta}(t)}{\theta(t)} \right) \quad [\text{Equation 2.4}]$$

Where $\dot{\theta}$ is normalised angular velocity, θ is the normalised angular displacement and φ is the phase angle at time t .

The CRP between the two segments can then be calculated as the difference between the segment phase angles and is usually achieved by subtracting the distal segment phase angle from that of the proximal segment. For example, the CRP between the shank and the foot during running would be calculated using the following equation:

$$\phi(t) = \varphi_{shank}(t) - \varphi_{foot}(t) \quad [\text{Equation 2.5}]$$

Where φ_{shank} is the shank segment phase angle, φ_{foot} is the foot segment phase angle and ϕ is the CRP at time t .

With regards to the calculation of the component phase angle, it is important to note that the output of $\tan^{-1} \left(\frac{y}{x} \right)$ takes on values between -90° and $+90^\circ$. Therefore, the output data needs manipulating to ensure that the component phase angles are calculated within a suitable range. A discrepancy exists in the literature between studies using CRP in terms of the definition of the range of component phase angles used. In the area of motor control (e.g. see Scholz, 1990 for an

overview), a phase angle range of $0^\circ \leq \varphi \leq 360^\circ$ has typically been used but the recent application of this technique to the area of biomechanics has brought with it a new $0^\circ \leq \varphi \leq 180^\circ$ definition. Hamill *et al.* (2000) suggested that this new definition was necessary because there is redundancy in angles in the original definition (i.e. 0° and 360° mean the same thing). Presumably, this new definition is preferred because it avoids discontinuities in the component phase angles, which can be problematic if conventional linear statistical analyses are used.

Changes in the component phase angle definitions, however, affect the values of the computed CRP. Wheat *et al.*, (2003) investigated the effect of different component phase angle definitions on CRP. In this study, test data were created for two ‘segments’ using sine and cosine functions. Data from a sine function served as the angular displacement of one segment. Similarly, as cosine is the first derivative of sine, data from a cosine function served as the angular velocity of the same segment. The CRP of two ‘segments’ was manipulated by simply adding a given amount to the angles inputted into the sine and cosine functions for the second segment. Three conditions were tested in which the segments were 180° , 90° and 45° out-of-phase with each other. In the 180° out-of-phase conditions, constant CRP values instantaneously switching between 180° and -180° were apparent for the $0^\circ \leq \varphi \leq 360^\circ$ range, but this was not evident when the phase angles were defined in a range of $0^\circ \leq \varphi \leq 180^\circ$ (Figure 2.7). Instead, a gradual shift between 180° and -180° was seen. Similar results were obtained for the 90° and 45° out-of-phase conditions.

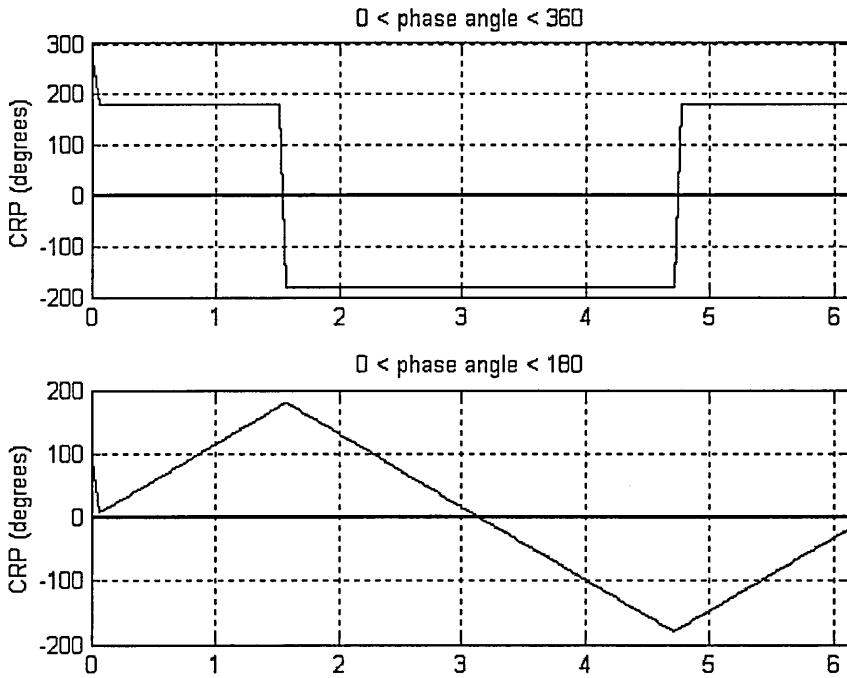


Figure 2.7: Continuous relative phase calculated using two different component phase angle definitions – top pane, $0^\circ \leq \theta \leq 360^\circ$, bottom pane, $0^\circ \leq \theta \leq 180^\circ$

It would appear that, if information about the coordination between the segments is required, the $0^\circ \leq \varphi \leq 360^\circ$ range is most suitable because the $0^\circ \leq \varphi \leq 180^\circ$ range does not yield correct results. In other words, if Kelso had used the $0^\circ \leq \varphi \leq 180^\circ$ definition in his work monitoring non-linear phase transitions in finger movement, he would never have been able to record the anti-phase ($\pm 180^\circ$ out-of-phase) relationship. Recently, other range definitions have also been used. These include: $180^\circ \leq \varphi \leq 180^\circ$ (e.g. Lamothe *et al.*, 2002), which is effectively the same as $0^\circ \leq \varphi \leq 360^\circ$, and $0^\circ \leq \varphi \leq 90^\circ$ (e.g. Kurz and Stergiou, 2002), which has the same problems as the $0^\circ \leq \varphi \leq 180^\circ$ definition. As already suggested above, presumably the $0^\circ \leq \varphi \leq 180^\circ$ (and $0^\circ \leq \varphi \leq 90^\circ$) definition has been used to avoid the discontinuities in the component phase angles and, subsequently, the CRP data during the

analysis of most movements. These discontinuities could introduce anomalies that might cause erroneously high variability to be calculated using conventional linear statistical techniques. However, this problem is easily solved if the recommended circular statistical techniques are used (see Burgess-Limerick *et al.*, 1991; Lamoth *et al.*, 2002). If researchers wish to present CRP data graphically as a function of time but do not want to present data containing discontinuities, the CRP data should be manipulated to a suitable range after calculation, similarly to Lamoth *et al.* (2002). It is certainly clear that authors should state their choice of phase angle definition for the reader to make an informed and correct interpretation of the results (Wheat *et al.*, 2003).

Interestingly, the correct choice of phase angle range definition would not be an issue if the data for each component oscillator were transformed onto the same phase plane. The continuous relative phase between the two oscillators could then be calculated using the Law of Cosines as recommended by Williams (1997). The result would be similar to that which would be obtained by calculating component phase angles with the $0^\circ \leq \varphi \leq 360^\circ$ phase angle definition but lead-lag information would not be available. Further work is warranted to determine the effectiveness of this method for calculating continuous relative phase. However, whenever continuous relative phase is calculated in this thesis, it is computed using component phase angles in the range $0^\circ \leq \varphi \leq 360^\circ$.

As mentioned previously, the need to normalise the data in a phase-plane portrait has been identified (e.g. Hamill *et al.*, 2000). The normalisation procedure adjusts

for amplitude differences in the ranges of motion and centres the phase plane portraits about the origin (Hamill *et al.*, 2000; Lamoth *et al.*, 2002). Hamill *et al.* (2000) presented data highlighting the effect of different normalisation techniques on CRP and CRP variability. Differences were seen, in both CRP and CRP variability, between the four techniques discussed. The authors suggested that, ultimately, the choice of normalisation procedure is likely to be dependent upon specific aspects of the research question. Conversely, Kurz and Stergiou (2002) suggested that no normalisation is required when calculating component phase angles. They investigated the effect of three different normalisation conditions - two normalisation techniques and no normalisation condition - and calculating phase angles within two different ranges, which appear to be $0^\circ \leq \varphi \leq 180^\circ$ and $0^\circ \leq \varphi \leq 90^\circ$. They suggested that certain combinations of parameters produced 'errors' in the calculated CRP. They proposed that normalisation of the data on a phase-plane is not required because of the properties of the arc tangent function used in the calculation of the component phase angles. They suggested that CRP is not affected by differences in amplitude between segments since the arc tangent function is based on a ratio (velocity:displacement) and the differences in amplitude are removed during phase angle calculation (Kurz and Stergiou, 2002). However, Peters *et al.*, (2003) presented data, using distorted sine waves with a known phase relationship, which suggested that CRP, calculated without normalising the data on a phase plane, produced erroneous results. Even when two sine waves with a frequency other than $0.5/\pi$ were tested, questionable CRP values were obtained without normalisation. Another reason for normalising the data on a phase-plane is to centre the trajectory around the origin (Hamill *et al.*,

2000). This increases the extent to which the motions are sinusoidal - an assumption of CRP. Therefore, normalisation appears to be useful. It should be noted that, in their analysis, Kurz and Stergiou (2002) used phase angle definitions of $0^\circ \leq \varphi \leq 180^\circ$ and $0^\circ \leq \varphi \leq 90^\circ$, which have been shown to produce questionable CRP results (Wheat *et al.*, 2003). Additionally, not normalising the data might affect CRP variability. As Heiderscheit (2000) indicated, when highlighting the reasons why non-sinusoidal data affects CRP variability, the proximity of a data point to the origin of the phase plane can directly influence the calculated coordination variability. Two data points of fixed distance will exhibit a greater phase angle difference the closer they are to the origin of the phase plane (Figure 2.8). This suggests that, when the data on phase planes are not normalised, erroneously high variability will be observed in segments with small amplitudes.

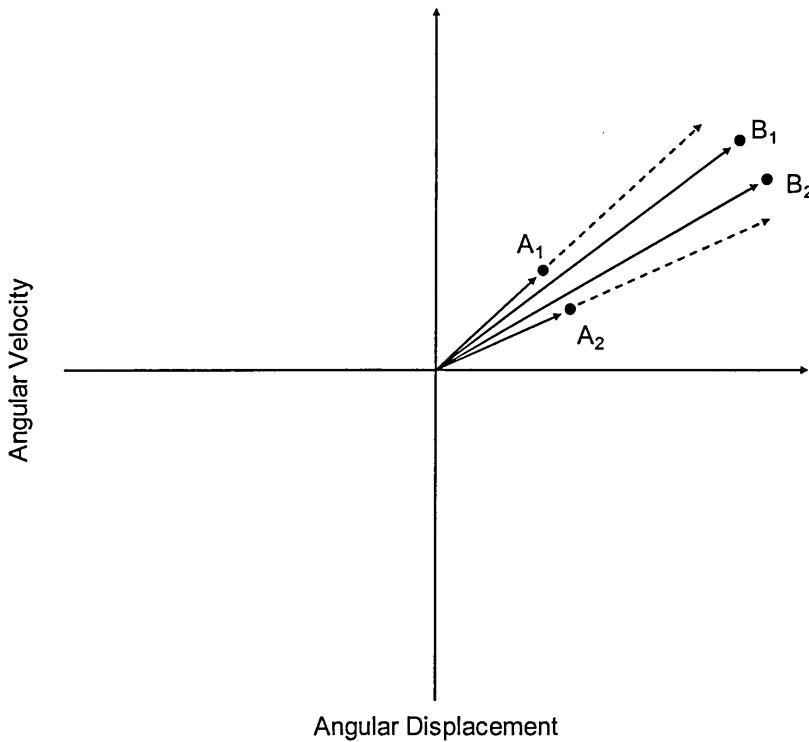


Figure 2.8: The influence of the proximity of data points to the origin. Data points A₁ and A₂ will have a greater phase angle difference than data points B₁ and B₂, even though each pair are the same distance apart - adapted from Heiderscheit (2000).

In summary of the discussion of the normalisation of raw data for continuous relative phase analysis, it is clear that the issue has received much attention in the literature. Although some authors have argued that normalisation is not required (Kurz and Stergiou, 2002), others have presented convincing arguments to suggest that it is needed (Peters et al., 2003). Certainly, it would seem that some sort of procedure has to be applied to the raw data in order to centre the phase plane about the origin. With most frequently used angle conventions, not centring the data about the origin confines the phase plane to two quadrants and the phase angle to a range even smaller than $0^\circ \leq \varphi \leq 180^\circ$. Because of this, the data are

normalised using the method outlined by Hamill et al. (1999) whenever continuous relative phase is used in this thesis. Finally, it is apparent that more work is required to determine the effect of different normalisation techniques on calculated CRP and CRP variability, using both circular statistics and suitable definitions of component phase angles.

The use of CRP has many advantages in quantifying both coordination and coordination variability. As angular velocity is included in the calculation of the component phase angles, CRP contains both spatial and temporal information (Hamill et al., 1999; Hamill et al., 2000), which gives a higher dimensional and more detailed analysis of the behavior (Hamill et al., 1999). Additionally, the inclusion of velocity in the calculation might make CRP a more sensitive measure of coordination variability (Wheat et al., 2002) than other techniques. However, the inclusion of a higher derivative (angular velocity) into the calculation will propagate any errors - whether they be due to skin marker movement, the recording system or any other source - in the displacement data. Therefore, it might also introduce a greater error into the CRP data, which could be interpreted as increased coordination variability. A further advantage reported by several authors (Burgess-Limerick et al., 1991; Stergiou et al., 2001a), is that the use of CRP is advantageous since there is evidence to suggest that receptors exist within muscles and tendons that control both the position and the velocity of the respective body segment (McCloskey, 1978). Finally, another advantage of CRP is that it provides a continuous measure of coordination and coordination variability throughout the entire movement. CRP and CRP variability can therefore be

calculated for different phases of the gait cycle (e.g. Hamill *et al.*, 1999; Heiderscheit *et al.*, 1999; Stergiou *et al.*, 2001a; Stergiou *et al.*, 2001b). This is particularly useful, considering the change in the functional demands on the lower extremity throughout the stride cycle (Heiderscheit *et al.*, 2002).

There are also limitations to consider when calculating CRP. A fundamental limitation is the assumption that the time histories of the joint motions are sinusoidal. Clearly, this is not always the case with joint motion during some activities. However, in some cases, centring the phase-plane trajectory around the origin through the use of a normalisation procedure, helps make it more sinusoidal. As mentioned previously, the issue of whether the data on a phase-plane need normalising is a contentious one and requires further clarification. Also, there are techniques available that effectively transform the data so that it has a more sinusoidal time history. Relative Fourier Phase, for example, is a technique that essentially transforms the data into the frequency domain and discards any frequencies other than the fundamental frequency. When the data are reconstructed in the time domain, the displacement trace is sinusoidal and CRP calculations can be made with confidence. In their study of pelvic-thorax coordination during pathological walking, Lamothe *et al.* (2002) justified this approach because 'movements of the pelvis are affected by forceful contacts between the feet and the support surface ... which induce oscillations affecting the phase progression of the pelvis rotation' (p.112). In other words, they argued that oscillations, other than those at the fundamental frequency, are not relevant to the coordination of the pelvis and thorax. However, when applying this technique to

other coordinative structures, care should be taken not to disregard potentially relevant information. A similar technique originally presented by Rosenblum and Kurths (1998) involves the calculation of continuous relative phase using the Hilbert transform. Briefly, the Hilbert transform shifts the frequency components of the original angular displacement data by $\pi/2$ radians. Data from the original and phase shifted signals of two joints are then used to calculate the continuous relative phase using the following equation:

$$CRP = \tan^{-1} \left(\frac{\tilde{s}_1(t)s_2(t) - s_1(t)\tilde{s}_2(t)}{s_1(t)s_2(t) - \tilde{s}_1(t)\tilde{s}_2(t)} \right) \quad [\text{Equation 2.6}]$$

Where $\tilde{s}_1(t)$ and $\tilde{s}_2(t)$ are the imaginary components of the Hilbert transform of the two joint angles and $s_1(t)$ and $s_2(t)$ are the original joint angles.

Another potential problem with CRP, which has been raised previously by other authors (Tepavac and Field-Fote, 2001; Mullineaux and Wheat, 2002), is that it is hard to relate to conceptually. This is mainly a problem for practitioners trying to interpret the type and nature of relationship between joints and body segments. However, this issue is resolved to a certain extent if the variability in coordination is the primary focus of an investigation.

2.6.2.2 Vector coding techniques

Several vector coding techniques have been introduced to quantify the data in relative motion plots and the variability in angle-angle trajectories (e.g. Whiting and Zernicke, 1982; Sparrow *et al.*, 1987; Tepavac and Field-Fote, 2001; Heiderscheit *et al.*, 2002). These techniques stem from the early work of Freeman (1961) who devised a chain-encoding technique to quantify an angle-angle curve. The procedure involves using a superimposed grid to transform the angle-angle trajectory into digital elements (see Figure 2.9). Subsequently, a chain of digits based on the direction of the line segment formed by the frame-to-frame interval between two consecutive data points is created, which approximates the shape of the original analog curve. Pairs of these integer chains from two different cycles are then cross-correlated to obtain what was termed a recognition coefficient, and is the peak value of the cross-correlation function. This technique has been used in studies of locomotion (Hershler and Milner, 1980; Whiting and Zernicke, 1982). However, as Tepavac and Field-Fote (2001) suggested, a flaw with this technique is that it converts ratio scale data to the nominal scale, risking the loss of important information and limiting the types of statistical analyses that can be applied. Also, a motivation behind Freeman's (1961) technique, that was one of computer efficiency, is no longer applicable owing to advances in modern computer hardware and software technology such as increased processor speeds and more sophisticated statistical analysis packages (Sparrow *et al.*, 1987).

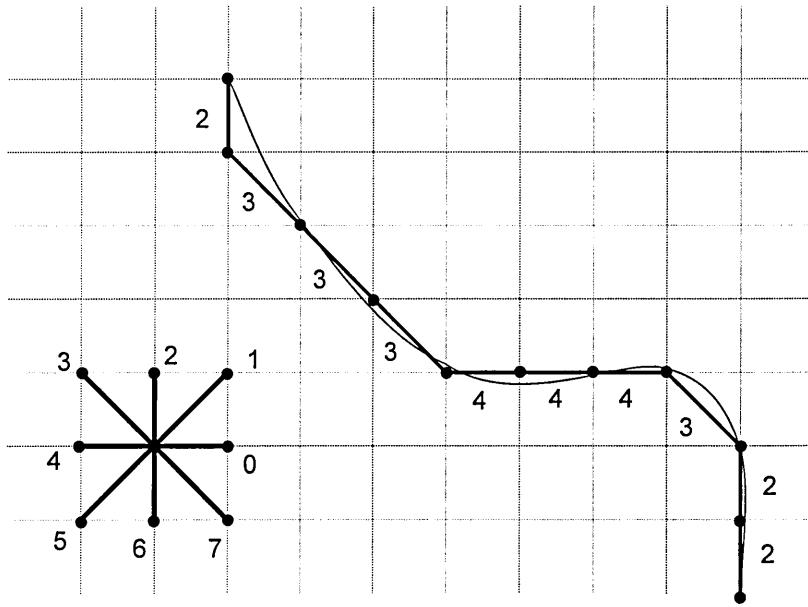


Figure 2.9: Freeman's (1961) Chain Encoding technique

Another problem with this technique is that it requires the data points to be equally spaced. Sparrow *et al.* (1987) recognised that this is not always the case in the human movement sciences and proposed a revised cross-correlation formula that takes into account the length of the frame-to-frame interval. However, Tepavac and Field-Fote (2001) identified two problems with Sparrow *et al.*'s (1987) technique; (i) the two trajectories of interest must consist of an equal number of data points and (ii), it can only compare two trajectories at a time (multiple cycles must be compared in a pairwise manner).

Tepavac and Field-Fote (2001) proposed a revised technique to address the above limitations. The authors presented a ratio-scale, vector-based coding scheme for the quantification and analysis of relative motion data. Field-Fote and Tepavac (2002) suggested that the technique provides an alternative to relative phase analysis. The technique was designed to assist in interpretation of the data

because they believed that practitioners are more likely to think of movement in terms of joint angles and not phase values. In a similar way to Sparrow *et al.* (1987), Tepavac and Field-Fote (2001) measured the direction and magnitude of the frame-to-frame intervals on the angle-angle trajectories and calculated the magnitude and direction of the vector connecting the two points of the relative motion plot. However, as opposed to pair-wise comparisons using cross-correlations, the standard deviation of the direction of the vector was calculated at each frame-to-frame interval ($a_{i,i+1}$), using circular statistics. This meant that the variability in the angular component of the angle-angle trajectory could be calculated at each individual frame-to-frame interval for multiple cycles. The variability in the magnitude of the frame-to-frame vector was also calculated ($m_{i,i+1}$). Finally, a measure of the overall variability of the angle-angle trajectories was proposed, which was simply the product of $a_{i,i+1}$ and $m_{i,i+1}$ ($r_{i,i+1}$) and was called the coefficient of correspondence. Field-Fote and Tepavac (2002) contended that this revised technique was mathematically equivalent to the technique of Sparrow *et al.* (1987). These three separate measurements mean that, when using the Vector Coding algorithm (Tepavac and Field-Fote, 2001), it is possible to analyse separately a relative motion plot based on its shape - angles, its magnitude - the length of the frame-to-frame intervals, or the frame-to-frame vector deviation - a combination of shape and magnitude. Field-Fote and Tepavac (2002) used their Vector Coding technique to assess the consistency or variability of the pattern of the hip-knee coupling over multiple cycles of treadmill walking, in patients with incomplete spinal cord injury before and after a period of training. They used only the measurement of shape (a) to assess variability, presumably because they

thought magnitude changes i.e. changes in the range of motion at each joint from cycle to cycle, would not significantly affect the cycle-to-cycle variability relative to the changes in the shape of the angle-angle diagram in this population.

Both Hamill *et al.* (2000) and Heiderscheit *et al.* (2002) proposed subtle alternatives to the technique of Tepavac and Field-Fote (2001), which were presented as modifications of the Sparrow *et al.* (1987) method. In both techniques a 'coupling angle' was defined as the orientation of the vector between two adjacent points on the angle-angle plot relative to the right horizontal (see Figure 2.10). This is similar to the measurement of shape provided by Tepavac and Field-Fote (2001).

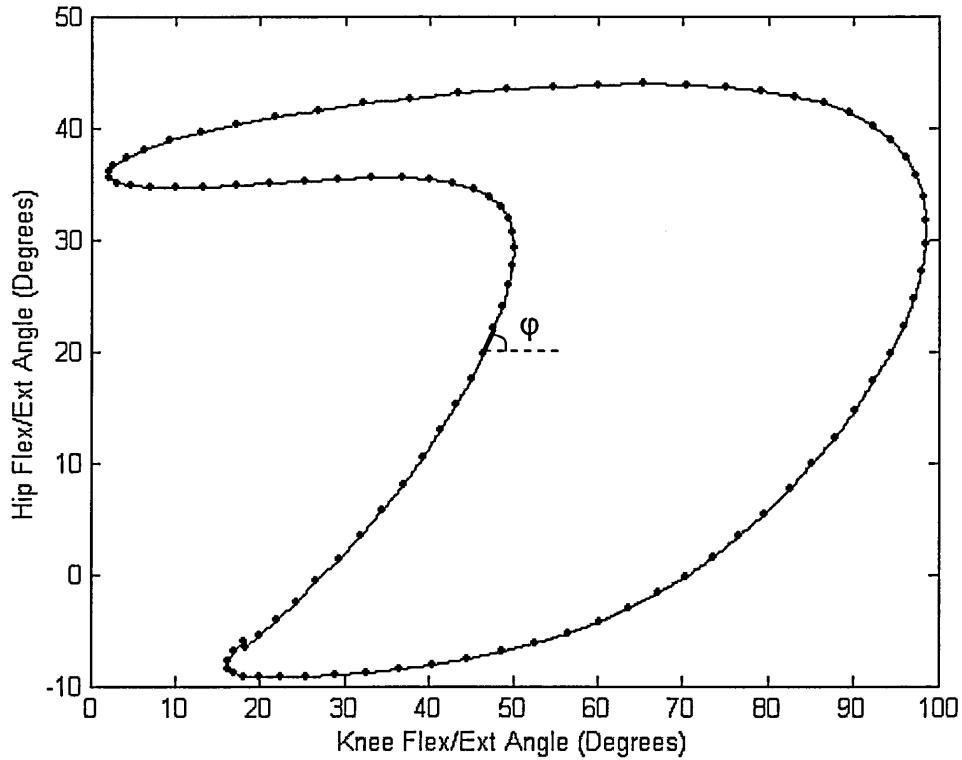


Figure 2.10: An illustration of the coupling angle calculation from an angle-angle plot

The variability in the coupling angle over multiple cycles was then calculated at each frame-to-frame interval using circular statistical techniques. No measurement was made, however, of the magnitude of the frame-to-frame vectors, which may represent a limitation of this technique. This vector coding method was also used by Heiderscheit *et al.* (2002) to compare joint coordination variability during treadmill running in participants with and without patellofemoral pain.

The advantages of this collection of techniques include no requirement for normalisation (Hamill *et al.*, 2000) and the maintenance of true spatial information in the data. This seems especially advantageous when using the technique of

Tepavac and Field-Fote (2001), which incorporates a measure of both the shape and magnitude of the angle-angle trajectories. Additionally, Field-Fote and Tepavac (2002) contended that vector coding techniques are more suitable than other methods, such as relative phase analysis because 'clinicians ... are more likely to think of movement in terms of joint angles as opposed to phase values' (p.710). The notion that vector coding techniques are easier to relate to than relative phase seems reasonable. However, the output data from these techniques are not joint angles but the direction and magnitude of frame-to-frame vectors, which may still be hard to understand conceptually.

There are also disadvantages of vector coding techniques. A potential limitation of this group of techniques is that they include only spatial information with no regard to temporal information (Hamill *et al.*, 2000) – although the technique of Tepavac and Field-Fote (2001) addresses this limitation to an extent with its measure of the length of the frame-to-frame vectors. Also, Heiderscheit *et al.* (2002) suggested a potential problem with vector coding techniques around times at which joint motions change direction. Clark and Phillips (1993) hypothesised that these periods of movement reversal are critical in the study of movement coordination as are the apparent increases in coordination variability (Ghez and Sainberg, 1995; Heiderscheit *et al.*, 2002). However, as Heiderscheit *et al.* (2002) noted, during this phase of the gait cycle, there is minimal joint displacement and, therefore, a clustering of data points on the relative motion diagram. Therefore, the apparent increase in coordination variability - often seen during these periods - might simply

be an artefact of the greater proximity of consecutive data points and the inherently greater sensitivity to slight changes in displacement.

2.6.2.3 Other techniques

Techniques other than relative phase and vector coding techniques have been used to quantify the coordination and coordination variability between two body segments or joints. Two techniques that have received varying amounts of coverage in the literature are cross-correlations (Amblard *et al.*, 1994) and normalised root mean squared difference (NoRMS: Sidaway *et al.*, 1995).

Cross-correlations have been used, for example, to measure changes in coordination of an elite javelin thrower's technique over a five year period (Morriess, 1998), assess changes in coordination during the learning of a ski-simulator task (Vereijken *et al.*, 1992; Whiting and Vereijken, 1993), monitor the acquisition of coordination during a handwriting task (Newell and van Emmerik, 1989), examine the effects of practice on coordination during dart throwing (McDonald *et al.*, 1989) and analyse the differences in intra-limb coordination between expert and novice volleyball players performing a serve (Temprado *et al.*, 1997). Cross-correlations are based on the assumption that a linear relationship exists between segments or joints. However, they do not assume that these segments or joints move in synchrony throughout the movement (Mullineaux *et al.*, 2001). By introducing 'time lags' to the data - shifting one segment's data forward or back in relation to another segment's data by a given number of data points - it is possible to find high correlations between two segments between which there is a constant time lag

(Mullineaux *et al.*, 2001). Amblard *et al.* (1994) suggested that cross-correlations are particularly relevant for the analysis of human movement as there are often time lags between coordinated segments.

Cross-correlations appear, in some respects, to be similar to DRP as the lag time value from cross-correlation, if expressed relative to the oscillation period, gives an indication of the phase relationship between the two segments (Temprado *et al.*, 1997). However, determining whether or not the relationship is a phase lag or a phase lead can be problematic (Amblard *et al.*, 1994). Advantages of cross-correlations include that, if the data are linear, no normalisation procedure is required and, similar to DRP, no further manipulations are required other than would normally be carried out in the calculation of joint angles. However, if the data are non-linear, a transformation procedure, such as a log-log transformation, is required to linearize the data (see Snedecor and Cochran, 1989). Nonetheless, if, after the transformation, the cross-correlation coefficient were still small - i.e. there was still no linear relationship between the segments of interest - cross-correlations would be unsuitable. Another disadvantage of the technique is that it provides only one measurement per movement cycle.

Sidaway *et al.* (1995) presented a technique to measure the consistency or variability of several angle-angle trajectories (NoRMS). By measuring the resultant distance between the angle-angle coordinate of a curve and the angle-angle coordinate of the mean curve at each instant, a root-mean-square difference value is calculated at each point in time. These values are then averaged across the

entire trial and subsequently normalised with respect to the number of cycles and the excursion of the mean plot using the equations presented by Sidaway *et al.* (1995) which were reduced by Mullineaux *et al.* (2001) into the following:

$$NoRMS = 100 * \left(\frac{\sum_{j=1}^k \sqrt{\sum_{i=1}^n (\bar{x}_A - x_{Ai})^2 + (\bar{x}_B - x_{Bi})^2}}{R} \right)^{n_j}$$
[Equation 2.7]

where A and B denote the two variables of interest, k is the number of cycles, n is the number of data points, R is the resultant excursion of the mean angle-angle curve over the entire cycle, \bar{x} is the mean position of a given variable at the i th data point and x is the position of a given variable at the i th data point on the j th cycle.

Sidaway *et al.* (1995) suggested that multiplication by 100 is used to make the resulting scores more manageable. However, the authors highlighted that, because linear statistical techniques are used on directional data, joint angles need to be greater than 0° and less than 360° and that the technique is not valid should joints, in an unusual situation, rotate through 360° . Sidaway *et al.* (1995) also suggested that it is more appropriate to express joint angles in relative terms to avoid neutral joint positions.

The NoRMS technique appears to offer a good measure of the variability in angle-angle traces that takes account of changes in the magnitude and shape of the plots. However, it gives no indication of the coordination between the segments of interest. Furthermore, there are some issues that need to be considered before using the NoRMS technique. First, NoRMS in the form outlined by Sidaway *et al.* (1995) only provides one measure of coordination variability over the entire duration of the movement of interest. This might limit the use of the technique to analyse the variability in coordination during movements in which changes in the functional demands of the task over its duration might alter the magnitude of the variability during different phases - e.g. throughout the stance period of running (Heiderscheit *et al.*, 2002). Moreover, the stage of the calculation during which the average cycle root mean square is divided by the resultant excursion of the mean angle-angle curve (see equation 4) appears to be similar to dividing the standard deviation by the mean value during the calculation of the coefficient of variation. Mullineaux (2000) stated that normalising data to the mean is appropriate if the means of the two sets of measurements are similar in size but it should not be done if the means are dissimilar as the results can be misleading. Therefore, in some instances where resultant excursions of the mean angle-angle curves are different between data sets under investigation, NoRMS might be inappropriate in the form presented by Sidaway *et al.* (1995). This technique has received little attention in the biomechanics and motor control literature.

In summary, the techniques available for the quantification of coordination variability during locomotion are diverse and they each have associated

advantages and disadvantages. Indeed, it seems that differing answers to research questions might be uncovered if different techniques are used to quantify coordination variability. For example, in a preliminary study, Wheat *et al.* (2002) investigated the effects of using both standard deviation in continuous relative phase and the coefficient of correspondence (Tepavac and Field-Fote, 2001) to quantify coordination variability during running. Results indicated that, particularly during the second half of the stance phase, the two techniques gave a contrasting indication of coordination variability.

2.6.3 Errors in the measurement of coordination and coordination variability

Another important consideration for the measurement of coordination and coordination variability is the error associated with the measurement of these variables. The total variability measured in any system – the variability that would be measured experimentally - can be represented by the following equation (adapted from James, 2004):

$$Var_T = Var_h + Var_{er} + Var_{en} \quad [\text{Equation 2.8}]$$

Where Var_T is the total variability, Var_h is the variability inherent in normal human movement made up of a random and deterministic component (*c.f.* Slifkin and Newell, 1998; Newell and Slifkin, 1998; Riley and Turvey, 2002), Var_{er} is the variability due to errors in the measurement and Var_{en} is the variability due to other sources – e.g. changes in the environmental and task constraints. Obviously, it is reasonable to assume that, in studies of coordination variability during locomotion,

Var_h is an important component in which a researcher might be interested and their aim would be to separate Var_h from the other components. In a well controlled research environment Var_{en} will be minimal and relatively easy to isolate. This is because either task or environmental constraints are likely to be manipulated as an independent variable whilst any constraints that are not manipulated should be controlled to minimise the affect of extraneous variables on the results of the study. However, the process of separating the inherent human variability from Var_{er} is far from trivial.

Before the coordination between body segments can be assessed, kinematic measurements must be made of the position and orientation of the segments of interest. Generally, these measurements are made by attaching markers directly onto the skin of the participant and calculating the position and orientation of body segments based on the position of the markers⁵. If data are collected in this manner, errors in the measurements (Var_{er}) originate from two sources (Cappozzo *et al.*, 1996):

- 1. Instrument errors** - which represent the errors in the reconstruction of the marker coordinates in the global/laboratory coordinate system.
- 2. Skin movement errors** - which are due to the relative movement between the markers and the underlying bone.

⁵ This method was used in all studies reported in this thesis and is most commonly used in studies of coordination variability in the literature (e.g. Hamill *et al.*, 1999; Stergiou *et al.*, 2001a,b; Heiderscheit *et al.*, 2002)

Certainly, when variability is the focus of study, an appreciation of the magnitude and nature of these sources of measurement error is required. The mensuration and estimation of skin marker movement errors has received a great deal of attention in the literature – the remainder of this section provides a detailed review of this literature.

To date, several studies have compared the motion of the lower extremity during gait using both skin- and bone-mounted markers simultaneously. Cappozzo *et al.* (1996) investigated the magnitude and pattern of the 'skin movement artefact' in the thigh and shank during various motor tasks. This was achieved using participants recovering from a fracture of the femur or tibia. A set of axes ('fixator technical frame'), rigidly attached to the bone embedded frame, were defined by placing markers on a unilateral external fracture-fixation device. The positions of the additional skin markers, placed on various anatomical landmarks and at other locations on the segments, were determined in the bone embedded frame. The bone embedded frame could be assumed accurate because it was calculated using the fixator technical frame. Movement of these markers in the three cardinal planes was then monitored. The motor tasks analysed included several cycles of walking at a natural cadence and/or several repetitions of cycling on a cycle ergometer. The 'position artefact' was defined as the root-mean-square error of the bone embedded frame coordinate for each marker. Also, for a given bone, anatomical frames were calculated using technical frames based on different clusters of three markers among the available skin markers. The RMS of the

orientation vectors of the 'noisy' frames, relative to the known anatomical frame, represented a measure of the inaccuracy with which the orientation of the bone was reconstructed using skin markers and was named the orientation artefact. Cappozzo *et al.* (1996) suggested that skin marker displacements with respect to the underlying bone might be of a 'remarkable magnitude': from a few millimetres up to 40 mm, with the largest artefacts being most likely for markers located above anatomical landmarks. This indicates that these locations are unsuitable for marker placement (Cappozzo *et al.*, 1996). They also concluded that skin markers placed on the lateral aspect of the thigh and shank, away from the joints and distally on the thigh, exhibit smaller artefact movements and, therefore, allow more reliable results. Maximal errors in bone orientation of between 6° and 20° in the femur and 4° and 10° for the tibia during walking were noted. It was also reported that different marker clusters yielded different orientation artefacts depending on both the geometry of the cluster and the position artefact of the individual markers. When the orientation artefacts were combined, it could be estimated that knee kinematics may be affected by inaccuracies which amount to 10% of flexion-extension, 50% of adduction-abduction and 100% of internal-external rotation. This led the authors to conclude that 'minimisation of skin movement must be the prevailing criterion' in marker set design.

