THE EFFECT OF BREAST SUPPORT ON RUNNING BIOMECHANICS

Alexandra Kiri Milligan

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ABSTRACT

Whilst sports bras have been reported to significantly reduce breast kinematics and exercise-related breast pain, little is known about the effect of breast support on running biomechanics. This research area has novel applications and many potential benefits to female athletes. Papers available within this area hypothesise that the reduction of breast kinematics and exercise-related breast pain, provided by a high breast support, ensures running biomechanics are maintained and potentially enhanced, however, few have provided evidence of this. To investigate this area this thesis explored biomechanical measures during running including; breast biomechanics, full body running kinematics, and an examination of upper body muscle activity during a five kilometre treadmill run, in low and high breast support conditions.

An integrated programme of work was conducted with multiple variables collected and presented in chapter four to seven. Chapter three identified significant changes in breast kinematics during a prolonged treadmill run, and defined the run duration for this programme of work. Chapter four examined breast biomechanics during a five kilometre treadmill run, in different breast support conditions. In line with previous publications, the high breast support provided superior magnitudes of support to the breasts (up to 75% reduction) compared to the lower breast support conditions, and significant reductions in exercise-related breast pain throughout treadmill running. Increases in multiplanar breast displacement, velocity, acceleration, and approximated force were reported from the start to the end of the five kilometre run in both low (increases of 7 mm, 0.10 $m.s^{-1}$, 5.6 $m.s^{-2}$, 3 N) and high (5 mm, 0.07 $m.s^{-1}$, 2.7 $m.s^{-2}$, 1 N) breast supports. These novel findings demonstrate that breast kinematics increase during a five kilometre treadmill run, which may directly affect an individual's running biomechanics.

Assessing the magnitude of variance associated with breast biomechanics data ensures accurate interpretation of the reported findings. To achieve this, within- and between-participant variance in multiplanar breast kinematics were quantified utilising the coefficient of variance (C_v%). The smallest differences in breast kinematics reported in the third chapter exceeded the reported within-participant variance in both low (12 C_v%) and high (15 C_v%) breast supports, and were therefore defined as meaningful differences. Between-participant variance in multiplanar breast kinematics in low (23 C_v%) and high (29 C_v%) breast supports was greater than the within-participant variance, and should be considered in future for research designs and sample sizes.

To assess running kinematics between breast supports, a full body kinematic analysis was conducted including the quantification of step length and full body Cardan joint angles. When running in the lower breast support conditions, costly running mechanics such as greater thorax flexion, shorter step length, less acute knee angle, greater arm swing mechanics, and greater axial rotation of the thorax and pelvis were reported. However, the high breast support exhibited a kinematic profile more closely aligned with a desirable, economic running style previously defined within the literature. These findings support claims that the breast support worn may impact upon biomechanical parameters, with high breast support eliciting advantageous running kinematics. This unique work found female runners will alter their running kinematics depending upon the breast support worn.

Changes in running kinematics away from an individual’s natural kinematics have been linked to changes in the activation of muscles driving these movements. Therefore, given the reported differences in upper body running kinematics, the effect of breast support on the activity of six upper body muscles central to running was examined and reported. Reductions in normalised peak activity of the pectoralis major (37% reduction), anterior deltoid (26% reduction) and medial deltoid (30% reduction) were reported in the high breast support; suggesting that a high breast support significantly reduces the peak activation of these three muscles compared to lower breast support conditions during running. Furthermore, the differences in activity of these muscles are thought to be associated with the changes in upper body kinematics, specifically arm swing mechanics.

The research design of this programme of work enabled relationships between the key biomechanical measures to be explored, providing a holistic view of the effect of breast support on the biomechanics of the female runner. Relationships were identified between the magnitude of breast kinematics, which is governed by the breast support worn, and the following biomechanical measures investigated; exercise-related breast pain, upper and lower body running kinematics and upper body muscle activity. Furthermore, certain running kinematics demonstrated significant relationships to muscle activity.

This research has shown that breast biomechanics, running kinematics and upper body activity are affected by the breast support worn during treadmill running. The use of high breast support has demonstrated the potential of this breast support to benefit running biomechanics. This novel programme of work has progressed the knowledge of the effect of breast support on both breast and body biomechanics during treadmill running.

CONTENTS

DECLARATION

Whilst registered as a candidate for the above degree, I have not been registered for any other research award. The results and conclusions embodied in this thesis are the work of the named candidate and have not been submitted for any other academic award.

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ABBREVIATIONS

TERMINOLOGY	DEFINITION
COM	Centre of Mass
3D	Three-Dimensions
SD	Standard Deviation
GRF	Ground Reaction Forces
C_v%	Percentage Coefficient of Variance
LCS	Local Coordinate System
GCS	Global Coordinate System
SCS	Segment Coordinate System
ISB	International Society of Biomechanics
POSE	Position and Orientation of a segment
QTM	Qualisys Track Manager
FFT	Fast Fourier Transform
ROM	Range of Motion
CODA	Codamotion, Charnwood Dynamics, Ltd.
EMG	Electromyography
MVC	Maximal Voluntary Contraction
RMS	Root-Mean-Square
*i*EMG	Integrated Electromyography
RPE	Rating of Perceived Exertion

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DISSEMINATION

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3. **Milligan, A.** (2012). The effect of breast support on multiplanar breast kinematics during a five kilometre treadmill run. University of Chichester and University of Portsmouth Postgraduate Research Conference, 18th June 2012.
4. **Milligan, A**., Scurr, J., Mills, C., & Wood, L. (2011). An investigation into the within and between participant variance in three-dimensional breast kinematic data during a 5 km run. In J. Paulo Vilas-Boas, L. Machado, K. Wangdo, & A. Velosa (Eds.), *Proceedings of the 29th Conference of the International Society of Biomechanics in Sports, ISBS, July 2011, Porto, Portugal.*
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CHAPTER ONE. INTRODUCTION

1.1 Justification

In recent years the potential benefits of wearing a sports bra during exercise have been investigated, with emphasis placed upon the sports bra's ability to reduce multiplanar breast kinematics relative to the thorax (Mason, Page, & Fallon, 1999; Scurr, White, & Hedger, 2009; Scurr, White, & Hedger, 2010a; White, Scurr, & Smith, 2009), reduce exercise-related breast pain (Mason et al., 1999; Scurr et al., 2010a), and potentially reduce the risk of strain to the anatomical restraints of the breast (i.e. the Cooper's ligaments and overlying skin) (Bowles & Steele, 2003; Scurr, Bridgeman, White, & Hedger, 2009b). These empirical studies have demonstrated that a sports bra can significantly reduce negative factors associated with independent movement of the breast tissue, however, a high percentage of exercising females do not wear a sports bra during physical activities such as running (Bowles, Steele, & Munroe, 2008). Many of the empirical publications highlighted and hypothesised the potential benefits of wearing a sports bra for sporting performance (Starr et al., 2005; Mason et al., 1999; White et al., 2009), suggesting that the reduction of negative factors such as the magnitude of relative breast movement and associated breast pain, would ensure performance is maintained and potentially enhanced. Currently, no publications exist which have quantified performance directly (e.g. finishing time, running pace). However, there are a few publications which have quantified and monitored biomechanical and physiological variables across multiple breast support conditions, which have previously been shown to influence performance.

Biomechanical analyses have previously been employed to address this question, which enabled the quantification of human movement via kinematic and kinetic analyses. Both Shivitz (2001) and White et al., (2009) postulated that increases in peak ground reaction forces (GRF) in a low breast support condition were as a result of changes in running kinematics. White et al., (2009) suggested these differences were made in an attempt to reduce the magnitude of breast kinematics and to increase comfort, and these changes may also influence physiological parameters. While this conclusion may seem logical, running kinematic parameters were not measured, and therefore it is unknown which gait parameter/s may be changing. Boschma, Smith, and Lawson, (1995) explored the effect of differing breast support conditions on kinematic analysis of treadmill running, and concluded that when examined on a case-by-case basis individuals altered certain running kinematics dependent upon the breast support worn.

Changes in running kinematics away from an individual's natural kinematics have been linked to changes in the activation patterns of the muscles driving these movements (Basmajian & De Luca, 1985; Higham, Biewener, & Delp, 2011; Komi, 2003), metabolic cost of running, and running economy (Cavanagh & Williams, 1982; Williams & Cavanagh, 1987). If a high breast support can reduce costly mechanical alterations during running such as; increases in centre of mass (CoM) displacements (Inman, 1966), changes in natural step lengths and step frequencies (Donelan, Kram, & Kuo, 2002; Martin & Morgan, 1992; Williams & Cavanagh, 1987), and greater flexion of the trunk (Saunders, Pyne, Telford, & Hawley, 2004; Williams & Cavanagh, 1987), running performance may be enhanced when exercising in this breast support condition.

Previous work by Scurr, Bridgman, and Hedger (2010b) reported significantly less muscle activity in the pectoralis major when wearing external breast support during treadmill running. These findings may be associated with potential differences in arm swing mechanics previously highlighted by Boschma et al., (2005), and may have potential benefits for female runners. Furthermore, Scurr et al., (2010b) hypothesised that this unique finding may indicate a contribution of anatomical support to the breast from this muscle, which has previously not been explored in-depth. Further exploration of the affect of breast support on upper body muscle activity would extend the knowledge of the influence of breast support on biomechanical measures of running and develop our understanding of the relationship between the pectoralis muscle and breast biomechanics.

Although these initial studies provide an insight into how a female may alter her biomechanical running performance in different breast supports, it is important to consider the external validity of these studies, and essential to assess the application of these findings. Firstly, these data are a collective from multiple papers and abstracts available within this area, therefore it is unknown if more than one biomechanical measure is affected by different breast support conditions. Examining multiple biomechanical parameters with one cohort will provide a more holistic view of the female runner in different breast supports. Secondly, running kinematics have been shown to change over time (Williams & Cavanagh, 1987; Williams, Snow, & Argruss, 1991; Hardin, Van Den Bogert, & Hamill, 2004; Hunter & Smith, 2007), and therefore the length of previously examined trials (< 7 minutes) may not be representative of common running distances, which restricts the application of these data.

The potential benefits of furthering the knowledge within this area are wide reaching, with applications to the maintenance and enhancement of training and performance for female athletes, product design, and influential methodological progressions. Examining multiplanar breast kinematics, full body running kinematics, and muscle activity simultaneously over a five kilometre treadmill run is novel research and would enhance the external validity of the effect of breast support on biomechanical measures during treadmill running.

1.2 Thesis Aims

The overall aim of this thesis is to investigate the effect of breast support on multiplanar breast biomechanics and biomechanical running parameters during a five kilometre treadmill run. The discrete objectives of the research study were;

- To investigate the effect of breast support on multiplanar breast biomechanics during a five kilometre run, and to assess the magnitude of within- and between-participant variance within this data.
- To investigate the effect of breast support on full body running kinematics during a five kilometre run.
- To investigate the effect of breast support on upper body muscle activity during a five kilometre run.

1.3 Outline of thesis

This thesis commences with an introduction to the research area comprising a review of relevant breast biomechanics literature and related biomechanical research, presented in chapter two. This review established gaps within the literature and provides rationale for the current research questions.

Magnitude of breast support has previously been reported to effect the magnitude of breast kinematics and breast pain, with recommendations on breast support and product design made based on these publications, however, these findings have currently only been reported over short duration runs. In order to clarify the relevance of these findings for prolonged running, chapter three investigated the effect of three breast support conditions (bare-breasted, low and high) on breast kinematics during a five kilometre run. The run distance was selected based upon the government guidelines for exercise prescription and in order to assess any potential changes in breast kinematics over a prolonged treadmill run. This pilot study defined the run distance set for this programme of work, confirmed differences in breast kinematics over time, and compared the findings with breast kinematics collected over shorter running bouts.

The work reported in chapter four examined breast biomechanics and exercise-related breast pain over a five kilometre treadmill run. As these data were some of the first to be reported over a prolonged treadmill run and to confirm the accuracy of these data, section two of this chapter assessed the variance in multiplanar breast kinematics in the three breast support conditions over the five kilometre run. These data helped ascertain the different components of total error in breast kinematics and define the significance of the differences reported in the first section of this chapter.

Previous literature has postulated that breast support may influence running biomechanics and performance. The work presented in chapter five examined the effect of breast support on full body running kinematics during a five kilometre run. Having established differences in running kinematics between breast support conditions, and considering the link between muscle activity and segmental movement, chapter six explored the effect of breast support on myoelectric activity in upper body muscles during prolonged treadmill running. The findings of these two chapters help determine the effect of breast support on biomechanical running parameters.

Since the data collected within this programme of work utilised the same participants over two testing sessions, chapter seven explored the relationships between breast and body biomechanical variables, to gain a holistic view of the female runner in different breast support conditions.

Chapter eight of this thesis provides a general discussion of the programme of research, considering the unique findings of the work conducted, progressions in methodologies, development of knowledge in this research domain, assumption, limitations and delimitations of this work, recommendations for future work, and final conclusions of the thesis.

CHAPTER TWO. LITERATURE REVIEW

The following methods were used to identify key publications within the area of interest. Firstly, literature was searched and reviewed using evidence based journals and academic databases, such as Pub Med and Elsevier/Science Direct. Secondly, all references retrieved were scanned for relevant citations to expand the literature search.

This literature review explores the area of breast biomechanics and influential factors associated with relative breast kinematics that may impact upon running biomechanics. It begins by describing the anatomy of the female breast, detailing the unique anatomical make up. Following this, factors related to the breast that may affect the exercising female including exercise-related breast pain, level of external breast support worn and the resulting magnitude of breast kinematics are reviewed. In the final section, the literature which has investigated factors affecting biomechanical parameters during running, and unique research investigating the effect of breast support on running biomechanics were reviewed.

2.1 The female breast

2.1.1 Breast anatomy

The breasts are located on the anterior aspect of the chest wall, typically from the level of the second rib to the level of the sixth rib. The breast tissue is situated within the superficial layer of the thoracic wall, anteriorly to the pectoralis major muscle (Hamdi, et al, 2005; Macéa & Fregani, 2006; Gefen & Dilmoney, 2007) (Figure 1). Beneath the deep layer of the superficial fascia, an area occupied by loose areolar tissue enables the breast tissue to an extent, to move over the pectoral fascia. The overlying skin is known to be highly non-linear and viscoelastic and is reported to vary substantially due to age and hydration status (Gefen & Dilmoney, 2007). The three layers of the skin are intimately connected but are very distinct in their nature, structure and properties. The epidermis protects the organism from the environment, while the fibrous dermis together with the hypodermis plays an essential role in protecting the skin from mechanical stress. The skin is required to accompany the mulitplanar movements of the body and to withstand a certain degree of mechanical constraint (Escoffier, Pharm, Rigal, Rochefort, Pharm, Vasselet, et al., 1989). The reported mechanical properties of the skin vary substantially within the literature, which has predominantly been attributed to wide variety in methods and devices employed (Finlay, 1970; Agache, Monneur, Leveque, & De Rigal, 1980; Escoffier et al., 1989; Clark, Cheng, & Leung, 1996; Silver, Freema, & De Vore, 2001).

The subcutaneous tissue and the corpus mammae have been identified as the two major structural components that make up the breast mass. The corpus mammae can be divided into two subcomponents: the parenchyma and the stroma, which are identified as the functional part of the breast (Page & Steele, 1999). The parenchyma is composed of ductular, lobular and alveolar structures, and commonly referred to as the glandular aspect of the breast. These glandular components are surrounded by dense connective tissues, known as the stroma, which acts as a supporting framework, composed of connective tissues, fat tissue, lymphatics, blood vessels and nerves (Hamdi et al., 2005; Macéa & Fregani, 2006; Gefen & Dilmoney, 2007).

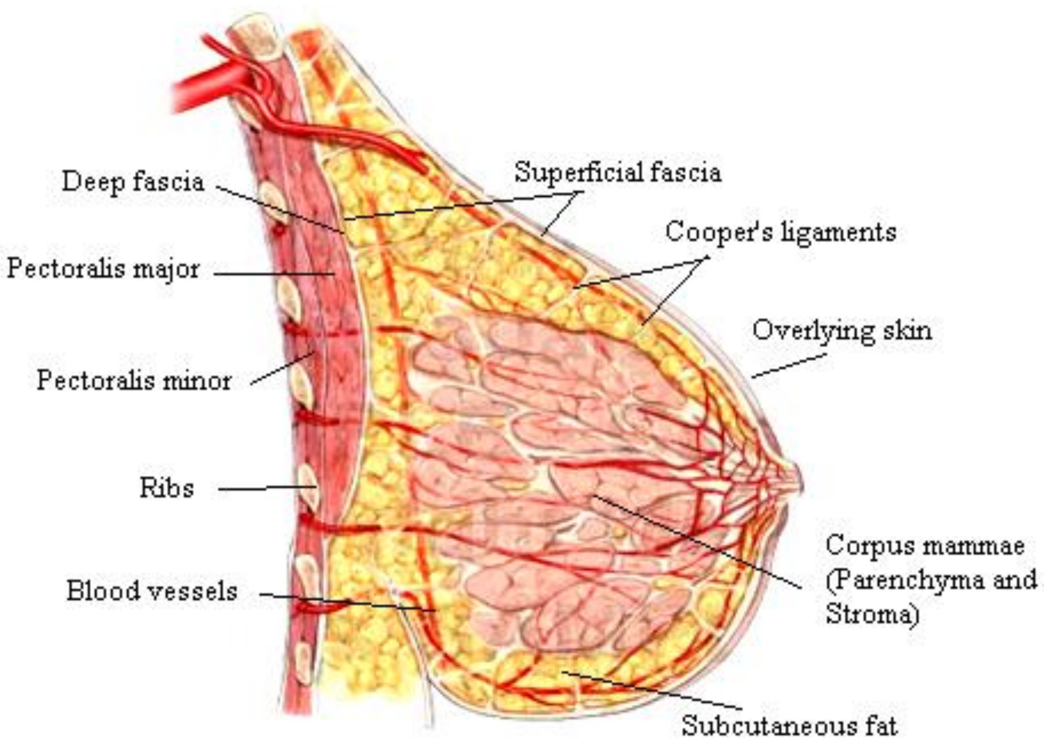

Figure 1. Sagittal view of the tissues of the breast.

The connective tissue in the stroma commonly referred to as the Coopers Ligaments, have been identified as providing suspensory support to the breast. However, their role in restricting overall breast movement and their actual mechanical properties are yet to be accurately defined (Page & Steele, 1999), and have only been approximated in the literature (Gefen & Dilmoney, 2007). Broadening the data available on the kinematics and forces subjected to the breast during physical activity will inform future research interested in the mechanical properties of the supportive breast tissues.

When considering the anatomy of the breast, it is apparent that it is like no other soft tissue within the body. There are certain negative factors associated with the breast that are considered influential to the female athlete such as; exercise-related breast pain and the magnitude of independent breast movement.

2.2 Exercise-related breast pain

The limited anatomical support within the breast enables the breast tissue to move, to an extent, over the chest wall, specifically driven by movements of the thorax (Bowles & Munro, 2009; Haake & Scurr, 2010). This independent breast movement commonly results in exercise-related breast pain, and has previously been reported to affect 72% of the exercising female population (Gehlsen & Albohm, 1980). A more recent study by Brown, White, Brasher, and Scurr (2013), identified a third of marathon runners (*n* = 1285) experienced exercise-related breast pain, and of that sample 17% of the symptomatic runners stated that exercise-related breast pain affected their exercise behaviour. With high percentages of females experiencing exercise-related breast pain and the potential barrier to exercise, one of the most common focuses of breast biomechanics research has been to gain a better understanding of the cause and measures taken to reduce exercise-related breast pain during exercise.

Little is known of the exact cause of exercise-related breast pain, however Mason, Page and Fallon (1999) reported a significant correlation between breast displacement and breast pain, and hypothesised that exercise-related breast pain occurs due to the tension placed on the skin, fascia, and nerves of the breast during large breast displacement, concluding that the relationship between these two variables may be due to at least one of the anatomical structures being stretched. Bowles and Steele (2003) supported the hypothesis made by Mason et al., (1999) when a significant increase in the distance between the clavicle and nipple was reported following a five minute treadmill run in an everyday bra (low breast support), when compared to the distance taken prior to the run. Bowles and Steele (2003) suggested that the increased magnitude of relative breast displacement in the lower breast support condition, when compared to a sports bra may be a result of internal damage to the supportive tissues within the breast. However, there are many factors that could have influenced the reported findings, such as position and orientation of the thorax and potential technical error in the video analysis system and software. The assumption proposed by Bowles and Steele (2003) can only be clarified through *ex vivo* examination or MRI examination, therefore the possible acute damage to the breast due to the magnitude of breast displacement during exercise remains unknown.

If damage was present when running in reduced breast support, it may be hypothesised that a female may alter their biomechanics to accommodate increases in breast kinematics and associated breast pain within this support condition. Furthermore, the examination of breast kinematics and strain on the breast within these studies were examined during short duration exercise (up to five minutes of running), and therefore may not represent the kinematics and forces subjected to the breast during prolonged run durations. Examining these data over more common running distances would provide a greater understanding of if/how breast biomechanics change over prolonged running.

In line with the suggestion of potential damage and strain to the breast by Bowles and Steele (2003), Scurr, Bridgman, Hedger, and White (2009b) explored the relationship of exercise-related breast pain and soft tissue strain between the clavicle and nipple during running. Mean peak strain between the clavicle and the nipple was 10% greater during running than in the static position. Furthermore, peak breast strain demonstrated a significant moderate correlation with breast comfort (r = 0.34), as soft tissue strain increased, breast pain also increased. Scurr et al., (2009b) suggested that these findings support Mason et al., (1999) previous hypothesis that pain may be caused by tension on the skin and fascia of the breast during motion. However, the r value reported for this correlation is low and may not be the key dependent variable when examining exercise-related breast pain.

A unique study by McGhee, Steele, and Power (2007) examined the effect of deep water running on three variables; breast displacement, breast velocity and exercise-related breast pain. The authors postulated that the buoyant forces associated with deep water running might help counteract the gravitational forces that accelerate the breast downward during land-based running, and therefore will decrease the exercise-related breast pain felt. The results suggest that while deep-water running elicited a greater perception of physical exertion, the perceptual responses of breast discomfort were significantly reduced. Furthermore, the magnitude of relative breast displacement was not different between land-based running and water running, but a significant reduction in breast velocity was reported. Although there are obvious mechanical differences between these two modes of running (velocity of segment movement and ranges of motion), which could have affected the reported findings, it is important to reiterate the magnitudes of breast displacement were not different, and it was the velocity of the breast which correlated to breast discomfort. In addition, it is important to consider the natural starting position of the breast in these two conditions, the breasts would have been lifted in the water due to the buoyant force, which may have limited the peak downward displacement and velocity.

This study provides recommendations of different exercise modalities (water running) where breast pain is substantially reduced for females who associate exercise-related breast pain as a barrier to exercise, enabling training and performance to be preserved.

Until recently, breast kinematics were examined within the frontal plane, which does not account for certain rotational movements of the thorax which can significantly influence the magnitude of relative breast movement (Scurr et al., 2009a). This could have resulted in misleading relationships reported between breast kinematics and breast pain. Furthermore, previous publications had only examined one or two kinematic variables. It was suggested that the exploration of the first and second derivatives of displacement may help determine the cause of breast pain. Scurr, White, and Hedger (2010a) progressed the methods for quantifying breast kinematics in multiple planes of movement (anteroposterior, mediolateral and vertical), and explored the relationship between multiplanar breast displacement, velocity, and acceleration to breast discomfort/pain. Multiplanar breast velocity displayed the strongest correlation (r = 0.61) to breast discomfort/pain, supporting the previous findings of McGhee et al., (2007). Due to the strength of the relationship between breast pain and relative breast acceleration, Scurr et al., (2010a) suggested that breast acceleration was not as effective for monitoring breast pain and a greater emphasis should be placed on the bras ability to reduce breast displacement and breast velocity. However, in line with the second law of motion (F=ma), a force is created by a change in the acceleration of an object, and is frequently related to stress and strain of an object. The force subjected to the breast tissues is therefore dependent upon its mass and acceleration. As the mass of the breast remains constant during movement, the acceleration will be the determining factor influencing the magnitude of force and the potential resulting pain experienced.

The identified relationships between breast kinematics and exercise-related breast pain during exercise highlights the importance of reducing multiplanar breast kinematics through adequate breast support (Bowles & Steele, 2003; Mason et al., 1999; Scurr et al., 2010) for exercising females. Although the exact cause of exercise-related breast pain remains unknown, these analyses provide an indication of the strength of the relationship between biomechanical variables of the breast and exercise-related breast pain. Previous publications have only examined breast kinematics and breast pain over short exercise durations. If exercise-related breast pain is caused by tension placed upon the skin and fascia of the breast, then exercising for a longer period of time may heighten the risk of strain and potential permanent damage to the breast tissues. This will not only increase the magnitude of breast discomfort and pain felt, but potentially negatively influence a female performer, and ultimately deter a female from exercising. It is unknown to what extent breast pain influences the biomechanics of a female runner, further exploration of these variables over common running distances will help progress knowledge on the effect of exercise-related breast pain on sports performance. In order to reduce the impact of these negative factors on female athletes, high breast supports such as sports bras are currently promoted within the literature.

2.3 The evolution of a sports bra

Sport bra design has significantly progressed in recent years with the aim to minimise breast motion during exercise (Starr et al., 2005). In addition to reducing breast motion, reducing exercise-related breast pain and improving overall comfort have been regarded as important considerations for sports bra design (Bowles et al., 2011). It is assumed that the reduction in these two negative factors will ensure females are not deterred from exercising, are able to perform without restraints, and importantly that sporting performance may be enhanced. The first prototype sports bra was created in the late 1970s, with two American women sewing two jockstraps together, and suggested that this would provide more support whilst they exercised than everyday bras (Schuster, 1979).

The first documented consideration of the effect of breast movement on the female athlete dates back to 1977. Haycock (1977), cited in Gehlsen and Albohm (1980), investigated the occurrence of injuries to the breast in a survey of 115 colleges and universities. Few breast injuries were reported, however, the prevalence of breast tenderness and pain during and post-exercise was high, with 72% of female athletes reporting this during various exercise modalities. Haycock's (1977) findings prompted the following recommendations for sports bra design; to provide firm support, limit motion of the breast, and to be made of firm elastic, nonabrasive, sturdy, and non-allergenic material. However, these recommendations were not evidence based with the magnitude and trajectories of the breast not considered during different exercise modalities. Examining these variables would have provided manufacturers with the fundamental movement patterns of the breast during exercise, and enable them to design sports bras which reduce breast movement in this pattern. Furthermore, validation of the magnitude of breast movement reduction would ensure the performance of the sports bras could be monitored.

Within recent years the structural components and material properties of sports bras have evolved dramatically from the first sports bra, with Lycra®, Elastane and Cool Max® materials incorporated into the more complex sports bras (Bowles et al., 2011). These materials are known to be lightweight, have high levels of both elastic and recovery properties, and are suggested to provide more support to the breast than a bra made out of Cotton (Zhou, 2011).

Currently, there are three major sports brassiere designs; encapsulation, compression, and combination (Yip & Yu, 2006; Krenzer, Starr, & Branson, 2005; Bowles et al., 2008; 2011) (Figure 2). Compression sports bras have been designed to restrict the amount of movement of the breast by compressing and flattening the breast tissue against the body, and redistributing the mass evenly. On the other hand, an encapsulation bra provides more structured support, harnessing each of the breasts individually. This design is thought to be more effective for larger breast sizes (C-cup upwards) (Starr et al., 2005).

Figure 2. Shock Absorber, UK, compression (B5064), encapsulation (N109), combination (B4490), and mode specific (RUN bra) sports bras.

A more recent sports bra design is the combination sports bra. This bra incorporates both compressive and encapsulating traits. The encapsulation aspect usually sits beneath the compressive component, combining the most effective components of each bra. Until recent years, sports bra design were dominated by these three types of bra, however, in recent years, mode specific sports bras have been manufactured, such as running, ball and racket sports bras (Shock Absorber, UK).

Alongside the design and shape of the sports bra, it is suggested that the fabric used largely affects the quality and effectiveness of support provided (Zhou, 2012). Sports bras need to possess diverse mechanical properties. Specifically they need to encompass both elasticity to enable upper body movement, and enough stiffness to prevent breast movement (Page & Steele, 1999). To enable natural breathing the sports bra needs to have a sufficient amount of elastic material along the horizontal plane (Bowles, Steele & Chaunchaiyakul, 2005). Conversely, the elastic material through the vertical plane must be kept to a minimum to prevent vertical displacement of the breasts (Page & Steele, 1999).

Although sports bras have been reported to significantly reduce breast kinematics (Scurr et al., 2009a; 2010a; 2011; White et al., 2009), Bowles, Steele, and Munroe (2011) identified that females can be deterred from wearing a sports bra during physical activity due to factors such as; shoulder slippage, straps digging into shoulders and the tightness of the sports bras, specifically around the chest band. These data are of importance to bra designers and manufacturers and should be considered, however, the effect of different breast supports on sports performance is of interest within applied research, with wide reaching applications to exercising females.

Currently, breast biomechanics literature has focussed on the effect of different breast support conditions on breast biomechanics. It has been shown that different designs and types of breast support can significantly influence many aspects of breast biomechanics, specifically the magnitude of breast kinematics. However, the effect of breast support and changes in breast biomechanics on running biomechanics has received little attention.

2.4 The effect of breast support on breast biomechanics

Quantitative investigation of breast movement during exercise dates back to the 1980s, with breast displacement first quantified and reported in the frontal plane (Gehlsen & Albohm, 1980; Lorentzen & Lawson, 1987; Lawson & Lorentzen, 1990). The primary aim of these three studies was to quantify the level of support provided to the breast in the available sports bras. Gehlsen and Albohm (1980) were the first to highlight the trajectory of the breast during a running stride. In order to calculate breast displacement, one marker was positioned on the centre of the breast, over the bra, and one marker positioned on the centre of the left clavicle, the difference in movement between these two markers was used to define the amount of breast movement allowed by each bra. A horizontal figure-of-eight pattern represented the frontal plane displacements of the body and the breast. These data helped inform sports bra manufacturers of effective garment design, with the underlying movement of the breast quantified during running defining where the structural support components were required.

Mason et al., (1999) demonstrated that the level of breast support worn during running can significantly influence the magnitude of vertical breast displacement and acceleration. One of the key findings of this paper was that the sports bra condition was more effective at reducing vertical breast displacement and acceleration than a crop top support and a fashion everyday bra. Furthermore, Mason et al., (1999) speculated that the magnitude of breast movement and associated breast pain in lower levels of breast support are disincentives to exercise, and the use of sports bras may enhance the enjoyment and may assist in improvements of sporting performance in a significant proportion of the female population.

Early publications within breast biomechanics examined breast kinematics within the frontal plane only; an important consideration for these data was the influence of thorax kinematics on the magnitude of multiplanar breast kinematics. The three rotations of the thorax (thorax pitch, roll and yaw) are not accounted for within these publications and could substantially influence the relative movement of the breasts. A more complex laboratory set up, incorporating more than one camera and more complex marker positioning on the relative body segment are required to gain a more accurate 3D representation of breast kinematics during running. To eliminate the movement of the body from that of the breast, previous studies used a single body reference marker; the displacement of this marker was then subtracted from the displacement of the breast (Gehlsen and Albohm, 1980; Mason et al., 1999; Starr et al., 2005). This method only gives an indication of the influence of the thorax on frontal plane breast movement. However, both the body and breast move in more than one dimension during running, with the upper body known to have six-degrees-of-freedom (6 *dof*) (three translations and three rotations) (Scurr et al., 2009a; Zhou et al., 2011).

The importance of reporting all components of relative breast kinematics was not emphasised nor reported until the work of Scurr, Galbraith, Hedger, and White (2007). Scurr et al., (2007) emphasised the need for a valid and reliable method to quantify relative breast kinematics in multiple planes of movement, by eliminating the 6 *dof* of the body. Scurr et al., (2007) stated that the presentation of only the vertical component of breast kinematics would result in a misinterpretation of overall breast kinematics, and previous recommendations provided for optimising sports bra design, and relationships to subjective measures such as breast pain would be lacking.

Scurr et al., (2009a) detailed the method for quantifying relative multiplanar breast kinematics during the running gait cycle in the first full paper quantifying relative multiplanar breast displacement. The aim was to assess the magnitude and trajectory of relative and absolute breast kinematics in three-dimensions (3D), with the intention to determine the influence of the body on breast kinematics (Figure 3). The key finding of this study was the description of the multiplanar breast displacement relative to the trunk during the gait cycle. Four phases of multiplanar breast displacement were identified during each gait cycle, and the calculation of relative breast displacement significantly reduced the magnitude of breast displacement when compared to the absolute magnitudes. These progressions in methodologies ensure breast biomechanics are reported accurately, and that all components of breast kinematics are presented.

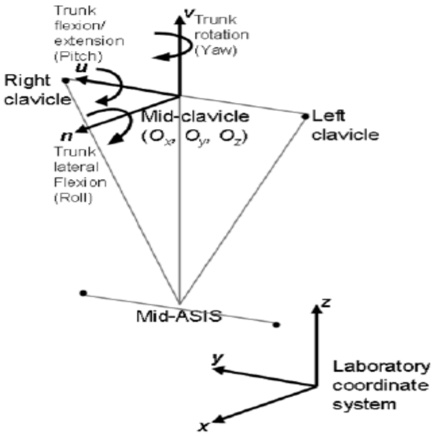

Figure 3. Axis of global and local coordinate systems. *U* defines an axis from the left to the right clavicle, *v* defines an axis from the mid-ASIS (virtual point) to the mid clavicle (virtual point), and *n* defines an axis of the cross-product of *u* and *v*. Axes were established using a left-hand coordinate system (Scurr et al., 2009a).

Scurr et al., (2009a) suggested that the elimination of the 6 *dof* movement of the trunk is essential for reporting relative breast displacement, which may be overestimated without this analysis. Scurr et al., (2009a) highlighted a crucial consideration of the reference points utilised to define the trunk segment. The ASIS and clavicle reference points belong to two separate segments, therefore counter rotation can occur between them (e.g. at the shoulder and pelvis). Scurr et al., (2009a) stated that previously published trunk marker sets (Nguyen & Baker, 2004; Sartor, Alderink, Greenwald, & Elders, 1999) had to be excluded as the design of bras may mean that markers were obscured. Although the restricted use of certain reference points is evident for studies examining breast supports, it was important to progress this marker set to ensure reference points from two different segments are not included. Scurr et al., (2010a) made progressions from the previous trunk marker set by identifying three non-collinear reference marker positions on the same segment, which would not be occluded by the different breast supports. Retro-reflective markers were positioned on the suprasternal notch, and on the left and right anterioinferior aspect of the 10th ribs.

As mentioned previously, Scurr et al., (2010a) was the first to present relative multiplanar breast displacement, velocity and acceleration during running. These data extended upon the knowledge of breast biomechanics, with a greater understanding of how the breast move in greater detail. Currently, these data have only been reported over short exercise durations (up to five minutes). Considering the government guidelines for exercise prescription recommending 30 minutes of exercise (equivalent of a five kilometre run paced at 10 km.h^{-1}) five times a week to maintain a healthy lifestyle (Department of Health, UK, July 2011), previous publications have not examined breast biomechanics over a common exercise duration. Examining breast kinematics and forces over prolonged exercise durations should be a consideration when examining the effect on sporting performance and product testing, since these values may increase with repeated loading over time.

Bowles and Steele (2004) reported significant increases in the magnitude of relative vertical breast displacement after five minutes of running when compared to the third and fourth minute in a 'poor' breast support condition, and reported a significant inferior extension in the relative static position of the nipple post run. Bowles and Steele (2004) promoted the use of a sports bra during prolonged running and postulated that inadequate breast support could pose a greater risk of damage to the breast for female runners. Based upon the findings of Bowles and Steele (2003), it is hypothesised that the magnitude of breast kinematics may increase over an extended run distance, and the magnitude of strain placed on these tissues may increase over time, due to the repeated loading over extended running distances. These findings may have possible implications for running biomechanics, product design, and potential strain and damage to the breast tissues. Investigation of breast biomechanics and the effect on running biomechanics over a more common running distance is warranted to further examine the implications of wearing different breast supports during prolonged running.

2.5 The effect of breast support biomechanical and physiological variables

Sports bras are promoted as a beneficial piece of sporting apparel for the female athlete, with significant reductions in two commonly identified negative factors; the magnitude of relative breast kinematics and exercise-related breast pain. However, there is little empirical evidence to suggest sports bras are beneficial to sporting performance, or how changes in breast biomechanics during exercise may influence the biomechanics of the body and vice versa.

The available published studies on the effect of breast support and breast biomechanics on sports performance have focussed on biomechanical parameters, such as ground reaction forces (Verscheure, 1999; Shivitz, 2001; White et al., 2009), kinematic analysis of running (Eden et al., 1992; Boschma et al., 1995) and electromyography (Scurr et al., 2010b), and a few papers available on certain physiological measures (Bowles et al., 2005; White, Lunt, & Scurr, 2012). The modes of exercise examined in these papers, abstracts and

Masters Theses are; jumping, overground (White et al., 2009) and treadmill running (Eden et al., 1992; Boschma et al., 1995; Shivitz, 2001; White et al., 2012), drop jumps (Verscheure, 1999), and cycling (Bowles et al., 2005).

Eden et al., (1992) presented an abstract that supported the hypothesis that breast kinematics may significantly differ when stride mechanics are altered. Significant increases were reported in the magnitude of downward and medial breast displacement when participants ran with their natural stride rate (mean data = 86 strides per minute), when compared to a forced quicker stride rate of 96 strides. This study was the first to report significant changes in breast kinematics due to alterations in running kinematic parameters, and suggests a relationship between breast and body biomechanics. The implications of these findings for the female athlete, specifically during running, need to be considered in future research. If changes in stride mechanics can significantly influence the magnitude of breast kinematics, it is hypothesised that other mechanical alterations may affect breast biomechanics, such as segmental movement patterns. Investigation within this area will further the understanding of the relationship between the body and the breast during running.

While the aim of the study by McGhee et al., (2007) was not to examine the effect of breast support on biomechanical performance variables, McGhee et al., (2007) reported differences in stride frequency between breast support conditions during the treadmill running. These data suggest that females altered their stride mechanics dependent upon the breast support worn. McGhee et al., (2007) hypothesised that the slower stride frequency adopted in the lower level of breast support was a strategy used by the participants to minimise breast discomfort. When running on a treadmill at a constant velocity, reductions in stride frequencies may indicate alterations in additional gait parameters. Longer time spent in the stance or swing phase, and changes in segment kinematics have been associated with changes in stride parameters. Research has identified that alterations in step characteristics (step length/frequency) and knee flexion have previously been linked to a reduction in running economy during a given task (Cavanagh & Williams, 1982; Saunders, Pyne, Telford, & Hawley, 2004), which could significantly impact upon running performance. McGhee et al., (2007) did not examine any additional kinematics parameters, and therefore it was unknown if different breast support conditions elicited changes in key kinematic parameters of running as well as stride parameters.

Previous research by Boschma (1995) did however explore the effect of three different breast supports (no support, moderate support, and full support) on breast kinematics and the following running kinematic parameters: stride rate, stride length, vertical trunk displacement, front arm angle range of motion, arm angle range of motion, and vertical breast displacement, during treadmill running. This study was based upon previous hypotheses suggesting breast motion may affect sports performance and reduce exercise adherence. Breast support conditions did not alter the kinematic running parameters across the sample (n = 15); however emphasis was placed upon individual differences following similar trends, with vertical trunk displacement reducing as breast support decreased. Boschma (1995) suggested that data should be presented for individual participants when examining the effect of breast motion on kinematic variables. However, the consideration of this type of analysis contrasts the aim of statistical analysis carried out on mean data from a sample of participants, with a large sample strengthening the generalisability of the conclusions drawn.

Considering the literature within this area, the most prevalent adaptive kinematic response to running in different breast supports appears to be alterations in step/stride parameters (Eden et al., 1992; McGhee et al., 2007; Boschma, 1995). This common finding could be due to the limited kinematic variables measured to date. Due to the location of the breast tissue on the thorax, it is assumed that the kinematics of the thorax will influence the relative breast kinematics. Boschma (1995) did not employ an in-depth kinematic analysis of the thorax segment, with only vertical displacement of this segment reported; therefore, influential movement patterns (i.e. three rotational degrees of freedom) of this segment were not examined. In order to gain a better understanding of the link between the thorax and relative breast kinematics, the quantification of thorax kinematics alongside breast kinematics is imperative. Haake and Scurr (2010) suggested that the thorax segment drives breast kinematics; further exploration of the relationship between these two variables will inform breast biomechanics research and progress the understanding of the links between the breast and body. In addition, it may be advantageous to consider the movement patterns of segments along the kinetic chain and the relationships to the breast in different breast support conditions to gain a holistic view of changes in an individual's kinematic profile. A full body kinematic analysis in different breast support conditions may indicate potential benefits or detriments to biomechanical running performance depending upon the breast support worn.

Not only has it been postulated that running kinematics may be affected by level of breast support, Shivitz (2001) and White et al., (2009) also found differences in running ground reaction forces (GRF) in different breast support conditions. Shivitz (2001) investigated the active vertical ground reaction forces, vertical stiffness, and stride frequency during running. Shivitz (2001) detailed the importance of understanding how different breast supports may influence the biomechanics of a female runner and potential injury risks. Taking on the suggestion from Boschma (1995), Shivitz (2001) incorporated a multiple single-subject design making it possible to detect significant changes on a case by case basis. The results indicated significant increases in active vertical GRF as level of support increased for thirteen out of the seventeen participants. Shivitz (2001) was the first to identify differences in kinetic parameters of running in different breast support, and suggested that participants' altered their running kinematics due to the magnitude of breast displacement experienced, and that these changes could significantly affect their running performance. Changes in running kinematic parameters may be an obvious explanation for differences in GRFs, however, kinematic parameters were not examined within this study, and therefore it is unknown which parameters were altered to elicit the changes in GRFs.

White et al., (2009) also investigated changes in kinetic parameters of gait between breast support conditions during overground running. Mediolateral force was significantly greater in the no breast support condition compared to the compression sports bra. White et al., (2009) suggested this difference could be related to the significantly greater magnitudes of mediolateral breast displacement in the no support condition compared to the compression sports bra, which may have altered the participants running biomechanics. White et al., (2009) identified a trend similar to Shivitz (2001), in the vertical peak impact force, with the no support condition eliciting a lower force compared to the high levels of breast support. Based upon these data, the authors suggested the participants may experience high levels of stress within the high breast support condition, which could lead to increased physiological demand and over time may have injury implications. Although the GRFs increased in the sports bra condition, these values may still fall within normative GRF values during running. Hreljac (2004) stated individuals will experience impact forces ranging from 1.2 to 5 times body weight (BW) during running. Therefore, the impact forces reported within both Shivitz (2001) and White et al., (2009) (< 2.5 BW) remain within normative values, and the potential of greater risk of injury or potential detriment to performance in this condition is unlikely.

Both White et al., (2009) and Shivitz (2001) discuss the possibility of kinematic alterations (e.g. attenuating the force through flexion of the lower extremity joints) during running when wearing insufficient breast support, in an effort to reduce the breast movement and discomfort experienced, however, kinematic analyses were not conducted. Gaining more data on the effect of the magnitude of breast support on running biomechanics will provide key recommendations for the most appropriate breast support for maintaining running performance. However, to ensure these findings are applicable and generalisable, it is important to consider the exercise length and protocol implemented for data collection.

Electromyography (EMG) enables the quantification of the physiological process of a muscle to generate force and create movement (De Luca, 1997). This biomechanical tool provides information regarding the neural system driving the 6 *dof* movements of body segments. Due to the link between segmental movements and muscle activity, and supporting the hypothesis which states that segmental kinematics may differ between breast support conditions (Shivitz, 2001; White et al., 2009), it would be beneficial to investigate both kinematics and EMG during running. Running kinematics that enables an individual to reduce demand on the active muscles has been associated with reduced energy costs and more economic running (Abe et al., 2007; Bourdin et al., 1995).

Currently, one abstract is available which examines EMG of upper body muscles in different breast support conditions during treadmill running. Scurr, Bridgman and Hedger (2010b) suggested that reductions in upper body muscle activity in high levels of breast support may benefit performers. The level of breast support worn did not affect EMG activity of the upper and lower trapezius, erector spinae, and anterior deltoid. However, the higher breast support conditions did significantly reduce pectoralis major activity, when compared to the no bra condition. The level of anatomical support provided by the pectoralis major to the breast remains unknown, Scurr et al., (2010b) proposed that an increase in pectoralis major activity when the level of support is reduced suggests that this muscle may contribute to internal support of the breast during active movement. The Cooper's ligaments, along with the skin represent the primary supportive structures for the breast tissue. These ligaments extend inwards from the outer skin and attach to the deep fascia of the pectoralis major muscle (Hamdi et al., 2005). Therefore, it may be relevant to examine the relationship between breast kinematics and the pectoralis muscle further, since activation of upper body muscles may be affected by the magnitude of breast kinematics, alterations in segmental movements, and a potential 'tensing' response brought on by exercise-related breast pain. In conclusion, Scurr et al., (2010b) suggested that differences in muscle activity seen in different breast support conditions may be linked to alterations in upper body kinematics during running, and could influence an individual's running economy, which has been defined as a crucial parameter for determining running performance (Saunders, Pyne, Telford, & Hawley, 2004; Foster & Lucia, 2007).

Alongside biomechanical measures, certain physiological measures may be influenced by the breast support worn during exercise. To examine the common complaint of tight chest bands within these garments, Bowles et al., (2005) investigated whether sports bras impede respiratory function. During treadmill running and maximal cycle ergometry, respiratory functions were measured in different breast support conditions. The results indicated that the pressure of the sports bra on the chest was significantly greater than the everyday bra; however no significant difference in lung volumes were reported between the breast support conditions. The investigators professionally bra fitted the participants, and therefore the results suggest that a correctly fitting sports bra did not impede respiratory function. However, the method for bra fitting was not reported and may have influenced the results presented. Women wearing ill-fitting sports bras could therefore still experience tightness around the chest, and may find their respiratory function impaired, which may be detrimental to performance. Confirmation on the effect of breast supports on respiratory function remains unknown for ill-fitting bras.

In line with the work of Bowles et al., (2005), White, Lunt, and Scurr, (2011) explored the effect of breast support on ventilation during treadmill running. Breathing frequency and ventilatory equivalents were lower without breast support when compared to an everyday bra and a sports bra, and tidal volume was greater when participants ran without breast support. The results suggest that wearing breast support changed ventilatory variables at the onset of running, compared to bare-breasted running. However, it is important to consider the application of these results, firstly, few women of the breast size examined (mode of 34 DD) are expected to run without breast support, and secondly, the run duration examined was not representative of a common running distance. These findings are interesting and suggest that the magnitude of breast kinematics and exercise-related breast pain may drive changes in physiological measures. Future research could extend upon this work with progressions in the experimental design to provide more ecologically valid results.

The aforementioned publications provided the first data examining the effect of breast support on both biomechanical and physiological measures of running. However, these publications have examined these variables over two to five minutes of exercise. It is suggested that the criteria for a steady state of running is between three and five minutes, based upon the limitations of the oxidative system (Hardin, Van Den Bogert, & Hamill, 2004). Data available on the criteria for a biomechanical steady state of running are sparse, although Campbell et al., (2007) suggested that two minutes was long enough to achieve a consistent gait pattern. However, changes in running kinematics have been linked to the activation pattern of the working muscles and the cost of running (Abe et al., 2007), and therefore will influence when a global steady state is reached. Furthermore, Lavcanska, Taylor, and Schache (2005) proposed six minutes as a criteria for ensuring participants are familiarised with treadmill running, and recommended that kinematic data should not be collected prior to this time to ensure that any changes are not due to the familiarisation period. It is therefore suggested that previous data collected on the effect of breast biomechanics on biomechanical and physiological parameters may not be representative of a steady state.

