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Breast Support is Associated with Altered Neuromuscular System Stability During Treadmill Running

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BREAST SUPPORT IS ASSOCIATED WITH ALTERED NEUROMUSCULAR SYSTEM STABILITY DURING TREADMILL RUNNING

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ABSTRACT

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A barrier to exercise for female runners is exercise-induced breast pain or discomfort. Increasing levels of breast support has been shown to reduce exercise-induced breast motion, reduce breast pain and discomfort, and may alter running kinematics in female runners, but its effects on the organization of the neuromuscular system is unknown. The purpose of this study was to evaluate the effect of breast support on the organization of the neuromuscular system and quantify LDS (LDS) at the ankle, knee, and hip during running tasks. 13 female recreational athletes performed three minutes of running in three different breast support conditions (CON, LOW, HIGH), at three different running speeds (SLOW, PREF, FAST). The Largest Lyapunov's Exponent was used to analyze the fluctuations in lower extremity joint angles and quantify LDS. A 3x3 repeated measure analysis of covariance was used to determine the effect of support and speed, while covarying for self-reported breast size. No significant effect of support LDS was observed at the ankle, knee; however, an effect of support was observed at the hip, with LOW have greater LDS than HIGH and CON. An effect of speed was observed at the ankle, with both CON and PREF having less LDS than FAST. These findings suggest that both breast support and speed alter LDS during running, and that the neuromuscular system adopts a proximal-to-distal strategy for perturbation attenuation.

PREFACE

The findings from this thesis will be submitted for publication to *Chaos: An Interdisciplinary Journal of Nonlinear Science* and the formatted manuscript for this journal is presented in Chapter III. References are specifically formatted for this journal.

Table of Contents

List of Tables

List of Figures

Key to Abbreviations

LyE	Largest Lyapunov's exponent
SLOW	15% less than preferred running speed
PREF	Preferred running speed
FAST	15% more than preferred running speed
CON	Control or no support condition
LOW	Low breast support condition
HIGH	High breast support condition
LDS	Local Dynamic Stability
DST	Dynamical Systems Theory

CHAPTER I

INTRODUCTION

1.1 Statement of the Problem

Female participation in sports and running has greatly increased since 1972. The implementation of title IX sparked an dramatic increase in female participation in high school and collegiate athletics, leading to a ten-fold increase in high school participation over the following four decades [1]. Collegiate running-related sports, such as cross-country and track, have seen a four-fold increase in participation over the past three decades [2]. With this increase in the of female athletes comes a need to better understand what factors may contribute to running performance.

As female participation in athletics has increased over the previous four decades, so has the focus of female-related athletic research. An area that has begun to see an increase in research focus is the effect of sports bras or external support on breast motion. The female breast has been described as a “wobbling mass on a rigid torso” and lacks strong intrinsic support [3]. This lack of support and the inertial properties of the breast tissue make the breast susceptible to excessive exercise-induced breast motion. Excessive breast motion, particularly vertical breast motion, during athletic tasks has been linked to breast pain and discomfort, embarrassment, and unwillingness to exercise [4-6]. Increasing breast support has been shown to limit multiplanar breast motion, decrease exercise-induced breast pain or discomfort, decrease embarrassment, and increase willingness to exercise [4-7]. As with the changes in breast kinematics, breast support has also been shown to alter running kinematics and temporospatial variables associated with running.

Increasing breast support has been shown to impact running kinematics and running temporospatial variables. Increasing support has been shown increase pelvic and trunk rotation and range of motion, increase vertical trunk oscillations, and increase stride length [6, 7]. Similarly, increasing breast support has been shown to alter running kinetics, with no support having greater mediolateral ground reaction forces compared to support conditions [8]. These findings show that breast motion alters movement patterns; however, while research has observed running kinetic and kinematic changes with the addition of breast support, no previous research has investigated the effect of breast motion and support on the adaptability of the neuromuscular system.

Local dynamic stability (LDS) represents the neuromuscular system's ability to respond and adapt to small, continuous perturbations and produce stable movement patterns [9, 10]. LDS has been used to evaluate neuromuscular stability in gait and running research in healthy and diseased adults, trained and novice runners, and runners with lower-limb amputations [11-14]. Decreases in LDS indicate worse control over the motor pattern or behavioral stability [15]. Increased neuromuscular demand to attenuate a perturbation, decreases in LDS, and may lead to decreases in running performance. Previous literature has suggested that reductions neuromuscular stability or efficiency would lead to decreases in running performance and efficiency, however this has not directly been test [15, 16].

LDS describes the neuromuscular system's robustness to small perturbations that occur during locomotion and is quantified through the calculation of the Largest Lyapunov's Exponent (Lye). LyE quantifies the ability of the neuromuscular system to react to small perturbations by calculating the exponential rate of divergence for neighboring trajectories in a state-space [10, 14].These divergences, or fluctuations, in kinematic data, such as joint angles, represent mechanical disturbances or motor control errors [17]. This measurement is inversely related to LDS, with high LyE values being indicative of less neuromuscular stability. LyE has also been shown to be sensitive to walking and running velocities, with LyE values increasing as speed increases [12, 18]. This cause-and-effect indicates that LDS decreases as running speed increases.

The first purpose of this study is to investigate the effect of breast support on lower extremity LDS. LyE will be used to assess the LDS of the hip, knee, and ankle by analyzing the fluctuations in sagittal plane joint angles during running bouts at different speeds and in different levels breast support. In the presences of LDS differences between conditions, temporospatial gait variables will be analyzed to assess the manifestation of altered motor patterns. For breast support, it is hypothesized that LDS at the ankle, knee, and hip will increase with increasing breast support. Conversely, it is hypothesized that LDS will decrease as running speed increases. The second aim of this study is to examine the temporospatial manifestation of altered neuromuscular control during running caused by changes in breast support. It is hypothesized that temporospatial variables will be altered as breast support increases, as seen by increased stride length, increased cadence, and decreased ground contact time with increasing breast support. Similarly, it is hypothesized these variables will change with increases in running speed, as seen by increased stride length, increased cadence, and decreased ground contact time and stride time interval with increasing running speed.

CHAPTER II

2.1 Literature Review

Female participation in sport has increased substantially since the implementation of Title IX in 1972. With this change in policy, female participation in high school athletics increased from less than 300,000 in 1972 to an estimated three million as of 2010 [1]. A survey in 2019 found that over 1.2 million female athletes participated in high school volleyball, basketball, and soccer, collectively [19]. At the collegiate level for these same three sports, participation increased from 25,000 in 1981-1982 to over 62,000 in 2018-2019 [2]. As of the 2012 London Olympics, 44.3% of the 10,903 athletes were female [20].

Overall increases in female participation in sport has resulted in a subsequent increase in female runners. Collegiately, female participation in cross-country, indoor track, and outdoor track has increased nearly four-fold from just under 19,000 in 1981-1982 to just under 74,000 as of 2018-2019 [2]. Likewise, participation in running as a mode of exercise or competition has increased from under 20% of total runners in 1986 to accounting for over 50% of all runners in 2018 [21].

2.1.1 Breast motion and Running

With the increase in female participation in running, it is important to understand different factors that may have an impact on running performance. Over the past several decades, more and more research has been done assessing the impact of breast motion, and breast support in female runners. The development of breast tissue as secondary sexual trait is an important factor which may impact running performance in female athletes. Sexual dimorphic breast development is affected by the circulating hormones, such as estrogen [22]. These dimorphic changes in breast tissue can be separated into five stages, with the final stage of maturation, complete maturation, being attained by average age of 15 years old [23]. The anatomic structure of maturely developed female breast tissue is noteworthy when looking at the impact breast tissue and breast motion may have on female runners. Typically covering the area from the second rib to the sixth rib and overlaying the pectoralis major muscle, the breast structure is composed of skin and subcutaneous tissue, ducts and lobules, stroma, and adipose tissue with an accompanying network of ligaments, blood vessels, nerves, and lymphatic vessels [24]. Breast size, shape, stiffness, and density vary between individuals, with differences in size being typically attributed to differing amounts of adipose tissue. Lacking muscular tissue, the breast is supported by fibrous connective tissues often referred to as Cooper's ligaments. These thin fibrous bands are located within the superficial fascia and are attached to the deeper fascia that covers the pectoralis muscles and provides limited support to the breast [25]. It is suggested that the skin covering the breast, as well as these ligaments, provide the greatest, albeit weak, amount of structural support [26].

This lack of strong intrinsic support leads to distinct breast motion patterns during running. Scurr et. al observed that breast tissue moves in a figure-eight pattern during running, following the movements of the torso and its inertial properties are restricted only by the weak, intrinsic support of the skin and Cooper's ligaments [27]. These researchers found that of the total displacement experienced during running, vertical displacement accounted for an average of 56% of overall breast displacement, while mediolateral and anteroposterior motion contributed to 41% of total displacement. To control or mitigate this multiplanar breast motion, external breast support, such as a sports bra, is worn. Increased breast support has been shown to significantly reduce breast excursion during athletic activities, including running [5, 6, 8, 28]. Scurr et. al also observed that peak vertical breast displacement, measured using the vertical displacement of the nipple, occurred later in the gait cycle than peak vertical trunk displacement, measured using the vertical displacement of the sternal notch. This time difference between peak breast displacement and peak trunk displacement was termed as *time lag* [27]. Similarly, this *time lag* has been shown to reduce with increasing levels breast support [5, 7]. During running, increasing breast support has also been shown to increase pelvis and trunk rotation and range of motion, increase stride length, increase peak swing phase knee flexion, increase thigh range-of-motion, and increase vertical trunk oscillations [6, 7].