Manal *et al.* (2000) investigated the effect of surface mounted markers and attachment methods on tibial rotation estimates during walking. The purpose of the study was to determine the effect of the location, physical characteristics and attachment method of 11 different marker sets on angular kinematics of the shank,

while also comparing the results to data collected with bone anchored markers, using a percutaneous skeletal tracker. Markers were attached to the skin either directly or using an 'underwrapped' or 'overwrapped' contour moulded shell. The markers were also placed either proximally or distally on the segment to assess the effect of the location of the marker sets. The 11 marker sets were ranked by the RMS values of the tibial rotation estimates. Better estimates of tibial rotation were obtained by placing the marker arrays more distally than proximally over the lateral shank (Manal, *et al.*, 2000). The 'best' marker set was an underwrapped rigid shell of markers placed on the distal-lateral aspect of the shank. However, even when the best marker set was used, the results still demonstrated that soft tissue movement of the shank can affect estimates of internal and external tibial rotation. In fact, when using this marker set, rotational deviations of ± 2 , ± 2 and $\pm 4^\circ$ about the medio-lateral, anterior-posterior and the longitudinal axes respectively were noted.

Holden *et al.* (1997) conducted an investigation to determine the magnitude and pattern of skin movement errors in shank skeletal motion and knee joint kinematics during walking. Holden *et al.* (1997), like Manal *et al.* (2000), used a percutaneous skeletal tracker to infer the true pose of the shank segment. The shank anatomical frame was calculated through the technical frame created by the skin mounted markers and compared to the bone embedded frame created using the markers mounted on the skeletal tracker. The differences in the six degrees of freedom (three translations, three rotations) between the shank anatomical and bone embedded frames were calculated as the 'kinematic surface movement errors'.

The greatest relative rotations occurred about the shank longitudinal axis. Large erroneous rotations about this axis were noted in early and terminal stance. Holden *et al.* (1997) suggested that these erroneous rotations were caused, at least in part, by soft tissue movement resulting from the phasing of muscle contraction and relaxation during the transition phases. Large erroneous longitudinal rotations were also noted during the swing phase, which were due to oscillations of the quiescent muscle mass during that period (Holden *et al.*, 1997). Errors were also noted in the estimation of the knee joint centre and the position of the shank anatomical coordinate system origin of up to 9.0, 10.2 and 10.5 mm, in the medio-lateral, anterior-posterior and longitudinal directions respectively. However, Holden *et al.* (1997) pointed out that these maximum errors using the neoprene wrap with a marker shell were smaller than those reported previously by Angeloni *et al.* (1993) and Karlsson and Lundberg (1994), who placed markers directly onto the skin of the segment.

Reinschmidt *et al.* (1997a) compared tibiocalcaneal motion during running from bone and skin mounted markers. The intracortical bone pins were inserted into the posterolateral aspect of the calcaneus and into the lateral tibial condyle to determine foot and shank segment pose respectively. Shank and foot pose were also calculated using skin markers placed on both the skin of the shank and the running shoe. In all participants and rotations, rotations based on skin markers exhibited a similar pattern to the rotations based on bone markers, but with higher amplitude (Reinschmidt *et al.*, 1997a). Root-mean-square errors, relative to the range of motion (occurring during the stance phase), were 14.1%, 34.7% and

51.2%, for plantar flexion-dorsiflexion, inversion-eversion and abduction-adduction respectively. Surprisingly, the absolute RMSs for the difference between the skin- and bone-based rotations for running were only 1.1° higher for abduction-adduction, 1.2° higher for inversion-eversion and 1.6° higher for plantar flexion-dorsiflexion, than results determined for walking (Reinschmidt *et al.*, 1997c). It was also evident from a 'segmental error analysis', that, for all rotations, the error due to the shoe mounted markers was greater than the error due to the shank mounted markers. This finding led Reinschmidt *et al.* (1997a) to conclude that markers placed on shoes should not be used to quantify movement at the ankle joint. As an alternative, they suggested that markers should be placed on the skin of the foot through windows cut into the shoe. In some participants maximal eversion was highly over estimated (over 100% higher in some participants), suggesting that all maximal eversion values should be considered as over-estimations of the true calcaneus to tibia eversion. They also speculated that the relationship between maximal eversion derived from skin- and bone-based markers is likely to be shoe dependent, which suggests a need for control over footwear.

Reinschmidt *et al.* (1997b) studied the effect of skin movement on the analysis of knee joint motion during running. Using the same method as Reinschmidt *et al* (1997a), they determined the pose of the thigh and shank using skin- and bone-based markers. The shape and magnitude of the knee flexion-extension curves were similar for all participants (Reinschmidt *et al.*, 1997b). Nevertheless, agreement between kinematics for skin- and bone-based markers was poor. No similar error patterns were present in the data for all the rotations, making it difficult

to develop a general algorithm to correct for skin marker movement (Reinschmidt, *et al.*, 1997b). In their study, markers were placed onto the skin directly and Reinschmidt *et al.* (1997b) suggested that any method, other than direct bone-pin measurements (e.g. rigid marker clusters and mathematical algorithms), only partially solves the problem of skin marker movement. Reinschmidt *et al.* (1997b) reduced the motion of the markers relative to each other, but the marker set might still move relative to the underlying bone (Reinschmidt *et al.*, 1997b). These authors also reported that markers should not be placed on the bellies of large muscles. This is especially relevant on the thigh segment, where 'most thigh errors were caused by muscle activity, and errors due to inertial effects were rather small' (Reinschmidt *et al.*, 1997b).

In a further study, Reinschmidt *et al.* (1997c) assessed the skin marker movement error in both tibiofemoral and tibiocalcaneal motion during walking. They achieved this using similar methods to their previous studies (Reinschmidt *et al.*, 1997a,b). Reinschmidt *et al.* (1997c) concluded that, during walking, skin markers could reliably determine only flexion-extension movements at the knee joint. They noted that, when looking at motion in the transverse and frontal planes at the knee joint, the error introduced by the movement of the skin can be almost as large as the actual joint motion. At the ankle joint complex, kinematics based on shoe or skin markers gave a relatively good estimate of the actual tibiocalcaneal motion when compared to the skin-based estimates of tibiofemoral motion.

2.6.4 The comparison of overground and treadmill locomotion

The treadmill is often used in both clinical and research settings to simulate the mechanics of overground locomotion. Its convenience makes the treadmill an attractive instrument for investigating human locomotion (Schache *et al.*, 2001). Space requirements are constrained, considerably reducing the size of the required calibration volume for capturing kinematic data. Additionally, environmental factors can be controlled, steady locomotion speeds are easily maintained and the participant stays in close proximity to other biomechanical equipment throughout testing. A further advantage, that is extremely relevant for studies of locomotion variability, is that multiple consecutive strides can be captured in one trial. However, a problem of ecological validity could exist if kinematic, kinetic and metabolic variables, for example, differ between the two forms of locomotion. Despite the fact that van Ingen Schenau (1980) presented evidence, using a theoretical mathematical approach, that the mechanics of treadmill and overground locomotion are basically the same, the experimental evidence in the literature is equivocal. Nigg *et al.* (1995) suggested that differences of opinion exist in the literature about the degree of similarity between overground and treadmill gait. This section outlines the results of studies that have compared overground and treadmill locomotion.

2.6.4.1 Kinematic comparisons of overground and treadmill locomotion

In an early study, Brookes *et al.* (1971) investigated possible biomechanical and physiological differences between overground and treadmill running. Using a very small sample size ($n = 3$) Brookes *et al.* (1971) measured heart rate, oxygen

uptake, ventilation, respiratory rate and stride rate. They reported no significant difference between overground and treadmill running for stride rate in addition to the physiological variables. The lack of significant differences reported by Brookes *et al.* (1971) could be explained purely by the study's lack of statistical power ($n = 3$). However, Nelson *et al.* (1972) did report significantly lower stride rates during treadmill running but only at high running velocities ($6.4 \text{ m}\cdot\text{s}^{-1}$). As might be expected, this decrease in stride rate was accompanied by a significant increase in stride length during treadmill running at $6.4 \text{ m}\cdot\text{s}^{-1}$. The study by Nelson *et al.* (1972) investigated the differences between overground and treadmill running on uphill and downhill gradients in addition to level running at varying velocities; only the results from the comparison of level treadmill and overground running are discussed here. Their study also showed evidence of longer periods of support during treadmill locomotion at $3.35 \text{ m}\cdot\text{s}^{-1}$ and $4.88 \text{ m}\cdot\text{s}^{-1}$. Additionally, participants exhibited decreased vertical velocity of the centre-of-mass and decreased variability in the vertical and horizontal velocity in treadmill running. Nelson *et al.* (1972) suggested that two modifications occur during treadmill running which are interrelated. The participants tended to place their foot down further in front of their centre-of-mass - which would lead to an increased deceleration of the centre-of-mass during overground running. They then allow the moving belt to run their feet underneath them (Nelson *et al.*, 1972). For a given velocity, this tended to increase the time of support and decrease the time of non-support. Nelson *et al.* (1972) concluded that the biomechanics of treadmill running differ significantly from those associated with overground running.

Dal Monte *et al.* (1974), like Brookes *et al.* (1971), compared overground and treadmill running using only a small sample size ($n = 3$). In agreement with Nelson *et al.* (1972), they reported less vertical movement of the centre-of-mass and a decreased period of non-support on the treadmill. Dal Monte *et al.* (1974) also reported that treadmill running resulted in decreased stride length. Despite these apparent differences between overground and treadmill running, they concluded that the treadmill can be used as a specific simulator for middle-distance running at speeds typically employed during competitions.

Elliot and Blanksby (1976) conducted a cinematographic analysis of overground and treadmill running using male ($n = 12$) and female ($n = 12$) participants. Unlike Nelson *et al.* (1992) and Dal Monte *et al.* (1973), the participants in this study were not trained athletes but simply college students with experience of treadmill locomotion. At running speeds of greater than $4.8 \text{ m}\cdot\text{s}^{-1}$, treadmill running elicited a decreased period of non-support, less vertical movement of the centre-of-mass and decreased stride length, when compared to the overground condition. Therefore, the authors concluded that, at speeds in excess of $5 \text{ m}\cdot\text{s}^{-1}$ but not below, biomechanical differences do exist between overground and treadmill running.

In addition to the studies cited above which compared overground and treadmill locomotion using spatio-temporal parameters, more recently, supplementary angular kinematics variables have been included as dependent measures (e.g. Frishberg, 1983; Strathy *et al.*, 1983; Nigg *et al.*, 1995; Wank *et al.*, 1998; Alton *et*

al., 1998). Frishberg (1983) compared overground and treadmill sprinting using a cohort ($n = 5$) of college athletes. It was reported that while sprinting ($9.2 \pm 0.1 \text{ m}\cdot\text{s}^{-1}$) no significant differences were apparent for spatio-temporal parameters between overground and treadmill running. However, significant differences were seen between the two modes of sprinting for angular kinematics variables. Most of these differences occurred in or just before the support phase and concerned the supporting leg (Frishberg, 1983). At footstrike, participants exhibited a decrease in leg angle, increased thigh angle, more extended knee angle and reduced forward trunk lean at foot strike in the treadmill running condition.

Nigg *et al.* (1995) compared overground and treadmill running using runners of differing running experience (11 runners with experience of treadmill locomotion, 11 non-runners without experience of treadmill locomotion). The authors reported differences between overground and treadmill running - when the two skill groups were collapsed - including leg angle and shoe sole angle at footstrike, range of rear-foot inversion from maximum eversion position to toe-off and range of rear-foot eversion from footstrike to maximum eversion. Also, while running on the treadmill, the shoe sole angle decreased when compared to overground running. These differences were only seen at speeds of under $6 \text{ m}\cdot\text{s}^{-1}$ which contradicts the findings of other authors (e.g. Elliot and Blanksby, 1976; Frishberg, 1983) and the suggestion of Williams (1985) that differences between overground and treadmill running are only apparent at speeds greater than $5 \text{ m}\cdot\text{s}^{-1}$. Single-individual comparisons between overground and treadmill locomotion were also discussed and many different patterns of adaptation were apparent. This phenomenon of

large inter-individual variability is consistent with other studies in biomechanics that incorporated kinematic and kinetic dependent variables (e.g. Bates *et al.*, 1983; DeVita and Bates, 1988; Dufek and Bates, 1991, Hreljac, 1998, Dixon *et al.*, 2000).

Wank *et al.* (1998) also conducted a comparison of overground and treadmill running and analysed many kinematic variables in addition to electromyographic (EMG) measurements - only the kinematic results will be discussed here - at both 4 and $6 \text{ m}\cdot\text{s}^{-1}$. While stride rate increased significantly, both stride length and contact time decreased during treadmill running at both speeds. Similarly to Nelson *et al.* (1972) and Dal Monte *et al.* (1973), treadmill running was characterised by a decrease in the vertical displacement of the centre-of-mass. Differences in knee joint angles were also seen between the two modes at both running speeds. For example, the knee joint angle at footstrike was significantly more extended during treadmill locomotion at both $4 \text{ m}\cdot\text{s}^{-1}$ and $6 \text{ m}\cdot\text{s}^{-1}$, which contradicts the findings of Frishberg (1983) who compared overground and treadmill sprinting. Differences were also reported for trunk angle in addition to maximum and minimum hip joint angle.

Kinematics of the lumbo-pelvic-hip complex have also been investigated during overground and treadmill locomotion. For example, Schache *et al.* (2001) reported that only three out of 25 kinematic variables that they measured for the lumbar spine and pelvis exhibited significant differences between overground and treadmill running. These were: lumbar extension at footstrike, anterior pelvic tilt at footstrike and the first maximum anterior pelvic tilt. Also, of the 17 hip parameters analysed,

only hip extension at toe-off, maximum hip extension, hip flexion-extension range of motion and hip flexion at footstrike were significantly different between the modes of locomotion. Also, in agreement with Wank *et al.* (1998), stride length decreased, whereas stride rate and contact time increased in the treadmill condition when compared to the overground condition.

All of the studies cited above reported a comparison of overground and treadmill running. Further to this, authors have reported both similarities and differences in kinematics during overground and treadmill walking (e.g. Strathy *et al.*, 1983; Lemke *et al.*, 1995; Alton *et al.*, 1998; Vogt *et al.*, 2002). Strathy *et al.* (1983) measured changes in knee joint function associated with treadmill walking using an electro-goniometer. Significant differences between overground and treadmill walking were seen for knee flexion-extension range of motion during the full stride and during the swing phase in isolation. Greater range of motion was seen in both of these phases during overground walking. Additionally, the knee was reported to be in significantly more extended position at footstrike and greater swing phase extension was reported in the overground condition. Strathy *et al.* (1983) found no differences in knee joint motion in the frontal and transverse planes. In addition to the knee kinematics data, Strathy *et al.* (1983) reported that average heel contact time on the treadmill was reduced compared to during overground walking, while the average toe contact time increased. Also, consistent with some studies of running (e.g. Dal Monte *et al.*, 1974; Wank *et al.*, 1998), treadmill walking was associated with an increase in stride rate compared to overground walking.

In a more recent study, Lemke *et al.* (1995) specifically conducted a comparison of rearfoot motion between overground and treadmill walking and reported no significant differences between the two conditions for any dependent variable. These included: stance phase duration, time-to-heel-off, maximum rearfoot pronation angle, time-to-maximum rearfoot pronation angle, maximum tibial angle, time-to-maximum tibial angle, maximum calcaneal angle and time-to-maximum calcaneal angle. Lemke *et al.* (1995) supplemented the statistical analysis with paired *t*-tests using Pearson Product-moment Correlation Coefficients between the two walking conditions for all dependent variables. For all temporal values the correlation between overground and treadmill walking was poor, whereas good agreement was noted for the amplitude of the rotations. This led the authors to conclude that the treadmill should not be used to simulate overground walking when the timing of rearfoot movements is of interest. However, the efficacy of using Pearson Product-moment Correlation Coefficients to measure agreement has previously been questioned (see Atkinson and Nevill, 1998).

Alton *et al.* (1998) noted differences between overground and treadmill walking. When the participants were collapsed across gender, significant increases were reported during treadmill walking in hip range of motion, maximum hip flexion joint angle and stride rate - a concurrent significant decrease in stride length was also noted. However, it should be noted that Alton *et al.* (1998) questioned the validity of their hip angle measurements as they used a marker placed on the acromion process of the shoulder. It is possible that artefacts were introduced into the calculation of hip joint angle due to rotations of the trunk and movements of the

scapula. Further studies are required to confirm or refute the findings related to hip joint motion by using better marker placement schemes for detecting hip angles (Alton *et al.*, 1998).

Similarly to comparisons of overground and treadmill running, differences between the two modes of walking have been reported for rotations in the lumbar spine and pelvis (e.g. Vogt *et al.*, 2002). Vogt *et al.* (2002) reported significant reductions in pelvis and upper lumbar rotation amplitudes in the transverse and frontal planes during treadmill walking compared to overground walking. The authors suggested that the observed differences should be taken into account when treadmill based pelvis and lumbar spine kinematics are to be extended or compared to overground walking.

2.6.4.2 Summary of kinematic differences between overground and treadmill locomotion

The literature presented in the previous sub-section reiterates the comment of Nigg *et al.* (1995) that there is little consensus on the differences or similarities between overground and treadmill locomotion. Each comparison of the two modes of locomotion revealed differences in angular kinematics variables between overground and treadmill locomotion that were largely inconsistent with other studies. Indeed, some studies revealed no significant differences between the two modes. However, it is questionable whether or not many of these studies (e.g. Brookes *et al.*, 1971; Lemke *et al.*, 1995) had sufficient power to detect differences between treadmill and overground conditions. Consistent findings have been

reported in the literature for spatio-temporal variables describing locomotion. For example, treadmill locomotion has often been associated with an increase in stride rate compared with overground locomotion (e.g. Dal Monte *et al.*, 1974; Strathy *et al.*, 1983; Alton *et al.*, 1998; Wank *et al.*, 1998; Schache *et al.*, 2001). However, evidence to the contrary has also been presented (Nelson *et al.*, 1972).

Where differences have been reported between overground and treadmill locomotion, many authors have hypothesised why they might have occurred. For example, Frishberg (1983) postulated that the participants might experience lower ground reaction forces during treadmill locomotion due to different surface characteristics. Also, since the participant is stationary on the treadmill, the changes in air resistance acting on the participant might cause differences in locomotion. This might be especially important at fast, sprinting speeds, which were the focus for Frishberg (1983). A further reason cited by Nelson *et al.* (1972) and Frishberg (1983) was that, during treadmill locomotion, the treadmill belt moves the supporting foot back under the trunk rather than having the trunk move over the supporting foot, as in overground locomotion. An additional interesting factor, first reported by van Ingen Schenau (1980), is a change in the perceptual information available to the participant whilst running or walking on a treadmill; whether it be changes in visual, or possibly to a lesser extent, auditory information. During overground locomotion, the participants experience changing optical flow information which is not the case in treadmill locomotion. As perception and action are tightly coupled (Williams *et al.*, 1999), changes in perceptual information could cause a difference in the regulation of the movement pattern resulting in

differences in the kinematics of locomotion (van Ingen Schenau, 1980). Further factors that have been cited in the literature that might explain the differences between overground and treadmill locomotion include the intra-stride belt speed variations of the treadmill and the level of participant familiarity with treadmill locomotion. Indeed, these factors, along with differences in the speed of locomotion and the type of population from which the participants are sampled, have been implicated as potential reasons for the disparity between studies that have compared overground and treadmill locomotion (Savelberg *et al.*, 1998; Matsas *et al.*, 2000; Wass *et al.*, 2004). They have also been identified as important factors to control in any study involving treadmill locomotion and they are reviewed in detail in sections 2.6.4.3 and 2.6.4.4 below.

2.6.4.3 Treadmill habituation

It has previously been suggested that a lack of familiarity with treadmill locomotion might be a reason for differences between overground and treadmill locomotion that have been observed in the literature. Indeed, Matsas *et al.* (2000 p.52) suggested that 'failing to consider treadmill familiarisation appears to be the major limitation of studies which had previously unsuccessfully attempted to validate measurements obtained from the treadmill'. This section provides a review of studies that have attempted to determine the affects of treadmill familiarisation, or lack thereof, on the kinematics of locomotion.

Wall and Charteris (1980) suggested that the process of treadmill familiarisation might be a long one. They demonstrated that, even after 10 minutes of treadmill

walking, a steady state had not been achieved by any of their groups of male participants. This confirmed the findings of a previous study on female participants (Charteris and Taves, 1978). In a follow-up investigation, Wall and Charteris (1981) examined the long term habituation to treadmill walking. Their results suggested that, when a motorised treadmill is to be used to analyse kinematics during gait, participants should be previously familiarised in distributed practice sessions for a total of one hour and then not measured within the first two minutes of performance. Over the familiarisation period in these studies, conducted on participants that were naïve to treadmill walking, differences were noted in spatiotemporal parameters of gait, such as stride time and heel contact time. Furthermore, differences were also noted for angular kinematic variables such as knee flexion-extension angle.

Scheib (1986) investigated the effects of treadmill training on the familiarisation process of novice treadmill runners. Six dependent variables were determined, including stride length, stride rate, stride time, time of support, time of non-support and vertical and lateral displacement of the centre-of-mass during ten 15-minute runs on consecutive days. Estimates of each of the variables were determined at minutes one, eight, and fourteen. Scheib (1986) reported significant adaptations both within and between days for stride length, stride rate, stride time and vertical displacement of the centre-of-mass. This led Scheib (1986) to conclude that misleading or inaccurate information might result if a novice treadmill participant is used for treadmill data acquisition. It was stated that an experienced overground runner, but novice treadmill runner, can become familiar with the treadmill device

after 8 minutes of treadmill running. However, the process of 'complete habituation' may take a minimum of three 15 minute training sessions or a combination of sessions totaling approximately 45 minutes.

In a more recent study, Matsas *et al.* (2000) investigated the familiarisation period required to obtain reliable sagittal-plane knee kinematics and spatio-temporal gait measurements during treadmill walking. Using participants that were unfamiliar to treadmill locomotion, highly reliable knee kinematics variables were found after only four minutes of treadmill walking. Reliable spatio-temporal variables were reported after six minutes of treadmill walking. These periods of treadmill familiarisation are considerably shorter than those suggested by the previous studies considered in this section. However, Taylor *et al.* (1996) similarly reported that measurements of the angular motion of the lumber spine and pelvis were reliable after only four minutes of treadmill familiarisation – intraclass correlation coefficients > 0.83 .

Wass *et al.* (2004) investigated familiarisation to treadmill walking in older adults. They reported that, contrary to the findings of Matsas *et al.* (2000) using younger adults, older adults had not familiarised to treadmill walking after 15 minutes. Knee kinematics and spatiotemporal parameters were not reliable after this period of familiarisation in this population. Furthermore, treadmill walking after 15 minutes was not 'closely related' to overground walking. This provides evidence for the concerns of Taylor *et al.* (1996) who suggested that one should be cautious about

generalising the suggestions for familiarisation periods to pathological and elderly populations.

The investigations presented in this section highlight the importance of treadmill familiarisation in studies of the kinematics of locomotion using a treadmill. Furthermore, it is evident that a lack of treadmill familiarity in participants taking part in comparisons between locomotion on treadmill and overground might account for some of the differences presented in section 2.6.4.1 and the disparity in the results of such studies. However, it should be noted that all studies reported in this section used participants, whether young or elderly, that were naïve to treadmill locomotion and it is reasonable to assume that the length of these periods would not apply to participants that had prior experience and were accustomed to treadmill locomotion.

2.6.4.4 The effect of intra-stride belt speed variations on treadmill locomotion

It is often assumed that the effective energy transfer between the treadmill and the participant in a study equals zero (Savelberg *et al.*, 1998). However, this is only the case in a situation in which the treadmill belt speed is constant and the moving coordinate system used to describe the movement has the averaged treadmill belt speed. As Savelberg *et al.* (1998) stated, however, due to the anterior-posterior component of the ground reaction force and the imperfect speed control of the treadmill engine, the treadmill belt speed might be expected to change throughout the stance phase of walking or running. These intra-stride belt speed variations

result in energy transfer between the treadmill and the participant. Savelberg (1998) hypothesised that these variations might interfere with normal locomotion mechanics and be a reason for potential differences between overground and treadmill locomotion. In a similar study, Nigg *et al.* (1995) indicated that differences between a large treadmill and overground running were greater than the differences between a small treadmill and overground running. Possibly, this indicates that the power of the treadmill has no influence on the difference between overground and treadmill running. However, it should be noted that the large treadmill used by Nigg *et al.* (1995) had a 2.5 kW motor which is less powerful than the smaller of the two motors used by Savelberg *et al.* (1998) – see below. Nigg *et al.* (1995) did not state the magnitude of the intra-stride belt speed variations for any of the treadmills used in their study.

Savelberg *et al.* (1998) investigated the effects of walking and running on a high-powered (22 kW motor) treadmill, low-powered (3.4 kW motor) treadmill and overground on various kinematic parameters. Of 43 kinematic parameters considered, six were significantly different between the high-powered treadmill and overground locomotion. Eight were significantly different between the low-powered treadmill and overground locomotion. Furthermore, when comparing both treadmills, 11 parameters were significantly different.

An analysis of covariance (ANCOVA) was used to investigate the dependence of the spatial and temporal kinematic factors on treadmill belt-speed variation. The ANCOVA allowed Savelberg *et al.* (1998) to conclude that the low powered

treadmill appeared to mainly change the patterns at the proximal joints, whereas the high-powered treadmill appeared to affect distal joint kinematics. Savelberg *et al.* (1998) concluded that the changes in energy flow between the treadmill and the participant due to intra-stride belt speed variations are a concern for studies using a treadmill to monitor locomotion. The authors also suggested that the intra-stride belt speed variations might be a reason for the differences seen between treadmill and overground locomotion and the obvious disparity in the results of different studies that was highlighted in section 2.6.4.1 of this thesis.

A further important finding from the study of Savelberg *et al.* (1998) was that the power of the motor of the treadmill was significantly related to the intra-stride belt speed variations ($r^2 = 0.55$). Other factors investigated, including participant mass and speed of locomotion, had a relatively small effect on intra-stride belt speed variations. As Savelberg *et al.* (1998) pointed out though, other factors, such as the treadmill belt speed control unit, might also influence the intra-stride belt speed variations but they appear to be strongly related to the power of the treadmill motor. However, Paul (2001) noted that, although augmenting the power of a treadmill could reduce the magnitude of the intra-stride belt speed variations, the power of the treadmill is not a direct factor. Paul (2001) suggested that the reduction in the intra-stride belt speed variations in higher powered treadmills is related to the fact that these treadmills generally have larger fly wheels than lower powered treadmills. It is these larger fly wheels that reduced the magnitude of the intra-stride belt speed variations.

A possible limitation of the study of Savelberg *et al.* (1998), however, is the fact that they used anterior-posterior ground reaction forces taken from an overground locomotion condition to calculate the change in energy flow between the treadmill and the participant. They did so because they believed that techniques to measure such forces on a treadmill were not yet available. Treadmills with in-built force platforms capable of the collection of three-dimensional ground reaction forces have now been developed (for example, see Kram *et al.*, 1998 or Determan *et al.*, 2004). Although Kram *et al.* (1998) found very little difference in anterior-posterior forces between overground and treadmill walking, in a more recent study, Determan *et al.* (2004) reported significantly reduced braking impulses during treadmill running. However, as Determan *et al.* (2004) highlighted, the participants in the study tended to slow down slightly during overground locomotion which could account for the greater braking impulse reported in this condition. Nonetheless, a study replicating the methods of Savelberg *et al.* (1998) with treadmills capable of measuring three-dimensional ground reaction forces is warranted.

2.6.4.5 Kinematic variability during overground and treadmill locomotion

In comparison to kinematics, there are very few studies that have investigated similarities or differences between overground and treadmill locomotion in terms of movement variability. No investigations have compared coordination variability between overground and treadmill locomotion. Limited evidence exists of differences in the variability of kinematic variables between the two forms of locomotion. Dingwell *et al.* (1999) investigated the effect of diabetes mellitus and

peripheral neuropathy on sagittal plane kinematics during treadmill walking. There was a trend for patients with diabetes mellitus and peripheral neuropathy to exhibit decreased kinematic variability. Additionally, Dingwell (1999) presented results suggesting that treadmill walking elicited less variable kinematics than overground walking. Coefficients of variation were 'substantially less than those reported by Winter (1983) for a group of young healthy adults walking overground' (Dingwell *et al.*, 1999, p.27). The coefficients of variation for the knee flexion-extension and ankle plantar flexion-dorsiflexion presented by Dingwell *et al.* (1999) were 43% and 57% less than those presented by Winter (1983) for the same joints, respectively. Dingwell *et al.* (1999) suggested that the differences were due to the fact that they had collected treadmill walking data whereas Winter (1983) collected data from overground walking.

In a further study, Dingwell *et al.* (2001) compared sagittal plane ankle, knee and hip angles collected during overground and treadmill walking. Treadmill walking was associated with small but significant reductions in variability compared to overground walking (Dingwell *et al.*, 2001). These results are in agreement with previous findings (Nelson *et al.*, 1972; Wank *et al.*, 1998). Nelson *et al.* (1972) reported less variability in the horizontal and vertical velocities of the centre-of-mass while running on a treadmill as opposed to overground running. Results similar to this were presented by Wank *et al.* (1998) who also reported decreased variability in the horizontal and vertical velocities of the centre-of-mass on the treadmill when compared to overground running. Dingwell *et al.* (2001) suggested that treadmills should not be used to study locomotion in certain circumstances.

Other work by Dingwell and colleagues was cited by Dingwell *et al.* (2001) as being pertinent to this argument. For example, kinematic variability was not significantly different between patients with diabetic neuropathy and a control group when participants walked on a treadmill (Dingwell *et al.*, 1999). However, when a similar cohort of participants walked overground, significant differences between groups were apparent (Dingwell *et al.*, 2000).

Although there is some evidence in the literature of differences between overground and treadmill walking in terms of kinematic variability, there is very little regarding running. This dearth of literature needs addressing. There is a clear need for studies that compare coordination variability during overground and treadmill locomotion as studies measuring coordination variability during walking and running have employed both overground (e.g. Hamill *et al.*, 1999) and treadmill (e.g. Heiderscheit *et al.*, 2002) methods.

Treadmills are very attractive for studies of variability as they enable the collection of multiple consecutive strides which are difficult to obtain overground using the commonly used stereophotogrammic techniques. However, an obvious problem exists if coordination variability measured during treadmill locomotion is not representative of coordination variability measured overground. If treadmill locomotion is inherently less variable than overground locomotion it would, potentially, be a large problem in studies in which a treadmill has been used. For example, the reduced variability due to the treadmill could mask either differences between experimental groups or differences due to an intervention.

2.7 Summary of the literature review

This literature review has attempted to provide a critical overview of research into movement variability from traditional and contemporary viewpoints. The review has directed the reader through the developments in the approach to movement variability over the past 30 years. Specifically, there has been a change in conceptualisation of movement variability from being noise or error in the data (in the traditional, cognitive view point) to accepting that variability is inherent and should be expected, to some degree, in movement. Indeed, research presented in sections 2.4 and 2.5 highlighted potentially functional roles for variability in a wide variety of human systems. The major focus of this thesis is the possible relationship between variability and overuse injury. The second part of the review highlighted considerations for the study of variability in movement during locomotion in this context.

Four main methodological considerations for the study of variability during locomotion were identified and reviewed; measures of variability, measurement errors, measures of coordination and coordination variability and the use of a treadmill to simulate overground locomotion. Gaps in the literature were highlighted which provides a rationale for the studies presented in this thesis. Furthermore, they warrant the formulation of the aims of the thesis presented in section 1.1. Particularly, it is clear that the errors associated with any study, especially those related to the variability in movement, should be quantified. Also, it is clear that determining whether the errors due to skin marker movement are systematic or

random would be beneficial to studies of variability during locomotion. Furthermore, it remains unclear which of the techniques available for the quantification of coordination and coordination variability is most appropriate. Finally, it is evident that there is a dearth of literature addressing whether measuring coordination variability on a treadmill represents that which would be measured overground. The overall purpose of this thesis was to address these issues – see section 1.1 for specific aims of each chapter.

CHAPTER III

3 The effect of error on the measurement of coordination

variability

3.1 Introduction

In section 2.6.3 of the literature review, experimental errors that might affect the measurement of movement variability were considered. It was suggested that the total variability measured in any system – the variability that would be measured experimentally – is made up of three components: Var_h , inherent human movement variability, Var_{er} , variability due to errors in the measurement and Var_{en} , variability due to other sources, e.g. changes in the environmental and task constraints – see Equation 2.3. It is reasonable to assume that, in studies of coordination variability during locomotion, Var_h is an important component in which a researcher might be interested and their aim would be to separate Var_h from the other components. In a well controlled research environment Var_{en} will be minimal and relatively easy to isolate. This is because either task or environmental constraints are likely to be manipulated as independent variables whilst any constraints that are not manipulated should be controlled to minimise the effect of extraneous variables on the results of the study. However, the process of separating the inherent human variability from Var_{er} is far from trivial.

Before the coordination between body segments can be assessed, kinematic measurements must be made of the position and orientation of the segments of

interest. This requires the determination of the instantaneous position and orientation, hereinafter referred to as pose, of systems of axes which are considered to be rigid with the bones under consideration (Cappozzo *et al.*, 1996). Image-based motion analysis systems are generally used to obtain three-dimensional coordinates of groups of markers attached to the segments of interest. Subsequently, the instantaneous pose of the marker cluster is estimated and associated with the underlying bone (Cappozzo, *et al.*, 1996). These markers may be attached either to the skin of the segment or by a direct link to the underlying bone, using bone pins for example. Although *in vivo* measurement of the pose of bone using markers attached to the bones directly is considered more accurate, it is highly invasive and not normally used (Cappozzo *et al.*, 1996). Ramsey *et al.* (1999) recently outlined some of the problems and methodological considerations associated with the measurement of kinematics using bone pins. The techniques used can restrict or interfere with gait or the natural movement of soft tissues, and their invasive nature has limited the number of participants studied (Holden *et al.*, 1997). For this reason the pose of bone is generally determined using markers attached to the skin directly (e.g. McClay and Manal, 1998; Mosely *et al.*, 1996; Lafourture *et al.*, 1994) or through some kind of fixture (e.g. Holden *et al.*, 1997; Cappozzo *et al.*, 1995; Stacoff *et al.*, 2000). Cappozzo *et al.* (1996) reported that when markers are attached to skin, inaccuracies in the estimation of the pose of the underlying bone are due to:

- 1. Instrument errors** - which represent the errors in the reconstruction of the marker coordinates in the global coordinate system.
- 2. Skin movement errors** - which are due to the relative movement between the markers and the underlying bone.

The following two sections consider separately both instrument and skin movement errors. Subsequently, the effect of these errors on the study of coordination variability is discussed.

3.2 Instrument errors

Photogrammetry has been used in the movement sciences since Muybridge's well-known sequence of horse movement first published in 1878 (Chiari *et al.*, 2005). Technology has developed over time and equipment that is capable of automated tracking of human motion is now available. Automated image-based motion analysis devices can be divided into passive marker systems, that reflect light emitted by the sensors, and active marker systems, the markers of which incorporate their own light source. Both active and passive systems offer particular benefits and limitations. As Chiari *et al.* (2005) suggested, the accuracy and sampling rates of passive marker systems are, generally, not as good as those for active marker systems. However, the absence of wires, batteries and pulsing circuitry attached to the participant with passive systems is a major advantage, making these systems more appropriate in most instances of human movement analysis (Chiari *et al.*, 2005). In recent years the measurement characteristics of

these systems have been studied (Linden *et al.*, 1992; Ehara *et al.*, 1995; Klein and De Haven, 1995; Ehara *et al.*, 1997; Richards, 1999; Kadaba and Stine, 2000). Kadaba and Stine (2000) suggested that the measurement of movement, based on video systems using passive retro-reflective markers, continues to be popular due to their high accuracy, resolution and precision, as well as their flexible and non-intrusive nature. Furthermore, with the significant technological advances made in the last ten years it is now possible to measure movement in real time, providing an opportunity to improve existing methods of motion measurement (Kadaba and Stine, 2000). A passive marker system, capable of real time capture, was used during this programme of research.