Furthermore, it has been found that running kinematics may change over time (Williams & Cavanagh, 1987; Williams, Snow, & Argruss, 1991; Hardin, Van Den Bogert, & Hamill, 2004; Abe et al., 2007; Candau, Belli, Millet, Georges, Barbier, & Rouillon, 1998). The biomechanical parameters most frequently reported to change over distance running are step and stride characteristics (Hunter & Smith, 2007; Williams et al., 1991; Candau et al., 1998), greater forward lean of the trunk (i.e. thorax) (Elliot & Ackland, 1981; Elliot & Roberts, 1980; Williams et al., 1991), and changes in knee flexion (Dierks, Davis, & Hamill, 2010; Williams et al., 1991). These changes have been linked to increases in metabolic cost and poor running economy. The changes in biomechanical parameters and running economy vary within the literature with contradictory results reported. Hunter and Smith (2007) suggest that the disparity in this research area is as a result of differences in running protocols employed, and considerable inter-individual differences with some runners being noticeably more sensitive to mechanical alterations, while others maintain a constant mechanical running form. To progress the external validity of the effect of breast support on breast and body biomechanics it is suggested that testing protocols should be extended from these short durations to more common running distances. In line with the government guidelines for a healthy lifestyle, (30 minutes of exercise, five times a week) a 30 minute run is equal to a five kilometre run at 10 $km.h^{-1}$ pace.

Additional research areas within biomechanics such as load carriage, footwear, and gait manipulations have established the impact of changes in segmental running kinematics, electromyography, and other biomechanical measures during running. Research investigating the effect of load carriage on walking and running has predominantly focused on energy costs, specifically focussing on load distribution (Abe, Yanagawa, & Niihata, 2004; Datta & Ramanathan, 1971; Knapik, Harman, & Reynolds, 1996). Changes in the distribution of load has been linked to greater forward lean of the thorax (Knapik et al., 1996), greater angular accelerations of the torso (Bobet & Norman, 1984), and changes in step characteristics (Harman et al., 1992), which have been associated with increased energy costs and earlier onset of muscular fatigue. Although the mass of these loads are substantially greater than the mass of the breast tissue, potential links could be made to the distribution of breast mass on the thorax in different breast support conditions, taking the different structure and design of breast supports into consideration (e.g. compression vs. encapsulation).

A wealth of information is available on the effect of footwear on biomechanical measures, with publications focussing on the reduction of GRFs (Kersting & Brüggemann, 2006), alterations in ankle and knee kinematics (Cheung & Ng. 2007; Lilley, Stiles, & Dixon, 2013), and changes in muscle activation (Kerr et al., 2008; Divert et al., 2005). It has been reported that individuals employ different biomechanical strategies (i.e. alterations in running kinematics) to account for modifications in footwear (e.g. cushioning properties) (Kersting & Brüggemann, 2006). It is of interest to relate the findings of this research area to breast biomechanics from the perspective of ergonomic aids. The reduction in negative factors of independent breast movement in a sports bra may ensure females are not required to employ different biomechanical strategies to accommodate magnitudes of breast kinematics and breast discomfort and pain. Ensuring running biomechanics are maintained could facilitate the maintenance and preservation of training and may benefit female runners.

2.6 Summary of literature review

Currently no publications are available investigating the effect of breast support conditions on breast biomechanics, upper and lower running kinematics and electromyographical analysis simultaneously. An integrated examination of these biomechanical tools would provide a holistic biomechanical understanding of the female runner in different breast support conditions.

Previous publications have explored biomechanical running parameters in different breast support conditions, however, the application of these findings are restricted to shorter exercise durations and therefore limit the external validity of this work. Examining breast and body biomechanics over a more common running distance would progress the work conducted within breast biomechanics, and would extend the knowledge of the effect of breast support on running biomechanics.

CHAPTER THREE. PILOT STUDY

MULTIPLANAR BREAST KINEMATICS DURING A PROLONGED TREADMILL RUN

3.1 Introduction

Previous publications within breast biomechanics have established differences in the magnitude of breast kinematics between breast support conditions during running (Mason et al., 1999; Scurr et al., 2009; 2010; White et al., 2009), with a high breast support (i.e. a bra designed to reduced breast motion) frequently promoted as an effective and important part of a females sport kit. While these publications provide recommendations for effective breast support for exercising females, the findings can only be applied to short running bouts (two to five minutes of running) due to the duration of the runs completed in the previous publications (Mason et al., 1999; Scurr et al., 2009; 2010; Starr et al., 2005).

Bowles and Steele (2003) reported significant increases in vertical breast displacement when wearing an everyday bra during a five minute treadmill run, with increases reported between the first minute and the third, fourth, and fifth minute of running. These data indicate a significant increase in breast displacement between the start and end of a five minute run. Bowles and Steele (2003) postulated that the differences were elicited by the repeated loading and potential strain on the anatomical constraints of the breast in the low breast support condition. With previous recommendations for breast support design, and quantification of sports bra performance previously based upon the findings collected only up to five minutes of running (Starr et al., 2005, Mason et al., 1999), it is of importance to understand if the magnitude of breast kinematics continues to increase over a prolonged run, in order to increase the external validity of this research area. Furthermore, if breast kinematics continue to increase over a prolonged run, it is important to investigate the influence of this on running biomechanics.

The focus of this programme of research was to investigate the effect of breast support on breast and body biomechanics during treadmill running. An influential decision prior to data collection was the length or distance of run implemented during the testing sessions. The government guidelines for exercise prescription currently recommend 30 minutes of exercise five times a week in order to maintain a healthy lifestyle (Department of Health, 2011). Thirty minutes of running paced at 10 km.hr^{-1} would meet the guidelines for one of these weekly activities, and the distance completed would be five kilometres. Based upon this example and to progress the external validity (more commonly run distances) of this research area, a steady state five kilometre treadmill run was implemented for the current pilot study. There were two aims of this study, firstly, to determine if a five kilometre treadmill run was an appropriate run distance to define a prolonged run for the current programme of work, and secondly, to determine if breast kinematics change over a prolonged treadmill run within and between low and high breast support conditions. These data will help inform the protocol of the current programme of work, and inform future research protocols for sports bra product testing and breast biomechanics research.

3.2 Hypotheses

H_1 - Based upon the work of Bowles and Steele (2003), it was hypothesised that multiplanar breast kinematics would increase over the duration of the five kilometre run in both a low and high breast support conditions.

3.3 Methods

3.3.1 Participants

Nine females (exercising for 30 minutes at least five times a week) participated in the study. Without effect sizes and power statistics available within previously published breast biomechanics research, the sample size of the current study was based upon sample sizes of the available literature. In order to inform future research within this area, both effect and power statistics will be presented *post-hoc* within this thesis. Participants had not had any children, had not experienced any surgical procedures to the breast, and were of either a 34B or 34D bra size. Participants had an average (SD) age of 21 years (1 year), body mass 65.4 kg (6.8 kg), and height 1.70 m (0.10 m).

3.3.2 Procedures

Participants performed two five kilometre treadmill run trials on separate days. To ensure participants time in the menstrual cycle did not vary substantially these two laboratory sessions were conducted from 24 to 72 hours apart, once in a low breast support (everyday M&S t-shirt bra) and once in a high breast support (B4490, Shock Absorber, sports bra). These two bras have been employed previously in breast biomechanics literature to define a low and high breast support (Scurr et al., 2010a; White et al., 2009), with the high breast support proposed as the market leader at the time of testing. Participants were required to perform an additional bare-breasted (BB) treadmill run, due to the discomfort associated with this condition, participants ran without breast support for only two minutes (Scurr et al., 2009; McGhee et al., 2012). Participants selected a comfortable running speed, which they felt they could maintain for the duration of the run, this ranged from 9 $km.hr^{-1}$ to 10 $km.h^{-1}$. Once selected, this speed remained constant throughout all trials for each participant. The treadmill was set level (0% gradient) based upon the findings of a pilot study presented in appendix A, which demonstrated no differences in upper body kinematics between a treadmill set at a 1% incline and 0% level treadmill.

Four retro-reflective markers (12 mm in diameter) were positioned on the following anatomical landmarks; the suprasternal notch, the right nipple, and the left and right anterioinferior aspect of the 10th ribs (Scurr et al., 2009; White et al., 2009; Scurr et al., 2010). During the bra conditions, participants repositioned the markers on the material of the bras, directly over the nipple via visual inspection (Starr et al., 2005; White et al., 2009; Scurr et al., 2010). A fifth marker was positioned on the lateral aspect of the left heel to determine gait cycles.

Three-dimensional coordinates of the five markers were tracked by eight 200 Hz calibrated Oqus infrared cameras (Qualisys, Sweden). The eight cameras positioned in an arc around the treadmill, and in the centre of the laboratory to maximise the field of view of all cameras. Cameras recorded for the final ten seconds of the initial two minutes of the five kilometre runs, and for ten seconds within the final 100 m of each kilometre interval following this (e.g. 900 m, 1900 m, etc.). Data collected at each kilometre interval enabled comparisons with previous publications in breast biomechanics which have collected breast kinematics after five minutes of running (i.e. first kilometre run at 9 $km.hr^{-1}$ would take 6.6 minutes), and examine breast kinematics over a distance previously not investigated, which will determine if breast kinematics change over a prolonged run.

3.3.3 Data processing

The markers were identified and three-dimensional data reconstructed in Qualisys Track Manager (QTM) software. Three-dimensional coordinates were exported to a frequency analysis program in MATLAB (MathWorks, UK). The frequency component of the data was assessed using a Fast Fourier Transformation (FFT) in MATLAB. The FFT showed the amplitude of the data point plotted against the frequency component, enabling the identification of data that should be retained and the noise component that is attenuated

(Winter, 1990). A cut-off frequency of 13 Hz was selected for the low pass filter based on this process.

The global coordinate system (GCS) identified x as the line of progression on the treadmill (anterioposterior), y as mediolateral, and z as vertical (Figure 4). In order to establish relative breast kinematics, independent to the 6*df* movement of the thorax, a mutually orthogonal local coordinate system (LCS) converted absolute right nipple coordinates (x′, y′, z) to relative coordinates (x″, y″, z″) using a transformation matrix (Foley, van Dam, Feiner, & Hughes, 1995; Scurr, *et al.*, 2010).

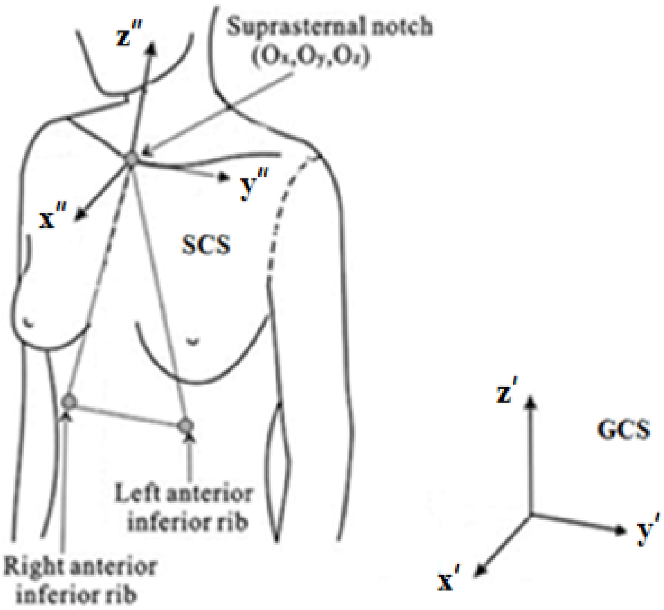

Figure 4. Orientation of the global and local coordinate systems, and marker locations of the thorax segment.

In order to determine gait cycles, instantaneous velocity of the heel marker was derived from the anteroposterior coordinates. Heel strike for each running gait cycle was identified as the velocity of the heel marker reached a peak positive progression (Zeni, Richards, & Higginson, 2008), with a full gait cycle identified as heel strike to heel strike of the ipsilateral heel. Using the relative nipple coordinates, minima positional coordinates were subtracted from maxima coordinates to calculate breast displacement in each plane, normalised to the percentage of each gait cycle (n = 5), at each interval of the five kilometre run. First (velocity, $m.s^{-1}$) and second (acceleration, $m.s^{-2}$) derivatives of the 3D coordinates were calculated for each sample, with the peak value recorded for both of these variables averaged over five gait cycles.

3.3.4 Statistical analysis

Relative breast displacement, velocity, and acceleration data over the five kilometre and two minute treadmill runs, in the three support conditions were checked for normality using the Kolmogorov-Smirnov and Shapiro-Wilk tests of normality, with normality assumed when $p > .05$. Data was accepted as normally distributed and therefore parametric analyses were performed.

Two-way repeated measures ANOVAs were performed to assess any differences in relative breast kinematics between the three support conditions, and across the six intervals of the five kilometre run, with an alpha level set at $p < .05$. *Post hoc* pairwise comparisons with Bonferroni adjustment were performed following the two-way repeated measures ANOVAs. Effect size and observed power were calculated to characterise the strength of all results, where a small effect ≤ .10, medium effect ≤ .30, large effect ≤ .50, and a high power ≥ .80 (Field, 2009).

3.4 Results

Multiplanar breast displacement was greatest when participants ran without breast support during two minutes of running (Table 1). Statistical analysis demonstrated a significant reduction in the magnitude of anteroposterior ($F_{(1)} = 32.413$, $p = .000$, $\eta^2 = .802$, $1\text{-}\beta = .998$), mediolateral ($\mathrm{F}_{(1)} = 6.171$, $p = .038$, $\eta^2 = .535$, $1\text{-}\beta = .588$), and vertical ($F_{(5)} = 8.568$, $p = .000$, $\eta^2 = .463$, $1\text{-}\beta = .996$) breast displacement when participants ran in the high breast support condition during two minutes of running.

Across the kilometre intervals of the five kilometre run, the anteroposterior ($F_{(1)} = 32.413$, $p = .000$, $\eta^2 = .802$, $1\text{-}\beta = .998$) and vertical ($F(1) = 44.292$, $p = .000$, $\eta^2 = .847$, $1\text{-}\beta = 1.000$) breast displacement were significantly different between the low and high breast support conditions, with the high breast support significantly reducing the magnitude of breast displacement compared to the low breast support.

Significant increases in vertical breast displacement were reported in the low ($F_{(5)} = 6.905$, $p = .000$, $\eta^2 = .682$, $1\text{-}\beta = 1.000$), and high ($\mathrm{F}(3.498) = 12.099$, $p = .000$ $\eta^2 = .602$, $1\text{-}\beta = 1.000$) breast support conditions, from the first two minutes to the third, fourth, and fifth kilometre intervals.

Table 1. Mean (± SD) anteroposterior, mediolateral, and vertical breast displacement (mm) in three breast supports, during six intervals across the five kilometre run (n = 9).

INTERVAL	ANTEROPOSTERIOR			MEDIOLATERAL			VERTICAL		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MIN	44 ± 14[*ab]	33 ± 8*[a]	30 ± 6[*b]	35 ± 13[*ab]	22 ± 7[*a]	19 ± 8[*b]	57 ± 17[*ab]	40 ± 12[*ac]	25 ± 8[*bc]
1 KM		36 ± 6[*c]	29 ± 5[*c]		25 ± 8	19 ± 7		44 ± 13[*c]	26 ± 8[*c]
2 KM		35 ± 7[*c]	30 ± 6[*c]		25 ± 9	21 ± 7		46 ± 16[*c]	28 ± 8[*c]
3 KM		37 ± 9[*c]	31 ± 8[*c]		26 ± 11	22 ± 8		46 ± 15 [*c†]	29 ± 9[*c†]
4 KM		39 ± 12[*c]	30 ± 9[*c]		27 ± 12	21 ± 8		48 ± 17[*c†]	29 ± 10[*c†]
5 KM		39 ± 12[*c]	33 ± 8[*c]		27 ± 12	22 ± 8		46 ± 15[*c†]	28 ± 8[*c†]
MEAN	44 ± 14	37 ± 9	31 ± 7	35 ± 13	25 ± 10	21 ± 8	57 ± 17	45 ± 15	28 ± 10

*[a] Denotes a significant difference between BB and low breast support conditions, $p < .05$
*[b] Denotes a significant difference between BB and high breast support conditions, p < .05
*[c] Denotes a significant difference between low and high breast support conditions, $p < .05$
†Denotes a significant difference between the first two minutes and kilometre distance intervals, within a support condition, $p < .05$

Table 2. Mean (± SD) anteroposterior, mediolateral, and vertical breast velocity ($m.s^{-1}$) in three breast supports, during six intervals across the five kilometre run (n = 9).

INTERVAL	ANTEROPOSTERIOR			MEDIOLATERAL			VERTICAL		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MIN	$0.7 \pm 0.3^{*ab}$	$0.5 \pm 0.2^{*ac}$	$0.3 \pm 0.1^{*bc}$	$0.8 \pm 0.3^{*ab}$	$0.5 \pm 0.2^{*bc}$	$0.4 \pm 0.1^{*bc}$	$1.3 \pm 0.3^{*ab}$	$0.9 \pm 0.3^{*ac}$	$0.5 \pm 0.1^{*bc}$
1 KM		$0.5 \pm 0.2^{*c}$	$0.3 \pm 0.1^{*c}$		$0.5 \pm 0.2^{*c}$	$0.4 \pm 0.2^{*c}$		$1.0 \pm 0.4^{*c}$	$0.6 \pm 0.1^{*c}$
2 KM		$0.6 \pm 0.3^{*c\dagger}$	$0.3 \pm 0.1^{*c}$		$0.6 \pm 0.2^{*c}$	$0.5 \pm 0.1^{*c}$		$1.0 \pm 0.4^{*c}$	$0.6 \pm 0.2^{*c}$
3 KM		$0.6 \pm 0.2^{*c\dagger}$	$0.4 \pm 0.2^{*c\dagger}$		$0.6 \pm 0.2^{*c\dagger}$	$0.5 \pm 0.2^{*c\dagger}$		$1.1 \pm 0.4^{*c}$	$0.6 \pm 0.2^{*c\dagger}$
4 KM		$0.6 \pm 0.3^{*c\dagger}$	$0.4 \pm 0.1^{*c\dagger}$		$0.7 \pm 0.3^{*c\dagger}$	$0.5 \pm 0.2^{*c\dagger}$		$1.1 \pm 0.4^{*c}$	$0.6 \pm 0.2^{*c\dagger}$
5 KM		$0.6 \pm 0.3^{*c\dagger}$	$0.4 \pm 0.1^{*c\dagger}$		$0.7 \pm 0.3^{*c\dagger}$	$0.5 \pm 0.2^{*c\dagger}$		$1.0 \pm 0.4^{*c}$	$0.7 \pm 0.2^{*c\dagger}$
MEAN	0.7 ± 0.3	0.6 ± 0.3	0.4 ± 0.1	0.8 ± 0.3	0.6 ± 0.2	0.5 ± 0.2	1.3 ± 0.3	1.0 ± 0.4	0.6 ± 0.2

^{*a} Denotes a significant difference between BB and low breast support conditions, $p < .05$
^{*b} Denotes a significant difference between BB and high breast support conditions, $p < .05$
^{*c} Denotes a significant difference between low and high breast support conditions, $p < .05$
†Denotes a significant difference between the first two minutes and kilometre distance intervals, within a support condition, $p < .05$

Table 3. Mean (± SD) anteroposterior, mediolateral, and vertical breast acceleration (m.s^{-2}) in three breast supports, during six intervals across the five kilometre run (n = 9).

INTERVAL	ANTEROPOSTERIOR			MEDIOLATERAL			VERTICAL		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MIN	42.4 ± 26.1^{*ab}	23.5 ± 14.7^{*ac}	15.7 ± 4.9^{*bc}	37.0 ± 18.0^{*ab}	26.5 ± 8.8^{*ac}	22.6 ± 7.8^{*bc}	50.8 ± 17.1^{*ab}	35.3 ± 12.8^{*ac}	22.6 ± 7.8^{*ab}
1 KM		26.5 ± 16.7^{*c}	15.7 ± 2.9^{*c}		28.4 ± 9.8^{*c}	23.5 ± 6.9^{*c}		41.2 ± 15.7^{*c†}	22.6 ± 5.9^{*c}
2 KM		30.4 ± 16.7^{*c}	18.6 ± 5.9^{*c†}		31.4 ± 10.8^{*c}	27.5 ± 6.9^{*c}		40.2 ± 16.7^{*c}	25.5 ± 6.9^{*c}
3 KM		30.4 ± 17.7^{*c†}	22.6 ± 10.8^{*c†}		34.3 ± 12.8^{*c}	27.5 ± 8.8^{*c†}		41.2 ± 15.7^{*c†}	25.5 ± 6.3^{*c}
4 KM		31.4 ± 19.6^{*c†}	17.7 ± 2.9^{*c†}		36.3 ± 14.7^{*c†}	28.4 ± 10.8^{*c†}		43.2 ± 16.7^{*c†}	25.5 ± 5.9^{*c}
5 KM		29.4 ± 19.6^{*c†}	19.6 ± 6.9^{*c†}		37.3 ± 14.7^{*c†}	29.4 ± 10.8^{*c†}		40.2 ± 15.7^{*c†}	26.5 ± 6.9^{*c†}
MEAN	42.4 ± 26	28.6 ± 17.6	18.3 ± 5.7	37.0 ± 18.0	32.3 ± 11.9	26.5 ± 8.6	50.8 ± 17.1	40.2 ± 15.5	21.3 ± 6.6

$*^{a}$ Denotes a significant difference between BB and low breast support conditions, $p < .05$
$*^{b}$ Denotes a significant difference between BB and high breast support conditions, p < .05
$*^{c}$ Denotes a significant difference between low and high breast support conditions, $p < .05$
†Denotes a significant difference between the first two minutes and kilometre distance intervals, within a support condition, $p < .05$

Multiplanar breast velocity was greatest when participants ran without breast support during two minutes of running (Table 2). Statistical analysis demonstrated a significant main effect of breast support for anteroposterior ($F_{(5)}$ = 64.039, p = .000, η^2 = .598, 1-β = 1.000), mediolateral ($F_{(5)}$ = 64.458, p = .000, η^2 = .678, 1-β = 1.000), and vertical ($F_{(1)}$ = 21.874, p = .002, η^2 = .732, 1-β = .982) breast velocity, with significant reductions in multiplanar breast velocity when participants ran in the high breast support condition compared to the low and barebreasted conditions.

Significant increases were reported in the anteroposterior ($F_{(5)}$ = 17.146, p = .004, η^2 = .528, 1-β = .912) and mediolateral ($F_{(5)}$ = 11.567, p = .000, η^2 = .591, 1-β = 1.000) breast velocity when participants ran in the low breast support condition, consistently from the first two minutes to the third, fourth, and fifth kilometre intervals. However, no differences were reported in the vertical breast velocity in the low support condition ($F_{(2.606)}$ =2.798, p = .072, η^2 = .259, 1-β = .552). When participants ran in the high breast support significant increases were reported in the anteroposterior ($F_{(5)}$ = 11.173, p = .000, η^2 = .583, 1-β = 1.000), mediolateral ($F_{(5)}$ = 18.592, p = .000, η^2 = .699, 1-β = 1.000), and vertical ($F_{(5)}$ = 10.920, p = .000, η^2 = .577, 1-β = 1.000) breast velocity over the intervals of the five kilometre run.

Multiplanar breast acceleration was greatest in the barebreasted condition, with significant reductions reported in the magnitude of mulitplanar breast acceleration when participants ran in the low and high breast support conditions (Table 3). During the five kilometre run, the high breast support significantly reduced the magnitude of anteroposterior ($F_{(5)}$ = 57.646, p = .001, η^2 = .492, 1-β = .851), mediolateral ($F_{(5)}$ = 3.307, p = .004, η^2 = .532, 1-β = .851), and vertical ($F_{(1)}$ = 18.098, p = .003, η^2 = .693, 1-β = .960) breast acceleration at every interval when compared to the low breast support condition.

Across the five kilometre run the anteroposterior ($F_{(5)}$ = 23.875, p = .001, η^2 = .543, 1-β = .977), mediolateral ($F_{(2.744)}$ = 9.509, p = .000, η^2 = .543, 1-β = .987), and vertical ($F_{(5)}$ = 4.944, p = .001, η^2 = .382, 1-β = .966) breast acceleration significantly increased in the low breast support condition. Similarly, significant increases were reported in the anteroposterior ($F_{(5)}$ = 17.698, p = .003, η^2 = .654, 1-β = 1.000), mediolateral ($F_{(5)}$ = 15.632, p = .000, η^2 = .661, 1-β = 1.000), and vertical ($F_{(5)}$ = 3.369, p = .012, η^2 = .296, 1-β = .858) breast acceleration in the high breast support over the duration of the five kilometre run.

3.4.1 Power and effect size

Power and effect sizes were reported throughout this pilot study. Of the significant differences reported within this chapter, the effect sizes were defined as large effects (> .50). The associated power was reported as high power (> .80), excluding one statistical difference (.588). These values indicate the strength of the effect of the independent measures (breast support and run duration) on the dependent measures (multiplanar breast kinematics). Both statistics rely upon the sample size and the variance in the data. With large effect sizes and high power associated with the significant differences reported within this chapter, it is assumed that the sample size (n = 9) employed was large enough to determine the effect of breast support on multiplanar breast kinematics during a five kilometre run. It is suggested that future work within breast biomechanics employ sample sizes of nine or more to ensure statistical significance with high power and effect sizes.

3.5 Discussion

There were two aims of the current pilot study, firstly, to determine an appropriate run distance to define a prolonged treadmill run for the current programme of work, and secondly, to determine if breast kinematics change over a prolonged treadmill run. The five kilometre run distance ensured that the participants were running for longer duration than previously examined within breast biomechanics literature, and at a distance that fell in line with the government guidelines for exercise prescription. The magnitude of multiplanar breast kinematics increased over the five kilometre run with increases reported as soon as the first kilometre interval. These data demonstrate the firstly breast kinematics do increase over a prolonged run in both low and high breast support conditions, and secondly, the recommendations for sports bra design and reporting of breast biomechanics previously presented within the literature cannot be extended to prolonged running.

On average participants ran the five kilometre run at 9.5 km.hr^{-1}. When considering the time taken to run the first kilometre at this speed (6.3 minutes), comparisons of breast kinematics can be made with previous breast biomechanics publications. Until now, previous literature had examined breast kinematics during constant treadmill speeds at two, five, and seven minutes (Scurr et al., 2009, 2010; Mason et al., 1999; Boschma, 1995); however these data were most commonly collected at the end of these time points, with only one publication measuring preceding intervals (Bowles & Steele, 2003). Collecting and reporting breast kinematics at intervals during a constant speed run may develop the understanding of breast kinematics during constant prolonged running. Within the current study the results demonstrate a significant increase in multiplanar breast displacement, velocity, and acceleration as the runner's progress through the five kilometre run. A vast majority of the significant increases in breast kinematics were reported at the third kilometre interval, when compared to the first two minutes of running. When considering the range of treadmill speeds (9 to 10 km.hr^{-1}) performed in the current study, these increases were occurring from 18 to 20 minutes of running. Vertical breast acceleration was the only kinematic variable to increase as early as the first kilometre of running (on average 6.3 minutes of running) in the low breast support condition only.

Based upon the results of the current study, recommendations of product design and the quantification of sports bra performance (Starr et al., 2005; Scurr et al., 2010) previously presented in the literature cannot be extended to common running distances such as a five kilometre run. It is imperative to implement a protocol that represents the external environment as closely as possible to increase the validity of the research conducted and to progress methodologies employed within this research area. It is suggested that protocols designed to quantify the performance of a sports bra should incorporate a longer duration run, and based upon the results of the current study, it is suggested that participants should run for a minimum of 20 minutes. It should be noted that the majority of breast kinematics were at the greatest magnitude at the fifth kilometre interval, and it is hypothesised that breast kinematics may continue to increase over a run exceeding this distance (e.g. 10 km). The cause for the increase in breast kinematics is currently unknown, however the forth chapter of this thesis will explore this further.

This programme of work will implement a five kilometre treadmill run to define a prolonged run; this distance enabled the identification of changes in breast kinematics previously not reported, and represents an activity recommended by the government for exercise prescription. The reported increase in breast kinematics over this exercise duration may influence the magnitude of breast discomfort or pain experienced, and/or impact on running biomechanics, specifically in the upper body. The subsequent chapters of this thesis aim to investigate breast and body biomechanics in different breast support conditions during a five kilometre treadmill run.

CHAPTER FOUR: SECTION ONE.

THE EFFECT OF BREAST SUPPORT ON MULTIPLANAR BREAST KINEMATICS DURING A FIVE KILOMETRE RUN

4.1 Introduction

Regardless of the recommendation to wear a sports bra when exercising (Mason et al., 1999; McGhee et al., 2007; McGhee & Steele, 2010; McGhee et al., 2012; Scurr et al., 2009ab; 2010ab; 2011), it has been reported that 60% of females taking part in exercise (e.g. running) do not wear a sports bra (Bowles et al., 2008). The impact of this on breast biomechanics has been emphasised within the previous literature, with crop top support bras and everyday fashion bras shown to elevate the magnitude of relative breast kinematics (Mason et al., 1999; Scurr et al., 2011) and exercise-related breast pain (Scurr et al., 2010a) when compared to high breast supports such as sports bras. However, the implications of wearing different breast supports on biomechanical parameters of running are yet to be explored in depth. Before attempts can be made to investigate this question, it is important to understand how breast biomechanics are influenced by different breast supports over exercise durations commonly performed by female athletes, and whether potential changes in breast biomechanics could impact upon running biomechanics.

Three breast support conditions have commonly been employed to represent different magnitudes of breast support; a bare-breasted trial, an everyday bra to represent a low breast support condition, and a sports bra to represent a high breast support condition (Scurr et al., 2010a; 2011; White et al., 2009). Though the findings of these papers helped progress the knowledge in this area, promoted the use of a high breast support during exercise due to substantial reductions in breast kinematics (Scurr et al., 2010a; 2011), and emphasised potential benefits to running performance (White et al., 2009), the conclusions are restricted to short running bouts. In order to progress this area, breast biomechanics in different breast supports should be investigated over common running distances. Understanding the direction, magnitude, and trajectories of breast kinematics in different breast supports over longer running distances may provide vital information for the progression of sports bra design for distance running. Ensuring negative factors associated with poor breast support such as; increased magnitudes of breast kinematics, exercise-related breast pain, and embarrassment, are reduced.

Based upon the findings of Bowles and Steele (2003), who reported significant increases in breast displacement at the end of a five minute run when females wore poor breast support, and the pilot data reported in chapter three, it is hypothesised that breast kinematics will continue to increase over the duration of an extended run. Bowles and Steele (2003) suggested that damage to the internal tissues of the breast may be occurring due to the repeated loading on these delicate structures. The implications of this work are wide reaching; firstly, these findings may have implications for running biomechanics. An individual may alter their upper and lower body kinematics in an attempt to reduce the independent movement of the breast; for example, a restricted ROM of the thorax segment may achieve this. Secondly, this work has implications for conclusions based upon previous publications within breast biomechanics. For example, previous publications have provided recommendations for sports bra design based upon data collected over short running durations (two to five minutes). The work of Bowles and Steele (2003) established increases in breast kinematics during a five minute treadmill run, therefore, recommendations for sports bras designed for exercise exceeding five minutes in duration should not be based upon these available data. Furthermore, these findings have implications for fundamental breast biomechanics research. If breast kinematics continues to increase over a prolonged run, the risk of stress and strain on the breast tissues would be increased, which may lead to greater discomfort and pain experienced.

Mason et al., (1999) hypothesised that exercise-related breast pain arises due to the tension placed on the skin and fascia, as the breasts move relative to the thorax. This hypothesis is supported by Gerard (1960) who states that the stretching of almost any tissue that resists stretching can produce pain. Therefore, it is suggested that when tension is placed upon the skin and the internal tissues of the breast during exercise, the pain receptors associated with these tissues are stimulated and pain is experienced. Currently, breast displacement and velocity have demonstrated the strongest relationship to pain (McGhee et al., 2007; Scurr et al., 2010a). However, in line with Newton's second law of motion, (F=ma) a force is created by a change in the acceleration of an object, and is frequently related to stress and strain of an object. The force subjected to the breast tissues is therefore dependent upon its mass and acceleration. As the mass of the breast is assumed to be constant during movement, the acceleration will be the determining factor influencing the magnitude of force, and potentially influencing the pain experienced. Reporting exercise-related breast pain and general comfort of the female runner helps to inform the effect of different breast supports on running from a psychological perspective. If a female is experiencing breast pain when running due to large magnitudes of breast kinematics, it is hypothesised that running biomechanics may be influenced; alterations to running biomechanics may be adopted in an attempt to reduce the magnitude of relative breast kinematics.

In order to understand the effect of different breast supports on running biomechanics, breast biomechanics research should first establish the effect of different breast supports on breast biomechanics during running distances commonly performed by females, such as a five kilometre run. If breast biomechanics differ within and between breast support conditions over a prolonged run, this may significantly affect the biomechanics of a female runner. Importantly, examination of breast biomechanics over a common running distance would increase the external validity of this research and possibly widen the application of the findings.

4.2 Aims and research hypotheses

The aim of this study was to investigate the effect of breast support on breast biomechanics during a five kilometre run.

H_1 Increasing the breast support will significantly reduce the magnitude of multiplanar breast biomechanics across the five kilometre run.

H_2 The magnitude of multiplanar breast biomechanics will significantly increase over the duration of the five kilometre run.

H_3 Multiplanar breast acceleration will demonstrate significant positive correlations to self-reported exercise-related breast pain.

4.3 Methods

Following University of Portsmouth, ethics approval (Science Faculty Ethics Committee), ten regularly exercising female volunteers, (experienced treadmill and outdoor runners currently training ≥ 30 min, ≥ five times per week), participated in this study. In an attempt to reduce the magnitude of between participant variance, participants had not had any children and not experienced any surgical procedures to the breast. Prior to the practical laboratory sessions, participants' bra size was measured by a trained bra fitter employing the best fit criteria recommended by White and Scurr (2012). Participants were required to fit either of the two cross-graded bra sizes of 34D and 32DD. These two sizes were selected for comparisons with previous research (Gehlsen & Albohm, 1980; Lorentzen & Lawson, 1987; Verscheure, 2000; White et al., 2009; Scurr et al., 2009a;

2010a; 2011), and a 34 D bra size sits within the cross-grading range of the UK average (36C) (Treleaven, 2007).

4.4.1 Participants

Participants had a mean (± SD) age of 23 years (± 2 years), body mass 62.1 kg (± 5.4 kg), and height 1.60 m (± 0.05 m). All participants provided written informed consent to participate in this study. Blood pressure was taken using a portable electronic sphygmomanometer (HEM-705C, Omron, Netherlands). Blood pressure values which fell between 150 to 90 mmHg (systolic pressure) and 90 to 60 mmHg (diastolic pressure) were deemed as acceptable.

4.4.2 Procedures

Participants performed two five kilometre treadmill run trials on separate days. To ensure participants time in the menstrual cycle did not vary substantially (within the luteal phase, days 5 to 15 of the 28 day menstrual cycle) these two laboratory sessions were carried out from 24 to 72 hours apart; once in a 'low' breast support (everyday t-shirt bra) and once in a 'high' breast support (B4490 sports bra) (Figure 5).

Participants were required to perform an additional bare-breasted (BB) treadmill run, but due to the discomfort associated with this condition, participants ran without breast support for only two minutes (Scurr et al., 2009; 2010a; McGhee et al., 2012). A random number generator (http://www.random.org/) was used to calculate the order for the support conditions for each participant to ensure order effects were reduced.

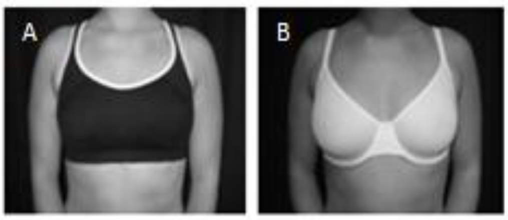

Figure 5. (A) High support condition sports bra: B4490, Shock Absorber level 4 support, made from 57% polyester, 34% polyamide, and 9% elastane. (B) Low support conditions everyday bra: Marks and Spencer Seamfree Plain Under wired T-Shirt Bra, non-padded, made from 88% polyamide and 22% elastane lycra.

The participants selected a comfortable running speed, which they felt they could maintain for the duration of the run, this ranged from 8.5 km·h^{-1} to 10.5 km·h^{-1}, with an average of 9 ± 1 km·h^{-1}. The treadmill was level (0% gradient), with no incline. Once selected, this speed remained constant throughout all run trials, for that participant. This meant that the distance covered in two minutes and the final five kilometre completion time varied between the ten participants. These data are shown when averaged over the ten participants in Table 4.

Table 4. The average treadmill speed, time taken to complete the five kilometre run, and the distance covered within the first two minutes of running averaged for the ten participants.

	Treadmill speed	Time taken to run 5 km	Distance covered in two minutes
Mean	9 km.h^{-1}	32 minutes	322 m
SD	1 km.h^{-1}	4 minutes	64 m

In order to carry out comparisons between breast support conditions, participants performed the bare-breasted run at the same speed as the two five kilometre run trials. Participants wore the same footwear and lower body clothing for all trials.

Five retro-reflective semi-spherical markers (diameter of 12 mm) were positioned with hyper-allergenic tape on the following anatomical landmarks; the suprasternal notch, the left and right anterioinferior aspect of the 10th ribs, the right nipple (Figure 5) (Scurr et al., 2010a; 2011), and one positioned on the lateral aspect of the left heel to identify gait cycles (Scurr et al., 2009a; 2010a; 2011). The nipple marker is assumed to represent the gross movement of the breast, and the resulting kinematics of this marker will be referred to as breast kinematics from here onwards. During the two bra conditions, participants repositioned the markers on the bra, directly over the nipple using visual inspection (White et al., 2009; Scurr et al., 2010a).

Participants were asked to verbally rate their exercise-related breast pain at the end of two minutes of running in all three support conditions, and at the end of the five kilometre run in the low and high breast supports, using an adapted version of the numerical analogue scale presented by Mason et al., (1999). The eleven point scale used in the current study defined zero as ‘no pain’, five as ‘moderate pain’ and ten as ‘excruciating pain’ (Appendix A). These adaptations ensured the participants had only descriptors related to magnitude of pain, whereas the previous scale included the descriptor ‘uncomfortable’ as five on the scale. The scale presented by Mason et al., (1999) therefore includes an additional measure of comfort. Participants walked at a self-selected speed for up to five minutes to cool down after the five kilometre run.

Three-dimensional coordinates of the five markers were tracked by eight 200 Hz calibrated Oqus infrared cameras (Qualisys, Sweden), operating with tracking parameters of 0.25 mm. The eight cameras positioned in an arc around the treadmill, and in the centre of the laboratory to maximise the field of view of all cameras. Cameras recorded for the final ten seconds of the initial two minutes of the five kilometre runs, and for ten seconds within the final 100 m of each kilometre interval following this (e.g. 900 m, 1900 m, etc).

4.4.3 Data Processing

Markers were identified and 3D data reconstructed in the Qualisys Track Manager (QTM) Software (Qualisys, Sweden). Untransformed 3D coordinate data were exported from QTM as a TSV file and imported to a frequency analysis program in MATLAB (MathWorks, UK). The frequency content of the data was assessed using a Fast Fourier transformation (FFT) in MATLAB. The FFT showed the amplitude of the data point plotted against the frequency component, enabling the identification of data that should be retained and the noise component that is attenuated (Winter, 1990). A cut-off frequency of 13 Hz was selected for the low pass filter based upon this process. The global coordinate system (GCS) identified x' as the line of progression on the treadmill (anteroposterior), y' as mediolateral, and z' as vertical (Figure 6).

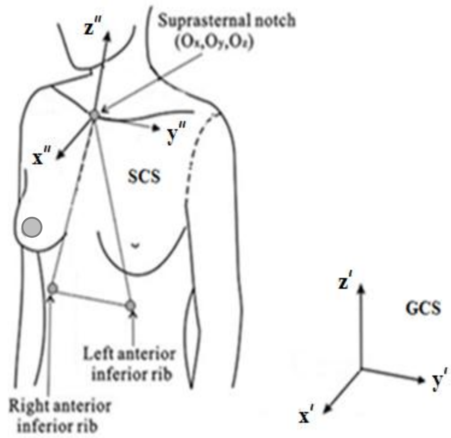

Figure 6. Marker locations, axes and coordinate systems for the global coordinate system (GCS) (x', y', z') and segment coordinate system (SCS) (x'', y'', z'').

Three-dimensional coordinates for the markers on the thorax, nipple, and heel were exported to Visual3D (c-motion, Inc) as a C3D file from QTM, and filtered at 13 Hz using a fourth-order zero-phase shift low pass Butterworth filter. The fourth order zero-phase shift eliminates the noise component of the signal with a sharp cut off, due to the two stage filtering processes (forward and reverse), creating a filtered signal that is in-phase with the raw data (Winter, 1990).

To determine running gait cycles, instantaneous velocity of the heel was derived from the anteroposterior coordinates. Heel strike[1] for each running gait cycle was identified as the velocity of the heel marker changed from positive to negative and then back to positive (Zeni, Richards, & Higginson, 2008), with a full gait cycle taken from heel strike to heel strike of the ipsilateral heel. To establish relative breast kinematics, independent to the 6 *dof* movement of the thorax, an orthogonal segment coordinate system (SCS) converted absolute coordinates of the breast to relative coordinates using a transformation matrix within Visual3D. The three non-collinear markers positioned on the thorax were used to define the SCS, with the anteroinferior ribs identified as the medial and lateral locations of the distal end of the segment and the suprasternal notch as the proximal end. A virtual mid-point was established between the medial and lateral points of the distal end (ribs) which extended to the suprasternal notch (proximal end and origin of SCS) creating the vertical axis (z"). The reference frontal plane (y"-z") was then defined using the three markers, with vector y" perpendicular to the z axis. Vector x" was directed anterior to this plane, and using the right hand rule was perpendicular to z" and y". Using these relative breast coordinates, minima positional coordinates were subtracted from maxima coordinates of the right nipple (Scurr et al., 2010a; 2011) to calculate breast displacement in each plane, normalised to the percentage of each gait cycle (n = 5) (Figure 6a), at each interval of the five kilometre run.

Percentage distribution of the breast displacement relative to the thorax were calculated in each direction (anteroposterior, mediolateral, and vertical), to illustrate the percentage breakdown of multiplanar breast displacement. Breast movement trajectories were calculated relative to the thorax to show the relative movement of breast, in the three levels of breast support, during the two minute treadmill runs. The 2D coordinates (mm) of the breast were graphically presented within the three planes of movement; frontal (y"-z" coordinates), sagittal (x"-z" coordinates) and transverse (x"-y" coordinates) relative to the thorax. First (velocity, $m \cdot s^{-1}$) and second (acceleration, $m \cdot s^{-2}$) (Figure 7b and c) derivatives of 3D relative breast coordinates were calculated for each sample, with the peak value recorded for each of these variables. Employing Newton's second law of motion, $F = m*a$, where F = force (N), m = mass (kg), and a = acceleration ($m.s^{-2}$), an approximation of the force measured at the breast was calculated. Breast mass was estimated at 0.52 kg for the bra sizes in this study (32 DD and 34 D), using estimates from Turner and Dujon (2005).

[1] The author is aware that heel strike may differ between participants dependent upon their footfall pattern.

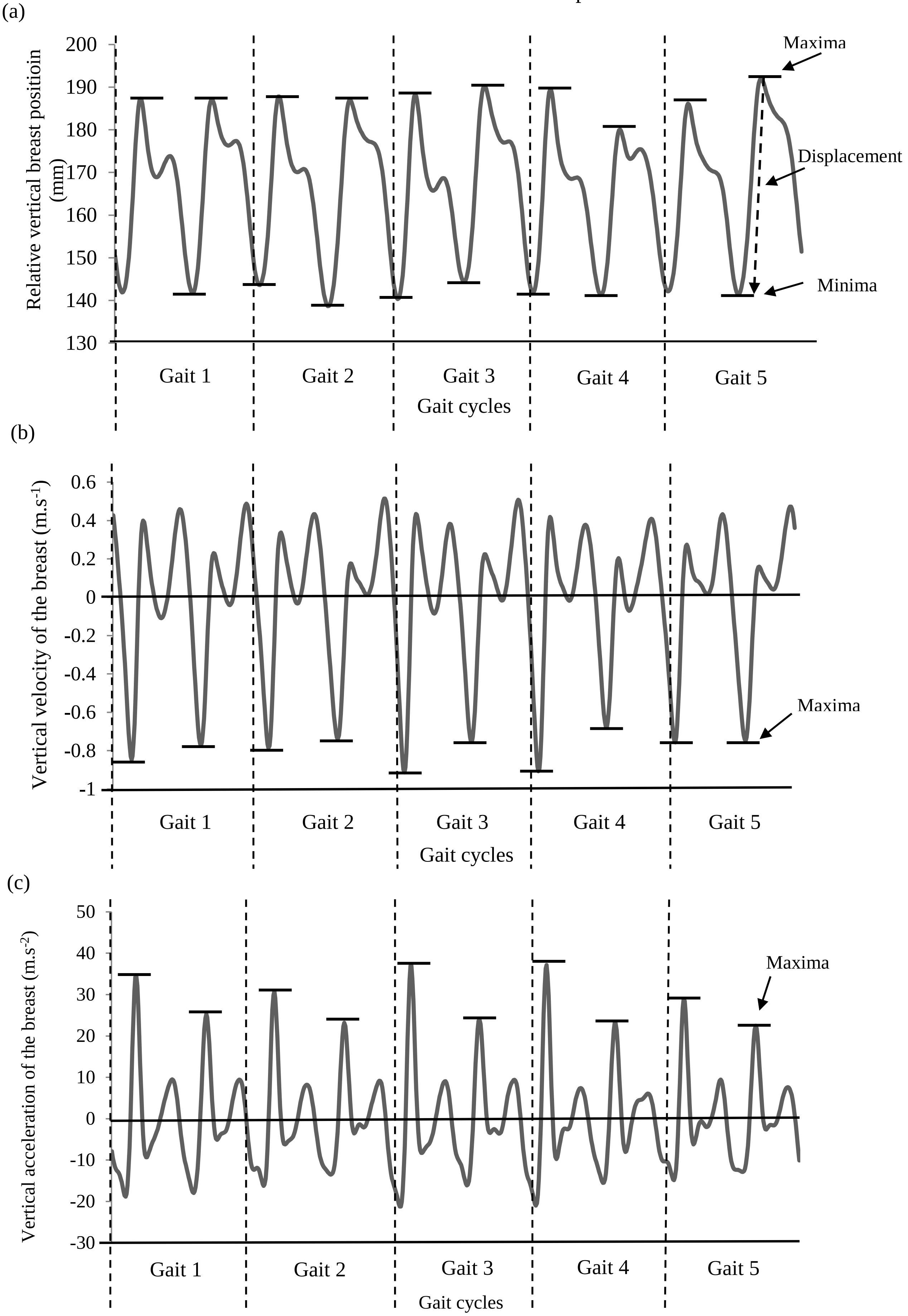

Figure 7. Example of relative vertical breast position (n = 1) (a), breast velocity ($m.s^{-1}$) (b), and acceleration ($m.s^{-2}$) (c) over five gait cycles, with maxima and minima (displacement) and peak values (velocity and acceleration) identified for each gait cycle.

4.4.4 Statistical Analyses

Relative breast displacement, velocity, acceleration, and force data over the five kilometre and two minute treadmill runs, in the three support conditions were checked for normality using the Kolmogorov-Smirnov and Shapiro-Wilk tests of normality, with normality assumed when $p > .05$. Homogeneity of variance was assessed using Mauchly's test of Sphericity, with homogenous data assumed when $p > .05$. Data were then accepted as normally distributed and displaying homogeneity and therefore defined as parametric.

One-way repeated measures ANOVAs were performed to assess the differences in relative breast kinematics and approximated force, for each plane of movement (anteroposterior, mediolateral and vertical), between the three breast support conditions (bare-breasted, low and high), for the two minute data sets. Separate two-way repeated measures ANOVAs were performed to assess any differences in relative breast kinematics and approximated force, for each plane of movement, between the two breast support conditions (high and low support); across the six intervals of the five kilometre run (two minutes, and the first to the fifth kilometre). The alpha level was set at $p < .05$. *Post hoc* pairwise comparisons with Bonferroni adjustment were performed following the two-way repeated measures ANOVAs.