As with running kinematics, breast kinematics have been shown to greatly change with increasing levels of breast support. Risius et. al found the vertical breast displacement was reduced by an average of close to 4cm between the no support and high support conditions [7]. Similarly, with reduced motion, vertical breast velocity significantly decreased by an average of nearly 0.6 m/s. Because vertical breast motion and vertical breast velocity are closely correlated to breast pain and discomfort, subjective perception of breast pain dropped from an average score of 6 to just above 0 on a scale from 1-10. These findings are corroborated by the findings of Scurr et. al, who observed significant reductions in multiplanar breast motion with increasing breast support, and a drop from an average score of 7 to 1.4 in perceived breast pain between no support and sports bra support conditions on a scale from 1-10 [5]. While each of these studies found significant reduction in vertical breast displacement, it should be noted that similar reductions in breast motion were observed in the mediolateral and anteroposterior directions, as well, as breast support increased.

While previous research has shown possible effects of breast support on running kinematics, temporospatial variables, and breast kinematics, it is not yet understood how breast motion may affect the neuromuscular system's ability produce stable movement patterns. A factor caused by exercise-induced breast motion that may place a constraint on the neuromuscular system is breast pain or discomfort. Breast pain/discomfort, as well as unwillingness to exercise, are common consequences of excessive, exercise-induced, breast motion [4]. Both Risius et. al and White et. al found that as breast support increased, willingness to exercise and breast comfort increased significantly, with increased breast motion being correlated with increased perception of breast pain [5-7]. It has been suggested that pain and discomfort may affect the organization of neuromuscular system. A study investigating the impact of postural perturbations on center of pressure stability in individuals with chronic lower back pain, observed that in the presence of pain or discomfort, individuals altered their motor pattern strategies [29]. Similarly, Dubois et. al found that induced lower back pain altered movement patterns and neuromuscular control of the trunk in healthy adults and those with chronic lower back pain [30]. A third study investing the effects of chemically induced back pain on the LDS of the trunk found that trunk LDS was significantly less during flexion-extension task than when no pain was present [31]. The findings from these studies suggest that in the presence of breast pain or discomfort due to exercise-induced breast motion, female runners may also alter their motor pattern strategies, which may have further implications in running performance.

2.1.2 Variability and Fluctuation Analysis

There are several ways of analyzing the motor pattern output variability, or organization, of the neuromuscular system. The term *variability* traditionally refers to a magnitude of deviation around a central point or mean (e.g. standard deviation, variance, range, or coefficient of variation). Large values of these measures are viewed as motor pattern error or mistake. For example, consider the game of darts, where accuracy of throws is key. When aiming for a bullseye, or the center of the dart board, large amounts of variability, or deviation from the center point for each throw would be indicative of a less skilled thrower. Conversely, small variability from the center would be indicative of a skilled thrower. In gait and running research, traditional measures of variability are used to describe a sequence of strides over time that are then filtered, averaged, and time-normalized, resulting in "typical gait". While this method is useful for describing what is occurring during the gait cycle compared to the mean, it eliminates the ability to observe the natural fluctuations of locomotion that occur stride-to-stride. These fluctuations are typically viewed as random noise in a signal to be filtered out; however, it has been suggested that the observed noise found in biological systems is not random, but instead has a deterministic origin, and that information regarding a system's behavior can be gained by analyzing this noise [32, 33].

A prominent theory that is used to explain the behavior of biological systems is the Dynamical Systems Theory. The Dynamical Systems Theory suggests that biological systems are self-organizing and that motor behavior is developed through the collective dynamics of the contributing subsystems, as well as environmental and biomechanical constraints [34]. The function of the neuromuscular system in human locomotion is to find the most stable solution for executing a movement pattern. This preferred movement pattern is called an "attractor", and this attractor represents the stable state or equilibrium state of the system. When internal or external constraints perturb the system, the system must either 1) attenuate the perturbation and return to the attractor state, 2) switch to a new, more stable attractor state, or 3) fail to deal with the perturbation resulting in failure, such as falling [10, 32, 35]. These adaptations are the seen as the fluctuations in a signal and are the reaction to an experienced perturbation [36]. The fluctuations of a signal, associated with this perturbation-response, are closely related with a rich behavioral state, have been linked to the health of biological systems, and can be measured using nonlinear mathematics [35].

Nonlinear mathematics have been developed to analyze the fluctuations of signal found in dynamical systems. These fluctuations can be characterized as the normal changes that occur in motor performance across multiple iterations of a task and are inherent in all biological beings [37, 38]. One mathematical concept that is frequently used in nonlinear gait and running research is the calculation of the largest Lyapunov's exponent (LyE). LyE is used to measure the neuromuscular system's ability to continuously respond to perturbations [9, 14]. These perturbations are revealed through diverging trajectories in the state space, with LyE being the exponential rate of divergence for neighboring trajectories [10, 14]. An example of a state space is present in **Figure I**. LyE has been used to evaluate gait stability and risk of falls in healthy and diseased populations as well as the impact of shoe type, running speed, lower-limb amputations, and transitioning from shod to barefoot running in adult runners [11, 12, 15, 39-41].

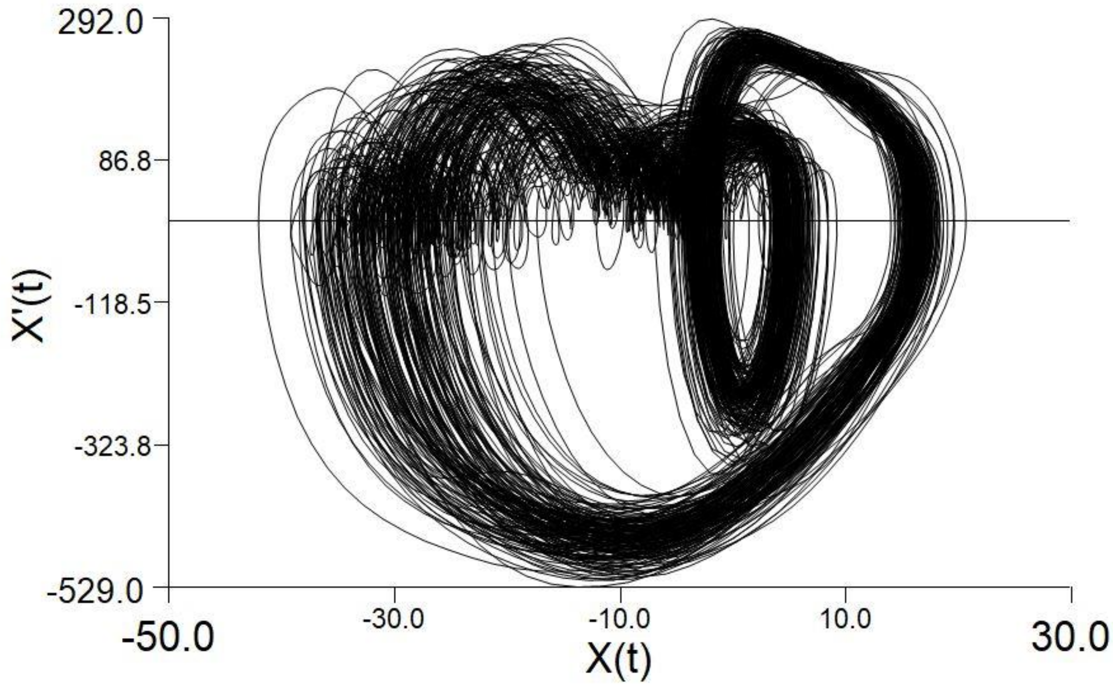

Figure I. Two-dimensional state space created by plotting the sagittal position, *X(t)*, and speed, *X'(t)*, time series for the ankle.

The LyE algorithm most frequently used in gait research is that presented by Rosenstein et. al [42]. This calculation first begins with reconstruction of attractor dynamics, X, for the time series, x:

$$X = [X_i \; X_2 \; \ldots \; X_M]^T \qquad \text{(Eq. 1)}$$

where X_i is the state of the system at time i. X_i is:

$$X_i = \left[x_i \; x_{i+j} \; \ldots \; x_{i+(m-1)J}\right] \qquad \text{(Eq. 2)}$$

where J is the reconstruction delay and m is the embedding dimension. X is an $M \times m$ matrix, where M is calculated as:

$$M = N - (m-1)J \qquad \text{(Eq. 3)}$$

and N is the number of data points in the time series. After attractor dynamics have been constructed, the nearest neighbor of each point on the trajectory is located. The nearest neighbor, $X_{\hat{j}}$, is the point the minimizes the distance to X_j, the reference point, and is expressed as:

$$d_j(0) = \min_{X_{\hat{j}}} \left\| X_j - X_{\hat{j}} \right\| \quad \text{(Eq. 4)}$$

where $d_j(0)$ is the initial distance from the j^{th} point to its nearest neighbor, and $\|..\|$ is the Euclidean norm. The j^{th} pair of nearest neighbors diverge approximately at a rate given by the largest LyE:

$$d_j(i) \approx C_j e^{\lambda_1 (i \cdot \Delta t)} \quad \text{(Eq. 5)}$$

where C_j is the initial separation, λ_1 is the largest Lyapunov's exponent, and $i \cdot \Delta t$ is the discrete-time step. Taking the logarithm of both sides gives:

$$ln d_j(i) = ln C_j + \lambda_1 (i \cdot \Delta t) \quad \text{(Eq. 6)}$$

The largest Lyapunov's exponent can then be calculated using a least-squares fit regression of the line:

$$y(i) = \frac{1}{\Delta t} \langle ln d_j(i) \rangle \quad \text{(Eq. 7)}$$

where $y(i)$ is the average diverging trajectory, $ln d_j$ represents the natural logarithm of the diverging trajectory, and $\langle .. \rangle$ denotes the average over all the values of j [42]. The reconstruction delay, J, used for creating the attractor dynamics can be determined using an average mutual information algorithm, while a false nearest neighbors algorithm can be used to determine the embedding dimension, m [43, 44].