To be worthwhile, any coordinate reconstruction technique needs to be accurate and precise (Challis *et al.*, 1997). Accuracy is the difference between a true and an observed value, whereas precision is the difference between an observed value and the mean of all of the observed values. Essentially the accuracy and precision of a motion analysis device can be equated to the errors in the motion measurement (instrument errors). The equation used to calculate root-mean-square error (RMS) is presented in Equation 3.1 below.

$$\text{RMS} = \sqrt{\frac{\sum_{i=1}^n (x_c - x_i)^2}{n}} \quad [\text{Equation 3.1}]$$

Where x_i is the value of the variable x at the i th data point, x_c is the criterion or known value of the variable x and n is the number of data points. As the measure

of root-mean-square error incorporates a known criterion value - the 'true' value, it can be used to infer the accuracy of a system. On the other hand, standard deviation is given by Equation 3.2 and is the measure of the variability of the measurement about the mean of the measurement.

$$\text{Standard Deviation} = \frac{\sqrt{\sum (x_m - x_i)^2}}{(n - 1)} \quad [\text{Equation 3.2}]$$

Where x_i is the value of the variable x at the i th data point, x_m is the mean measurement of x and n is the number of data points. Unlike root-mean-square error, the measure of standard deviation gives an indication of the deviation from a mean value, which is synonymous with a measurement of precision. The accuracy of an image-based motion analysis system has commonly been reported as the difference between the true location of a set of control points and their measured values (Challis *et al.*, 1997). However, Challis and Kerwin (1992) showed that using the same control points for calibration and accuracy assessment over-estimates accuracy. Therefore, an independent set of control points is essential for accuracy assessment (Challis *et al.*, 1997).

Richards (1999) recently used a device independent of the calibration instrument to measure and compare the 'accuracy' of seven commercially available image-based motion analysis systems. The device used to test the seven systems incorporated known distances of 500 mm and 90 mm, along with a known angle of 95.8°, which was defined using a triangle of markers. The average measured

distance or angle and the maximum absolute error were calculated for each variable in each trial. The 'root-mean-square difference' was also calculated. However, although it was termed root-mean-square difference, Richards (1999) did not use a criterion or known value in the calculation, using instead the 'average measured distance'. This is the same as taking the population standard deviation of the data, and suggests that Richards (1999) measured the precision and not the accuracy of the motion analysis systems. Another criticism of Richards' (1999) study is that data were collected from only one position in the volume. In fact, the size of the volumes used for each of the motion analysis systems was not common (the volume lengths ranged from 1.73 - 4.60 m). It is reasonable to assume that the accuracy/precision of the system would change at different points in the volume due to the nature of the calibration procedures used with these types of systems. Therefore, a study that investigates the accuracy and precision of these systems at different points in the calibration volume is warranted. The aim of this study was to determine the accuracy and precision of the motion analysis system used in the present programme of research using a procedure similar to that outlined by Richards (1999).

3.2.1 The motion analysis system testing device

A modified version of the device used by Richards (1999) and Kadaba and Stine (2000) was constructed and is shown in Figure 3.1 below:

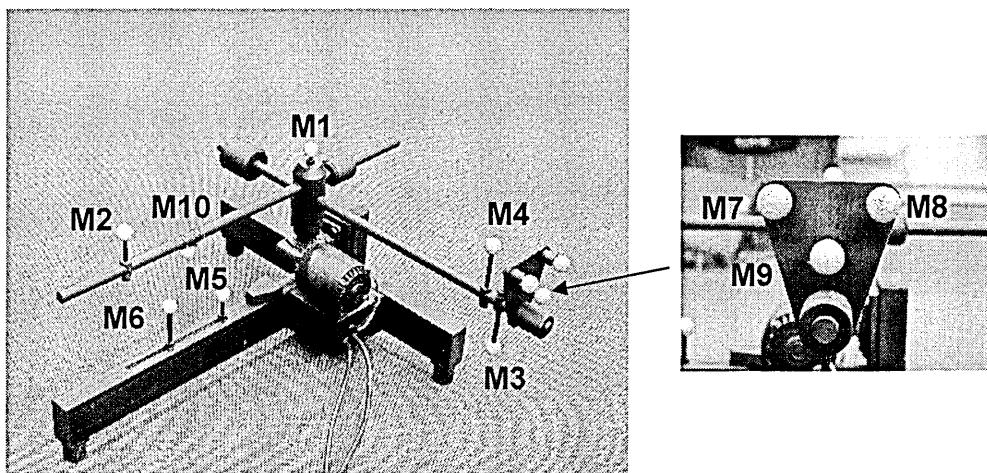


Figure 3.1: The testing device - M1 to M10 represent markers 1 to 10 respectively

Marker 1 served as the origin of the device and was mounted on the axis of rotation of the two perpendicular arms. Mounted on arm_1 was a marker (marker 2) placed 400 mm away from the origin. Arm_2 supported a further five markers (markers 3, 4, 7, 8 and 9). Marker 4, like marker 2, was placed 400 mm away from the origin of the device. Markers 7, 8 and 9 were placed in a triangular pattern on a metal plate mounted on the end of arm_2. The distance between markers 7 and 8 was 80 mm. Due to unforeseen problems with marker drop-out owing to marker 9 obscuring markers 7 and 8, marker 9 was covered during all trials and was omitted from any analysis. Marker 10 was placed on arm_1 to enable the arm's local coordinate system to be defined during the calculation of the Cardan angle - see section 3.2.3. Markers 1, 2, 4, 7 and 8 were set approximately in the xz plane of

the laboratory using a spirit level located on the device. Marker 3 was mounted on a post at the bottom of arm_1. This post placed marker 3 at the same vertical height as an adjustable marker (marker 6 on the base of the device). The final marker (marker 5) was non-adjustable and was mounted on the base of the device. During dynamic trials, the rigidly connected arms 1 and 2 rotated at a rate of 60 revolutions per minute. The rotation was driven by a synchronous motor mounted on the base of the device. The manufacturers of the device (SCEPTRE, Sheffield, UK) gave assurance that the device was constructed to an accuracy of 0.25 mm.

3.2.2 Procedures

The motion capture system tested here was an eight digital camera system (Motion Analysis Corporation, Santa Rosa, CA, USA). The eight cameras were placed in optimal positions around a measurement volume of length = 4.4 m, width = 2.1 m and height = 1.6 m. The cameras were judged to be in optimal positions when the largest possible amount of measurement volume was in view of the cameras with a minimum amount of 'dead space' (regions of the camera view outside of the measurement volume). Calibration of the motion analysis system involved a two-step procedure. First, a 'seed' calibration was performed during which an L-shaped device containing four markers at known locations was placed in the measurement volume to define the global coordinate system. Second, camera properties were refined during the 'wand' calibration during which a rod containing three markers at known locations was offered around the measurement volume. The system was calibrated and eight trials were collected at 120 Hz for a duration of 4 seconds with the device positioned at six points in the measurement volume. Both static and

dynamic conditions were collected. In position one the testing device was placed on the floor of the laboratory with its origin approximately over the global origin and the long axis of the base of the device aligned with the global x axis. In positions two and three, the orientation of the testing device was the same as in position one but the origin of the device was located 1.15 m and -1.35 m away from the global origin, along the x axis of the laboratory respectively. Positions four, five and six were a repeat of positions one, two and three respectively but the device was placed on a table of height 0.86 m. Placing the device on a table of this height meant that the markers at the greatest vertical height were 1.26 m above the origin of the lab. All positions were at the same location on the laboratory z axis.

3.2.3 Data analysis

Using the raw coordinate data, the accuracy and precision of the motion analysis system was determined by calculating the root-mean-square error and standard deviation, respectively, for the following variables in each position and in both the static and dynamic conditions:

1. **Long:** The absolute distance between m1 and m4, calculated as the magnitude of the vector **m1m4** (known distance = 400 mm).
2. **Short:** The absolute distance between m7 and m8, calculated as the magnitude of the vector (known distance = 80 mm).
3. **Three-point angle:** The angle between arm_1 and arm_2, calculated as the absolute angle between **m1m2** and **m1m4** (known angle = 90°).

4. **Cardan angle:** The angle between arm_1 and arm_2 was also calculated using the Cardan convention. Local coordinate systems were defined for both arms. Marker 1 was defined as the origin of both coordinate systems. The x-axis of each arm coordinate system was coincident with the long axis of each arm respectively. The y-axes were orthogonal to the respective x axes and pointed vertically upward. Furthermore, the z-axis of each arm coordinate system was defined as the cross product of the x and y axis unit vectors of each local coordinate system – the direction of the z-axis obeyed the right-hand rule. The angles between arm 1 and arm 2 were then calculated using a y-z-x rotation sequence. Rotations about the y-axis (known angle of 90°) were recorded and used in the analysis.

These data were then averaged across the eight trials to give a mean root-mean-square and standard deviation value of the variable at each position in the laboratory.

3.2.4 Results

The values for root-mean-square difference and standard deviation in the static and dynamic conditions are given in Table 3.1 and Table 3.2 respectively. Also presented in Table 3.2 are the standard deviation values for seven different motion capture systems that were provided by Richards (1999).

Table 3.1: Mean root-mean-square error (RMS) and standard deviation (SD) for the four dependent variables in each position, averaged across eight trials in the static condition.

Position	Long (mm)		Short (mm)		3-Point Angle (°) Criterion = 90°		Cardan Angle (°) Criterion = 90°	
	RMS	SD	RMS	SD	RMS	SD	RMS	SD
1	0.17	0.05	0.44	0.10	0.04	0.01	0.04	0.01
2	0.05	0.05	1.05	0.21	0.07	0.01	0.07	0.01
3	0.59	0.07	1.19	0.13	0.04	0.01	0.04	0.01
4	0.24	0.04	1.19	0.13	0.04	0.01	0.04	0.01
5	0.37	0.05	0.59	0.05	0.07	0.01	0.07	0.01
6	0.53	0.08	0.94	0.30	0.02	0.01	0.02	0.01
Mean	0.32	0.16	0.90	0.16	0.05	0.01	0.05	0.01

Table 3.2: Top: Mean root-mean-square error (RMS) and standard deviation (SD) for the four dependent variables in each position, averaged across eight trials in the dynamic condition. Bottom: Standard deviation data presented by Richards (1999) for the long, short and three-point angle variables

Position	Long (mm)		Short (mm)		3-Point Angle (°)		Cardan Angle (°)	
	RMS	Criterion = 400 mm SD	RMS	Criterion = 80 mm SD	RMS	Criterion = 90° SD	RMS	Criterion = 90° SD
1	0.58	0.41	0.74	0.46	0.12	0.11	0.12	0.11
2	0.67	0.47	0.99	0.70	0.12	0.12	0.12	0.12
3	0.85	0.43	0.95	0.62	0.10	0.10	0.10	0.10
4	0.62	0.42	1.58	0.92	0.11	0.11	0.11	0.11
5	0.62	0.34	4.54	4.18	0.10	0.09	0.10	0.09
6	0.71	0.48	1.70	1.35	0.12	0.11	0.12	0.11
Mean	0.67	0.43	1.75	1.37	0.11	0.11	0.11	0.11

<u>Richards' (1999) data</u>								
Criterion = 500 mm		Criterion = 90 mm		Criterion = 95.8°		Criterion = 95.8°		
	RMS	SD	RMS	SD	RMS	SD	RMS	SD
Ariel	4.27		1.51		2.11			
CODA	4.87		2.25		3.39			
Elite	1.73		4.46		4.29			
Motion	0.59		1.49		1.76			
Peak	0.91		1.77		3.77			
Qualisys	0.80		2.21		4.50			
Vicon	0.62		1.29		1.42			

3.2.5 Discussion

It can be seen from the results presented in Table 3.1 and Table 3.2 that the measures of RMS and standard deviation varied across the six positions in the lab. This might indicate that an incomplete picture of the accuracy or precision of a motion analysis device is obtained from only one position in the measurement volume, as has been reported in the past (Richards, 1999; Kadaba and Stine, 2000). Furthermore, the disparity between RMS and standard deviation values taken at different positions in the measurement volume might be especially relevant when the volume is quite large. It would seem appropriate to take several measurements at different positions in the volume and calculate an average value to describe the accuracy and precision of the system set up for that volume size. For this reason only the averages across the six positions in the volume will be discussed further here.

As would be expected for all variables, the root-mean-square error and standard deviation values were greater in the dynamic than the static conditions. This is similar to the results reported by Kadaba and Stein (2000), where static measurements of a known distance were closer to the actual distance than the dynamic estimates. However, most analyses in sport and exercise biomechanics involve movement. Therefore, the dynamic results provide a better representation of these activities and, therefore, only dynamic accuracy and precision will be discussed further.

The dynamic results obtained in the present study of a Motion Analysis Corporation (MAC) image-based motion analysis device appear similar to those previously reported by Richards (1999). For the long distance (400 mm), a root-mean-square error of 0.67 mm was obtained, which was slightly greater than the value of 0.59 mm reported by Richards (1999) for a similar distance (500 mm). The value of 1.80 mm for the RMS for the short distance (80 mm) in this study was also slightly higher than the value of 1.49 mm presented by Richards (1999). It can be seen that the RMSs for the larger distance of 400 mm are smaller than for the lesser distance of 80 mm. Sometimes problems occurred in the tracking of the markers that made up the smaller distance, m7 and m8, due to marker occlusion by the covered m9. Consequently, in some positions in the volume, a high root-mean-square-error was observed, inflating the mean value.

A potential reason for the greater RMS values in the present study than those presented by Richards (1999) is that Richards' measure of root-mean-square difference did not include a criterion or known value and was actually a measure of standard deviation. It can be seen (Table 3.1 and Table 3.2) that standard deviation values for all variables were smaller than the RMSs in this study. Richards (1999) actually measured the precision and not the accuracy of the systems. Another possible reason could be that the measurement volume was larger in the present study. A larger measurement volume can have the effect of decreasing the accuracy and precision of the reconstructed marker coordinates because the resolution of the system is reduced.

The RMSs and standard deviations for the known 90° angle in the present study, whether calculated using the three-point or Cardan methods, are smaller than those presented by Richards (1999). An average root-mean-square-error of 0.11° was apparent in this study as opposed to a value of 1.76° presented by Richards (1999). This could be because the sides of the triangle of markers used to construct the angles were much smaller in the Richards (1999) study than in the present study. Therefore, any minor variations in the reconstructed marker position would have a large effect on the measured angle. However, in this study, the angle was constructed from two ‘segments’, with lengths similar to those of lower extremity body segments. Therefore, the angle included in this study better simulates angles measured between lower extremity body segments during measurement of coordination variability.

Notwithstanding the larger measurement volume used in this programme of research compared to those used by Richards (1999), comparable RMS and standard deviation values were obtained. Therefore, the system exhibited very good accuracy and precision; the instrumentation errors were small. Certainly, instrumentation errors are a great deal smaller than the magnitude of the errors due to skin movement – see section 2.6.3.

3.3 Skin movement errors

When markers are directly applied to the skin, the cluster of markers used to determine the pose of bone may undergo deformation, i.e. the inter-marker

distances change. This deformation will result in errors in the reconstruction of the local anatomical reference frame. Also, the markers might move relative to the bone, resulting in erroneous alignment of the local frame with the underlying bone. Nigg and Herzog (1999) classified these sources of error as relative and absolute marker error respectively.

Many methods have been used in an attempt to reduce skin movement errors, when the rigid body assumption has been violated. Some studies have taken the highly invasive approach of using bone pins to negate the need to place markers on skin and attach them directly to bone creating a rigid cluster of markers (e.g. Holden *et al.*, 1997; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b; Manal *et al.*, 2000). However, these techniques are highly invasive and are not generally used (Cappozzo, *et al.*, 1996).

Mathematical algorithms have also been proposed as a technique for reducing relative marker movement errors (e.g. Woltring *et al.*, 1985; Veldpaus *et al.*, 1988). Some of these techniques require the use of smoothing algorithms, which are based on the assumption that noise is additive and random, with a mean value of zero (white noise) (Woltring *et al.*, 1985). However, Nigg and Herzog (1999) suggested that errors due to skin marker movement may not have zero means, which led them to conclude that mathematical smoothing algorithms are inappropriate approaches for removing the error due to marker movement. This is reiterated by Cappozzo *et al.* (1996) who suggested that skin marker movement has the same frequency content as the bone absolute movement, and there is no

way of separating the artefact from the actual bone movement by using a simple filter. Many highly complex algorithms have been developed to try to compensate for skin movement errors and they have recently been reviewed by Leardini *et al.* (2004). However, these techniques are complex, computationally intensive and not used routinely. Therefore, for these reasons, they are not discussed here and were not used in this programme of research.

One approach to reducing relative marker error is to apply the markers to the skin using a rigid fixture (e.g. Ronsky and Nigg, 1992; Manal *et al.*, 2000; Manal *et al.*, 2002). There are fewer errors due to deformation because the marker mounting is rigid. However, the absolute error, or movement of the marker cluster as a whole relative to the underlying bone, remains; which is the case with all techniques used to reduce relative error. The techniques used most frequently to quantify the movement between a marker cluster and the underlying bone are invasive bone pin experiments (e.g. Holden *et al.*, 1997; Reinschmidt *et al.*, 1997a; Reinschmidt *et al.*, 1997b; Reinschmidt *et al.*, 1997c; Manal *et al.*, 2000). Other techniques such as video fluoroscopy have also been used (e.g. Baltzopoulos, 1995; Cappozzo *et al.*, 1996) but, again, ethical issues associated with exposing participants to radiation are inherent in these analyses.

In section 2.6.3 of this thesis, a detailed review of studies which have attempted to quantify the errors due to skin marker movement was provided. In summary, this review highlighted that, in comparison to the instrument errors reported in section 3.2, errors due to skin marker movement are large. For example, Cappozzo *et al.*

(1996) suggested that knee kinematics can be affected by inaccuracies which amount to 10% of flexion-extension, 50 % of abduction-adduction and 100% of internal–external rotation.

3.4 Implications for studies of coordination variability

Inaccuracies when using markers placed on the skin to estimate the position and orientation of underlying bone, in the process of calculating angular kinematics for example, are due to instrument errors and skin movement errors (Cappozzo *et al.*, 1996). From the information presented in this Chapter in addition to the literature reviewed in section 2.6.3, it is clearly apparent that skin movement errors are far greater than instrument errors. In the study presented in section 3.2, errors in the calculation of angles due to the measurement system were about 0.1°. In comparison to instrument errors, the magnitude of skin movement errors appears to be very large. For example, errors due to skin marker movement during plantar flexion-dorsiflexion in running have been reported to be approximately 5° (Reinschmidt *et al.*, 1997a). Also skin movement errors increase in magnitude when the secondary planes of motion (frontal and transverse planes) are analysed (e.g. Cappozzo *et al.*, 1996). This suggests that skin movement errors are likely to have a much greater influence on studies of coordination variability than instrument errors. In fact, when measurement devices with similar resolution and accuracy to the system used in this programme of research are used, instrument errors might be considered negligible when compared to errors due to skin movement.

Skin movement errors could be extremely detrimental to the study of variability if they were completely random. It would be very difficult to conclude with certainty that any observed variability was due to the inherent natural human variability of interest or whether it was solely, or partly, caused by the inaccuracies due to skin movement errors. For example, if the errors presented by Reinschmidt *et al.* (1997a) were entirely random, a study reporting plantar flexion-dorsiflexion variability values of less than 5° could not conclude with certainty that the variability was the inherent variability of the motor system or solely due to error. Conversely, if the inaccuracies due to skin movement were entirely systematic, errors would be equivalent on each repetition. This would mean that the errors due to skin movement would not affect the observed variability. Furthermore, the researcher could be confident that the observed variability was due to the inherent variability of the participant. Subsequently, the effect of manipulating an independent variable on variability could be assessed with confidence.

Leardini *et al.* (2004) suggested that skin movement errors are reproducible within, but not among, participants and that skin movement introduces systematic as well as random errors. Additionally, qualitative assessment of graphs presented by Reinschmidt *et al.* (1997c) indicates that the errors have a systematic component – see Figure 3.2. Furthermore, Reinschmidt *et al.* (1997c) suggested that there is evidence that skin based marker systems systematically over-estimate the motion of the underlying bone at the ankle joint complex. Unfortunately, no study has addressed the issue of whether skin movement errors are systematic or random using appropriate statistical techniques. Examples of such techniques include

Bland and Altman's (1986) limits of agreement, and least products regression (Ludbrook, 1997). A study that investigates skin movement error using these or similar techniques is certainly warranted. Regrettably, such a study was not possible in this programme of research, due to the invasive nature of such experiments.

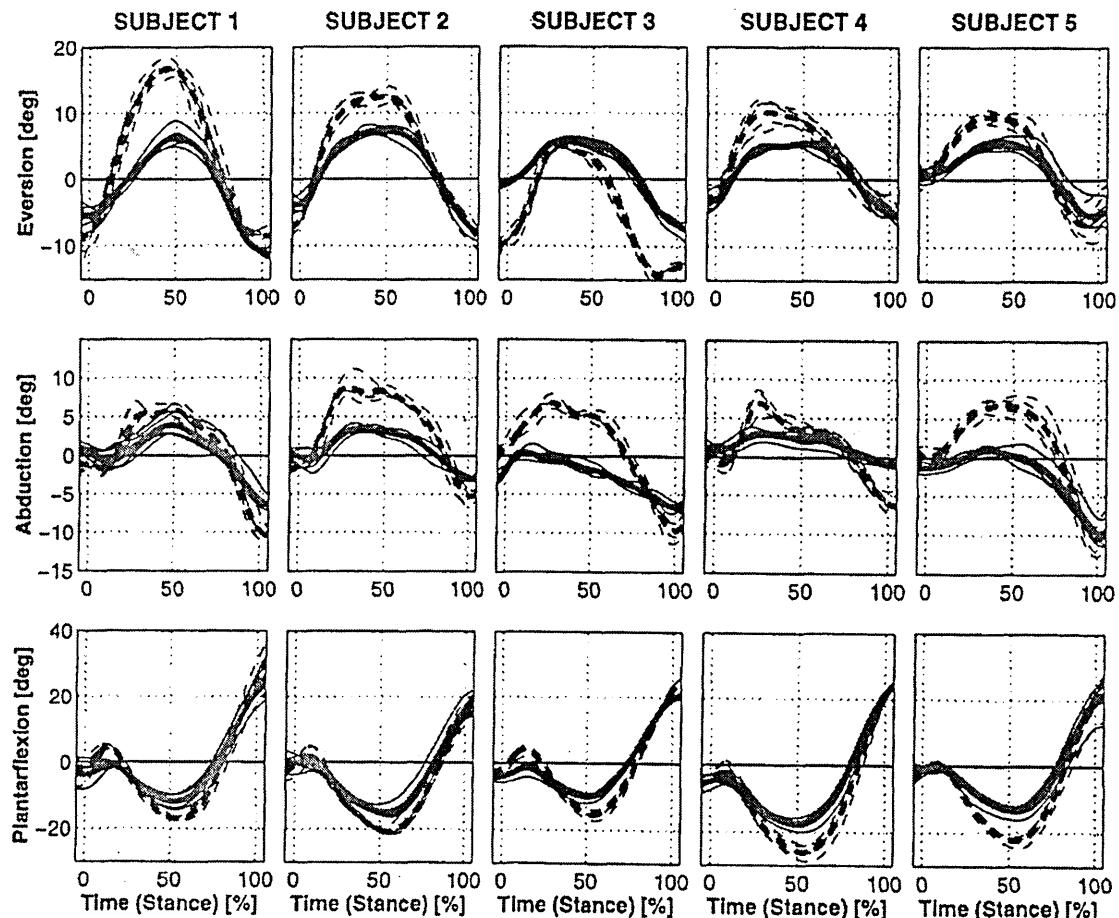


Figure 3.2: Tibiocalcaneal angles during the stance phase of walking for five trials of five participants. Solid lines represent kinematics based on bone pins and dashed lines represent kinematics based on skin markers - adapted from Reinschmidt *et al.* (1997c).

However, it is possible to perform a limits of agreement analysis on data presented by Holden *et al.* (1997). These authors presented the means (and standard deviations) of the differences between the position of the knee joint centre determined using skin and bone mounted markers. These data – means and standard deviation of the differences between measurement techniques - are needed for the calculation of limits of agreement (Bland and Altman, 1986). As described by Bland and Altman (1986) the limits of agreement are defined by the systematic error (mean of the differences between measurement techniques) \pm the random error, such that the limits of agreement are given by: mean \pm 1.96*standard deviation, of the differences. The limits of agreement between the position of the knee joint centre determined using skin and bone mounted markers, from the data presented by Holden *et al.* (1997), are given in Table 3.3 below.

Table 3.3: Limits of agreement between the position of the knee joint centre determined using skin and bone mounted markers, from the data presented by Holden *et al.* (1997). Data are presented for the medio-lateral, anterior-posterior and inferior-superior directions (systematic \pm random error).

Participant	Medio-lateral (mm)	Anterior-posterior (mm)	Inferior-superior (mm)
A	8.4 \pm 2.74	10.2 \pm 2.94	4.9 \pm 0.59
B	9.0 \pm 1.96	10.0 \pm 2.94	10.5 \pm 1.18
C	5.7 \pm 2.94	5.3 \pm 0.98	6.7 \pm 0.59

For all participants and directions the systematic component of the error was larger than the random component. The results of the crude analysis of data presented by Holden *et al.* (1997) suggest that, in the main, as was noted through qualitative

examination of the data presented by Reinschmidt *et al.* (1997a), the errors due to skin marker movement are systematic within participants. However, these results should be treated with caution because it was not possible to determine whether the data presented by Holden *et al.* (1997) met the assumptions of a limits of agreement analysis; most notably, normal distribution of the errors and heteroscedasticity – see Atkinson and Neville (1998). Therefore, there is a clear need for a well designed study employing appropriate statistical techniques, e.g. limits of agreement and least products regression, to ascertain whether the errors due to skin marker movement are systematic or random.

Previous literature – see section 2.6.3 - has highlighted that markers attached to the skin using rigid marker clusters described by Cappozzo *et al.* (1996) and Manal *et al.* (2000) most effectively minimise the errors due to skin movement. Furthermore, the attachment technique described as 'optimal' by Manal *et al.* (2000), appears to be most effective. This technique involves attaching a pre-moulded, Velcro™-backed thermoplastic shell to an under-wrapped elastic band placed on the distal-lateral portion of a segment. The distal-lateral location of the cluster of markers using this technique concurs with Cappozzo *et al.*'s (1996) suggestion and the design criteria presented by Cappozzo *et al.* (1997). This method of attachment will be used in subsequent chapters of this thesis to provide the most accurate available representation of bone pose.

3.5 Conclusion

In conclusion to this Chapter, it seems clear that errors in the estimation of bone pose which are due to skin marker movement are greater than those due to the instrument. The large magnitude of these errors – especially in the frontal and transverse planes of motion – would be especially detrimental to studies of coordination variability if they were entirely random. Results of a crude analysis of data presented by Holden *et al.* (1997) using Bland and Altman's Limits of Agreement technique suggested that errors due to skin marker movement were, primarily, systematic. Therefore, errors due to skin marker movement appear to be less problematic for studies of coordination variability. However, further work is required to substantiate the preliminary findings presented in this Chapter.

CHAPTER IV

4 Techniques for measuring coordination and coordination variability

Traditionally in biomechanics, data from isolated joints are presented as a function of time (displacement, velocity, angle, force etc.) with much research focusing on the magnitude and relative timing of discrete events. However, literature reviewed in section 2.4 of this thesis has implicated the coordination or coupling relationships between segments as being important for study. Furthermore, literature reviewed throughout Chapter II suggested a functional role for variability in human movement. As highlighted in Chapter II, there are several proposed functions of variability, each with many, potentially different, methodological issues to consider. To provide a clear direction for this thesis, the proposed relationship between variability and overuse injury (*c.f.* James, 2004) was taken as a context upon which to focus. Further, as both the coordination between body segments - see section 2.6.2 - and movement variability have recently been proposed as important in understanding the aetiology of injury, this thesis specifically focuses on the measurement of coordination variability during locomotion.

In section 2.6.2, issues related to the quantification of coordination and coordination variability were outlined. Amongst other issues, it was suggested that the choice of technique used to quantify coordination and coordination variability was important and could have implications for the interpretation of the results of a given research study. In the following sections of this Chapter the results of an

investigation of the effect of an unstable shoe construction on coordination variability are provided. By focusing on main effects and descriptive statistics, the aim of providing the experimental data was to highlight the potential for researchers to obtain disparate results depending on the coordination quantification technique used. A brief introduction to the investigation is given. Subsequently, methods and results sections are reported and the Chapter concludes with a discussion. However, it was not the purpose of this Chapter to provide a detailed discussion of the results of the investigation e.g. implications, limitations, future directions etc. Instead, the discussion section (section 4.4) focuses on addressing reasons for, and implications of, obtaining disparate results from different coordination quantification techniques.

4.1 Introduction

Stability is an important aspect of human walking and many shoes are typically constructed in order to provide stability (Nigg *et al.*, 2004). However, as Nigg *et al.* (2004) highlighted, it is reasonable to assume that a result of wearing highly stable shoes might be that the muscles that contribute to dynamic stability during locomotion get weaker over time due to reduced recruitment. Often these under-used muscles are trained using unstable situations, e.g. wobble boards, and examples of the efficacy of the use of such activities can be found in the literature (e.g. Caraffa *et al.*, 1996; Waddington and Adams, 2004). Obviously, however, devices such as the wobble board train the dynamic stability muscles in static conditions. Masai Barefoot Technologies (MBT) has recently developed a device

which, potentially, allows training of the dynamic stability muscles during a dynamic activity such as walking. The shoe developed by MBT has a rounded sole in the anterior-posterior direction (Figure 4.1) creating an unstable base. Because of this unstable base, walking in MBT shoes should elicit greater coordination variability than walking in normal shoes.



Figure 4.1: The MBT shoe with its curved sole in the anterior-posterior direction

There is also evidence to suggest that differences in coordination variability between the normal and MBT shoes might change over the stance period. For example, Heiderscheit *et al.* (2002) suggested that variability should be calculated over various phases of the stride cycle as the change in functional demands on the lower extremity over different periods is likely to affect the observed variability in coordination. This was also advocated by Holt *et al.* (1995) who presented data suggesting that variability in the coordination between body segments of the lower extremity changes during different phases of the stride cycle.

4.2 Methods

Fourteen participants (7 male, 7 female) volunteered to take part in the study. Their age, stature and body mass were (mean \pm s) 26.5 ± 4 years, 1.71 ± 0.06 m and 71 ± 10 kg respectively. All participants were physically active and free from injury at the time of testing. The local ethics committee approved the procedures, and written informed consent was gained from each participant before data collection – see Appendix E for copies of the informed consent form and participant information sheet. On volunteering to take part in the study, each participant was provided with a pair of Masai Barefoot Technology (MBT) shoes which were worn during the data collection sessions in the MBT condition in addition to tight-fitting shorts and vests. The MBT shoe has a curved sole in the anterior-posterior direction which provides an unstable base (see Figure 4.1). In the normal shoes condition participants were required to wear the same tight-fitting clothing and their normal exercise shoes.

All kinematic data were collected using an eight-camera digital motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 120 Hz. The eight cameras of the motion capture system were placed in optimal positions around a calibrated measurement volume of dimensions $4.4 \times 2.1 \times 1.8$ m in the anterior-posterior, vertical and medio-lateral directions respectively. Ground reaction force data were also collected using a Kistler Type 9281CA force platform (Kistler Instrumente AG Winterthur, Switzerland) which was embedded in the laboratory floor and covered with a surface common to the entire laboratory. The force platform sampled data at 1200 Hz and was time-synchronised with the motion capture system. Infra-red timing (Brower Timing Systems) gates were

placed 3 m apart either side of the floor-mounted force platform to monitor walking speed during the trials.

Twenty-two retro-reflective markers (12.5 mm diameter) were attached to each participant in locations consistent with the Helen Hayes Marker Set (Kadaba *et al.*, 1990). Markers were attached to the participants' left and right 2nd metatarsal head (closest approximation on the shoe), posterior aspect of the calcaneus (closest approximation on the shoe), lateral malleolus, lateral epicondyle, anterior superior iliac spines, acromion processes, lateral humeral epicondyles and midpoints of the radius and ulnar styloid processes. Further markers were attached to the sacrum, the seventh cervical vertebrae and the right scapula. In addition to these markers, four supplementary markers - mounted on 100 mm posts, were attached to the lateral aspect of each participant's left and right shank and thigh.

Before testing, participants were required to attend a tutorial session in which they were provided with their MBT shoes and instructions on their use. Participants were then asked to endeavour to use the shoes as frequently as possible in the interval between the tutorial and data collection (approximately five days).

After preparation and attachment of the markers, each participant was required to traverse the laboratory, approximately 16 m in length, at their preferred speed while making contact with the force platform with their right foot. The participants completed five 'good' trials in both the MBT and normal shoe conditions. The order in which the participants completed the conditions was randomised. Trials were

accepted when the whole of the participant's right foot contacted the force platform, without any obvious alterations to their gait. Participants were permitted as many practice trials as they required to become able to consistently achieve this prior to the onset of data collection. Kinematic data were collected for five seconds using the motion capture system along with the kinetic data from the force platform for every trial. Participants were required to walk at their preferred speed and this was measured during every trial using the infrared timing gates.

At the end of the data collection session a further static calibration trial was performed to allow correct anatomical reference frame alignment. Additional markers were attached to each participant's left and right medial malleolus and medial femoral epicondyles. Kinematic data were collected for 3 seconds with the participant in the anatomical position.

The three-dimensional coordinate data were filtered using a fourth order low-pass Butterworth filter (6 Hz: Antonsson and Mann, 1985). Right knee flexion-extension and ankle plantar flexion-dorsiflexion joint coordinate system (JCS) angles (Grood and Suntay, 1983) were then calculated using Orthotrak software (Motion Analysis Corporation, Santa Rosa, CA, USA). The resulting angular displacement profiles for each trial were then cropped to the duration of one foot contact – foot-strike to toe-off. The vertical component of the ground reaction force was used to determine foot contact events – thresholds of 20 N and 10 N were used to determine footstrike and toe-off respectively. The angular displacement profiles were then numerically differentiated to obtain angular velocity values. Subsequently, the

cropped profiles were interpolated to 101 data points using a cubic spline procedure, where 0 represented foot-strike and 100 represented toe-off.

The coordination between ankle plantar flexion-dorsiflexion and knee flexion-extension was quantified using four techniques: standard deviation in continuous relative phase, the coefficient of correspondence from Vector Coding (Tepavac and Field-Fote, 2001), the standard deviation in relative motion angles (*c.f.* Heiderscheit *et al.*, 2002) and the modified NoRMS (mNoRMS) technique – see Appendix B for details of how these parameters are calculated. Values of coordination variability at each data point were then averaged across the entire stance or stride phase. This gave an indication of the coordination variability over the duration of the activities using the four different techniques. Additionally, due to the changing functional demand placed on the body during different periods of stance (Heiderscheit *et al.*, 2002), coordination variability was assessed over four quarters of the stance phase.

Differences in coordination variability – quantified using continuous relative phase standard deviation, the coefficient of correspondence, standard deviation in relative motion angles and mNoRMS - between the MBT and normal shoe conditions were assessed using a series of two-factor (condition, interval) analyses of variance (ANOVA), with repeated measures on both factors. The alpha level of significance was set at 0.05. The aim of Chapter IV was to assess the effects of using four different techniques for quantifying coordination variability to address the same research question. The analysis of main effects and interactions was deemed

adequate to address this aim. Therefore, only the interaction effects and main effects for the condition and interval factors and between the independent variables and are presented and discussed.

4.3 Results

Figure 4.2 displays the knee flexion-extension, ankle plantar flexion-dorsiflexion angle-angle diagram for a representative participant. Visual inspection of the diagram suggests that walking in MBT shoes elicited greater coordination variability than walking in normal shoes. However, as can be seen in Table 4.1, only the NoRMS, relative motion and vector coding techniques elicited a significant main effect for the type of shoe factor. With these techniques, coordination variability was significantly higher during locomotion in the unstable shoes than the normal shoes.

All techniques, with the exception of mNoRMS, elicited a significant main effect for the interval factor. Furthermore, only the relative motion angles technique exhibited a significant interaction between independent variables. Figure 4.3 represents the coordination variability calculated using each of the techniques in both unstable and normal shoes across each phase of stance.

Table 4.1: Results of the analyses of variance comparing main effects for the type of shoe and of the intervals of the walking cycle for the continuous relative phase standard deviation (CRPsd), mNoRMS, relative motion angle standard deviation (RMsd) and coefficient of correspondence (CC) techniques.