Non-parametric Friedman test of difference was employed to assess the differences in exercise-related breast pain between the three breast supports. *Post hoc* Wilcoxon comparisons were employed to determine where differences lay. Non-parametric Spearman's Rho correlations were performed to assess the relationship between exercise-related breast pain and multiplanar breast kinematic and approximated force data, with a small relationship defined as $\pm \leq .10$, medium relationship as $\pm \leq .30$, and large relationship as $\pm \geq .50$ (Field, 2009). Effect size (η^2) and observed power (1-β) were calculated to characterize the strength of all results, where a small effect $\leq .10$, medium effect $\leq .30$, large effect $\leq .50$, and a high power = $\geq .80$ (Field, 2009).

4.4 Results

4.4.1 Breast movement trajectories relative to the thorax

Examples of breast movement trajectories relative to the thorax (with the suprasternal notch as the origin = 0) are presented graphically for each plane of movement (frontal (x''-y''), sagittal (y''-z''), and transverse (x''-z'')) during the initial two minute run and the fifth kilometre of the five kilometre run (Figure 8 a, b, c), in the three breast support conditions, over an average of five gait cycles, averaged participants (n = 10).

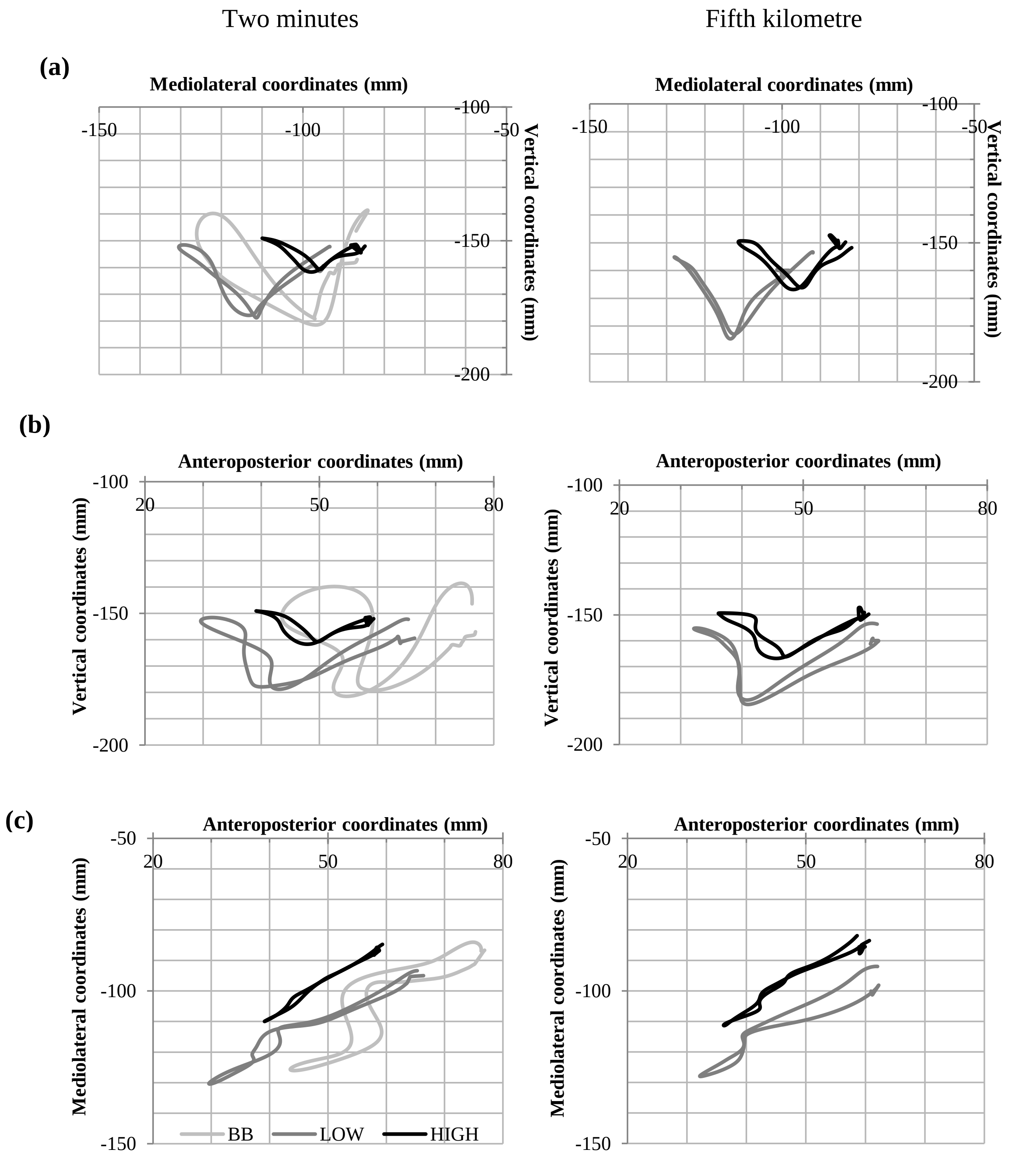

Figure 8. Breast movement trajectories relative to the thorax in the (a) frontal (b) sagittal and (c) transverse plane, in the different breast supports, averaged over five gait cycles, at the end of two minutes and the fifth kilometre of a five kilometre run (n = 10).

During the first two minutes of running, breast support is shown to alter the breast trajectory (Figure 8), with the high support reducing the more erratic pattern seen in the bare-breasted running. Within the frontal plane, the breast is brought medially in the high breast support and slightly lifted, compared to the lower breast support conditions. A considerable reduction in the breast trajectory in the sagittal plane can be seen in the high breast support, with the anteroposterior position and trajectory path reduced. As breast support is increased, the trajectory of the breast in the transverse plane is reduced, with the high breast support reducing the range of movement when compared to the low and bare-breasted support conditions.

At the five kilometre distance interval the trajectory of the breast in the low and high breast support appear to demonstrate similar shapes in each plane of movement, however, the magnitude of the trajectories are reduced in the high breast support. When comparing the breast trajectories in the low and high support between the two minute date collection and the fifth kilometre, small changes in shape and magnitude can be seen (Figure 8). At the fifth kilometre the prominent ‘V’ shape trajectory within the frontal plane, in the low and high support, appear slightly different when compared to the two minute data, noticeably when examining the vertical trajectory path of the breast. Similarly, differences can be seen between these two data collections when examining the breast trajectories in the sagittal and transverse planes, with a less erratic path presented.

4.4.2 Relative multiplanar breast displacement (mm)

The magnitude of breast support demonstrated a significant main effect on mulitplanar breast displacement during the two minute and five kilometre treadmill runs (Table 5). The greatest reduction in multiplanar breast displacement was reported in the high breast support compared to the bare-breasted condition during the first two minutes, significantly reducing the magnitude of anteroposterior, mediolateral and vertical breast displacement by 42%, 48%, and 70%, respectively.

During the five kilometre run, the high breast support provided significantly greater reductions in multiplanar breast displacement compared to the low breast support at each kilometre interval. On average, during the five kilometre run, the high level breast support further reduced the anteroposterior, mediolateral, and vertical breast displacement reported in the low breast support by 28%, 21%, and 55%, respectively.

Table 5. Mean (± SD) anteroposterior, mediolateral, and vertical breast displacement (mm) in three breast supports, during six intervals across the five kilometre run (*n* = 10).

INTERVAL	ANTEROPOSTERIOR			MEDIOLATERAL			VERTICAL		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MIN	41 ± 5*[ab]	34 ± 3*[c]	24 ± 6*[c]	50 ± 11*[ab]	36 ± 7*[c]	26 ± 5 *[c]	60 ± 13*[ab]	34 ± 9*[c] †	18 ± 4*[c]
1 KM		36 ± 3*[c]	27 ± 6*[c]†		38 ± 6*[c]	30 ± 5*[c]		38 ± 9*[c]†	21 ± 6*[c]
2 KM		38 ± 4*[c]	27 ± 6*[c]		40 ± 8*[c]	31 ± 7*[c]		39 ± 9*[c]	22 ± 5*[c]†
3 KM		38 ± 5*[c]	27 ± 5*[c]		40 ± 7*[c]	32 ± 7*[c]		40 ± 8*[c]	22 ± 5*[c]†
4 KM		37 ± 5*[c]	27 ± 6*[c]		40 ± 8*[c]	32 ± 6*[c]†		40 ± 8*[c]†	22 ± 5*[c]†
5 KM		37 ± 6*[c]	28 ± 6*[c]†		40 ± 9*[c]	32 ± 5*[c]†		41 ± 8*[c]†	23 ± 6*[c]†
MEAN	41 ± 5	36 ± 2	26 ± 1	50 ± 11	39 ± 2	31 ± 2	60 ± 13	38 ± 3	21 ± 2

*[a] Denotes a significant difference between BB and low breast support conditions, *p* < .05
*[b] Denotes a significant difference between BB and high breast support conditions, p < .05
*[c] Denotes a significant difference between low and high breast support conditions, *p* < .05
†Denotes a significant difference between the first two minutes and kilometre distance intervals, within a support condition, *p* < .05

N.B. Significant main effect of breast support for anteroposterior ($F_{(2)} = 40.782$, $p = .001$, $\eta^2 = .819$, $1\text{-}\beta = 1.000$), mediolateral ($F_{(2)} = 40.782$, $p = .001$, $\eta^2 = .819$, $1\text{-}\beta = 1.000$), and vertical ($F_{(2)} = 69.638$, $p = .001$, $\eta^2 = .886$, $1\text{-}\beta = 1.000$) breast displacement for the two minute data.
Significant main effect of breast support for the anteroposterior ($F_{(1)} = 68.868$, $p = .001$, $\eta^2 = .884$, $1\text{-}\beta = 1.000$), mediolateral ($F_{(1)} = 66.937$, $p = .001$, $\eta^2 = .881$, $1\text{-}\beta = 1.000$), and vertical ($F_{(1)} = 83.465$, $p = .001$, $\eta^2 = .903$, $1\text{-}\beta = 1.000$) breast displacement for the five kilometre data.
Interaction effect of breast support and run interval for the anteroposterior ($F_{(5)} = 5.240$, $p = .001$, $\eta^2 = .368$, $1\text{-}\beta = .977$), mediolateral ($F_{(5)} = 6.671$, $p = .001$, $\eta^2 = .426$, $1\text{-}\beta = .995$), and vertical ($F_{(5)} = 13.140$, $p = .001$, $\eta^2 = .593$, $1\text{-}\beta = 1.000$) breast displacement for the five kilometre data.

An interaction effect was reported between breast support and distance intervals, with significant increases reported in the magnitude of relative mulitplanar breast displacement within the low and high breast support conditions from the start to the end of the five kilometre run. In the high breast support condition the anteroposterior and mediolateral breast displacement increased between the initial two minutes to the fifth kilometre, with percentage increases of 17% and 23%, respectively. Increases in the vertical breast displacement of 21% and 28% were reported from the initial two minutes to the fifth kilomtre in the low and high breast support, respectively.

4.4.3 Plane of movement distribution of breast displacement (%)

The percentage distribution of the breast displacement in each plane of movement was calculated for the three breast supports during the two minute treadmill run (Table 6). The percentage distribution was also calculated for the low and high breast supports during the fifth kilometre interval, to examine any differences in the distribution of breast displacement at the end of the five kilometre run compared to the first two minutes.

Table 6. Percentage distribution of relative multiplanar breast displacement (%), in three breast supports during treadmill running (over two minutes and five kilometre run) (n = 10).

Plane of movement	Two minutes			Fifth kilometre	
	BB	LOW	HIGH	LOW	HIGH
ANTEROPOSTERIOR	27%	33%	35%	31%	33%
MEDIOLATERAL	33%	34%	38%	34%	39%
VERTICAL	40%	33%	27%	35%	28%

N.B. BB = Bare-breasted

When running without external breast support, the greatest percentage of movement occurs in the vertical plane of movement. However, the percentage distribution of relative breast movement in each plane changes when external breast support (low and high) is worn. Within the high breast support the distribution was greatest in the mediolateral direction of movement. No differences were seen in the distribution of breast movement in the low and high breast support from two minutes to the fifth kilometre.

4.4.4 Relative multiplanar breast velocity (m.s^{-1})

The bare-breasted activity demonstrated the greatest magnitude of multiplanar breast velocity (Table 7). The greatest reductions in breast velocity were reported in the high breast support compared to the bare-breasted condition, with significant reductions in the anteroposterior, mediolateral, and vertical directions of 58%, 68%, and 75%, respectively.

During the five kilometre run the high breast support provided superior magnitudes of support compared to the low breast support, with significant reductions in mulitplanar breast velocity at each kilometre interval. On average the high breast support provided 40% more reduction in multiplanar breast velocity when compared to the low breast support, over the five kilometre run.

Increases in breast velocity were only reported in the vertical direction, from the first two minutes to the latter intervals of the five kilometre run (third, fourth, and fifth km). The greatest increase was reported between two minutes and third kilometre for the high breast support, an increase of 32%, and first two minutes and fifth kilometre for the low support, an increase of 17%.

4.4.5 Relative multiplanar breast acceleration ($m.s^{-2}$)

Breast support demonstrated a significant main effect on multiplanar breast acceleration (Table 8). The greatest magnitude of breast acceleration was reported in the bare-breasted condition and smallest in the high breast support for all directions. The high breast support reduced the magnitude of anteroposterior, mediolateral, and vertical breast acceleration by 68%, 68%, and 72%, respectively.

During the five kilometre run, the high breast support significantly reduced the magnitude of multiplanar breast acceleration compared to the low breast support. On average this reduction in anteroposterior, mediolateral, and vertical breast acceleration was 47%, 38%, and 59%, respectively.

Vertical breast acceleration in the low and high breast support condition significantly increased from two minutes to the fifth kilometre of the run, by 24% and 28%, respectively.

Table 7. Mean peak (± SD) anteroposterior, mediolateral, and vertical breast velocity ($m.s^{-1}$) in the three breast support conditions, during six intervals across the five kilometre run (n = 10).

INTERVAL	ANTEROPOSTERIOR			MEDIOLATERAL			VERTICAL		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MIN	0.43 ± 0.12*[ab]	0.28 ± 0.07*[c]	0.18 ± 0.06*[c]	0.50 ± 0.15*[ab]	0.24 ± 0.04*[c]	0.16 ± 0.04*[c]	0.88 ± 0.29*[ab]	0.47 ± 0.15*[c]	0.22 ± 0.10*[c]
1 KM		0.29 ± 0.05*[c]	0.19 ± 0.07*[c]		0.26 ± 0.04*[c]	0.19 ± 0.05*[c]		0.54 ± 0.17*[c]	0.26 ± 0.11*[c]
2 KM		0.36 ± 0.16*[c]	0.20 ± 0.06*[c]		0.27 ± 0.06*[c]	0.18 ± 0.04*[c]		0.54 ± 0.19*[c]	0.27 ± 0.10*[c]
3 KM		0.32 ± 0.08*[c]	0.21 ± 0.06*[c]		0.26 ± 0.04*[c]	0.18 ± 0.04*[c]		0.57 ± 0.1*[c]	0.29 ± 0.10*[c]†
4 KM		0.34 ± 0.09*[c]	0.19 ± 0.04*[c]		0.26 ± 0.04*[c]	0.18 ± 0.05*[c]		0.55 ± 0.18*[c]†	0.28 ± 0.11*[c]†
5 KM		0.32 ± 0.08*[c]	0.20 ± 0.06*[c]		0.26 ± 0.05*[c]	0.18 ± 0.05*[c]		0.57 ± 0.16*[c]†	0.28 ± 0.10*[c]†
MEAN	0.43 ± 0.12	0.32 ± 0.03	0.20 ± 0.01	0.50 ± 0.15	0.26 ± 0.01	0.18 ± 0.01	0.88 ± 0.29	0.54 ± 0.03	0.26 ± 0.04

*[a] Denotes a significant difference between BB and low breast support conditions, $p < .05$
*[b] Denotes a significant difference between BB and high breast support conditions, p < .05
*[c] Denotes a significant difference between low and high breast support conditions, $p < .05$
†Denotes a significant difference between the first two minutes and kilometre distance intervals, within a support condition, $p < .05$

N.B. Significant main effect of breast support for anteroposterior ($F_{(2)} = 29.592$, $p = .001$, $\eta^2 = .767$, 1-β = 1.000), mediolateral ($F_{(2)} = 35.079$, $p = .001$, $\eta^2 = .796$, 1-β = 1.000), and vertical ($F_{(2)} = 27.507$, $p = .001$, $\eta^2 = .753$, 1-β = 1.000) breast velocity for the two minute data.
Significant main effect of breast support for the anteroposterior ($F_{(1)} = 26.009$, $p = .001$, $\eta^2 = .743$, 1-β = .995), mediolateral ($F_{(1)} = 54.627$, $p = .001$, $\eta^2 = .859$, 1-β = 1.000), and vertical ($F_{(1)} = 60.252$, $p = .001$, $\eta^2 = .870$, 1-β = 1.000) breast velocity for the five kilometre data.
Interaction effect of breast support and distance intervals for vertical ($F_{(5)} = 11.074$, $p = .001$, $\eta^2 = .552$, 1-β = 1.000) breast velocity for the five kilometre data.

Table 8. Mean peak (± SD) of anteroposterior, mediolateral, and vertical breast acceleration ($m.s^{-2}$) in the three breast support conditions, during six intervals across the five kilometre run (*n* = 10).

INTERVAL	ANTEROPOSTERIOR			MEDIOLATERAL			VERTICAL		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MIN	22.0 ± 8.4*[ab]	13.9 ± 5.2*[c]	7.0 ± 3.1*[c]	15.8 ± 5.4*[ab]	7.9 ± 2.1*[c]	5.0 ± 1.3*[c]	34.1 ± 11.0*[ab]	23.2 ± 6.9*[c]	9.5 ± 5.3*[c]
1 KM		13.9 ± 5.1*[c]	7.6 ± 3.3*[c]		8.2 ± 2.1*[c]	5.4 ± 1.7*[c]		26.8 ± 7.5*[c]	11.3 ± 6.0*[c]
2 KM		16.3 ± 6.8*[c]	8.7 ± 2.9*[c]		8.5 ± 2.5*[c]	5.6 ± 1.7*[c]		27.1 ± 8.8*[c]	11.2 ± 5.0*[c]
3 KM		14.2 ± 6.2*[c]	7.6 ± 3.0*[c]		8.9 ± 2.8*[c]	5.0 ± 1.3*[c]		28.2 ± 8.3*[c]	11.5 ± 5.1*[c]
4 KM		15.4 ± 5.2*[c]	7.9 ± 2.7*[c]		8.7 ± 2.4*[c]	5.1 ± 1.8*[c]		28.4 ± 8.0*[c]	11.7 ± 2.2*[c]
5 KM		14.4 ± 5.8*[c]	7.9 ± 3.2*[c]		8.0 ± 2.0*[c]	5.2 ± 1.4*[c]		28.8 ± 6.5*[c]†	12.2 ± 4.6*[c]†
MEAN	22.0 ± 8.4	14.7 ± 1.0	7.8 ± 0.6	15.8 ± 5.4	8.4 ± 0.4	5.2 ± 0.2	34.1 ± 11.0	27.1 ± 2.1	11.2 ± 0.9

*[a] Denotes a significant difference between BB and low breast support conditions, *p* < .05
*[b] Denotes a significant difference between BB and high breast support conditions, p < .05
*[c] Denotes a significant difference between low and high breast support conditions, *p* < .05
†Denotes a significant difference between the first two minutes and kilometre distance intervals, within a support condition, *p* < .05

N.B. Significant main effect of breast support for anteroposterior ($F_{(1.227)}$ =26.728, p = .001, η^2 = .748, 1-β = .999), mediolateral ($F_{(1.357)}$ = 25.924, p = .001, η^2 = .742, 1-β = .999), and vertical ($F_{(2)}$ = 60.573, p = .001, η^2 = .871, 1-β = 1.000) breast acceleration for the two minute data. Significant main effect of breast support for the anteroposterior ($F_{(1)}$ = 19.747, p = .002, η^2 = .687, 1-β = .976), mediolateral ($F_{(1)}$ = 16.633, p = .003, η^2 = .649, 1-β = .951), and vertical ($F_{(1)}$ = 18.586, p = .001, η^2 = .923, 1-β = 1.000) breast acceleration for the five kilometre data. Interaction effect of breast support and distance intervals for vertical ($F_{(5)}$ = 5.831, p = .001, η^2 = .393, 1-β = .987) breast acceleration for the five kilometre data.

Table 9. Mean peak (± SD) anteroposterior, mediolateral, and vertical approximated breast force (N) in the three breast support conditions, during six intervals across the five kilometre run (n = 10).

INTERVAL	ANTEROPOSTERIOR			MEDIOLATERAL			VERTICAL		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MIN	12 ± 4*[ab]	7 ± 3*[c]	4 ± 2*[c]	8 ± 3*[ab]	4 ± 1*[c]	3 ± 1*[c]	18 ± 6*[ab]	12 ± 4*[c]	5 ± 3*[c]
1 KM		7 ± 3*[c]	4 ± 2*[c]		4 ± 1*[c]	3 ± 1*[c]		14 ± 4*[c]	6 ± 3*[c]
2 KM		7 ± 3*[c]	4 ± 2*[c]		5 ± 1*[c]	3 ± 1*[c]		15 ± 5 *[c]	6 ± 3*[c]
3 KM		7 ± 3*[c]	4 ± 2*[c]		5 ± 1*[c]	3 ± 1*[c]		15 ± 5*[c]	6 ± 3*[c]†
4 KM		8 ± 3*[c]	4 ± 1*[c]		5 ± 1*[c]	3 ± 1*[c]		15 ± 4*[c]†	6 ± 3*[c]
5 KM		8 ± 3*[c]	4 ± 2*[c]		4 ± 1*[c]	3 ± 1*[c]		15 ± 4*[c]†	6 ± 3*[c]
MEAN	12 ± 4	7 ± 1	4 ± 0	8 ± 3	5 ± 1	3 ± 0	18 ± 6	14 ± 1	6 ± 0

*[a] Denotes a significant difference between BB and low breast support conditions, $p < .05$
*[b] Denotes a significant difference between BB and high breast support conditions, p < .05
*[c] Denotes a significant difference between low and high breast support conditions, $p < .05$
†Denotes a significant difference between the first two minutes and kilometre distance intervals, within a support condition, $p < .05$

N.B. Significant main effect of breast support for anteroposterior ($F_{(1.229)}$ = 1.229, p = .001, η^2 = .747, 1-β = .999), mediolateral ($F_{(2)}$ = 25.880, p = .001, η^2 = .742, 1-β = 1.000), and vertical ($F_{(2)}$ = 60.487, p = .001, η^2 = .870, 1-β = 1.000) breast force for the two minute data. Significant main effect of breast support for the anteroposterior ($F_{(1)}$ = 20.711, p = .001, η^2 = .697, 1-β = .981), mediolateral ($F_{(1)}$ = 18.546, p = .002, η^2 = .673, 1-β = .968) and vertical ($F_{(1)}$ = 109.230, p = .001, η^2 = .924, 1-β = 1.000) breast force for the five kilometre data. Interaction effect of breast support and distance intervals for vertical ($F_{(5)}$ = 7.184, p = .001, η^2 = .444, 1-β = .997) breast force for the five kilometre data.

4.4.6 Approximation of force (N)

A significant main effect was reported for breast support when examining multiplanar breast force during the two minute treadmill runs, with the greatest force reported in the bare-breasted condition followed by the low breast support, and the smallest magnitude in the high breast support. When compared to the bare-breasted condition, the low breast support demonstrated a percentage reduction in anteroposterior, mediolateral and vertical breast forces of 42%, 50%, and 33%, respectively (Table 9). Whereas, the high breast support provided a superior percentage reduction of breast force in each direction, with percentage reductions of 67%, 62%, and 72%, respectively.

During the five kilometre run the high breast support continued to demonstrate the greatest percentage reduction in multiplanar breast force at each kilometre interval, with an average reduction of 47%, when compared to the low breast support. Nevertheless, within both the low and high breast support the magnitude of relative vertical breast force significantly increased from the first two minute interval to the fifth kilometre interval of the run. The greatest increase during the run was reported within the low breast support, from the first two minute to the fifth kilometre (3 N), an increase of 25%.

4.4.7 Relationship of breast kinematics and approximated force to breast pain

Exercise-related breast pain obtained at the end of the first two minutes of running in the three breast support conditions; bare-breasted (BB), low and high support, and again after the five kilometre run in the low and high breast supports are presented in Figure 9.

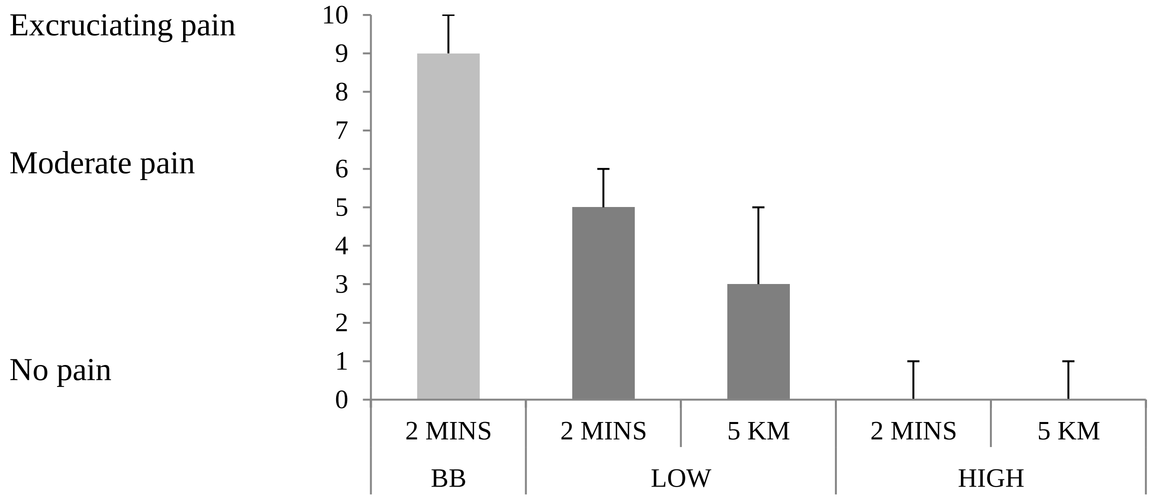

Figure 9. Mean ratings of exercise-related breast pain during the two minute and fifth kilometre interval of the five kilometre treadmill run in three breast support conditions (*n* = 10).

Exercise-related breast pain was significantly different between the three breast support conditions during the first two minutes of running (χ^2 (2) = 20.000, p = .001), with the bare-breasted support eliciting significantly greater breast pain than the low (p = .005) and high (p = .005) breast support conditions. Furthermore, the high breast support significantly reduced the exercise-related breast pain compared to the low breast support during the two minute (p = .005), and five kilometre treadmill run (p = .009). Interestingly, the participants rated their exercise-related breast pain as significantly greater in the low breast support during the first two minutes when compared to their five kilometre rating (p = .016).

Spearman's Rho correlations were carried out to examine the relationship between exercise-related breast pain and multiplanar breast displacement, velocity, acceleration and approximated force in all breast support conditions. With no differences in exercise-related breast pain between the first two minutes and fifth kilometre of the run the two minute data were employed (Table 10).

Table 10. Mean ranked Spearman's Rho correlations between exercise-related breast pain and multiplanar breast kinematics and approximated force in all breast support conditions, during five gait cycles over the first two minutes of running (n = 10).

Breast kinematic	Spearman Rho correlation coefficient (*r*)	Sig. (2 tailed) *P* value
ML acceleration	.834	.001
ML velocity	.826	.001
V velocity	.811	.001
V displacement	.788	.001
V acceleration	.781	.001
AP velocity	.748	.001
AP acceleration	.744	.001
V force	.744	.001
ML force	.731	.001
AP displacement	.716	.001
ML displacement	.707	.001
AP force	.700	.001

N.B. AP = anteroposterior, ML = mediolateral, V = vertical

Mediolateral breast acceleration was reported as having the strongest correlation to breast pain (r = .834, p = .001) (Table 10). However, all multiplanar breast kinematic variables were significantly correlated to exercise-related breast with the corresponding r values ranging from .700 to .834.

4.4.8 Effect sizes and power

Effect size and power were calculated alongside the one-way and two-way repeated measures ANOVAs carried out within this chapter. Of the differences reported within this chapter, the effect sizes were defined as large effects (> .50), with the exception of three results which fell between .30 and .50. The associated power was reported as high power (> .80). These values indicate the strength of the effect of the independent measures (breast support and run duration) on the dependent measures (multiplanar breast kinematics). Both statistics rely upon the sample size and the variance in the data. With large effect sizes and high power associated with the significant differences reported within this chapter, it is assumed that the sample size (n = 10) employed was large enough to determine the effect of breast support on multiplanar breast kinematics during a five kilometre run.

4.5 Discussion

This is the first study to examine multiplanar breast kinematics and breast force, and subjective ratings of breast pain in different breast support conditions over a five kilometre run. The key findings demonstrated that the high breast support condition provided superior support to the breast when compared with the low breast support condition over the five kilometre run distance. Significant reductions in multiplanar breast kinematics, approximated force, and exercise-related breast pain were identified when participants wore the high breast support. Furthermore, this study demonstrated significant increases in mulitplanar breast kinematics as the participants progressed through the five kilometre run in both low and high breast support conditions. These findings may have implications for sports performance, breast biomechanics methodologies, product design and product testing protocols.

Since previous publications (Scurr et al., 2009a; 2010a; 2011; White et al., 2009) reported significant differences in the magnitude of breast kinematics between low and high breast support conditions, over short running durations, it was hypothesised that the high breast support condition would significantly reduce the magnitude of multiplanar breast kinematics at all intervals of the five kilometre run when compared to the low support condition. The first key finding demonstrated that as breast support increased from low to high, multiplanar breast kinematics significantly reduced. Furthermore, the superior support provided by the high breast support condition was prevalent throughout all intervals of the five kilometre run, regardless of the significant increases in magnitude of breast kinematics within both low and high breast support conditions, therefore, hypothesis one was accepted. The greatest difference in breast kinematics between breast support conditions was consistently reported within the first two minutes, between the bare-breasted and high breast support. This result is consistent with Scurr et al., (2010a), and further promotes the use of a high breast support during prolonged running to reduce negative factors associated with the breast to maintain running performance.

Breast movement trajectories were calculated in the SCS and therefore represent the relative movement of the breast. A proposed explanation for the reported reduction in anterior position, within the high breast support condition, is due to the compressive aspect and material incorporated in the high breast support. Within the high support condition, the mass of the breast is more evenly distributed across the chest wall when compared to the low breast support and bare-breasted condition. Qualitatively, the trajectories of the breast appear to change within both low and high breast support conditions from the start to the end of the five kilometre run. These reported findings have only been considered from a qualitative perspective, and therefore for further confirmation on distribution of movement in each plane the percentage distribution has been explored.

When running without breast support, vertical breast displacement relative to the thorax was greatest (40%), followed by mediolateral (33%) and then anteroposterior (27%). These findings are similar to Scurr et al., (2009a), who reported 50% of total multiplanar breast displacement occurred in the vertical direction during treadmill running. One reason for this commonly reported finding is the influence of the thorax segment, which has been proposed as the segment driving relative breast movement (Haake & Scurr, 2010). The mode and intensity of the exercise will govern the movement of the thorax. During treadmill running the movement of the thorax incorporates both translations and rotations in three-dimensions, with the greatest translation occurring in the vertical plane (Thorstensson, Nilsson, & Zomlefer, 1984). Therefore, it is not surprising that the greatest relative movement of the breast is reported within the vertical direction.

When external breast support was worn (low and high breast supports) during treadmill running, the distribution of breast movement was more evenly distributed between the planes of movement. Interestingly, the greatest percentage distribution of breast displacement in the low and high breast supports was in the mediolateral direction. This finding may indicate that sports bras designed for running should increase the stiffness in the material used for supportive components in the medial and lateral directions. Furthermore, in the low and high breast support conditions the smallest distribution of breast displacement, and greatest reductions in breast displacement were reported in the vertical direction. This finding may be explained by the recommendations of designs for breast support in previous years (Gehlsen & Albohm, 1980; Lorentzen & Lawson, 1987; Lawson & Lorentzen, 1990; Mason et al., 1999; Starr et al., 2005; McGhee et al., 2007). Before the publications of Scurr et al., (2009a; 2010a; 2011), who revealed the importance of examining multiplanar breast kinematics, breast movement was predominantly examined in the frontal plane, reporting only vertical and mediolateral breast displacement. Bra manufacturers may have based the design and structural components of a sports bra on these data. However, considering the results of breast biomechanics research within recent years (Scurr et al., 2010; White et al., 2009), and the results reported within this chapter, it is apparent that sports bras require structural support to reduce breast kinematics within the three planes of motion.

The second key finding of the current study was the reported increases in multiplanar breast kinematics from the first two minutes to the final distance intervals of the five kilometre run. These findings have implications for previous publications reporting the magnitude of breast kinematics over shorter run durations (two to five minutes), which have been used to define requirements for breast support during running (Lorentzen & Lawson, 1987; Lawson & Lorentzen, 1990; Mason et al., 1999; Starr et al., 2005; Scurr et al., 2009a; 2010a; 2011). Product testing for sports bra has been conducted over shorter exercise durations in the past (Starr et al., 2005). The results in the present chapter demonstrates that inferences regarding breast kinematics and the effectiveness of breast supports obtained from brief exercise trials cannot be extended to longer running distances. Significant increases in mulitplanar breast kinematics were reported from the first two minutes to the fifth kilometre in both low and high breast support conditions, therefore hypothesis two is accepted. Although the increases in breast kinematics were statistically significant, it is important to consider whether these differences were meaningful, and to what extent could these increases influence the biomechanics of a female runner. Based upon these findings it is suggested that breast biomechanics do not remain constant over a five kilometre run in a low and high breast support condition, and that increases were reported as early as the second kilometre of the five kilometre run. Therefore, it is imperative that future research within breast biomechanics carefully consider the protocol duration, as the duration of run examined will influence the magnitude of breast kinematics and the effectiveness (percentage reduction of movement) of the breast supports examined.

There are a number of possible explanations for these findings. Firstly, as the runner progresses through the five kilometre run, changes in running kinematics (e.g. failure to maintain optimum kinematics or alterations due to the onset of muscular fatigue) may be apparent (Williams, Snow, & Agruss, 1991). The thorax has been identified as the driving force of relative breast kinematics (Haake & Scurr, 2010); therefore, any alterations in the kinematics of the thorax, which may be brought about by mechanical alterations in other segments, may impact upon the magnitude and trajectory of multiplanar breast kinematics. Previous research has established changes in stride length, stride rate, increased foot contact during support period, and greater forward lean of the trunk, from the start intervals to the final intervals of 10 km (Elliot & Ackland, 1981) and 3 km (Elliot & Roberts, 1980) runs. However, there is disparity in the literature when assessing the relationship between run performance and mechanical fatigue. Cavanagh, Andrew, Kram, Rodgers, Sanderson, and Hennig (1985) concluded that individuals may adopt very different mechanical running forms to accommodate the effects of prolonged running, and that this may explain why there are disparities in the literature.

Secondly, the stress on the supportive tissues of the breast, caused by cyclic loading of this tissue relative to the thorax, may cause a degree of strain. In this instance, the ability of these tissues to restrict breast movement may be affected. Bowles and Steele (2003) reported a significant increase in vertical distance between the sternal notch and the nipple after a five minute run in a ‘poor’ breast support condition. Unfortunately no data are available to determine the practical implication of this increase; however, these data suggest the position of the breast has changed after exercise. The global position of the upper body during the static images was omitted from the abstract; therefore, it is unclear if this affected the reported results. If the orientation of the thorax was different between the two data collections then the position of the breast could be altered due to this, leading to a misinterpretation of these data. Alternative methods for assessing mechanical properties of human skin, such as rotational sensors (Agache, Monneur, Leveque, & De Rigal, 1980; Escoffier et al., 1989) and extensometers (Clark, Cheng, & Leung, 1996) used to quantify elasticity and strain rate of skin, respectively, may be considered as more appropriate measures for further examination of this hypothesis.

Thirdly, the performance and durability of the textile properties incorporated into the bras may differ over the course of the run. As the performer progresses through the run, body temperature and sweat rates are likely to increase, which may influence the mechanical properties of the materials, such as elasticity, stretch ability, recovery and strength (Shishoo, 2008). To ensure optimum performance, the materials and fibres of sports bras incorporate diverse mechanical properties (Page & Steele, 1999), where the ability of recovery is just as important as the ability of stretching, utilising materials such as intelligent elastane fibres. These intelligent textiles ensure that when the garment is exposed to high temperatures, the fibres undergo self-crimping and long-lasting stretch and recovery; ensuring heat does not reduce the performance of the bra (Shishoo, 2008; Senthilkumar, Anbumani, & Hayavadana, 2011). Furthermore, sports bras now contain materials which facilitate sweat evaporation, such as CoolMax® and Lycra®. The high breast support used within this study contains polyester, polyamide, and elastane, whereas the low breast support incorporates only polyamide and elastane Lycra. The blending of these fibres within the high breast support ensures the bra contains diverse mechanical properties. Polyester is the single most common fibre used for sportswear (Shishoo, 2008); the inclusion of polyester in the high breast support ensures a high level of strength. Consequently the low breast support may be subjected to greater stretch rate over time, due to the repeated stretch of the material at the cup and straps, as the breast tissue displaces throughout each gait cycle. Therefore, the material properties of the low and high breast support conditions may be subject to stretch over the five kilometre run, due to the interaction between the increased skin temperature of the individual and the temperature of the garment.

Fourthly, the global position of the bra on the thorax is another consideration for the reported increases in breast kinematics. The straps of a sports bra should include minimal elasticity for two reasons; firstly, to assist in the reduction of breast movement, and secondly, to minimize the occurrence of strap and bra slippage (Page & Steele, 1999). With this in mind, the possibility of bra slippage is reduced when considering the design and strap configuration of the high breast support used in the current chapter. However, bra slippage may have been prevalent within the low breast support condition, due to the materials and the classic U-back strap configuration of this bra. Bra slippage was not measured in the current study, but could account for the reported increases in magnitude of breast kinematics from the start to the end of the run within the two breast support conditions. Future research should monitor the global position of the bra when assessing the performance of external breast support over prolonged exercise.

One final consideration is the variance in these data. The kinematic data within the current study were associated with high standard deviations; therefore it is important to explore the within- and between-participant variance in these data. The breast is defined as a soft tissue, therefore, when considering rigid body mechanics these data may be more prone to sources of error; both measurement and random. Scurr et al., (2009a) reported low within-participant variance in resultant breast displacement over five gait cycles (6.2%), but up to 72% variance in breast displacement between-participants (2011). The greater variance between-participants may be attributed to the non-rigid characteristic of breast tissue, or the differences in composition and distribution of breast tissues (e.g. glandular, fat, and connective). The quantification and exploration of the variance in breast kinematic data presented in this chapter will help define the difference between a statistical difference and a meaningful difference, ensuring appropriate conclusions are drawn. Knudson (2009) detailed common errors and appropriate methodologies for interpreting effects in biomechanics research, defining the difference between a statistical difference and a meaningful difference.

Breast acceleration is an important variable when considering the forces acting on the breast, and the mechanical properties of the supportive breast tissues. An approximation of force was calculated within the current study, with the reported magnitudes similar to those reported by McGhee et al., (2012), who implemented the same method of breast force approximation. The force measured at the breast was reported to significantly decrease as breast support increased, which may reduce the chance of strain and damage to the supportive tissues of the breast. Relative vertical breast acceleration and approximated force were reported as greater than anteroposterior and mediolateral directions across the three breast support conditions. One fundamental explanation for this finding is gravitational acceleration acting solely within the vertical axis of the GCS (Mason et al., 1999; Gefen and Dilmoney, 2007; McGhee et al., 2012). The acceleration of 9.81 $m.s^{-2}$ subjected to the body within the vertical axis therefore contributes to the vertical acceleration and forces measured when the thorax is aligned with the vertical axis of the GCS. Within the current study, breast acceleration was calculated relative to the thorax and reported in metres per second per second. Mason et al., (1999), Scurr et al., (2010a), and Bridgman et al., (2010) have previously presented the relative acceleration of the breast in gravitational units (g), by dividing the acceleration in metres per second per second by the constant acceleration of gravity (9.81 $m.s^{-2}$). However, the orientation of the thorax should be considered when interpreting breast acceleration presented in these units. The thorax may not be perpendicular to the ground during the gait cycle, and therefore the vertical axis of the SCS may not be aligned with the vertical axis of the GCS, and the proportion of the gravity vector will change depending upon the thorax orientation. If presented in gravitational units, this should be accounted for by quantifying the orientation of the thorax and the degree at which the thorax is not aligned to the vertical axes of the GCS.

Breast velocity has previously been strongly correlated with exercise-related breast pain (McGhee et al., 2007; Scurr et al., 2010a). It was hypothesised that breast pain arises due to tension on the skin and fascia as the breasts move relative to the thorax (Mason et al., 1999), which activates the nerves at these sites. Within the current study, multiplanar breast kinematics positively correlated to exercise-related breast pain, with mediolateral breast acceleration demonstrating the strongest relationship. Reductions in multiplanar breast acceleration may decrease the internal forces on the supportive breast tissues, and therefore reduce the risk of damage. It is interesting to note that the mediolateral breast acceleration demonstrated the strongest relationship to breast pain. Although it was hypothesised that breast acceleration would demonstrate the strongest relationship, it was assumed that is would be the direction with the greatest magnitude (i.e. vertical), however, in line with the results presented, hypothesis three is accepted. Based upon these findings, sports bras designed for distance running may benefit from increased support in the mediolateral plane, as this may reduce exercise-related breast pain, and ensure females are running comfortably.

Participants rated their exercise-related breast pain as significantly less at the end of the five kilometre run in comparison to the first two minutes of running within the low breast support condition. This finding is interesting when considering the reported increases in multiplanar breast kinematics over the five kilometre run under this breast support condition. It was hypothesised that the cyclic repetitions over the prolonged run may cause greater tension on the breast tissues, and therefore may increase exercise-related breast pain. However, this was not evident within the current study. It might be hypothesised that the participants become accustomed to the tension placed upon these structures during the cyclic loading, and the pain is reduced. Previous literature has identified an increase in pain thresholds, and a decrease in pain ratings after a bout of aerobic (Janal, 1996; Hoffman & Hoffman, 2007) or resistance exercise (Koltyn & Arbogast, 1998). It is hypothesised that exercise can alter an individual’s perception of pain, and can act as effective and healthy pain management. However, these studies and review articles demonstrate the assessment of pain elicited by an external method (e.g. pressure stimulus, Forgione-Barber pain stimulator) before and after exercise, and not pain brought about by the movement itself (i.e. independent breast movement) during the activity. Taking the findings of the current study and the aforementioned literature into consideration, future work should consider the order of breast support conditions and the study design when examining exercise-related breast pain during prolonged running.

The sensitivity of the method employed to measure breast pain may need to be examined further. Subjective rating scales have been criticised for their lack of sensitivity (Downie, Leatham, Rhind, Wright, Branco, & Anderson, 1978), and the influence the participant's previous experiences have on the subjective rating. However, to enable comparisons with previous literature this method was employed, with the results suggesting that reductions in multiplanar breast displacement, velocity, acceleration, and approximated force provided by an external breast support, alleviate the magnitude of exercise-related breast pain. This reduction in exercise-related breast pain is beneficial to exercising females, ensuring they are able to maintain exercise intensities for longer durations, which may enhance sporting performance.

4.6 Conclusion

The work reported within the current chapter is the first within breast biomechanics research to examine multiplanar breast kinematics and approximated force in different breast support conditions over both short and prolonged running distances. The results are novel and add external validity to this crucial research area, ensuring the conclusions drawn are applicable to females exercising for a five kilometre run distance. The greatest reductions in multiplanar breast kinematics and breast pain were reported in the high breast support during short and prolonged running, which further promotes the use of the high breast support condition during running.

Multiplanar breast kinematics were reported to increase during the final distance intervals of the five kilometre run when compared to the initial two minute interval in both low and high breast supports. These findings have wide-reaching implications, the results of chapter three and four demonstrate that a high breast support (sports bra) is not as effective at the end of a five kilometre run compared to the first two minutes (a run duration commonly used in previous breast biomechanics research). Based upon these findings, it is suggested that protocols for breast biomechanics and sports bra product validation are extended to commonly performed run durations and distances. Interestingly, breast pain did not increase in conjunction with the increases in breast kinematics, and were actually reported to decrease in the low breast support condition at the end of the five kilometre run.

It is important for future work to consider the sources driving the reported increases and how these changes may influence a female runner. Alterations in running kinematics and/or differences in muscle activity profiles during the five kilometre run in different breast support conditions can be monitored through three-dimensional kinematic analysis and electromyographical analysis. Examination of these variables would provide a holistic view of the way in which external breast supports, and the associated magnitudes of breast kinematics, may influence biomechanical aspects of running for the female runner.

The work reported within this chapter is first to examine multiplanar breast kinematics over a five kilometre run distance. Quantification of the within- and between-participant variance will provide a profile of the variability in these data, which has not previously been presented in depth. Furthermore, these data will facilitate the interpretation of true differences for changes in magnitudes of breast kinematics due to the breast support worn, and the reported increases in breast kinematics over the five kilometre run reported in section one.

CHAPTER FOUR: SECTION TWO

WITHIN- AND BETWEEN-PARTICIPANT VARIANCE IN MULTIPLANAR BREAST KINEMATICS DURING A FIVE KILOMETRE RUN

4.7 Introduction

All tests and measurements include measurement error (Batterham & George, 2003; Payton & Bartlett, 2008), quantification of this error is required to conduct and interpret results within research (Hopkins, 2000; Batterham & George, 2003). The following sources of error constitute total error in a measurement; systematic bias and random error. Systematic bias refers to a general trend in the differences of measurements in a certain direction; this phenomenon commonly includes a ‘learning effect’ (Batterham & George, 2003), which may incorporate effects of fatigue in prolonged study designs and possible order effects when employing more than one condition (Atkinson & Nevill, 1998; Hopkins, 2000). For many measurements in sports science, the magnitude of error increases as the measured value increases (Atkinson & Nevill, 1998), known as heteroscedascity.

When quantifying the magnitude of breast kinematics the order of breast support should be considered, and randomised where possible. For instance, the increased magnitude of multiplanar breast kinematics reported during the bare-breasted and low breast support conditions within section one of the current chapter, may influence the measured outcome in later breast support conditions, due to the repetitive loading and possible strain on the anatomical restraints. Systematic bias can be a threat to the internal validity of a study, whereby the data collected prior can influence the data obtained in later tests (Batterham & George, 2003).

The second component of the total error, which is usually larger than the error due to systematic bias, is defined as random error. Random error can occur due to inherent biological or mechanical (measurement tool) variance (Atkinson & Nevill, 1998; Batterham & George, 2003). Unfortunately, the researcher can do little to reduce the biological aspect of random error because of its source, for example an individual's muscular strength may differ between tests, due to psychological factors such as motivation or due to physiological adaptations (Batterham & George, 2003). Day to day changes can be reduced by the standardisation of time of day and detailed pre-test guidelines. However, the mechanical error of measurement tools can be reduced to a certain extent by standardisation procedures and calibration processes (Atkinson & Nevill, 1998; Hopkins, 2000). Researchers should aim to calculate and report these data to ensure these sources are monitored and do not mask important effects.

Hopkins (2000) suggests that within-participant variance is the most important measure to consider when examining total error, as it can affect the precision of estimates of change in the measured variable. For example, providing the magnitude of the difference is greater than the within-participant variance reported, a meaningful difference will be reported. Therefore, the smaller the variance, the more confident we can be in the differences identified. A simple statistic which captures the notion for variance in within-participant repeated measures study designs is that of the standard deviation, which illustrates the spread of data about the mean (Atkinson & Nevill, 1998; Altman & Bland, 2005). The coefficient of variation (C_v), which is the ratio of the standard deviation to the mean expressed as a percentage (Atkinson, & Nevill, 1998; Batterham & George, 2003), enables comparisons between studies utilising different tools and measurements, regardless of units or calibration procedures (Hopkins, 2000). The C_v depends upon the magnitude of measured values and the agreement between these values, in other words it assumes the largest variance occurs in the measurement with the greatest values (Atkinson

& Neville, 1998). As mentioned previously, heteroscedascity is common in sports science data, however, it is important that this is explored and quantified before assuming it is present. Quantifying the within- and between-participant variance will ensure inferences made regarding the changes in magnitudes of breast kinematics within and between breast support conditions, are meaningful differences and enable accurate conclusions to be drawn.