It is important to note that LyE calculation for gait stability is sensitive to the number of data points used for the entire time series and for each stride [45]. It is suggested that the same number of data points for the entire time series, as well as for each stride, should be used for each condition and subject [45, 46]. Normalizing both the total length of the time series, as well as each stride, allows for comparison between conditions of differing speeds where stride duration may be different [14, 18]. While this does remove stride-to-stride temporal variations, this method of normalization has been suggested to provide the greatest sensitivity when using Rosenstein's algorithm during both walking and running conditions [45]**.**

2.1.3 Local Dynamic Stability

LDS represents the ability of the neuromuscular system to recover from small perturbations and can be quantified through the calculation of LyE [10, 18]. Bruijn et. al gives three, partially-independent, qualifications of stable gait: 1) the system must be able to recover from small stride-to-stride perturbations, 2) the system must be able to adapt to large, behavior-changing perturbations, and 3) the largest perturbation the system can recover from must be larger than all other perturbation experienced [10]. LDS has been used to evaluate neuromuscular robustness to small kinematic disturbances that are experienced stride-to-stride during walking and running. Changes in LDS are indicative of movement pattern efficiency, with decreased LDS, or decreased behavioral stability, indicating worse movement efficiency. England et. al analyzed the effect of speed on gait stability in adults to evaluate the suggested notion that elderly adults, and those with disease or neuropathy, adopt a slower gait speed to improve stability [18]. Analyzing joint angles, this study found that LDS was inversely related to gait speed, finding that as speed increased dynamic stability at all three joints decreased, resulting in less stable gait. Look et. al replicated these finding in runners, observing that LDS at the ankle, knee, and hip decreased as running speed increased [12].

While speed has been shown to impact LDS, there does not exist much research investigating the effect of an external wearable on running stability. A study by Ezikos et. al, investigated the differences in LDS of the trunk between shod and barefoot running, arguing that a transition to barefoot running would increase the demand on the neuromuscular system resulting in greater motor control errors quantified using LyE [15]. Investigating this transitional effect in twenty young adults, this study found that a transition from shod to barefoot running resulted in significantly less LDS of the trunk. This reduced stability was interpreted as barefoot running placing a greater strain on the neuromuscular system, decreasing its robustness to perturbations, and eliciting more motor control errors than shod running. A similar study attempted to assess the impact of shoe type on lower extremity LDS. Comparing novice and trained runners using shoes with varying midsole thickness and hardness, Frank et. al found that changing shoe type had no effect on the LDS at the ankle and knee between shoe types, but an interaction between shoe type and running group was observed at the hip [11]. These findings suggest that the small variations between shoe types were not large enough perturbations on the neuromuscular system, rather that running experience has a greater impact on LDS. This then leads to the question of "are changes in breast motion, due to breast support and running speed, large enough to perturb the system and elicit changes to lower extremity LDS?" and if so, "how will these changes be visible?".

2.1.4 Perturbation-response to breast motion

Perturbations are revealed through diverging trajectories in a state space and then quantified by LyE. Under the influence of external or internal perturbations, the neuromuscular system must either adapt or fail [10]. Dynamical system's theory would suggest that the mechanism of adaptation would be perturbation-dependent, meaning different types of perturbations elicit different neuromuscular responses. The purpose of this study is to assess the effect of breast support on lower extremity dynamic stability, or the ability of the neuromuscular system to respond to changes in breast motion (breast motion being the perturbation) with changes in levels of breast support. To explore the lower extremity joint dynamics, sagittal joint angles of the ankle, knee, and hip have been previously used to assess the LDS of each joint [11, 12]. Joint angles provide a kinematic explanation to human movement, and the natural variances in kinematic data can be attributed to mechanical disturbances or motor control errors [18].

A comprehensive systematic review of the use of LyE in gait research observed that the knee tended to have larger LyE values than the hip and ankle when Rosenstein's algorithm was used to analyze lower extremity joint angles [14]. This trend would suggest that the knee exhibits a greater reaction to perturbations than at the hip and ankle. The knee functions as the strut of the lower body, and plays key role in load attenuation and dampening during running [17, 47]. As the knee assumes a more flexed position, vertical excursion of the of the pelvis increases. During running, this dampening role may be a mechanism by which female runners attempt to limit the excessive vertical breast motion and vertical breast velocity that is correlated with breast pain and discomfort [5]. These findings suggest that neuromuscular adaptation to breast motion may be seen in the kinematic fluctuations of the knee joint.

To better assess the mechanical manifestation of lower extremity dynamic instability, temporospatial variables such as stride length, cadence, and ground contact time can be analyzed. Changes in lower extremity dynamics as a response to changes in joint LDS should be manifested in these temporospatial variables, as they are influenced by lower extremity kinematics. Ground contact time, for example, has been shown to increase with increased knee flexion during the stance phase, and is also inversely related to cadence [48]. Should the perturbations caused by breast motion be large enough, motor behavior should change [10].

2.3 Research Question and Hypotheses

Purpose: The purpose of this study is to evaluate the effects of breast support on lower extremity LDS in female runners.

Aim #1 Evaluate the effects of breast support and running speed on the LDS of the ankle, knee, and hip.

Hypothesis #1 Increases in breast support will be result in greater LDS at the ankle, knee, and hip.

Hypothesis #2: Faster running speed will result in decreased LDS at the ankle, knee, and hip compared to slower running velocities.

Aim #2 Evaluate the effects of breast support and running speed on temporospatial variables in running.

Hypothesis #1: Temporospatial variables will be altered with changes in breast support, seen by increased stride length, increased cadence, and decreased ground contact time as breast support increases.

Hypothesis #2: Temporospatial variables will be altered with changes in running speed, seen by increased in stride length, increased cadence, and decreased ground contact time as running speed increases.

CHAPTER III

Breast Support Is Associated With Altered Neuromuscular System Stability During Treadmill Running.

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Introduction

Female participation in sports and running has greatly increased since 1972. The implementation of title IX sparked an dramatic increase in female participation in high school and collegiate athletics, leading to a ten-fold increase in high school participation over the following four decades[1]. Collegiate running-related sports, such as cross-country and track, have seen a four-fold increase in participation over the past three decades [2]. With this increase in female athlete participation comes a need to focus sport science research on female-specific questions.

A major challenge for female athletes and runners is exercised-induced breast pain and discomfort. Described as a "wobbling mass on a rigid torso", the female breast lack strong intrinsic support and its motion during exercise is only dictated by its inertial properties and connective tissue, such as skin [3]. Excessive breast motion experienced during exercise, despite external support from sports bras, has been linked to breast pain and discomfort, and contributes to unwillingness to exercise in female runners[4, 6, 7]. To mitigate this motion and pain, more supportive sports bras are often worn. Greater breast support has been shown to reduce multiplanar breast motion, reduce breast pain and discomfort, and increase willingness to exercise as breast support level increases[4, 6, 27]. Along with changes to breast kinematics, increasing levels of breast support has been shown to increase peak trunk and pelvis rotation, increase trunk and pelvis range-of-motion, increase stride length, and increase vertical trunk oscillations [6, 7]. These differences in whole-body kinematics between external support levels may alter the overall neuromuscular strategy during running.

The human neuromuscular system adopts different strategies in the presence of pain or discomfort[29, 30]. It is suggested that in the presence of breast pain caused by exercise-induced breast motion, female runners will adopt different neuromuscular strategies in order to attenuate the perturbation and run pain-free. Because of its function as the strut of the lower body, it is suggested that neuromuscular adaptations may be seen at the knee as the system attempts to find a way to attenuate breast motion perturbations and limit breast discomfort during lower support conditions. Previous research has suggested that changes in neuromuscular strategy may have implications for running economy and performance[15, 16].

To understand the organization of the neuromuscular system and its response to experienced perturbations, nonlinear mathematics have been developed. The Largest Lyapunov's Exponent (LyE) is a mathematical tool that calculates the exponential rate of divergence of neighboring trajectories in a state space and assess the neuromuscular system's ability to continuously respond to perturbations during locomotion [10, 49, 50]. LyE can be used to analyze fluctuations kinematic data and quantify the local dynamic stability (LDS) of a system which represents the neuromuscular system's ability to respond and adapt to small, continuous perturbations and produce stable movement patterns[9, 10]. Changes in LDS can be seen in the preferred movement, or attractor dynamics, of lower extremity joint angles, with reduced LDS challenging the system to find a new, more stable movement pattern[32]. It is suggested that excessive breast motion, with less supportive breast support, may challenge to the neuromuscular system, alter neuromuscular strategy, and result in changed motor behavior.

The purpose of this study was to investigate the effects of increasing breast support on lower extremity LDS during running. To exacerbate breast motion, different running speeds were used, as breast motion increases with running speed, thus increasing the imposed perturbation. First, it was hypothesized that as running speed increased, the divergence in lower extremity joint LDS between breast support conditions would increase, with LDS being greatest with more breast support. Second, it was hypothesized that as both breast support and running speed increased, stride length and cadence would increase, while ground contact time and stride time would decrease.