	Shoe Main		Interval Main		Interaction	
	Effect		Effect			
	F	p	F	P	F	p
CRPsd	0.49	0.499	64.09	<0.001	1.31	0.287
mNoRMS	19.66	0.001	2.72	0.059	1.22	0.317
RMsd	19.23	0.001	44.93	<0.001	22.87	0.002
CC	55.42	<0.001	25.05	<0.001	4.33	0.051

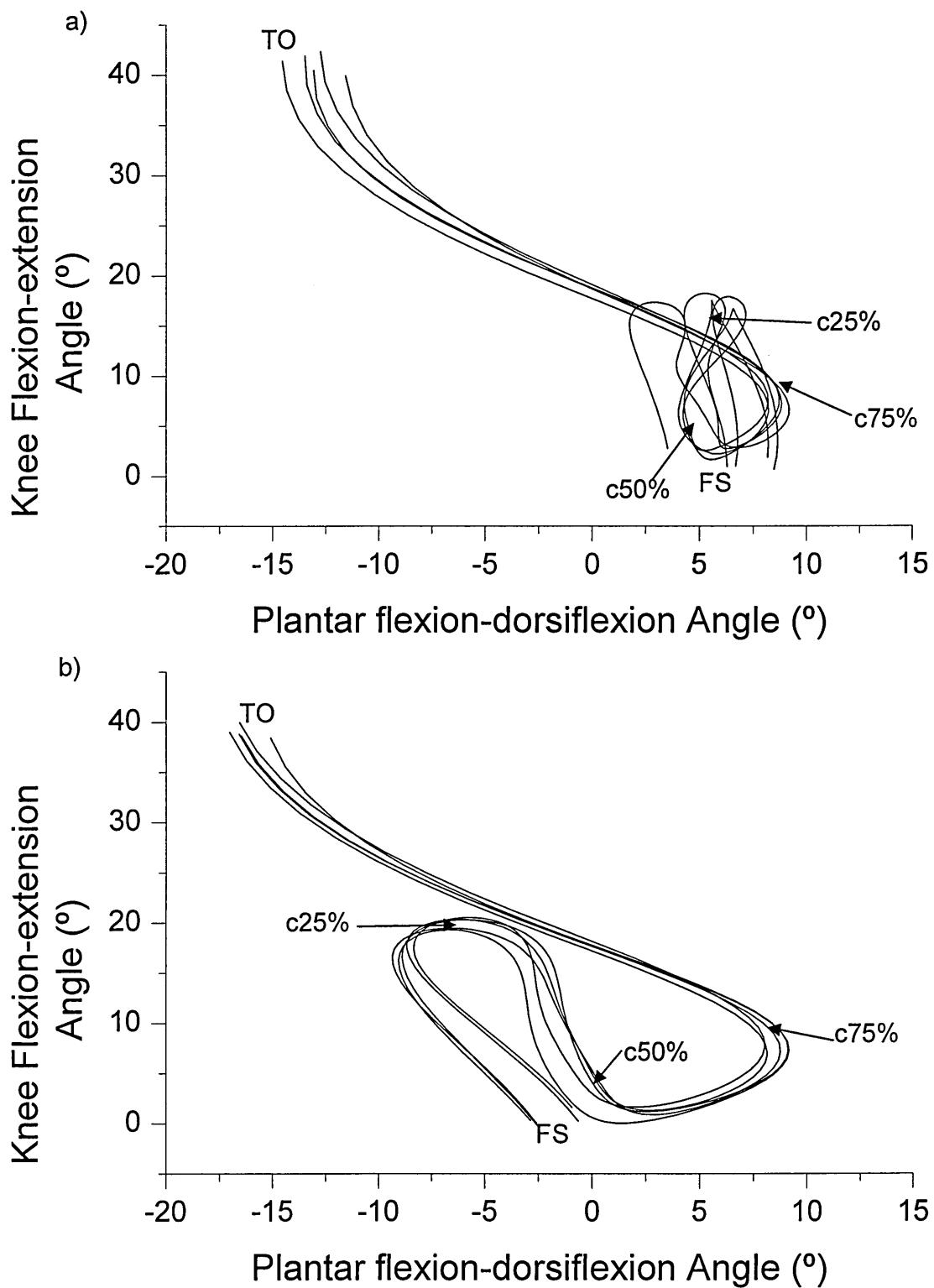


Figure 4.2: Angle-angle plot of knee flexion-extension against plantar flexion-dorsiflexion during the stance phase of walking in unstable (a) and normal (b) shoes for a representative participant. FS = footstrike, TO = toe-off.

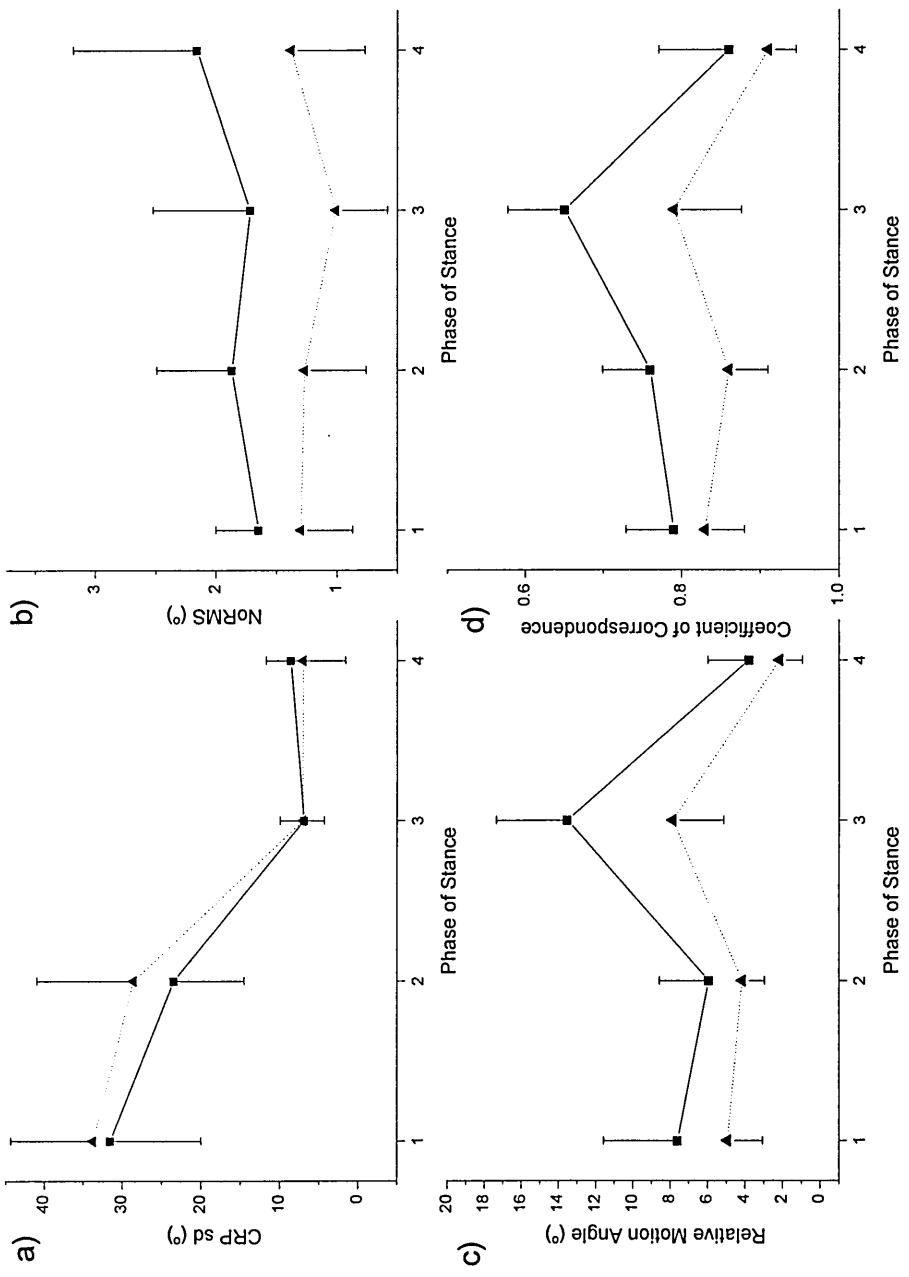


Figure 4.3: Coordination variability measured using a) continuous relative phase standard deviation, b) mNoRMS, c) relative motion angle standard deviation and d) the coefficient of correspondence in both unstable (solid line with squares) and normal shoes (dotted line with triangles). Phase of Stance 1, 2, 3 and 4 are 0-25%, 26-50%, 51-75% and 76-100% of the stance phase respectively

4.4 Discussion

The data presented in this Chapter suggests that contrasting results might be obtained by researchers investigating coordination variability dependent on their choice of quantification technique. It appears that different findings could be obtained in studies of locomotion when different techniques for quantifying coordination variability are used. In this section, potential reasons for the differences observed between quantification techniques in this Chapter are highlighted. Also, the implications of the findings for studies of coordination variability during locomotion are discussed.

It is widely recognised that the techniques available for quantifying coordination variability are associated with benefits and limitations (e.g. Hamill *et al.*, 2000; Kurz and Stergiou, 2002, Peters *et al.*, 2003; van Emmerik *et al.*, 2004). Further, some of the techniques used in this Chapter also require that the data meet certain assumptions. The advantages, disadvantages and assumptions of each technique were discussed in section 2.6.2. In order to highlight the potential for the choice of quantification technique to influence the results of a study of coordination variability, example experimental data were included in this Chapter. Specifically, continuous relative phase, relative motion angles, Vector Coding and mNoRMS were used to assess the difference in coordination variability between walking in normal and MBT shoes.

Inspection of the knee flexion-ankle dorsiflexion angle-angle diagram (Figure 4.3) suggests that walking in MBT shoes was associated with greater coordination

variability than walking in normal shoes. This qualitative finding is consistent with the assumption that, due to the relative instability of the shoe construction, walking in MBT shoes would be associated with increased variability. Relative motion angles, mNoRMS and Vector Coding indicated a significant main effect for the shoe factor which corroborates the qualitative findings. However, for the same factor, continuous relative phase suggested a non-significant main effect. A potential reason for the disparity between measurement techniques is related to small differences in the specific detail of how each technique quantifies coordination. A major problem with the use of continuous relative phase in the context of the investigation reported in this Chapter is that the data did not meet the assumption of the joint motions having a one-to-one sinusoidal time history. With the exception of hip flexion-extension, the joint motions of the lower extremity during locomotion are inappropriate for continuous relative phase analysis. They are inappropriate because, generally, joint motions during gait, although periodic, contain more than one frequency component. The multiple frequency components in the joint motions increase the likelihood of the data violating the assumptions of continuous relative phase. Many authors have highlighted that, if the data do not meet the assumptions of sinusoidal time history and one-to-one frequency ratio, artefacts could be introduced into the measures of coordination (Fuchs and Kelso, 1994; Rosenblum and Kurths, 1998; Peters *et al.*, 2003). Indeed, the use of continuous relative phase as a measure of coordination and coordination variability during gait has been questioned by many authors (e.g. Heiderscheit *et al.*, 2002; Wheat *et al.*, 2002; Ferber *et al.*, 2005; van Emmerik *et al.*, 2004). Although most normalisation procedures centre the phase plot about the origin of the phase plane

- which can help in the calculation of continuous relative phase – some of these artefacts would be present regardless of the normalisation procedure used. It should also be noted that, continuous relative phase was originally used in studies of continuous, cyclical movements such as rhythmical bimanual coordination (*c.f.* Kelso, 1981, 1984, 1995). Joint motions during locomotion are cyclical but, often, as in the investigation presented in this Chapter, the stance phase is studied in isolation. Potentially, continuous relative phase might have produced results that were more consistent with the other quantification techniques if the entire stride had been included.

Although relative motion angles, Vector Coding and mNoRMS all indicated a significant main effect for the condition factor – suggesting that coordination variability, averaged over the entire stance phase, was greater in the normal condition than the MBT condition – visual inspection of Figure 4.3 suggests that the pattern of differences over the four quarters of the stance phase are dissimilar between techniques. Again, this can be explained by the small differences in the calculation of each technique. For example, as noted in section 2.6.2.2, a potential problem with relative motion angles is that they only give an indication of changes in the shape of the angle-angle trace and give no indication of changes in the ‘size’ of the trace – no measure of the variability in the magnitude of the point-to-point vectors is included. Because of this, important information might be neglected. An example how this might happen is given in Figure 4.4:

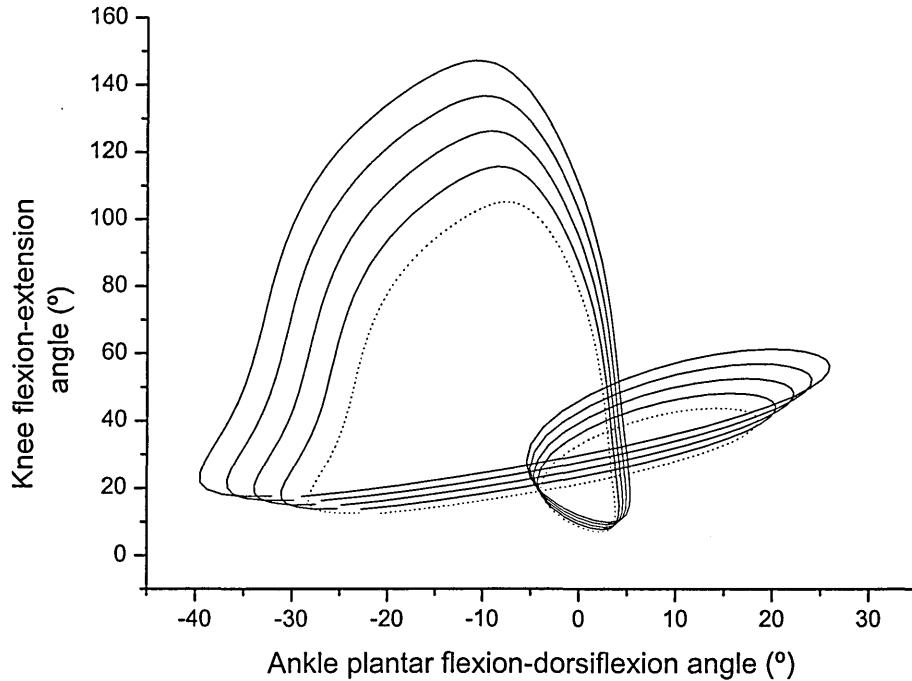


Figure 4.4: A representation of the effect increasing the magnitude of the joint angles represented in an angle-angle plot. The dotted line is taken from a participant during a treadmill running trial. The remaining four traces are a representation of the original increased by a factor of 1.1, 1.2, 1.3 and 1.4 respectively.

In Figure 4.4, the dotted line represents the original data for knee flexion-extension and ankle plantar flexion-dorsiflexion. For the remaining four (solid) lines the magnitude of both joint angles is greater than the original by a factor of 1.1, 1.2, 1.3 and 1.4 respectively. Qualitative assessment of the angle-angle trace over each ‘artificial’ stride suggests that there is evidence of variability in coordination. However, as relative motion angles only measure the change in shape of the trace, they would indicate that the trials were entirely consistent.

The issue outlined in Figure 4.4 is also a problem for the continuous relative phase technique. With the normalisation procedure used for continuous relative phase

calculation in this Chapter⁶, angular displacement is normalised to the maximum and minimum values on each individual stance phase. This particular normalisation procedure would have the same effect of neglecting changes in the magnitude of the joint motions. Other normalisation techniques could be used to address this issue (*c.f.* Hamill *et al.*, 2000).

Regardless of the reason for the differences between quantification techniques, it is clear that authors might draw contradictory conclusions about the differences in coordination variability between walking in normal and MBT shoes depending on their choice of technique. For example, if Vector Coding, relative motion angles or mNoRMS were used, an investigator would likely conclude that MBT shoes significantly effected coordination variability. Whereas, if continuous relative phase had been used, it would probably be concluded that MBT shoes had no effect on coordination variability.

As mentioned previously in this section, a potential reason for the disparity between measurement techniques lies in the details of their calculation. As another example of an important difference, relative motion angles, Vector Coding and continuous relative phase - calculated using the normalisation algorithms adopted in this thesis - give an indication of the variability in the shape of the traces on the angle-angle diagram, from cycle-to-cycle. Vector Coding also gives an indication of the variability in the size of the traces i.e. magnitude of the point-to-point vectors.

⁶ Which has been employed in previous studies (e.g. Hamill *et al.*, 1999)

MNoRMS gives an indication of both of these factors, in addition to the variability caused by the entire angle-angle trace 'shifting' in the angle-angle diagram on each cycle. This 'shift' of an angle-angle trace could be apparent even when the range-of-motion and relative timing of each joint motion was consistent from cycle-to-cycle. As an example of this 'shift', Figure 4.5 is an angle-angle diagram in which the angle-angle trace for each successive cycle is moved along the x- and y-axes by two degrees.

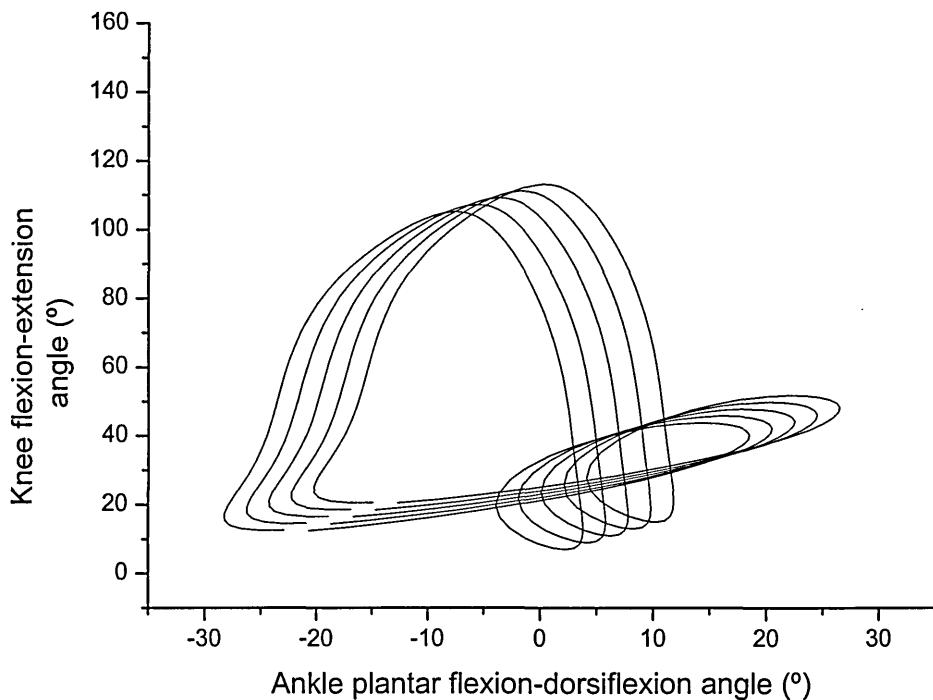


Figure 4.5: A representation of the entire angle-angle trace 'shifting' in the angle-angle diagram. The dotted line is taken from a participant during a treadmill running trial. The remaining four trials are the original ankle plantar flexion-dorsiflexion and knee flexion-extension angles plus 2°, 4°, 6° and 8° respectively.

As the range-of-motion and relative timing of the joint actions are identical over each successive cycle, relative motion angles, Vector Coding and continuous

relative phase (calculated using the normalisation method employed in this Chapter) would indicate that the trials in Figure 4.5 were entirely consistent. Clearly, however, the cycles in Figure 4.5 are not identical and each trace is in a different position on the x - and y -axes. Unlike the other techniques, mNoRMS takes account of this absolute 'shift' of the trace and would indicate variability. However, importantly, it could be argued that the coordination between knee flexion-extension and ankle plantar flexion-dorsiflexion is not affected by the 'shifting' of the traces seen in Figure 4.5. Indeed, the same argument could apply to the circumstances outlined in Figure 4.4. Certainly, some research questions might dictate that the variability in the angle-angle traces represented in Figure 4.4 and Figure 4.5 is not important. Conversely, in some circumstances, such as the investigation of the variability-overuse injury hypothesis, the variability caused by the 'shifting' of the angle-angle trace might be relevant.

Many of the techniques that are available for quantifying coordination and, arguably more importantly, coordination variability were reviewed in section 2.6.2 of this thesis. It was clear from this review that each technique is associated with various benefits and limitations. In this Chapter, experimental data were presented which highlighted that different techniques for quantifying coordination variability provide different answers to some research questions. These differences were suggested to be related to the small differences in the calculation of each technique. It seems clear that, in agreement with other authors (Fuchs and Kelso, 1994; Rosenblum and Kurths, 1998; Peters *et al.*, 2003), continuous relative phase should be used with caution in studies of coordination variability during locomotion.

It should be noted that, as reported in section 2.6.2.1.2, other variations of continuous relative phase are available, which address the issue of data not meeting the assumption of the data being sinusoidal and of a one-to-one frequency. Relative Fourier phase (Lamothe *et al.*, 2002) and continuous relative phase calculated using the Hilbert transform (Rosenblum and Kurths, 1998) are examples of such techniques.

It is also clear that comparisons between studies of coordination variability that have used different quantification techniques should be made with caution. Authors should make clear exactly what technique was used and how it was calculated, to enable an informed reader to make appropriate judgments about the study and compare the results to similar work. Finally, as Hamill *et al.* (2000) suggested, before choosing a particular technique, the researcher should be aware of the benefits and limitations of each and understand which is most suited to the movement or activity of interest. Furthermore, the choice of technique should be based on the research question of interest. For example, Ferber *et al.* (2005) recently conducted an investigation into the effect of foot orthoses on the coordination between rearfoot inversion-eversion and tibial internal-external rotation using relative motion angles. This was an appropriate choice of analysis technique because, first, continuous relative phase would have been compromised by the individual joint angle time histories not meeting the necessary assumptions. Second, mNoRMS and Vector Coding provide only an indication of coordination variability and not coordination which was also required. However, the context of the studies in this thesis is the future exploration of the variability-overuse injury

hypothesis. Although, it could be argued that the ‘shift’ of the angle-angle trace (see Figure 4.5) does not cause a change in coordination, in the context of the variability-overuse injury hypothesis the variability introduced by the ‘shift’ of the angle-angle traces could be, potentially, very important. Therefore, in addition to continuous relative phase, which is inappropriate because the data would not meet the sinusoidal assumption, relative motion angles and Vector Coding are also inappropriate. Consequently, the mNoRMS technique will be used as a measure of coordination variability in the remaining studies of this thesis. With the mNoRMS technique, a measure of the variability due to the ‘shifting’ of the angle-angle traces (see Figure 4.5) will be obtained, in addition to that caused by changes in the shape and magnitude of the traces.

4.4.1 Conclusion

This data presented in this Chapter highlighted differences between techniques previously used to quantify the variability in coordination between two body segments. These differences might lead authors to draw contradictory conclusions about coordination variability from two independent studies using different techniques. It is suggested that direct inter-study comparisons of coordination variability between studies incorporating different techniques should be made with caution. The choice of technique should be based on the research question of interest and researchers should be aware of the benefits and limitations of each technique and understand which is most suited to the movement or activity of interest.

CHAPTER V

5 Coordination variability during overground and treadmill running

5.1 Introduction

In studies of gait, treadmills are frequently used to simulate overground locomotion as they are often deemed to be more convenient. Space requirements are constrained, environmental factors can be controlled, and steady-state locomotion speeds are selectable (White *et al.*, 1998). Obviously, a problem of ecological validity exists if kinematic, kinetic, and metabolic variables, for example, differ between the two forms of locomotion. Nigg *et al.* (1995) suggested there is inconsistency in the literature about the similarity between overground and treadmill running. For example, whilst some studies have reported differences between kinematic and kinetic parameters during overground and treadmill locomotion (e.g. Sykes 1975; Elliot and Blanskby, 1976; White *et al.*, 1998; Schache *et al.*, 2001) others have reported no statistically significant differences (e.g. Murray *et al.*, 1985; Lemke *et al.*, 1995).

The investigations that reported no differences between the modes of locomotion are in agreement with a theoretical study conducted by van Ingen Schenau (1980). Van Ingen Schenau (1980) used a mathematical approach to suggest that the mechanics and energetics of overground and treadmill running are the same, if the speed of the treadmill belt is constant. Potential reasons for the reported

differences between the modes of locomotion include both non-mechanical factors, such as treadmill familiarity, and mechanical factors, such as intra-stride belt speed variations. In studies in which these factors have been controlled, treadmill locomotion was reported to effectively simulate overground locomotion. For example, Matsas *et al.* (2000) reported that knee joint kinematics during treadmill walking in young unimpaired participants who had undertaken a four-minute treadmill habituation session could be generalised to overground walking.

Recently, concepts and techniques from the dynamical systems approach to movement coordination and control have been applied to biomechanical research (see van Emmerik *et al.*, 2004, for a review). Examples of such research include investigations into the dynamics of the walk-to-run transition (Diedrich and Warren, 1995; Kao *et al.*, 2003), obstacle clearance during running (Stergiou *et al.*, 2001a,b) and the causes of patellofemoral pain (Hamill *et al.*, 1999; Heiderscheit *et al.*, 2002). A major tenet of the dynamical systems approach is that it affords a function to movement variability. This is in contrast to traditional paradigms, including information processing theory, in which variability is viewed negatively and seen as a noise which must be eliminated. Because of the numerous biomechanical degrees of freedom⁷ available during movement, proponents of the dynamical systems approach have argued that variability is an inherent component both within and between all biological systems (Newell and Corcos, 1993).

⁷ Biomechanical degrees of freedom are at the observable level e.g. segment rotations and translations as opposed to the active degrees of freedom (see Mitra, 1998) which capture the attractor dynamics of the movement system in the state space (van Emmerik *et al.*, 2005)

Variability has, for example, been seen as essential in inducing a coordination change (Kelso, 1984; Diedrich and Warren, 1995) and been hypothesised to be related to the aetiology of overuse injury (Hamill *et al.*, 1999; James, 2004). The idea that variability is functional is also gaining recognition in a wide variety of disciplines including cardiac dynamics (e.g. Goldberger *et al.*, 1990) and brain pathology (e.g. Stam *et al.*, 1994).

A related influence of the dynamical systems approach on some recent research into human locomotion is the assertion that the coordination or coupling between joints of the lower extremity is important. Running is a complex motor skill that involves many degrees of freedom. In order to produce coordinated movement and master the myriad of interacting components in the human body, the runner must solve what Bernstein (1967) termed the 'degrees of freedom problem'. Dynamical systems theorists have proposed that, to aid in managing the degrees of freedom in the system, synergies emerge between the different interacting components. These synergies are known as coordinative structures which are an important facet of the dynamical systems approach and allow for a solution to the degrees of freedom problem (Turvey, 1990). Traditionally in biomechanics, time series data from isolated joints are presented, with much research focusing on the magnitude and timing of discrete events. Some authors have recognised that these approaches do not effectively capture the complexity of the coordinated motions of components of the body. Therefore, recently, suitable techniques have been used to measure the coordinated actions of lower extremity joints during running

(Diedrich and Warren, 1995; Hamill *et al.*, 1999; Heiderscheit *et al.*, 1999; Heiderscheit *et al.*, 2002; Ferber *et al.*, 2005).

Investigations of human locomotion have also been reported which addressed both important facets (variability and coordination) of the dynamical systems approach. Examples of such studies include investigations into the relationship between coordination variability and joint pain (Hamill *et al.*, 1999; Heiderscheit, 2000; Heiderscheit *et al.*, 2002), the effect of orthoses on coordination and coordination variability (Ferber *et al.*, 2005), the dynamics of the walk-to-run transition (Diedrich and Warren, 1995; Kao *et al.*, 2003) and the relationship between spinal cord injury and coordination variability (Field-Fote and Tepavac, 2002). Some of these studies were conducted overground (Ferber *et al.*, 2005) while many involved treadmill locomotion (Diedrich and Warren, 1995; Hamill *et al.*, 1999; Heiderscheit, 2000; Field-Fote and Tepavac, 2002; Heiderscheit *et al.*, 2002; Kao *et al.*, 2003).

Clearly, a problem of ecological validity exists if differences in coordination and coordination variability exist between the two modes of locomotion. An assumption of the studies in which a treadmill was used was that treadmill locomotion effectively simulates overground locomotion in terms of coordination variability. The limited examples of studies that have compared variability in kinematics between overground and treadmill locomotion were reviewed in section 2.6.4.5. Briefly, Dingwell *et al.* (2001) compared sagittal plane ankle, knee and hip angles collected during overground and treadmill walking. Treadmill walking was associated with small but significant reductions in variability compared to overground walking

(Dingwell *et al.*, 2001). These results are in agreement with previous findings (Nelson *et al.*, 1972; Wank *et al.*, 1998). Both Nelson *et al.* (1972) and Wank *et al.* (1998) reported less variability in the horizontal and vertical velocities of the centre-of-mass while running on a treadmill as opposed to overground running. The study by Dingwell *et al.* (2001) is the only investigation that specifically compared the variability in angular kinematics of lower extremity joints. However, they studied joint motions in isolation while walking and no studies have compared coordination variability between overground and treadmill running. Therefore, the purpose of this study was to assess differences in the variability of lower extremity joint couplings between overground and treadmill running. Significantly reduced variability in lower extremity coordination in the treadmill condition was hypothesised. Specifically, the following research hypotheses were formulated:

H_1 – mNoRMS values for the hip flexion-knee flexion coupling, averaged over the entire stance period and various phases of stance, will be lower in the treadmill condition than the overground condition.

H_2 – mNoRMS values for the hip flexion-ankle dorsiflexion coupling, averaged over the entire stance period and various phases of stance, will be lower in the treadmill condition than the overground condition.

H_3 – mNoRMS values for the knee flexion-rearfoot inversion coupling, averaged over the entire stance period and various phases of stance, will be lower in the treadmill condition than the overground condition.

5.2 Methods

5.2.1 Participants

Thirteen male participants volunteered to take part in the study. *A priori* power calculations for repeated measures analysis of variance (Park and Schutz, 1999) based on pilot test data ($\alpha = 0.05$, $\beta = 0.20$) indicated that this number of participants gave the study sufficient power to detect differences between overground and treadmill conditions. Participants had an average ($\pm s$) age of 25.9 \pm 4.1 years, stature of 1.78 ± 0.09 m and body mass of 73.7 ± 8.4 kg. All participants exhibited a rear-foot striking pattern (determined through visual inspection), were experienced in treadmill running, physically active and free from injury at the time of testing. The University's Ethics Committee approved the procedures, and written informed consent was gained from each participant before data collection – see Appendix G for example copies of the participant information sheet and informed consent form. Participants were required to wear only their own running shoes and a pair of tight-fitting running shorts throughout testing.

5.2.2 Experimental set-up

All kinematic data were collected using an eight-digital camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 120 Hz. The orientation of the global coordinate system was such that the positive x axis pointed in the direction of forward progression, the positive y axis pointed vertically upward and the positive z axis pointed to the right. The eight cameras of the motion capture system were placed in optimal positions around a calibrated

measurement volume of dimensions $4.4 \times 1.6 \times 2.1$ m in the x, y and z directions respectively. The measurement volume was this size to incorporate the Kistler Gaitway treadmill and the Kistler Type 9281CA force platform (Kistler Instrumente AG Winterthur, Switzerland) embedded in the laboratory floor. The two force platforms built into the Gaitway treadmill and the force platform embedded in the ground collected ground reaction force data at 1200 Hz and were time-synchronised with the motion capture system. Infra-red timing gates (Brower Timing Systems, South Draper, UT, USA) were placed 3.8 m apart either side of the floor-mounted force platform to monitor running speed during the overground trials.

Pre-moulded, Velcro™-backed thermoplastic shells equipped with four 12.5 mm retro-reflective markers were attached to each participant's left shank and thigh. The markers were placed non-collinearly on the thermoplastic shells with inter-marker distances of greater than 100 mm. The thermoplastic shells were attached to the participant using a technique (Figure 5.1) describe as 'optimal' by Manal *et al.* (2000). The shells were attached in distal-lateral locations by fastening the Velcro™ to a securely fastened, under-wrapped elastic bandage (SuperWrap™, Fabriflora, Exton, PA, USA). Markers were also attached directly to the participant's left running shoe at the fifth metatarsal head, superior navicular, toe (second metatarsal head) and the most posterior aspect of the heel. Four further retro-reflective markers were attached to the participant's pelvis at the left and right anterior superior iliac spines, left iliac crest and left posterior superior iliac spine.

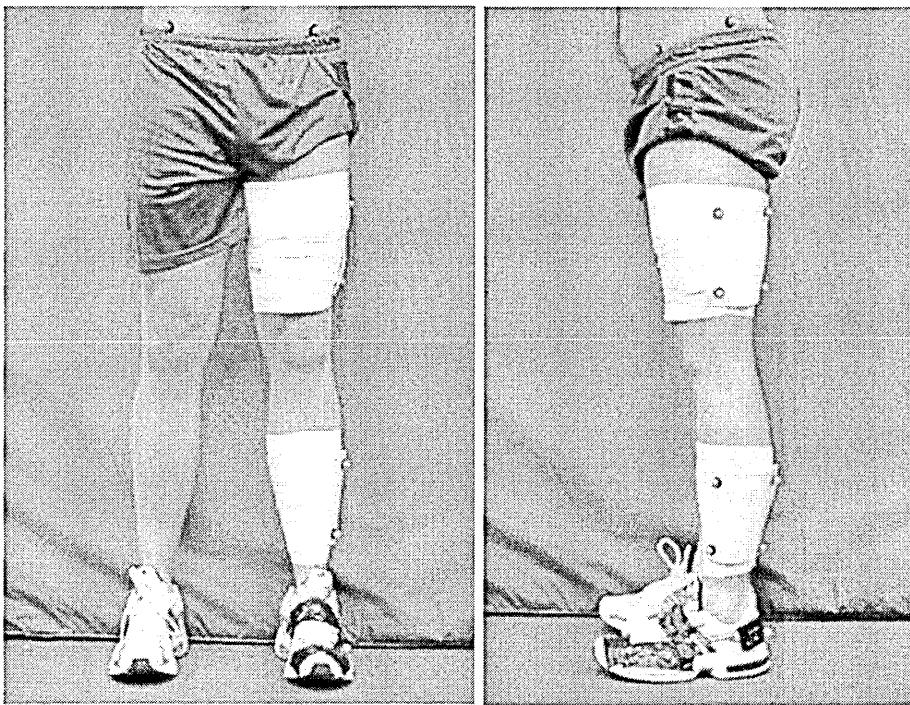


Figure 5.1: The marker set configuration

5.2.3 Procedures

After participant preparation, data were collected during overground and treadmill running; the order in which the conditions were presented to the participants was randomised. In the overground condition, each participant was required to run across the laboratory, approximately 16 m in length, at $3.8 \text{ m}\cdot\text{s}^{-1}$ ($\pm 5\%$) while hitting the force platform with his left foot. Before testing, participants were permitted as many practice trials as they required to be able to achieve this with no alterations to their normal running gait. During testing, trials were accepted when the whole of the participant's left foot contacted the force platform, without any obvious

alterations to running stride, while running at the desired speed. A total of eight acceptable overground trials were collected for each participant.

In the treadmill condition, all participants were required to complete a treadmill habituation period before data were collected. This habituation consisted of a five minute level treadmill run at a speed of $3.8 \text{ m}\cdot\text{s}^{-1}$. This was deemed acceptable because it is consistent with the recommendations of Matsas *et al.* (2000) and the participants were experienced treadmill runners. Participants were then given a rest period of approximately ten minutes between the habituation period and data collection. The treadmill trial consisted of a one minute warm-up period at $3.8 \text{ m}\cdot\text{s}^{-1}$ during which the participants were able to settle into a natural gait. At the end of the warm-up 15 seconds of data, containing at least nine left leg strides, were collected.

At the end of the data collection session, a further static calibration trial was performed to allow correct anatomical reference frame alignment. Additional markers were attached to each participant's left leg at the medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter and right posterior superior iliac spine. Kinematic data were collected for 3 seconds with the participant in the anatomical position.

5.2.4 Data analysis

The data were exported to Matlab (Mathworks Inc., Natick, MA, USA) and smoothed using a fourth order low-pass Butterworth filter. It was always ensured

that greater than 20 data points were included either side of the data of interest before filtering to avoid deleterious end effects due to insufficient padding (Smith, 1989). Because of the possibility of differing frequency contents of each marker (see Giakis, 2004), separate cut-off frequencies (9-12 Hz) for each individual marker were selected using residual analysis (Winter, 1990). Hip, knee and ankle Joint Coordinate System (JCS) angles (Grood and Suntay, 1983) were then calculated using MARey software (Cavanagh *et al.*, 2001), written for Matlab. The JCS used by the MARey software was based on the standardisation paper of Cole *et al.* (1993) in which the first axis is the flexion axis of the proximal (reference) segment, the third axis is the longitudinal axis of the distal segment (target) and the second (floating) axis is the cross product of the third by first axes - see Appendix A for further details of the joint coordinate systems used.

The resulting angular displacement profiles were then cropped to the length of one left foot stance phase using the force platform data. In both conditions, the vertical component of the ground reaction force was used to determine foot contact events – thresholds of 20 N and 10 N were used to determine foot-strike and toe-off respectively. The cropped profiles were then interpolated to 101 data points using a cubic spline procedure, such that 0 and 100 were foot-strike and toe-off respectively.