Within human kinematic studies, quantifying the movement of the skeleton is commonly the primary aim (Lu & O'Connor, 1999; Winter, 1990; Wu & Cavanagh, 1995). A large magnitude of soft tissue movement relative to the skeleton is considered to be artefact, and considered to be the most critical source of error in human movement analysis (Leardini, Chiari, Della Croce, & Capozzo, 2005). To overcome this artefact, stringent methods are employed in an attempt to minimise them, such as optimisation methods (Lu & O'Connor, 1999), where distances between measured and modelled marker positions are optimised. However, when examining the independent movement of the breasts, it is this soft tissue movement that is considered and quantified relative to markers positioned on the thorax, and therefore these methods cannot be directly applied to the anatomical position of interest (the nipple marker which represents the global movement of the breast). With no muscles within the breast to damp and reduce oscillations, the breast may demonstrate non-uniform movement patterns, specifically during the contact phase of the gait cycle, which creates an impact force between the body and ground. The shock wave that is transmitted from the heel to the head is attenuated by deformation of biological tissues in the body (Derrick, Hamill, & Caldwell, 1998; Hamill, Derrick, & Holt, 1995). Therefore, it could be argued that when examining the kinematic parameters of the breast, moderate levels of variance will be present, due to the inability of the tissue to dampen oscillations.

One source of within-participant variance may be a result of biological changes to the composition of the breast between testing sessions; it is documented that breast volume may increase and density of the tissues alter during the luteal phase (day 14 to 28) of the menstrual cycle due to hormonal shifts (Warren, 2004; White et al., 1998; Page & Steele, 1999). The effect these changes may have on breast kinematics is currently unknown; however, in an attempt to minimise the likelihood of this, participants should be tested during days 5 to 15 of the 28 day menstrual cycle.

Scurr et al., (2009a) were the first to report within-participant variance in the measurement of resultant breast displacement during walking and running. Employing typical error measurement percentage coefficient of variance (TEM CV %), Scurr et al., (2009a) reported average C_v percentages over five gait cycles as 0.9% (walking at 5 km.h^{-1}) and 1.3%, (running at 10 km.h^{-1}), indicating a very consistent soft tissue movement pattern.

Another source of within-participant variance, which is difficult to overcome during running studies, is variance in individual running technique. The thorax has been identified as the ‘driving force’ behind breast movement (Haake & Scurr, 2010); therefore alterations in the kinematics of the thorax, other body segments, and gait parameters may impact upon the relative kinematics of the breast, which may be more evident during longer duration running as the performer tires.

Differences in gait kinematics between-participants, such as stride parameters and magnitude of segment degrees of freedom, specifically the upper body, will also influence breast kinematic data when grouped to create a mean data set. Between-participant variance in relative resultant breast displacement has been reported by Scurr et al., (2011) during a two-minute incremental speed treadmill test (ranging in speed from 5 to 14 km.h^{-1}) and found up to 72% variance between-participants. It was concluded that the high between-participant variance may be explained by the range of chest band sizes within the sample, resulting in a range of breast volumes among the participants, however this cannot be confirmed as the relationship of breast volume and variance was not reported.

Another crucial source of between-participant variance to consider is the difference in the composition of the breast. The dimensions, density and mass of a breast vary substantially between individuals (Vandeweyer & Hertens, 2002; Boston et al., 2005; Gefen & Dilmoney, 2007), moreover, the composition of these tissues is likely to vary in distribution between breasts of the same breast volume (e.g. the 34 B, is the equivalent breast volume as a 32 C and 36 A) (Hardaker & Fozzard, 1997). Boston et al., (2005) reported a breast composition ratio of 69.9 ± 22.9% and 30.1 ± 22.9% of fat to glandular tissue, respectively, in a cohort of females aged 38 to 70 years using magnetic resonance imaging (MRI). From the large standard deviations reported it can be seen that the ratio is highly variable between individuals, this may be explained due to the non-standardisation of breast size and age. Gefen and Dilmoney (2007) stated that the ratio of fat to connective tissue will determine the firmness of the breast, which may be related to the movement of the breast. The two anatomical restraints to the breast, the overlying skin and Cooper’s ligaments are affected by hydration and age status with the skin losing elasticity becoming more lax, and ultimately leading to breast ptosis with increasing age (Page & Steele, 1999; Gefen & Dilmoney, 2007). Due to differences in breast composition between-participants, the anatomical support provided to the breast may vary substantially even between individuals of the same breast volume, which therefore could lead to large between-participant variance in breast kinematic data.

Scurr et al., (2009a; 2011) reported the within- and between-participant variance in resultant breast displacement over constant and incremental speed treadmill protocols, lasting only two minutes in duration. These are the only data available on variance in breast kinematics and therefore, variance in breast kinematics has not been quantified over prolonged running distances. Alterations in running kinematics during a run (Williams, Snow, & Argruss, 1991; Kyröläinen, Pullinen, Candau, Arela, Huttenen, & Komi, 2000; Hardin, Van Den Bogert, & Hamill, 2004) may therefore influence the magnitude of variance in breast kinematics. Furthermore, within- and between-participant variance has only been reported for resultant breast displacement; therefore no inferences can be made regarding the variance in multiplanar breast kinematics. Magnitudes of breast kinematics have been reported to differ between the three directions of movement (Scurr et al., 2010a); therefore the associated variance may also differ between the three directions of movement. It is currently unknown whether the magnitude of variance in mulitplanar breast kinematics is homogenous across directions, or whether it follows a heterogeneous pattern. Exploring these data would help establish the variance in multiplanar breast kinematic data. Furthermore, magnitudes of multiplanar breast velocity and acceleration have been reported during running (Scurr et al., 2010a), but the associated variance in these data is yet to be examined. The derivative calculation of velocity and acceleration may exhibit greater magnitudes of variance due to the magnification of any error in the coordinate data during the calculation (magnification of 20 times at the second derivative) (Pezzack, Norman, & Winter, 1977). Therefore, the magnitude of variance in acceleration should be considered and reported, as the variance may mask any important effects in these data between conditions and trials.

4.8 Aims and research hypotheses

Using the data acquired previously in this chapter, the aim of this section was to explore the within- and between-participant variance in multiplanar breast kinematics in the three breast support conditions, during the two minute and five kilometre treadmill runs. Furthermore, this section aimed to identify if meaningful differences were reported.

H_1 The magnitude of breast support will have a significant effect on the magnitude of within-participant variance in multiplanar breast kinematics.

H_2 Within-participant variance in multiplanar breast kinematics will significantly increase from the first two minutes to the final kilometre of the five kilometre run.

H_3 The direction of movement will significantly affect the magnitude of within-participant variance in multiplanar breast kinematics.

H_4 Between-participant variance in multiplanar breast kinematics will be greater than the within-participant variance in multiplanar breast kinematics.

4.9 Methods

4.9.1 Procedures

Further data analysis were conducted on the data presented in section one of chapter four. See chapter four, section one, 4.4.1 for data collection methods.

4.9.2 Data processing

Coefficient of variance (C_v), reported as a relative percentage of the mean (Equation 1) was used to quantify the within- and between-participant variance in relative multiplanar breast displacement, velocity and acceleration. The within-participant variance was calculated for five gait cycles, during the first two minutes of running in the three breast support conditions (bare-breasted, low, and high), and at each interval of the five kilometre run, for the low and high breast supports, for each participant and then averaged across participants (n=10). The between-participant variance was calculated across the ten participants using the mean data of the first two minutes, and again at each interval of the five kilometre run, in the low and high breast support conditions.

Equation 1. $C_v = \sigma / \mu * 100$

Where, σ is the standard deviation and μ is the mean.

4.9.3 Estimation of technical error in the motion capture system

The accuracy and precision of the motion capture and analysis systems (eight cameras, sampling at 200 Hz) were determined by recording two markers on a rigid calibration wand (Qualysis, Sweden), with a known inter-marker distance of 750.7 mm. The accuracy of the system was defined as the difference between the known inter-marker distance and the mean reported inter-marker distance recorded over three, 10 second trials (Table 11).

During the three trials the movement of the rod imitated rotation of the shoulders during running. The precision of the motion capture system was defined as the mean of the standard deviations (SD) of these three trials. The mean accuracy of the system was measured at 0.4 mm, and the precision of the system was measured at 0.2 mm. Therefore, the technical error of the motion capture system was defined as less than 1 mm.

Table 11. Accuracy (mm) and precision (mm) of the motion capture system.

	Trial 1	**Trial 2**	**Trial 3**	**Mean**
Mean inter-marker distance (mm)	750.2	750.2	750.4	750.3
SD of inter-marker distance (mm) (Precision)	0.1	0.2	0.3	0.2
Accuracy (mm)	0.5	0.5	0.3	0.4

4.9.4 Statistical analyses

Within-participant variance (C_v) in breast kinematic data within the three breast support conditions were checked for normality (Kolmogorov-Smirnov and Shapiro-Wilk) and homogeneity of variance (Mauchly’s test of Sphericity), during the first two minutes of running and at each distance interval of the five kilometre run, where normality was assumed when $p > .05$. One-way repeated measures ANOVAs were performed to assess the effect of breast support on the within-participant variance in breast kinematic data during the first two minutes of running. Following this, two-way repeated measures ANOVAs with *post hoc* pairwise comparisons (with Bonferroni adjustment) were performed to assess the effect of breast support and run distance (kilometre intervals) on the magnitude of within-participant variance in mulitplanar breast kinematics. To examine the effect of the direction of movement on within-participant variance of breast kinematics during the two minute and five kilometre treadmill run, Wilcoxon signed rank tests were employed. A Bonferroni adjustment was calculated dependent upon the number of pairs tested (Fields, 2009). Due to the calculation of the between-participant variance, statistical analysis could not be performed on a single value data set.

4.10 Results

4.10.1 Variance in breast displacement (mm)

Breast support did not significantly affect the magnitude of within-participant variance in anteroposterior and mediolateral breast displacement (Table 12). Variance in the vertical breast displacement was however affected by breast support worn, with the high breast support showing greater variance than the low breast support at the second (p = .023), third (p = .006), and fourth kilometre (p = .016), increasing the within-participant variance by 8%, 7%, 6%, respectively. The within-participant variance in multiplanar breast displacement did not differ between distance intervals of the five kilometre run.

Within-participant variance in multiplanar breast displacement was significantly different dependent upon the direction of movement during the five kilometre run, in both low and high breast support conditions. Greater levels of within-participant variance were reported in the anteroposterior direction when compared to the mediolateral direction in the low (Z = -4.340, p = .001) and high (Z = -3.001, p = .001) breast supports. Additionally, the variance in the vertical breast displacement was significantly greater than the mediolateral displacement (Z = -4.697, p = .001).

Between-participant variance in multiplanar breast displacement was greater than the within-participant variance, with the greatest between-participant variance reported in the vertical direction (33%) in the high breast support condition, during the first kilometre.

4.10.2 Variance in breast velocity ($m.s^{-1}$)

Breast support did not significantly affect the magnitude of within-participant variance in anteroposterior and mediolateral breast velocity (Table 13). However, the within-participant variance in vertical breast velocity was affected by the breast support worn. Greater magnitudes of within-participant variance were reported in the high breast support compared to the low breast support during the second (p =.003), third (p < .001), and fourth (p = 0.17) kilometre of the run, increases in variance of 7%, 10%, and 8%. The within-participant variance in multiplanar breast velocity did not differ across the distance intervals of the five kilometre run.

Within the low breast support condition, the magnitude of within-participant variance was significantly different between the three directions of movement, with variance in the anteroposterior velocity demonstrating greater magnitudes than the variance in the mediolateral (Z = -2.179, p = .029) and vertical (Z = -4.741, p < .001) breast velocity. Furthermore, within-participant variance in the mediolateral breast velocity was significantly greater than the variance in the vertical velocity (Z = -3.129, p = .002). Within the high breast support condition, the within-participant variance was significantly greater in the anteroposterior direction when compared to the mediolateral direction (Z = -2.168, p = .030).

Between-participant variance in multiplanar breast velocity was greater than the within-participant variance, with the greatest variance reported in the vertical direction (46%) in the high breast support during the first two minutes. Similarly, the greatest within-participant variance in multiplanar breast velocity was also reported in the vertical direction (22%) in the high breast support, during the third kilometre interval.

4.10.3 Variance in breast acceleration ($m.s^{-2}$)

Breast support condition did not significantly affect the within-participant variance in anteroposterior and mediolateral breast acceleration (Table 14). However, within-participant variance in vertical breast acceleration was significantly greater in the high breast support compared to the low breast support during the second (p = .020), third (p = .001), fourth (p = .005), and fifth (p = .049) kilometre intervals, increasing by 9%, 10%, 11%, and 8%, respectively. Within-participant variance in multiplanar breast acceleration did not differ across the kilometre intervals of the five kilometre run.

Within the low breast support conditions greater magnitudes of within-participant variance were reported in the anteroposterior direction when compared to the vertical plane (Z = -5.956, p = .001). Similarly, variance in mediolateral breast acceleration was significantly greater than variance in vertical breast acceleration (Z = -5.013, p = .001) within the low breast support condition.

Between-participant variance in multiplanar breast acceleration was greatest in the vertical direction in the high breast support, during the first two minutes. The greatest within-participant variance in multiplanar breast acceleration was reported in the anteroposterior direction (30%) in the high breast support.

Table 12. Mean (± SD) multiplanar breast displacement (mm) during the five kilometre run, in the three breast support conditions, and the associated within- and between-participant coefficient of variance (%) (*n* = 10).

DISPLACEMENT	INTERVALS														
	2 MINS			1 KM		2 KM		3 KM		4 KM		5 KM		MEAN	
	BB	L	H	L	H	L	H	L	H	L	H	L	H	L	H
AP (mm)	41	34	24	36	27	38	27	38	28	37	26	37	27	36	27
SD (mm)	5	3	6	3	6	4	6	5	6	5	6	7	7	5	6
WITHIN (C_v%)	9	7	10	7	10	8	9	9	9	10	9	11	9	9	9
BETWEEN (C_v%)	12	9	25	8	22	11	23	13	21	14	22	19	26	12	23
ML (mm)	50	36	26	38	30	41	31	40	32	39	32	39	31	38	30
SD (mm)	11	7	5	7	5	9	7	7	7	8	7	10	5	8	6
WITHIN (C_v%)	8	6	8	5	6	6	7	6	7	6	6	6	6	6	7
BETWEEN (C_v%)	21	19	19	18	17	22	23	18	22	21	22	26	16	21	20
V (mm)	60	34	18	38	21	39	22	40	23	40	23	41	31	39	23
SD (mm)	13	10	5	9	7	10	5	9	6	9	6	9	6	9	6
WITHIN (C_v%)	5	8	11	7	11	6*[c]	14*[c]	6*[c]	13*[c]	4*[c]	10*[c]	10	12	7	13
BETWEEN (C_v%)	25	29	28	24	33	26	21	23	26	23	26	22	19	25	26

*[c] Denotes a significant difference between low and high breast support conditions ($p < .05$).
N.B. AP = anteroposterior, ML = mediolateral, V = vertical, BB = bare-breasted, L = low, H = high.

N.B. Significant main effect of breast support on the within-participant variance in vertical breast displacement ($F_{(1)} = 22.382$, $p = .001$, $\eta^2 = .713$, $1-\beta = .987$) at the 2nd, 3rd, and 4th kilometre of the run.

Table 13. Mean (± SD) multiplanar breast velocity (m.s^{-1}) during the five kilometre run, in the three breast support conditions, and the associated within- and between-participant coefficient of variance (%) (*n* = 10).

VELOCITY	INTERVALS														
	2 MIN			1 KM		2 KM		3 KM		4 KM		5 KM		MEAN	
	BB	L	H	L	H	L	H	L	H	L	H	L	H	L	H
AP (m.s^{-1})	0.43	0.28	0.18	0.29	0.19	0.36	0.20	0.32	0.21	0.34	0.19	0.32	0.20	0.32	0.20
SD (m.s^{-1})	0.12	0.07	0.06	0.05	0.05	0.08	0.06	0.08	0.06	0.09	0.04	0.08	0.06	0.08	0.06
WITHIN (C_v%)	12	19	16	17	14	12	15	14	13	15	17	19	14	16	15
BETWEEN (C_v%)	32	25	36	17	37	22	32	27	29	26	19	25	31	24	31
ML (m.s^{-1})	0.50	0.24	0.16	0.26	0.19	0.27	0.18	0.26	0.18	0.26	0.18	0.26	0.18	0.26	0.18
SD (m.s^{-1})	0.15	0.04	0.04	0.04	0.05	0.06	0.04	0.04	0.04	0.04	0.05	0.05	0.05	0.05	0.05
WITHIN (C_v%)	16	17	10	14	15	12	15	12	12	13	16	12	11	13	13
BETWEEN (C_v%)	43	16	36	15	28	22	24	14	21	15	30	21	27	17	28
V (m.s^{-1})	0.88	0.47	0.22	0.54	0.26	0.54	0.27	0.57	0.29	0.55	0.28	0.57	0.28	0.54	0.27
SD (m.s^{-1})	0.29	0.15	0.10	0.17	0.11	0.19	0.10	0.18	0.11	0.18	0.11	0.16	0.10	0.17	0.11
WITHIN (C_v%)	9	11	14	10	13	9*^{c}	16*^{c}	12*^{c}	22*^{c}	8*^{c}	16*^{c}	10	11	10	15
BETWEEN (C_v%)	32	32	46	32	42	36	37	32	38	32	40	29	36	32	40

*^{c} Denotes a significant difference between low and high breast support conditions ($p < .05$).
N.B. AP = anteroposterior, ML = mediolateral, V = vertical, BB = bare-breasted, L = low, H = high.

N.B. Significant main effect of breast support on the within-participant variance in vertical breast velocity ($F_{(1)} = 64.404$, $p < .001$, $\eta^2 = .877$, $1\text{-}\beta = 1.000$) at the 2nd, 3rd, and 4th kilometre intervals of the run.

Table 14. Mean (± SD) multiplanar breast acceleration (m.s^{-2}) during the five kilometre run, in three breast support conditions, and the associated within- and between-participant coefficient of variance (%) (*n* = 10).

ACCELERATION	INTERVAL													
	2 MIN			1 KM		2 KM		3 KM		4 KM		5 KM		MEAN	
	BB	L	H	L	H	L	H	L	H	L	H	L	H	L	H
AP (m.s^{-2})	22.0	13.9	7.0	13.9	7.0	16.3	8.7	14.2	7.6	15.4	7.9	14.4	7.9	14.7	7.8
SD (m.s^{-2})	8.4	5.2	3.1	5.1	3.3	6.8	2.9	6.2	3.0	5.2	2.7	5.8	3.2	1.0	0.6
WITHIN (C_v%)	12	23	20	18	20	16	30	19	21	21	20	21	23	20	22
BETWEEN (C_v%)	38	37	44	37	44	42	33	44	40	34	35	41	40	44	39
ML (m.s^{-2})	15.8	7.9	5.0	8.2	5.4	8.5	5.6	8.9	5.0	8.7	5.1	8.0	5.2	8.4	5.2
SD (m.s^{-2})	5.4	2.1	1.3	2.1	1.7	2.5	1.7	2.8	1.3	2.4	1.8	2.0	1.4	0.4	0.2
WITHIN (C_v%)	16	21	26	18	23	17	23	23	20	17	23	20	21	21	23
BETWEEN (C_v%)	34	27	25	25	32	29	29	31	25	27	34	25	27	27	28
V (m.s^{-2})	34.1	23.2	9.5	26.8	11.3	27.1	11.2	28.2	11.5	28.4	11.7	28.8	12.2	27.1	11.2
SD (m.s^{-2})	11.0	6.9	5.3	7.5	6.0	8.8	5.0	8.3	5.1	8.0	2.2	6.5	4.6	2.1	0.9
WITHIN (C_v%)	8	12	18	12	16	11*^{c}	20*^{c}	12*^{c}	22*^{c}	8*^{c}	19*^{c}	11*^{c}	19*^{c}	11	19
BETWEEN (C_v%)	32	30	56	28	53	32	45	30	44	28	47	23	37	29	47

*^{c} Denotes a significant difference between low and high breast support conditions ($p < .05$).
N.B. AP = anteroposterior, ML = mediolateral, V = vertical, BB = bare-breasted, L = low, H = high.

N.B. Significant main effect of breast support level on the within-participant variance in vertical breast acceleration ($F_{(1)} = 18.701$, $p = .002$, $\eta^2 = .675$, $1\text{-}\beta = .969$) at the 2nd, 3rd, 4th, and 5th kilometre intervals.

4.11 Discussion

This is the first study to explore the magnitude of within- and between-participant variance in multiplanar breast kinematics between different breast support conditions, during prolonged treadmill running. Importantly, the work in this section sought to establish whether the magnitude of differences reported in section one of chapter four could be accepted as meaningful differences, whereby the differences reported exceed the total variance in the data. The key findings were; i) breast support was shown to significantly influence the magnitude of within-participant variance, ii) within-participant variance in multiplanar breast kinematics did not differ across the distance intervals of the five kilometre run, iii) the magnitude of within-participant variance in breast kinematics differed between the three directions of movement, iv) between-participant variance was greater than the within-participant variance in multiplanar breast kinematics. An important consideration was the technical error of the Qualisys camera system, which was established as less than 1 mm, therefore, the majority of the reported variance in breast kinematics was assumed to be comprised of biological variance and systematic bias, and as such has implications for identifying meaningful differences in breast kinematics data.

Within-participant variance in multiplanar breast displacement, over the five kilometre run averaged 7% in the low breast support, and 10% in the high breast support. As these data were derived for breast velocity the average variance increased to 13% (low) and 14% (high), and increased again to 17% (low) and 21% (high) for acceleration. These within-participant variance values are equivalent to a total error in breast kinematics of 3 mm, 0.04 $m.s^{-1}$, and 2.8 $m.s^{-2}$ in the low breast support condition, and 3 mm, 0.03 $m.s^{-1}$ and 1.7 $m.s^{-2}$ in the high breast support condition. When considering the magnitude of differences reported within the previous chapter, meaningful differences can only be assumed when these values exceed the technical error (1 mm) and total error reported. Therefore, within the current population, within-participant differences of 3 mm (or less) in displacement, 0.04 $m.s^{-1}$ (or less) in velocity, and 2.8 $m.s^{-2}$ (or less) in acceleration within and between high and low breast support conditions are equivalent to the magnitude of identified total error, and therefore would not be considered as a meaningful difference. The smallest differences reported in multiplanar breast kinematic data, within and between the breast support conditions, for the previous chapter have been summarised in Table 15. The greatest differences in breast kinematics were reported consistently within vertical direction, when compared to mediolateral and anteroposterior directions. The differences reported in the previous chapter within and between breast support conditions all exceed the total error, and therefore can be accepted as meaningful differences.

Table 15. Smallest magnitude of differences reported in breast kinematic data in chapter four, section one, within and between the low and high breast support conditions over the five kilometre run (n = 10).

Multiplanar kinematics		**Within support conditions**		**Between support conditions**
		Low	*High*	*Low vs. High*
Displacement	*A-P*	ND	3 mm	10 mm
	M-L	ND	6 mm	8 mm
	V	4 mm	4 mm	17 mm
Velocity	*A-P*	ND	ND	$0.10\ m.s^{-1}$
	M-L	ND	ND	$0.08\ m.s^{-1}$
	V	$0.08\ m.s^{-1}$	$0.06\ m.s^{-1}$	$0.25\ m.s^{-1}$
Acceleration	*A-P*	ND	ND	$6.9\ m.s^{-2}$
	M-L	ND	ND	$2.9\ m.s^{-2}$
	V	$5.6\ m.s^{-2}$	$2.7\ m.s^{-2}$	$13.7\ m.s^{-2}$

N.B. ND = No difference reported

Significantly greater within-participant variance in vertical breast displacement, velocity and acceleration were reported in the high breast support, when compared to the low breast support during the five kilometre run, therefore H_1 can be accepted. Due to the calculation of the coefficient of variance, the magnitude of the mean has the potential to either elevate or reduce the absolute value reported (Hopkins, 2000). This is emphasised when small magnitudes are reported as a relative statistic. This effect may be prevalent within the current study when considering the magnitudes of breast kinematics reported in the different breast supports employed. For example, when considering the mean and standard deviations of the anteroposterior velocity in the low and high supports during the first two minutes ($0.24\ m.s^{-1} \pm 0.04$ and $0.16\ m.s^{-1} \pm 0.04$, respectively), the standard deviations are the same, but mean values different, however, the relative C_v statistics are reported as 17% and 25%, respectively. This may explain why the high support was commonly reported to have higher variance than the low support when reported using this statistic.

An arbitrary criterion for an acceptable level of variance is a C_v percentage of 10% has been proposed, and is frequently employed within sports science research studies

(Atkinson & Neville, 1998). However this value is by no means definitive and many studies within sports science research would not employ this guideline as a criterion for acceptable variance. Prior knowledge and implementation of this arbitrary value may influence the reader's interpretation of results across studies, utilising different methods and measurement tools. If the 10% criterion was employed as an 'acceptable' limit of variance within the current study, many of the differences reported could not be considered as meaningful. When assessing the smallest differences reported in section one of this chapter in absolute units, all differences reported were meaningful. However, the C_v percentages exceeded 10%. Therefore, it is imperative to also consider the total error in absolute units of the measurement. Atkinson and Neville (1998) suggested that the definition of an acceptable degree of variance should be approached by many within the disciplines of sports science, and that the statistical method sensitive for that research area should be agreed upon. However, until more data are available on variance in multiplanar breast kinematics across different breast sizes and exercise modalities, this cannot be accomplished in this research domain.

Within-participant variance of relative multiplanar breast kinematics was not reported to differ over the five kilometre run within the low or high breast support conditions. With no differences across the kilometre intervals, it is suggested that the magnitude of variance is consistent across prolonged treadmill running (five kilometres). This finding rejects H_2 and suggests that variance could be examined within the first two minutes of data collection. The decision for further examination of variance in these data is ultimately left to the investigator.

The magnitude of within-participant variance within the current study was substantially greater than the within-participant variance reported by Scurr et al., (2009a). A log transformation was applied to the data of Scurr et al., (2009a), which is advocated when the data do not follow normal distribution (Atkinson & Neville, 1998; Hopkins, 2000; Bland & Altman, 1996) and demonstrate heteroscedascity. This type of transformation uniforms the variance to produce a homogenous data set; however it should be acknowledged that many biological parameters do not follow equal distributions due to the inherent random biological error, making these data unsuitable for log transformation (Bland & Altman, 1996). For the efficacy of future research in the area and to enable conclusive findings to be reported, within-participant variance in breast kinematics should be reported without log transformation. Moreover, future research on larger cohorts of participants, across multiple breast sizes, should establish generalised boundaries of error.

Within the current study the between-participant variance in breast displacement, across the three directions of movement, in the low and high support conditions averaged 20% and 23%, respectively. Again, as these data were derived for breast velocity and acceleration the average between-participant variance increased to 24% and 25% for the low breast support, and 33% and 32% for the high breast support, respectively. These between-participant percentage values are equivalent to a total error of 8 mm, 0.08 $m.s^{-1}$, 5.5 $m.s^{-2}$ for the low breast support, and 6 mm, 0.06 $m.s^{-1}$, 2.6 $m.s^{-2}$ for the high breast support. These findings have important implications for establishing participant sample sizes in future studies using between-group designs. These data can help inform future studies of appropriate sample sizes, which are large enough to generate acceptable statistical power and effect size, and reduce type 2 errors. The earlier findings reported in section one of chapter four, suggest that a sample of ten participants is large enough to identify differences in breast kinematics within and between breast support conditions, during short and prolonged run distances.

Separating breast kinematics data into individual directions of movement (anteroposterior, mediolateral, and vertical) enables a greater understanding of which direction has larger magnitudes of variance. Anteroposterior breast kinematics frequently demonstrated significantly greater magnitudes of within–participant variance than the mediolateral and vertical directions, indicating a more sporadic movement pattern in this direction, and leading to acceptance of H_3. It is unclear why anteroposterior breast kinematics displayed higher levels of within-participant variance; one proposed explanation for this finding is linked to a discussion point within the previous chapter. The reduction of movement within the anteroposterior direction, provided by the external breast support, may not be as effective as the vertical and mediolateral directions, due to the majority of breast biomechanics research previously carried out within the frontal plane of movement. Therefore, the anteroposterior movement may be more sporadic due to the inability of the bra to reduce movement in this direction. This explanation for the greater variance in this direction is supported by the variance in the bare-breasted condition, since the greatest variance was not seen in this direction when the breast was unsupported. A second reason for this finding is the influence of thorax kinematics during running. The variance in thorax kinematics within- and between-participants will directly influence the relative movement of the breast tissue, since the thorax is the segment driving this movement. Further exploration of this segment alongside breast kinematic data, may help to establish the relationship between the thorax and the breast.

Greater magnitudes of variance were reported between-participants when compared to within-participant, therefore H_4 can be accepted. When examining the magnitudes of between-participant variance in breast kinematic data, it is important to consider the implications of grouping data from multiple participants to establish a mean value. The between-participant C_v percentages within the current study were similar to those reported by Scurr et al., (2011), however, the peak between-participant variance reported by Scurr et al., (2011) was 26% greater than the peak value reported within the current study. It is important to consider why these values are different. Firstly, Scurr et al., (2011) reported the between-participant variance in resultant breast displacement and not multiplanar kinematics, therefore, the combination of all planes of movement within the calculation of resultant breast kinematics may increase the reported percentages. Secondly, Scurr et al., (2011) examined these data over a range of treadmill speeds for an incremental treadmill test, and therefore the time spent at each speed (five gait cycles) cannot be assumed as long enough to establish a biomechanical steady state of running. The quick change in treadmill speed could have affected the participants running kinematics, which in turn could influence relative kinematics of the breast. Thirdly, the cup size of the sample recruited by Scurr et al., (2011) was a D, however, the band size ranged from 32 to 36 inches. Therefore, the volume of breast tissue differed between participants, further adding to the sources of between-participant variance. Finally, the cut off frequency of the Butterworth filter applied to these data was different between studies, with the current study employing a cut off frequency of 13 Hz, and Scurr et al, (2011) at 10 Hz. The cut off frequency of the filter will affect the magnitude of 'noise' that is attenuated, and therefore the signal that is passed will differ between these two frequencies (Winter, 1990).

4.12 Conclusion

The work within this section explored the within- and between-participant variance in breast kinematic data, and established the differences reported within section one of this chapter as meaningful (i.e. the differences reported exceeded the variance in these data). Firstly, it was found that within-participant variance in breast kinematics was significantly affected by the breast support worn, with greater variance reported in the high breast support. Secondly, the magnitude of within-participant variance in breast kinematics remained constant over a five kilometre treadmill run in both low and high breast supports. Thirdly, the within-participant variance in anteroposterior breast kinematics was significantly greater than the mediolateral and vertical, indicating a more sporadic movement in this direction. Finally, greater magnitudes of variance were reported between-participants than within-participants despite similar breast volumes. As a result of the exploration of the variance in breast kinematics data within the present programme of work, for any effect to be considered as meaningful, it is recommended that the smallest reported differences exceed the total error reported.

Conclusions drawn from this study may influence future study designs, sample sizes, data collection, and analysis procedures used within breast biomechanics research. It is important for the progression of this research area that the presence and sources of within- and between-participant variance in breast kinematics are identified and quantified, and the margin for meaningful differences is defined. The results of chapter four suggest that a sample of ten participants is large enough to identify differences in breast kinematics within and between breast support conditions during short and prolonged run distances.

Changes in running kinematics were proposed as an explanation for the differences reported in multiplanar breast kinematics across the five kilometre run during the first section of the current chapter of this thesis. Monitoring running kinematics over the five kilometre run in different breast supports, and quantifying the magnitude of variance in these data, will help establish the effect of this source of within- (e.g. within gait cycles, distance intervals) and between-participant variance (e.g. between participants) in breast kinematics. The fifth chapter of this thesis aims to quantify running kinematics in different breast support conditions during a five kilometre run. These data will provide crucial data on female running biomechanics in different breast support conditions, and provide the first in-depth exploration of the relationship between the body and the breast during running.

CHAPTER FIVE.

THE EFFECT OF BREAST SUPPORT ON UPPER AND LOWER BODY SEGMENTAL KINEMATICS DURING FIVE KILOMETRE RUNNING

5.1 Introduction

Presently, the work of Boschma et al., (1995) is the only study to examine the effect of breast support on running kinematics. Based upon anecdotal evidence, Boschma et al., (1995) hypothesised that females who experience large breast displacements and exercise-related breast pain may be deterred from exercise, and more specifically that a female's running biomechanics may be significantly affected by these factors. Boschma et al., (1995) investigated stride rate, stride length, vertical trunk displacement, front arm range of motion, and vertical breast motion during a five minute treadmill run, in different breast support conditions. No differences were found in the kinematic parameters measured between breast support conditions, however, it was reported that a decreased level of breast support elicited a significant decrease in the magnitude of vertical trunk displacement in certain participants within the sample. It was suggested that participants running kinematics were affected by the breast support worn on an individual basis. The work of Boschma et al., (1995) identified unique differences in trunk kinematics (i.e. vertical displacement) when wearing different breast supports. However, a more in-depth analysis of this segment, and investigation into the cause and effect relationship between the thorax and breast kinematics will further the understanding of the segment previously identified as the primary segment driving relative breast kinematics (Haake & Scurr, 2010).

When considering the moment of inertia of a segment, it is established that a segment with a greater distribution of mass from the axis of rotation, will have a greater moment of inertia. Therefore, it is expected that the change in distribution of the breast mass in different breast supports may influence the kinematics of the thorax. In the bare-breasted condition, the distribution of breast mass is assumed to be the furthest from the long axis of the thorax, when compared to low and high breast support conditions. It is therefore expected that the bare-breasted condition could elicit the greatest change in thorax kinematics and vice versa.

Furthermore, the breast does not encompass any bones or muscles, but contains both glandular and fat tissue, and therefore is considered as a wobbling mass unique to the female athlete. The inertial effects of the breast tissue causes a time-lag between the movement of the thorax and the movement of the breast during treadmill running (Scurr et al., 2009a; Haake & Scurr, 2010), creating an out-of-phase movement pattern, which may influence thorax mechanics. Currently it is unknown which is more influential; the movement of the thorax driving the relative breast kinematics, or the inertial effects of the breast mass influencing the kinematics of the thorax, or a combination of both. It is hypothesised that the magnitude of breast support worn will directly influence the distribution of breast mass and therefore the inertial effects of the breast, and indirectly influence the kinematics of the thorax segment. It is important to consider the type of breast support worn and the distribution of breast mass when investigating this cause and effect relationship between breast and thorax. Increased understanding of the relationship between thorax and breast biomechanics could facilitate sports bra manufacturers in optimising key components of sports bras for specific exercise modalities (i.e. including greater medial and lateral support for sports that elicit large thorax rotations about the long axis).

The work of Boschma et al., (1995) was conducted during a five minute treadmill run, and it is well established that running kinematics change over time (Hardin, Van Den Bogert, Hamill, 2004; Stirling, Tscharner, Hoon Kim, & Nigg, 2008; Williams & Cavanagh, 1987; Williams, Snow, & Argruss, 1991). Stirling et al., (2008) reported greater forward leaning and an increased stride period at the end of a 60 minute constant velocity treadmill run, in 15 female participants. Stirling et al., (2008) suggested that the forward lean was a compensation mechanism for increased demands for oxygen at the later stages of the run. Given that both Stirling et al., (2008) and Williams and Cavanagh (1987) have reported this body position to be associated with lower oxygen consumption than those who maintain a more upright posture, it is of interest when examining alterations in running mechanics and running performance.

Williams, Snow, and Agruss (1991) investigated changes in distance running kinematics with fatigue during treadmill protocols. Williams et al., (1991) reported marked changes in kinematic variables including differences in step length, maximum angle of the thigh during hip flexion, and maximal angle of the knee during swing. Differences were identified for individual participants but these differences did not consistently match the differences reported for the entire sample, suggesting that there are considerable inter-individual differences, with some runners being noticeably more sensitive to mechanical fatigue, while others maintain a constant mechanical running form.

The majority of literature on human locomotion is centred on lower limb kinematics (Diss, 2001; Ferber, McClay Davis, & Williams, 2003). In early publications the upper body was modelled as a single rigid segment (Winter, 1987; Kubo & Ulrich, 2006). Syczewska, Öberg, and Karlsson (1999) identified the importance of quantifying and reporting the individual segmental movements of the upper body, stating that it is hard to believe that the upper body is only passively carried by the legs since the upper body accounts for more than 60% of total body mass. Within recent years, more research has focussed on the roles of upper body segments (thorax, spine, and arm) during locomotion, further stressing the importance of assessing the upper body as individual rigid segments (Wu et al., 2005; Rau, Disselhorst-Klug, & Schmidt, 2000). However, these studies were predominantly driven by clinical interest, specifically considering gait pathologies and the rehabilitation and prevention of pathologies (Schache, Bennell, Blanch, & Wrigley, 1999; Lamoth, Beek, & Meijer, 2002; Nguyen & Baker, 2004; Sartor, Alderink, Greenwald, & Elders, 1999). When considering breast biomechanics and the female runner, examining the thorax and upper body limbs is of great value for understanding the role they play in driving the relative breast kinematics and vice versa.

Regardless of a runner's experience, individuals try to optimise their running kinematics in order to preserve energy (Williams, 1990). Any significant alterations in an individual's self-selected running kinematics may have an effect on running performance both biomechanically (Dugan & Bhat, 2005) and physiologically (Williams, Snow, & Agruss, 1991; Dallam, Wilber, Jadelis, Fletcher, & Romanov, 2005). Since the early work of Cavanagh and Williams (1982), many have attempted to develop a universally accepted description of optimum running kinematics for the most economical running style (Cavanagh & Williams, 1982; Williams & Cavanagh, 1987; Egbuonu, Cavanagh, & Miller, 1990; Sartor, Alderink, Greenwald, & Elders, 1999; Williams, 2007). However, due to the considerable difference in mechanical running form between individuals (Williams, 1993) the clarification of 'optimum' running kinematics is not easily identifiable; a given change that might be detrimental for one individual, may be another runner's optimum (Williams et al., 1991). For example, greater rotation of the thorax and a more vigorous arm swing will enable one individual to maintain a high velocity; however when considering the female athlete, these movement patterns may elicit greater magnitudes of relative breast kinematics, and lead to breast discomfort causing the participant to alter certain movement patterns or refrain from running at that velocity.

Two key considerations when examining running kinematics are energy conservation and the importance of understanding the synchronous relationship between upper and lower body segments (Novacheck, 1998; Williams & Cavanagh, 1987). The relationship between these factors is crucial when considering metabolic cost of exercise. Efficient energy transfer between segments during the running cycle ensures a reduced metabolic cost (Williams & Cavanagh, 1987). In order to maintain a constant velocity during running, counter-rotation occurs between the pelvis and thorax, this rotation enables the individual to maintain a constant step length and step frequency (Novacheck, 1998; Bruijn, Meijer, van Dieën, Kingma, & Lamoth, 2008). Novacheck (1998) suggested the rotation of the pelvis about the vertical axis functions as a pivot between leg swing and the counter-rotating thorax and arms during running. The degree of rotation of the upper body is therefore partially governed by this counter-rotation, and the velocity at which the individual is running.

The role of the pelvis in energy conservation has been emphasised by Schache, Bennell, Blanch, and Wrigley, (1999), suggesting that the degree of anteroposterior tilt at the pelvis should be minimised to conserve energy and maintain efficiency in running. Furthermore, Schache et al., (1999) proposed that lateral tilt of the pelvis is thought to play a role in shock absorption and in controlling the centre of gravity (CoG) at the stance phase of the gait cycle. The combined rotational movements occurring in the frontal plane about the anteroposterior axis between the thorax and pelvis, are thought to play a vital role in decoupling the intense lower extremity motion from the shoulder and head, allowing equilibrium to be maintained (Stokes, Andersson, & Forssberg, 1989; Schache et al., 1999). Therefore, any alteration in either of these two segments during running may impact upon an individual’s running performance.

Stokes et al., (1989) stated that it is important to note the inertial effects of the upper body limbs (e.g. arm swing) on the movements of the thorax (trunk), as this will significantly influence the kinematics of these segments during motion. Arm swing is a distinctive characteristic of walking and running, with the magnitude and frequency defined as compensatory and synchronous with the action of the legs (Hinrichs, 1990; Pontzer, Holloway, Raichlen, & Lieberman, 2009; Eke-Okoro, Gregoric, & Larsson, 1997). For example, during sprinting leg mechanics are forceful and explosive, the arms must move in large controlled flexion and extensions at the shoulder to support the increase in velocity. As the pace is slowed, the arms move through shorter arcs and swing across the thorax towards the midline of the body (Hinrichs, 1990). There are many benefits of arm swing reported in the literature; it has been shown that the arms serve to reduce fluctuations in mediolateral and anteroposterior displacement of the centre of mass during running, which may result in a reduction of energy cost (Hinrichs, 1990; Pontzer et al., 2009; Bruijn, Meijer, Beek, & van Dieën, 2010). In addition, arm swing and shoulder rotation counteract the torque seen in the upper body about the vertical axis that is imparted by the rotation of the pelvis to put the legs through their alternating patterns of stance and swing (Kuhtz-Busvhbeck & Jing, 2012; Hinrichs, 1990). Arm swing mechanics is an under investigated area, with little research published in comparison to the lower body limbs, particularly for female runners. Moreover, whilst the link between arm swing mechanics and thorax rotations has been documented, the influence of breast support on arm swing mechanics during running is unknown.

Accordingly, a detailed three-dimensional (3D) kinematic investigation of the upper and lower body segments is warranted on female runners. This area of research is lacking within the literature, a greater understanding of how breast support and breast movement may affect full body kinematics, and vice versa, would broaden the knowledge of breast kinematics and some of the unique issues of the female runner.

5.2 Aims and research hypotheses

The primary aim of this work was to explore the effect of breast support on peak orientation and ROM of upper and lower body segments during a five kilometre treadmill run. A secondary aim was to examine the relationship between thorax orientation and relative multiplanar breast displacement.

H_1 Breast support will significantly influence three-dimensional (3D) peak orientation and ROM of the upper and lower body segments during the five kilometre run.

H_2 Significant changes in 3D peak orientation and ROM of upper and lower body segments will be reported between the start and end of the five kilometre run.

H_3 A significant positive relationship will be reported between thorax kinematics and multiplanar breast kinematics.

5.3 Methods

5.3.1 Participants

The participants within the current chapter were the same as those previously described in chapter four, with additional data analysis detailed within this chapter.

5.3.2 Procedures

As described previously, participants visited the biomechanics laboratory on two separate occasions (up to 72 hours apart). Participants were required to perform two five kilometre treadmill run trials, once in a low breast support and once in a high breast support (Chapter 4, section 4.3.2). Participants were required to perform an additional bare-breasted (BB) treadmill run, but due to the discomfort associated with this condition the participants ran without breast support for only two minutes. A random number generator (http://www.random.org/) calculated an order for the breast support conditions for each participant to ensure order effects were reduced.

Participants conducted a five minute self-paced warm-up on a treadmill in the high breast support condition. Once completed, the participants were asked to select a comfortable running speed, which they felt they could maintain comfortably for the duration of the run, this ranged from 8.5 km·h^{-1} to 10.5 km·h^{-1}, with an average of 9.0 km·h^{-1} ± 1.0 km·h^{-1}. The same running speed was used in both breast support conditions. In order to carry out comparisons between breast support conditions, participants performed the bare-breasted run at the same speed as the two five kilometre run trials. Participants wore the same footwear and clothing for all trials.

Retro-reflective semi-spherical markers (marker diameter 12 mm) were positioned on the anatomical landmarks of the body detailed in Table 16 and presented in Figure 10 (Visual3D, segment models, C-motion). Three-dimensional coordinates of the markers were tracked by eight 200 Hz calibrated Oqus infrared cameras (Qualisys, Sweden), positioned around the treadmill. Prior to the dynamic running trials, participants were required to stand in the anatomical position for two seconds in all three breast support conditions. The purpose of these data was to provide a reference position for the 3D orientation of the segments during the dynamic running trials.

Cameras were set to record for the final ten seconds of the initial two minutes of the five kilometre run, and for ten second captures at each kilometre interval following this. The markers were identified and 3D data reconstructed in the Qualisys Track Manager (QTM) Software (Qualisys, Sweden).

Table 16. Proximal and distal end points of the anatomical segments on both sides of the body, with each segment defined by at least three non-collinear markers (Visual3D, C-motion).

Segment	Anatomical positions
Forearm segment	*Proximal end:* Medial and lateral condyles of the humerus at the radial-humeral junction *Distal end*: Lateral styloid process of the radius and medial styloid process of the ulna
Upper arm segment	*Proximal end:* Acromion process *Distal end:* Medial and lateral condyles of the humerus at the radial-humeral junction
Thorax segment	*Proximal end:* Suprasternal notch *Distal end*: Right and left anterioinferior aspect of the 10^{th} rib
Pelvis segment	Left and right anterior superior iliac spines (ASIS) Left and right posterior superior iliac spines (PSIS).
Thigh segment	*Proximal end:* Hip joint centre created in Visual3D from the Coda pelvis construction *Distal end:* Medial and lateral epicondyles of the femur
Shank segment	*Proximal end:* Medial and lateral epicondyle of the femur *Distal end:* Medial and lateral malleolus
Heel marker	Positioned on the heel on the participant’s left trainer.

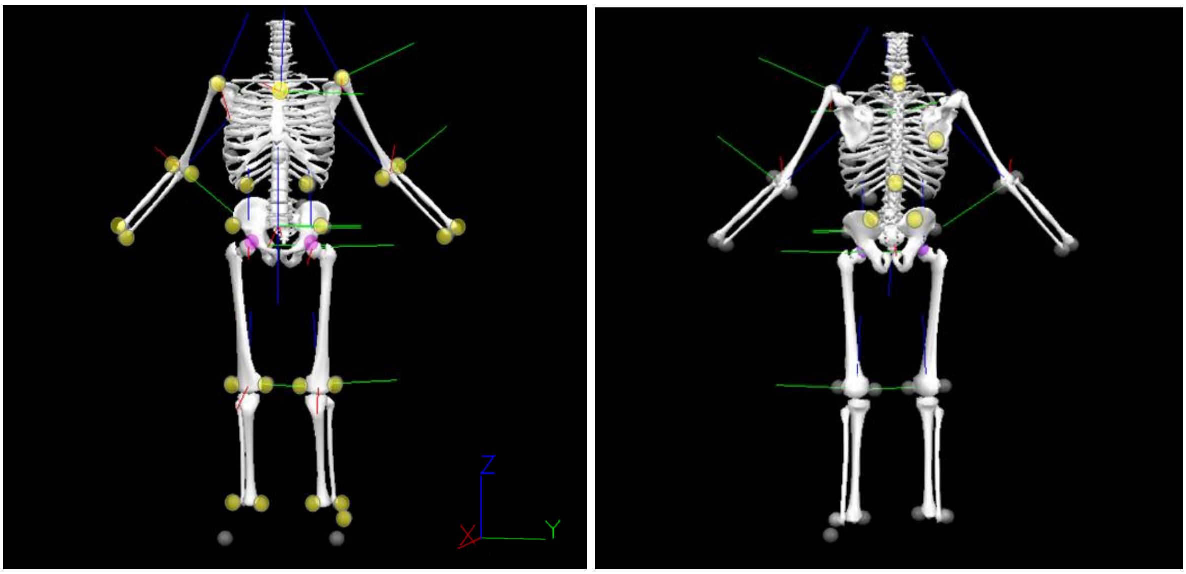

Figure 10. Anterior and posterior anatomical landmarks of the reflective markers and the Segment Coordinate Systems (SCS) and GCS axes, created in Visual3Ds model build. Orientation of axes of SCS and GCS are x (anteroposterior), y (mediolateral), and z (vertical).