Materials and Methods

Participants

15 female runners were originally recruited to participate in this study; however, data from two participants were excluded from analysis. Of these two participants, one failed to complete all three conditions, while the second individual switched between rearfoot and forefoot running throughout each trial determined by foot angle at initial contact. This inconsistency will affect LyE calculation and artificially decrease LDS. Therefore, 13 female runners (24 ± 3 years) completed the study and were included for analysis. A *post hoc* power analysis was conducted to determine the observed power using the support condition effect size

(*f*) for hip LyE of 0.23. Observed power for this study was then determined to be 0. 16, or low statistical power. Inclusion criteria to participate included (1) being 18-35 years of age, (2) having a self-reported breast size of B-cup to D-Cup, (3) no medical conditions as screened using a Physical Activity Readiness Questionnaire (PAR-Q), (4) were a recreational athlete as defined by participating in moderate to vigorous exercise for 30 or more minutes, three or more times per week (5) and were currently free from musculoskeletal or neuromuscular injury for the previous 6 months. Foot strike angle (FSA) at initial contact was used to determine foot strike pattern consistency throughout the running trials. Anthropometric measurements were taken for each participant including height, mass, self-reported breast size, bust size, and ribcage measurement. Bust size was measured as the circumference around the largest portion of the breast and the torso. Rib cage circumference was measured as the circumference around the rib cage just below the bust. Participants signed a voluntary consent form approved by the University Institutional Review Board for Human Participants Research.

Experimental Protocol

Participants visited the University of Memphis Musculoskeletal Analysis Laboratory once for examination and testing. Individuals were screened for inclusion criteria as well as provided informed consent and a completed a PAR-Q form. Prior to testing, participants were fitted for two sports bras including a high support (Ultimate, SheFit Inc., Hudsonville, MI, USA) and low support sports bra (Flex, SheFit Inc., Hudsonville, MI, USA). Bra Fittings were conducted per the manufacturer's instructions.

Participants then completed a 10-minute warm-up of light aerobic activity and dynamic stretching. After warming up, each participant's preferred running speed (PREF) was determined by having participants run across a 25-meter runway at their preferred running pace. Electronic timing gaits (63501 IR; Lafayette Instruments Inc., Lafayette, IN) and electronic timer (Model 54035A; Lafayette Instruments Inc., Lafayette, IN) were used to measure the time interval to cross a 3-meter section located in the middle of the 25-meter runway. The average running speed from three trials was used to calculate each participant's PREF. The SLOW and FAST running speeds were PREF ± 15%, respectively.

Once preferred running speed was determined, retroreflective markers placed on participants. 14 mm Retro-reflective markers were placed bilaterally on the participant's lower extremities and trunk for motion capture. Anatomical markers were placed, bilaterally, on the first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral epicondyles, greater trochanter of the femurs, and iliac crests to define each segment. Rigid clusters of four markers were used to track the left and right shank and thigh segments, as well as the pelvis. The left and right feet were tracked using three individual markers secured on the posterior superior, inferior, and lateral aspects of the shoe. Retro-reflective markers were placed on the upper body, including the: T1 vertebrae, left and right acromion processes (of the scapula), sternum, spine, and left and right nipples. Nipple markers were self-placed by participants and were used to track breast displacement [5].

Running trials were performed on a Bertec instrument treadmill (1200 Hz, Bertec, Inc., Columbus, OH). Kinematic data were recorded using a 10-camera motion capture system (240Hz, Qualisys AB, Goteborg, Sweden). Three breast support conditions were used: bare-chested or no support (CON), low support (LOW), and high support (HIGH). Each participant then performed three-minutes of running on an instrumented treadmill for each speed-support combination, accumulating to 27 minutes of total running. Support conditions HIGH and LOW were randomized for each participant with CON always occurring last. No data were collected during the first minute of treadmill running for each trial to allow the participants to acclimate to the treadmill. 120s of kinematic and kinetic data were collected during minutes two and three of running. Participants were given several minutes of rest between each running bout to avoid fatigue.

Data Analyses

Breast size and displacement

A criterion for inclusion into this study was a self-reported breast size of B-cup to D-cup. Bust and ribcage circumferences were taken for each participant. Measured breast size was then determined as the difference between bust and ribcage measurements. This difference was then used to classify breast sizes into the following categories: A-cup = 1-inch difference, B-cup = 2-inch difference, C-cup = 3-inch difference, D-cup = 4-inch difference, and DD-cup = 5-inch difference. None of the participants in this study had a measured breast size greater than DD.

A novel method was used to measure breast displacement. First, the trunk was defined as a rigid segment identified by retroreflective markers on the acromion processes of the left and right scapula and the sternum. Two virtual markers were then created using Visual3D (C-Motion Inc., Bethesda, MD, USA) located directly behind the right nipple and left nipple, with their anteroposterior positioning being the same as the sternum marker. The virtual markers moved in relationship to the rigid torso segment mentioned above, allowing calculated planar breast displacement to be relative to planar trunk motion. Virtual marker and right nipple kinematic data were filtered at 10Hz using a 4^{th} order low-pass Butterworth filter [5]. Anteroposterior, mediolateral, and vertical breast displacement were then calculated for each stride, for each speed-support condition. Strides were time normalized to 100 percent of stride time and displacement in each plane was calculated as the distance between the minimum right nipple location relative to the virtual marker and the maximum right nipple location relative to the virtual marker. Average displacement was calculated over 15 strides for each speed-support condition, for each participant. This method of calculating breast displacement allows for displacement, in each plane, to be calculated relative to trunk motion.

Temporospatial Variables

Cadence, average stride length, stride time, and ground contact times for the right foot were calculated for each participant, in each support-condition combination. Average stride length was determined by multiplying each participant's stride frequency (strides/s) by their running speed. Ground contact time was determined as the time between initial foot contact and toe off for the right foot and was averaged over 10 strides. Stride time was calculated as the time between consecutive heel strikes of the right foot and was averaged over 10 strides.

Largest Lyapunov's Exponent

To avoid the removal of small fluctuations that may be meaningful biological information, the kinematic data used for nonlinear analysis were not filtered. Unfiltered sagittal plane joint angles for the hip, knee, and ankle were normalized to 100 data points per stride, and 134 strides from each speed-support condition combination were used for analysis, as this was the greatest number of strides shared between all participants/conditions. This normalization method allowed for each time series to have the same length, number of strides, and number of data points per stride [10, 45]. Each stride was defined from right foot heel strike to right foot heel strike, determined using ground reaction force assignments with 50N threshold. Joint angles were calculated using X-Y-Z Cardan rotation and the ankle, knee, and hip angles were expressed in the shank, thigh, and pelvis coordinate systems respectfully. Angular position polarity was defined using the right-hand rule.

The Largest Lyapunov's Exponent (LyE) was then calculated using the *lyapunovExponent.m* MATLAB function according to the methods given by Rosenstein et. al where attractor dynamics for the normalized time series were first constructed as X: [42, 51]

$$X = [X_i \; X_2 \; \ldots \; X_M]^T \qquad \text{(Eq. 1)}$$

Reconstruction delay, *J*, and embedding dimension, *m*, for each times series were determined using an average mutual information algorithm and false nearest neighbors algorithm through the *phaseSpaceReconstruction.m* function [43, 51]. The time lag and embedding dimension was calculated, for the ankle, knee, and hip, individually, for each speed-support condition. LyE was extracted from 0-1 strides for each speed-support combination, for each participant [10].

Statistical Analysis

A Shapiro-Wilk test of normality was used to assess the distribution of the dependent variables and determined that the selected biomechanical data were normally distributed. A 3x3 repeated measures analysis of covariance was used to determine the effect of breast support and speed while covarying for self-reported breast size. Because of the low observed statistical power, one-tailed t-tests were used to for *post hoc* comparison of group means if a significant effect was observed. A significance level of $\alpha = 0.05$ was used. Self-reported breast was used as the covariate, as it was determined to be a better representation of the participant population than measured breast support. Cohen's *d* effect size was used to determine effect size in the *post hoc* comparisons using Hopkin's interpretations [52,53] : small: $|d| < 0.6$; moderate: $0.6 < |d| < 1.2$; large: $|d| > 1.2$.

Results

Participant Information and Running Speeds

Participant information, anthropometric variables, and running speeds are provided in **Table I**.

Table I. Participant average (± SD) anthropometrics and running speeds

Age (yrs.)	**Height (m)**	**Mass (kg)**	**Bust (cm)**	**Ribcage (cm)**	**SLOW (m/s)**	**PREF (m/s)**	**FAST (m/s)**
24 ± 3	1.65 ± 0.05	59.86 ± 4.41	85.13 ± 4.27	75.17 ± 5.04	2.17 ± 0.17	2.55 ± 0.21	2.93 ± 0.24

Breast Size and Displacement

Self-reported breast size and measured breast size results are provided in **Table II,** as well as the number of participants in each group that had the same self-reported and measured breast size**.**

No speed-by-support interaction was observed for relative vertical breast displacement ($F = 0.35$, $p = 0.84$). There was a significant effect of support and speed ($F = 141.56$, $p < 0.01$, and $F = 7.00$, $p < 0.01$, respectively). *Post hoc* analysis for the effect of support revealed that each condition was significantly different from the other, with vertical displacement decreasing as support level increased ($p < 0.01$). Comparison of means for the effect of speed on vertical displacement found no difference between SLOW and PREF ($p = 0.12$); however, FAST was different than both SLOW and PREF, with displacement increasing with increasing running speed (p <0.01, and p <0.01, respectively). See **Figure II** for vertical relative breast displacement.

No interaction between speed and support was found for mediolateral relative breast displacement (F = 0.48, p =0.75). An effect of support and an effect of speed were observed for (F = 101.75, p <0.01, and F= 8.73, p <0.01, respectively). *Post hoc* analysis for the effect of support found all support conditions to be significantly different (p <0.01). Pairwise comparison of the means for the effect of speed found no difference between SLOW and PREF (p = 0.21) but FAST was significantly different that SLOW and PREF, with mediolateral displacement increasing with increasing running speed (p <0.01 and p <0.01, respectively). See **Figure III** for mediolateral relative breast displacement.