Both qualitative and quantitative analyses of coordination variability were incorporated into the study. Coordination variability was assessed qualitatively from inspection of angle-angle plots of the joint motions of interest. Many techniques are

available for quantifying the variability in coordination between body segments and their benefits and limitations were reviewed in Chapter IV. In Chapter IV it was concluded that, in the context of this thesis, the modified ‘normalised root mean squared difference’ (mNoRMS) method was most appropriate. The mNoRMS provides a measure of the variability in the coordination between two joint angles plotted on an angle-angle diagram at each data point of the stance phase – see Appendix B for details of its calculation.

A mean mNoRMS value was calculated over the entire stance phase, which is analogous to the NoRMS value first introduced by Sidaway *et al.*, (1995). Due to the changing functional demands placed on the lower extremity during the stance phase (e.g. weight acceptance, propulsion), calculating the mean coordination variability within specific regions of stance rather than across the entire stance phase might provide a more sensitive measure for detecting between-condition differences (Heiderscheit *et al.*, 2002). Therefore, mean mNoRMS values were calculated over four quarters of the stance phase. This procedure was repeated for each of the following inter-joint couplings: hip flexion-knee flexion, hip flexion-ankle dorsiflexion and knee flexion-rearfoot inversion. The knee flexion-rearfoot inversion coupling was chosen because it has previously been indicated as useful in the study of the relationship between joint coordination variability and patellofemoral pain (e.g. Hamill *et al.*, 1999; Heiderscheit *et al.*, 2002). The hip flexion-knee flexion and hip flexion-ankle dorsiflexion couplings were included because they are sagittal plane couplings that do not include rotations from the secondary planes

(frontal and transverse planes) which have been associated with large skin movement errors – see section 2.6.3.

Differences between mNoRMS averages calculated over the entire stance phase and various phases of stance for each joint coupling were tested using three, two-factor, repeated measures (condition, interval) analyses of variance (ANOVA). In accordance with the hypotheses of the study, only selected comparisons were analysed *post-hoc*. Specifically, pair-wise comparisons of the differences between coordination variability measured overground and on the treadmill were investigated during each quarter of the stance period. Paired *t*-tests were used as *post-hoc* tests. To assess the meaningfulness of any differences, estimates of effect size were also calculated to supplement the inferential statistics, as supported by various authors (e.g. Mullineaux *et al.*, 2001). The effect size statistics were calculated using Equation 5.1:

$$ES = \frac{\bar{x}_2 - \bar{x}_1}{s_c} \quad [\text{Equation 5.1}]$$

Where, \bar{x}_2 is the mean for condition two, \bar{x}_1 is the mean for condition one and s_c is the pooled standard deviation. Effect size statistics were interpreted based on Cohen's (1988) criteria - effect sizes of 0.2, 0.5 and 0.8 represent small, moderate and large differences respectively - in addition to comparison with effect sizes cited in relevant literature. Before performing the statistical tests, the data were screened

to ensure that the data met the assumptions of normal distribution and sphericity - see Appendix D.

5.3 Results

The differences in coordination variability between treadmill and overground locomotion can be seen qualitatively in Figure 5.2. It is apparent that, for all three joint couplings, variability in coordination was lower in the treadmill than the overground running condition. The inferential (Table 5.1) and effect size statistics reveal similar results to these qualitative findings.

Table 5.1: Results of the ANOVAs for the hip flexion-knee flexion, hip flexion-ankle dorsiflexion and knee flexion-rearfoot inversion couplings

	Main effect					
	Mode		Interval		Interaction	
	F	P	F	P	F	P
Hip flexion-knee Flexion	8.18	0.014	6.33	0.001	0.67	0.570
Hip flexion-ankle dorsiflexion	9.74	0.009	24.97	<0.001	2.86	0.092
Knee flexion-rearfoot inversion	9.62	0.009	11.28	<0.001	2.694	0.101

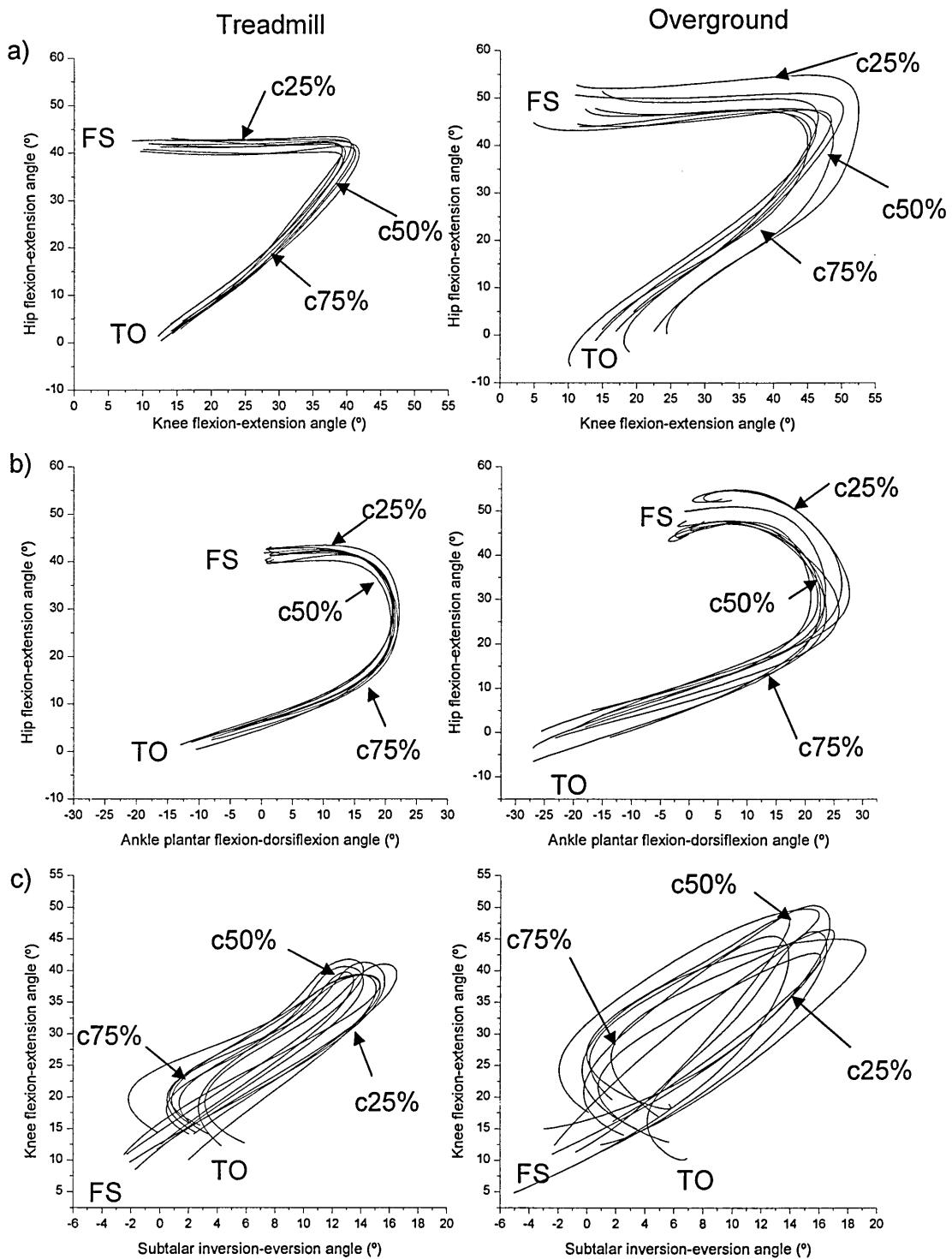


Figure 5.2: Coordination variability during treadmill (left) and overground (right) running for the joint couplings of a) hip flexion-knee flexion, b) hip flexion-ankle dorsiflexion and c) knee flexion-rearfoot inversion in a representative participant. FS = footstrike, TO = toe-off.

Table 5.2. Descriptive statistics for the mNoRMS values averaged across the entire stance phase.

	mNoRMS (°)	
	Treadmill	Overground
Hip flexion-knee flexion	2.31 ± 0.49	3.34 ± 1.18
Hip flexion- ankle dorsiflexion	2.05 ± 0.47	2.97 ± 0.93
Knee flexion-rearfoot inversion	2.20 ± 0.51	3.26 ± 1.13

5.3.1 Hip flexion-knee flexion coupling

The descriptive statistics for the mNoRMS values for the hip flexion-knee flexion coupling averaged over the entire stance phase are given in Table 5.2. The significant main effect for the condition factor (Table 5.1) indicated that, for the mNoRMS values averaged over the entire stance phase, coordination variability was significantly greater in the overground condition than in the treadmill condition (effect size = 0.91). Due to the changing functional demand placed on the lower extremity throughout the stance phase (Heiderscheit *et al.*, 2002), differences in coordination variability were also assessed over different periods of the stance phase and the descriptive statistics for the hip flexion-knee flexion coupling are displayed in Figure 5.3 (top panel). From inspection of Figure 5.3 it is apparent that there was a pattern of greater coordination variability in the overground condition in all phases of stance than in the treadmill condition. Post-hoc tests performed on the selected comparisons indicated that overground running was associated with significantly higher coordination variability during the 0-25% ($p = 0.034$, effect size

= 0.68), 26-50% ($p = 0.023$, effect size = 0.93), 51-75% ($p = 0.031$, effect size = 0.87) and 76-100% ($p = 0.013$, effect size = 0.80) phases of the stance period.

5.3.2 Hip flexion-ankle dorsiflexion coupling

The descriptive statistics for the mNoRMS values for the hip flexion-ankle dorsiflexion coupling averaged over the entire stance phase are given in Table 5.2. The significant main effect for the condition factor (Table 5.1) indicated that, for the mNoRMS values averaged over the entire stance phase, coordination variability was significantly higher in the overground condition than in the treadmill condition (effect size = 1.14). The descriptive statistics for the hip flexion-ankle dorsiflexion coupling over the four quarters of the stance phase are displayed in Figure 5.3 (middle panel). Post-hoc tests performed on the selected comparisons indicated that overground running was associated with significantly higher coordination variability during all phases of stance; 0-25% ($p = 0.010$, effect size = 0.95), 26-50% ($p = 0.037$, effect size = 0.86), 51-75% ($p = 0.020$, effect size = 1.03) and 76-100% ($p = 0.013$, effect size = 0.95) phases of stance.

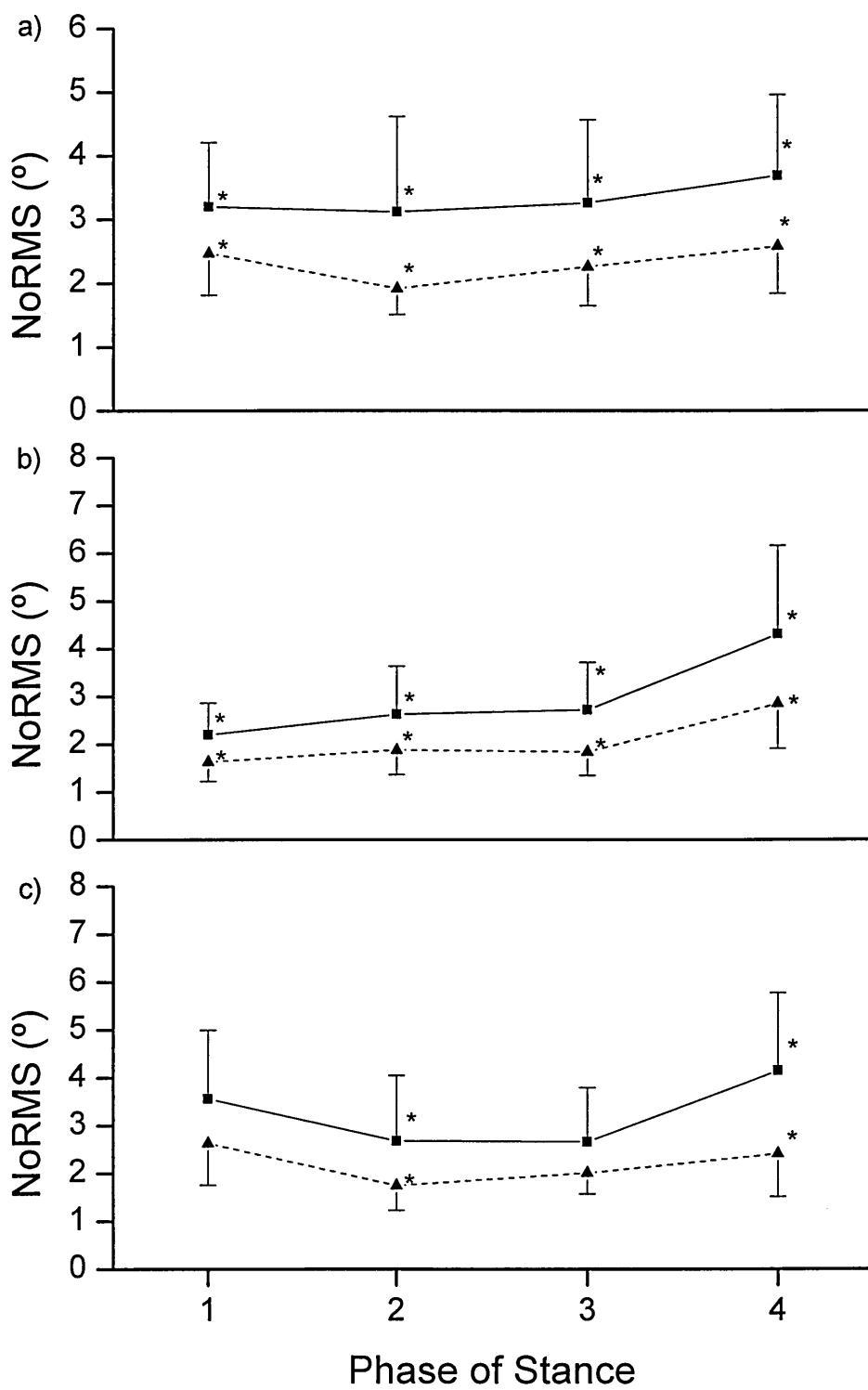


Figure 5.3: Average coordination variability during treadmill (dotted line with triangles) and overground (solid line with squares) running for the joint couplings of a) hip flexion-knee flexion, b) hip flexion-ankle dorsiflexion and c) knee flexion-rearfoot inversion over the four quarters of the stance phase, where phase 1,2,3 and 4 are 0-25%, 26-50%, 51-75% and 76-100% of stance respectively.

* significant difference between conditions ($p < 0.05$)

5.3.3 Knee flexion-rearfoot inversion coupling

The descriptive statistics for the mNoRMS values for the knee flexion-rearfoot inversion coupling averaged over the entire stance phase are given in Table 5.2. The significant main effect for condition factor (Table 5.1) indicated that, for the mNoRMS values averaged over the entire stance phase, coordination variability was significantly higher in the overground condition than in the treadmill condition (effect size = 1.13). The descriptive statistics for the knee flexion-rearfoot inversion coupling over the four quarters of the stance phase are displayed in Figure 5.3 (bottom panel). Post-hoc tests performed on the selected comparisons indicated various significant differences. Overground running was associated with significantly higher coordination variability than treadmill running in the 26-50% ($p = 0.027$, effect size = 0.85), and 76-100% ($p = 0.005$, effect size = 1.30) phases of stance. The differences between conditions were not statistically significant during the 0-25% ($p = 0.053$, effect size = 0.72) and 51-75% ($p = 0.092$, effect size = 0.69) stance periods.

5.4 Discussion

The purpose of this study was to assess differences in the variability of lower extremity coordination between overground and treadmill running. Significantly reduced variability in lower extremity coordination in the treadmill condition was hypothesised. Significantly lower coordination variability ($p < 0.05$) was observed in the treadmill condition than the overground condition, over the entire stance phase as well as during various phases of stance, for all joint couplings studied. Effect

size statistics were also used in an attempt to determine the meaningfulness of the statistically significant differences; all effect sizes for the significant comparisons were large - based on Cohen's (1988) criteria - ranging from 0.80 to 1.30. Even in the various phases of the stance period for each joint coupling in which the differences between overground and treadmill locomotion were not statistically significant ($p > 0.05$), the pattern of increased coordination variability during overground running could be seen in all couplings during each phase stance (see Figure 5.3). Indeed, estimates of effect size indicated that the non-significant differences were of moderate magnitude.

No studies have previously investigated differences in coordination variability between overground and treadmill running. However, the results of this study are consistent with data presented on the variability in the velocity of the centre-of-mass during running (Nelson *et al.*, 1972; Wank *et al.*, 1998) and angular kinematics variability during walking (Dingwell *et al.*, 2001). Dingwell *et al.* (2001) reported differences between overground and treadmill locomotion in lower extremity angular kinematics variability. These authors examined kinematic variability of sagittal plane hip, knee and ankle joint angles during overground and treadmill walking. Significantly smaller standard deviations in joint angles were reported in the treadmill condition than the overground condition at the ankle and knee joint. Dingwell *et al.* (2001) also reported that the differences between modes of locomotion in terms of kinematic variability became 'systematically more significant' from the proximal to the distal joints. Further, although the differences were not statistically significant at the hip joint, the pattern of decreased variability

during treadmill walking was still apparent. Unlike Dingwell *et al.* (2001), in the present study, significant differences were seen even for proximal joint couplings. Potentially, this subtle difference in findings is because the focus of this study was the variability in the coordination between joints. As two, as opposed to one, joints were free to vary, it possibly made the dependent measures more sensitive to changes in the variability of the lower extremity. However, evidence of larger differences at the distal joints was seen in this study with larger effect sizes reported for the two couplings incorporating the ankle joint – hip flexion-ankle dorsiflexion and knee flexion-ankle inversion - compared to the hip flexion-knee flexion coupling – see sections 5.3.1, 5.3.2 and 5.3.3.

It has previously been suggested that coordination variability might provide an adaptive mechanism to potential external perturbations such as uneven ground (Holt *et al.*, 1995). The present study lends some support to this hypothesis because reduced coordination variability was apparent in the treadmill condition and it is likely that less threat of an external perturbation is perceived when running on a treadmill. However, other potential reasons for the observed reduction in coordination variability during treadmill running can be postulated. For example, it is possible that intra-stride belt speed fluctuations could cause differences between the two modes of locomotion (Savelberg *et al.*, 1998). However, neither Savelberg *et al.* (1998) nor any other author have reported the effects of intra-stride belt speed variations on coordination variability. Potentially, an increase in coordination variability might be expected as a result of intra-stride belt speed variations, not the decrease observed in this and other studies (e.g. Dingwell *et al.*, 2001). It is also

possible that participant familiarity with treadmill locomotion could have an effect on differences between the two modes (Scheib, 1986). However, it is unlikely that this accounted for the differences seen in the present study because the participants were experienced treadmill runners and received a five minute habituation period which has been deemed adequate in walking (Matsas *et al.*, 2000).

Further potential reasons for the differences between the modes of locomotion include differences in the mechanical characteristics of the treadmill and overground surfaces (Frishberg, 1981; Dingwell *et al.*, 2001) and reductions in the air resistance experienced by the participants in the treadmill condition (Frishberg, 1981; van Ingen Schenau, 1980). However, there is no evidence in the literature to suggest what the effect of changes in these parameters might have on coordination variability. Another viable explanation for the differences between the modes of locomotion is the altered perceptual information available during treadmill running (van Ingen Schenau, 1980). Possibly most importantly, participants were exposed to different optical flow information in the two conditions. As van Ingen Schenau (1980) highlighted, during overground running the surroundings move with respect to the participant, which is not the case during treadmill locomotion. In other words, optical flow information was present during overground running, whereas, in the treadmill condition, optical flow information was absent. A fundamental notion of ecological theories of movement control is the idea that perception and action are tightly coupled - see Williams *et al.* (1999) for a detailed review. From this theoretical standpoint it should be expected that a change in the

perceptual optical flow information would result in a change in the action or outcome of the movement i.e. coordination and coordination variability. Limited evidence exists in the literature to suggest that the lack of optical flow information during treadmill locomotion has little effect on the variability of gait (Masson and Pailhous, 1994). However, no studies have assessed the effects of optical flow information on coordination variability during locomotion. An interesting direction for future research would be to evaluate the effects of introducing optical flow information, comparable to that in overground locomotion, into a treadmill running condition.

Dingwell *et al.* (2001) suggested that a reason for decreased kinematic variability in the treadmill condition was the treadmill belt imposing an artificially constant speed, externally driving the participant's feet throughout the stance phase of each stride cycle. This might provide another plausible explanation for the lower coordination variability noted during the stance phase of running on a treadmill than running overground in this study. An interesting point that was noted earlier in this section is that larger magnitudes of differences were reported for the two couplings incorporating the ankle joint – hip flexion-ankle dorsiflexion (effect size = 1.14) and knee flexion-ankle inversion (effect size = 1.13) - compared to the hip flexion-knee flexion coupling (effect size = 0.91). This result seems intuitive, as the ankle is closest to the belt that is constraining the movement in the treadmill condition and, therefore, has the least potential for variability during treadmill running. This is again consistent with the results of Dingwell *et al.* (2001), who reported that differences between overground and treadmill walking systematically became

'more significant' from the proximal to the distal joints. Recently, Minetti *et al.* (2003) presented details of a feedback controlled treadmill (treadmill-on-demand), the speed of which is continuously changed according to the participant's preference to accelerate, decelerate or keep a constant speed. On this type of treadmill, participants are not constrained to run at a constant speed. Consequently, locomotion on the treadmill-on-demand might better resemble overground locomotion in terms of coordination variability than conventional motorised treadmill locomotion. This is an interesting topic for future research which needs to be addressed appropriately.

Whether the reasons for the differences in coordination variability between the two modes of locomotion are related to changes in the mechanical constraints, perceptual information or any other factors, there are important implications of the findings of this study. The first relates to overuse injury risk. In section 2.5.1 of this thesis an hypothesised link between variability and overuse injury (James, 2004) was presented. Briefly, James (2004) suggested that variability might play a positive role in preventing overuse injuries. Variability might provide a broader distribution of stresses among different tissues or alter the stress magnitude, direction and frequency within the same tissue (James, 2004). In other words, a person might be at an increased risk of overuse injury if they exhibited less locomotion variability. There is limited experimental evidence to support this hypothesis. For example, participants with a history of lower extremity injury exhibited lower variability in various internal joint moment dependent variables than participants with no history of lower extremity injury in a drop jumping task (James

et al., 2000). Also, individuals with patellofemoral pain syndrome have been reported to exhibit less coordination variability than healthy controls (Hamill *et al.*, 1999; Heiderscheit *et al.*, 2002).

The significantly reduced coordination variability exhibited during treadmill running could indicate that runners are at greater risk of overuse injury in comparison to running overground. Indeed, the size of the effects for the differences in coordination variability between modes of locomotion in this study (0.80-1.30) were larger than the magnitudes of the differences between patellofemoral pain syndrome and control participants reported by Heiderscheit *et al.* (2002) – effect sizes⁸ < 0.55. However, it should be noted that no conclusive evidence exists to link decreased variability with overuse injury and a prospective study is required to establish cause and effect (Hamill *et al.*, 1999; Heiderscheit *et al.*, 2002). Therefore, it is not possible to make a judgment about the clinical significance of these findings. It should also be noted that Milgrom *et al.* (2003) recently reported that tibial axial compression and tension strain and strain rates were 48-285% greater during overground running than during treadmill running. This again highlights that caution must be applied in concluding that runners would be at greater risk of injury during treadmill locomotion. Although the larger variability during overground running will, potentially, better distribute the loads over different internal structures, it also appears the magnitude of the loads is reduced in treadmill running. Parenthetically, it is possible that loads on the lower extremity

⁸ Calculated using the descriptive statistics provided in the paper and Equation 5.1

themselves could be a constraint that causes increased variability during overground running, in order to provide a broader distribution of the increased load. If an appropriate experimental design could be developed, an investigation into the relationship between load and locomotion variability would be an interesting direction for future work.

Second, it would appear that performing studies of coordination variability on a motorised treadmill might mask differences between experimental groups. This is especially relevant in situations where differences in neuromuscular control are likely to result in changes in the variability of locomotion (Dingwell *et al.*, 2001). An example of treadmill walking producing misleading results with regards to locomotion variability is apparent in other work of Dingwell and colleagues (Dingwell *et al.*, 1999; Dingwell *et al.*, 2000). Dingwell *et al.* (1999) assessed kinematic variability in patients with diabetic neuropathy and matched controls walking on a treadmill and reported that the differences between groups were not significant. However, differences between groups were statistically significant when a similar cohort of participants walked overground (Dingwell *et al.*, 2000). This led Dingwell *et al.* (2001) to conclude that treadmills should not be used to study locomotion variability in certain circumstances. The results of this study conform to the findings of Dingwell *et al.* (2001) and suggest that their conclusions regarding walking can be extended to running. However, although differences between experimental groups which resulted from modifications in neuromuscular control have the potential to be masked if a motorised treadmill is used it is unclear whether differences that are the result of changed mechanical constraints would be

concealed. Examples of such mechanical constraints include the use of orthotics, changed shoe/surface characteristics and the use of bracing. The effects of treadmill analysis on variability due to the addition of mechanical constraints would also be an interesting avenue for future research.

5.4.1 Conclusion

Reduced coordination variability during treadmill locomotion compared to overground locomotion was hypothesised and the results presented in this Chapter support this. Therefore, performing studies of joint coordination variability on a motorised treadmill might mask differences between experimental groups. Also, caution should be applied when comparing results from studies using overground and treadmill analysis. Further work to determine the cause of the large differences in coordination variability between the two conditions is certainly warranted.

CHAPTER VI

6 Coordination variability during overground, treadmill and treadmill-on-demand running

6.1 Introduction

The dynamical systems approach affords a positive, functional role to movement variability as opposed to the traditional interpretation of variability as error - see sections 2.4 and 5.4. Recently, as described in section 2.4, authors have employed dynamical systems methods to address a range of research questions in locomotion. Examples of such studies include investigations into the relationship between coordination variability and joint pain (Hamill *et al.*, 1999; Heiderscheit, 2000; Heiderscheit *et al.*, 2002), the effect of orthoses on coordination and coordination variability (Ferber *et al.*, 2005), the dynamics of the walk-to-run transition (Diedrich and Warren, 1995; Kao *et al.*, 2003) and the relationship between spinal cord injury and coordination variability (Field-Fote and Tepavac, 2002). Some of these studies were conducted overground (Ferber *et al.*, 2005), whilst many involved treadmill locomotion (Diedrich and Warren, 1995; Hamill *et al.*, 1999; Heiderscheit, 2000; Field-Fote and Tepavac, 2002; Heiderscheit *et al.*, 2002; Kao *et al.*, 2003). Obviously, there is a problem of ecological validity if differences in coordination and coordination variability exist between the two modes of locomotion.

The literature regarding the comparison of the kinematics of overground and treadmill locomotion is inconsistent (Nigg *et al.*, 1995). For example, whilst some studies have reported differences between kinematic and kinetic parameters during overground and treadmill locomotion (e.g. Sykes 1975; Elliot and Blanskby, 1976; Nigg *et al.*, 1995; White *et al.*, 1998; Schache *et al.*, 2001) others have reported no statistically significant differences (e.g. Murray *et al.*, 1985; Lemke *et al.*, 1995) - see section 2.6.4.1 for a detailed review of these studies.

There are many potential reasons for the disparity between studies that have compared the two modes of locomotion. First, small sample sizes have often been used ($n < 10$) without the assurance that they were determined based on *a priori* statistical power calculations, potentially limiting the ability to detect differences between modes. Second, Savelberg (1998) suggested that differences observed between overground and treadmill locomotion were, at least in part, due to intra-stride variations in treadmill belt speed - see section 2.6.4.4 for a detailed review of the effects of intra-stride belt speed variations on the differences between overground and treadmill locomotion. However, very few studies that compared overground and treadmill locomotion reported the magnitude of the belt speed variations or discussed their effect on the dependent variables under investigation. Third, participant habituation to treadmill locomotion might also explain the irregularity in findings in the literature. Many studies have addressed this issue (e.g. Charteris and Taves, 1978; Wall and Charteris, 1980, 1981; Scheib, 1986; Taylor *et al.*, 1996; Matsas *et al.*, 2000; Wass *et al.*, 2004) which were reviewed in detail in section 2.6.4.3 of this thesis. Briefly, authors have suggested that novice

treadmill walkers/runners need between 4 (Matsas *et al.*, 2000) and 45 minutes (Scheib, 1986) to become habituated to treadmill locomotion. It has been recognised that a lack of participant habituation might account for some of the differences reported between overground and treadmill locomotion. Therefore, it is possible that the inconsistency between studies that have compared overground and treadmill locomotion could be explained by differences in the familiarity of different cohorts to treadmill locomotion.

Although the findings of studies that have compared overground and treadmill kinematics are equivocal, the results of investigations of movement variability are more consistent. The limited examples of studies that have compared variability in kinematics between overground and treadmill locomotion were reviewed in detail in section 2.6.4.5. In an early study comparing overground and treadmill locomotion, Nelson *et al.* (1972) reported lower variability in the horizontal and vertical velocities of the centre-of-mass while running on a treadmill than running overground. Similar findings were also reported by Wank *et al.* (1998). The only study in the literature to have specifically compared the variability in kinematic measurements during overground and treadmill locomotion is that of Dingwell *et al.* (2001). These authors compared sagittal plane ankle, knee and hip angles collected during overground and treadmill walking. Treadmill walking was associated with significantly lower variability than overground walking (Dingwell *et al.*, 2001). The lack of literature comparing movement variability during overground and treadmill running provided a rationale for the study presented in Chapter V of this thesis; the results of which confirmed the findings of Dingwell *et al.* (2001) as

treadmill running was associated with decreased variability in lower extremity joint couplings. Several reasons for the observed reductions in the variability in coordination during treadmill locomotion were postulated in section 5.4. These reasons included altered mechanical constraints on performance, such as the surface characteristics, in the two conditions and changes in the air resistance experienced by participants during treadmill locomotion. Further reasons were related to changes in the perceptual information available in the two modes of locomotion. For example, various authors have hypothesised that the changes in visual information available to participants in the treadmill condition might account for any differences seen between overground and treadmill locomotion (e.g. Frishberg, 1983; van Ingen Schenau, 1980). Optical flow information is available during overground locomotion whereas it is absent during locomotion on a treadmill.

Another potential reason for the decreased coordination variability during treadmill locomotion given in section 5.4, which was also cited by Dingwell *et al.* (2001), is the treadmill belt imposing an artificially constant speed, externally driving the participant's feet throughout the stance phase of each stride cycle. The constant speed of the treadmill belt might have served to constrain the potential for variability in joint couplings of the lower extremity. An interesting point noted in section 5.4 was that larger magnitudes of differences were reported for the two couplings incorporating the ankle joint – hip flexion-ankle dorsiflexion (effect size = 1.14) and knee flexion-ankle inversion (effect size = 1.13) - compared to the hip flexion-knee flexion coupling (effect size = 0.91). This result seems intuitive, as the

ankle is closest to the belt that is constraining the movement in the treadmill condition and, therefore, has the least potential for variability during treadmill running. This is again consistent with the results of Dingwell *et al.* (2001), who reported that differences between overground and treadmill walking systematically became 'more significant' from the proximal to the distal joints.

Minetti *et al.* (2003) recently presented an innovative type of treadmill that might better resemble overground locomotion in terms of movement variability than conventional treadmill locomotion. The 'treadmill-on-demand', which is a feedback controlled treadmill that changes speed continuously according to the participant's preference to accelerate, decelerate or keep a constant speed, is represented in 

Figure 6.1.

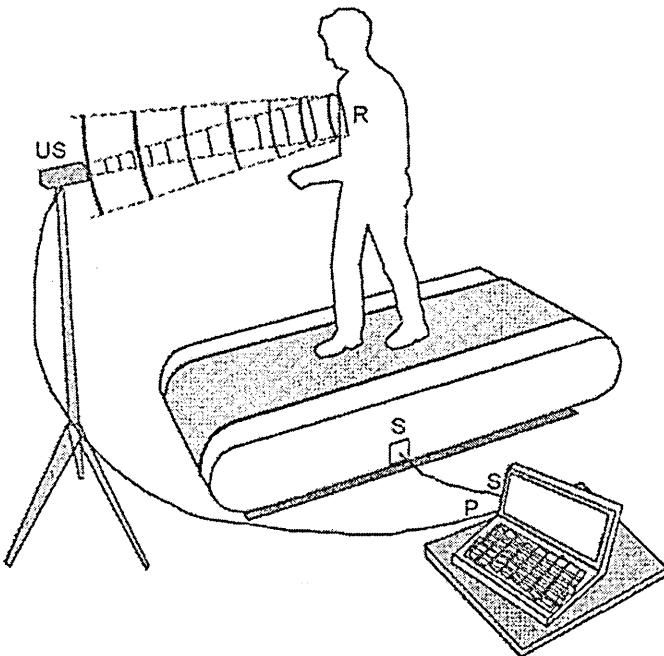


Figure 6.1: The treadmill-on-demand, consisting of an ultrasonic range finder, the treadmill and a computer (Minetti *et al.*, 2003). The ultrasonic range finder is controlled by a computer via the bidirectional parallel port. The range finder (US) emits ultrasonic bursts that bounce off the passive reflector (R) located on the chest of the participant. Subsequently, the reflected bursts are detected by the same transducer. The distance between the participant and the probe is calculated at a rate of 10 Hz based on the time of flight of the pulse. A program processes the distance data and controls the treadmill speed via the serial port. The system is complemented by a safety harness, a bar circling the participant and the emergency stop switch - adapted from Minetti *et al.* (2003).

On this type of treadmill, the participants are not constrained to run at a constant speed. Consequently, locomotion on the treadmill-on-demand might better resemble overground locomotion in terms of coordination variability than conventional treadmill locomotion. Therefore, the purpose of this study was to compare joint coordination variability during overground, conventional treadmill and treadmill-on-demand running. It was hypothesised that running on the treadmill-on-demand would better resemble overground running than running on the conventional treadmill in terms of coordination variability. Specifically, the following research hypotheses were formulated:

H_1 – mNoRMS values for the hip flexion-knee flexion, coupling averaged over the entire stance period and various phases of stance, will be lower in the conventional treadmill condition than the overground and treadmill-on-demand conditions.

H_2 – mNoRMS values for the hip flexion-ankle dorsiflexion, coupling averaged over the entire stance period and various phases of stance, will be lower in the conventional treadmill condition than the overground and treadmill-on-demand conditions.

H_3 – mNoRMS values for the knee flexion-rearfoot inversion, coupling averaged over the entire stance period and various phases of stance, will be lower in the conventional treadmill condition than the overground and treadmill-on-demand conditions.

6.2 Methods

6.2.1 Participants

Eleven male participants volunteered to take part in the study. *A priori* power calculations for the repeated measures analyses of variance (Park and Schutz, 1999) based on data presented in Chapter V of this thesis ($\alpha = 0.05$, $\beta = 0.20$) indicated that this number of participants gave the study sufficient power to detect differences between the modes of locomotion. Participants had an average ($\pm s$) age of 23.1 ± 4.3 years, stature of 1.81 ± 0.07 m and body mass of 74.3 ± 7.8 kg. All participants exhibited a rear-foot striking pattern (determined through visual inspection), were experienced in treadmill running, physically active and free from

injury at the time of testing. The Local Research Ethics Committee approved the procedures, and written informed consent was gained from each participant before data collection – see Appendix H. Participants were required to wear only their own running shoes and a pair of tight fitting running shorts throughout testing.

6.2.2 Experimental set-up

All kinematic data were collected using a nine-camera motion capture system (VICON, Oxford Metrics, Oxford, UK) sampling at 120 Hz. The orientation of the global coordinate system was such that the positive x axis pointed in the direction of forward progression, the positive y axis pointed vertically upward and the positive z axis pointed to the right. The nine cameras of the motion capture system were placed in optimal positions around a calibrated measurement volume of dimensions $4.4 \times 1.6 \times 2.1$ m in the anterior-posterior, vertical and medio-lateral directions respectively. The measurement volume was made this size to incorporate the treadmill (Ergo LG70, Woodway, Germany) and to allow the participants to complete one stance period at the correct speed within the measurement volume. Infra-red timing gates (Brower Timing Systems, South Draper, UT, USA) were placed 3.5 m apart either side of the floor-mounted force platform to monitor running speed during the overground trials.