Participants completed a subjective questionnaire after each run trial. The questionnaire asked participants to rate their perception of overall comfort during the run, with a scale ranging from 0 (comfortable, relaxed) to 10 (very uncomfortable, tense), with 5 rated as uncomfortable (Appendix B). This question referred to the participant's comfort throughout the two five kilometre run trials. An additional open ended question was included, which asked the participants; did you notice any differences in your running style during the run?

5.3.3 Data Processing

Three-dimensional coordinate data for each body marker were imported into a Fast Fourier Transform (FFT) program in MATLAB (MathWorks, UK). The frequency content of the data was assessed using a Fast Fourier transformation (FFT) program within MATLAB. The FFT shows the amplitude of the data plotted against the frequency component, enabling the identification of data that should be retained and the noise component that is attenuated (Winter, 1990). A cut-off frequency of 8 Hz was selected based upon this process.

Three-dimensional coordinates of all markers were then imported to Visual3D (c-motion, v4, Inc) in C3D file format, for further analysis. The static trial for each participant was used to create the upper and lower body segment coordinate systems (SCS) (Table 16)

(Cappozzo, Della Croce, Leardini, & Chiari, 2005). The orientation of the SCS axes followed the same right-hand rule orientation as the GCS, when the runner was in the anatomical position, z was defined as pointing along the distal to proximal segment axis (vertical), x was defined as the pointing to the front (anteroposterior), and y pointing to the left (mediolateral) (Figure 11) (Schache et al., 2001). The origin of each SCS was assumed at the proximal end of the segment, with the only exception being the pelvis segment (Figure 10).

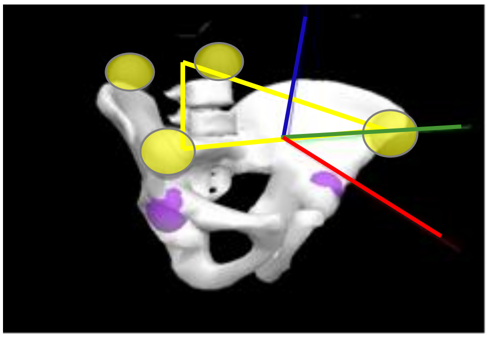

Figure 11. The CODA pelvis SCS convention within Visual3D (C-motion).

The CODA pelvis conventions were employed with the origin positioned at the midpoint of the left and right ASIS on the plane created between these four markers. The hip joint centres are created for the left and right side of the body at a set distance from the left and right ASIS, using the following equations; RHJC = (0.36*ASIS_distance, -0.19*ASIS_distance, -0.3*ASIS distance) and LHJC = (0.36*ASIS_distance, -0.19*ASIS_distance, -0.3*ASIS distance) (Visual3D, C-motion).

Assuming each of these segments to be rigid, the three-dimensional position and orientation (POSE) of each SCS could be determined at any given time using the marker coordinates relative to the GCS or another SCS. The markers were filtered within Visual3D, employing a fourth-order zero-phase shift Butterworth filter, at a cut off of 8 Hz. Once filtered, Cardan joint angles were calculated, this process enables the calculation of the segment rotation about three orthogonal axes of one particular segment with respect to a reference set of axes (Chiari, Croce, Leardini, Cappozzo, 2005).

Due to the orientation of the coordinate systems (GCS and SCS) within the current study the output for the cardan joint angles were influenced by the definition of the coordinate axes. The ISB recommendation and the most commonly implemented sequence of the Cardan joint angles is XYZ, with the X axes defined as the mediolateral axes, with flexion/extension occurring about this axis, Y axes defined as the anteroposterior with abduction/adduction occurring about this axis, and Z as vertical with axial rotation occurring about this final axis (Cappozzo, Della Croce, Leardini, & Chiari, 2005). The following Cardan sequence was employed within the current chapter; YXZ (Figure 12), with Y defined as the mediolateral axes (flexion/extension), X defined as the anteroposterior axes (abduction/adduction), and Z defined as the vertical (axial rotation). Although the Cardan sequence differs from that of the standard, the order of the corresponding anatomical movements is identical to previous publications (Tupling & Pierrynowski, 1987; Schache et al., 2001; Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2004).

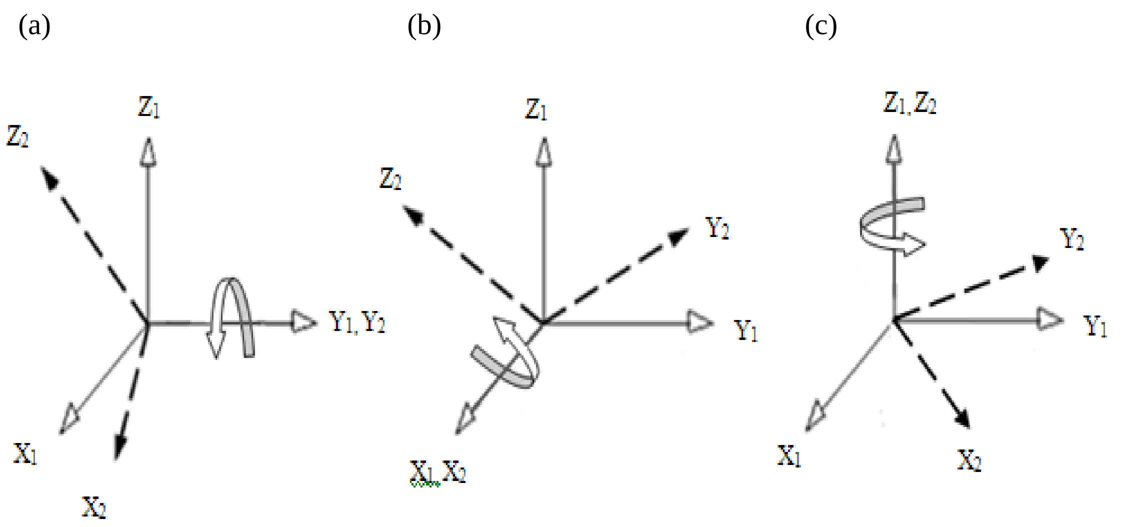

Figure 12. Cardan sequence of rotations about the (a) Y axis, (b) X axis, and (c) Z axis of the SCS. The initial orientation of the SCS axes (Y_1, X_1, Z_1) are illustrated in the figure above, axes are then rotated about the Cardan sequence (YXZ) to their second orientation (Y_2, X_2, Z_2).

These data were time normalised to each gait cycle at 1% intervals, with five gait cycles identified. Gait cycles were determined using the method detailed in chapter four; section 4.3.3 (Zeni, Richards, & Higginson, 2008). Peak orientation and range of motion (ROM) were calculated for all selected 3D joint angles. Peak orientation was calculated by identifying the maxima value within each plane of movement, for each segment during a gait cycle. Additionally, ROM was calculated by taking the minima orientation angle of the segment away from the maxima about each axes of rotation during each gait cycle, an example for peak knee flexion and ROM of the knee about the mediolateral axis during five gait cycles, are detailed in Figure 13.

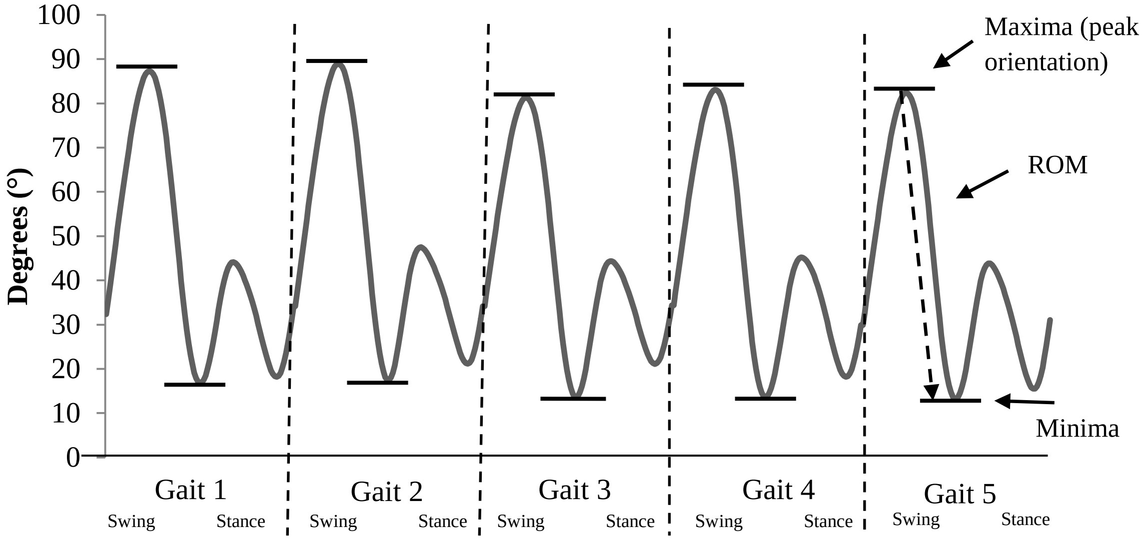

Figure 13. Example of knee flexion (n = 1) relative to the thigh segment over five gait cycles, with maxima and minima values identified for the calculation of peak orientation and joint ROM, for each gait cycle.

Step length (m) was calculated utilising the anteroposterior coordinates of the heel marker, the distance travelled was taken from the initial contact at heel strike to heel strike of the ipsilateral heel (Hunter, Marshall, & McNair, 2004).

5.3.4 Statistical Analyses

All data were checked for normality using the Kolmogorov-Smirnov and Shapiro-Wilk tests of normality, with normality assumed when $p > .05$. Homogeneity of variance was assessed using Mauchly's test of Sphericity, with homogenous data assumed when $p >$ .05. Data was accepted as normally distributed and displaying homogeneity, and therefore defined as parametric. Independent variables examined were breast support conditions and the intervals of the five kilometre run, and the dependent variables examined were the kinematic parameters. One-way repeated measures ANOVAs were performed to examine the effect of breast support conditions on the kinematic parameters for the two minute treadmill run data. Two-way repeated measures ANOVAs were performed to examine the main and interaction effects of breast support conditions and run distance on the running kinematic parameters. *Post-hoc* pairwise comparisons, with Bonferroni adjustment, were performed alongside the two-way repeated measures ANOVAs. Pearson's moment product correlations (r) were performed to explore the relationship between thorax rotations relative to the GCS and breast kinematics relative to the thorax, where a small relationship $\pm \leq .10$, medium relationship $\pm \leq .30$, and large relationship $\pm \geq .50$ (Field, 2009). The coefficient of determination (R^2) was calculated using the r value from the

Pearson correlations and converted to a percentage. This statistic details the percentage of variability shared by two variables and provides an indication of how much of each variable may account for the other. Effect size (η^2) and observed power (1-β) are presented to indicate the strength of the results, where small effect ≤ .10, medium effect ≤ .30, large effect ≤ .50, and a high power ≥ .80 (Field, 2009).

5.4 Results

5.4.1 Presentation of data

Orientation of the thorax segment is presented in each of the planes of motion; rotation about the anteroposterior axis in the frontal plane, rotation about the mediolateral axis in the sagittal plane, and rotation about the vertical axis in the transverse plane. The remaining upper and lower body segments are inter-segmental angles, with rotations about the three axes reported. The layout of the segment orientations within the results section is used for clarity, but it is acknowledged that the rotations may occur simultaneously during the running gait cycle. The orientation of the upper and lower body segments are presented as ensemble angle-time graphs across all participants, during the two minute data collection (Figures 14 and 15). In addition, peak orientation and ROM of the upper and lower body joint angles, over the two minute and five kilometre treadmill run, in three breast supports are presented in Tables 18 to 29.

5.4.2 Peak orientation and ROM of the body segments

The orientation of the thorax relative to the GCS in each plane of motion is graphically presented in Figure 14. The zero line represents the projected axes within each plane of motion (e.g. thorax pitch occurs within the sagittal plane, with vertical axis of the GCS defined as zero). The greatest ROM of the thorax occurs in the transverse plane, about the vertical axis (defined as thorax yaw within this chapter), and is greatest within the low breast support condition (27.4°) during the first two minutes of running. The magnitude of thorax roll and pitch relative to the GCS are relatively small in comparison to thorax yaw, 5.4° and 7.1° when averaged across breast support conditions during two minutes of running. This indicates that thorax yaw is the most dominant rotation of this segment during the running cycle, and therefore suggested that the vertical axis is the primary axis. The positive and negative peaks of thorax yaw occur at heel strike of each side of the body, with the peak rotation occurring in the opposite direction to the side of the body in contact with the ground, due to the counter rotation between the upper and lower body. Thorax pitch can be seen to cross the vertical with a positive peak representing a forward flexion and a negative peak representing a backwards extension. Within the current study, the gait cycles were identified with the left heel and therefore the initial negative peak occurs as the left heel comes into contact with the ground (breaking). The thorax then reaches a positive peak with the event of left foot stance, whilst the right leg is in swing.

The rotation of the pelvis about the anteroposterior axis (pelvic obliquity) relative to the thorax was shown to have the greatest ROM, 18.7° when averaged across breast support conditions. Pelvic obliquity occurred about the anteroposterior axis when one side of the body came into contact with the ground (Figure 14). The double peak (M-shape) seen is proposed as a stabilising mechanism of the pelvis over the leg in stance. During the stance phase of the gait cycle, as full foot contact progresses to toe off, the pelvis and thorax align along the vertical axis and are in-phase (zero rotation about the vertical axis) for this short time, as soon as the contact leg moves posteriorly into swing, these two segments are then out-of-phase and back to the counter-rotation relationship commonly reported during the running gait cycle. The average ROM in axial rotation of the pelvis across breast support conditions is 16.2°. Rotation of the pelvis about the mediolateral axis (pelvic tilt) peaks anteriorly during heel strike of each foot, and ranges over approximately 12°, however this is different between breast support conditions, with a smaller ROM seen in the bare-breasted condition compared to low and high breast supports.

The orientation of the upper arm was quantified relative to the thorax segment. The greatest rotation occurred about the mediolateral axis (29.1° averaged across the breast support conditions), and defined as extension within the current chapter (Figure 14). With the left heel strike defined as the start of one gait cycle, the right arm is shown to be in the smallest angle of extension relative to the thorax at this point. As the left foot progresses from full-foot contact to toe-off, the right upper arm is swung posteriorly in extension and reaches peak extension (42.6° average across breast support conditions) as the right heel comes into contact with the ground. During the gait cycle the upper arm is slightly abducted and internally rotated relative to the thorax.

The orientation of the forearm about the mediolateral axis, referred to as flexion, remains relatively stable during the gait cycle between the breast support conditions, with the angle ranging from 70° to 100°. As the left heel strikes the ground, initiating the gait cycle, the right forearm is brought medially towards the thorax and reaches peak adduction. However, as the left foot progresses through the stance phase of the gait cycle to toe-off the magnitude of right forearm adduction is reduced and is brought through a small degree of abduction as the right heel strikes the ground. The right arm is held in an externally rotated orientation throughout the entire gait cycle.

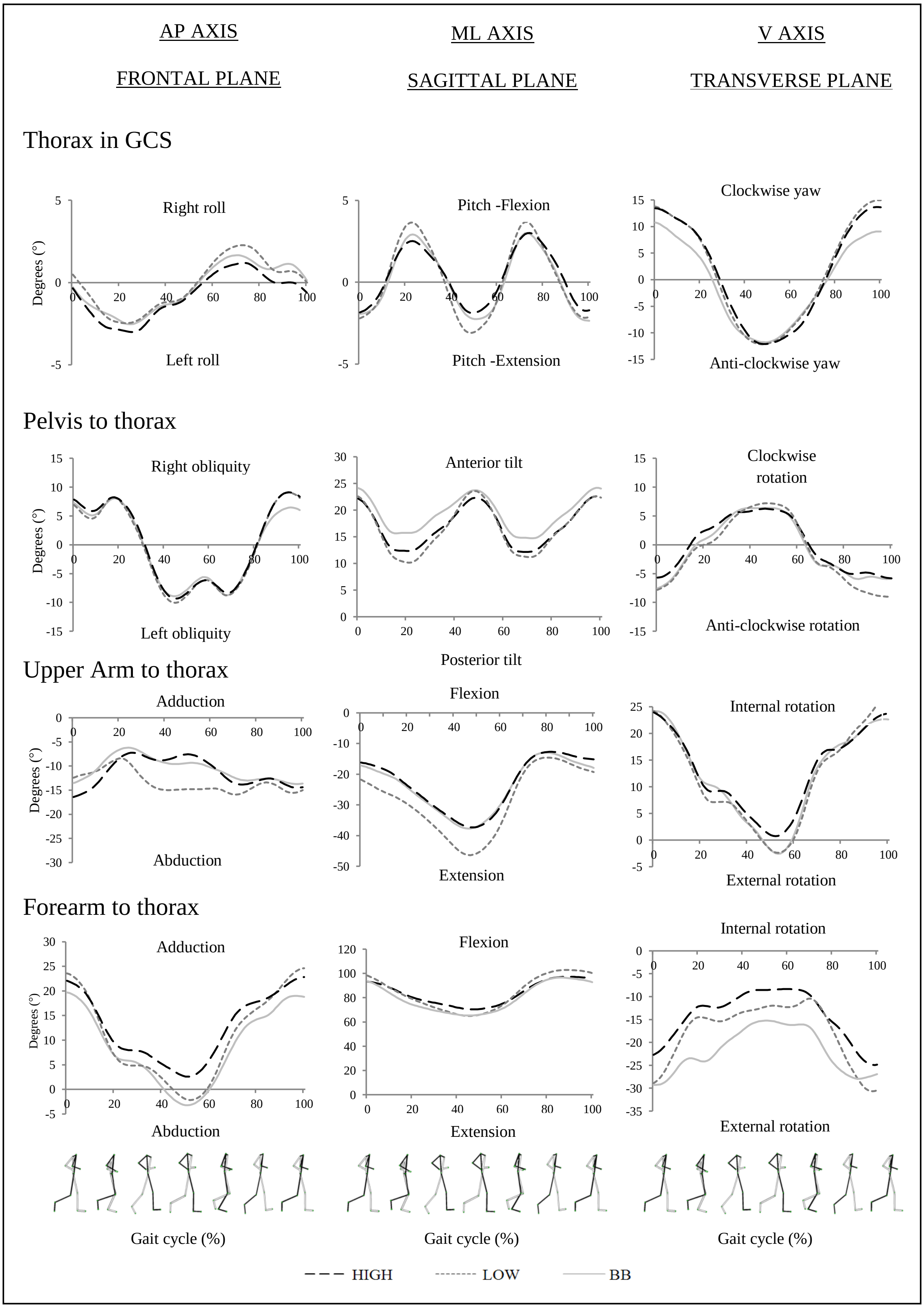

Figure 14. Mean orientation and ROM of upper body segments, averaged over five gait cycles during the first two minutes of the five kilometre run in three breast support conditions. Stick figure adapted from QTM output of bone segments (n = 10).

Thigh to pelvis abduction and adduction ranged from ~ -10° to 10° throughout the gait cycle (Figure 15). As the left side of the body is in stance the right thigh is abducted from the pelvis during the swing phase of the gait cycle. As the right heel comes into contact with the ground the thigh is then adducted towards the midline of the body and peaks at this moment in time. Rotation of the thigh about the mediolateral axis, referred to as flexion in the current chapter, peaks just before the right heel comes into contact with the ground (average peak of 48.2° across breast support conditions). The ROM of thigh flexion relative to the pelvis is approximately 50° in the three breast support conditions. Rotation of the thigh about the vertical axis, referred to as internal/external rotation within the current chapter, ranged from ~15° to -8°, with peak rotations occurring during the stance phase of the gait cycle, and was similar for all breast support conditions.

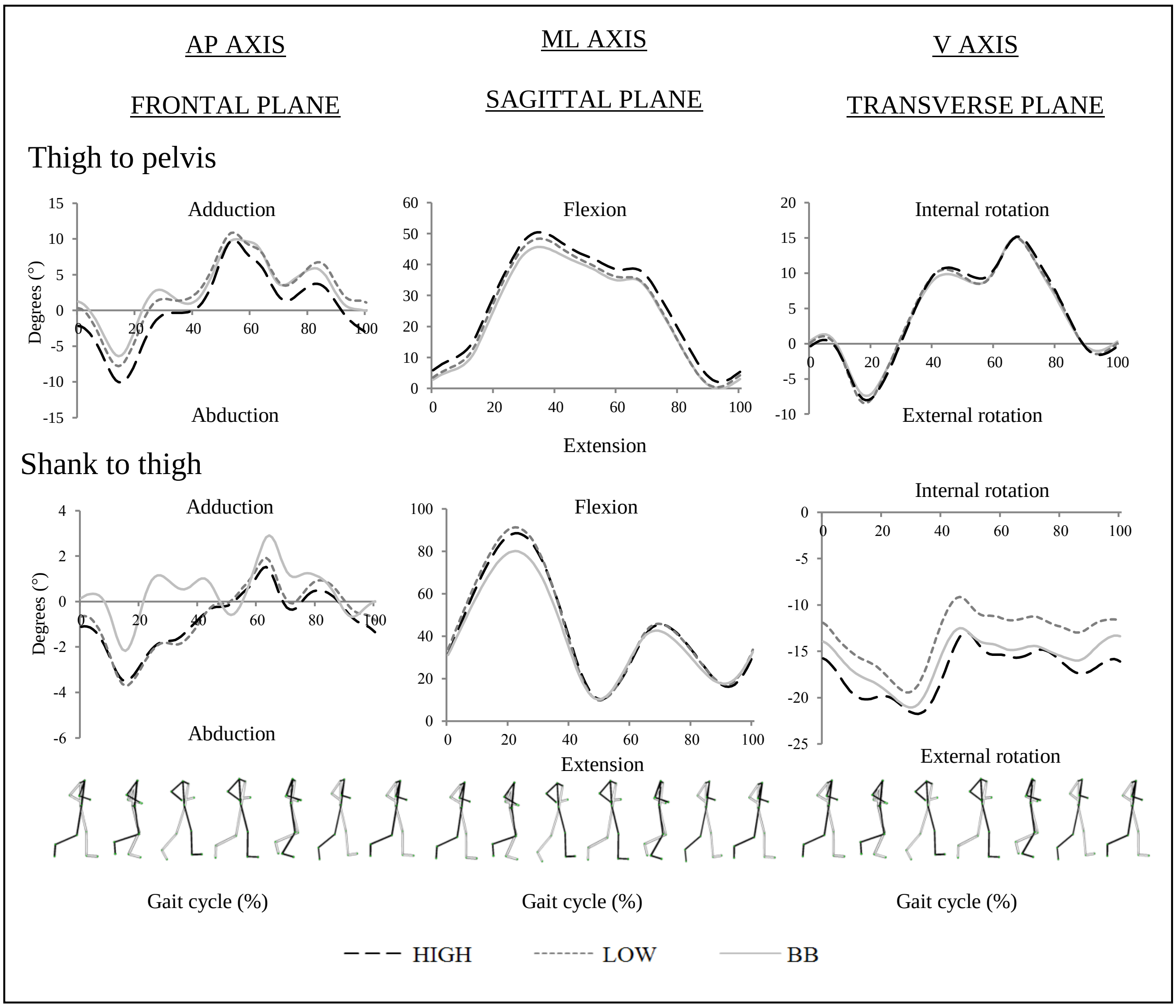

Figure 15. Mean orientation and ROM of lower body segments, averaged over five gait cycles, during the first two minutes of the five kilometre run, in the three breast support conditions (n = 10).

A very small degree of rotation occurs about the anteroposterior axis, referred to as shank adduction/abduction, during the gait cycle, and follows a similar pattern as the thigh segment. The greatest ROM at the shank relative to the thigh is rotation about the mediolateral axis, and is referred to as shank flexion, ranging from 10° to 90° during the gait cycle. Peak flexion of the right shank relative to the thigh (90°) occurs during the flight phase of the right side of the lower body, as the left foot is in full contact with the ground. At this time, peak flexion is approximately 10° smaller in the bare-breasted condition compared to the low and high breast supports. The shank segment is always in a degree of external rotation.

Step length was significantly shorter (0.03 m) in the bare-breasted compared to the high breast support condition, during the two minute data collection (Table 17). No differences in step length were reported between the intervals of the five kilometre run, within or between breast support conditions.

Table 17. Mean (± SD) step length (m) in three breast support conditions over five gait cycles of the first two minutes and the five kilometre run (*n* = 10).

INTERVALS	SUPPORT LEVEL		
	BB	LOW	HIGH
2 MINS	0.69 ± 0.05*[b]	0.71 ± 0.04	0.72 ± 0.05*[b]
1 KM		0.70 ± 0.04	0.71 ± 0.05
2 KM		0.70 ± 0.05	0.71 ± 0.05
3 KM		0.70 ± 0.04	0.71 ± 0.03
4 KM		0.70 ± 0.03	0.71 ± 0.04
5 KM		0.70 ± 0.04	0.71 ± 0.03

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Significant main effect of breast support level on step length during the two minute treadmill run ($F_{(2)}$ = 24.380, p = .001, η^2 = .730, 1-β = 1.000).

Table 18. Mean (± SD) peak orientation (°) of the thorax relative to the GCS over five gait cycles at each interval of the five kilometre run, in three breast support conditions (*n* = 10).

INTERVALS	THORAX ROLL			THORAX FLEXION			CLOCKWISE THORAX YAW		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	3.2 ± 2.2	3.0 ± 2.0	3.8 ± 2.1	4.2 ± 1.8	4.4 ± 1.8	4.1 ± 2.7	11.9 ± 5.7*[a]	15.9 ± 4.8*[a]	13.0 ± 4.0
1 KM		3.1 ± 2.0	3.8 ± 2.2		5.4 ± 2.7	4.2 ± 2.4		16.2 ± 6.2*[c]	13.0 ± 4.6*[c]
2 KM		3.2 ± 1.9	3.7 ± 1.8		5.6 ± 3.0	5.0 ± 2.9		17.2 ± 6.7*[c]	12.7 ± 4.9*[c]
3 KM		3.0 ± 2.0	4.1 ± 1.9		5.7 ± 2.8†	4.8 ± 2.7		16.0 ± 6.2*[c]	12.9 ± 4.5*[c]
4 KM		2.8 ± 1.7	3.3 ± 1.9		5.9 ± 3.0†	5.0 ± 3.0		16.0 ± 5.3*[c]	12.3 ± 3.3*[c]
5 KM		3.0 ± 1.4	3.5 ± 2.1		5.5 ± 2.5	5.4 ± 3.3		15.4 ± 4.7*[c]	13.6 ± 4.6*[c]
MEAN	3.2 ± 2.2	3.0 ± 0.1	3.7 ± 0.2	4.2 ± 1.8	5.5 ± 0.5	4.7 ± 0.5	11.9 ± 5.7	16.1 ± 0.6	12.9 ± 0.4

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Breast support significantly influenced peak thorax yaw during the two minute ($F_{(2)}$ = 6.732, p = .007, η^2 = .428, 1-β = .863) and the five kilometre run ($F_{(1)}$ = 9.856, p = .012, η^2 = .523, 1-β = .797), on average the high breast support reduced thorax yaw by 3.3° when compared to the low breast support. Peak flexion of the thorax significantly increased from the first two minutes to the third and fourth kilometre ($F_{(2.239)}$ = 7.157, p = .004, η^2 = .443, 1-β = .912), with increases of 1.3° and 1.5°, respectively.

Table 19. Mean (± SD) ROM (°) of the thorax relative to the GCS over five gait cycles at each interval of the five kilometre run, in three breast support conditions (*n* = 10).

INTERVALS	THORAX ROLL			THORAX PITCH			THORAX YAW		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	5.0 ± 0.8	5.7 ± 1.9	5.4 ± 1.9	6.5 ± 2.6	7.8 ± 1.6*[c]	6.8 ± 1.5*[c]	24.3 ± 3.8*[b]	27.4 ± 3.3*[c]	26.6 ± 3.1*[bc]
1 KM		5.7 ± 1.9	5.6 ± 1.7		7.7 ± 2.1*[c]	6.9 ± 1.9*[c]		27.4 ± 5.2	26.4 ± 5.2
2 KM		6.0 ± 2.1	5.7 ± 1.6		7.9 ± 1.9*[c]	6.6 ± 1.7*[c]		26.7 ± 4.9	26.5 ± 4.8
3 KM		5.8 ± 2.1	6.0 ± 1.7		7.8 ± 2.4	7.5 ± 1.9		26.5 ± 4.6	26.6 ± 5.3
4 KM		5.4 ± 1.4	5.5 ± 1.7		7.8 ± 2.3	6.8 ± 1.4		25.6 ± 4.0*[c]	28.5 ± 6.1*[c]
5 KM		5.2 ± 1.4	5.5 ± 2.3		7.4 ± 2.1	7.1 ± 1.7		25.7 ± 4.7*[c]	28.4 ± 5.0*[c]
MEAN	5.0 ± 0.8	5.6 ± 0.3	5.6 ± 0.2	6.5 ± 2.6	7.7 ± 0.2	7.1 ± 0.3	24.3 ± 3.8	26.6 ± 0.8	27.2 ± 1.0

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Breast support significantly influenced the ROM in thorax pitch during the five kilometre run ($F_{(1)} = 6.011$, $p = .037$, $\eta^2 = .400$, $1-\beta = .590$), with the high breast support significantly reducing the thorax pitch by 1.0° and 1.3°, respectively. The ROM of thorax yaw during the two minute ($F_{(2)} = 6.109$, $p = .009$, $\eta^2 = .404$, $1-\beta = .827$) and five kilometre run ($F_{(1)} = 6.550$, $p = .031$, $\eta^2 = .421$, $1-\beta = .629$) were significantly affected by the level of breast support worn, with the greater ROM in the low breast support compared to the high support during two minutes of running, however, during the fourth and fifth kilometre the high support elicited a greater ROM of thorax yaw when compared to the low breast support, on average a difference of 2.9° and 2.7°, respectively.

Table 20. Mean (± SD) peak orientation (°) of the pelvis relative to the thorax over five gait cycles at each interval of the five kilometre run, in three breast support conditions (n = 10).

INTERVALS	PELVIC OBLIQUITY			ANTERIOR PELVIC TILT			AXIAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	9.5 ± 2.9	10.6 ± 2.3	9.5 ± 1.3	22.8 ± 8.0	23.7 ± 6.3	24.7 ± 7.4	7.0 ± 5.4	10.1 ± 5.3*[c]	6.8 ± 4.2*[c]
1 KM		11.0 ± 1.5*[c]	9.5 ± 1.1*[c]		22.4 ± 6.4	24.7 ± 7.9		10.4 ± 5.0*[c]	7.2 ± 3.9*[c]
2 KM		10.8 ± 2.1*[c]	9.5 ± 1.2*[c]		21.8 ± 6.3	23.9 ± 8.0		11.8 ± 4.9*[c]	7.3 ± 3.8*[c]
3 KM		11.2 ± 1.8*[c]	9.1 ± 1.5*[c]		21.7 ± 6.7	24.3 ± 8.0		10.4 ± 4.9*[c]	7.2 ± 3.3*[c]
4 KM		10.7 ± 2.1*[c]	9.7 ± 1.4*[c]		21.3 ± 7.1	21.3 ± 8.1		10.5 ± 4.7*[c]	6.4 ± 3.3*[c]
5 KM		10.8 ± 2.6	9.1 ± 1.2		23.5 ± 9.0	23.4 ± 8.1		9.6 ± 3.8	7.9 ± 4.2
MEAN	9.5 ± 2.9	10.8 ± 0.3	9.4 ± 0.3	22.8 ± 8.0	22.4 ± 1.0	23.7 ± 1.3	7.0 ± 5.4	10.5 ± 0.7	7.1 ± 0.5

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Breast support significantly affected peak pelvic obliquity during the five kilometre run ($F_{(1.000)}$ = 10.247, p = .011, η^2 = .532, 1-β = .812), with the high support reducing peak obliquity by 1.5° on average. Peak pelvic axial rotation was also different between breast support conditions during the two minute run ($F_{(2)}$ = 6.025, p = .010, η^2 = .401, 1-β = .821) and the five kilometre run ($F_{(1)}$ = 5.950, p = .037, η^2 = .398, 1-β = .585), with the high support reducing the peak rotation by 3.8° on average when compared to the low breast support.

Table 21. Mean (± SD) ROM (°) of the pelvis relative to the thorax over five gait cycles at each interval of the five kilometre run, in three breast support conditions (*n* = 10).

INTERVALS	PELVIC OBLIQUITY			PELVIC TILT			AXIAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	17.0 ± 4.2*[ab]	19.1 ± 3.2*[a]	20.1 ± 3.2*[b]	11.2 ± 3.2*[a]	13.4 ± 2.7*[a]	12.2 ± 2.7	14.7 ± 3.1*[a]	18.8 ± 4.7*[a]	15.4 ± 3.1
1 KM		20.5 ± 3.0	19.1 ± 2.6		13.7 ± 1.9	12.7 ± 2.7		18.3 ± 4.0*[c]	15.3 ± 3.1*[c]
2 KM		20.5 ± 2.9	19.5 ± 2.9		13.6 ± 2.6	12.1 ± 2.2		19.5 ± 4.3*[c]	15.5 ± 2.3*[c]
3 KM		20.5 ± 3.1	19.1 ± 3.0		13.9 ± 2.5	12.6 ± 2.1		18.4 ± 3.8	15.7 ± 3.0
4 KM		18.8 ± 3.1	18.9 ± 3.5		13.8 ± 3.2	12.4 ± 2.2		18.2 ± 4.3*[c]	15.4 ± 2.2*[c]
5 KM		19.1 ± 2.9	18.8 ± 2.8		12.9 ± 2.6	12.8 ± 2.1		17.3 ± 3.1	16.0 ± 3.1
MEAN	17.0 ± 4.2	19.8 ± 0.8	19.3 ± 0.5	11.2 ± 3.2	13.6 ± 0.4	12.5 ± 0.3	14.7 ± 3.1	18.4 ± 0.7	15.6 ± 0.3

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. During the two minute run, the ROM of pelvic obliquity ($F_{(2)}$ = 12.195, *p* = .001, η^2 = .575, 1-β = .987), pelvic tilt ($F_{(2)}$ = 4.586, *p* = .025, η^2 = .338, 1-β = .702), and axial rotation of the pelvis ($F_{(2)}$ = 27.789, *p* = .001, η^2 = .755, 1-β = 1.000) were significantly different dependent upon the breast support worn. Range of motion in axial rotation of the pelvis was significantly smaller in the high breast support during the five kilometre run ($F_{(1)}$ = 7.066, *p* = .026, η^2 = .440, 1-β = .659), at the first, second and fourth kilometre, a reduction of 3.3° on average.

Table 22. Mean (± SD) peak orientation (°) of the upper-arm relative to the thorax over five gait cycles at each interval of the five kilometre run, in three breast support conditions (*n* = 10).

INTERVALS	ABDUCTION			EXTENSION			INTERNAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	14.7 ± 4.1	16.7 ± 3.4	17.0 ± 2.7	44.1 ± 5.6*[b]	43.3 ± 6.3	40.5 ± 7.6*[b]	24.7 ± 9.4	24.9 ± 9.1	26.4 ± 5.3
1 KM		18.0 ± 3.6	17.7 ± 3.0		44.9 ± 5.0	41.8 ± 6.6		28.8 ± 9.7	26.3 ± 7.8
2 KM		18.4 ± 3.4	18.2 ± 2.9		44.3 ± 6.3	40.6 ± 6.4		29.5 ± 12.6	32.2 ± 11.9
3 KM		17.6 ± 3.1	18.7 ± 3.2		42.3 ± 5.4	41.6 ± 6.5		28.8 ± 14.2	29.4 ± 12.4
4 KM		17.7 ± 2.8	18.4 ± 3.7		41.6 ± 5.9	41.6 ± 6.5		31.2 ± 15.6	33.5 ± 10.6
5 KM		17.5 ± 2.6	18.4 ± 3.8		40.0 ± 5.8	41.7 ± 5.3		28.6 ± 2.1	26.7 ± 11.3
MEAN	14.7 ± 4.1	17.6 ± 0.6	18.1 ± 0.6	44.1 ± 5.6	42.7 ± 1.8	41.1 ± 0.7	24.7 ± 9.4	28.6 ± 2.1	29.7 ± 3.1

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Breast support significantly affected peak upper arm extension during the two minute run ($F_{(2)} = 3.236$, $p = .043$, $\eta^2 = .264$, $1\text{-}\beta = .542$), with the high breast support reducing peak extension by 3.6°.

Table 23. Mean (± SD) ROM (°) of the upper-arm relative to the thorax over five gait cycles at each interval of the five kilometre run, in three breast support conditions (*n* = 10).

INTERVALS	ABDUCTION			EXTENSION			INTERNAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	7.1 ± 1.5*[b]	9.7 ± 3.2	11.3 ± 3.3*[b]	27.0 ± 8.7	32.6 ± 5.5*[c]	26.2 ± 3.9*[c]	27.2 ± 12.0	30.6 ± 9.7	26.3 ± 5.3
1 KM		11.0 ± 3.6	11.1 ± 1.9		35.1 ± 7.3*[c]	27.8 ± 4.8*[c]		33.7 ± 8.5	28.3 ± 7.0
2 KM		11.6 ± 3.5	10.4 ± 3.2		36.7 ± 8.3*[c]	28.6 ± 7.8*[c]		34.5 ± 8.4	31.0 ± 10.6
3 KM		11.8 ± 3.7	10.8 ± 2.5		34.1 ± 11.1	31.4 ± 8.3		34.2 ± 9.6	30.0 ± 10.4
4 KM		11.0 ± 2.7	10.5 ± 1.3		32.8 ± 8.6	34.8 ± 10.2		31.0 ± 11.3	31.0 ± 9.0
5 KM		10.8 ± 3.2	11.0 ± 3.0		30.9 ± 7.3	30.1 ± 9.4		28.1 ± 7.7	30.3 ± 9.4
MEAN	7.1 ± 1.5	11.0 ± 0.7	10.9 ± 0.4	27.0 ± 8.7	33.7 ± 2.0	29.8 ± 3.0	27.2 ± 12.0	32.0 ± 2.5	29.5 ± 1.8

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. The ROM in upper-arm abduction was significantly greater in the high level of breast support when compared to the bare-breasted condition ($F_{(2)}$ = 7.879, p = .003, η^2 = .467, 1-β = .913) during the two minute run. The ROM in upper-arm extension during the five kilometre run distance ($F_{(1)}$ = 16.578, p = .003, η^2 = .648, 1-β = .950), was reduced in the high level of breast support compared to the low level of support, with an average reduction of 7.3°.

Table 24. Mean (± SD) peak orientation (°) of the forearm relative to the thorax over five gait cycles at each interval of the five kilometre run, in three breast support conditions (*n* = 10).

INTERVALS	ADDUCTION			FLEXION			EXTERNAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	23.4 ± 9.9	27.0 ± 4.6	25.9 ± 5.0	102.1 ± 12.7	105.1 ± 12.7*[c]	102.4 ± 17.6*[c]	26.7 ± 7.7	29.6 ± 7.0	24.8 ± 7.7
1 KM		29.6 ± 7.2	26.9 ± 5.7		109.3 ± 14.3*[c]	102.7 ± 12.5*[c]		26.5 ± 7.9	21.3 ± 9.0
2 KM		30.0 ± 6.1	30.7 ± 8.9		111.1 ± 15.5*[c]	107.3 ± 12.9*[c]		27.2 ± 9.8	21.8 ± 8.1
3 KM		28.6 ± 6.5*[c]	31.5 ± 6.8*[c]		112.5 ± 17.5*[c]	109.0 ± 11.7*[c]		27.8 ± 9.1	24.1 ± 8.5
4 KM		31.8 ± 10.0*[c]	33.4 ± 8.5*[c]		111.6 ± 15.9*[c]	110.7 ± 11.2*[c]		29.4 ± 7.1	26.4 ± 7.1
5 KM		25.7 ± 12.5*[c]	28.1 ± 6.9*[c]		111.9 ± 18.3*[c]	107.6 ± 12.6*[c]		28.3 ± 8.5	24.7 ± 5.6
MEAN	23.4 ± 9.9	28.8 ± 2.2	29.4 ± 2.4	102.1 ± 12.7	110.3 ± 2.7	106.6 ± 3.6	26.7 ± 7.7	28.2 ± 1.2	23.8 ± 1.9

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Peak forearm adduction was significantly greater in the high level of breast support during the five kilometre run ($F_{(1)}$ = 2.774, p = .029, η^2 = .236, 1-β = .780) when compared to the low breast support, a difference of 2.3°. Peak forearm flexion was significantly smaller in the high breast support compared to the low breast support at every measured interval of the five kilometre run ($F_{(1)}$ = 67.423, p = .001, η^2 = .882, 1-β = 1.000), a reduction of 3.6° on average.

Table 25. Mean (± SD) ROM (°) of the forearm relative to the thorax over five gait cycles at each interval of the five kilometre run, in three breast support conditions (n = 10).

INTERVALS	ADDUCTION			FLEXION			EXTERNAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	24.0 ± 11.4	27.3 ± 8.8	23.6 ± 8.2	33.4 ± 13.5	36.3 ± 9.9	32.0 ± 11.0	15.1 ± 5.1	21.0 ± 10.7	15.5 ± 5.3
1 KM		30.8 ± 8.2	25.8 ± 6.4		40.2 ± 10.6*[c]	31.9 ± 7.1*[c]		19.1 ± 7.7	17.2 ± 4.8
2 KM		31.2 ± 6.9	26.2 ± 10.3		41.6 ± 13.0*[c]	33.2 ± 10.1*[c]		22.7 ± 10.7	16.6 ± 5.7
3 KM		30.3 ± 7.8	28.7 ± 9.7		38.9 ± 17.7	35.1 ± 12.4		22.8 ± 12.0	18.8 ± 8.9
4 KM		27.6 ± 9.1	27.0 ± 8.5		38.0 ± 15.3	36.8 ± 11.1		18.7 ± 8.1	19.3 ± 7.9
5 KM		28.2 ± 9.0	27.4 ± 8.6		36.8 ± 14.2	32.9 ± 13.1		19.9 ± 9.1	18.9 ± 8.8
MEAN	24.0 ± 11.4	29.2 ± 1.7	26.5 ± 1.7	33.4 ± 13.5	38.6 ± 2.0	33.7 ± 1.9	15.1 ± 5.1	20.7 ± 1.8	17.7 ± 1.5

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Breast support significantly influenced the ROM in forearm flexion ($F_{(1)}$ = 10.272, p = .011, η^2 = .533, 1-β = .813) during the five kilometre run, with a reduction in the ROM of forearm flexion in the high breast support, a difference of 8.4°.

Table 26. Mean (± SD) peak orientation (°) of the thigh relative to the pelvis over five gait cycles at each interval of the five kilometre run, in three breast support conditions (n = 10).

INTERVALS	ADDUCTION			FLEXION			INTERNAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	15.4 ± 2.0	15.7 ± 2.4	15.3 ± 2.5	48.0 ± 10.4	48.2 ± 8.5	49.8 ± 8.6	11.4 ± 5.6	11.0 ± 7.5	11.1 ± 5.7
1 KM		16.2 ± 2.1	15.0 ± 2.1		47.3 ± 7.6	49.6 ± 7.5		11.5 ± 8.2	11.0 ± 4.8
2 KM		16.2 ± 2.7	15.3 ± 2.6		47.7 ± 8.6	50.1 ± 8.0		11..9 ± 8.4	11.5 ± 5.0
3 KM		16.0 ± 2.6	15.6 ± 2.0		47.7 ± 8.2	48.8 ± 7.9		11.8 ± 7.9	14.8 ± 9.0
4 KM		16.1 ± 2.7	15.1 ± 2.4		47.8 ± 7.9	48.4 ± 6.8		11.5 ± 8.5	9.9 ± 5.1
5 KM		16.4 ± 2.8	14. 4 ± 2.2		48.3 ± 7.4	46.9 ± 12.3		11.4 ± 8.0	11.3 ± 5.1
MEAN	15.4 ± 2.0	16.1 ± 0.2	15.1 ± 0.4	48.0 ± 10.4	47.8 ± 0.4	48. 9 ± 1..2	11.4 ± 5.6	11.5 ± 0.3	11.6 ± 1.6

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

Table 27. Mean (± SD) ROM (°) of the thigh relative to the pelvis over five gait cycles at each interval of the five kilometre run, in three breast support conditions (*n* = 10).

INTERVALS	ADDUCTION/ABDUCTION			FLEXION			INT/EXT ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	21.5 ± 4.0	23.4 ± 2.7	22.1 ± 1.9	46.8 ± 6.0	47.5 ± 5.7	47.6 ± 5.2	19.4 ± 2.9	19.9 ± 4.3	20.8 ± 5.0
1 KM		23.2 ± 2.3*[c]	21.9 ± 2.5*[c]		47.1 ± 5.1	47.4 ± 4.7		21.0 ± 4.4	21.8 ± 5.4
2 KM		23.1 ± 2.4	22.1 ± 2.6		47.2 ± 5.7	48.1 ± 5.7		21.4 ± 3.9	21.7 ± 4.9
3 KM		22.6 ± 2.5	22.4 ± 2.8		47.2 ± 5.3	47.4 ± 3.7		20.8 ± 4.4	22.0 ± 5.2
4 KM		22.9 ± 2.6	22.4 ± 3.3		47.0 ± 4.7	46.9 ± 3.5		21.0 ± 4.7	21.1 ± 5.1
5 KM		23.7 ± 2.9*[c]	21.7 ± 2.3*[c]		45.2 ± 6.6	47.7 ± 4.5		21.6 ± 4.7	21.8 ± 4.9
MEAN	21.5 ± 4.0	23.2 ± 0.4	22.1 ± 0.3	46.8 ± 6.0	46.9 ± 0.8	47.5 ± 0.4	19.4 ± 2.9	21.0 ± 0.6	21.5 ± 0.5

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Breast support significantly influenced the ROM in adduction and abduction of the thigh relative to the pelvis ($F_{(1)}$ = 10.758, p = .010, η^2 = .544, 1-β = .830) during the five kilometre run, with the high support reducing the ROM on average by 1.7°.

Table 28. Mean (± SD) peak orientation (°) of the shank relative to the thigh over five gait cycles at each interval of the five kilometre run, in three breast support conditions (n = 10).

INTERVALS	ABDUCTION			FLEXION			EXTERNAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	3.2 ± 2.1	3.9 ± 3.3	4.4 ± 2.3	82.5 ± 6.1*[ab]	91.5 ± 8.8*[a]	90.1 ± 8.4*[b]	13.0 ± 4.6	14.6 ± 4.2	14.2 ± 6.0
1 KM		3.8 ± 3.2	4.3 ± 2.0		91.6 ± 9.5	91.4 ± 8.5		15.3 ± 4.6	14.5 ± 5.5
2 KM		3.8 ± 3.1	4.3 ± 1.7		92.2 ±10.3	90.5 ± 7.3		15.7 ± 4.8	14.9 ± 5.5
3 KM		3.7 ± 2.9	5.0 ± 2.5		90.7 ± 8.8	90.0 ± 6.4		16.0 ± 4.7	14.4 ± 5.5
4 KM		4.7 ± 3.4	5.2 ± 2.1		89.9 ± 8.8	87.5 ± 6.1		16.1 ± 4.9	15.6 ± 5.9
5 KM		4.8 ± 3.7	4.5 ± 1.6		90.4 ± 8.4	87.3 ± 10.5		15.4 ± 4.9	14.3 ± 5.8
MEAN	3.2 ± 2.1	4.1 ± 0.5	4.6 ± 0.4	82.5 ± 6.1	91.0 ± 0.9	89.5 ± 1.7	13.0 ± 4.6	15.5 ± 0.6	14.7 ± 0.5

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Peak shank flexion was significantly smaller in the bare-breasted condition compared to the low and high breast support during the two minute run ($F_{(2)}$ = 19.248, p = .001, η^2 = .681, 1-β = 1.000), a difference of 7.6°.

Table 29. Mean (± SD) ROM (°) of the shank relative to the thigh over five gait cycles at each interval of the five kilometre run, in three breast support conditions (n = 10).