No interaction between speed and support was observed for anteroposterior relative displacement (F = 0.58, p = 0.68). An effect of support was observed but no effect of speed was found (F = 3.69, p < 0.01, and F = 2.16, p = 0.12). *Post hoc* analysis for the effect of support found no significance between LOW and HIGH (p = 0.19) but found that CON was significantly greater than LOW and HIGH (p <0.01 and p = 0.01, respectively). See **Figure IV** for anteroposterior relative breast displacement. See **Table III** for breast displacement effect sizes.

Table II. Self-reported and measured breast sizes

	Self-reported	**Measured**	**Same Measured and Self-reported**
A-cup	0	1	0
B-cup	4	1	0
C-cup	6	2	1
D-cup	1	5	1
DD-cup	2	4	1

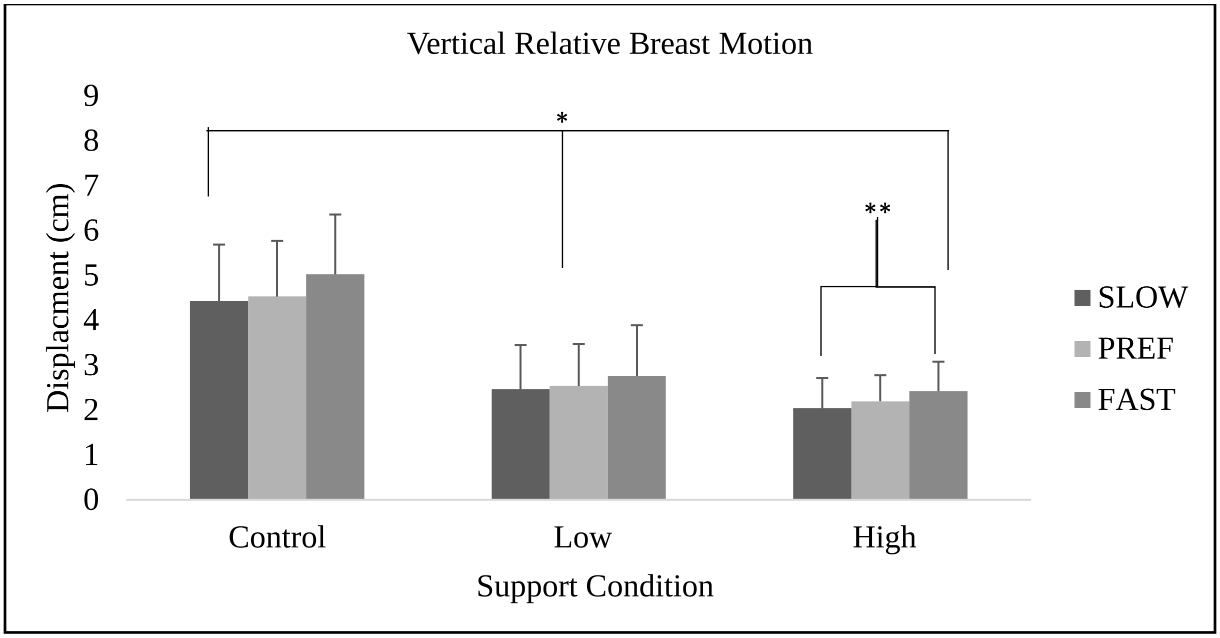

Figure II. Vertical relative breast displacement for each speed-support condition (mean ± SD). * denotes significant effect of support. ** denotes significant effect of speed.

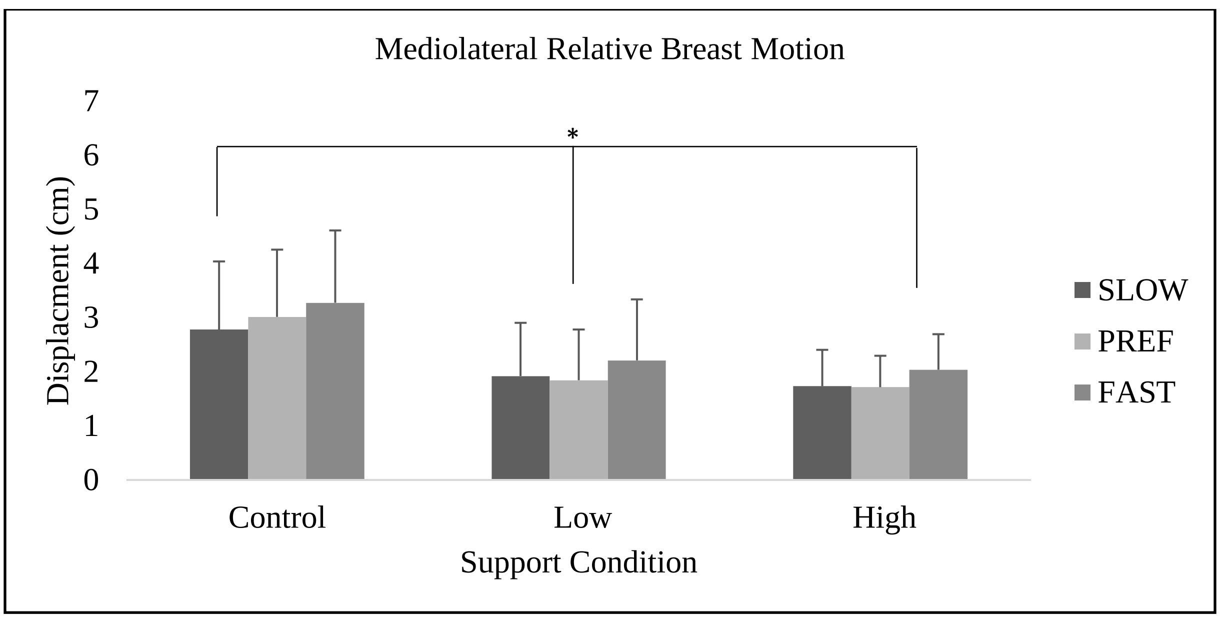

Figure III. Mediolateral relative breast displacement for each speed-support condition (mean ± SD). * denotes significant effect of support.

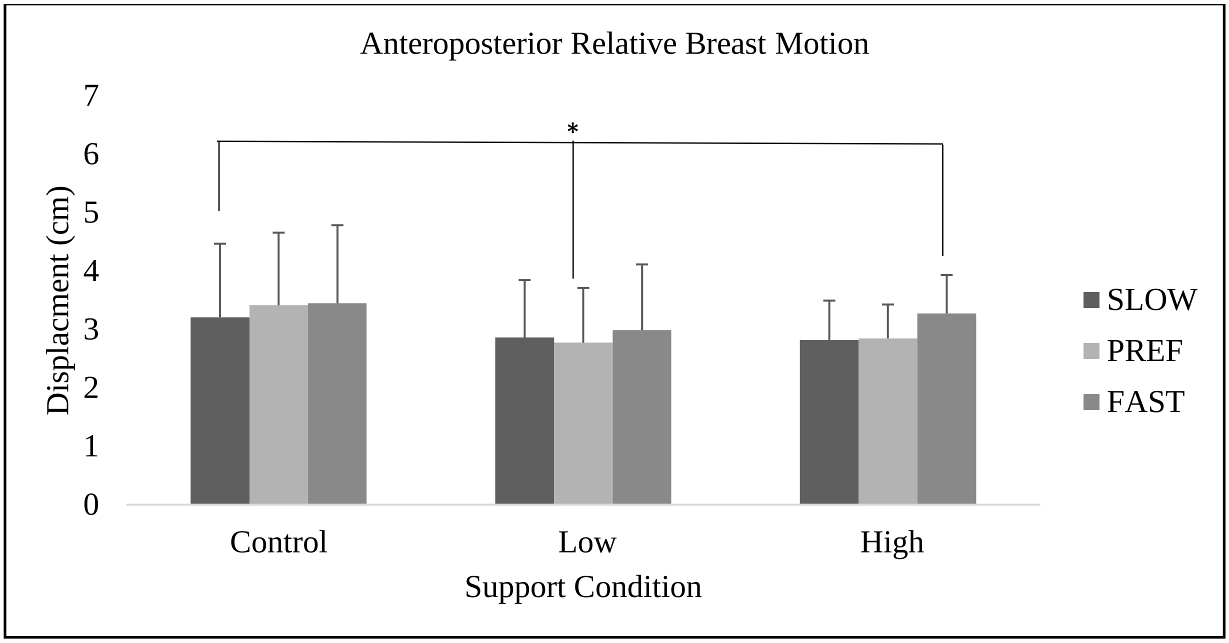

Figure IV. Anteroposterior relative breast displacement for each speed-support condition (mean ± SD). * denotes significant effect of support.

Table III. Significant breast displacement variables and effect sizes

	p	***d***
Vertical Relative Displacement (cm)		
SLOW-PREF	0.12	- 0.05
SLOW-FAST	< **0.01**	-0.26
PREF-FAST	< **0.01**	-0.21
CON-LOW	< **0.01**	1.32
CON-HIGH	< **0.01**	1.52
LOW-HIGH	< **0.01**	0.43
Mediolateral Relative Displacement (cm)		
SLOW-PREF	0.211	1.09
SLOW-FAST	< **0.01**	1.16
PREF-FAST	< **0.01**	0.22
CON-LOW	< **0.01**	-0.05
CON-HIGH	< **0.01**	-0.38
LOW-HIGH	< **0.01**	-0.32
Anteroposterior Relative Displacement (cm)		
CON-LOW	< **0.01**	0.50
CON-HIGH	**0.01**	0.42
LOW-HIGH	0.19	-0.13

Ankle LyE

No speed-by-support interaction was observed at the ankle (F = 0.20, $p = 0.94$). No effect of breast support was found, but an effect of speed was observed (F = 0.22, $p = 0.81$, and F = 8.44, $p < 0.01$, respectively). *Post hoc* analysis revealed no significant difference between SLOW and PREF (p 0.24, $d = 0.01$), but that FAST was significantly lower than both SLOW and PREF ($p < 0.01$, $d = 0.54$ and $p < 0.01$, $d = 0.45$, respectively).