Pre-moulded, Velcro™-backed thermoplastic shells equipped with four 12.5 mm retro-reflective markers were attached to each participant's right shank and thigh. The markers were placed non-collinearly on the thermoplastic shells with inter-

marker distances of greater than 100 mm. The thermoplastic shells were attached to the participant using a technique described as 'optimal' by Manal *et al.* (2000) (Figure 5.1). The shells were attached in distal-lateral locations by fastening the Velcro™ to a securely fastened, under-wrapped elastic bandage (SuperWrap™, Fabrifoom, Exton, PN, USA). Markers were also attached directly to the participant's right running shoe at the fifth metatarsal head, superior navicular, toe (second metatarsal head) and the most posterior aspect of the heel. Four further retro-reflective markers were attached to the participant's pelvis at the left and right anterior superior iliac spines, right iliac crest and right posterior superior iliac spine.

6.2.3 Procedures

After participant preparation, data were collected during overground, treadmill and treadmill-on-demand running. The order in which the conditions were presented to the participants was randomised. In the overground condition, each participant was required to run across the laboratory, approximately 30 m in length, at $3.5 \text{ m}\cdot\text{s}^{-1}$ ($\pm 5\%$) while completing one full right stance phase within the measurement volume. Before testing, participants were permitted as many practice trials as they required to be able to achieve this with no alterations to their normal running gait. During testing, trials were accepted when a full right foot stance period occurred within the measurement volume, without any obvious alterations to running gait and while running at the desired speed. A total of eight acceptable overground trials were collected for each participant.

In the treadmill condition, all participants were required to complete a treadmill habituation session before data were collected. This habituation consisted of a five min level treadmill run at $3.5 \text{ m}\cdot\text{s}^{-1}$. This was deemed acceptable because the participants were experienced treadmill runners and the duration of the session was consistent with the recommendations of Matsas *et al.* (2000). After the treadmill and treadmill-on-demand acclimatisation participants were given a rest period of approximately ten minutes. The treadmill trial consisted of a one minute warm up period at $3.5 \text{ m}\cdot\text{s}^{-1}$ during which the participants were able to settle into a natural gait. At the end of the warm-up 15 seconds of data, containing at least nine right leg strides, were collected.

For treadmill-on-demand running, participants were required to complete an habituation period of approximately ten minutes whilst running at their preferred speed. The treadmill-on-demand trial consisted of a one minute warm up period at $3.5 \text{ m}\cdot\text{s}^{-1}$ during which the participants were able to settle into a natural gait. At the end of the warm-up 15 seconds of data, containing at least nine right leg strides, were collected. As running speed was not fixed in the treadmill-on-demand trials, similar to the overground trials, a $\pm 5\%$ boundary of acceptable speeds was used. This was monitored using a real-time display of treadmill belt speed and any trials in which the speed strayed outside of these limits were discarded.

At the end of the data collection session a static calibration trial was performed to allow correct anatomical reference frame alignment. Additional markers were attached to each participant's right leg at the medial and lateral malleoli, medial

and lateral femoral epicondyles, greater trochanter and left posterior superior iliac spine. Kinematic data were collected for 3 seconds with the participant in the anatomical position.

6.2.4 Data analysis

The raw three-dimensional coordinate data were smoothed using generalised cross-validated quintic splines (Woltring, 1986). Subsequently, hip, knee and ankle Joint Coordinate System (JCS) angles (Grood and Suntay, 1983) were calculated using MARey software (Cavanagh *et al.*, 2001) written for Matlab (Mathworks Inc., Natick, MA, USA). The JCSs used by the MARey software were based on the standardisation paper of Cole *et al.* (1993) in which the first axis is the flexion axis of the proximal (reference) segment, the third axis is the longitudinal axis of the distal segment (target) and the second (floating) axis is the cross product of the third by first axes - see Appendix A for further details of the joint coordinate systems used.

The resulting angular displacement profiles were cropped to produce eight individual right leg stance phases for the overground, treadmill and treadmill-on-demand conditions. Foot-strike and toe-off gait events were determined using custom-written algorithms based on the vertical displacement and velocity of the calcaneus and toe marker respectively (Milner *et al.*, 2002; Wheat and Milner, 2004). The cropped profiles were then interpolated to 101 data points using a cubic spline procedure.

Both qualitative and quantitative analyses of coordination variability were incorporated into the study. Coordination variability was assessed qualitatively from inspection of angle-angle plots. Many techniques are available for quantifying the variability in coordination between body segments and their benefits and limitations were reviewed in Chapter IV. In Chapter IV it was established that, in the context of this thesis, the modified 'normalised root mean squared difference' (mNoRMS) method was most appropriate. The mNoRMS provides a measure of the variability in the coordination between two joint angles plotted on an angle-angle diagram at each data point of the stance period – see Appendix B for details of its calculation.

A mean mNoRMS value was calculated over the entire stance period, which is analogous to the NoRMS value first introduced by Sidaway *et al.*, (1995). Further to this, due to the changing functional demands placed on the lower extremity during the stance phase (e.g. weight acceptance, propulsion), averaging the coordination variability within specific regions of stance rather than across the entire stance phase might provide a more sensitive measure for detecting between-condition differences (Heiderscheit *et al.*, 2002). Therefore, average mNoRMS values were calculated over the four quarters of the stance phase. This procedure was repeated for each of the following inter-joint couplings: hip flexion-knee flexion, hip flexion-ankle dorsiflexion and knee flexion-rearfoot inversion. The knee flexion-rearfoot inversion coupling was chosen because it has previously been indicated as useful in the study of the relationship between joint coordination variability and patellofemoral pain (e.g. Hamill *et al.*, 1999; Heiderscheit *et al.*, 2002). The hip flexion-knee flexion and hip flexion-ankle dorsiflexion couplings

were included because they are sagittal plane couplings which do not include rotations from the secondary planes (frontal and transverse planes) which have been associated with large skin movement errors – see section 2.6.3.

Differences between mNoRMS means calculated over the entire stance period and various phases of stance for each joint coupling were tested using three, two-factor, repeated measures (condition, interval) analyses of variance (ANOVA). In accordance with the hypotheses of the study, only selected comparisons were analysed *post-hoc*. Specifically, differences between overground and treadmill, overground and treadmill-on-demand as well as treadmill and treadmill-on-demand conditions were investigated during each period of stance for each coupling. Paired *t*-tests were used as *post-hoc* tests. To assess the meaningfulness of any differences, estimates of effect size were also calculated to supplement the inferential statistics, as supported by various authors (e.g. Mullineaux *et al.*, 2001). The effect size statistics were calculated using Equation 5.1. Effect sizes were interpreted based on Cohen's (1988) criteria - effect sizes of 0.2, 0.5 and 0.8 represent small, moderate and large differences respectively - in addition to comparison with effect sizes cited in relevant literature. Before performing the statistical tests, the data were screened to ensure that they met the assumptions of normal distribution and sphericity - see Appendix D.

6.3 Results

Differences in coordination variability between the three modes of locomotion for the hip flexion-ankle dorsiflexion, hip flexion-knee flexion and knee flexion-rearfoot inversion couplings can be seen qualitatively in Figure 6.2, Figure 6.3 and Figure 6.4 respectively. The angle-angle traces appear less consistent in the overground condition compared to the other modes. Therefore, both the treadmill and treadmill-on-demand condition appear to have elicited decreased coordination variability compared to overground running. The inferential and effect size statistics reveal similar results to these qualitative findings.

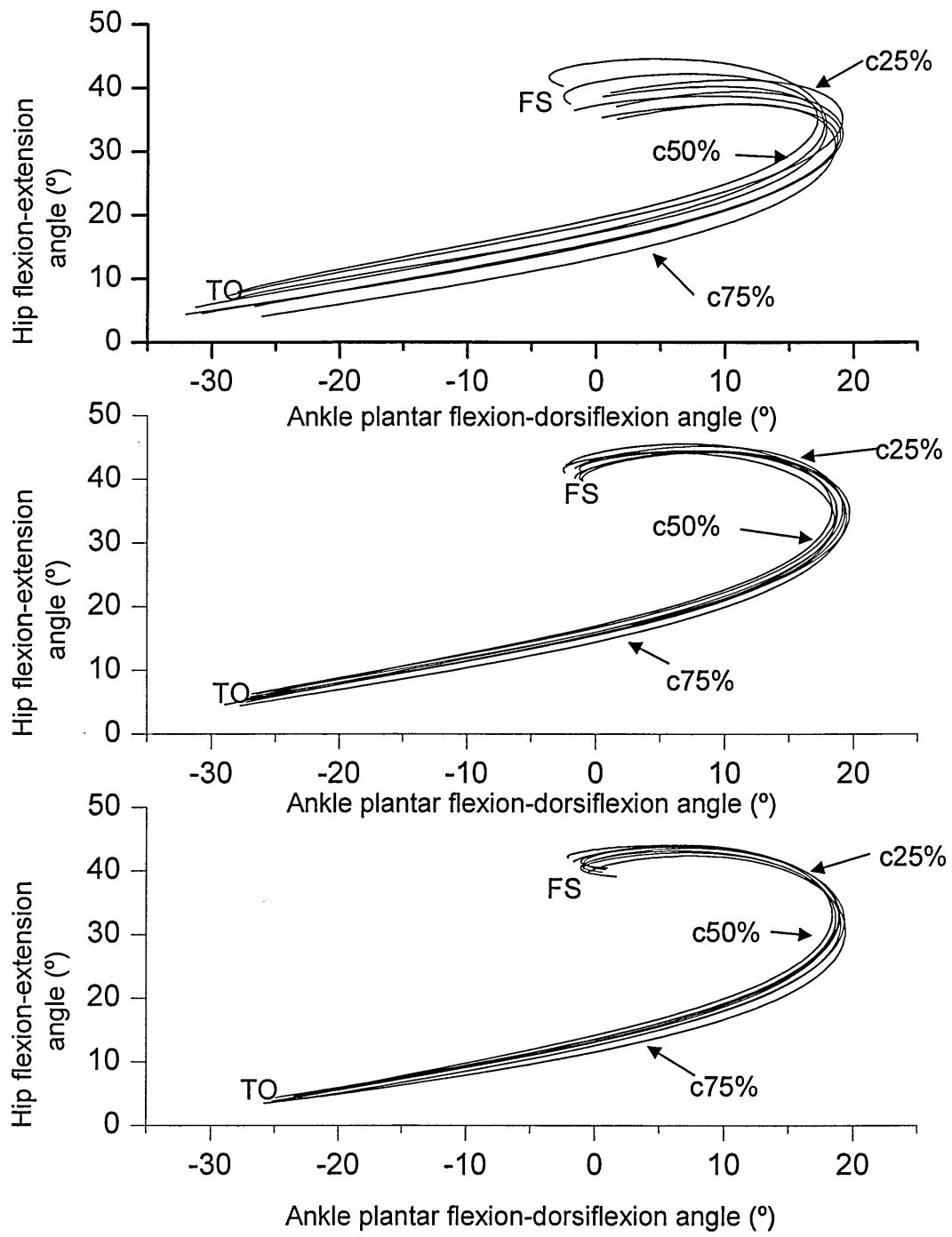


Figure 6.2: Coordination variability during overground (top), treadmill (middle) and treadmill-on-demand (bottom) running for the hip flexion-ankle dorsiflexion joint coupling for a representative participant. FS = footstrike, TO = toe-off.

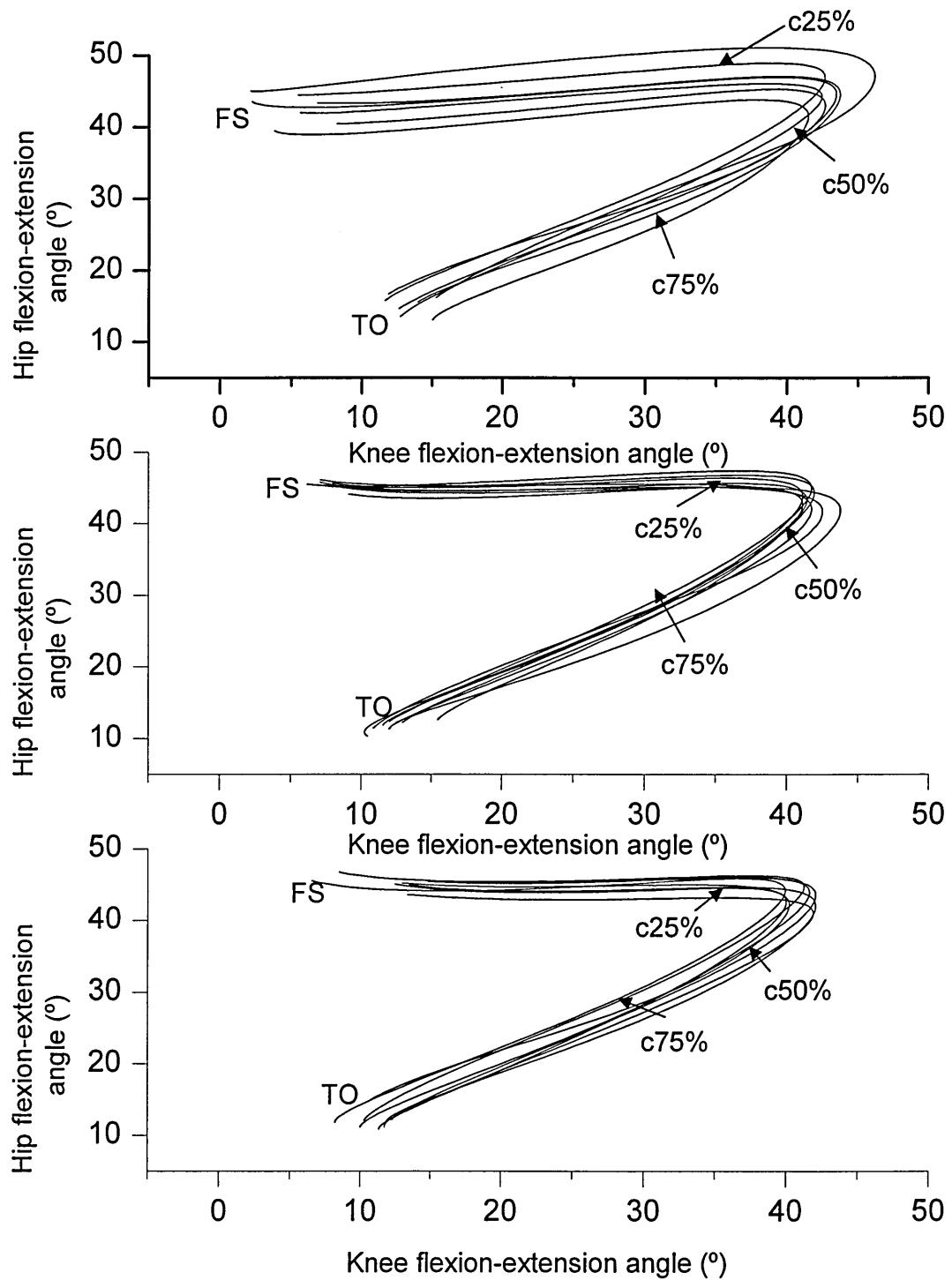


Figure 6.3: Coordination variability during overground (top), treadmill (middle) and treadmill-on-demand (bottom) running for the hip flexion-knee flexion joint coupling for a representative participant. FS = footstrike, TO = toe-off.

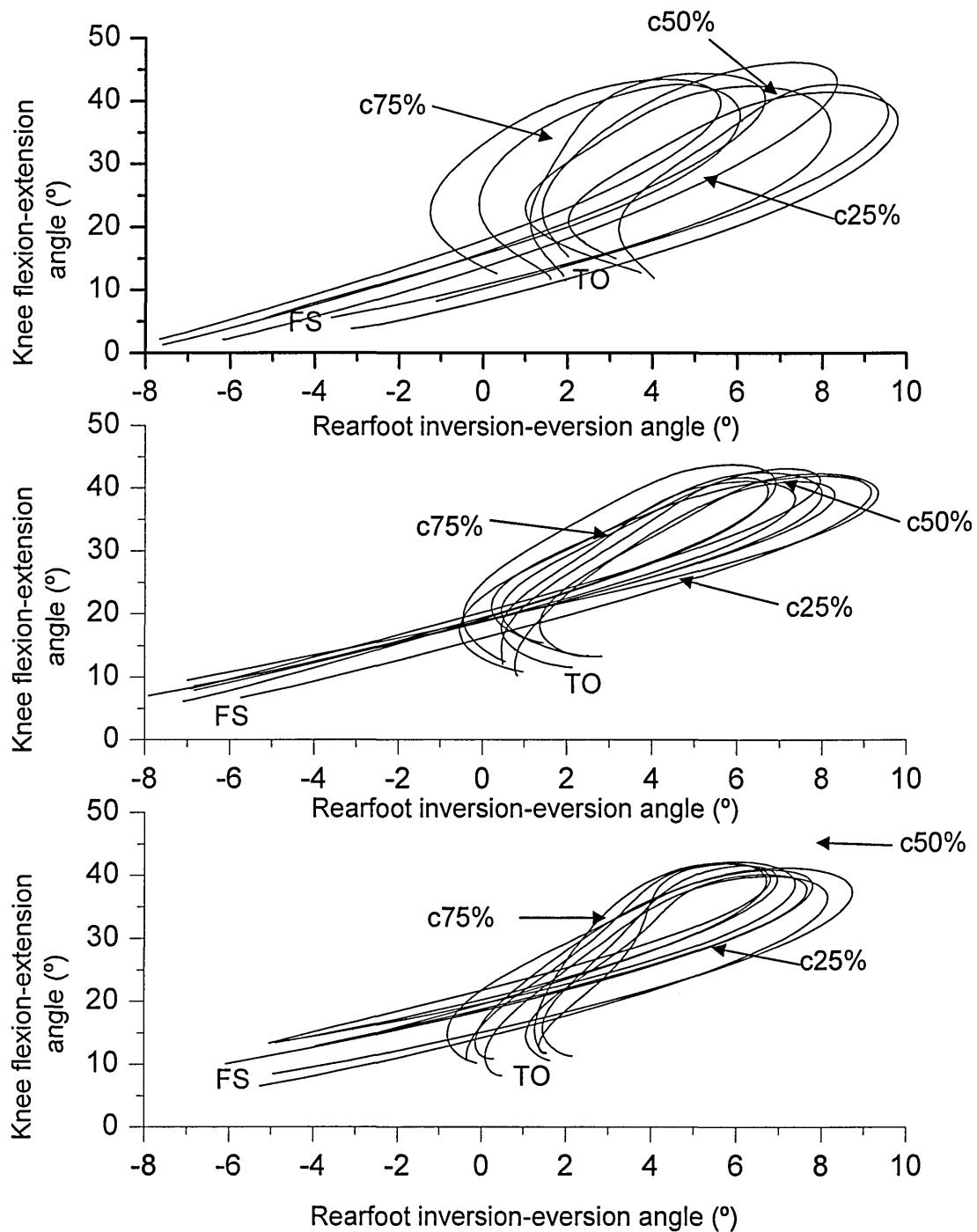


Figure 6.4: Coordination variability during overground (top), treadmill (middle) and treadmill-on-demand (bottom) running for the knee flexion-rearfoot inversion joint coupling for a representative participant. FS = footstrike, TO = toe-off.

Table 6.1: Descriptive statistics for the mNoRMS values averaged across the entire stance phase

	mNoRMS (°)		
	Overground	Treadmill	Treadmill-on-demand
Hip flexion-knee flexion	3.09 ± 0.69	2.19 ± 0.55	2.16 ± 0.48
Hip flexion- ankle dorsiflexion	2.95 ± 0.52	2.05 ± 0.53	2.03 ± 0.52
Knee flexion-rearfoot inversion	2.87 ± 0.90	1.99 ± 0.47	1.97 ± 0.44

Table 6.2: Results of the two factor (condition, interval) ANOVAs for the hip flexion-knee flexion, hip flexion-ankle dorsiflexion and knee flexion-rearfoot inversion couplings

	Main effect					
	Mode		Interval		Interaction	
	F	P	F	P	F	P
Hip flexion-knee flexion	17.29	<0.001	5.78	0.003	0.46	0.830
Hip flexion- ankle dorsiflexion	21.74	<0.001	7.63	0.012	0.33	0.771
Knee flexion-rearfoot inversion	10.82	0.001	18.85	<0.001	1.85	0.157

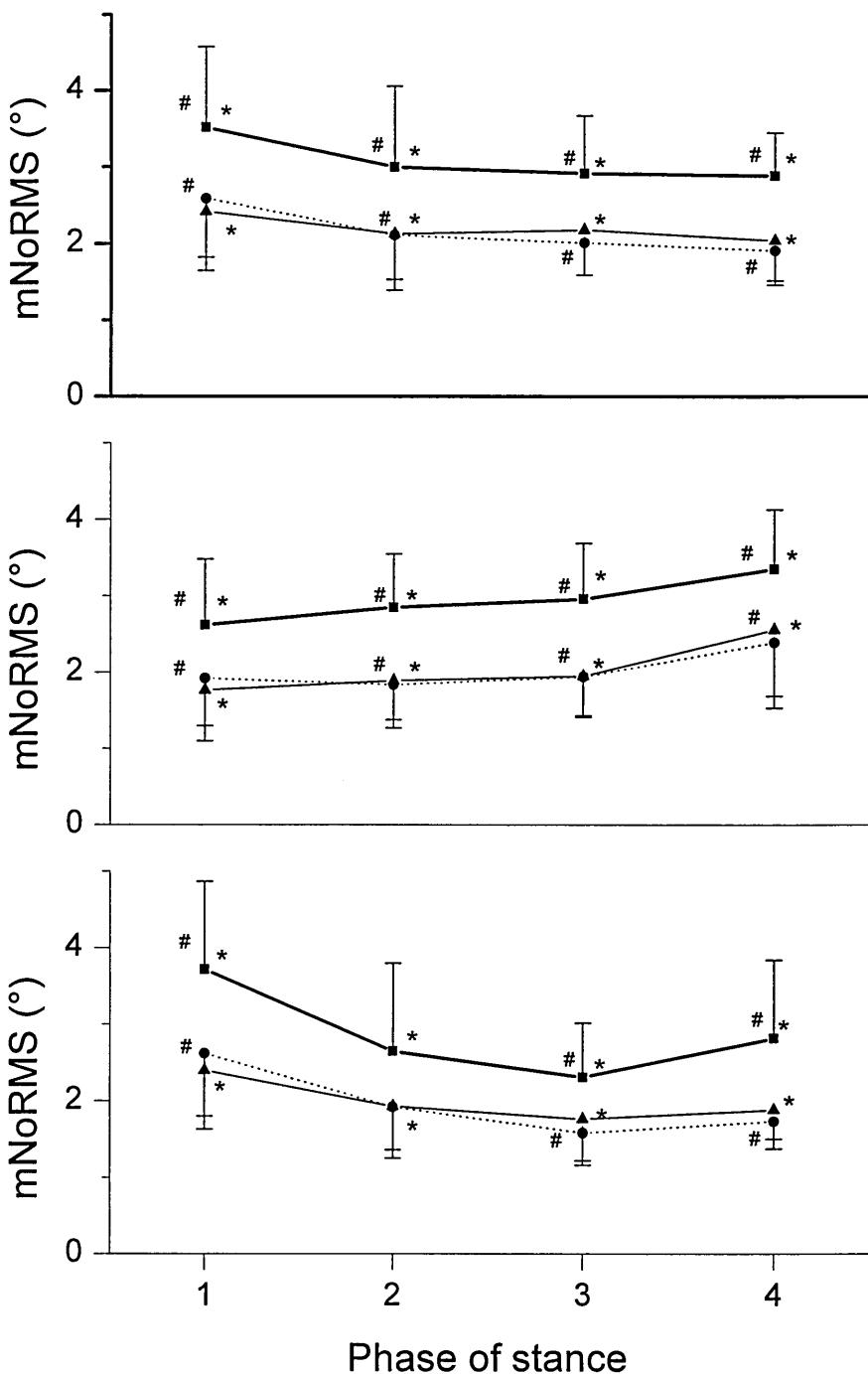


Figure 6.5: Average coordination variability during treadmill (thin solid line with triangles), treadmill-on-demand (dotted line with circles) overground (thick solid line with squares) running for the joint couplings of hip flexion-knee flexion (top), hip flexion-ankle dorsiflexion (middle) and knee flexion-rearfoot inversion (bottom) over the four quarters of the stance phase, where phase 1,2,3 and 4 are 0-25%, 26-50%, 51-75% and 76-100% of stance respectively. *significant difference between overground and treadmill ($p < 0.05$) #significant difference between overground and treadmill-on-demand conditions ($p < 0.05$)

6.3.1 Hip flexion-knee flexion coupling

The descriptive statistics for the mNoRMS values for the hip flexion-knee flexion coupling averaged over the entire stance period are given in Table 6.1. Results of the ANOVA for the hip flexion-knee flexion coupling indicated a significant main effect for the condition factor (see Table 6.2). Pair-wise follow-up tests indicated that mNoRMS values averaged across the entire stance phase were significantly higher in the overground condition compared to both the treadmill ($p = 0.001$, effect size = 1.44) and treadmill-on-demand ($p = 0.001$, effect size = 1.56) conditions. However, coordination variability was not significantly different between treadmill and treadmill-on-demand conditions ($p = 0.854$, effect size = 0.06). Due to the changing functional demand placed on the lower extremity throughout the stance phase (Heiderscheit *et al.*, 2002), differences in coordination variability were also assessed over different periods of stance and the descriptive statistics for the hip flexion-knee flexion coupling are displayed in Figure 6.5 (top panel). From inspection of Figure 6.5 it is apparent that there was a pattern of greater coordination variability in the overground condition in all phases of stance than in the treadmill and treadmill-on-demand conditions. Post-hoc tests performed on the selected comparisons for the different stance phases indicated various significant differences. First, coordination variability was significantly lower in the treadmill than in the overground condition during 0-25% ($p = 0.009$, effect size = 1.19), 26-50% ($p = 0.001$, effect size = 0.95), 51-75% ($p = 0.009$, effect size = 1.10) and 76-100% ($p = 0.002$, effect size = 1.59) phases of stance. The overground condition was also associated with significantly greater coordination variability than the treadmill-on-demand condition during 0-25% ($p = 0.026$, effect size = 1.00), 26-

50% ($p = 0.014$, effect size = 1.04), 51-75% ($p < 0.001$, effect size = 1.49) and 76-100% ($p < 0.001$, effect size = 1.95) phases of stance. Coordination variability was not significantly different between the treadmill and treadmill-on-demand during any phases of stance (p 's > 0.234, effect sizes < 0.33).

6.3.2 Hip flexion-ankle dorsiflexion coupling

The descriptive statistics for the mNoRMS values for the hip flexion-ankle dorsiflexion coupling averaged over the entire stance period are also given in Table 6.1. Results of the ANOVA for the hip flexion-ankle dorsiflexion coupling indicated a significant main effect for the condition factor (see Table 6.2). Pair-wise follow-up tests indicated that mNoRMS values averaged across the entire stance phase were significantly higher in the overground condition than in the treadmill ($p < 0.001$, effect size = 1.71) and treadmill-on-demand ($p < 0.001$, effect size = 1.79) conditions. However, coordination variability was not significantly different between treadmill and treadmill-on-demand conditions ($p = 0.895$, effect size = 0.06). Descriptive statistics for the hip flexion-ankle dorsiflexion coupling over the different phases of stance are displayed in Figure 6.5 (middle panel). From inspection of Figure 6.5 it is apparent that there was a pattern of greater coordination variability in the overground condition in all stance phases than in the treadmill and treadmill-on-demand conditions. Post-hoc tests performed on the selected comparisons for the different stance phases indicated various significant differences. The overground condition exhibited significantly greater coordination variability than the treadmill condition during 0-25% ($p = 0.032$, effect size = 1.10), 26-50% ($p = 0.001$, effect size = 1.46), 51-75% ($p = 0.001$, effect size = 1.61) and 76-100% stance

period ($p = 0.042$, effect size = 0.90) of the stance phase. Coordination variability was significantly lower in the treadmill-on-demand than in the overground condition during 0-25% ($p = 0.042$, effect size = 0.94), 26-50% ($p < 0.001$, effect size = 1.72), 51-75% ($p < 0.001$, effect size = 1.60) and 76-100% stance period ($p < 0.001$, effect size = 1.16) of the stance phase. Similarly to the hip flexion-knee flexion coupling, coordination variability was not significantly different between the treadmill and treadmill-on-demand during any stance phases (p 's > 0.421 , effect sizes < 0.23).

6.3.3 Knee flexion-rearfoot inversion coupling

The descriptive statistics for the mNoRMS values for the knee flexion-rearfoot inversion coupling averaged over the entire stance period are also given in Table 6.1. Results of the ANOVA for the knee flexion-rearfoot inversion coupling indicated a significant main effect for the condition factor (see Table 6.2). Pair-wise follow-up tests indicated that mNoRMS values averaged across the entire stance phase were significantly higher in the overground condition than in the treadmill ($p = 0.003$, effect size = 1.08) and treadmill-on-demand ($p = 0.008$, effect size = 1.30) condition, for the knee flexion-rearfoot inversion coupling. However, coordination variability was not significantly different between treadmill and treadmill-on-demand conditions ($p = 0.838$, effect size = 0.05). Descriptive statistics for the knee flexion-rearfoot inversion coupling over the different stance phases are displayed in Figure 6.5 (bottom panel). From inspection of Figure 6.5 it is apparent that there was a pattern of greater coordination variability in the overground condition in all phases of the stance period compared to the treadmill and treadmill-on-demand conditions.

Post-hoc tests performed on the selected comparisons for the different phases of stance indicated various significant differences. The overground condition exhibited significantly greater coordination variability than the treadmill condition during 0-25% ($p = 0.001$, effect size = 1.17), 26-50% ($p = 0.008$, effect sizes = 0.77), 51-75% ($p = 0.037$, effect size = 0.84) and 76-100% ($p = 0.016$, effect size = 1.22) of the stance phase. Treadmill-on-demand locomotion was associated with significantly lower coordination variability than overground running during 0-25% ($p = 0.022$, effect size = 1.11), 51-75% ($p = 0.013$, effect size = 1.23) and 76-100% ($p = 0.005$, effect size = 1.43) of the stance phase. Differences between the treadmill-on-demand and overground conditions were not significant during 26-50% ($p = 0.076$, effect size = 0.81) of the stance phase. Coordination variability was not significantly different between the treadmill and treadmill-on-demand during any stance phases (p 's > 0.222, effect sizes < 0.41).

6.4 Discussion

The purpose of this study was to investigate differences in coordination variability measured during overground, conventional treadmill and treadmill-on-demand running. In comparison to overground running, significantly reduced coordination variability ($p < 0.05$) was observed in the treadmill and treadmill-on-demand conditions, over the entire stance period as well as various phases of stance, for all joint couplings. Further, the effect size statistics indicated that, for these

comparisons⁹, differences between overground and both treadmill (effect sizes: 0.84-1.71) and treadmill-on-demand (effect sizes: 0.94-1.95) running were large. Indeed, even during the period of stance in which the difference between overground and the treadmill-on-demand conditions was non-significant, the effect size statistic indicated that the magnitude of the difference was large.

The decreased variability in lower extremity coordination during treadmill running seen in this study is consistent with previous investigations (Nelson *et al.*, 1972; Wank *et al.*, 1998; Dingwell *et al.*, 2001) and Chapter V. The artificially constant speed of the treadmill belt was cited in section 5.4 as a potential reason for the observed differences in coordination variability between overground and treadmill locomotion. In addition to the conventional treadmill, an innovative treadmill was studied in this Chapter. The treadmill-on-demand was recently introduced by Minetti *et al.* (2003) and its speed is continuously changed according to the participant's preference to accelerate, decelerate or keep a constant speed (see Figure 6.1). Consequently, in the present investigation, it was hypothesised that coordination variability measured on the treadmill-on-demand, in which the belt speed is not constant, would better resemble that measured overground than the conventional treadmill. However, the results of this investigation do not support this hypothesis as the differences between the treadmill-on-demand and conventional treadmill were non-significant ($p<0.05$) and effect sizes were small (0.00-0.41).

⁹ With the exception of the 26-50% stance period for the comparison of the knee flexion-rearfoot inversion coupling between overground and treadmill running, for which the effect size was 0.77 – a moderate effect.

Further, the differences in coordination variability between overground and both treadmill (effect sizes: 0.77-1.71) and treadmill-on-demand (effect sizes: 0.94-1.95) running were similar in magnitude. These effect sizes are also similar in magnitude to those that were presented for the differences in overground and treadmill coordination variability in Chapter V. The results presented in this Chapter, therefore, suggest that the constant speed of the treadmill belt during conventional treadmill locomotion does not account for the differences in coordination variability seen between overground and treadmill locomotion. As it appears that the constant treadmill belt speed is not accountable for the differences in coordination variability consistently observed between overground and treadmill locomotion, some other factor or factors must be responsible.

Various factors have been implicated as potential reasons for differences between overground and treadmill kinematics which could also be responsible for the differences in coordination variability. Examples include differences in the mechanical characteristics of the treadmill and overground surfaces (Frishberg, 1981; Dingwell *et al.*, 2001) and reductions in air resistance experienced by the participants in the treadmill condition (Frishberg, 1981; van Ingen Schenau, 1980). However, there is no evidence in the literature to suggest how changes in these parameters might affect coordination variability. Also, Savelberg (1998) suggested that the differences observed between overground and treadmill kinematics could be explained by changes in the speed of the treadmill belt throughout the stance phase of treadmill locomotion. Again, however, neither Savelberg *et al.* (1998) nor any other author have reported the effects of intra-stride belt speed variations on

coordination variability. Because of the lack of evidence in the literature it is not possible to determine what the effect of these factors might be on coordination variability.

The differences between overground and treadmill coordination variability might also be related to the suggestion of Holt *et al.* (1995) that, potentially, coordination variability provides an adaptive mechanism to possible external perturbations such as uneven ground. It seems reasonable that, during treadmill locomotion, there is less threat of an external perturbation and, therefore, less of a requirement for coordination variability. A further viable explanation for the differences in coordination variability between the modes of locomotion is the altered perceptual information available during treadmill running. As van Ingen Schenau (1980) highlighted, during overground running the surroundings move with respect to the participant, which is not the case during treadmill locomotion. A fundamental notion of ecological theories of movement control is the idea that perception and action are tightly coupled - see Williams *et al.* (1999) for a detailed review. From this theoretical standpoint it should be expected that a change in the perceptual optical flow information would result in a change in the action or outcome of the movement, i.e. coordination. As suggested in section 5.4, limited evidence exists in the literature regarding the effects of reducing optical flow information on locomotion variability (e.g. Masson and Pailhous, 1994). An interesting direction for future research would be to evaluate the effects on coordination variability of introducing optical flow information, comparable to that in overground locomotion, into a treadmill running condition. It is clear that further work is required to

determine which, if any, of these factors – that were consistent across treadmill and treadmill-on-demand conditions – are responsible for the differences in variability between treadmill and overground locomotion.

As the treadmill and treadmill-on-demand conditions were not significantly different, the implications of the findings of this study for the treadmill-on-demand are very similar to those formulated in Chapter V. The first is related to the hypothesised link between variability and overuse injury (James, 2004) which was reviewed in section 2.5.1. Briefly, James (2004) postulated that variability might play a positive role in preventing overuse injuries. Further, variability might provide a broader distribution of stresses among different tissues or alter the stress magnitude, direction and frequency within the same tissue (James, 2004). In other words, a person might be at an increased risk of overuse injury if they exhibited less locomotion variability. In Chapter V, it was suggested that, due to decreased variability, runners might be at a greater risk of overuse injury during treadmill locomotion than during overground locomotion. It seems that this also applies to running on the treadmill-on-demand. However, various confounding factors were detailed in section 5.4 that dictate this conclusion should be made with caution. Briefly, it should be noted that no conclusive evidence exists to link decreased variability with overuse injury and a prospective study is required to establish cause and effect (Hamill *et al.*, 1999; Heiderscheit *et al.*, 2002). Also, Milgrom *et al.* (2003) recently reported that loads on the tibia were significantly reduced during treadmill compared to overground running. Therefore, although the higher variability during overground running will, potentially, better distribute the loads

over different internal structures, it also appears the magnitude of the loads is reduced during treadmill locomotion.

In their paper, Minetti *et al.* (2003) suggested that the treadmill-on-demand would be useful whenever reliable measurements of the spontaneous speed of locomotion are required. Examples of such studies include those related to the transition between walking and running, the economy of gait and the effects of load carriage on gait mechanics (*c.f.* Minetti *et al.*, 2003). However, the treadmill-on-demand does not appear to be superior to the conventional treadmill for the study of coordination variability. Similarly to the conventional treadmill, using the treadmill-on-demand to monitor changes in coordination variability might mask differences between experimental groups in certain circumstances – see section 5.4 for a detailed discussion of the circumstances in which this might arise.