INTERVALS	ADDUCTION/ABDUCTION			FLEXION			EXTERNAL ROTATION		
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH
2 MINS	5.4 ± 1.9	6.0 ± 1.9	5.9 ± 2.5	71.5 ± 6.2*[b]	80.7 ± 8.3	81.4 ± 7.2*[b]	7.8 ± 2.9	10.2 ± 5.0	9.7 ± 4.5
1 KM		5.8 ± 1.9	6.2 ± 2.2		81.4 ± 8.8	81.2 ± 8.3		10.4 ± 4.6	9.2 ± 4.6
2 KM		6.0 ± 2.1	6.2 ± 2.0		81.4 ± 9.7	80.1 ± 7.4		10.8 ± 4.7	8.3 ± 2.3
3 KM		6.0 ± 1.6	6.3 ± 2.1		79.4 ± 8.1	79.7 ± 6.9		9.7 ± 3.8	8.5 ± 1.6
4 KM		6.2 ± 1.7	6.7 ± 1.8		78.0 ± 8.5	77.0 ± 7.8		10.5 ± 4.4	9.5 ± 3.2
5 KM		7.3 ± 2.5	6.4 ± 1.9		76.5 ± 7.1	79.8 ± 8.1		11.7 ± 5.8	8.5 ± 2.2
MEAN	5.4 ± 1.9	6.2 ± 0.5	6.3 ± 0.3	71.5 ± 6.2	79.6 ± 2.0	79.9 ± 1.6	7.8 ± 2.9	10.6 ± 0.7	9.0 ± 0.6

*[a] Denotes a significant difference between the BB and low breast support conditions.
*[b] Denotes a significant difference between the BB and high breast support conditions.
*[c] Denotes a significant difference between the low and high breast support conditions.
† Denotes a significant difference between the first two minutes and the intervals of the five kilometre run.

N.B. Range of motion in shank flexion was significantly greater in the high breast support compared to the low level support ($F_{(1.201)}$ = 30.370, p = .001, η^2 = .771, 1-β = 1.000) during two minute running, a difference of 9.9°.

5.4.3 Relationship between thorax kinematics and multiplanar breast kinematics

Peak orientation of the thorax did not correlate to multiplanar breast kinematics, however, ROM in thorax yaw demonstrated a significant negative relationship with both mediolateral and vertical breast displacement (Table 30).

Table 30. Pearson correlation coefficients between the thorax ROM relative to the GCS and multiplanar breast displacement relative to the thorax in the three levels of breast support (bare-breasted, low and high), during the first two minutes of the five kilometre run (n = 10).

Thorax ROM relative to the GCS	Multiplanar breast displacement relative to the thorax		
	A-P displacement	M-L displacement	V displacement
Thorax roll	r = .043, p = .411	r = -.080, p = .674	r = -.181, p = .340
Thorax pitch	r = .082, p = .667	r = -.056, p = .770	r = -.098, p = .605
Thorax yaw	r = -.264, p = .083	r = -.382, p = .037*	r = -.697, p = .001*

*Denotes a significant correlation, where $p < .05$.

By calculating the coefficient of determination (R^2) (Table 31) and converting the value to a percentage (r^2*100), the percentage of variance shared by these two variables can be calculated. From these data, it can be seen that a large percentage (almost 50%) of the variance in vertical breast displacement can be accounted for by the ROM in thorax yaw.

Table 31. Coefficient of determination (R^2) presented as a percentage (%).

Correlation variables	Coefficient of determination (R^2)
Thorax yaw and M/L breast displacement	15%
Thorax yaw and V breast displacement	49%

5.4.4 Perceived comfort scores

Perceived overall comfort scores recorded following the five kilometre run in the low and high breast supports are detailed in Figure 16. Participants rated the low breast support condition as uncomfortable, whereas the participants rated running in the high breast support condition closer to comfortable, relaxed.

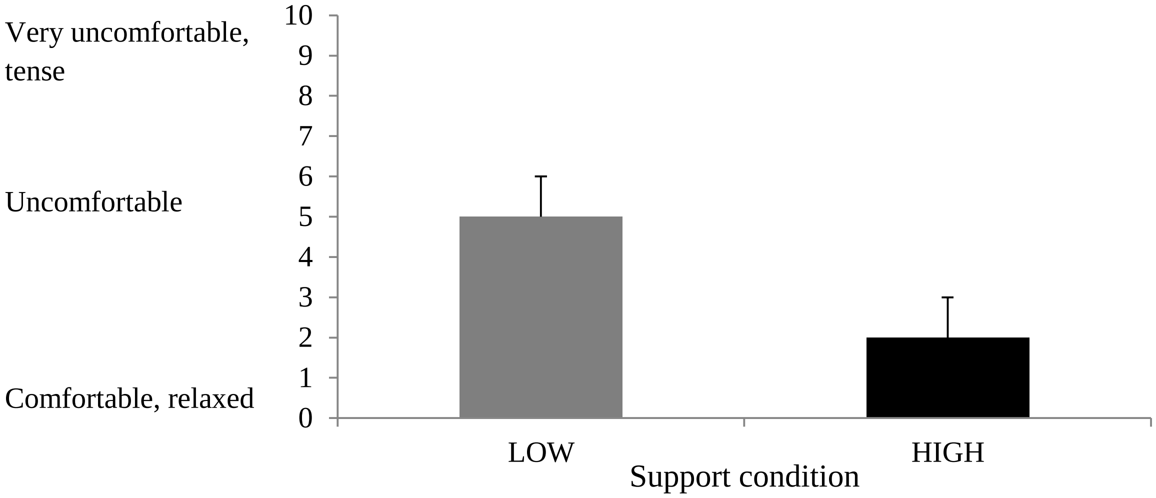

Figure 16. Mean (SD) overall comfort ratings during the five kilometre run, in the low and high breast support conditions (*n*=10).

Participants provided details of any perceived differences in their running style during the five kilometre run, these are presented in Table 32. Only seven participants provided comments for this question.

Table 32. Participant's subjective comments on perception of running style over the five kilometre run, in each breast support condition.

LOW SUPPORT	HIGH SUPPORT
Participant one: At times I felt like I took longer strides.	
Participant three: Tried to place feet down lighter to avoid breast pain.	
Participant five: felt slightly different, mainly to accommodate markers.	
Participant six: Slight difference to my normal running style.	
Participant seven: leant forward more, shoulders felt really far forwards. Back felt really tense.	**Participant seven**: More upright, back was straighter.
Participant eight: Felt a bit awkward in my running style at first.	**Participant eight**: felt sluggish, generally tired at end.
Participant ten: I felt a bit more rigid.	**Participant nine**: my sports bra isn't as supportive as this, and so felt more comfortable.

5.4.5 Effect sizes, power, and variance

Effect sizes and power were calculated alongside the one-way and two-way repeated measures ANOVAs reported in this chapter. Of the significant differences reported the effect sizes were all defined as medium (> .30) to large effects (> .05), with only two results falling below an effect size of .44. The power calculations were all deemed as high (> .80) except three values which were all greater than .59. With medium to large effect sizes and a majority of high power presented for the differences reported in this chapter, it is assumed that these differences are meaningful and the sample (*n* =10) is large enough to determine the effect of breast support on joint angles during treadmill running.

The within-participant variance (C_v%) in joint angles was calculated over five gait cycles at each kilometre interval within each breast support condition, for all participants and then averaged across this sample (*n* =10) and across the five kilometre intervals (Table 33). Within-participant variance was greatest in the upper-arm segment (average = 17%), specifically rotation about the vertical axis across the three levels of breast support. The flexion of the shank relative to the thigh demonstrated the smallest magnitude of variance, eliciting only 2% in all breast support conditions. However, the thigh segment demonstrated the smallest overall within-participant variance (average = 7%).

Table 33. Within-participant variance (C_v%) in joint angles averaged over the five kilometre run in the low and high breast support conditions (*n* = 10).

SEGMENTS	AP AXIS			ML AXIS			V AXIS			MEAN
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH	
Thorax to GCS	16	16	16	15	15	15	9	8	8	13
Pelvis to thorax	9	7	8	10	10	10	10	11	10	9
Upper-arm to thorax	28	23	26	13	10	11	13	16	14	17
Forearm to thorax	14	15	14	15	14	15	22	25	23	14
Thigh to pelvis	8	8	8	4	4	4	10	10	10	7
Shank to thigh	10	10	10	2	2	2	14	14	14	9
MEAN	14	13	14	10	9	10	13	14	13	12

Within-participant variance was noticeably smaller in the lower body segments than the upper body segments. Within-participant variance in step length was 2% across all breast support conditions, demonstrating consistent values over the five gait cycles analysed.

Between-participant variance (C_v%) in joint angles was calculated over each kilometre interval within each breast support condition, across all participants in the sample (n = 10) (Table 34). The greatest magnitude of between-participant variance was reported in the forearm segment (average = 38%), with the greatest variance shown in abduction/adduction within the bare-breasted condition. The smallest magnitude of between-participant variance was reported in the thigh segment relative to the pelvis (average = 15%). Shank flexion relative to the thigh elicited a low between-participant variance across the three levels of breast support. Between-participant variance in step length was 6% across breast support conditions, demonstrating a low magnitude (<10%) of variance across participants.

Table 34. Between-participant variance (C_v%) in joint angles averaged over the five kilometre run in the low and high breast support conditions (n = 10).

SEGMENTS	AP AXIS			ML AXIS			V AXIS			MEAN
	BB	LOW	HIGH	BB	LOW	HIGH	BB	LOW	HIGH	
Thorax to GCS	16	32	32	39	27	24	16	17	18	25
Pelvis to thorax	25	15	16	27	18	19	21	22	18	20
Upper-arm to thorax	21	30	23	32	24	24	44	29	29	28
Forearm to thorax	48	29	33	41	35	32	34	47	39	38
Thigh to pelvis	18	11	11	13	12	10	15	21	24	15
Shank to thigh	35	31	33	9	10	10	37	44	33	27
MEAN	27	25	25	27	21	20	28	30	27	26

5.5 Discussion

The work presented within this chapter is the first to provide an in-depth description of female running kinematics in different breast support conditions, over short or prolonged treadmill running. The key findings of the current study indicate that firstly; peak orientation and ROM of upper and lower body segments are different between breast support conditions, during both short (two minutes) and prolonged (five kilometre) treadmill running. Secondly, peak flexion of the thorax was the only kinematic variable to increase over the distance of the five kilometre run within the low breast support condition. Finally, the ROM in thorax yaw demonstrated significant relationships with breast displacement during treadmill running.

It is suggested that the differences in peak orientation and ROM of the upper and lower body segments reported in the bare-breasted conditions, when compared to the kinematics reported in the low and high breast support conditions, were driven by a protective response, driven by the significantly greater magnitude of breast kinematics and the associated breast pain (data reported in chapter four, section 4.4.7) when running without external support. This suggestion is supported by the participant's comments on their elected running style in the bare-breasted condition. The participants stated that they acknowledged a change in running kinematics, and that those alterations were attempts to reduce the independent breast movement causing the breast pain experienced under this condition. The key differences in the kinematic parameters in the bare-breasted condition when compared to the low and high breast supports were; a shorter step length, less rotation about the vertical and anteroposterior axis in the thorax and pelvis, less abduction of the upper arm, greater extension of the upper-arm, and reduced flexion of the shank relative to the thigh. The majority of these results indicate a suppression of peak orientation values, corresponding to the reduced ROM of the examined segments when running bare-breasted.

The significant reduction in step length in the bare-breasted condition occurred with reduced flexion at the knee and hip during the swing phase, and reduced rotation about the vertical axis of the pelvis to thorax, indicating changes along the kinetic chain. Saunders, Inman, and Eberhart (1953) and Schache, Bennell, Blanch, and Wrigley (1999) both reported the significance of damping vertical oscillations by less pelvic obliquity, and minimising the amount of anteroposterior tilt, which conserves energy by reducing the vertical displacement of the centre of gravity (CoG), and at the same time reduces the vertical movement of the thorax. These kinematics alterations when running in the barebreasted condition may enable a female runner to conserve energy for a given distance, whilst also reducing the magnitude of breast kinematics. However, this positive effect may be outweighed by the possible increase in energy cost of maintaining a constant running velocity with the reported reduction in step length (Hamill et al., 1995), and the increase in moment of inertia associated with reduced knee flexion during swing, as the mass is situated further away from the axis of rotation (Robertson et al., 2004). Changes in kinematic variables are known to influence a runner's metabolic cost, with changes in stride/step length and frequency away from the self-selected length and frequency reported to increase metabolic cost and affecting running economy (Hunter & Smith, 2007; Williams & Cavanagh, 1982). Research by Moore, Jones, and Dixon (2012) emphasised the influence of running kinematics on running economy, with results demonstrating 94.3% of the variance in running economy in novice female runners could be explained by alterations in the following kinematic parameters; less extended knee at toe off, peak dorsiflexion occurring later in stance, and slower eversion velocity at touchdown.

Changes in joint rotations have also been shown to affect impacts on the body during running (Cole, Nigg, van den Bogert, 1996). White et al., (2009) and Shivitz (2001) suggested that differences in ground reaction forces, when running in different breast support conditions, may be due to changes in running kinematics. The reported differences in the magnitude of flexion of the shank relative to thigh, between breast support conditions, support the hypotheses of White et al., (2009) and Shivitz (2001). When running without breast support the ROM in flexion of the shank was ~10° less, with a less acute peak flexion angle (82.5°) during swing, than in the high breast support condition, which may help to explain the previously published differences in GRFs reported between different breast supports.

The impact of the alterations reported in the lower body kinematics between breast support conditions should be considered. Firstly, it is important to consider the potential drive behind these alterations. It is assumed that the shorter step length and less acute knee angle reported in the bare-breasted condition enabled the participants to spend a longer time in contact with the ground, potentially reducing the vertical oscillation of the upper body. This may have influenced the forces subjected to the body, with greater natural cushioning due to the change in running mechanics. With each foot strike a shock wave is transmitted throughout the body, ultimately reaching the upper body and head (Hamill, Derrick, & Holt, 1995; Mercer, Vance, Hreljac, & Hamill, 2002). Therefore, it is suggested that the participants were altering the lower body kinematics to reduce the force and shock transmitted to the upper body during the unsupported condition. Future work could look at GRFs and kinematic analyses simultaneously to gain a clearer understanding of the effect of breast support during running on both kinetic and kinematic analyses.

Secondly, it is important to consider the potential benefits of the differences reported. A review article by Saunders et al., (2004) suggests that a more acute knee angle is a key biomechanical attribute of an economical runner, therefore, it is suggested that wearing a high level sports bra may ensure an economical running style is maintained during short and prolonged running. However, these affects may only be apparent during the initial stages of the run, since shank flexion did not differ over time within or between the low and high breast support conditions.

The kinematic differences reported within the bare-breasted condition are interesting and detail the potential alterations in running kinematics driven by large magnitudes of relative breast kinematics and breast pain during running. It is assumed that few females exercise under this condition, and therefore these results cannot be generalised to the active female population. However, dependent upon the size of their breasts or sensitivity and prevalence of breast pain, it may be possible to infer these findings for females who may experience similar magnitudes of breast kinematics and exercise-related breast pain during running when wearing an external breast support.

More importantly, the significant differences in running kinematics between the low and high breast support conditions can be extended to the exercising population, and provide crucial information on the effects of breast support on female running kinematics during a five kilometre run. Within the current chapter the high breast support elicited the following kinematic profile when compared to the low breast support condition; greater step length, less thorax yaw (two minutes to third kilometre), less thorax pitch, less axial rotation of the pelvis, less extension of the upper-arm, less abduction of the upper-arm, less peak adduction of the forearm, less flexion of the forearm, and less adduction/abduction of the thigh.

Smaller ROM in thorax pitch (relative to the GCS) was reported when wearing the high breast support when compared to the low breast support condition, during the initial stages of the five kilometre run (two minutes to the second kilometre). The mean differences were relatively low (1° to 1.3°), and the smallest standard deviations were greater than these differences (± 1.5°). Whether these differences in thorax pitch would have an effect on final running performance is unclear at present, however, it is suggested that the ROM in thorax pitch can affect breast kinematics (e.g. greater ROM in thorax pitch, greater ROM in anteroposterior breast displacement). The difference in ROM of thorax pitch between the breast support conditions appeared to be present when examining the peak flexion values. The low breast support condition elicited a greater peak flexion of the thorax compared to the high breast support. Furthermore, peak flexion of the thorax was shown to significantly increase from the first two minutes to the third and fourth kilometre under the low breast support condition. Greater thorax flexion has previously been shown to increase the cost of running and been associated with a less economical running style (Saunders, Pyne, Telford, & Hawley, 2004; Williams & Cavanagh, 1987). It is suggested that this alteration in thorax kinematics under the low breast support condition may be detrimental to females running under this support condition. With the ROM of thorax pitch and peak flexion of the thorax reduced in the high breast support, it is suggested that this breast support condition could be beneficial to female runners, and potentially reduce costly running kinematics associated with the upper body.

In addition, the reported differences in thorax kinematics may facilitate the interpretation of the increases in breast kinematics over the five kilometre run (reported in chapter four) within the low breast support condition. Greater peak flexion of the thorax in the low breast support condition may have significantly influenced the distribution and magnitude of relative breast kinematics. A greater forward lean (peak flexion) of the thorax would mean the vertical axis of the thorax SCS would not be directly aligned with the vertical axis of the GCS, and therefore the gravity vector (9.81 $m.s^{-2}$) would not be solely acting within the vertical direction of the relative breast kinematics. Because of this, the magnitude and contribution of breast kinematics in each direction will differ. Greater magnitudes of anteroposterior breast kinematics may be prevalent in this situation when compared to a thorax that is orientated directly in line with the vertical axis of the GCS.

Another alteration to thorax kinematics between the low and high breast supports was reported in the peak orientation and ROM of thorax yaw. Peak thorax yaw was significantly reduced across all intervals of the five kilometre run in the high breast support when compared to the low support condition. Differences were reported in the ROM in thorax yaw; however the direction of these differences change as the runners progressed through the five kilometre run. Initially, the high breast support reduced the ROM in thorax yaw from the first two minutes to the second kilometre intervals. However, at the final two kilometres of the run the high breast support elicited a greater ROM of thorax yaw than the low breast support. With no significant change in peak rotation over the five kilometre run, within either support condition, this finding is difficult to interpret. The ROM in thorax yaw does however follow a trend within both support conditions, with the ROM progressively increasing in the high breast support, and progressively decreasing in the low breast support as the runners progress through the kilometres of the run.

Peak and ROM in axial rotation of the pelvis relative to the thorax, was significantly less in the high breast support when compared to the low breast support at certain intervals of the five kilometre run. It is interesting to note that the magnitude of rotation between these two segments was more comparable between the high and bare-breasted support condition, whereas significantly greater ROM was reported in the low breast support. It is hypothesised that the reduced axial rotation of the pelvis is due to the previously mentioned alterations in knee flexion and step length, reported under the bare-breasted condition in an attempt to reduce the magnitude of breast kinematics. In contrast, the relative breast kinematics in the high breast support are effectively reduced (chapter four, Section 4.4.2), and therefore the reduced magnitude of pelvic and thorax rotation about the vertical axis in the high breast support might indicate a running style that enables the preservation of energy (Saunders et al., 2004). Another beneficial kinematic trait reported when participants ran in the high breast support.

With arm swing mechanics reported to facilitate stabilisation and reduce angular motion about the vertical axis (Hinrichs, 1990; Park, 2008), any alterations in arm swing mechanics may affect the mechanical profile of a female runner. Furthermore, Williams and Cavanagh (1987) proposed that more economical runners exhibited less arm movement. Similarly, Saunders et al., (2004. p. 472) review article included '*arm motion that is not excessive'* to a list of desirable biomechanical variables for an economical running style. With reduced ROM in upper-arm extension in the high breast support condition, it is suggested that the arm swing mechanics in this condition may preserve more energy. Furthermore, peak abduction was greater in the high breast support, suggesting that participants held their arms further away from the mid-line of the body in the higher breast support conditions.

The aforementioned differences indicate that participants will alter the magnitude of upper-arm extension and abduction dependent upon the breast support worn. Due to the synchronous relationship between the upper and lower limbs, these differences may be prevalent as a result of the differences reported in the lower body (i.e. increased step length and changes in pelvic and thorax rotations in the high breast support condition). Umberger (2008) reported increased metabolic cost, and Eke-Okoro et al., (1997) reported significant alterations in step characteristics and running velocity when arm swing mechanics were suppressed during running. The alteration to arm hold positions in these studies were quite radical, (e.g. folded arms in front of chest, holding hands on head, hips, and behind back, or no arm swing), however, these papers highlight the sensitivity of adaptations to arm swing mechanics on physiological measures that influence running performance (i.e. running economy). The work within the current chapter demonstrates that arm swing mechanics can be affected by the breast support worn, with the high breast support condition reducing excessive arm swing mechanics that could increase the cost of running, and affect an individual's running economy.

The work within the current chapter suggest that breast support can significantly influence both upper and lower body kinematics during short and prolonged treadmill running, we can therefore accept hypothesis one. However, these differences were not consistent across the five kilometre run. Peak thorax flexion was the only kinematic variable reported to increase over the five kilometre run in the low breast support condition, and therefore these findings do not support hypothesis two. The majority of the studies examining changes in running kinematics over time have examined runners over long distance running using protocols designed to elicit fatigue. The distance selected in the current study was not employed with the aim of eliciting undue fatigue, but to ensure the investigation of breast and running biomechanics were examined over an externally valid run distance. Therefore, it is difficult to make comparisons with past work within this area. Furthermore, the participants recruited in the current study were training at this running distance, and therefore any differences reported could be attributed to the breast support worn and not due to mechanical fatigue.

Thorax yaw ROM was negatively correlated with mediolateral and vertical breast displacement. However, no relationships were reported between thorax ROM and relative breast velocity and accelerations, nor peak orientation of the thorax and multiplanar breast kinematics. The relationship to relative vertical breast displacement (large) displayed a stronger relationship than the mediolateral direction (moderate). It was hypothesised that the relationship between these two variables would be positive, i.e. greater thorax rotation would elicit a greater magnitude of breast kinematics. However, when considering the reduction in magnitude of relative breast kinematics and the greatest ROM in thorax yaw reported under the high breast support, a negative relationship is evident. Therefore, hypothesis three can be partially accepted. During running, the bare-breasted condition elicited the smallest degree of thorax yaw, suggesting that during a set-paced treadmill protocol the participants are restricting the thorax range of motion when the breasts are unsupported. The inertia of the breast and the magnitude of pain experienced under this condition are proposed as the variables influencing this change in kinematics and the relationships reported within the current chapter.

The calculation of the coefficient of determination enabled the percentage of variability shared by these two variables to be reported, providing an indication of how much of each variable accounts for the other. The coefficient of determination suggested that thorax yaw accounted for 15% of mediolateral breast displacement. Furthermore, thorax yaw accounted for 49% of breast displacement in the vertical direction. These data facilitate the design of sports bras. The suggestion that thorax yaw can account for almost 50% of vertical breast displacement during running informs manufacturers that sports bras designed for running should include sufficient support on the lateral and upper pole regions of the bra.

5.6 Conclusion

The work presented in this chapter is the first to examine the effect of breast support on peak orientation and ROM of upper and lower body segments, during short (two minutes) and prolonged (five kilometre) treadmill running. It has been shown that reduced levels of breast support (bare-breasted and everyday bra) elicit certain alterations in running kinematics that have previously been related to less economical running styles and potentially detrimental to performance (e.g. greater arm extension, suppressed arm abduction, more forward lean of the thorax, shorter step length, and less acute knee flexion). Conversely, in a high level of breast support, the mechanical profile represented a more economical running style (e.g. reduced arm extension, greater step length, and a more acute knee flexion). Peak flexion of the thorax was the only kinematic variable to change over the five kilometre run, with a significantly greater forward lean in the low breast support condition. It is assumed that greater peak flexion of the thorax may have influenced the distribution and increases in magnitude of relative breast kinematics over the five kilometre run within the low breast support condition, however this finding cannot explain the differences reported under the high breast support condition within chapter four, section one.

Human movement is driven by the associated functional muscles; shortening and lengthening to produce and dissipate energy, and/or stabilise a joint to create mechanical movement (Higham, Biewener, & Delp, 2011). Therefore, any changes in segmental movements are driven by changes in neural drive of the neuromuscular system (e.g. modulations in muscle fibre firing rates) (Basmajian & De Luca, 1985). Since differences in peak orientation and ROM of upper body segments were apparent during running when wearing different breast supports, it is logical to explore changes in muscle activity of the muscles which drive these movements. The impact of breast support on upper body muscle activity during running has received little attention; examining these variables will further the knowledge of the effect breast support has on biomechanical measures.

CHAPTER SIX.

THE INFLUENCE OF BREAST SUPPORT ON UPPER-BODY MUSCLE ACTIVITY DURING FIVE KILOMETRE TREADMILL RUNNING

6.1 Introduction

The electromyographical profile and characteristics of lower body muscles during running has been extensively researched (Gazendam & Hof, 2007; Rand & Ohtsuki, 2000; Yokozawa, Fujii, & Ae, 2007), however, the study of electromyography (EMG) of the upper body during running has received considerably less attention (Newton et al., 1997; Smoliga, Myers, Redfern, & Lephart, 2010). Furthermore, there is little published literature which explores EMG of the upper body during running in female participants. When considering the additional mass and magnitude of soft tissue movement of the breast for female runners (Scurr et al., 2010a; Haake and Scurr, 2010), a question that remains unanswered is whether this additional mass and relative soft tissue movement affects the recruitment of motor units and the magnitude of myoelectric activity, specifically within the upper body. A 34 D cup participant has an approximated breast mass of 460 g per breast (Turner & Dujon, 2005), and on average may experience vertical breast displacement up to 80 mm (Scurr et al., 2009a) when unsupported during treadmill running. However the effect of this additional wobbling mass on the neuromuscular system during running has received little attention.

Martin and Morgan (1992) suggest that the distribution of mass on a segment will influence the metabolic cost of locomotion, assuming factors such as; velocity, body mass, and running style remain constant, segments with smaller inertial loads with the mass closer to the primary axes of rotation require less muscular effort to accelerate the limbs. Differences in bra structure, shape and materials utilised for the garment will influence the amount of compression the bra provides, ultimately changing the distribution of the breast mass over the chest. This may alter the inertial properties of the breast and the moment of inertia of the thorax during running, which may help to explain the differences reported in thorax kinematics between breast support conditions (chapter five). Furthermore, during running it is suggested that in the high breast support condition the muscle activity associated with thorax kinematics will be less.

Bennett (2009) explored differences in postural muscle activity in females with larger breasts (defined as D or larger) during a range of simple tasks, such as step up, sitting and picking up a pencil. Muscles of the cervico-thoracic region were investigated as it was found that common complaints of neck, back, and shoulder pain were as a result of increased tension (activation) of these muscles due to the mass of the breast tissue. It is interesting that the pectoralis major was not examined within this work due to the location and anatomical connections to the breast. Higher percentages of muscle activation in females with larger breasts were reported when compared to smaller cup sizes, during static postural trials (Bennet, 2009). When considering the results from Bennet (2009), it is important to consider how movement of the breast mass may affect the muscles of the upper body during dynamic tasks, such as running, and what impact this may have on the neuromuscular system during physical activity.

To date only one abstract has examined the effect of breast support (and the associated magnitude of breast movement) on muscle activity in the upper body during a treadmill protocol, lasting two minutes in duration (10 $km.h^{-1}$) (Scurr et al., 2010b). The following upper body muscles were examined; pectoralis major, anterior deltoid, upper and lower trapezius, and erector spinae in a bare-breasted condition, an everyday T-shirt bra, and a combination sports bra. The raw EMG data were rectified and then processed using integrated EMG (*i*EMG); the results indicated no differences in *i*EMG in the majority of investigated muscles across breast support conditions. However, differences were reported in pectoralis major activity when running with and without breast support. Scurr et al., (2010b), proposed that the associated increase in pectoralis major muscle activity when breast support was removed may indicate a contribution of this muscle to the anatomical support of the breast during running. However, the relationship between pectoralis major activity and breast kinematics was not explored further. If the pectoralis major muscle activity is greater in lower breast supports during two minute running, the implications of this over a prolonged run distance could be detrimental to performance. The increase in muscle activity will increase the metabolic demand over a given exercise period, and may result in earlier muscular fatigue, and reductions in running economy.

The relationship between the pectoralis major and the breast is of interest. The pectoralis major is situated underneath the breast tissue and is responsible for a combination of movements of the upper arm during running, such as adduction and flexion (Basmajian & De Luca, 1985). As detailed in the introduction of chapter three, the two proposed anatomical supportive tissues to the breast are the Cooper's ligaments and the overlying skin, the pectoralis major is currently not thought to provide any additional support to the breast tissue. The anatomical connection of the pectoralis major muscle to the breast is minimal with the Cooper's ligament extending inward from the skin and attaching on to the deep pectoral fascia (Hamdi et al., 2005; Gefen & Dilmoney, 2007). When breast tissue oscillates during running the tension placed upon the attachment site of the Cooper's ligaments to the pectoral fascia can only be approximated as currently no published data exist on the mechanical properties of these two tissues (Gefen & Dilmoney, 2007). It could be hypothesised that the tension placed upon the connection site (due to the weighted oscillating tissue) between the Cooper's ligaments to the deep superficial fascia of the breast, which is fused to the pectoralis fascia, may cause the pectoralis major muscle to activate in an attempt to reduce this movement of the breast and the tension at this site. The deep superficial fascia is a dense fibrous tissue that surrounds the pectoralis fascia and muscle, and provides connections and support to the projected ligaments, nerves and blood vessels from the superficial layer as they pass through the retromammary space to the pectoralis fascia. Breast parenchyma can accompany these tissues to the pectoralis major muscle itself (Hamdi et al., 2005). The significance of these connections are emphasised when considering breast surgeries which require the removal of the entire breast (mastectomy). Hamdi et al., (2005) emphasise the necessity of excision of the pectoralis fascia and a layer of the pectoralis major muscle during these procedures, confirming the connection between the breast tissue and the pectoralis major muscle.

Two processing techniques that are commonly employed to assess the EMG signal in the time domain are root-mean-square (RMS) and the integral of the EMG signal (*i*EMG). The outcomes of both techniques are dependent upon the number of recruited motor units and the firing rate of the innervated muscle fibres (Basmajian & De Luca, 1985). However, the quantification and information gained from these two techniques differ. The RMS processing technique quantifies the amplitude of the signal and represents signal power and therefore has physical meaning (De Luca, 1997). The *i*EMG processing technique sums the total activity in a period of time so the total accumulated activity can be computed (Kamen, 2004), with *i*EMG processing previously utilised as a method to determine total work (Abrabadzhiev, Dimitrov, Dimitrova, & Dimitrov, 2010; Edwards & Lippold, 1956).

Smoliga et al., (2010) suggested it should not be assumed that a given EMG processing technique has the same reliability and precision for all muscles, and reported differences in precision between muscles with the similar functions and within the same anatomical region between processing techniques. This paper supports the presentation of multiple processing techniques in order to gain accurate interpretations of changes at the muscular level in multiple muscles. Furthermore, when examining the reliability and precision of the *i*EMG and RMS techniques, Smoliga et al., (2010) recommended the *i*EMG over the RMS technique during running, although both are frequently reported. Although it is important to consider the reliability of EMG processing techniques, it should be noted that these two techniques measure different parameters of muscle activity. The *i*EMG quantifies the area under the curve and represents the total amount of muscle activity present over a given time, whereas the RMS processing technique represents the amplitude of the muscle activity. Therefore, presenting these two methods provides a greater representation of the muscles EMG signal characteristics. If differences are reported in EMG time-domain analysis, between breast support conditions, this may signify a potential change in the neural drive of the muscles, such as the number of recruited motor units or the firing rate of the muscle fibres, both of which may influence the metabolic cost of running.

Because of the known variability of the EMG signal between testing sessions, trials, muscles, and participants, normalisation of the data is required (Burden, 2010; De Luca, 1997). This involves dividing the EMG from a specific task by a reference action of the same muscle, with the reference action processed using the same methods as the event EMG (Burden, 2010). The EMG is then presented as a percentage of the reference value. The most common method for eliciting the reference muscle action is a maximal voluntary contraction (MVC). However this method has received criticism as it does not provide good reliability, with different maxima values observed within the same participant. Furthermore, EMGs during dynamic tasks have been reported to greatly exceed the MVC value (Burden, 2010; Clarys, 2000). Another consideration when implementing the MVC method is the ability of the participants to stimulate certain muscles at maximum capacity; certain muscles of the upper body are difficult to activate maximally and the action does not always replicate the action of the dynamic task. Because of these criticisms Clarys (2000) suggested the MVC method should not be used for normalisation of dynamic activities. Although debate still exists between the most appropriate normalisation methods, it is apparent that the aim of the study should drive the decision (Burden, 2010). Clarys (2000) recommended the use of the peak dynamic method, since it replicates the exercise of interest. Furthermore, this method has been shown to reduce inter-individual variability (Yang and Winter, 1984), which is commonly high in EMG signals.

Within the previous chapter participants provided subjective feedback during the run trials in the different breast support conditions, which demonstrated that participants experienced feelings of ‘tensing’, ‘more rigid’, and that their shoulders felt ‘in a further forward, hunching position’ in the low support compared to the high support. It is postulated that these feelings and experiences may impact upon the individual’s rating of perceived exertion (RPE) and may elicit a feelings of greater perceived exertion. The aforementioned subjective responses are similar to those expressed in the study by Bennett (2009), and were postulated to be due to the increased magnitude of breast kinematics and breast pain reported in the low breast support condition, moreover these findings could be linked to similar changes in muscular activity in the cervico-thoracic region. Alongside objective measures of running biomechanics (i.e. kinematic analysis and EMG), subjective ratings of exertion can be monitored through a Borg score (Borg, 1990) to help interpret the influence of any differences in running biomechanics. The RPE scores are expected to increase linearly throughout constant load exercise (Noakes, 2004) and will provide information on the ability of an individual to maintain exercise at a given pace.

Examining the amplitude (peak RMS) and total (*i*EMG) muscle activity of key functional upper body muscles, in different breast support conditions, during a five kilometre run, will increase the understanding of the effect of breast support on the neuromuscular system during running. Investigating this area will broaden the knowledge of changes in crucial biomechanical parameters of running performance, dependent upon the level of breast support worn. Furthermore, greater exploration of the relationship between the pectoralis major and breast kinematics will provide an insight into the hypothesis made by Scurr et al., (2010a) regarding the role the pectoralis major may have in providing anatomical support to the breast.

6.2 Aims and research hypotheses

The primary aim of the study was to examine the effect of breast support on upper body myoelectric activity during a five kilometre run, through examination of the peak amplitude (RMS) and *i*EMG muscle activity. A secondary aim was to explore the relationship between breast kinematics and pectoralis major muscle activity.

H_1 There will be significantly greater peak amplitude (RMS) and total amount (*i*EMG) of upper body muscle activity in the bare-breasted condition, compared to the low and high breast support conditions.

H_2 There will be significant differences in peak RMS and *i*EMG values between the intervals of the five kilometre run in both low and high breast support conditions.

H_3 A significant positive relationship will be reported between breast kinematic data and pectoralis major muscle activity.

H_4 Participants will perceive their physical exertion as greater in the low breast support condition when compared to the high breast support condition.

6.3 Methods

6.3.1 Participants

The participants and procedures for chapter six were the same as chapters four and five, with additional data analysis conducted on upper body EMG, see chapter four, section 4.3.1 for general procedures.

6.3.2 Data collection

Electromyography data were recorded over 10 second periods at each sampling interval of the five kilometre run (at the last ten seconds of the first two minutes and then at each kilometre interval thereafter), and was time synchronised with the Oqus motion camera system by a wireless external start trigger (Flash RT-16, Neewer). The two systems were programmed to start recording when this trigger was pressed, ensuring the two systems were time synchronised. Borg’s (1990) rating of perceived exertion (RPE) scale was implemented with a verbal description of the RPE scale given to the participants prior to data collection. Participants were required to verbally state their rating from the Borg scale (Appendix C), after the first two minutes and at each interval of the five kilometre run trials.

6.3.3 Electromyography

Electromyography data were collected using an eight channel Datalink EMG system (Biometrics, UK). Electrodes were positioned parallel with the muscle fibres on the muscle belly (De Luca, 1997) on the right side of the body on the following muscles; pectoralis major, anterior and medial deltoid, upper trapezius, latissimus dorsi, and erector spinae, in accordance with the SENIAM (Surface EMG for a non-invasive assessment of muscles) recommendations (Table 35 and Figure 17).

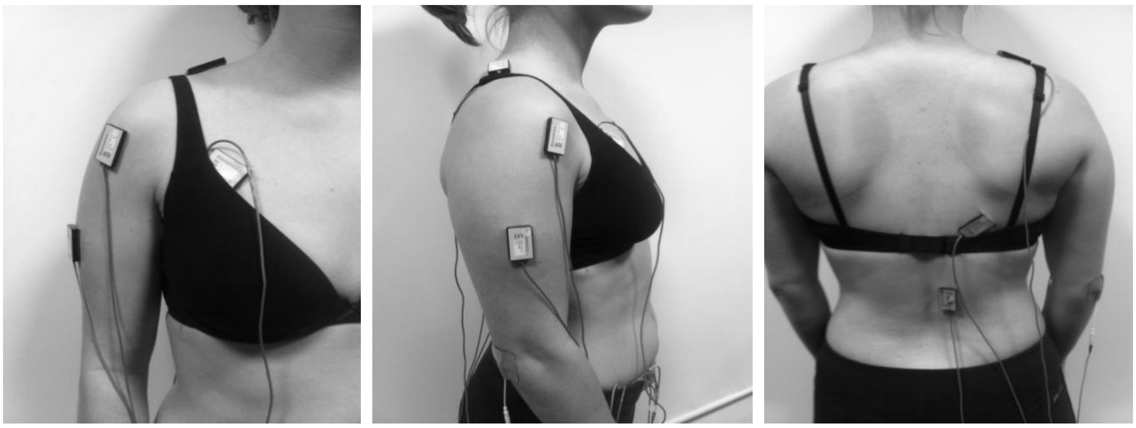

Figure 17. Electrode placement on the six upper body muscles.

Table 35. SENIAM recommendations for participant starting postures and electrode placement of the investigated upper body muscles.

Muscle	Function	Starting posture	Electrode placement
Pectoralis Major *	Forward flexion and adduction of the humerus	Erect, sitting with arms hanging vertically.	Centrally positioned over the pars clavicularis
Anterior Deltoid	Abduction of the shoulder joint and medially rotate the shoulder joint.	Sitting with arms hanging vertically and the palm pointing inwards.	One finger width distal and anterior to the acromion
Medial Deltoid	Abduction of the shoulder joint.	Sitting with the position of the trunk in relation to the arm such that a stable trunk will need no further stabilization.	Placed from the acromion to the lateral epicondyle of the elbow. Corresponding to the most prominent bulge of the muscle.
Trapezius Descendens (upper)	Adduction, rotation, and elevation of the scapula. Rotation of the head.	Erect, sitting with the arms hanging vertically.	Placed 50% on the line from the acromion to C7.
Erector Spinae	Trunk extension	Prone with the lumbar vertebral columns slightly flexed.	Placed two finger width lateral from L1.
Latissimus Dorsi	Adduction , extension, and internal rotation of the shoulder joint	Prone with the lumbar vertebral columns slightly flexed.	Placed two finger widths below the scapula centrally placed in line with the Trapezius electrode.

* For the pectoralis major the electrode was positioned centrally at the pars clavicularis to reduce the signal attenuation due to the impedance of the breast tissue (Kŕol, Sobota, & Nawrat, 2007).

To reduce skin impedance, the skin was prepared by shaving and cleansing the area with an isopropyl alcoholic swab (Medi-Swab, UK) (De Luca, 1997). Biometrics SX230 active (Ag/AgCl) bipolar pre-amplified disc electrodes (gain x 1000; input impedance >100 MΩ; common mode rejection ratio >96dB; with a 1 cm electrode contact surface, and 2 cm separation distance) were adhered to the site (De Luca, 1997) using a hypoallergenic adhesive tape (3M, UK). Electromyography signals were sampled at 1000 Hz. A passive reference electrode was positioned at an electronically neutral site on the olecranon process. The Datalink utilised both high-pass (18 dB/octave; <20 Hz) to remove DC offsets, and low pass filters for frequencies >450 Hz. The electrodes include an eighth order elliptical filter (-60 dB at 550 Hz). The Datalink system was zeroed before any data were collected; this involved the participants lying supine and relaxing. The electrode placement was verified by voluntary muscle actions. The electrodes were secured with clinical tape in an attempt to reduce relative movements of the electrodes during running. The electrodes were connected to the Datalink subject unit, which was securely attached to the side hand bar of the treadmill and wires grouped together to limit artefacts due to hardware movement.

6.3.4 Data Processing

Gait cycle identification was performed as defined in the previous chapters using a marker on the heel (chapter four, section 4.3.3). For comparisons between studies, identical gait cycles to the previous chapters were defined for the EMG analysis, with the heel strike event time noted from the kinematic data.

Electromyography data were uploaded onto Datalink analysis (Version 5.02, Biometrics, UK) for processing. The raw EMG signals (mV) were visually checked for artefacts and then processed using two processing techniques; (1) RMS (filter constant of 100 ms), and (2) full-wave rectified, followed by an *i*EMG (filter mV.s) performed over every sample. Processing techniques were employed to the raw data separately, for five gait cycles at each interval of the five kilometre run trials in each support condition. This was conducted for each muscle (six muscles) for all breast support conditions. The processed EMG signals (RMS and *i*EMG) were normalised using the bare-breasted data as the denominator (Scurr et al., 2010b), in line with the assumption that the peak RMS and *i*EMG values would be reported under the bare-breasted condition for each muscle (Equation 2). Where the peak EMG value within a gait cycle, under the bare-breasted condition is used as the denominator, then all peak values from five gait cycles (n=5) at each distance interval (n=6), for each muscle (n=6) within all breast support conditions are then quantified as a percentage of the denominator. The normalisation processes for both techniques are detailed in Figure 18.

Equation 2.

(Peak EMG value of gait cycle/ Peak EMG value in bare-breasted condition)*100

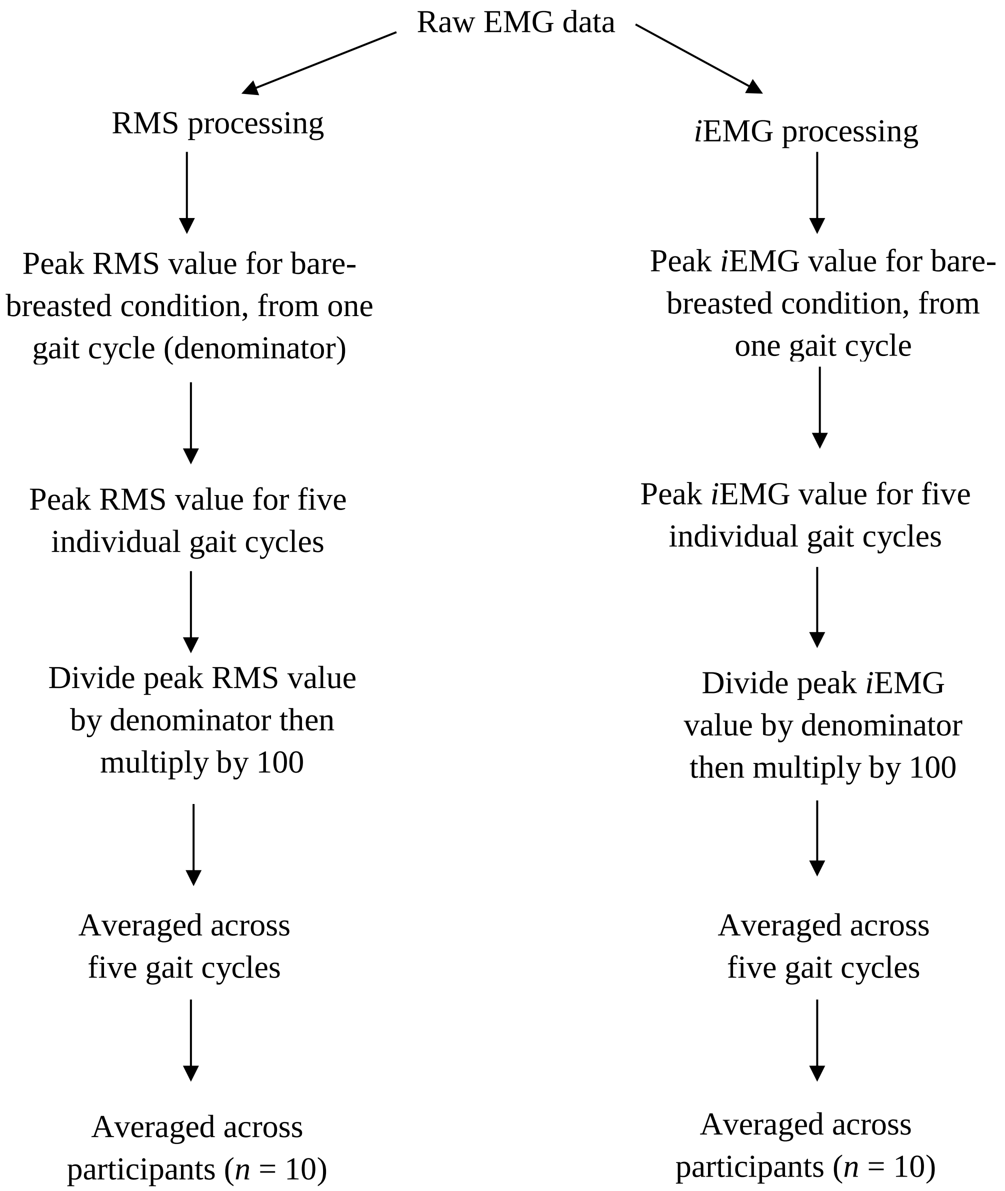

Figure 18. Flow chart of processing stages for both RMS and *i*EMG techniques.

6.3.5 Statistical Analyses

All normalised EMG data for the six investigated muscles were checked for normality using the Kolmogorov-Smirnov and Shapiro-Wilk tests of normality, with normality assumed when $p > .05$. Homogeneity of variance was assessed using Mauchly's test of Sphericity, with homogenous data assumed when $p > .05$. Data was accepted as normally distributed and displaying homogeneity and therefore defined as parametric. The independent variables examined were the three breast support conditions and the six intervals of the five kilometre run, and the dependent variables examined were the normalised peak RMS and *i*EMG signals. One-way repeated measures ANOVAs were performed to examine the effect of support conditions on EMG data for the two minute treadmill run data. Two-way repeated measures ANOVAs were performed to examine the main and interaction effects of breast support conditions and run distance on the EMG data during the five kilometre run trials. *Post-hoc* pairwise comparisons, with Bonferroni adjustment, were performed alongside the two-way repeated measures ANOVAs. Pearson's moment product correlations (r) were performed to explore the relationship between EMG data of the pectoralis major and multiplanar breast kinematics, where a small relationship = $\pm \leq .10$, medium relationship = $\pm \leq .30$, and large relationship = $\pm \geq .50$ (Field, 2009). Non-parametric Wilcoxon signed-rank tests were performed on the RPE responses across the five kilometre run in the low and high breast support conditions. Effect size (η^2) and observed power (1-β) are calculated to characterise the strength of the results, where a small effect = ≤ .10, a medium effect = ≤ .30, a large effect = ≤ .50, and a high power = ≥ .80 (Field, 2009).

6.4 Results

6.4.1 The effect of breast support on upper body muscle activity

6.4.1.1 Pectoralis major (PM)

During the two minute run, the high breast support reduced the peak RMS pectoralis major activity by 29% and 28% compared to the bare-breasted and low support conditions, respectively (Table 36). Peak RMS activity significantly reduced by 45% in the high support when compared to the low breast support at the fourth kilometre interval (p = .005). However, no differences were reported in peak RMS pectoralis major activity within the low and high breast support over the five kilometre run distance.

Table 36. Normalised (%) mean (± SD) peak RMS and *i*EMG of the pectoralis major during the two minute and five kilometre treadmill run trials, in three levels of breast support (n = 10).