Knee LyE

No speed-by-support interaction was observed at the knee (F =1.41, and $p = 0.24$). Similarly, no effect of support or speed was observed for knee joint LyE values (F = 2.34, $p = 0.10$, and F= 0.16, $p = 0.85$, respectively).

Hip LyE

At the hip, no support-by-speed interaction was observed (F = 0.85, $p = 0.50$). However, an effect of support was observed, with no effect of speed (F = 4.02, $p = 0.02$, and F = 0.96, $p = 0.38$, respectively). *Post hoc* revealed that LOW had significantly smaller LyE values than CON and HIGH, indicating greater LDS ($p < 0.01$, $d = 0.45$, and $p = 0.02$, $d = 0.42$). No difference was observed between CON and HIGH ($p = 0.48$, $d = 0.01$). Effect means, average LyE values, and input parameters, J and m, for each joint are presented in **Table III**, **Table IV**, and **Table V**, respectively. See **Appendix A** for individual LyE values for each participant, for each speed-support condition.

Table IV. Effect means (± SD) for Largest Lyapunov's Exponent

	Control	**Low**	**High**	**SLOW**	**PREF**	**FAST**	Interact. (*p*)	Support (*p*)	Speed (*p*)
Ankle (λ) †	0.77 ± 0.12	0.78 ± 0.11	0.78 ± 0.11	0.80 ± 0.10	0.79 ± 0.11	0.74[ab] ± 0.12	0.94	0.81	< **0.01**
Knee (λ)	1.11 ± 0.09	1.10 ± 0.07	1.13 ± 0.07	1.12 ± 0.08	1.11 ± 0.08	1.11 ± 0.08	0.24	0.10	0.85
Hip (λ) ††	0.95 ± 0.10	0.90[cd] ± 0.11	0.94 ± 0.11	0.94 ± 0.12	0.94 ± 0.12	0.91 ± 0.08	0.50	**0.02**	0.38

Note: †,†† denotes significant effect of speed and effect of support, respectively. a,b denotes significant difference from SLOW and PREF, respectively. c,d denotes significance difference compared to CON and HIGH, respectively.

Table V. Average (± SD) Largest Lyapunov's Exponent

Speed		**SLOW**			**PREF**			**FAST**	
Support	Control	Low	High	Control	Low	High	Control	Low	High
Ankle† (λ)	0.79 ± 0.10	0.81 ± 0.10	0.80 ± 0.10	0.78 ± 0.10	0.79 ± 0.10	0.80 ± 0.11	0.74 ± 0.15	0.74 ± 0.11	0.74 ± 0.10
Knee (λ)	1.11 ± 0.08	1.12 ± 0.07	1.12 ± 0.08	1.11 ± 0.07	1.08 ± 0.08	1.14 ± 0.08	1.13 ± 0.10	1.09 ± 0.06	1.12 ± 0.05
Hip †† (λ)	0.95 ± 0.09	0.90 ± 0.14	0.96 ± 0.11	0.96 ± 0.11	0.88 ± 0.10	0.97 ± 0.12	0.93 ± 0.10	0.91 ± 0.05	0.91 ± 0.08

Note: †,†† denotes significant effect of speed and effect of support, respectively.

Table VI. Average (± SD) lag, *J*, and embedding dimension, *m*, for LyE

		SLOW			PREF			FAST		
		Control	Low	High	Control	Low	High	Control	Low	High
Ankle	*M*	4.00 ± 0.00	4.00 ± 0.00	4.00 ± 0.00	3.77 ± 0.44	3.85 ± 0.38	3.77 ± 0.44	3.85 ± 0.38	3.92 ± 0.28	4.00 ± 0.00
	J	16.62 ± 2.99	16.92 ± 3.50	16.54 ± 3.80	15.85 ± 5.58	14.38 ± 4.05	16.08 ± 5.56	14.62 ± 3.75	14.69 ± 3.75	14.77 ± 3.83
Knee	*m*	3.08 ± 0.28	3.23 ± 0.44	3.15 ± 0.38	3.00 ± 0.00	3.00 ± 0.00	3.08 ± 0.28	3.08 ± 0.28	3.00 ± 0.00	3.08 ± 0.28
	J	12.23 ± 1.09	11.85 ± 1.46	12.15 ± 1.14	10.69 ± 2.21	11.00 ± 2.35	11.31 ± 2.32	10.77 ± 2.68	11.08 ± 1.61	11.69 ± 1.55
Hip	*m*	3.38 ± 0.51	3.54 ± 0.52	3.46 ± 0.52	3.23 ± 0.44	3.15 ± 0.38	3.08 ± 0.28	3.23 ± 0.44	3.15 ± 0.38	3.31 ± 0.48
	J	14.54 ± 2.88	17.31 ± 4.39	14.62 ± 3.01	13.69 ± 3.75	14.46 ± 4.68	14.08 ± 4.37	13.69 ± 4.77	14.00 ± 3.96	15.46 ± 3.78

Temporospatial Variables

Temporospatial variable effect means and speed-support means are found in **Table VII** and **Table VIII**, respectively. No speed-by-support interaction was observed for stride length ($F = 0.63$, $p = 0.64$). No effect of support was observed, however there was an effect of speed ($F = 0.71$, $p = 0.49$, and $F = 1066.97$, $p < 0.01$, respectively). *Post hoc* analysis revealed all speed conditions were significantly different from each other, as stride length increased with increasing running speed ($p < 0.01$).

For cadence, no speed-by-support interaction was observed ($F = 0.85$, $p = 0.50$). No effect of support was found, however there was an effect of speed for cadence ($F = 1.25$, $p = 0.29$, and $F = 63.56$, $p < 0.01$, respectively). *Post hoc* analysis revealed that all speed conditions were significantly different from one another, with cadence increasing with increasing running speed ($p < 0.01$). *Post hoc* analysis revealed a significant difference in stride time between all speeds, with stride time decreasing as running speed increases ($p < 0.01$).

No speed-by-support interaction for stride time was observed (F = 0.88, *p* = 0.48). No effect of support was found, however there was an observed effect of speed (F = 1.28, *p* = 0.28, and F = 54.52, p <0.01, respectively). Similarly, No speed-by-support interaction was observed for ground contact time, as well as no effect of support (F = 0.66, *p* =0.62, and F = 1.06, *p* =0.35, respectively). An effect of speed was observed for ground contact time (F=97.76, *p* < 0.01). *Post hoc* analysis revealed that all speeds were significantly different from each other, with ground contact time decreasing as speed increases (*p* < 0.01). See **Appendix B** for individual temporospatial variables for each participant, for each speed-support condition. See **Table IX.** for significant temporospatial variables and effect sizes.

Table VII. Effect means (± SD) for temporospatial variables

	Control	**Low**	**High**	**SLOW**	**PREF**	**FAST**	Interact. (*p*)	Support (*p*)	Speed (*p*)
Cadence (steps/min)	162.67 ± 10.72	162.28 ± 9.16	163.42 ± 9.93	158.30 ± 9.20	163.6[a] ± 9.56	166.44[ab] ± 9.33	0.50	0.29	**<0.01**
Stride length (m)	1.87 ± 0.24	1.87 ± 0.22	1.86 ± 0.24	1.64 ± 0.11	1.86[a] ± 0.14	2.10[ab] ± 0.15	0.64	0.49	**<0.01**
Stride time (s)	0.74 ± 0.05	0.74 ± 0.04	0.74 ± 0.05	0.76 ± 0.05	0.74[a] ± 0.04	0.72[ab] ± 0.04	0.48	0.28	**<0.01**
Ground contact time (s)	0.26 ± 0.03	0.26 ± 0.03	0.26 ± 0.03	0.28 ± 0.03	0.26[a] ± 0.02	0.24[ab] ± 0.02	0.62	0.35	**<0.01**

Note: †,†† denotes significant effect of speed and effect of support, respectively. [a,b] denotes significant difference from SLOW and PREF,

Table VIII. Average (± SD) temporospatial variables

Speed	SLOW			PREF			FAST		
Support	Control	Low	High	Control	Low	High	Control	Low	High
Cadence (steps/min)	158.42 ± 10.18	157.22 ± 8.44	159.26 ± 9.54	162.90 ± 10.82	163.07 ± 8.77	164.91 ± 8.68	166.68 ± 10.29	166.55 ± 8.34	166.10 ± 9.99
Stride Length (m)	1.64 ± 0.12	1.65 ± 0.11	1.63 ± 0.11	1.87 ± 0.15	1.86 ± 0.14	1.84 ± 0.14	2.10 ± 0.17	2.09 ± 0.14	2.10 ± 0.16
Stride time interval (s)	0.76 ± 0.05	0.77 ± 0.04	0.76 ± 0.05	0.74 ± 0.05	0.74 ± 0.04	0.73 ± 0.04	0.72 ± 0.05	0.72 ± 0.04	0.72 ± 0.05
Ground contact time (s)	0.28 ± 0.03	0.27 ± 0.03	0.28 ± 0.03	0.26 ± 0.03	0.26 ± 0.02	0.26 ± 0.03	0.24 ± 0.02	0.24 ± 0.03	0.24 ± 0.03