A possible limitation of this study is related to the length of the habituation period the participants completed on the treadmill-on-demand. Many authors have postulated that the differences seen between treadmill and overground kinematics might be due to the participants not being habituated to treadmill locomotion (e.g. Charteris and Taves, 1978; Wall and Charteris, 1981; Scheib, 1986; Matsas *et al.*, 2000; Wass *et al.*, 2004). Potentially, this would not account for the differences seen in this study, as all participants were experienced treadmill runners and were given time to become accustomed to the environment before testing began. Further, the time provided for the participants to become accustomed to treadmill-on-demand locomotion was more than has been deemed adequate for

conventional treadmill habituation (Matsas *et al.*, 2000). Additionally, at the end of the habituation period participants confirmed that they were comfortable running on the treadmill-on-demand. However, it is possible that the ten minutes provided was not sufficient to fully habituate the participants to treadmill-on-demand locomotion. Unfortunately, no studies have determined what is sufficient to fully habituate participants to the treadmill-on-demand and none have investigated the effect of a lack of participant familiarity on coordination variability. The process of becoming habituated to treadmill-on-demand locomotion and the effects of incomplete acclimatisation to this and conventional treadmill locomotion on coordination variability would be interesting avenues for future research.

A further possible limitation of this study is related to the way in which running speed was controlled in both the overground and treadmill-on-demand conditions. Because running speed was not constrained to be constant in the overground and treadmill-on-demand conditions a boundary of \pm 5% of $3.5 \text{ m}\cdot\text{s}^{-1}$ was used. It is possible that, in the overground condition, participants explored this bandwidth of acceptable speeds to a greater degree than in the treadmill-on-demand condition – effectively meaning the speeds were more consistent during treadmill-on-demand trials than the overground trials. If this was the case, the lower variability in the treadmill-on-demand condition might have been due to the more consistent running speed than in the overground condition. Unfortunately, running speeds were not recorded – they were only inspected at the time of testing – so this possibility can not be confirmed or refuted. However, treadmill-on-demand trials often had to be discarded – due to the participant's running speed straying outside of the \pm 5%

boundary of acceptable speeds – suggesting that participants were indeed exploring the full extent of the boundary of acceptable speeds. Also, it is reasonable to assume that, if the running speed variable was important, the data would show a trend towards higher coordination variability in the treadmill-on-demand condition than the treadmill condition whilst still being lower than the overground condition – as running speed in the conventional treadmill condition was entirely consistent. However, no such trend was apparent in the data presented in this Chapter suggesting that factors other than running speed – as outlined previously in this section – were responsible for the differences seen between the locomotion conditions.

6.4.1 Conclusion

Within its scope and limitations, this study has provided evidence to support the rejection of the hypothesis that the treadmill-on-demand would better resemble overground locomotion in terms of coordination variability than the conventional treadmill. It appears that the constant speed of the treadmill belt is not responsible for the lower variability consistently observed during treadmill than in overground locomotion. Further work is required to determine the cause of the difference. A particularly interesting follow-up study would be to investigate the effects on coordination variability of introducing optical flow information, comparable to that in overground locomotion, into a treadmill running condition.

CHAPTER VII

7 Summary and overall discussion

Over the past 25 years there has been a change in the conceptualisation of movement variability. Traditionally thought to be noise or error in data, it is now accepted, by some authors, that variability is inherent, and it should be expected to some degree in all human movement. Indeed, research presented in sections 2.4 and 2.5 highlighted potentially functional roles for variability in a wide variety of human systems. An interesting functional role for variability has recently been hypothesised in which a relationship between movement variability and overuse injury was proposed – see section 2.5.1. Briefly, James (2004) suggested that variability might play a positive role in preventing overuse injuries. Variability might provide a broader distribution of stresses among different tissues or alter the stress magnitude, direction and frequency within the same tissue (James, 2004). Potentially, a runner might be at an increased risk of overuse injury if they exhibited less locomotion variability. There is limited experimental evidence to support this hypothesis. For example, participants with a history of lower extremity injury exhibited less variability in various internal joint moment dependent variables than participants with no history of lower extremity injury in a drop jumping task (James *et al.*, 2000). Also, individuals with patellofemoral pain syndrome have been reported to exhibit less coordination variability than matched controls (Hamill *et al.*, 1999; Heiderscheit *et al.*, 2002).

As many authors have highlighted, more work is required to further investigate the link between movement variability and overuse injury (e.g. Hamill *et al.*, 1999; Heiderscheit, 2000a,b; James *et al.*, 2000; Heiderscheit *et al.*, 2002; James, 2004). However, as highlighted in Chapter II, this new area of research is associated with many methodological considerations that warrant appraisal. These include issues related to measurement errors, the suitability of techniques for quantifying the variability in coordination and the use of a treadmill to simulate overground locomotion variability. The overall purpose of this thesis was to address these issues.

This Chapter provides an overall discussion and summary of this thesis of studies and is divided into four sections. First, a brief summary of Chapters III, IV, V and VI is provided. Second, section in which the implications for future studies of the relationship between variability and overuse injury is given. Third, a discussion of the limitations of the thesis is provided which is followed by suggestions for future work in the area. Finally, a conclusion is presented.

7.1 Summary and implications of individual Chapters

7.1.1 Chapter III

Chapter III addressed issues related to errors in the measurement of human movement. As stated in section 2.6.3, the total variability measured in any system is made up of three components: inherent human variability, variability due to changing constraints on movement and variability due to measurement error. In a

well-controlled research environment, the variability due to changing constraints on movement will be minimal and relatively easy to isolate. To be able to address research questions related to variability, it is important to be able to separate the inherent human movement variability from that which is due to measurement error. Therefore, an in-depth account of the nature of the errors in the measurement of kinematics using skin-based markers was presented in Chapter III.

When data are collected using skin-based markers, errors in the measurements originate from inaccuracies of the measurement system (instrument errors) and the movement of the markers on the skin relative to the underlying bone (skin movement errors). The Chapter was divided into two sections to address both aspects of measurement error. First, the performance of the measurement system used in this thesis of studies was assessed using a testing device and procedure similar to that presented by Richards (1999). The results demonstrated that assessing the accuracy and precision of a motion capture system in only one position in the measurement volume is inadequate. Further, the results indicated that the accuracy of the measurement system used in this thesis was greater than the systems investigated by Richards (1999). It was also evident that the errors due to the measurement system were considerably smaller than those due to skin marker movement presented in the second half of Chapter III. Because studies of skin marker movement errors are inherently very invasive, an experimental study of these errors was not possible in this thesis due to ethical reasons. Instead, an in-depth review of the literature in the area was conducted. This review highlighted that, especially in the secondary planes of motion, the magnitude of skin marker

movement errors is large. For example, Cappozzo *et al.* (1996) suggested that knee kinematics might be affected by inaccuracies which amount to 50% and 100% of adduction-abduction and internal-external rotation respectively.

At the end of the Chapter, implications for the study of variability were discussed. It was suggested that skin movement errors have a far greater influence on studies of locomotor variability than instrument errors. In fact, when measurement devices of similar resolution and accuracy to the system used in this programme of research are used, instrument errors might be considered negligible in relation to errors due to skin movement. Further, it was recognised that skin movement errors could be extremely detrimental to the study of locomotion variability if they were completely random. Conversely, if these errors were entirely systematic they would not affect the observed variability. Preliminary evidence presented in Chapter III suggested that skin movement errors were predominantly systematic, making them less of a problem for studies of variability. However, further work is required to substantiate these findings.

7.1.2 Chapter IV

Many techniques have been used previously to quantify coordination and coordination variability. Each technique is associated with advantages and disadvantages for quantifying variability in coordination – see section 2.6.2. In Chapter IV, the results of an investigation of the effect of an unstable shoe construction on coordination variability were provided. By focusing on main effects and descriptive statistics in this investigation, the aim of Chapter IV was to highlight

the potential for researchers to obtain disparate results depending on the coordination quantification technique used.

Chapter IV highlighted that different techniques for quantifying coordination variability might provide different answers to some research questions. It was suggested that these differences were related to the small disparities in the calculation of each technique. It seems clear that, in agreement with other authors (Fuchs and Kelso, 1994; Rosenblum and Kurths, 1998; Peters *et al.*, 2003), continuous relative phase should be used with caution in studies of coordination variability during locomotion. It is also clear from Chapter IV that comparisons between studies of coordination variability that have used different analysis techniques should be made with caution. Authors should make clear exactly what technique was used and how it was calculated, to enable an informed reader to make appropriate judgments about the study and compare the results to similar work. Further, as Hamill *et al.* (2000) suggested, before choosing a particular technique, the researcher should be aware of the benefits and limitations of each and understand which is most suited to the movement or activity of interest. Finally, it was suggested that the mNoRMS technique was the most suitable for use in Chapters V and VI of this thesis.

7.1.3 Chapter V

Some studies of coordination variability during locomotion cited in the literature were conducted overground (Ferber *et al.*, 2005) whilst others used a treadmill (Diedrich and Warren, 1995; Hamill *et al.*, 1999; Field-Fote and Tepavac, 2002;

Kao *et al.*, 2003). Obviously, there is a problem of ecological validity if differences in coordination and coordination variability exist between the two modes of locomotion. However, no studies had examined differences in coordination variability between overground and treadmill running. Therefore, the purpose of Chapter V was to assess differences in the variability of lower extremity joint couplings between overground and treadmill running.

Reduced coordination variability during treadmill locomotion was hypothesised and the results presented in Chapter V support this. Specifically, treadmill running was associated with statistically significant reductions in coordination variability in the hip flexion-ankle dorsiflexion, hip flexion-knee flexion and knee flexion-ankle inversion couplings over the entire stance period and during various phases of stance. These results were consistent with a previous investigation conducted by Dingwell and colleagues (2001) who suggested that the variability in sagittal plane ankle and knee angles was significantly reduced during treadmill compared to overground walking. Various reasons for the observed differences were cited in section 5.4. These included, for example, intra-stride belt speed variations, changes in air resistance, changes in perception of the threat of an external perturbation and changes in optical flow information. An additional reason - which was also presented by Dingwell *et al.* (2001) - was the treadmill belt imposing an artificially constant speed, externally driving the participant's feet throughout the stance phase of each stride cycle.

Whether the reasons for the differences in coordination variability between the two modes of locomotion were related to changes in the mechanical constraints, perceptual information or any other factors, there are important implications of the findings of the study in Chapter V. First, it is possible that the reduced variability in the treadmill condition will result in an increased risk of overuse injury in this mode of locomotion. The higher variability in the overground condition, potentially, results in broader distribution of stresses among different tissues altering the stress magnitude, direction and frequency within internal structures. However, the hypothesised link between variability and injury is yet to be strongly supported experimentally - see sections 2.5.1 and 5.4. Second, it is possible that the artificially constant speed of the treadmill belt might constrain the movement variability. Therefore, performing studies of joint coordination variability on a motorised treadmill might mask differences between experimental groups. Caution should be applied when comparing results from studies of variability in coordination using overground and treadmill analysis.

7.1.4 Chapter VI

In the final experimental Chapter of this thesis, coordination variability was assessed on an innovative type of treadmill and compared to overground and conventional treadmill running. The 'treadmill-on-demand' is a feedback-controlled treadmill, the speed of which is continuously changed according to the participant's preference to accelerate, decelerate or keep a constant speed. On this type of treadmill, the participants are not constrained to run at a constant speed. Therefore, it was hypothesised that, in terms of coordination variability, running on

the treadmill-on-demand would better resemble overground running than running on the motorised treadmill. However, the results did not support this hypothesis; the differences between the treadmill-on-demand and conventional treadmill were non-significant and the magnitudes of the differences were small. Further, the significant differences in coordination variability between overground and both treadmill and treadmill-on-demand running were similar. Therefore, the results presented in Chapter VI suggested that the constant speed of the treadmill belt during conventional treadmill locomotion does not account for the differences in coordination variability seen between overground and treadmill locomotion.

As the treadmill-on-demand and the conventional treadmill appear to be similar in terms of coordination variability, the implications for the use of the treadmill-on-demand are comparable to those presented in Chapter V for the conventional treadmill. Finally, the treadmill-on-demand appears to be no more suited to studies of coordination variability than the conventional treadmill.

7.2 Implications of findings for future studies of the variability-overuse injury hypothesis

The previous subsections provide a concise summary of the experimental chapters of this thesis. Specifically, this thesis of studies has addressed methodological considerations pertinent to the study of variability in coordination during running. There are clear implications of the results of this thesis for studies of coordination variability. However, as any given result of the investigations in this thesis could

have different meaning for studies with different research questions, a context was provided. As was reviewed in section 2.5.1, James (2004) recently presented a hypothesis linking movement variability and overuse injury. The future investigation of the variability–overuse injury hypothesis was chosen as a context for the thesis and provided a framework within which to interpret the results.

The results of the investigations presented in Chapters III, IV, V, and VI have implications for further work investigating the variability-overuse injury hypothesis. First, it is clear from the data presented in Chapter III that measurement errors are an extremely important consideration. Measurement errors would be extremely detrimental to future studies of the variability-overuse injury hypothesis if they were entirely random. It is clear from the data presented in Chapter III that, relative to errors due to skin marker movement, errors due to the instrument – the motion capture system in all studies of this thesis – are of a much smaller magnitude and can be considered to be negligible. Evidence in the literature – presented in section 2.6.3 – suggests that, certainly in the secondary planes of motion, errors due to skin marker movement can be up to 100% of actual segment motion. Obviously, if these errors were entirely random, it could not be concluded with certainty that any observed coordination variability was inherent human movement variability or solely due to error. Further, it could not be stated with certainty that any differences observed between an injured group and a control group, for example, was a real difference. Therefore, the systematic or random nature of skin marker movement errors is an extremely important consideration for any study of the variability-overuse injury hypothesis. Results of a preliminary analysis of data presented by

Holden *et al.* (1997) suggested that skin marker movement errors are primarily systematic which would mean that these error are not a large problem for future studies of the variability-overuse injury hypothesis. It should be noted that further work is required to substantiate these preliminary findings. However, it is certainly clear that researchers in future studies of the relationship between variability and overuse injury should use a marker set developed with the reduction of skin marker movement error as the prevailing design criteria. As highlighted in section 2.6.3, Manal *et al.* (2000) identified the marker set design that was used in Chapters V and VI of this thesis as the most effective. Therefore, it is suggested that a marker set similar to this be used in future studies investigating the relationship between coordination variability and overuse injury.

It is clear from the review of techniques available for the quantification of coordination variability presented in section 2.6.2 that each technique has both advantages and disadvantages. It is also apparent from the results presented in Chapter IV that different answers to some research questions might be obtained dependent on the choice of technique used to quantify coordination variability. An important implication of this finding is that, before choosing a particular technique, the researcher should have a sound understanding of the calculation of each technique and have knowledge of their benefits and limitations. For the reasons cited in section 4.4, mNoRMS was chosen as the technique for quantifying coordination variability in Chapters V and VI of this thesis - which were carried out in the context of the variability overuse injury. The mNoRMS would appear to be very useful for future studies of the relationship between variability and overuse

injury. However, this is not to suggest that mNoRMS would be most suitable for addressing all research questions. Indeed, the major implication of the findings presented in Chapter IV is that the choice of technique used to quantify coordination variability should be based on the research question and activity of interest.

As summarised in section 7.1.3, Chapter V of this thesis investigated the degree to which treadmill running simulates overground locomotion in terms of coordination variability. It was reported that treadmill running is associated with significantly lower coordination variability than overground running. This has important implications for future studies of the link between variability and overuse injury. First, it would appear that caution should be applied when comparing the results of studies investigating the link between overuse injury and variability using treadmill and overground analysis. Certainly, the ecological validity of any future studies using a treadmill to investigate the variability-overuse injury hypothesis could be questioned.

Linked to this, it is also possible that any differences between experimental groups in future studies of the relationship between variability and overuse injury might be masked if the study were conducted on a treadmill. An example of how this might manifest is clear in the work of Dingwell and colleagues (Dingwell *et al.*, 1999; Dingwell *et al.*, 2000). Dingwell *et al.* (1999) assessed kinematic variability in patients with diabetic neuropathy and matched controls walking on a treadmill and reported that the differences between groups were not significant. However,

differences between groups were statistically significant when a similar cohort of participants walked overground (Dingwell *et al.*, 2000). It is possible that any differences between injured and control groups that might be apparent during overground locomotion are not seen during treadmill locomotion because coordination variability is artificially reduced on the treadmill. Any future study of the relationship between coordination variability and overuse injury should be mindful of the potential for treadmill analysis to mask differences in coordination variability between experimental groups.

In Chapter VI of this thesis differences, in terms of coordination variability, between overground and treadmill running were again investigated. However, because in Chapter V it was suggested that the artificially constant speed of the treadmill belt might be a reason for the differences seen between the two modes of locomotion, a further, novel type of treadmill was included in the analysis. The treadmill-on-demand is a new type of treadmill first introduced by Minetti *et al.* (2003) on which the participant is not constrained to run at a constant speed. Using an ultra-sonic range finder interfaced with a computer, the treadmill-on-demand is able to regulate the treadmill belt speed dependent on the participant's preference to accelerate, decelerate or keep a constant speed. Because of this, it was hypothesised that the treadmill-on-demand would better simulate coordination variability measured overground than the conventional treadmill. However, the data did not support this hypothesis. Indeed, there appeared to be no differences between coordination variability measured on the treadmill-on-demand than the conventional treadmill. Consequently, the treadmill-on-demand would seem no

more suited to the future study of the variability-overuse injury hypothesis than the conventional treadmill. Certainly, the problems outlined previously in this section regarding the use of the conventional treadmill for the study coordination variability apply to the use of the treadmill-on-demand in a similar setting.

In this section, the implications of the results presented in this thesis for future studies of the hypothesised link between variability and overuse injury have been explicitly identified. Hopefully, these results will be useful for authors who wish to perform such investigations.

7.3 Limitations

There are factors which might have influenced the results of the studies in each Chapter of this thesis. The first is specific to Chapter III. Unfortunately, due to ethical reasons, it was not possible to conduct a study to investigate the effects of the, relatively large, skin marker movement errors on measurements of coordination variability. Preliminary evidence suggested that skin marker movement errors were predominantly systematic. However, these results should be treated with caution because it was not possible to determine whether the data presented by Holden *et al.* (1997) met the assumptions of the limits of agreement analysis used.

Further, there are factors associated with Chapters IV, V and VI, the first of which relates to the statistical techniques used to quantify variability. All techniques to

quantify coordination variability used in this thesis were based on traditional measures of variability such as standard deviation and coefficient of variation. Various authors in movement coordination and motor control have advocated that studies investigate the structure rather than solely the magnitude of variability (*c.f.* Newell and Corcos, 1993). Others have suggested that, in certain circumstances, traditional techniques for quantifying variability should not be used because they might mask a potentially deterministic structure (Slifkin and Newell, 1998; Dingwell and Cavanagh, 2001; Riley and Turvey, 2002). In certain contexts such as the investigation of neuromuscular factors associated with aging and Huntington's disease (e.g. Hausdorff *et al.*, 1997b) and the effects of diabetes mellitus on the stability of movement (e.g. Dingwell *et al.*, 2000), the structure, rather than the magnitude, of the variability is important. However, this thesis was concerned with quantifying variability in coordination during locomotion in a manner that would be relevant to the investigation of the overuse injury-variability hypothesis. In this context, when the magnitude of the variability is important, the traditional techniques - upon which the techniques used to quantify coordination variability in this thesis are based - were most relevant. Further, in this thesis, only the stance phase was studied which also precluded the use of non-linear techniques.

Another potential limitation of the research presented in this thesis relates to the way in which data were collected during overground running. As opposed to the treadmill conditions in which data were collected from ten consecutive strides, non-consecutive strides were recorded in the overground condition. Some authors have highlighted that this is not appropriate (e.g. Dingwell *et al.*, 1999). However, a

justification for claiming it was inappropriate was that information about the structure of variability is lost when non-consecutive strides are analysed. For reasons cited above, however, this information was not required in this thesis, making the way in which data were collected in the overground condition less problematic. Finally, it is important to note that Dingwell *et al.* (2001), who measured consecutive strides in both the treadmill and overground walking conditions, reported results similar to those presented in Chapters V and VI of this thesis.

Last, it should be noted that the samples of participants used in all studies in this thesis were taken from a population of young, healthy adults. Therefore, some of the results should be applied to other populations, such as elderly and pathological groups, with caution.

7.4 Future directions

The results of the current series of studies also provide direction for future work. It is clear from Chapter III that there is a requirement for a study to ascertain whether the errors due to skin marker movement are systematic or random. Such a study should employ statistical techniques such as least products regression and limits of agreement which have been deemed appropriate for this type of analysis (Bland and Altman, 1986; Ludbrook, 1997). Additionally, more detailed investigation of the research question formulated in Chapter IV is warranted. Specifically, further

investigation of the effects of an unstable shoe construction on variability in coordination during locomotion is required.

Future work is also required to determine the cause of the consistent observation that treadmill locomotion is associated with significantly reduced locomotion variability than overground locomotion. For example, additional work is needed to determine the effects of factors such as the mechanical characteristics of the treadmill surface compared to overground surfaces, intra-stride belt speed variations and treadmill habituation on the differences between overground and treadmill coordination variability. A particularly interesting follow-up study would be to investigate the effects on coordination variability of introducing optical flow information, comparable to that in overground locomotion, into a treadmill running condition. Further, more work is required to ascertain whether the decreased variability observed during treadmill locomotion means that performing studies of variability on a treadmill masks differences between experimental groups. Dingwell *et al.* (2001) suggested that this is a problem when factors related to neuromuscular control and its relationship with variability are under investigation. However, it is unclear whether differences in coordination variability that are the result of changed mechanical constraints – including orthotic devices, bracing and footwear - would be concealed. The effects of treadmill analysis on changes in variability due to the addition of mechanical constraints would also be an interesting avenue for future research.

7.5 Conclusion

The purpose of this thesis was to address issues related to the methodological considerations for studies of variability in coordination during running. Specifically, these methodological considerations included issues related to measurement errors, the suitability of techniques for quantifying the variability in coordination and the use of a treadmill to simulate overground locomotion variability. To investigate the effects of these factors on coordination variability during running, four studies were reported in this thesis. Several important findings were noted in each study, which were summarised in sections 7.1.1, 7.1.2, 7.1.3 and 7.1.4. Although some unanswered questions remain, this series of studies has enhanced understanding of important methodological considerations for the study of coordination variability during running. It is hoped that the results presented in this thesis will be useful in future studies of variability in coordination during running and other activities.

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Appendix A: Joint Coordinate Systems

Hip, knee and ankle Joint Coordinate System (JCS) angles were calculated from the filtered data using MARey software (Cavanagh *et al.*, 2001), written for Matlab. The axis systems used by the MARey software in the calculation of the JCS angles are based on the standardisation paper of Cole *et al.* (1993) in which the first axis is the flexion axis of the proximal (reference) segment, the second axis is the longitudinal axis of the distal (target) segment and the third (floating) axis is the cross product of the second by first axes. In order for the JCS to be defined at each joint of interest, orthogonal segment coordinate systems for the foot, shank, thigh and pelvis were calculated using the data from the static trials. It should be noted that the segment coordinate systems are defined individually for the different joints in order to make the JCS as close to anatomical rotations as possible. For example, the segment coordinate system for the shank for use in the ankle JCS is different to the shank segment coordinate system used in the knee JCS. When calculating ankle joint motion, the shank is the reference segment so it is crucial that the flexion/extension axis is correctly aligned. However, when knee joint motion is of interest the shank is the target segment and it is therefore crucial that the longitudinal axis is aligned correctly. The same applies to the thigh segment when it is involved in the calculation of both knee and hip motion.

Joint Coordinate Systems and rotation calculation

The following section outlines the details of the segment coordinate systems and subsequently the JCSs used by the MARey software (Cavanagh *et al.*, 2001) at the ankle, knee and hip on the right side of the body¹⁰. Subsequently, the specifics of the mathematics used to calculate the rotations at each of the joints are described.

Nomenclature and definition of terms

X,Y,Z =	The three orthogonal axes of the proximal/reference coordinate system. Oriented approximately anterior-posteriorly, inferior-superiorly and medio-laterally respectively (all with the body in the anatomically neutral position)
x,y,z =	The three orthogonal axes of the distal/target coordinate system. Oriented approximately anterior-posteriorly, inferior-superiorly and medio-laterally respectively (all with the body in the anatomically neutral position)
e ₁ ,e ₂ ,e ₃ =	First, second and third axes of the non-orthogonal Joint Coordinate System (JCS: Grood and Suntay, 1983).
F,L,T =	Alternative labeling of segment axes, proposed by Cole <i>et al.</i> (1993). The flexion, longitudinal and third axes of the segment respectively.
MM:	Distal apex of the medial malleolus
LM:	Distal apex of the lateral malleolus
IM:	The inter-malleolar point located midway between the MM and LM (ankle joint centre)
MC:	The most medial point on the border of the medial tibial condyle
LC:	The most lateral point on the border of the lateral tibial condyle
IC:	The inter-condylar point located midway between the MC and LC (knee joint centre)
⊗	Segment centre of mass (CoM)
GT:	Greater trochanter
HJC:	Hip joint centre
ASIS:	Anterior superior iliac spine
PSIS:	Posterior superior iliac spine
NP:	Intermediate hip joint centre
IASIC:	The inter-ASIS point located midway between the RASIS and LASIS
IPSiS:	The inter-PSIS point located midway between the RPSIS and LPSIS

¹⁰ Relevant transformations were performed in the studies in which the left side of the body was studied.

The definitions of the shank and foot coordinate systems used in the ankle JCS are summarised in Figure A.1:

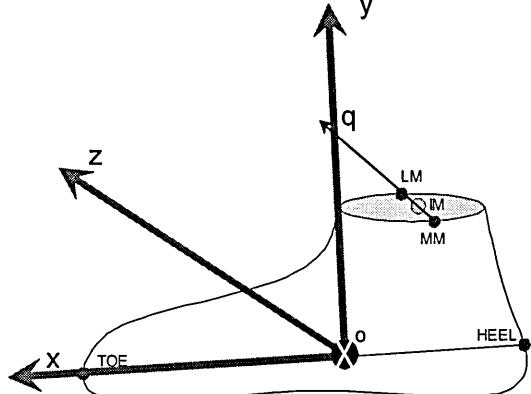
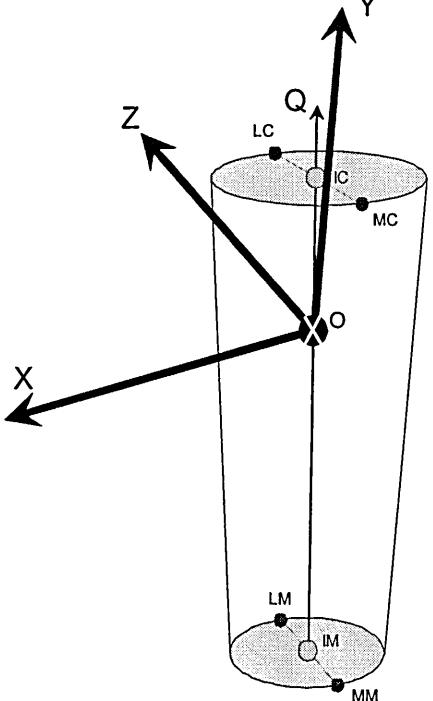
	
<p>o: Origin: The point coincident with the segment COM</p> <p>q: An intermediate vector passing between MM and LM and directed to the right</p> <p>x: x-axis: A vector coincident with the line passing from the Heel to the Toe</p> <p>y: y-axis: The cross product of $q \times x$</p> <p>z: z-axis: The cross product of $x \times y$</p>	<p>O: Origin: The point coincident with the segment COM.</p> <p>Q: An intermediate vector passing from IM to IC</p> <p>X: X-axis: The cross product of $Q \times Z$</p> <p>Y: Y-axis: The cross product of $Z \times X$</p> <p>Z: Z-axis: The line coincident with that passing through MM and LM, and pointing from left to right</p>

Figure A.1: The definition of the foot (left) and shank (right) segment coordinate systems for use in the ankle JCS.

Subsequently, the ankle JCS was defined as follows (Figure A.2):

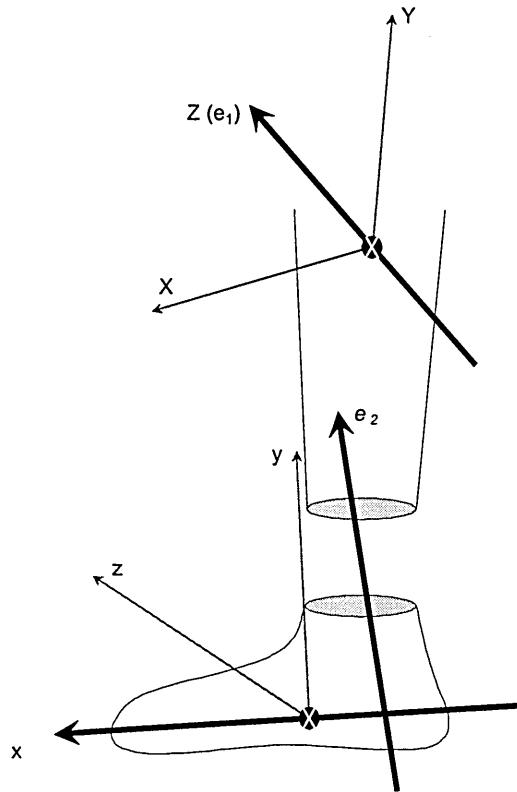


Figure A.2: The ankle Joint Coordinate System

Where:

- e_1 : Z-axis of the shank coordinate system - flexion/extension axis
- e_2 : The floating axis, defined as the cross product of the e_1 and e_3 axes - internal/external rotation (abduction/adduction) axis
- e_3 : x-axis of the foot coordinate system - inversion/eversion axis

The shank and thigh segment coordinate systems used in the knee JCS to calculate knee motion are given in Figure A.3:

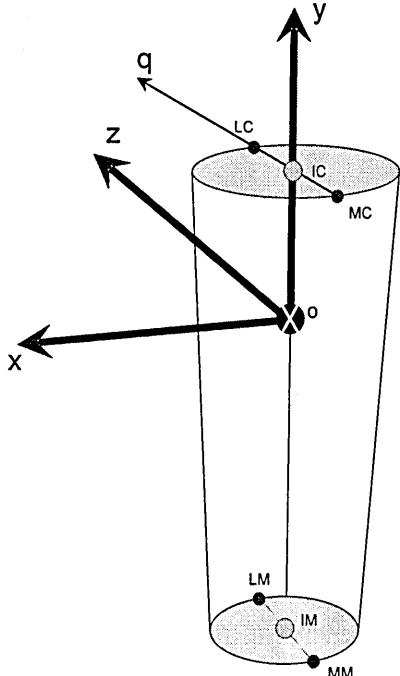
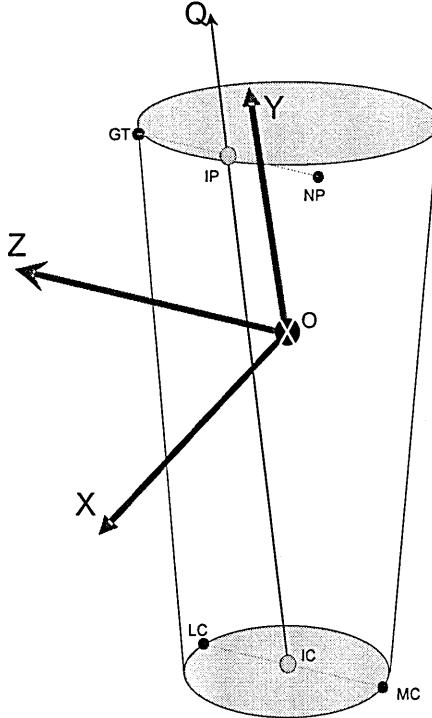
	
<p>o: Origin: The point coincident with the segment COM</p> <p>q: An intermediate vector passing between MC and LC and directed to the right</p> <p>x: x-axis: The cross product of $y \times q$</p> <p>y: y-axis: The line coincident with that passing through IM and IC, directed anteriorly</p> <p>z: z-axis: The cross product of $x \times y$</p>	<p>o: Origin: The point coincident with the segment COM.</p> <p>Q: An intermediate vector passing from IC to IP</p> <p>X: X-axis: The cross product of $Q \times Z$</p> <p>Y: Y-axis: The cross product of $Z \times X$</p> <p>Z: Z-axis: The line coincident with that passing through MC and LC, and pointing from left to right</p>

Figure A.3: The definition of the shank (left) and thigh (right) segment coordinate systems for use in the knee JCS.

Subsequently, the knee JCS was defined as follows (Figure A.4):

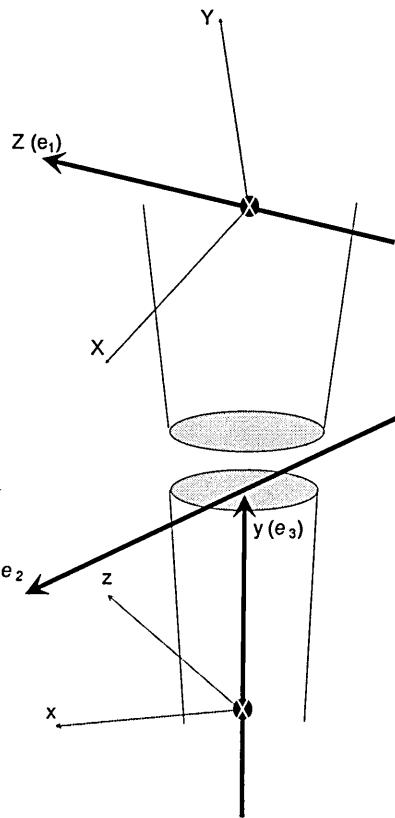


Figure A.4: The knee Joint Coordinate System

Where:

- e_1 : Z-axis of the thigh coordinate system - flexion/extension axis
- e_2 : The floating axis, defined as the cross product of the e_1 and e_3 axes - abduction/adduction axis
- e_3 : y-axis of the shank coordinate system - axial rotation axis

The thigh and pelvis segment coordinate systems used in the hip JCS to calculate hip motion are given in Figure A.5:

o: Origin: The point coincident with the segment COM q: An intermediate vector passing between MC and LC and directed to the right x: x-axis: The cross product of $y \times q$ y: y-axis: The line coincident with that passing through IC and HJC, directed anteriorly z: z-axis: The cross product of $x \times y$	O: Origin: The point coincident with the segment COM. Q: An intermediate vector passing from IPSIS to IASIS X: X-axis: The cross product of $Y \times Z$ Y: Y-axis: The cross product of $Z \times Q$ Z: Z-axis: The line coincident with that passing through LASIS and RASIS, directed from left to right

Figure A.5: The definition of the thigh (left) and pelvis (right) segment coordinate systems for use in the hip JCS.

Subsequently, the hip JCS was defined as follows (Figure A.6):

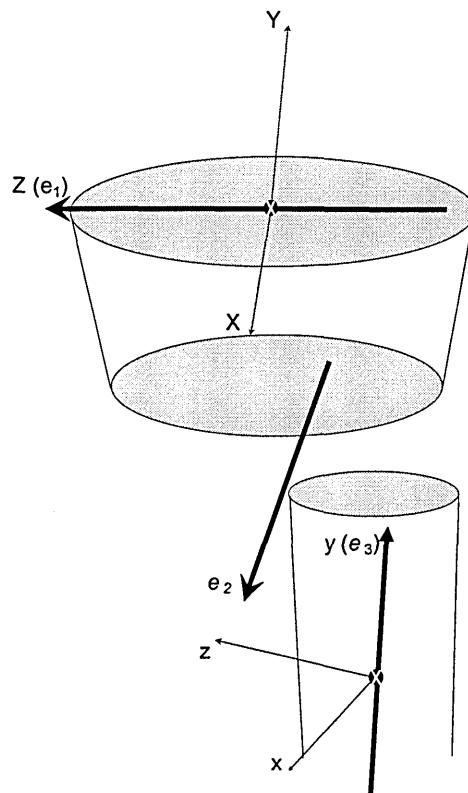


Figure A.6: The hip Joint Coordinate System

Where:

- e₁: Z-axis of the pelvis coordinate system - flexion/extension axis
- e₂: The floating axis, defined as the cross product of the e₁ and e₃ axes - abduction/adduction axis
- e₃: y-axis of the thigh coordinate system - axial rotation axis

Subsequently, the three angles that represent the three-dimensional orientation of the distal segment (j) relative to the proximal segment (i) are calculated as follows. It should be noted that, once the unit vectors that represent the orientation of the

axes of the JCS have been defined at each joint, the mathematics are exactly the same for the angles at the ankle, knee and hip joint (*c.f.* Cole *et al.*, 1993).