Intervals	RMS (%)			*i*EMG (%)		
	BB	LOW	HIGH	BB	LOW	HIGH
2 minutes	82 ± 11*[b]	81 ± 27*[c]	58 ± 39*[bc]	75 ± 7	93 ± 26	85 ± 33
1 km		71 ± 27	55 ± 35		95 ± 34	74 ± 32
2 km		71 ± 26	58 ± 47		95 ± 35	69 ± 30
3 km		69 ± 19	56 ± 40		86 ± 34	82 ± 43
4 km		86 ± 33*[c]	47 ± 24*[c]		87 ± 23	74 ± 35
5 km		61 ± 25	56 ± 43		85 ± 28	77 ± 33
Mean	82 ± 11	73 ± 9	55 ± 4	75 ± 7	90 ± 5	76 ± 6

*[a] Denotes a significant difference between BB and low breast support conditions, where p < .05.
*[b] Denotes a significant difference between BB and high breast support conditions, where p < .05.
*[c] Denotes a significant difference between the low and high breast support conditions, where p < .05.
†Denotes a significant difference between the first two minutes and the distance intervals, where p < .05.

N.B. Significant main effect of breast support on the peak RMS pectoralis major muscle during the two minute ($F_{(2)}$ = 3.662, p = .046, η = .289, 1-β = .598) and five kilometre ($F_{(1)}$ = 7.506, p = .023, η = .445, 1-β = .685) treadmill runs.

Total pectoralis major activity (*i*EMG) did not differ between breast support conditions during the two minute or five kilometre run (Table 33).

6.4.1.2 Anterior deltoid (AD)

Peak RMS anterior deltoid activity was affected by the breast support worn (Table 37), with the bare-breasted condition eliciting 60% more activity when compared to the low breast support (p = .035), and 36% more than the high breast support (p = .045). However, breast support did not affect peak RMS muscle activity of the anterior deltoid during the five kilometre run distance. Furthermore, no differences were reported over the five kilometre run within either breast support condition.

Total muscle activity (*i*EMG) of the anterior deltoid did not differ between breast support conditions over the two minute and five kilometre runs. However, differences were reported in the total activity (*i*EMG) of the anterior deltoid within the low and high breast support conditions over the five kilometre run. Between the first two minutes and the fourth kilometre interval (p = .031) a significant increase in total muscle activity in the anterior deltoid was reported within the low and high support, increases of 12% and 57%, respectively.

Table 37. Normalised (%) mean (± SD) peak RMS and *i*EMG of the anterior deltoid during the two minute and five kilometre treadmill run trials, in three levels of breast support (n = 10).

Intervals	RMS (%)			*i*EMG (%)		
	BB	LOW	HIGH	BB	LOW	HIGH
2 minutes	72 ± 16*[ab]	45 ± 26*[ac]	53 ± 32*[bc]	78 ± 13	74 ± 54	65 ± 39
1 km		45 ± 21	56 ± 25		77 ± 43	70 ± 35
2 km		34 ± 15	52 ± 32		72 ± 43	80 ± 44
3 km		40 ± 11	79 ± 32		86 ± 44	94 ± 34
4 km		45 ± 12	54 ± 23		83 ± 47 †	102 ± 40 †
5 km		52 ± 19	68 ± 39		90 ± 45	99 ± 42
Mean	72 ± 16	44 ± 6	60 ± 11	78 ± 13	80 ± 7	85 ± 16

*[a] Denotes a significant difference between BB and low breast support conditions, where p < .05.
*[b] Denotes a significant difference between BB and high breast support conditions, where p < .05.
*[c] Denotes a significant difference between the low and high breast support conditions, where p < .05.
†Denotes a significant difference between the first two minutes and the distance intervals, where p < .05.

N.B. Significant main effect of breast support on peak RMS anterior deltoid activity during the two minute run ($F_{(2)} = .359$, $p = .031$, $\eta = .353$, $1\text{-}\beta = .669$). Significant main effect of run duration on the *i*EMG anterior deltoid activity during the five kilometre run ($F_{(5)} = 4.018$, $p = .006$, $\eta = .365$, $1\text{-}\beta = .913$).

6.4.1.3 Medial deltoid (MD)

The bare-breasted condition elicited 54% greater peak RMS activity for the medial deltoid compared to the high breast support during the first two minutes of running (Table 38).

Table 38. Normalised (%) mean (± SD) peak RMS and *i*EMG of the medial deltoid during the two minute and five kilometre treadmill run trials, in three levels of breast support (n = 10).

Intervals	RMS (%)			*i*EMG (%)		
	BB	LOW	HIGH	BB	LOW	HIGH
2 minutes	83 ± 12*[b]	70 ± 20*[c]	54 ± 17*[bc]	82 ± 8*[b]	74 ± 27	62 ± 22*[b]
1 km		77 ± 20*[c]	55 ± 19*[c]		79 ± 32	63 ± 25
2 km		83 ± 31*[c]	63 ± 28*[c]		86 ± 44	67 ± 27
3 km		71 ± 19	59 ± 24		79 ± 44	71 ± 29
4 km		69 ± 21	56 ± 20		76 ± 29	65 ± 24
5 km		61 ± 14	65 ± 28		71 ± 28	70 ± 29
Mean	83 ± 12	72 ± 7	59 ± 5	82 ± 8	78 ± 5	66 ± 4

*[a] Denotes a significant difference between BB and low breast support conditions, where $p < .05$.
*[b] Denotes a significant difference between BB and high breast support conditions, where $p < .05$.
*[c] Denotes a significant difference between the low and high breast support conditions, where $p < .05$.
†Denotes a significant difference between the first two minutes and the distance intervals, where $p < .05$.

N.B. Significant main effect of breast support on peak RMS medial deltoid activity during two minutes ($F_{(2)} = 9.327$, $p = .002$, $\eta = .509$, $1\text{-}\beta = .953$) and five kilometre ($F_{(1)} = 7.101$, $p = .026$, $\eta = .441$, $1\text{-}\beta = .661$) run durations. Significant main effect of breast support level on *i*EMG of the medial deltoid during the two minute run duration ($F_{(2)} = 4.832$, $p = .021$, $\eta = .349$, $1\text{-}\beta = .726$).

Breast support also influenced the peak RMS value of the medial deltoid during the five kilometre treadmill run, with the low breast support eliciting greater peak values at the first ($p = .003$) and second ($p = .023$) kilometre intervals. Distance of the run was not shown to affect the peak RMS values during the five kilometre run within and between the breast support conditions.

Total activity (*i*EMG) of the medial deltoid was greater in the bare-breasted condition compared to the high support condition ($p = .028$) during the two minute run, a reduction of 24% by the high breast support. However, no differences were reported in total activity of the medial deltoid within or between the two breast support conditions over the five kilometre run.

6.4.1.4 Upper trapezius (UT)

Peak RMS and total activity (*i*EMG) of the upper trapezius did not differ between the three breast supports examined during the two minute and five kilometre treadmill runs (Table 39).

Table 39. Normalised (%) mean (± SD) peak RMS and *i*EMG of the upper trapezius during the two minute and five kilometre treadmill run trials, in three levels of breast support (n = 10).

Intervals	RMS (%)			*i*EMG (%)		
	BB	LOW	HIGH	BB	LOW	HIGH
2 minutes	81 ± 7	70 ± 19	77 ± 36	82 ± 9	78 ± 31	95 ± 60
1 km		75 ± 31	70 ± 34		70 ± 25	99 ± 53
2 km		67 ± 26	87 ± 36		66 ± 30	93 ± 36
3 km		69 ± 39	85 ± 36		70 ± 23	93 ± 37
4 km		71 ± 32	86 ± 47		73 ± 28	96 ± 38
5 km		78 ± 43	91 ± 46		79 ± 31	99 ± 40
Mean	81 ± 7	72 ± 4	83 ± 8	82 ± 9	73 ± 5	96 ± 3

*[a] Denotes a significant difference between BB and low breast support conditions, where p < .05.
*[b] Denotes a significant difference between BB and high breast support conditions, where p < .05.
*[c] Denotes a significant difference between the low and high breast support conditions, where p < .05.
†Denotes a significant difference between the first two minutes and the distance intervals, where p < .05.

6.4.1.5 Erector spinae (ES)

Peak RMS and total activity (*i*EMG) of the erector spinae did not differ between the three breast supports examined during the two minute and five kilometre treadmill runs (Table 40).

Table 40. Normalised (%) mean (± SD) peak RMS and *i*EMG of the erector spinae during the two minute and five kilometre treadmill run trials, in three levels of breast support (n = 10).

Intervals	RMS (%)			*i*EMG (%)		
	BB	LOW	HIGH	BB	LOW	HIGH
2 minutes	80 ± 8	83 ± 36	82 ± 30	84 ± 6	84 ± 30	77 ± 21
1 km		81 ± 31	67 ± 26		86 ± 31	67 ± 19
2 km		76 ± 21	68 ± 30		79 ± 25	72 ± 21
3 km		78 ± 28	69 ± 24		83 ± 19	77 ± 24
4 km		85 ± 30	70 ± 31		98 ± 40	76 ± 28
5 km		84 ± 35	75 ± 33		92 ± 26	89 ± 38
Mean	80 ± 8	81 ± 4	72 ± 6	84 ± 6	87 ± 7	76 ± 7

*[a] Denotes a significant difference between BB and low breast support conditions, where $p < .05$.
*[b] Denotes a significant difference between BB and high breast support conditions, where $p < .05$.
*[c] Denotes a significant difference between the low and high breast support conditions, where $p < .05$.
†Denotes a significant difference between the first two minutes and the distance intervals, where $p < .05$.

6.4.1.6 Latissimus dorsi (LD)

Peak RMS and total activity (*i*EMG) of the latissimus dorsi did not differ between the three breast supports examined during the two minute and five kilometre data (Table 41).

Table 41. Normalised (%) mean (± SD) peak RMS and *i*EMG of the latissimus dorsi during the two minute and five kilometre treadmill run trials, in three levels of breast support ($n = 10$).

Intervals	RMS (%)			*i*EMG (%)		
	BB	LOW	HIGH	BB	LOW	HIGH
2 minutes	79 ± 8	77 ± 27	65 ± 19	80 ± 6	72 ± 22	65 ± 23
1 km		67 ± 29	71 ± 31		65 ± 22	67 ± 23
2 km		59 ± 19	67 ± 31		64 ± 20	61 ± 23
3 km		62 ± 19	65 ± 25		66 ± 20	63 ± 24
4 km		62 ± 21	67 ± 23		64 ± 18	65 ± 23
5 km		63 ± 23	67 ± 31		67 ± 24	64 ± 25
Mean	79 ± 8	65 ± 6	67 ± 2	80 ± 6	66 ± 3	64 ± 2

*[a] Denotes a significant difference between BB and low breast support conditions, where $p < .05$.
*[b] Denotes a significant difference between BB and high breast support conditions, where $p < .05$.
*[c] Denotes a significant difference between the low and high breast support conditions, where $p < .05$.
†Denotes a significant difference between the first two minutes and the distance intervals, where $p < .05$.

In summary, the magnitude of breast support worn influenced the peak RMS activity of the pectoralis major, anterior deltoid and medial deltoid during two minutes of running and certain intervals of the five kilometre run. The low and high breast support conditions significantly reduced the peak RMS values of the aforementioned muscles compared to the bare-breasted condition during the first stages of the five kilometre run (2 minutes to second kilometre). Differences were also reported in the total activity (*i*EMG) of the anterior and medial deltoid, with significant increases in the anterior deltoid reported over the five kilometre run in both low and high breast supports, and the high breast support significantly reducing the *i*EMG in the medial deltoid during the two minute run compared to the low breast support.

6.4.1.7 Ranking of muscle activity

Normalised peak RMS and *i*EMG activity for each muscle were averaged over the two minute run duration for the bare-breasted condition, and over the five kilometre run intervals for the low and high breast supports. The most active muscle under the RMS (Table 42) and *i*EMG (Table 43) processing techniques were then reported to compare muscle activity profiles between breast support conditions.

The pectoralis major demonstrated high levels of RMS activity for the bare-breasted and low breast support, but was identified as the least active muscle within the high breast support condition. The erector spinae elicited the second greatest RMS activity in both the low and high breast support conditions. When the peak RMS activity was averaged across all muscles within each breast support condition, the high breast support elicited the lowest peak RMS values when compared to the bare-breasted and low support conditions.

Table 42. Normalised (%) mean peak RMS muscle activity ranked in order of greatest amplitude over the two minute and five kilometre runs, within each breast support condition (n = 10).

BB		LOW		HIGH	
MUSCLES	%	MUSCLES	%	MUSCLES	%
MD	83	PM	78	UT	83
PM	82	ES	78	ES	72
UT	81	MD	74	LD	67
ES	80	UT	71	AD	60
LD	79	LD	67	MD	59
AD	72	AD	60	PM	55
MEAN	80	MEAN	71	MEAN	66

The pectoralis major muscle demonstrated the greatest total activity (*i*EMG) in the low breast support, whereas the upper trapezius was the greatest total activity for the bare-breasted and the high breast support conditions. Within all support conditions the latissimus dorsi was reported as having the least total activity during running.

Table 43. Normalised (%) mean *i*EMG muscle activity ranked in order of greatest total activity over the two minute and five kilometre runs, averaged within each breast support condition (n = 10).

BB		LOW		HIGH	
MUSCLES	%	MUSCLES	%	MUSCLES	%
UT	84	PM	90	UT	96
ES	82	ES	87	AD	85
PM	81	AD	80	PM	77
AD	81	MD	77	ES	76
MD	75	UT	73	MD	66
LD	70	LD	66	LD	64
MEAN	80	MEAN	79	MEAN	77

When the peak RMS and *i*EMG were averaged across all muscles within each support condition, the bare-breasted condition elicited the greatest total activity compared to the low and high breast support conditions.

6.4.1.8 Correlations of pectoralis major muscle and multiplanar breast kinematics

These data demonstrate that as the magnitude of certain breast kinematic variables increased, so did the peak RMS values of the pectoralis major muscle (Figures 19 to 21). The anteroposterior breast displacement demonstrated the strongest relationship to pectoralis major activity.

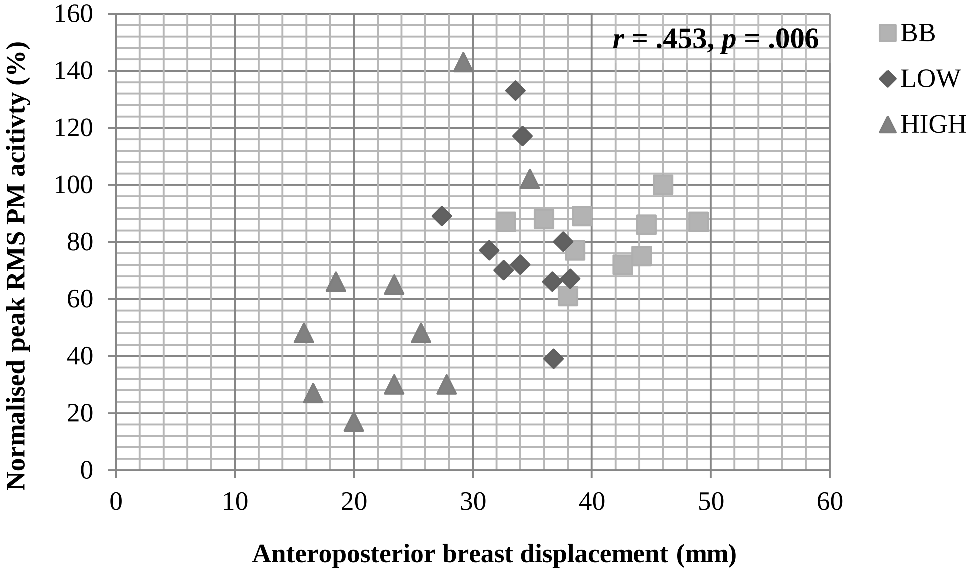

Figure 19. Anteroposterior breast displacement (mm) and peak pectoralis major muscle activity (%) during the first two minute run in the bare-breasted, low and high breast support conditions (n = 10 per condition).

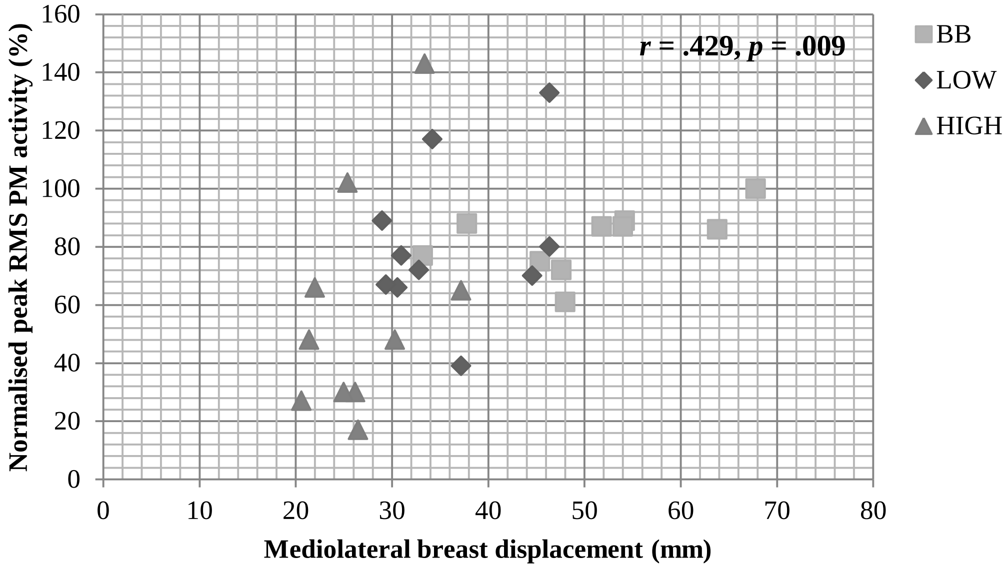

Figure 20. Mediolateral breast displacement (mm) and peak pectoralis major muscle activity (%) during the first two minute run in the bare-breasted, low and high breast support conditions (n = 10 per condition).

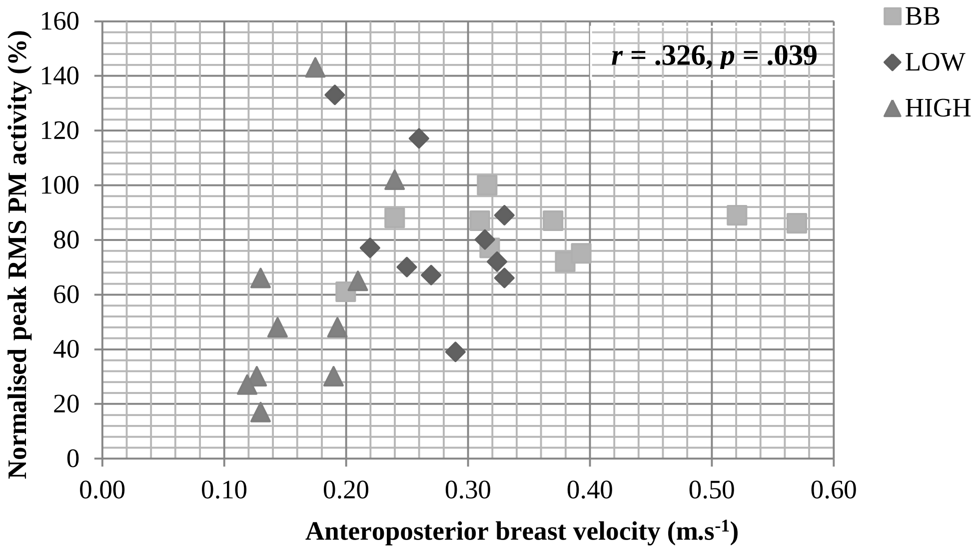

Figure 21. Mediolateral breast displacement (mm) and peak pectoralis major muscle activity (%) during the first two minute run in the bare-breasted, low and high breast support conditions (n = 10 per condition).

The data presented in Table 44 demonstrates the percentage of variance shared by the variables previously correlated. The strongest relationship presenting previously was between the pectoralis major activity and the anteroposterior breast displacement. From the R^2 value we can identify that the anteroposterior breast displacement shares 21% of its variance with pectoralis major activity. Although a moderate relationship was seen between these two variables, 79% of the variability is unaccounted for.

Table 44. Coefficient of determination (%) reported for the three significant correlations.

Correlated variables	Coefficient of determination (R^2)
PM activity and A/P breast displacement	21%
PM activity and M/L breast displacement	18%
PM activity and A/P breast velocity	11%

6.4.1.9 Rating of perceived exertion (RPE)

Participants provided a rating of perceived exertion (RPE) at each interval of the five kilometre run (Figure 22). No significant differences were reported in the participants RPE scores between the low and high breast support conditions. However, as expected, the participants RPE scores increased from the first two minutes to the final kilometre of the run within both breast support conditions.

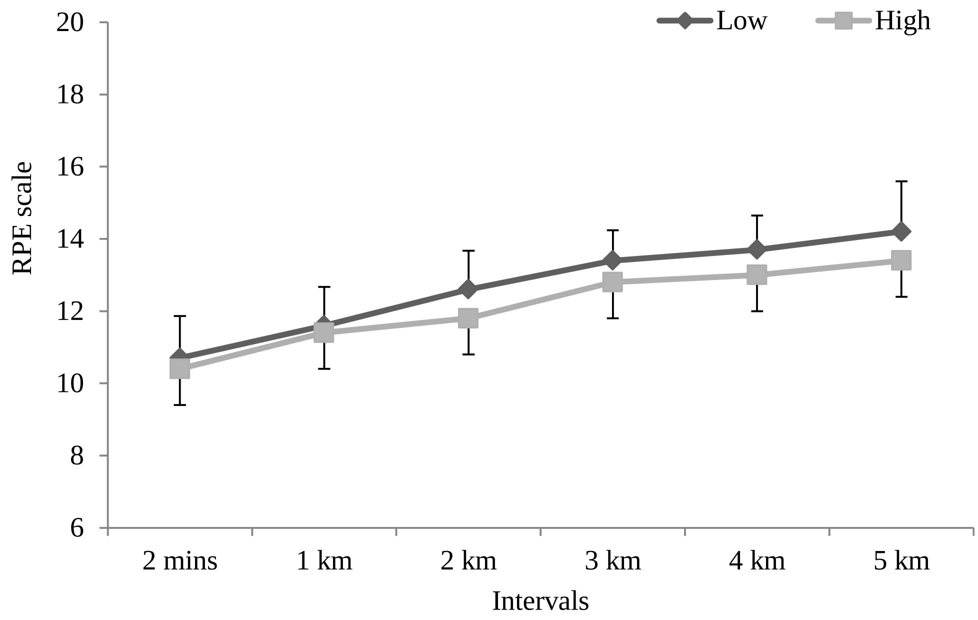

Figure 22. Subjective responses for RPE at each interval of the five kilometre run in the low and high level breast support (n = 10).

6.4.1.10 Effect sizes, power and variance

Effect sizes and power were calculated alongside the one-way and two-way repeated measures ANOVAs conducted in this chapter. Of the significant differences reported within this chapter, the effects sizes were deemed to range from medium (.30) to large (.50). Power ranged from .59 to .95, with the majority of power reported as high (> .80). The magnitude of variance within these data may contribute to the effect sizes presented within this chapter. Within- and between-participant variance in the RMS and *i*EMG data is presented below (Table 45 and 46).

Within-participant variance in the upper-body muscles was quantified over five gait cycles within each interval of the run then averaged across the six intervals, in the three levels of breast support for each participant, and then averaged across the sample (n = 10).

Table 45. Within-participant variance (C_v%) in RMS and *i*EMG muscle activity in the investigated upper-body muscles in three breast support conditions (n =10).

MUSCLES	RMS				*i*EMG			
	BB	LOW	HIGH	MEAN	BB	LOW	HIGH	MEAN
AD	31	33	26	30	23	25	22	23
PM	19	19	21	20	24	17	20	20
UT	19	19	19	19	18	21	19	19
MD	19	20	19	19	18	20	17	18
ES	20	17	17	18	17	18	20	18
LD	19	16	18	18	18	18	16	17
MEAN	21	21	20	21	20	20	19	20

On average the anterior deltoid elicited the greatest within-participant variance in both RMS and *i*EMG processing methods, 30% and 23%, respectively, with the low breast support condition demonstrating the greatest variance in this muscle. The smallest within-participant variance was reported in the latissimus dorsi for both processing techniques, (RMS = 18% and *i*EMG = 17%). Variance in the six upper body muscles ranged from 16% to 33% in the peak RMS data, and from 16% to 25% in the *i*EMG data.

Between-participant variance was calculated for each muscle at each kilometre interval across the sample (n =10) (Table 46). Similar to the within-participant variance, the between-participant variance was greatest in the anterior deltoid for RMS and *i*EMG processing techniques, 39% and 43% on average, respectively. However, the smallest variance was reported in the medial deltoid (27%) for the RMS processing and in the erector spinae (25%) for the *i*EMG technique. This suggests that the processing technique employed may influence the variance in EMG data. Variance in the six upper body muscles ranged from 8% to 68% in the peak RMS data, and from 8% to 64% in the *i*EMG data.

Table 46. Between-participant variance (C_v%) in RMS and *i*EMG muscle activity in the investigated upper-body muscles (n =10).

MUSCLES	RMS				*i*EMG			
	BB	LOW	HIGH	MEAN	BB	LOW	HIGH	MEAN
AD	23	40	55	39	17	64	47	43
PM	13	36	68	39	9	33	45	29
UT	8	44	47	33	10	39	46	32
MD	15	28	38	27	10	43	39	31
LD	10	36	40	29	8	32	37	26
ES	10	37	41	29	8	33	33	25
MEAN	13	37	48	33	10	41	41	31

6.5 Discussion

The aim of the work reported in the present chapter was to examine the effect of breast support on myoelectric activity in the upper body over a five kilometre run. The key findings indicate that the peak amplitude (RMS) and total amount (*i*EMG) of myoelectric activity of the pectoralis major, anterior deltoid and medial deltoid were affected by the breast support worn during short (two minutes) and prolonged (five kilometre) treadmill running. Furthermore, significant moderate relationships (r = .326 to .453) were reported between peak RMS pectoralis major activity and anteroposterior and mediolateral breast displacement, and anteroposterior breast velocity when the data were examined across the three breast supports during two minute running. Only one difference was reported in the *i*EMG of the anterior deltoid over the intervals of the five kilometre run, with a significant increase from the start to the end of the run. No differences in the peak RMS or *i*EMG of the remaining investigated muscles were reported over the five kilometre run. Finally, when the activity of the muscles were ranked within and between each breast support condition, the bare-breasted condition and low breast support elicited greater activity than the high breast support.

The greater amplitudes (RMS) of pectoralis major, anterior and medial deltoid muscles in the lower breast support conditions (bare-breasted and low) may be a result of changes in the neuromuscular system required to maintain the cyclic actions (Winter, 1980). There have been many suggestions as to why an increase in EMG amplitude might be recorded (Dimitrova & Dimitrov, 2003; Holtermann & Roeleveld, 2006; Lowery & O'Malley, 2003), including an increase in motor unit recruitment and/or motor unit firing frequency modulation (Merletti & Parker, 2004), muscle fibre conduction velocity, and recruitment of additional muscle fibres (Basmajian & De Luca, 1985). In order to maintain the muscular action required to sustain mechanical running form, at the selected treadmill speed, it is postulated that one or more of these mechanisms may occur within these upper body muscles in the lower levels of breast support. Furthermore, participants experienced significantly more exercise-related breast pain in the lower breast support conditions compared to the high breast support; therefore, it is hypothesised that increases in tension in the upper body elicited by pain could explain the increase in activity within these muscles. Based upon these findings, it is apparent that the breast support worn, and potentially the exercise-related breast pain experienced, can influence peak RMS values of the pectoralis major, anterior and medial deltoid during treadmill running. Therefore, hypothesis one is partially accepted.

The deltoid muscle is responsible for driving movement of the upper arm at the glenohumeral joint, with the anterior and medial fibres facilitating abduction of the humerus (Smoliga et al., 2010). Furthermore, the anterior deltoid assists the pectoralis major in flexion of the humerus at the glenohumeral joint (Blasier, Soslowsky, Malicky, & Palmer, 1997). Significant reductions in peak RMS values of the anterior deltoid were reported in the low breast support condition during the first two minutes of running, when compared to the bare-breasted condition. Further reductions were reported in the peak RMS medial deltoid when the participants wore the high breast support when compared to the low breast support, during the first two minutes, and the first and second kilometre of the five kilometre run. The significant differences reported in the peak RMS value of these muscles between breast support conditions are consistent with differences reported in arm swing mechanics within the previous chapter of this thesis. It is postulated that the reported increases in muscle activity of the pectoralis major and deltoids, in the bare-breasted and low breast support conditions, are associated with increased tension and restricted ROM in this region due to the magnitude of relative breast kinematics and associated breast pain reported. It is hypothesised that the reductions in the peak RMS activity of the investigated upper body muscles in the high breast support could be beneficial to the female athlete. Griffin, Roberts, and Kram (2003) describe the metabolic cost of walking and running as the cost of generating muscular force. The reductions in the peak RMS values of the pectoralis major and deltoids could indicate a reduction in the muscular force generated, which may indicate reductions in the metabolic demands of these muscles during treadmill running. When considering the metabolic cost of running between breast support conditions, it is important to reflect on the magnitude of difference in peak muscle activity of these upper body muscles between breast support conditions, and consider the potential of these differences to cause significant changes in the cost of running. Future research within breast biomechanics could quantify the metabolic cost of a steady state five kilometre treadmill run in different breast support conditions concurrently with EMG data to investigate this proposed link.

The *i*EMG of the anterior deltoid was the only muscle found to change over the course of the five kilometre run, with a significant increase in total activity reported within the low and high breast supports, from the first two minutes to the fourth kilometre of the run. As mentioned earlier, *i*EMG processing technique can be used to determine total activity of a muscle. With a significant increase in *i*EMG towards the final stages of the five kilometre run, in both low and high breast supports, it is suggested that the reported differences in *i*EMG could be explained by a decreased contraction force. It has previously been stated that an observed increase in *i*EMG at a given intensity is the result of additional recruitment of muscle fibres due to the decreased contraction force associated with fatigue (Abrabadzhiev et al., 2010). However, no differences were reported over the five kilometre run in the remaining investigated muscles. Furthermore, no differences were reported in upper-arm mechanics over the run distance (in the previous chapter); therefore, it is unlikely that fatigue was present in this muscle.

Additionally, the magnitude of variance in the anterior deltoid should be considered when discussing this muscle. High within- and between-participant variance was reported for this muscle, which may be attributed to the position of the electrode and the relative soft tissue movement of this aspect of the arm. The increases in activity of the anterior deltoid within both low and high breast supports at the fourth kilometre are considered inconclusive due to the magnitude of variance. It is important to carefully consider differences based upon these data, and whether they are defined as meaningful. Based upon these findings hypothesis two is rejected.

During the first two minutes of running, the peak amplitude (RMS) of the pectoralis major muscle activity significantly reduced by 23%, and at the fourth kilometre of the five kilometre run by 39%, in the high breast support condition when compared to the low breast support condition. The reduction in pectoralis major muscle activity in high breast support is in accordance with previous findings by Scurr et al., (2010b). Scurr et al., (2010b) postulated that the increase in *i*EMG of the pectoralis major, associated with the reduction of external breast support, may indicate that this muscle is providing structural support to the breast tissue.

Following on from this finding, a secondary aim of the current study was to explore the relationship between the pectoralis major and multiplanar breast kinematics, and to establish if the pectoralis major plays an anatomical role in the support to the breast tissue. It was hypothesised that a significant positive relationship would be reported between these two variables. This hypothesis was based upon results from a previous abstract presented by Scurr et al (2010b) and would support the assumption that the pectoralis major is providing some structural support to the breast. Moderate positive relationships were reported between peak RMS pectoralis major activity and anteroposterior displacement, mediolateral displacement and anteroposterior breast velocity. This finding suggests that as anteroposterior and mediolateral breast kinematic variables increase, so does the pectoralis major activity. In comparison to literature examining the role of muscles for damping the vibrations and movement of soft tissue (i.e. greater muscle activity reduces the soft tissue movement) (Wakeling, Liphardt, & Nigg, 2003; Wakeling, Nigg, & Rozitis, 2002), it is interesting to see the opposite relationship shown with the soft tissue of the breast and the pectoralis major muscle. The majority of the literature examining the role of muscles for damping soft tissue vibrations has been conducted on the lower extremities. For example Wakeling et al., (2003) examined the influence of vastus lateralis, biceps femoris (long head), tibialis anterior and lateral gastrocnemius and the associated soft tissue vibrations during heel strike. The breast encompasses glandular tissue and the connection site of this tissue to the pectoralis major is unique, therefore it cannot be directly compared to the soft tissue previously explored in the lower limbs. It is suggested that the increase in pectoralis major activity with reduced breast support is a protective response, in an attempt to reduce any potential damage to this important tissue. With only three breast kinematic variables demonstrating a moderate relationship with pectoralis major activity, hypothesis three cannot be confirmed within the present study, and therefore is partially accepted.

When assessing the normalised activity of the six investigated muscles within each breast support condition, the bare-breasted condition elicited the greatest percentage of activity when averaged across the six muscles in both processing techniques. It could be suggested that the different breast support conditions may have elicited different recruitment of muscles when these were ranked. However, it is important to consider the variance associated with these data to determine meaningful differences. When examining the total *i*EMG the two most active muscles were the upper trapezius and the erector spinae. During running the upper trapezius supports the glenohumeral joint, incorporating elevation of the scapular and humerus, and assists with humerus adduction during the mechanics of arm swing (Basmajian & De Luca, 1985). Moreover, Fernandez Ballestros, Buchthal, and Rosenfalck (1965) reported continual electrical activity from the upper aspect of the trapezius during the walking gait cycle. The erector spinae, also reported as one of the most active muscles within the current study, supports the upper body posture through trunk extension and flexion, with thorax pitch dependent upon the support of this muscle. Due to their important postural and functional roles during running, it is unsurprising that these two upper body muscles are reported as most active during the running gait cycle. Furthermore, it is surprising that the changes in the ROM of thorax pitch between breast support conditions, and the increase in thorax flexion within the low breast support over the five kilometre run, reported in the previous chapter (chapter four, section 4.4.2), did not influence the EMG of the erector spinae. However, the change in ROM and peak flexion was relatively low (1º to 1.3º), and therefore may not have placed a greater demand on the erector spinae.

Participants provided ratings of perceived exertion (RPE) at each interval of the five kilometre run in the low and high breast support conditions. The RPE scores increase from the first two minutes to the fifth kilometre interval, which suggests that as the participants progressed through the run they perceived it as harder. No significant differences were reported in the RPE scores between the low and high breast supports, which is similar to the findings of McGhee et al., (2012). In light of these findings, hypothesis four was rejected.

Within the current study soft tissue movement artefact and potential increase in low-pass filtering due to the breast tissue was an important consideration for the pectoralis major EMG signal. The electrode placement for the pectoralis major muscle was positioned at the pars clavicularis in an attempt to reduce the influence of the breast tissue on this signal. Recommendations for the pectoralis major electrode placement are sparse in the literature. Król et al., (2007) examined the effect of electrode placement on the pectoralis major and proposed that to achieve the greatest *i*EMG activity, the electrode should be positioned medially on the abdominalis part of the muscle; however these data were collected from male participants and examined during an isometric barbell bench press. Currently no papers detail the influence of breast tissue on the output EMG signal from different sites of the pectoralis major for female participants during dynamic exercises. These data would be extremely beneficial for this area of research, with standardised electrode placement likely to reduce the chance of high variability among these data. Within the current study the pectoralis major muscle demonstrated an average within-participant variance of 20% when processed using the RMS method, with the greatest variance reported during the high breast support condition. These data demonstrate that an increase in magnitude of breast kinematics reported in the bare-breasted condition did not cause greater levels of variance, and is considered as an appropriate position when examining female participants.

Although differences in upper body muscle activity have been reported within the current study between breast support conditions during treadmill running, there are limitations that should be acknowledged when interpreting these data. Electromyographical signals are renowned for being noisy as a result of intrinsic and extrinsic variables (De Luca, 1997). Muscle fibre type, diameter, depth and location and amount of tissue between the muscle and the electrode are the most commonly reported intrinsic factors, which cannot be controlled (Burden & Bartlett, 1999). The orientation, location, area, and shape of the electrode, and the distances between electrodes are all extrinsic factors and can be influenced by the researcher (Burden & Bartlett, 1999; De Luca, 1997). Within the current study the extrinsic factors were considered in depth and action taken to reduce the influence of these factors between testing sessions, such as; set up (e.g. equipment selection, wire movement artefact reduced by securing the wires and pack to the treadmill), procedures (e.g. skin and electrode preparation), and electrode positioning (e.g. standardisation using SENIAM procedures, taping of electrodes). Due to the repeated measures design of the study, the positioning of the electrodes between testing session is considered a limitation of this study. Due to the time between sessions (up to 72 hours) applying an outline of the electrodes with an eye liner pencil was not practical. Therefore, the reliability of electrode reapplication for all muscles, except the pectoralis major, was based upon the standardised anatomical positions recommended by SENIAM. Difference in electrode location and orientation could result in measurement of different motor units in a given muscle, and may result in large variations in the signals recorded (Burden & Bartlett, 1999; De Luca, 1997; Hermens, Freriks, Disslehorst-Klug, & Rau (2000); Reaz, Hussain, & Mohd-Yasin, 2006). In an attempt to reduce this factor a normalisation method was applied to the data.

Another factor to consider is the influence the laboratory conditions had on the EMG data. When the skin becomes moist with sweat, which is conductive, the EMG signal can be affected (De Luca, 1997), with the signal amplitude reported to decrease, deteriorations in the signal to noise ratio and changes at the skin surface could filter out higher frequency components. This is a concern within the current study, the participants were running for approximately 32 minutes in a laboratory, and at times moisture on the skin surface and around the electrode was visible.

6.6 Conclusion

The findings of the current chapter are novel and detail the effect of breast support on EMG of upper body muscles during a five kilometre run. The findings indicate that the breast support worn can influence the activity of the pectoralis major, anterior and medial deltoids during short (two minutes) and prolonged (five kilometre) treadmill running. It is suggested that wearing a high breast support during running can significantly reduce the peak amplitude of EMG activity in these three upper body muscles when compared to lower levels of breast support. This is the first study to identify this, and these findings could have significant implications for female distance runners in terms of metabolic cost, potential onset of muscular fatigue, and subjective feelings of exertion. These reductions were reported during the initial stages (first two minutes, first and second kilometre) of the five kilometre run. Furthermore, these differences indicate a link between the mechanical differences reported in the previous chapter, with differences reported in the muscles driving these segmental movements.

The relationship between the EMG activity of the pectoralis major and breast kinematics was examined within the current study, and demonstrate a moderate positive relationship between the peak RMS pectoralis major activity and anteroposterior and mediolateral breast displacement and anteroposterior breast velocity. Although the amount of anatomical support provided by the pectoralis major to the breast tissue remains unclear, the results of the current study suggest that these two variables are related. The work reported in this chapter provide crucial provisional evidence in an area that has not been reported previously, and could have implications for earlier onset of muscular fatigue in lower levels of breast support in prolonged treadmill running, which may negatively influence running performance.

The work presented within chapters four and five highlight the link between running kinematics and muscle activity, and provide a better understanding of the effect of breast support conditions on running biomechanics. The integrated design of the work within this thesis ensures simultaneous changes in breast and body biomechanics, as a result of the breast support worn, are reported for the same participant, adding strength to these findings. Further exploration of the relationships between the dependent measures investigated within this thesis will provide a holistic biomechanical view of the effect of breast support on running biomechanics. Chapter seven aims to explore the relationships between the breast and body parameters examined within this thesis.

CHAPTER SEVEN.

RELATIONSHIPS BETWEEN BREAST AND BODY BIOMECHANICS

7.1 Introduction

The overall aim of this thesis was to investigate the effect of breast support on breast and body biomechanics during a five kilometre treadmill run. The work presented within the previous chapters (chapter four to six) of this thesis has identified the effect of breast support on multiplanar breast displacement, velocity, acceleration and approximated force, exercise-related breast pain, upper and lower body 3D joint kinematics, and upper body EMG during a five kilometre treadmill run.

In line with previous literature (Scurr et al., 2010a; White et al., 2009), as breast support increased, the magnitude of multiplanar breast displacement, velocity, and acceleration significantly decreased, with up to 75% reduction in the high breast support condition. Approximated breast force also decreased as breast support increased, further promoting the use of a high breast support during running. The reduction in multiplanar breast kinematics and approximated breast force when wearing a high breast support were highly correlated to reductions in exercise-related breast pain. Exercise-related breast pain has previously been related to the magnitude of breast displacement (Mason et al., 1999) and velocity (McGhee et al., 2007; Scurr et al., 2010a), however, the results of the current programme of work suggest it is more closely related to the acceleration of the breast. Presentation of breast acceleration and approximated breast force during exercise will help to inform future research focussing on the mechanical properties of the tissues supporting the breast.

An interesting and unique aspect of the work presented in chapter three was the significant increases in breast kinematics over the distance intervals of the five kilometre run in both the low and high breast support conditions. The anteroposterior, mediolateral and vertical displacement (mm), vertical velocity ($m.s^{-1}$), vertical acceleration ($m.s^{-2}$), and vertical force (N) of the breast significantly increased between the start (two minutes) and the end (fourth and fifth kilometres) of the five kilometre run. These unique findings help inform methods for breast biomechanics research and testing protocols for sports bras, ensuring the examination of breast biomechanics and the effectiveness of a product is examined within an externally valid environment. Furthermore, these data suggest that a combination of factors may influence changes over a prolonged run, including potential strain to the supporting tissues (Scurr et al., 2009b; Scurr, White, Milligan, Risius, Hedger, 2011), changes in the performance of the material properties of the bra (i.e. elastane) due to heat and sweat saturation (Ayres, White, Hedger, & Scurr, 2013) and, most prevalent to the current research; changes in running kinematics (Hardin et al., 2004; Williams et al., 1991), specifically the thorax segment (Haake & Scurr, 2010).

Participants exhibited significantly different kinematic running profiles when wearing different breast supports. The high breast support elicited the following kinematic profile when compared to the low breast support; greater step length, greater thorax yaw (ROM), less thorax pitch, less axial rotation of the pelvis, less extension of the upper-arm, greater abduction of the upper-arm, less peak adduction of the forearm, less flexion of the forearm, and less adduction/abduction of the thigh. Many of these kinematic variables have been associated with economical running styles (Saunders, Pyne, Telford, & Hawley, 2004; Williams & Cavanagh, 1987).

Reductions in the peak RMS activity of the pectoralis major and medial deltoid were reported in the high breast support compared the lower levels of breast support. The significant differences reported in the activity of these muscles suggests that certain neuromuscular adaptations occur when running in different breast support conditions, which may be associated with an altered metabolic cost of running. Furthermore, the alterations in peak RMS and *i*EMG of the anterior deltoid, medial deltoid, and the pectoralis major, between breast support conditions, align with the differences reported in the running kinematic parameters; such as differences in arm swing mechanics. These data provide a holistic biomechanical view of the female runner during a five kilometre run in different breast support conditions.

The research design of the current programme of work (within-participant, repeated measures) ensured influential variables previously presented across a range of publications were included in this integrated design, and relationships could be investigated between variables. It is of interest to determine the relationship between the dependent variables examined within this thesis. For example, it was commonly hypothesised that the magnitude of breast biomechanics and exercise-related breast pain were influential to changes in running biomechanics, and that certain running kinematic variables would be related to changes in muscle activity.

7.2 Aims and research hypotheses

In order to confirm these hypothesised relationships, the aim of the current chapter was to explore the relationships between breast and body biomechanics variables presented in chapters four to six, which were significantly affected by breast support during a five kilometre run.

H_1 – The magnitude of breast kinematics will be significantly correlated with key running kinematic parameters shown to vary between breast supports.

H_2 – The magnitude of breast kinematics will be significantly correlated with the activity of the three muscles which varied between breast supports.

H_3 – Key running kinematic parameters and activity of the three key muscles will demonstrate a significant correlation with exercise-related breast pain.

H_4 – Key running kinematics parameters will be significantly correlated with the three key muscles which varied between breast supports.

7.3 Methods

7.3.1 Data analyses

All participant information, data collection, and data processing procedures have been reported in the previous chapters of this programme of work. The following dependent variables measured in this programme of work demonstrated a significant difference between breast support conditions; multiplanar breast kinematics, exercise-related breast pain, thorax pitch, thorax yaw, pelvic tilt, pelvic obliquity, axial rotation of the pelvis, upper-arm abduction, upper-arm extension, forearm adduction, forearm flexion, thigh adduction/abduction, shank flexion, step length, peak RMS activity of the pectoralis major, and the anterior and medial deltoid.

Using Pearson correlations (*r*) the following relationships were examined between mulitplanar breast kinematics and thorax pitch, thorax yaw, pelvic tilt, pelvic obliquity, axial rotation of the pelvis, upper-arm abduction, upper-arm extension, forearm adduction, forearm flexion, thigh adduction/abduction, shank flexion, and step length. Secondly, Pearson correlations were performed to assess the relationship between multiplanar breast kinematics and the RMS activity of the muscles influenced by breast support; these were the pectoralis major, anterior deltoid, and the medial deltoid.

Thirdly, Spearman correlations were performed to assess the relationship between exercise-related breast pain and both kinematic running variables and RMS muscle activity. Following this, Pearson correlations were performed to assess the relationships between the kinematic running variables and the EMG for the three muscles of interest. These correlations determined the most influential biomechanical parameters examined within the current programme of work, providing a holistic view of the effect of breast support on the biomechanics of the female runner.

The following criteria for the strength of the relationships examined within this chapter was followed; a small relationship = ± .10, medium = ± .30, and a large relationship = ± .50 (Field, 2009).

7.4 Results

The relationships between multiplanar breast kinematics, running kinematic parameters, and EMG of upper body muscles are presented in Figure 23, with significant correlations presented on the connecting lines. The strongest relationship presented in this schematic was between breast kinematics and thorax yaw (r = -.697, p = .001).

The only dependent variable to correlate with breast pain was the pectoralis major muscle activity (r = .535, p = .002). Demonstrating an increase in exercise-related breast pain as pectoralis major muscle activity increased.

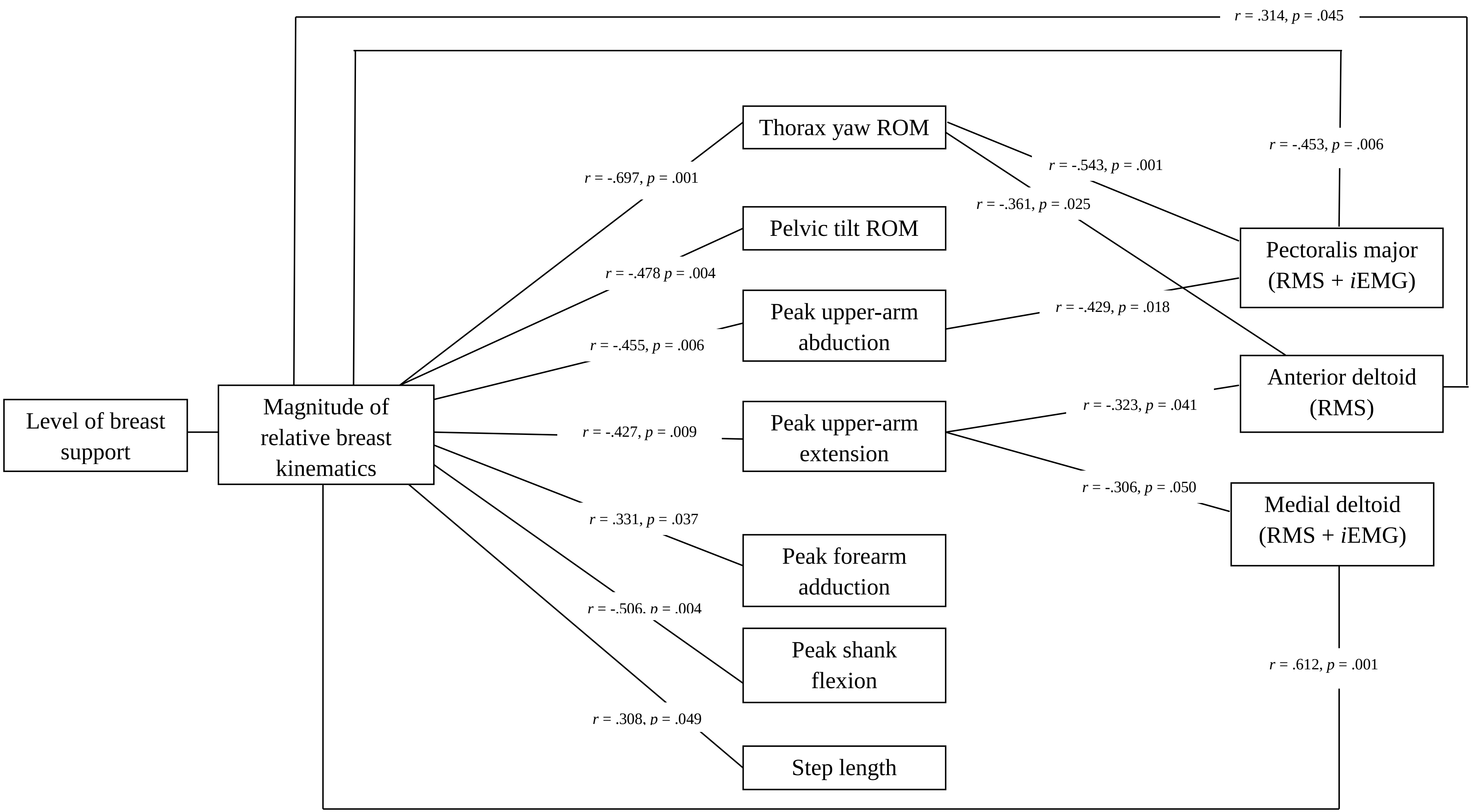

Figure 23. Schematic of the significant relationships between breast and body biomechanics examined within this programme of work.