Table IX. Dependent variables and effect sizes

	p	***d***
Cadence (steps/min)		
SLOW-PREF	< **0.01**	-0.55
SLOW-FAST	< **0.01**	-0.81
PREF-FAST	< **0.01**	-0.30
Stride Length (m)		
SLOW-PREF	< **0.01**	-1.33
SLOW-FAST	< **0.01**	-1.73
PREF-FAST	< **0.01**	-1.27
Stride Time Interval (s)		
SLOW-PREF	< **0.01**	0.53
SLOW-FAST	< **0.01**	0.78
PREF-FAST	< **0.01**	0.29
Ground Contact Time (s)		
SLOW-PREF	< **0.01**	0.67
SLOW-FAST	< **0.01**	1.11
PREF-FAST	< **0.01**	0.63

Discussion

The purpose of this study was to investigate the effects of breast support on lower extremity joint angle LDS in female runners. The first hypothesis was mostly unsupported as ankle and knee LDS were not different among support conditions, but hip LDS was found to be greater in LOW compared to both CON and HIGH support conditions. However, as expected, breast displacement did decrease with increasing levels of breast support which confirms the findings of previous research that as breast support increases, breast displacement in all three planes decreases [5-7]. It should be noted that the low support sports bra (Flex, SheFit Inc., Hudsonville, MI, USA) used in this study functions more closely to a typical high support sports bra. This can be reflected in the lack of difference found in mediolateral and anteroposterior displacement between the two support conditions and was commented on by participants. Similarly, the hypothesis that increasing running speed would decrease joint angle LDS was unsupported, as no differences at the hip or knee were observed between running speeds and ankle LDS was found to be greater at FAST compared to SLOW and PREF speed conditions. The second hypothesis of this study was also unsupported as changes in breast support did not lead to altered temporospatial variables of running. It was observed, however, that increases in running speed led to increases in stride length and cadence, as well as decreases in stride time and ground contact time, which was expected.

Local Dynamic Stability

The findings of this study suggest that the perturbations caused by breast motion are attenuated from the hip and above, and that LOW has greater hip joint angle LDS compared to HIGH and CON conditions. This finding is partially congruent with the proposed hypothesis, that LOW would have greater LDS than CON. This suggests that breast support influences neuromuscular strategy at the hip but not at the ankle or knee joint when using LDS.

During running, it is suggested that female runners will adopt a motor behavior that will best reduce exercise-induced breast pain. It was postulated that pain mitigation would be attenuated through the dampening of the imposed ground reaction forces by the knee acting as a strut. To mitigate vertical breast motion and resulting pain (particularly with less breast support), female runners would express less LDS at the knee. However, the current findings demonstrated that the imposed perturbation associated with less breast support may have affected LDS of local structures including the trunk and pelvis but did not alter the LDS of the knee or ankle. Given the location of breast tissue on the anterior portion of the trunk, it could be postulated that the system adopts a local-to-remote pattern of perturbation attenuation, meaning that the segment closest to the imposed perturbation would be most affected by the perturbation followed by the adjacent segment. If the perturbation is not dissipated by the local segment, then the imposed perturbation would alter LDS of the subsequent segment in the link-chain-model. This concept would therefore suggest that greater changes in LDS would be seen at the trunk, pelvis, and hip compared to the knee and ankle in response to changes in breast support. Current findings show only changes in hip LDS in response to changes in breast support and not at the ankle or knee. Findings from previous literature support this concept of a local-to-remote response showing that both trunk and pelvis rotation range-of-motion, as well as peak rotation, increase as breast support increases but that lower extremity joint kinematics do not change[7]. To better understand this local-to-remote response, future research should investigate the effects of breast support on trunk and pelvis LDS.

Current findings, as well as findings from previous research, shows that breast motion is greater in the control condition compared to supported conditions [3, 7, 27]. It was expected that greater breast motion would perturb the system more, as well as lead to greater breast pain and discomfort, and thus less LDS in CON and LOW compared to HIGH support. However, these findings suggested that LOW provides greater neuromuscular freedom, or robustness to perturbations, compared to CON and HIGH. As previously state, subjective participant feedback suggests that the low support sports bra used in this study functions more like a traditional high support sports bra, and that the high support bra used is even more constrictive. This feedback parallels the, although statistically different, similar breast displacements between LOW and HIGH conditions. The perceived increase in restrictiveness may have perturbed the system if it was unfamiliar to the participant or if considered uncomfortable. These data suggest a U-shaped relationship between support condition and LDS, suggesting that there may exist a preferred level of support. Dynamical Systems Theory (DST) would suggest that these decreases in LDS in CON and HIGH, compared to LOW, indicate a possible transition to a new, more stable movement pattern, where should the experienced perturbation be large enough, changes in attractor dynamics would occur as the neuromuscular system searches for a new, more stable behavior. However, the current findings show that while the experienced perturbations altered LDS, these perturbations, seen in less LDS in CON, were not great enough to elicit new attractor dynamics or temporospatial changes during running.

While increasing speed was shown to increase multiplanar breast motion, a speed effect was only observed at the ankle, and not at the knee or hip. It was hypothesized that the greatest differences in LDS between support conditions would be seen at FAST speed condition due to this increased breast motion, with greater support leading to greater LDS; however, no interaction between speed and support was observed. This finding of no interaction may be attributed to the U-shaped relationship between support and LDS, as well as the support similarities between low and high support sports bras and the perceived constrictiveness of the high support. These findings also suggest that the increases in breast motion due to running speed may not proportionally increase the imposed perturbation, as no effect of speed was observed at the hip which is the most local investigated segment to the breast tissue. The effect of speed found at the ankle may be more attributed to the foot interacting with the treadmill belt at higher speeds. Future research should investigate this possible interaction by using more traditional sports bras with greater contrasting support levels.

Similar to the small differences in support levels between LOW and HIGH, the differences in running speeds used my not have been large enough to adequetly perturb the system. Previous literature showing a speed effect for LDS during running used running speeds of 3m/s to 9m/s, with LDS decreasing as speed increases [12]. These changes in running speed are much greater than the fifteen percent changes used in the present study, where the fastest single speed ran was 3.22 m/s during the FAST condition. It is important to note, however, that Look et al. observed less than a 0.01 difference in right knee and hip LyE values between 3 m/s and 4 m/s for healthy adults and that large differences in LDS were not observed until faster speeds. Ultimately, while the changes in speed used in the current study were large enough to change the temporospatial components of running, these speeds may not have been fast enough to adequately challenge the neuromuscular system.

The differences seen at the ankle, due to speed, with FAST speed condition having greater LDS than the two slower conditions may be attributed to methodological flaws in the study design. Preferred running speed was determined by instructing the participant to run over a

25m runway at their preferred 5k training pace and running speed was measured over a 3m period in the middle. 25m may not be sufficient distance to accurately run at one's preferred pace, with the resulting speed being either faster or slower to their actual preferred speed. Should the actual preferred running speed be better reflected by the FAST condition speeds used, participants may actually be in their preferred attractor state, meaning deviation from these FAST speeds, either faster or slower, would result in decreased LDS. In a study looking at variability in running, Jordan et. al observed a U-shaped relationship between long-range correlations and percentage of preferred running speed for discrete gait variables such as stride length, stride interval, and impulse [54]. Long-range correlations were measured using detrended fluctuation analysis, a nonlinear measurement which determines if a discrete metric is dependent upon the same metric at remote previous times and assumes that this dependence decays over time [54]. The authors suggest that these findings indicate a less constrained stride and that the system is more readily adaptable compared to faster or slower speeds. For the present study, this interpretation may suggest that the preferred running speed used may be slower than the participant's *actual* preferred running speed.

A factor that may have influenced LyE analysis is foot strike pattern. A study in 2014 found that foot strike pattern significantly affected knee stiffness, with forefoot strikers exhibiting greater knee stiffness that rearfoot strikers [55]. This same study also found that rearfoot strikers exhibit significantly greater load absorption at the knee compared to forefoot strikers. Knee stiffness has been suggested to be a resultant of neuromuscular strategy and the findings from Hamill et. al may suggest that LDS at the knee may be influenced by strike pattern [56] . While no known research has investigated the impact of strike pattern on lower extremity LDS, Ekizos et. al did find that a transition from shod to barefoot running, resulting in a more forefoot strike approach, decreased LDS of the trunk [15]. The current study did not control for foot strike pattern in the inclusion criteria, resulting in a mixture of rearfoot and forefoot runners. Future research should be done to investigate the role of strike pattern on lower extremity LDS. Future investigations should also be done assessing the impact of breast support on knee stiffness.

Limitations

Several limitations for the current study have been mentioned such as foot strike pattern and the distance used to determine preferred running speed. Small sample size will also impact statistical analysis and interpretation due to the low statistical power. The sample size within each breast size category also effects statistical analysis and interpretation as breasts of differing sizes, shapes, and densities experience varying magnitudes of displacement during running due to the inertial properties of the breast tissue. Previous breast support literature focused only on females with size D-cup breasts [6, 7]. While females of with D-cup size breasts experience large amounts of breast displacement, these are not the only females that run, and females of differing breast sizes may have differing neuromuscular responses to breast motion. Additionally, another limitation to the current study was that the fifteen percent difference in running speed may not have been large enough to properly challenge the neuromuscular system.

Conclusion

The findings of this study suggest that using a low support sports bra increases hip joint angle LDS compared no or high breast support. Overall, the current data suggest that there may exist a preferred level of breast support, and that the neuromuscular system adopts a local-to-remote pattern of perturbation attenuation; however, future research is needed to understand the effects of breast support on trunk and pelvis LDS. Similarly, future research is needed to understand the performance implications of LDS and investigate the relationship between LDS and running efficiency and performance.