For the angle of rotation about the flexion-extension axis:

$$\phi_{1_{ij}} = \cos^{-1}(e_{2_{ij}} \bullet t_i) \cdot sign(e_{2_{ij}} \bullet l_i) \quad [\text{Equation A.1}]$$

For the angle of rotation about the abduction-adduction axis:

$$\phi_{2_{ij}} = \cos^{-1}(r \bullet l_i) \cdot sign(e_{1_{ij}} \bullet e_{3_{ij}}) \quad [\text{Equation A.2}]$$

For the rotation about the axial rotation axis:

$$\phi_{3_{ij}} = \cos^{-1}(e_{2_{ij}} \bullet t_i) \cdot sign(e_{2_{ij}} \bullet t_j) \quad [\text{Equation A.3}]$$

Where:

$$r = \left(\frac{e_{1_{ij}} \times e_{2_{ij}}}{|e_{1_{ij}} \times e_{2_{ij}}|} \right) \quad [\text{Equation A.4}]$$

$$sign(x) = \begin{cases} 1 & \text{if } x \geq 0 \\ 1 & \text{if } x < 0 \end{cases}$$

f = flexion axis of a given segment coordinate system

l = longitudinal axis of a given segment coordinate system

t = third (floating) axis of a given segment coordinate system (*l X f*)

Appendix B: Methods for quantifying coordination variability

Many different techniques exist for quantifying the variability in coordination. Chapter IV of this thesis provided a critique of four methods that have previously been used, namely, continuous relative phase (e.g. Hamill *et al.*, 1999), Vector Coding (Tepavac and Field-Fote, 2001, relative motion angles (e.g. Heiderscheit *et al.*, 2002) and NoRMS (e.g. Sidaway, 1995). The general considerations and aspects of their calculation are outlined in Chapter IV but this appendix provides specific details.

Modified Normalised RMS

Sidaway (1995) presented a technique known as Normalised Root Mean Squared Difference (NoRMS) that quantifies the variability in angle-angle traces. A brief overview of this technique and its benefits and limitations are provided in section 2.6.2.3 of this thesis. As mentioned in section 2.6.2.3, the major limitation of the NoRMS technique presented by Sidaway *et al.* (1995) is that it only provides one measure of coordination variability for the entire cycle. This limits its use in the analysis of movements during which changes in the functional demands of the task over its duration might alter the magnitude of the variability in coordination - e.g. throughout the stance period of running – see section 2.6.2.3. For this reason, a modified NoRMS technique (mNoRMS) was used in this programme of research. The procedure for the calculation of mNoRMS is outlined below.

First, the angular displacement data for the two rotations of interest were interpolated to 101 data points using a cubic spline procedure. Then, the ensemble average of each angular displacement profile was calculated. For each sample data point, the resultant deviation between the cycles of interest was calculated using the following equation:

$$R_{ij} = \sqrt{(\bar{x}_{ij} - x_{ij})^2 + (\bar{y}_{ij} - y_{ij})^2} \quad [\text{Equation B.5}]$$

Where R_{ij} is the resultant deviation for the i th sample and j th cycle, x_{ij} and y_{ij} are the values for the i th sample and j th cycle for the variables represented on the x - and y -axes respectively and \bar{x}_{ij} and \bar{y}_{ij} are the mean values for the i th data point for the variables represented on the x - and y -axes respectively. The root-mean-square deviation values at each data point across cycles were then calculated using the following equation:

$$RMS_i = \left(\frac{\sum_{j=1}^n R_{ij}^2}{n} \right)^{1/2} \quad [\text{Equation B.6}]$$

where RMS_i is the root-mean-square deviation for the i th sample, R_{ij} is the resultant deviation for the i th sample and j th cycle and n is the number of cycles. The procedure produced a value of variability at each of the 101 data points throughout the stride cycle. As James (2004) highlighted, the RMS_i value can also

be calculated by determining the magnitude of the individual joint angle component population standard deviation values as follows:

$$RMS_i = SDp_{Ri} = \sqrt{SDp_{xi}^2 + SDp_{yi}^2} \quad [\text{Equation B.7}]$$

where RMS_i is the root-mean-square deviation for the i th sample, SDp_{Ri} is the resultant deviation for the two angle variables for the i th sample and SDp_{xi} and SDp_{yi} are the population standard deviations for the variables on the x - and y -axes respectively, at the i th sample.

Unlike in the NoRMS calculation outlined by Sidaway *et al.* (1995) which was critiqued in section 2.6.2.3, the RMS values obtained in the mNoRMS technique used in this study were not normalised to the 'maximum resultant excursion of the angle-angle trace'. This normalisation appears to be similar to dividing the standard deviation by the mean value during the calculation of the coefficient of variation. Some authors have questioned the use of the coefficient of variation with some data sets (e.g. Mullineaux, 2000). Mullineaux (2000) stated that normalising data to the mean is appropriate if the means of the two sets of measurements are similar in size but it should not be done if the means are dissimilar as the results can be misleading. Therefore, in some instances where resultant excursions of the mean angle-angle curves are dissimilar, normalising the data to the maximum resultant excursion of the angle-angle trace might produce misleading results. Further, in the context of the studies in this thesis – the future exploration of the variability-

overuse injury hypothesis – it was deemed appropriate to measure absolute rather than relative variability.

Continuous relative phase

The CRP between two oscillating segments at any given point in time is defined as the difference between the respective phase angles of each segment. Phase angles are determined from data plotted on a phase plane (angular velocity on the ordinate axis against angular displacement on the abscissa axis). Therefore, angular velocities were calculated at each joint prior to CRP calculation using the first central difference method (Hamill and Knudson, 2003). Additionally, as discussed in section 2.6.2.1.2, both angular velocity and angular displacement need to be normalised prior to the assembly of the phase plane portrait to adjust for amplitude differences in the ranges of motion and centre the phase plane portraits about the origin. The issue of how to normalise the data on a phase plane is a contentious one – see section 2.6.2.1.2. The normalisation procedures used in this thesis were chosen because they have been used previously (e.g. Hamill *et al.*, 1999, 2000) and have been recommended (Peters *et al.*, 2003). The angular displacement, plotted on the horizontal axis, was normalised using equation B.1:

$$\bar{\theta}_i = \frac{2\theta_i - \max(\theta) - \min(\theta)}{(\max(\theta) - \min(\theta))} \quad [\text{Equation B.1}]$$

Where, $\bar{\theta}$ = normalised angular displacement, θ = angular displacement, i = data point of interest. This ensured that the angular displacement data was normalised

to a unit circle and maximum and minimum values were 1 and -1 respectively. The procedure also ensured that the mid-range of joint motion was represented at the origin of the phase plane portrait. The angular velocity, plotted on the vertical axis of the phase plane portrait, was normalised using equation B.2:

$$\bar{\omega}_i = \frac{\omega_i}{\max\{\max(\omega), \min(\omega)\}} \quad [\text{Equation B.2}]$$

Where, $\bar{\omega}$ = normalised angular displacement, ω = angular displacement, i = data point of interest. This normalisation procedure ensured that zero velocity was maintained at the origin of the phase plane portrait.

Subsequently, the phase angle for each segment was defined as the angle between the right horizontal axis of the phase plane portrait and a line drawn to each specific data point (θ, ω) . The phase angle was calculated using equation B.3:

$$\phi(t) = \tan^{-1}\left(\frac{\omega(t)}{\theta(t)}\right) \quad [\text{Equation B.3}]$$

However, it is important to note that the output of $\tan^{-1}\left(\frac{y}{x}\right)$ takes on values between -90° and $+90^\circ$. Therefore, the output data was manipulated depending on the quadrant in which the data point of interest lay (see Figure B.1). This definition meant that the component phase angles were calculated within a $0^\circ \leq \varphi \leq 360^\circ$ range. This range for the component phase angles has been shown to produce correct results, whereas other definitions have been questioned (Wheat *et al.*, 2002) – see section 2.6.2.1.2.

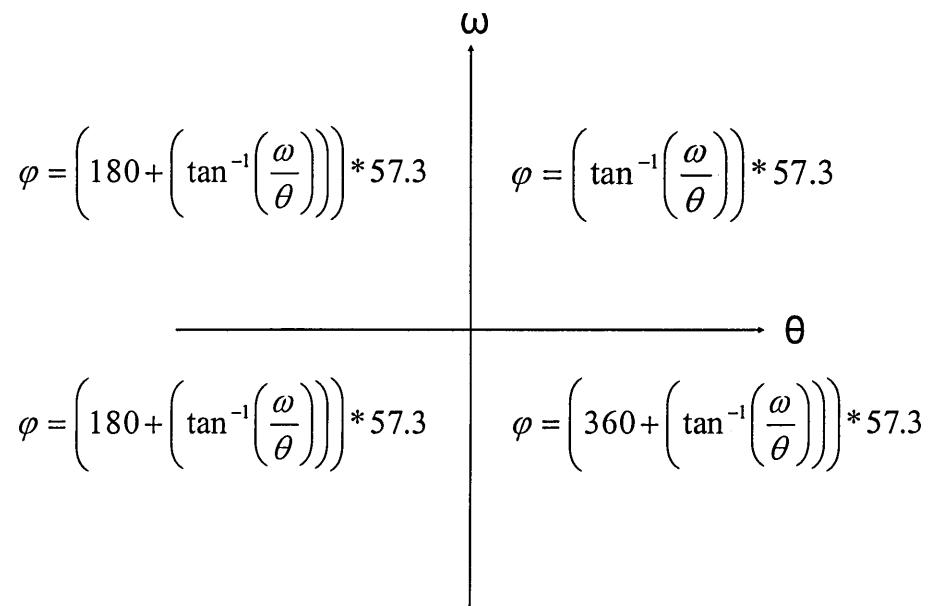


Figure B.1: Phase angle (φ) definition based on a phase plot of normalised angular displacement and normalised angular velocity. The resulting phase angle range is $0^\circ \leq \varphi \leq 360^\circ$.

The CRP (ϕ) between the two segments can then be calculated as the difference between the segment phase angles. In this thesis, this is achieved by subtracting the distal segment phase angle from that of the proximal segment (Equation B.4).

$$\phi(t) = \varphi_{prox}(t) - \varphi_{dist}(t) \quad [\text{Equation B.4}]$$

To enable the measurement of between trial variability of relative phase, each CRP profile was interpolated to 101 data points using a cubic spline procedure. As the CRP values were circular variables, to avoid phase wrapping (Burgess-Limerick *et al.*, 1991; Lamoth *et al.*, 2002), between trial means and standard deviations were calculated at each individual data point using circular statistics – see Appendix C.

Relative motion angles

Relative motion angles have been used by various authors to quantify the coordination between two body segments (e.g. Heiderscheit *et al.*, 2002; Pollard *et al.*, 2003; Ferber *et al.*, 2005). They are related to the vector coding technique introduced by Sparrow *et al.* (1987) and, hence, were calculated in a similar fashion. First, the two angular displacement traces in each of the cycles of interest were normalised to 101 data points using a cubic spline procedure. Second, angle data from two joints were plotted on an angle-angle diagram. Subsequently, the direction of each point-to-point vector on the angle-angle plot was calculated using the following equation:

$$\phi_i = \tan^{-1} \left(\frac{y_{i+1} - y_i}{x_{i+1} - x_i} \right) \quad [\text{Equation B.8}]$$

$$i = 1, 2, 3, \dots, n$$

where ϕ is the relative motion angle, y is the variable on the y -axis of the angle-angle diagram and x is the variable on the x -axis of the angle-angle diagram at the i th data point and n is the number of data points. Similarly to phase angles in the continuous relative phase calculations, the magnitude of the angle needs to be manipulated depending on the quadrant in which the data points lie. Throughout this thesis the relative motion angles were calculated within a $0 \leq \phi \leq 360$ range. Subsequently, the mean and standard deviation of the relative motion angle at each data point over all cycles were calculated. Like continuous relative phase, the relative motion angle is a circular variable so the means and standard deviations at each data point were calculated using circular statistics – see Appendix C for an overview of circular statistics.

Vector Coding

A Vector Coding technique (Tepavac and Field-Fote, 2001) was also used in this thesis. Like continuous relative phase, mNoRMS and relative motion angles, Vector Coding also quantifies the variability in angle-angle traces and it was calculated as described in this sub-section. First, the two angular displacement traces from the joint coupling of interest were normalised to 101 data points using a cubic spline procedure. Second, the difference between the angles represented on the x - and $-$ and y -axes of the angle-angle diagram at frame 1 and frame 2 were determined. The length of the vector between adjacent data points in the angle-angle trace was then calculated using the following equation:

$$l_{i,i+1} = \sqrt{(x_{i,i+1})^2 + (y_{i,i+1})^2} \quad [\text{Equation B.9}]$$

Where $l_{i,i+1}$ is length of the vector, $x_{i,i+1}$ and $y_{i,i+1}$ are the difference between the angles represented on the x- and y-axes respectively, at the i th and subsequent data point. The cosine and sine of the direction of $l_{i,i+1}$ were then calculated using Equation B.10 and Equation B.11 respectively.

$$\cos\theta_{i,i+1} = \frac{x_{i,i+1}}{l_{i,i+1}} \quad [\text{Equation B.10}]$$

$$\sin\theta_{i,i+1} = \frac{y_{i,i+1}}{l_{i,i+1}} \quad [\text{Equation B.11}]$$

The mean cosine ($\overline{\cos\theta_{i,i+1}}$) and mean sine ($\overline{\sin\theta_{i,i+1}}$) for each frame-to-frame interval over the cycles of interest were then calculated. Subsequently, the mean vector angle for each frame-to-frame interval was calculated using Equation B.12:

$$a_{i,i+1} = \sqrt{(\overline{\cos\theta_{i,i+1}})^2 + (\overline{\sin\theta_{i,i+1}})^2} \quad [\text{Equation B.12}]$$

Where $a_{i,i+1}$ denotes the dispersion of the angle-angle values about the mean, over multiple cycles (Batschelet, 1981) between the i th and subsequent data point. The arithmetic average of a over all frames in the cycle was then determined using Equation B.13:

$$\bar{a} = \frac{1}{N}(a_{1,2} + a_{2,3} + \dots + a_{N-1,N}) \quad [\text{Equation B.13}]$$

Where n is the total number of frames per cycle and \bar{a} signifies the overall variability (between 0 and 1) in the shape of the angle-angle trace for all cycles recorded. Tepavac and Field-Fote's (2001) vector coding technique also provides a measure of the variability in the size of the angle-angle traces i.e. frame-to-frame vector magnitudes. Firstly, in order to keep the variance below 1, the vector lengths during each cycle were normalised by dividing the all magnitudes within a given frame-to-frame interval by the maximum value for that interval:

$$l'_{i,i+1}[j] = \frac{l_{i,i+1}[j]}{\max(l_{i,2})} \quad j = 1, 2, \dots, M \quad [\text{Equation B.14}]$$

Where $l_{i,j+1}$ is the length of the frame-to-frame vector, $l'_{i,j+1}$ is the normalised length of the frame-to-frame vector and M is the number of cycles. Subsequently, the mean and standard deviation for the magnitudes within each frame-frame interval were calculated. As Tepevac and Field-Fote (2001) stated, the maximum value for

the standard deviation of an array of normalised samples is $\frac{1}{2}\sqrt{\frac{M+1}{M}}$ for an odd

number of samples and $\frac{1}{2}\sqrt{\frac{M}{M-1}}$ for an even number samples. In order to place

the measure of the variability in the magnitude of the frame-to-frame interval vector, m , within the range of 0 to 1, the standard deviation of $l_{i,i+1}$ was divided by the maximum standard deviation. Therefore, the magnitude deviation for each frame-to-frame interval was calculated as follows:

$$m_{i,i+1} = 1 - \frac{\sigma(l_{i,i+1})}{\sigma_{\max}} \quad [\text{Equation B.15}]$$

Where, $m_{i,i+1}$ is the magnitude deviation, $l_{i,i+1}$ is the length of the frame-to-frame vector between the i th and subsequent data point. The mean vector magnitude deviation for all frames in the cycle was then determined by:

$$\bar{m} = \frac{1}{N}(m_{1,2} + m_{2,3} + \dots + m_{N-1,N}) \quad [\text{Equation B.16}]$$

The larger the value of \bar{m} , the more the angle-angle traces have similar distances between consecutive frames. Finally, Tepavac and Field-Fote defined the coefficient of correspondence as the product of $a_{i,i+1}$ and $m_{i,i+1}$:

$$\bar{r} = \frac{1}{N}(a_{1,2} \cdot m_{1,2} + a_{2,3} \cdot m_{2,3} + \dots + a_{N-1,N} \cdot m_{N-1,N}) \quad [\text{Equation B.17}]$$

Where $r_{i,i+1}$ is the coefficient of correspondence (in the range 0 to 1)¹¹, $a_{i,i+1}$ is the deviation in shape and $m_{i,i+1}$ is the deviation in length of the vector defined by the i th and subsequent data point. It should be noted that average values of a , m or r can be easily calculate over any defined period of the cycle, in addition to the averages over the entire cycle given in Equation B.13, B.16 and B17 respectively.

¹¹ A value of 1 indicates that all cycles are identical, while a value close to 0 indicates that the data are near random.

Appendix C: Circular statistics

Since both continuous relative phase and relative motion angles are circular measures, means and standard deviations of these variables were calculated using circular statistics as recommended by various authors (e.g. Batschelet, 1981; Burgess-Limerick *et al.* 1991; Lamothe *et al.*, 2002). If conventional linear statistical techniques were applied to directional data erroneous means and standard deviations could be calculated. Using four angles that describe a direction as an example: 30° , 45° , 350° and 300° , it seems obvious that an appropriate mean angle is somewhere close to 0° (Figure C.1). However, if the mean angle is calculated using conventional linear statistical techniques, the answer (181.25°) is clearly incorrect.

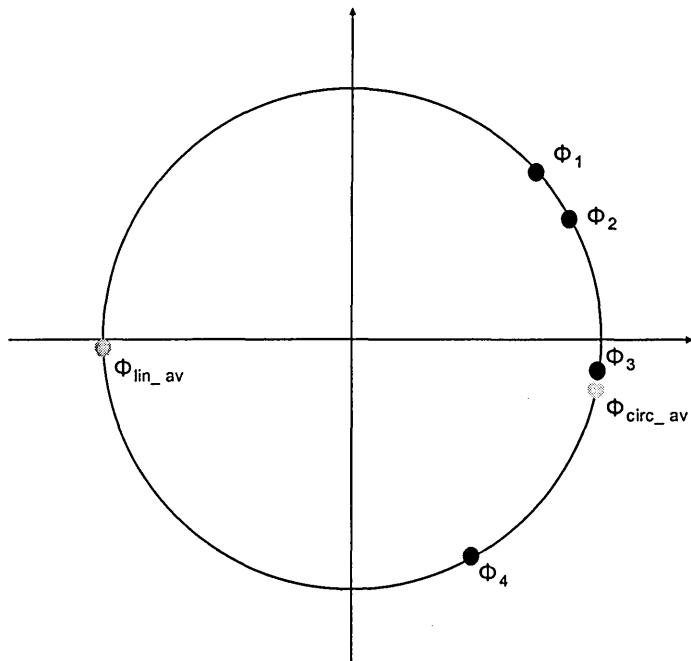


Figure C.1: The calculation of the mean of four angles (Φ_1 , Φ_2 , Φ_3 , Φ_4) using linear (Φ_{lin_av}) and circular (Φ_{circ_av}) statistical techniques.

The appropriate technique for calculating the mean angle is based on a quite different procedure. First, all angles are converted into unit vectors. Second, each angle is plotted on a *unit circle*. The mean direction is then simply calculated as the direction of the resultant of all of the vectors on the unit circle. Therefore, the mean direction of a given set of angles can be calculated as follows (example adapted from Lamothe *et al.*, 2002):

For a sample of directions: e_1, \dots, e_n . θ_i is one of the n observed angles, e_i is the corresponding unit vector and C_i and S_i are the rectangular components of θ_i calculated using Equation C.1 and Equation C.2 respectively.

$$C = \frac{1}{n} \sum_{i=1}^n \cos \theta_i \quad [\text{Equation C.1}]$$

$$S = \frac{1}{n} \sum_{i=1}^n \sin \theta_i \quad [\text{Equation C.2}]$$

The mean angle $\bar{\theta}$ of the sample is obtained using Equation C.3:

$$\bar{\theta} = \tan^{-1}\left(\frac{S}{C}\right) \quad [\text{Equation C.3}]$$

if $C > 0$ and $S > 0$; $\bar{\theta} = \bar{\theta}$

if $C < 0$; $\bar{\theta} = \bar{\theta} + \pi$

if $C > 0$ and $S < 0$; $\bar{\theta} = \bar{\theta} + 2\pi$

The circular variance is then calculated using Equation C.4:

$$R = \sqrt{C^2 + S^2} \quad [\text{Equation C.4}]$$

Where R is a measure of the circular variance of the sample and is the length of the resultant vector with components C and S . Finally, the standard deviation can be calculated using Equation C.5:

$$S.D. = \sqrt{2(1 - R)} \quad [\text{Equation C.5}]$$

Batschelet (1981) reported that the 2 in Equation C.5 makes the circular standard deviation conform to the linear standard deviation.

Appendix D: Further Statistical information

In Chapters IV, V and VI repeated measures research designs were employed. This meant that the statistical procedures were common to all three Chapters; analyses of variance (ANOVA) with repeated measures for each dependent variable. This appendix outlines the assumptions of the repeated measures ANOVA and details of the checks made to ensure that the data met these assumptions are also documented. Finally, a justification of the alpha level set throughout the thesis is provided.

Assumptions of the repeated measures analysis of variance

Vincent (1999) stated that the repeated measures ANOVA assumes that the data are parametric; all of the dependent variables measured in this thesis comply with this assumption as they are based on a ratio scale and samples were randomly selected. In addition to the data being parametric, the repeated measures ANOVA assumes normal distribution and sphericity.

Normal Distribution

There are two ways in which a given set of data could deviate from the normal distribution. The data are not normally distributed if they exhibit skewness or kurtosis. Kurtosis is a measure of the relative 'peakedness' of the curve of the data and skewness gives a measure of the direction of the 'hump' of the curve of distribution of data and the nature of the tails of the curve (Vincent, 1999).

Most researchers eyeball their sample data (using a frequency distribution) to assess normal distribution (Field, 2005). However, objective measurements are available to test normal distribution. In this thesis normal distribution was tested by computing the z scores for skewness and kurtosis. These z scores were calculated as follows:

$$z_{kurtosis} = \frac{kurt}{SE_{kurt}} \quad [\text{Equation D.1}]$$

Where $z_{kurtosis}$ is the kurtosis z score, $kurt$ is the kurtosis statistic and SE_{kurt} is the standard error for kurtosis.

$$z_{skewness} = \frac{skew}{SE_{skew}} \quad [\text{Equation D.2}]$$

Where $z_{skewness}$ is the skewness z score, $skew$ is the skewness statistic and SE_{skew} is the standard error for skewness. The sample data in this thesis were assumed to be normally distributed if $z_{skewness}$ and $z_{kurtosis}$ scores were within ± 2 as recommended by Vincent (1999).

Sphericity

A further assumption of repeated measures ANOVA is that the data exhibit sphericity, which is sometimes referred to as compound symmetry (Vincent, 1999). Sphericity requires that the repeated measurements demonstrate homogeneity of variance and homogeneity of covariance. Homogeneity of variance means that

variances (s^2) are equivalent between the repeated measures (Thomas and Nelson, 2004). Furthermore, homogeneity of covariance means that the relationship, or correlations, on the dependent variable among all three or more repeated measures are equal (Vincent, 1999). Sphericity was tested using Mauchly's Test. If Mauchly's test was significant ($P < 0.05$), the data did not exhibit sphericity and the affected F -ratio was adapted using the Greenhouse-Geisser correction.

Alpha Level of Significance

Frequently in human movement research the alpha level for inferential statistical tests is set at 0.05. This was adopted as an appropriate alpha level for the studies in this thesis. Also, in line with the recommendations of Cohen (1992), the beta level was set at 0.2. Therefore, the type I and type II error rates in all studies in this thesis were 0.05 and 0.2 respectively. However, in Chapters V and VI multiple tests were performed on the data. Some authors have suggested that, in situations of multiple significance testing, the type I error rate is inflated (e.g. Tukey, 1977; Bland and Altman, 1995). To account for this, it has been proposed that the alpha level should be adapted using some sort of correction procedure (e.g. Bland and Altman, 1995; Thomas and Nelson, 2004). A popular example of such a procedure is the Bonferroni correction (c.f. Bland and Altman, 1995), with which the alpha level is adapted by dividing the original alpha by the number of statistical comparisons performed in the study. This has the effect of maintaining the study-wide error rate at the original value of alpha.

Notwithstanding the possible benefits of a more stringent criterion – decreased risk of a type I statistical error - many authors have challenged the notion that a correction need be applied (e.g. Rothman, 1990; Savitz and Olshan, 1995; Perneger, 1998). Indeed, Perneger (1998, p. 1236) suggested that ‘Bonferroni adjustments are, at best, unnecessary and, at worst, deleterious to sound statistical inference’. In explanation of this statement, Perneger (1998) suggested that the main weakness of the Bonferroni adjustment is that the interpretation of a finding depends on the number of tests performed. Further, the type I error cannot decrease without inflating the type II error. As type II errors are no less important than type I errors (Perneger, 1998), this represents a large problem for the Bonferroni adjustment. For the reasons outlined by Perneger (1998), and as separate, *a priori* determined, hypotheses were tested in all Chapters, the Bonferroni correction was not used in this thesis. Therefore, for all statistical comparisons, alpha was set at 0.05.

Appendix E

Chapter IV

Participant information sheet and informed consent form



School of Sport and Leisure Management

Research Ethics Committee

Participant Information Sheet

Project Title	Biomechanical analysis of mBT shoes
Name of Participant	
Supervisor/Director of Studies	
Principal Investigator	
Purpose of Study and Brief Description of Procedures <i>(Not a legal explanation but a simple statement)</i>	

The aim of the proposed investigation is a comprehensive assessment of mBT shoes compared to traditional shoes. Three dimensional motion analysis, force plate analysis and electromyography will be used to assess your walking technique used during the two conditions. This will allow us to compare walking kinematics and kinetics when using mBT and traditional shoes.

You will firstly be required to complete a training session that will teach you to use the mBT shoes properly. The testing will take place in the Biomechanics Laboratory, A010 Collegiate Hall and should take no longer than one hour. At a separate testing session you will be required to perform eight good trials of walking at a set speed using mBT shoes and your normal shoes during which time 3-dimensional kinematic and kinetic data will be captured. EMG data will also be captured using a telemetred EMG system which will acquire muscle activation data from some of the superficial core musculature and gluteus maximus muscle. This will require the shaving of a small area of skin around where the EMG electrode will be placed. In order for the 3D analysis to take place you will be required to wear a number of reflective markers which will be positioned on specific parts of the body. Tight, close fitting clothes will be required to be worn.

As use of mBT sometimes can involve the realignment of the musculature, when mBT are first used users may experience some discomfort. You are free to withdraw at any time throughout the testing and if you have any questions about the procedures or any other aspects of the project please do not hesitate to ask. On completion you are allowed to keep the mBT shoes that you wore for the testing procedure.

It has been made clear to me that, should I feel that these Regulations are being infringed or that my interests are otherwise being ignored, neglected or denied, I should inform Professor Edward Winter, Chair of the School of Sport and Leisure Management Research Ethics Committee (Tel: 0114 225 4333) who will undertake to investigate my complaint.



School of Sport and Leisure Management

Research Ethics Committee

INFORMED CONSENT FORM

TITLE OF PROJECT: Biomechanical analysis of mBT shoes

The participant should complete the whole of this sheet himself/herself

Have you read the Participant Information Sheet?	YES/NO
Have you had an opportunity to ask questions and discuss this study?	YES/NO
Have you received satisfactory answers to all of your questions?	YES/NO
Have you received enough information about the study?	YES/NO
To whom have you spoken?	
Do you understand that you are free to withdraw from the study: <ul style="list-style-type: none">• at any time• without having to give a reason for withdrawing• and without affecting your future medical care	YES/NO
Have you had sufficient time to consider the nature of this project?	YES/NO
Do you agree to take part in this study?	YES/NO
Signed	Date
(NAME IN BLOCK LETTERS).....	
Signature of Parent / Guardian in the case of a minor	

FOR USE WHEN STILL OR MOVING IMAGES WILL BE RECORDED

Consent to scientific illustration

I hereby confirm that I give consent for photographic and/or videotape and sound recordings (the 'material') to be made of me. I confirm that the purpose for which the material would be used has been explained to me in terms which I have understood and I agree to the use of the material in such circumstances. I understand that if the material is required for use in any other way than that explained to me then my consent to this will be specifically sought.

1. I understand that the material will form part of my confidential records and has value in scientific assessment and I agree to this use of the material.

Signed..... Date.....

Signature of Parent / Guardian in the case of a minor

2. I understand the material has value in teaching and I consent to the material being shown to appropriate professional staff for the purpose of education, staff training and professional development.

Signed..... Date.....

Signature of Parent / Guardian in the case of a minor

I hereby give consent for the photographic recording made of me on..... to be published in an appropriate journal or textbook. It is understood that I have the right to withdraw consent at any time prior to publication but that once the images are in the public domain there may be no opportunity for the effective withdrawal of consent.

Signed Date

Signature of Parent / Guardian in the case of a minor

Appendix G

Chapter V

Participant information sheet and informed consent form



School of Sport and Leisure Management

Research Ethics Committee

Participant Information Sheet

Project Title	Joint coordination variability during overground and treadmill running
Name of Participant	
Supervisor/Director of Studies	Prof Roger Bartlett
Principal Investigator	Mr Jon Wheat
Purpose of Study and Brief Description of Procedures <i>(Not a legal explanation but a simple statement)</i>	
<p>The purpose of this study is to examine the small differences that exist from stride to stride when running at different speeds and thus quantify the variability of locomotion. This locomotion variability will be examined in two different running and walking situations: 1) running on the ground, 2) running on a treadmill. Your main aim is to try to maintain your speed constant irrespective of the mode of locomotion. In order to measure slight differences from stride to stride in your movement, reflective markers will be fixed on a moulded plate and this will be fixed on your left leg, one plate with four markers on your calf and another one on your thigh. These will be secured with bandages and special tape. Four additional reflective markers will be stuck with double-sided tape on your pelvis and a further four will be placed on your shoes. The position of these markers will be tracked during running and walking by nine special cameras so that the accurate position and orientation of your left lower limb in the three-dimensional space can be determined. You will be required to wear a special belt that is connected to the treadmill control box which is attached to a main switch, so that if you encounter any difficulties the treadmill will stop. During running overground you are required to hit a force platform that measures the forces exerted between your foot (shoe) and the ground so there will be several attempts until a clean contact can be obtained with a consistent speed that is within the target speed you need to maintain. You will be given a familiarisation period even if you have used a treadmill and a running track with a force plate embedded before.</p>	
<p>If you have any questions we will be happy to answer them before you agree to participate. You are under no obligation to participate in the study and you are free to withdraw at any time, without giving any reasons or explanation. All data collected will be treated confidentially and your name will not be identified in any reports or publications resulting from this study.</p>	
<p>It has been made clear to me that, should I feel that these Regulations are being infringed or that my interests are otherwise being ignored, neglected or denied, I should inform Professor Edward Winter, Chair of the School of Sport and Leisure Management Research Ethics Committee (Tel: 0114 225 4333) who will undertake to investigate my complaint.</p>	



School of Sport and Leisure Management

Research Ethics Committee

INFORMED CONSENT FORM

TITLE OF PROJECT: Joint coordination variability during overground and treadmill running

The participant should complete the whole of this sheet himself/herself

Have you read the Participant Information Sheet?	YES/NO
Have you had an opportunity to ask questions and discuss this study?	YES/NO
Have you received satisfactory answers to all of your questions?	YES/NO
Have you received enough information about the study?	YES/NO
To whom have you spoken?	
Do you understand that you are free to withdraw from the study: <ul style="list-style-type: none">• at any time• without having to give a reason for withdrawing• and without affecting your future medical care	YES/NO
Have you had sufficient time to consider the nature of this project?	YES/NO
Do you agree to take part in this study?	YES/NO

Signed Date

(NAME IN BLOCK LETTERS).....

Signature of Parent / Guardian in the case of a minor
.....

FOR USE WHEN STILL OR MOVING IMAGES WILL BE RECORDED

Consent to scientific illustration

I hereby confirm that I give consent for photographic and/or videotape and sound recordings (the 'material') to be made of me. I confirm that the purpose for which the material would be used has been explained to me in terms which I have understood and I agree to the use of the material in such circumstances. I understand that if the material is required for use in any other way than that explained to me then my consent to this will be specifically sought.

1. I understand that the material will form part of my confidential records and has value in scientific assessment and I agree to this use of the material.

Signed..... Date.....

Signature of Parent / Guardian in the case of a minor

2. I understand the material has value in teaching and I consent to the material being shown to appropriate professional staff for the purpose of education, staff training and professional development.

Signed..... Date.....

Signature of Parent / Guardian in the case of a minor

I hereby give consent for the photographic recording made of me on..... to be published in an appropriate journal or textbook. It is understood that I have the right to withdraw consent at any time prior to publication but that once the images are in the public domain there may be no opportunity for the effective withdrawal of consent.

Signed Date

Signature of Parent / Guardian in the case of a minor

Informed Consent Form (to be retained by the investigator)

Participant:

Name:

Sex: Male / Female

Date of Birth:

Supervisor/Principal Investigator: V. Baltzopoulos

Investigator/Collaborators: Jon Wheat

Ethics Committee Approval Number:

Project Title: Kinematic variability during overground, treadmill and treadmill on-demand locomotion

The purpose of this study is to examine the small differences that exist from stride to stride when running and thus quantify the variability of locomotion. This is very important information because any changes in the running technique must be above this normal variability threshold before any conclusions can be safely reached about the effects of training, pathology etc. This locomotion variability will be examined on three different running and walking situations: 1) normal running on the ground, 2) running on a treadmill where the speed of the belt is controlled and remains constant while you are on it and 3) running on a treadmill where you can move freely and the speed is adjusted automatically all the time so that you have control over your speed, in a similar fashion to overground locomotion. A speed of approximately $3.5 \text{ m}\cdot\text{s}^{-1}$ will be used. Your main aim is to try to maintain your speed constant irrespective of the mode of locomotion. In order to measure slight differences from stride to stride in your movement, reflective markers will be fixed on a moulded plate and this will be fixed on your left leg, one plate with four markers on your calf and another one on your thigh. These will be secured with bandages and special tape. Four additional reflective markers will be stuck with double-sided tape on your pelvis. The position of these markers will be tracked during running by nine special cameras so that the accurate position and orientation of your left lower limb in the three-dimensional space can be determined. The belt level of the treadmill is approximately 2 feet higher than the ground and for this reason you will be required to wear a special harness that is connected to a supporting crane and is attached to a main switch, so that if you accidentally trip or lose your balance you will be supported and the treadmill will stop. During running overground you are required to hit a force plate that measures the forces exerted between your foot (shoe) and the ground so there will be several attempts until a clean contact can be obtained with a consistent speed that is within the target speed you need to maintain. You will be given a familiarisation session even if you have used a treadmill and a running track with a force plate embedded before.

If you have any questions we will be happy to answer them before you agree to participate. You are under no obligation to participate in the study and you are free to withdraw at any time, without giving any reasons or explanation. All data collected will be treated confidentially and your name will not be identified in any reports or publications resulting from this study.

Participant Statement

I fully understand what is involved in taking part in this study. Any questions I have about the study, or my participation in it, have been answered to my satisfaction. I understand that I do not have to take part and that I may decide to withdraw from the study at any point without prejudice. I have had my attention drawn to the document 'Ethical Regulations for the Use of Humans in Research'. My concerns regarding this study have been answered and such further concerns as I have during the time of the study will be responded to. It has been made clear to me that, should I feel that these Regulations are

being infringed or that my interests are otherwise being ignored, neglected or denied, I should inform the Chair of the Ethics Committee of the Department of Exercise and Sport Science, Manchester Metropolitan University, Hassall Road, Alsager, Cheshire, ST7 2HL who will undertake to investigate my complaint.

Signed Date