7.5 Discussion

The aim of the current chapter was to explore the relationships between the key dependent variables in this programme of work. The novel findings highlight the strength of the integrated approach employed within this research. The key findings demonstrated that i) mulitplanar breast kinematics is significantly related to both kinematic and EMG parameters ii), pectoralis major activity is related to exercise-related breast pain, and iii) certain running kinematic parameters are significantly related to upper body EMG.

Multiplanar breast kinematics was significantly correlated to the following running kinematic parameters; thorax yaw, pelvis tilt, peak upper-arm abduction, peak upper-arm extension, peak forearm adduction, peak shank flexion, and step length. These findings support hypothesis one, and the link between breast and body biomechanics. Thorax yaw demonstrated the strongest relationship between breast kinematics and body kinematics, which emphasises the relationship between the breast and the thorax. Many publications have emphasised the influence of the thorax on the independent kinematics of the breast (Scurr et al., 2009; 2010; Haake & Scurr, 2010), however, these publications have neglected to report the kinematics of the thorax, or the relationship between the two. The work reported within this thesis demonstrates the importance of quantifying thorax kinematics alongside breast biomechanics and provides empirical evidence that suggests thorax yaw drives breast kinematics.

Not only did multiplanar breast kinematics demonstrate significant relationships to running kinematics, but also with pectoralis major, anterior deltoid, and medial deltoid muscle activity. The relationship between multiplanar breast kinematics and pectoralis major activity is of interest to breast biomechanics research due to the anatomical connections between the breast tissue and this muscle. However, the relationship between breast kinematics and the anterior and medial deltoid was a unique finding. Bennett (2009) identified higher muscle activity in the cervico-thoracic region during different static positions due to participant's breast sizes, and suggested that complaints of neck, back, and shoulder pain were as a result of increased tension placed on these muscles due to the large breast mass. Although the participant's bra sizes in the current programme of work fell within the cross-grading range of the UK average, and would not be classed as 'larger' breasts, it is of interest to consider the magnitude of independent breast kinematics on the activity of muscles in and around this region, since increased feelings of tension and increases in muscle activity may negatively impact upon sporting performance. The correlations reported between breast kinematics and muscle activity allows us to accept hypothesis two.

Exercise-related breast pain was identified as a negative factor to female runners, which could substantially impact on an individual's running biomechanics. The only dependent variable to correlate with breast pain was the pectoralis major activity, indicating a significant increase in breast pain with an increase in muscle activity. With only one variable demonstrating a significant relationship to breast pain, hypothesis three is only partially accepted. Though no other kinematic parameters correlated with breast pain, it is suggested that it should not be disregarded as an influential factor for female runners. The magnitude of independent breast movement and exercise-related breast pain experienced, in the different breast support conditions, were frequently mentioned by the participants, suggesting that these factors influenced their running biomechanics.

The link between muscle activity and kinematic parameters was highlighted within this programme of work, with changes in the activation of muscles associated with changes in running mechanics. The relationship between the running kinematics and EMG activity of upper body muscles that were affected by breast support were explored. A significant negative relationship was reported between thorax yaw and both pectoralis major and anterior deltoid muscle activity, with the ROM of thorax yaw decreasing as muscle activity increased. Furthermore, arm swing mechanics demonstrated significant negative relationships with upper-body muscles, with peak upper-arm abduction correlated with pectoralis major, and peak upper-arm extension correlated with both the anterior and medial deltoid. These relationships demonstrate the link between changes in segmental running kinematics and the muscle activity driving these changes, and allows hypothesis four to be accepted.

7.6 Conclusion

The relationships presented within this chapter illustrate the most influential biomechanical variables examined within this programme of work. The key findings of this chapter indicate multiplanar breast kinematics are related to both running kinematic parameters and upper body EMG, pectoralis major activity is related to exercise-related breast pain, and key running kinematics parameters are related to the EMG of associated upper body muscles. The integrated approach provides a more holistic understanding of the changes in female running biomechanics dependent upon the magnitude of multiplanar breast kinematics, which is directly influenced by the breast support worn.

CHAPTER EIGHT. GENERAL DISCUSSION AND CONCLUSION

The aim of this programme of work was to investigate breast biomechanics, upper and lower body running kinematics, and muscle activity in different breast supports during a five kilometre treadmill run. Research regarding the effect of breast support on running biomechanics is minimal. The importance of this research is emphasised when considering the percentage of females exercising in reduced breast supports (Bowles et al., 2011), and influential negative factors associated with lower breast support conditions, such as greater relative breast kinematics and exercise-related breast pain (Mason et al., 1999; Scurr et al., 2009; 2010; 2011; White et al., 2009). The results of this programme of work are novel and support the hypotheses that a high breast support can significantly reduce breast kinematics, exercise-related breast pain, ensure running kinematics remain unchanged over a prolonged run, and finally reduce upper body muscle activity. The research design employed for this programme of work ensured a holistic biomechanical view of the female runner was reported, with the data collected over two testing sessions, employing the same participants for the repeated trials. Furthermore, the distance of the run examined extends the breast biomechanics research previously published and ensured the research study possessed high external validity.

The first aim of this thesis was to investigate the effect of breast support on multiplanar breast kinematics during a five kilometre run, and to assess the magnitude of within- and between-participant variance within these data. In line with previous literature, the high breast support was reported to significantly reduce relative multiplanar breast kinematics and breast force (up to 75% reduction) during treadmill running, which was correlated with reductions in exercise-related breast pain. It was suggested that the reduction in approximated breast force under the high breast support condition ensured reduced loads subjected to the intricate structures of the breast, and reduce potential risks of stress and strain on these tissues during running. These findings further promote the use of a high breast support for a female runner, and support the notion that wearing a high breast support can reduce the relative kinematics of the breast and exercise-related breast pain, which in turn may improve a females comfort when exercise, and may enable runners to maintain long distance training regimes.

The unique progression of assessing breast kinematics over a five kilometre run ensured the current research study possessed higher external validity than previous research in the area. The results demonstrated significant increases in the magnitude of breast kinematics as the runners progressed through the intervals of the five kilometre run in both low and high breast support conditions. These novel findings suggest that neither the low or high breast support provided consistent magnitudes of support to the breast over this run distance. This result has implications for product design, materials used, and testing protocols for future breast biomechanics research and product evaluation. Previous validation of sports bras have been carried out over short run durations (two minutes to five minutes) (Starr et al., 2005), these testing protocols lack external validity, and the results of this work suggest that this may lead to inappropriate promotion of the effectiveness of a sports bra for longer durations and run distances. Females exercising for 30 minutes or more may therefore experience a reduction in breast support and an associated increase in exercise-related breast pain, which could impact on other biomechanical measures such as running kinematics and muscle activity (as shown in this research). These results address the first discreet objective of this thesis and provide the first quantitative description of breast biomechanics over a prolonged running distance.

Many publications reporting breast kinematics are emerging within the literature. These studies rarely present effect sizes or power, and to date only two studies have reported the variance in these data. Effect sizes and power are influential statistics which facilitate the interpretation of statistical analyses and help to determine sample sizes for future research (Hoenig & Heisey, 2001; Levine & Hullett, 2002). Despite this, important and wide reaching conclusions are drawn from the breast biomechanics data presented previously, including applications to product design, breast pain assessment, sports performance affects and more. The data presented in this thesis are the first to examine multiplanar breast kinematics over a common running distance (five kilometre). Therefore, assessing the magnitude of variance in multiplanar breast kinematics provided a description of the characteristics of these data. Moreover, defining the different components of total error in these data facilitate the interpretation of the results presented.

The smallest magnitudes of difference in breast kinematics were found to exceed the total variance in these data and were therefore confirmed as meaningful differences. In light of these findings, it is recommended that the smallest differences reported in breast kinematics exceed the total error. These findings ensure recommendations of data collection procedures (i.e. accuracy and precision of cameras and laboratory set up), and sample sizes can be proposed for future work. It is suggested that a sample of ten or more is large enough to identify differences in breast kinematics within and between breast support conditions during short and prolonged treadmill running. The interpretation and discussion of these data address the first objective of this thesis.

The kinematic analyses revealed differences in running gait associated with the breast support worn. Specifically, it was shown that when wearing a high breast support the following kinematic profile was reported when compared to the low breast support; greater step length, greater ROM in thorax yaw, less ROM in thorax pitch, less ROM in axial rotation of the pelvis, less peak extension of the upper-arm, greater peak abduction of the upper-arm, less peak adduction of the forearm, less ROM of flexion at the forearm, and less ROM in adduction/abduction of the thigh.

Peak flexion of the thorax significantly increased from the start to the end of the run when participants wore the low breast support. Greater peak flexion of the thorax has previously been shown to increase the cost of running, and been associated with a less economical running style (Saunders, Pyne, Telford, & Hawley, 2004; Williams & Cavanagh, 1987). With the ROM of thorax pitch and peak flexion of the thorax reduced in the high breast support condition, it is suggested that the high breast support could be beneficial to female runners, and potentially reduce costly running kinematics associated with the thorax. Additionally, the high breast support significantly reduced arm swing mechanics including reduced peak upper-arm extension. Arm motion that is not excessive, dependent upon the running velocity, has also been associated with economical running styles; therefore, this difference may indicate a potential energy saving mechanism when wearing the high breast support. These findings have implications for psychological, biomechanical and physiological aspects of running, which have previously been related to changes in running kinematics such as perceptions of effort and exertion (Milani, Hennig, & Lafortune, 1997; Messier & Cirillo, 1989), muscle activity (Nilsson, Thorstensson, & Halbertsma, 2008), running economy (Saunders et al., 2004; Cavanagh & Williams, 1982), and the metabolic cost of running (Candau et al., 1998; Williams and Cavanagh, 1987; Williams, 1990). When considering the differences in running kinematics reported within this thesis, it could be hypothesised that the mechanical running profile within a high breast support may be advantageous when compared to a low breast support, although this requires experimental confirmation via physiological testing. This work provides the first full body kinematic description of the female runner in different magnitudes of breast support. These data help determine the effect of breast support on running biomechanics, and answer the second objective of this thesis.

The third objective of this programme of work was to examine the effect of breast support on six upper body muscles central to running. The neuromuscular system drives the complex 3D movement of body segments, therefore alterations in running kinematics should be driven by changes in muscle activity. The findings of this research demonstrated that a high breast support significantly reduced the amplitude in myoelectric activity in the pectoralis major, anterior and medial deltoid during treadmill running. The muscles which demonstrated significant differences between breast support conditions are central to the segments which demonstrated mechanical changes, supporting the global alterations to running gait in different breast supports. These findings have large implications when considering the overall effect of different breast supports on the female runner. With reduced peak muscle activity in the pectoralis major, anterior and medial deltoid when running in the high breast support, it is assumed that these differences are beneficial to the female runner, and may indicate that the lower breast supports require a number of changes to the neuromuscular system to maintain the cyclic actions of these muscles during running.

The findings of this programme can be applied to three main areas. The application to changes in running biomechanics and outcome measures of running performance, informing sports bra design specifically for running, and developments in both sports bra testing and breast biomechanics research protocols. Firstly, it is important to consider the implications of the research findings for female running performance. Alterations in running kinematics have been linked to both detrimental and advantageous changes in physiological measures of running performance, such as running economy and metabolic cost, and outcome measures of performance, such as finish time and pacing. When participants completed the five kilometre run in the high breast support, the kinematic profile was more closely related to economical upper and lower body kinematics defined within the literature (Saunders et al., 2004; Cavanagh & Williams, 1982). Saunders et al., (2004) summarised the key mechanical variables that have been shown to affect running economy. Reduction in arm mechanics, faster rotation of the shoulders in the transverse plane, reduced forward lean of the upper body, and reduction in vertical oscillation of the centre of mass, have all been highlighted as influential mechanical variables related to better running economy. Not only did the participants demonstrate kinematics that were more closely related to an economical running style, significant reductions were reported in peak muscle activity of three upper body muscles during the five kilometre run in the high breast support condition. The reductions in the peak RMS values of the pectoralis major and deltoids demonstrate a reduction in the muscular force generated, which may indicate reductions in the metabolic demands of these muscles during treadmill running. Further research is required to confirm this; however, these data demonstrate a potential benefit for female runners when wearing a high breast support compared to a low breast support. Mechanical alterations that lead to a runner using less energy and a reduced metabolic cost at a given speed are advantageous to performance (Williams, 1990).

Secondly, the data collected during this programme of work can inform the development of sports bras designed for running. Three key data sets from the current programme of work help to provide recommendations for this. Firstly, the magnitude of breast kinematics in each direction of movement and the distribution of movement were monitored during barebreasted running. These results inform bra manufacturers of the locations of the bra that require more/less structural support. The greatest magnitude of breast kinematics, when running without breast support, occurred in the vertical direction, followed by the mediolateral, and the smallest movement in the anteroposterior direction during treadmill running. Secondly, the description and quantification of thorax and upper body kinematics during running informs sports bra manufacturers of the movement that drives the independent movement of the breast, with the greatest rotation of the thorax occurring about the vertical axis. This would suggest that sports bras designed for running may need to provide greater medial and lateral support panels to reduce the breast movement in these directions. Finally, it is also important to consider the greatest increase in breast kinematics over the five kilometre run within the low and high breast support conditions. These data help to determine which components of the bras examined within the current research (UK best-selling sports bra) provide the smallest resistance to movement over a prolonged run. The magnitude of the increase in the vertical breast kinematics was the greatest within both breast support conditions over the five kilometre run. Based upon these three sets of data, it is suggested that sports bra designed for prolonged running should incorporate stiffer materials across the width of the bra, and more structural support in the medial and lateral panels. In addition, it is suggested that sports bra for running have greater coverage at the superior aspect of the bra to restrict vertical breast kinematics effectively.

Thirdly, the work presented in this thesis has implications for research protocols. Developments and progressions in bra testing and breast biomechanics methodologies are crucial to this research area. It is important to note that a vast majority of the significant increases in breast kinematics were reported at the third kilometre interval of the five kilometre run, when compared to the first two minutes of running. These increases occurred from 18 to 20 minutes of treadmill running. It is suggested that protocols designed to quantify the performance of a sports bra should incorporate a longer duration run, and based upon the current research findings, it is suggested that participants should run for a minimum of 20 minutes to ensure the performance of a sports bra can be monitored as closely as possible to an externally valid environment. It should be noted that the majority of breast kinematics were at the greatest magnitude at the fifth kilometre interval, and it is hypothesised that breast kinematics may continue to increase over a run exceeding this distance (e.g. 10 km), however this is currently unknown, and requires confirmation.

The findings of the current programme of work indicate significant differences in multiplanar breast kinematics, exercise-related breast pain, upper and lower body running kinematics, and upper body muscle activity between breast supports. The high breast support provided superior support to the breast, significantly reduced exercise-related breast pain, elicited more economical running kinematics, and significantly reduced muscle activity. These results suggest that wearing a high breast support may be advantageous to a female runner, and ensure that negative factors associated with lower levels of breast support are reduced, enabling a female to exercise with minimal restriction or discomfort. The research design of this programme of work enabled relationships between crucial biomechanical measures to be explored, providing a holistic view of the effect of breast support on the female runner. Relationships were identified between the magnitude of breast kinematics, which was governed by the breast support worn, and the following dependent measures; exercise-related breast pain, upper and lower body running kinematics, and muscle activity. Furthermore, certain running kinematics demonstrated significant relationships to muscle activity, demonstrating changes in more than one biomechanical measure, and signifies the value of an integrated study design.

8.1 Delimitations and limitations

Within this section, the delimitations and limitations of the programme of work are discussed. Delimitations were defined as an aspect of the research that was under the control of the researcher and were considered when implementing the boundaries of the work. Limitations were defined as an aspect of the research that was out of the control of the research that could have potentially influenced the outcome.

A consideration for this programme of research was the decision to monitor the dependent variables of interest during treadmill running and not during overground running. The treadmill has long been used to provide a standardised, reproducible work performance task in biomechanical and physiological research (Nelson, Dillman, Lagasse, & Bickett, 1972). Due to the sensitive nature of the current research area (breast biomechanics) and the video analysis (optoelectronic) required for the collection of breast kinematics, the treadmill and laboratory set up was deemed as most appropriate. However, it is acknowledged that there are differences between treadmill and overground running. Within gait literature, the following differences have been reported between the treadmill and overground running; running kinematics, including step/stride characteristics (Alton, Baldey, Caplan, Morrissey, 1998), hip, knee, and ankle joint angles (Alton et al., 1998; Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007; Schache et al., 2001), running kinetics such as peak GRFs and joint moments (Riley et al., 2007). However, it is important to note that the magnitude of these differences are small when compared to the magnitude of variability in gait mechanics, and Riley et al., (2007) concluded that the mechanics of overground and treadmill running are similar. It is acknowledged that the findings of the current study cannot be directly applied to overground running.

Additionally, it has been shown that a 1% gradient on the treadmill can replicate the energetic cost of overground running for up to five minutes of running at speeds between 2.92 and 5 $m.s^{-1}$ (Jones and Doust, 1996). The potential differences in kinematic parameters between a 1% gradient and 0% gradient on the treadmill were considered for this programme of work, with data collected and presented in appendix C. With no differences reported between key kinematic variables of interest, 0% gradient was selected and it is suggested that either could be implemented for future work in this area.

Another consideration of this programme of work and research area is the absence of a barebreasted familiarisation session prior to data collection. Certain publications within the barefoot running research suggest and employ a familiarisation period prior to collection (Bonacci et al., 2013; Warne & Warrington, 2012). The aim of the familiarisation is to ensure any differences reported between conditions are not due to the participant familiarising their running mechanics to the barefoot running condition. The benefit of a barefoot familiarisation is clear for the validity of research in this area, however, it is currently unknown if a barebreasted familiarisation should be incorporated into breast biomechanics research. It is unknown if breast and body biomechanics will differ after repeated bouts of barebreasted running, and in order to answer this question and promote the use of a familiarisation within this research, great consideration it warranted due to the discomfort and pain reported under this condition. Furthermore, potential short and long term damage to the breast tissues could the results of repeated bouts of barebreasted running, and therefore this is an important ethical consideration for breast biomechanics research.

In line with classical mechanics, an assumption that is widely accepted within biomechanics research, that is focussed upon the quantification of the position and orientation (POSE) of body segments, is that the segment is non-deformable and therefore rigid (Cappozzo, Della Croce, Leardini, & Chiari, 2005; Chiari, Della Croce, Leardini, & Cappozzo, 2005). It is well established that when passive markers are positioned on the soft tissue of the body to represent the movement of the skeleton, soft tissue artefact is present and may influence the data (Cappozzo et al., 2005). To overcome this artefact, stringent processing techniques, such as segment optimisation, are employed within analysis programmes (Visual3D) (Lu & O'Connor, 1999). The segment optimisation technique employs a least squares method whereby the distance between the measured and modelled marker positions are minimised. Without these analyses methods, accurately quantifying the movement of the underlying skeleton is extremely difficult. The work detailed in the third chapter of this thesis quantified the multiplanar kinematics of the breast relative to the thorax. The thorax segment was defined using three non-collinear passive markers positioned on the suprasternal notch, and the left and right anterioinferior aspect of the 10th ribs as previously reported by Scurr et al., (2009a). The thorax is an extremely difficult segment to model/define due to the change in depth associated with breathing. One specific limitation for the current area of research is that the desirable anatomical landmarks recommended by ISB for modelling this segment (Wu et al., 2005) are obscured by many breast support garments. Therefore, different segment definitions were required for this programme of research. The employed marker set for the thorax segment within the current programme of work may be subject to greater magnitudes of STA and be at greater risk of deformability due to breathing mechanics, compared to the ISB marker set, and therefore may heighten the chance of error in the POSE of this segment. Defining an accurate and valid marker set for modelling the thorax was not an aim of this thesis, however it is acknowledged as a delimitation of the current research.

Electrode positioning between testing sessions was classed as a delimitation of the sixth chapter in this programme of work. Due the time duration between testing sessions, marking the electrode location was not deemed practical. Although standardisation methods for electrode positions were employed utilising the SENIAM guidelines, a slight difference in location and orientation of electrode positioning may result in different motor units are examined between testing sessions (Burden & Bartlett, 1999). This could have implications for the differences presented between breast support conditions and were considered when interpreting these data.

A delimitation that is a common consideration for studies that examine subjective measures, such as perceptual ratings of pain and exertion, is the effect of prior knowledge and experience. Literature has established the influence of prior knowledge of task duration on measures of RPE (Baden, McLean, Tucker, Noakes, & St Clair Gibson, 2005), suggesting that individuals are able to mentally prepare for a repeated task. Therefore, within this programme of work, the prior knowledge of run duration of the first testing session may influence a participant's ratings in a subsequent session. Randomisation of the breast supports was implemented within this current programme of work in an attempt to overcome the systematic bias associated with learning effects. This precaution reduces the presence of a learning effect for objective measures; however the impact of prior knowledge of the first session may significantly impact upon the subjective measures of the second testing session. For example, since the two treadmill runs were completed at the same speed, the influence of knowledge of the time of the run end point may impact upon the RPE scores provided.

A final delimitation which has implications for the entire programme of work is the restrictive characteristics of the population (age range, breast size, volunteers, training background) examined, which limits generalisation of the findings presented. The findings of the current thesis can only be applied to this specific population. On the other hand, due to unique aspects of breast biomechanics, examining a select population strengthens this programme of work, due to the confounding effects ageing on breast anatomy (Gefen & Dilmoney, 2007) as well as the impact of breast size on between-participant variance (Scurr et al., 2011).

One limitation to this programme of work is the participant's knowledge of the differences in the bras used to determine a low and high level of breast support. The materials used, bra structure, strap configuration and general styles of the bras were obviously different and commented on by the participants. It was clear one was an everyday t-shirt bra and the other designed as a sports bra. The influence of this knowledge on the measured variables is unknown, but it is suggested that this may have affected the perceptual measures reported. This should be a consideration between future research examining differences in both objective and subjective measures between support conditions.

8.2 Recommendations for future work

Since differences were reported in running kinematics and EMG between breast support conditions, it would be of interest to examine what effect these alterations have on physiological measures. Literature suggests that there is a strong relationship between alterations to biomechanical parameters and the metabolic cost and economy of running (Williams & Cavanagh, 1987). Quantifying physiological measures such as these would determine the impact of different breast supports on running bioenergetics. Furthermore, in order to confirm the effect of breast support on running performance, a time-trial running study could be conducted. Whereby, the participant performs a set distance time-trial in different breast support conditions. It is speculated that the significant differences reported in the step characteristics and running kinematics between breast support conditions at a fixed pace, within the current programme of work, indicate a desire to run at a different velocity and employ different running kinematics dependent upon the breast support worn.

Literature available on the standardisation of electrode placement for the pectoralis major muscle on female athletes is sparse. The work conducted within the current programme of work further promotes the pars clavicularis position when examining females. However, the influence of the breast tissue on the EMG signal of the different locations on the pectoralis major is undecided. Future research could investigate the most effective electrode position of the pectoralis major for female participants. This would provide a reliable and standardised method for investigating this muscle in females.

8.3 Conclusion

This programme of work is the first to investigate the effect of breast support on multiplanar breast kinematics, upper and lower body running kinematics and muscle activity over short and prolonged running. The work has demonstrated that a high breast support:

- Reduced multiplanar breast kinematics.
- Reduced exercise-related breast pain.
- Elicited running kinematics previously reported as desirable for economic running.
- Reduced upper body muscle activity.

Furthermore, the following variables significantly increased over the five kilometre run;

- Multiplanar breast kinematics.
- Peak thorax flexion.

The work conducted within this thesis has extended the knowledge of the effect of breast support on running biomechanics. A high breast support is further promoted as an essential piece of sports kit for females running short and prolonged distances, with significant reductions in negative factors associated with independent breast movement during treadmill running.

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10.0 APPENDICES

Appendix A: Upper body kinematics during treadmill running set level (0%) and at a 1% incline gradient.

Introduction

Previous research has demonstrated that a treadmill set at a 1% incline can replicate the energetic cost of overground running for up to five minutes at speeds between 2.92 and 5 $m.s^{-1}$ (Jones and Doust, 1996). Therefore, it is suggested that physiological data collected on a treadmill set at 1% incline can be applied to overground running. Within gait literature, the following differences have been reported between treadmill and overground running; running kinematics, including step/stride characteristics (Alton, Baldey, Caplan, Morrissey, 1998), hip, knee, and ankle joint angles (Alton et al., 1998; Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007; Schache et al., 2001), running kinetics such as peak GRFs and joint moments (Riley et al., 2007).

Due to the biomechanical focus of this programme of research, and the sensitive data collected (breast kinematics), an indoor treadmill protocol was considered as most appropriate for the data collection ruling out overground running. However, the gradient of the treadmill was an important consideration for the current research (include a 1% gradient to enable comparisons to overground running or set the treadmill at 0%), any kinematic differences between a level treadmill (0% incline) and a treadmill set at 1% incline may influence other measures of interest (e.g. breast kinematics and upper body muscle activity). Therefore, the aim of the first pilot was to investigate upper body kinematics between a treadmill set level (0% gradient) and at a 1% incline. This data would determine which treadmill level would be selected for the current programme of research. Based upon the magnitude of difference in the incline angle, it was hypothesised that no significant differences would be identified in the upper body kinematic variables between the two treadmill levels.

Methods

Participants

Nine females (exercising for 30 minutes at least five times a week) participated in the study. Participants had not had any children, had not experienced any surgical procedures to the breast, and were of either a 34B or 34D bra size. Participants had an average (SD) age of 21 years (1 year), body mass 65.4 kg (6.8 kg), and height 1.70 m (0.10 m).

Procedures

Retro-reflective markers (12 mm diameter) were positioned on the suprasternal notch and the left and right side of the body at the following anatomical landmarks; anterioinferior ribs, greater trochanter, anterior superior iliac spine (ASIS), acromion process, lateral epicondyle of the humerus at the radia-humural junction, and the lateral epicondyle of the radius in line with the third metacarpal bone (Figure 24). An additional marker was positioned on the lateral aspect of the left heel on the participant's trainers to track gait cycles.

Figure 24. Anatomical locations of the retro reflective markers on the upper body.

Participants completed two 10 minute treadmill runs on the same day in a high impact sports bra at a set speed of 10 $km.hr^{-1}$, one run completed at a 1% incline and the other on a level treadmill (0%). The two treadmill runs were separated by a 10 minute rest period. Participants wore the same footwear and clothing for both trials. Three-dimensional coordinates of the five markers were tracked by eight calibrated Oqus infrared cameras (Qualisys, Sweden), sampling at 200 Hz. The eight cameras were positioned in an arc around the treadmill, in the centre of the laboratory to maximise the field of view of all cameras. Cameras recorded for 10 seconds at three time intervals; two, five, and ten minutes.

Data processing

The markers were identified and three-dimensional data reconstructed in Qualisys Track Manager (QTM) software. Three-dimensional coordinates were exported to a frequency analysis program in MATLAB (MathWorks, UK). The frequency component of the data was assessed using a Fast Fourier Transformation (FFT) in MATLAB. The FFT showed the amplitude of the data point plotted against the frequency component, enabling the identification of data that should be retained and the noise component that is attenuated (Winter, 1990). A cut-off frequency of 8 Hz was selected for the low pass filter based on this process.

In order to determine gait cycles, instantaneous velocity of the heel marker was derived from the anteroposterior coordinates. Heel strike for each running gait cycle was identified as the velocity of the heel marker reached a peak positive progression (Zeni, Richards, & Higginson, 2008), with a full gait cycle identified as heel strike to heel strike of the ipsilateral heel. The following upper body kinematic variables were quantified in QTM for five gait cycles at each time point; vertical displacement of the thorax (Scurr & Haake, 2010), thorax pitch (Scurr, White, & Hedger, 2009; 2010; Segers, Lenoir, Aerts, & De Clercq, 2007), upper arm flexion and extension (Cavanagh and Williams, 1987; Pontzer, Holloway, Raichlen, & Lieberman, 2009), and transverse plane shoulder segment rotation (Frigo, Carabalona, Mura, & Negrini, 2003).

Vertical displacement of the thorax was quantified with the vertical coordinates of the suprasternal notch. The range of motion (ROM) in vertical displacement of the suprasternal notch was determined by subtracting the minima turning point from the maxima turning point of the sinusoidal oscillations (Haake & Scurr, 2010) and averaged over five gait cycles. Markers positioned on the right and left anterioinferior aspect of the 10^{th} ribs, and the suprasternal notch represent the rigid thorax (Scurr, *et al.*, 2009; 2010). To account for axial rotation of the thorax a mid-rib marker was created between the right and left markers. The projected angle between the line joining the mid-rib and suprasternal notch and the vertical axis of the global coordinate system (GCS) was calculated to gain the degree of thorax pitch. Peak and ROM values of thorax pitch were calculated and averaged over five gait cycles. Markers positioned on the acromion and lateral epicondyle of the elbow enabled the calculation of flexion and extension of the upper arm at the shoulder. Shoulder flexion was defined as line segment joining the acromion and lateral epicondyle of the elbow passed the vertical axis (90°) of the GCS, towards the anterior aspect of the body within the sagittal plane. Shoulder extension was defined when the line segment of the upper arm passed the vertical axis (90°) of the GCS towards the posterior aspect of the body within the sagittal plane. Range of motion values of upper arm extension were calculated and averaged over five gait cycles. The line segment created between the right and left acromion markers enabled the transverse plane rotation of this segment to be quantified. The projected angle between the line segment and the mediolateral axis of the GCS was quantified with the ROM values averaged over five gait cycles.

Statistical analysis

All data were checked for normality using the Kolmogorov-Smirnov and Shapiro-Wilk tests. Data met the normality assumptions ($p > .05$) and therefore parametric tests were employed. A two-way repeated measures ANOVA was employed to determine any differences in upper body kinematic variables between the 1% gradient and 0% level treadmill runs over the three time points (2, 5, and 10 minutes). The alpha level was set at $p < .05$. *Post hoc* pairwise comparisons with Bonferroni adjustments were performed following the two-way repeated measures ANOVAs. Effect size and observed power were calculated to characterise the strength of all results, where a small effect ≤ .10, medium effect ≤ .30, large effect ≤ .50, and a high power ≥ .80 (Field, 2009).

Results

On average the ROM in vertical displacement of the suprasternal notch was 13 cm (Figure 25), and did not differ between level treadmill running and treadmill running with a 1% incline at any time point during the ten minute trial ($F_{(1)} = .440$, $p = .526$, $\eta^2 = .052$, $1\text{-}\beta = .090$).

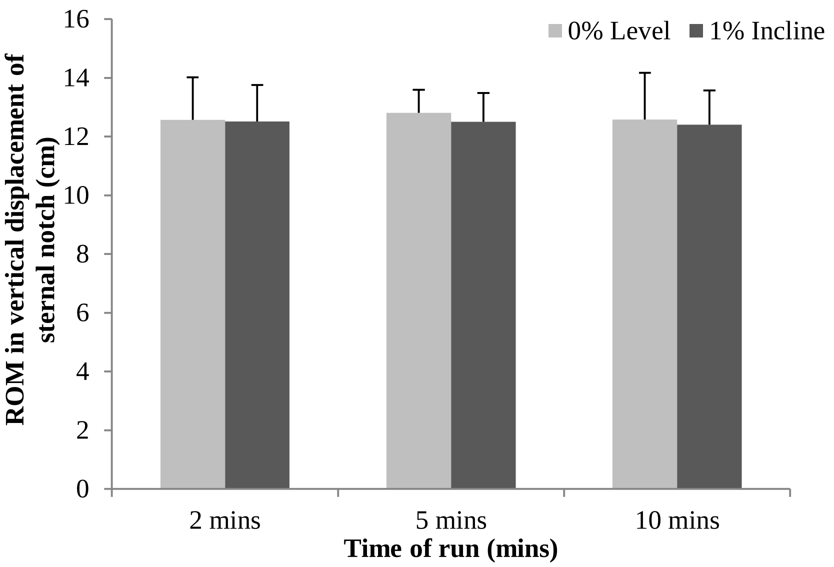


Figure 25. ROM in vertical displacement of the suprasternal notch during the three time intervals (2, 5 and 10 mins) of the two 10 minute treadmill runs (0% and 1% incline).

Range of motion in thorax pitch did not differ between treadmill running set at 0% and 1% incline over ten minutes of running ($F_{(1)}$ = 1.467, p = .260, η^2 = .155, 1-β = .188). On average the ROM in thorax pitch was 12° during level and 1% incline treadmill running (Figure 26).

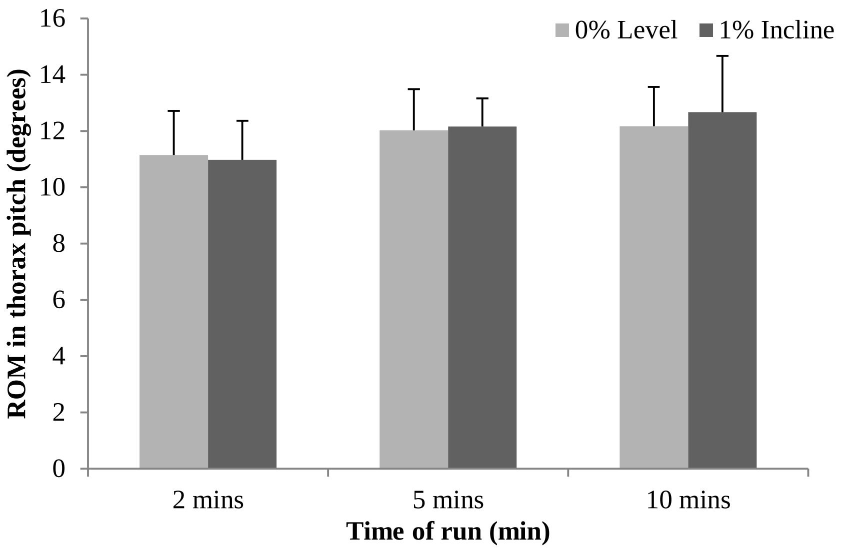


Figure 26. ROM in thorax pitch during the three time intervals (2, 5 and 10 mins) of the two 10 minute treadmill runs (0% and 1% incline).

On average forward flexion of the thorax was 5° during both level and 1% incline treadmill running (Figure 27). No differences were reported between the two treadmill runs (0% and 1%) during the ten minute run ($F_{(1)}$ = 2.633, p = .143, η^2 = .248, 1-β = .299).

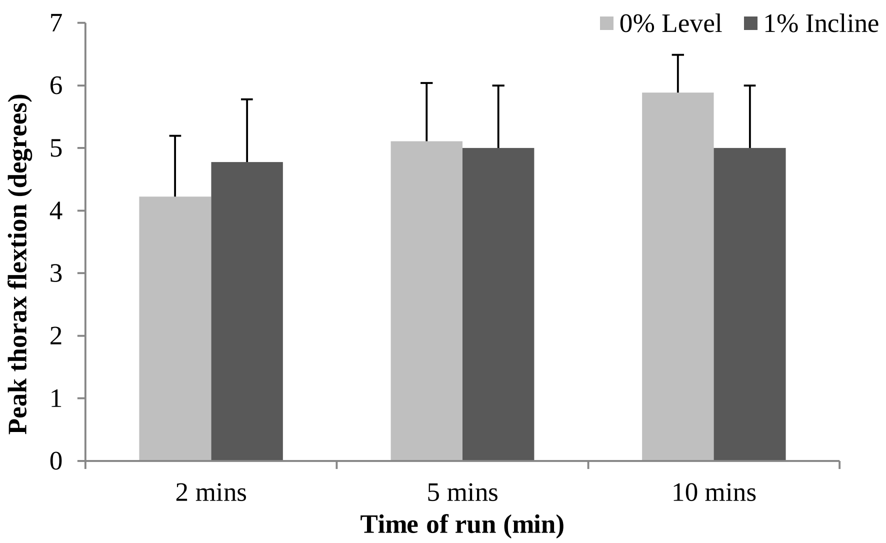


Figure 27. Peak thorax flexion during the three time intervals (2, 5 and 10 mins) of the two 10 minute treadmill runs (0% and 1% incline).

During level treadmill running (0%) upper arm extension was 45° on average, and 44° when running with a 1% incline (Figure 28), however, no significant differences were reported ($F_{(2.976)} = 2.466$, $p = .087$, $\eta^2 = .236$, $1\text{-}\beta = .537$).

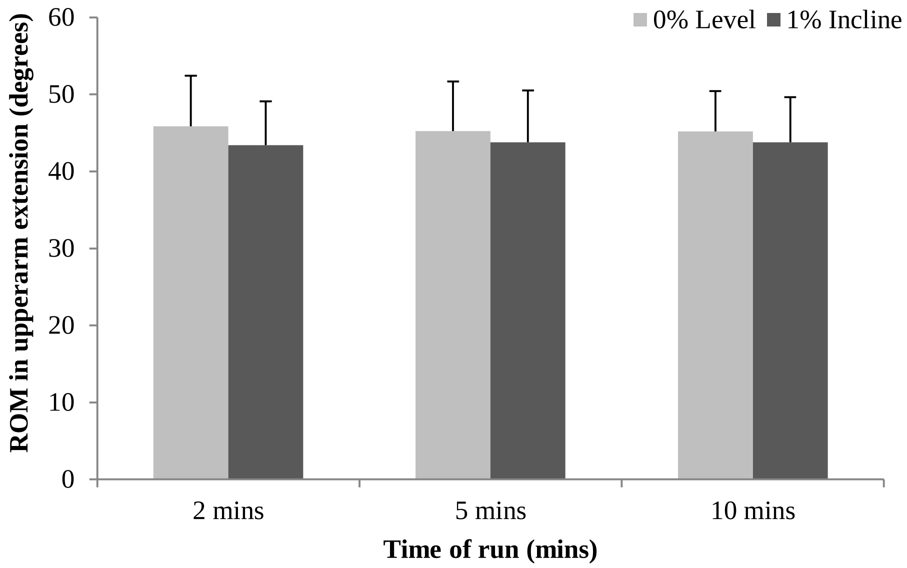


Figure 28. ROM in extension of the upper arm at the shoulder during the three time intervals (2, 5 and 10 mins) of the two 10 minute treadmill runs (0% and 1% incline).

On average the ROM in shoulder segment rotation was 36° and 39° for level and 1% incline treadmill runs, respectively (Figure 29). However, no significant differences were reported during the ten minute run trials ($F_{(1.600)} = 6.614$, $p = .064$, $\eta^2 = .353$, $1\text{-}\beta = .478$).

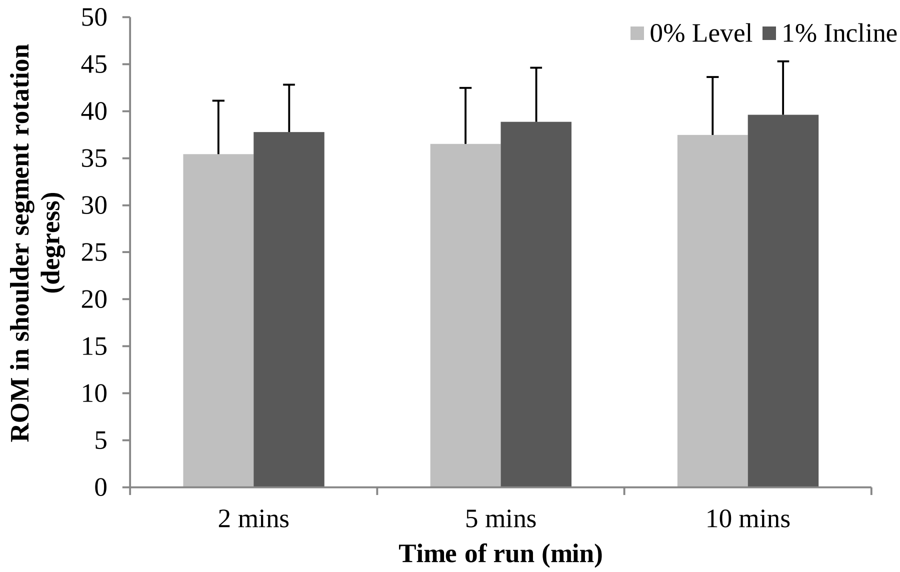


Figure 29. ROM in shoulder segment rotation during the three time intervals (2, 5 and 10 mins) of the two 10 minute treadmill runs (0% and 1% incline).

Discussion

The primary aim of this pilot study was determine if upper body kinematics differed between a treadmill set at 0% or at a 1% incline during a ten minute constant speed run trial. A secondary aim was to determine which treadmill orientation (0% or 1%) should be implemented for the remaining studies in the current programme of research. Vertical displacement of the suprasternal notch, ROM and peak thorax pitch, ROM of extension of the upper arm, and shoulder segment rotation remained the same across the two treadmill conditions (0% and 1% incline).

With no differences reported in the upper body kinematics examined, it is suggested that either treadmill orientation (0% or 1% incline) can be employed for this programme of research and that an incline of only 1% does not elicit significant changes to upper body kinematics. It is assumed that few females exercising on a treadmill would set the incline to 1%. Therefore, with the testing restrained to the laboratory due to the sensitive nature of breast kinematic data, and in order to apply the results of the current programme of research to females exercising on a treadmill, a level treadmill (0% level) orientation has been selected.

Although a 1% treadmill incline has been found to represent the energetic cost of overground running (Jones & Doust, 1990), significant differences have been reported in step characteristics and lower body kinematics (Alton, Baldey, Caplan, Morrissey, 1998), and GRFs (Riley et al., 2007) between overground and treadmill running based upon other factors such as environmental conditions and different surface-foot interactions (e.g. belt thickness and material used compared to road or trail running). Therefore, it is suggested that biomechanical comparisons in the results collected in this programme of research can be made between 0% and 1% treadmill orientations, but not between treadmill and overground running.

Appendix B: Five kilometre subjective questionnaire

University of Portsmouth

Questionnaire

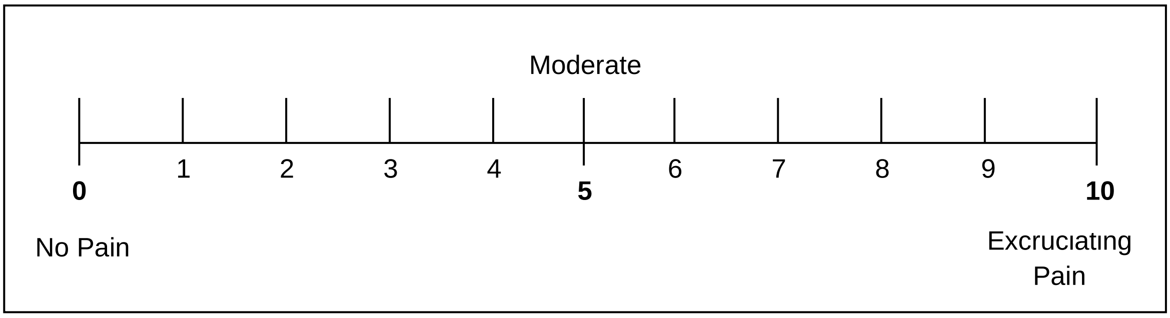
Participant Number: ____________________ Support Condition: __________________

After each breast support condition please rate:

1) Was there any breast pain throughout the duration of the run? (If yes, please specify and answer questions 2 and 3).

____

2) The **intensity** of breast pain felt during the run? *(Please circle)*

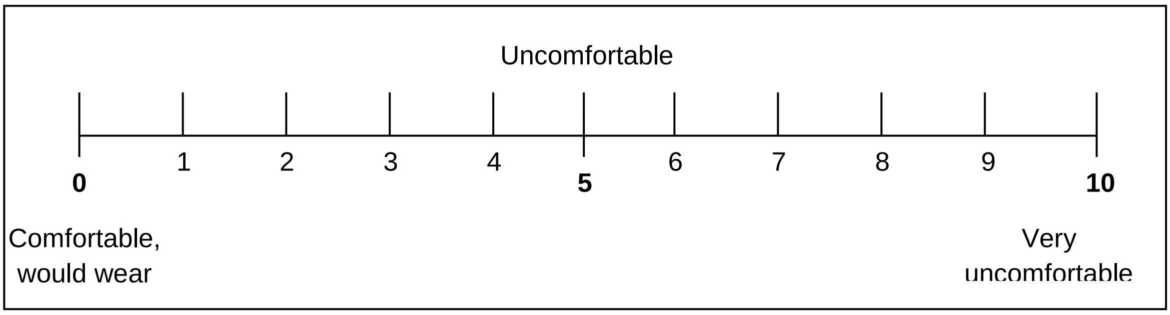


3) Did the breast pain change at any point of the run? (E.g. more painful/less painful once into the run or at specific times of the run?).

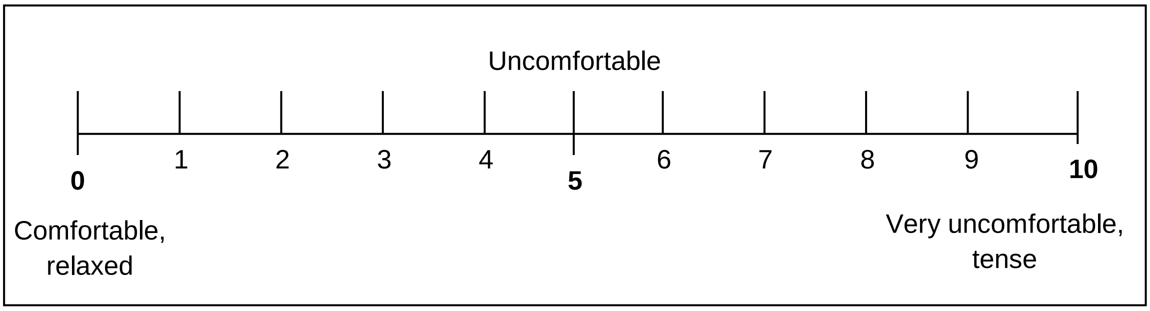
____

____

4) How comfortable you felt the **bra** was during the run if applicable? *(Please circle)*



5) How comfortable did you feel during this run?



Please provide any details if applicable:

6) Did you notice any differences in your running style during this 5km run?

Appendix C: RPE scale (Borg, 1982).

RATING OF PERCIEVED EXERTION (RPE) SCALE

6	NO EXERTION AT ALL
7	
8	EXTREMELY LIGHT
9	
10	
11	LIGHT
12	
13	SOMEWHAT HEAVY
14	
15	HARD (HEAVY)
16	
17	VERY HARD
18	
19	EXTREMELY HARD
20	MAXIMAL EXERTION