Appendix A

Table X. Individual ankle LyE values for each participant

	SLOW			PREF			FAST		
Subject	Control	Low	High	Control	Low	High	Control	Low	High
S1	0.80	0.73	0.73	0.69	0.74	0.63	0.65	0.77	0.80
S2	0.72	0.75	0.68	0.64	0.79	0.64	0.66	0.59	0.62
S3	0.74	0.78	0.75	0.79	0.88	0.81	0.68	0.72	0.75
S4	0.75	0.79	0.84	0.95	0.74	1.03	0.65	0.68	0.71
S5	0.91	0.93	0.94	0.89	0.86	0.90	0.88	0.76	0.83
S6	0.74	0.75	0.70	0.78	0.73	0.69	0.75	0.71	0.70
S7	0.85	0.82	0.83	0.82	0.84	0.82	0.67	0.86	0.75
S8	1.06	1.00	0.95	0.86	0.90	0.93	1.16	0.85	0.90
S9	0.84	0.87	0.93	0.73	0.78	0.85	0.80	0.77	0.83
S10	0.67	0.65	0.69	0.57	0.58	0.78	0.56	0.47	0.53
S11	0.81	0.83	0.77	0.74	0.80	0.80	0.57	0.78	0.57
S12	0.68	0.69	0.69	0.82	0.67	0.69	0.79	0.75	0.76
S13	0.76	0.94	0.85	0.81	0.99	0.87	0.78	0.87	0.82

Table XI. Individual knee LyE values for each participant

	SLOW			PREF			FAST		
Subject	Control	Low	High	Control	Low	High	Control	Low	High
S1	1.03	1.09	1.10	1.09	1.08	1.03	1.14	1.10	1.14
S2	1.17	1.17	1.10	1.10	1.21	1.14	1.09	1.10	1.05
S3	1.05	1.12	1.07	1.09	1.01	1.15	1.19	1.07	1.16
S4	0.98	1.11	1.06	1.11	1.00	1.30	1.03	1.03	1.12
S5	1.05	1.04	1.10	1.03	0.99	0.99	1.10	1.03	1.03
S6	1.09	1.05	1.04	1.13	1.06	1.04	1.12	1.06	1.05
S7	1.14	1.20	1.04	1.09	1.17	1.15	1.07	1.08	1.12
S8	1.23	1.15	1.12	1.22	1.05	1.23	1.39	1.14	1.18
S9	1.04	1.13	1.22	1.04	1.07	1.17	1.03	1.08	1.09
S10	1.11	1.08	1.12	0.99	1.05	1.19	1.03	0.98	1.14
S11	1.05	1.01	1.04	1.07	1.01	1.08	1.14	1.09	1.18
S12	1.25	1.21	1.31	1.22	1.05	1.16	1.13	1.21	1.16
S13	1.20	1.24	1.21	1.23	1.25	1.19	1.22	1.21	1.17

Table XII. Individual hip LyE values for each participant

	SLOW			PREF			FAST		
Subject	Control	Low	High	Control	Low	High	Control	Low	High
S1	0.91	0.98	1.09	0.99	1.01	1.08	0.96	0.93	0.92
S2	1.09	0.95	1.05	1.02	0.97	0.81	1.00	0.88	0.93
S3	0.86	0.80	0.77	0.85	0.74	0.81	0.84	0.84	0.82
S4	0.86	0.79	0.88	1.18	0.79	1.22	0.95	0.84	0.94
S5	1.00	0.73	1.03	0.92	0.69	0.89	0.76	0.81	0.70
S6	0.90	0.98	0.99	0.91	0.86	0.93	0.90	0.88	0.85
S7	0.89	0.97	0.80	0.80	0.96	0.85	0.82	0.92	0.89
S8	1.06	1.08	1.06	1.03	0.88	1.02	1.16	0.98	0.92
S9	1.07	1.11	1.04	0.97	0.97	1.08	1.02	0.98	1.02
S10	0.83	0.64	1.03	0.87	0.94	1.11	0.86	0.92	1.03
S11	0.86	0.79	0.76	0.80	0.75	0.87	0.85	0.88	0.87
S12	1.03	0.91	1.00	1.12	0.95	0.89	0.93	0.97	0.95
S13	0.97	1.04	0.99	1.03	0.95	0.98	1.01	0.95	0.96

Appendix B

Table XIII. Individual ground contact times (seconds) for each participant

	SLOW			PREF			FAST		
Subject	**Control**	**Low**	**High**	**Control**	**Low**	**High**	**Control**	**Low**	**High**
S1	0.26	0.27	0.27	0.25	0.24	0.24	0.23	0.23	0.22
S2	0.26	0.26	0.26	0.25	0.25	0.24	0.23	0.23	0.23
S3	0.24	0.26	0.27	0.24	0.24	0.23	0.22	0.22	0.20
S4	0.24	0.22	0.23	0.23	0.22	0.21	0.21	0.20	0.20
S5	0.32	0.28	0.35	0.29	0.28	0.29	0.26	0.29	0.26
S6	0.26	0.25	0.25	0.25	0.24	0.23	0.23	0.22	0.22
S7	0.29	0.26	0.28	0.25	0.25	0.26	0.25	0.25	0.24
S8	0.27	0.25	0.24	0.24	0.23	0.23	0.21	0.21	0.22
S9	0.30	0.26	0.27	0.27	0.27	0.23	0.25	0.25	0.23
S10	0.29	0.30	0.32	0.26	0.28	0.31	0.25	0.26	0.30
S11	0.31	0.31	0.30	0.29	0.28	0.28	0.27	0.26	0.27
S12	0.31	0.33	0.30	0.31	0.29	0.29	0.28	0.27	0.27
S13	0.27	0.27	0.27	0.26	0.26	0.26	0.24	0.24	0.24

Table XIV. Individual cadence (steps/min) for each participant

	SLOW			PREF			FAST		
Subject	**Control**	**Low**	**High**	**Control**	**Low**	**High**	**Control**	**Low**	**High**
S1	164.50	167.00	171.00	172.00	173.00	176.14	174.50	175.00	174.00
S2	163.50	161.18	162.00	167.00	165.50	168.50	170.83	168.24	170.28
S3	164.36	155.50	164.00	169.00	167.05	168.50	172.50	170.59	169.50
S4	167.00	166.18	164.50	169.50	169.50	170.09	172.00	173.50	171.00
S5	160.50	146.50	148.00	150.98	151.00	152.00	156.00	153.00	152.00
S6	149.60	150.50	150.50	151.04	154.50	157.00	156.50	159.50	159.13
S7	160.00	160.50	159.00	164.00	169.50	166.00	170.50	169.50	166.50
S8	171.00	166.50	168.50	173.00	170.50	172.00	175.00	175.00	172.50
S9	162.00	163.00	172.50	175.00	169.00	178.50	177.39	174.80	182.50
S10	135.50	147.50	142.50	138.68	145.04	145.00	142.00	151.54	145.00
S11	149.50	150.00	151.00	155.00	154.50	157.00	159.00	158.00	159.18
S12	146.00	144.50	152.04	163.00	164.86	165.04	167.09	166.00	166.00
S13	166.00	165.00	164.86	169.50	166.00	168.00	173.50	170.50	171.68

Table XV. Individual stride time interval (seconds) for each participant

	SLOW			PREF			FAST		
Subject	**Control**	**Low**	**High**	**Control**	**Low**	**High**	**Control**	**Low**	**High**
S1	0.73	0.72	0.70	0.70	0.69	0.68	0.69	0.69	0.69
S2	0.73	0.74	0.74	0.72	0.73	0.71	0.70	0.71	0.70
S3	0.73	0.77	0.73	0.71	0.72	0.71	0.70	0.70	0.71
S4	0.72	0.72	0.73	0.71	0.71	0.71	0.70	0.69	0.70
S5	0.75	0.82	0.81	0.79	0.79	0.79	0.77	0.78	0.79
S6	0.80	0.80	0.80	0.79	0.78	0.76	0.77	0.75	0.75
S7	0.75	0.75	0.75	0.73	0.71	0.72	0.70	0.71	0.72
S8	0.70	0.72	0.71	0.69	0.70	0.70	0.69	0.69	0.70
S9	0.74	0.74	0.70	0.69	0.71	0.67	0.68	0.69	0.66
S10	0.89	0.81	0.84	0.87	0.83	0.83	0.85	0.79	0.83
S11	0.80	0.80	0.79	0.77	0.78	0.76	0.75	0.76	0.75
S12	0.82	0.83	0.79	0.74	0.73	0.73	0.72	0.72	0.72
S13	0.72	0.73	0.73	0.71	0.72	0.71	0.69	0.70	0.70

Table XVI. Individual stride length (meters) for each participant

	SLOW			PREF			FAST		
Subject	**Control**	**Low**	**High**	**Control**	**Low**	**High**	**Control**	**Low**	**High**
S1	1.64	1.62	1.58	1.85	1.84	1.81	2.13	2.13	2.14
S2	1.43	1.45	1.44	1.62	1.63	1.60	1.81	1.84	1.82
S3	1.66	1.76	1.67	1.91	1.93	1.92	2.15	2.17	2.19
S4	1.65	1.66	1.68	1.92	1.92	1.91	2.18	2.16	2.19
S5	1.46	1.60	1.58	1.83	1.83	1.82	2.03	2.07	2.08
S6	1.93	1.91	1.91	2.25	2.20	2.16	2.49	2.45	2.45
S7	1.64	1.64	1.65	1.89	1.83	1.87	2.08	2.10	2.13
S8	1.59	1.64	1.62	1.86	1.89	1.87	2.11	2.11	2.14
S9	1.68	1.67	1.58	1.84	1.90	1.80	2.08	2.11	2.03
S10	1.75	1.61	1.67	2.02	1.93	1.93	2.26	2.12	2.22
S11	1.57	1.56	1.55	1.78	1.79	1.76	2.00	2.01	1.99
S12	1.61	1.63	1.55	1.69	1.67	1.67	1.90	1.92	1.92
S13	1.66	1.67	1.67	1.84	1.88	1.86	2.01	2.04	2.03

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