

Monitoring Decelerations: GPS Sensitivity, Specificity, and Biomechanical Responses

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ABSTRACT

Rapid decelerations are crucial in multi-directional sports, as an athlete's ability to reduce whole-body momentum contributes to performance outcomes and injury risks. Decelerations require large braking forces, which can cause high loading across joints and surrounding tissues of the lower extremities. Repeated deceleration actions can potentially decrease lower extremity coordinative proficiency and increase injury susceptibility due to altered mechanical loading. Global Positioning System (GPS) units are utilized to quantify deceleration frequency and intensity; however, GPS units do not provide information on the body's biomechanical responses associated with accumulated decelerations. This study (1) evaluated GPS unit's ability to classify detected decelerations into predefined absolute intensity zones using motion capture as the reference, (2) examined if accumulated high-intensity decelerations influenced horizontal deceleration ability, and (3) examined if accumulated decelerations altered lower extremity kinematics during standardized deceleration tasks. To address aim 1, 15 healthy, recreationally active college-aged females (age: 22yrs $\pm$ 2yrs) completed sets of squatting, walking, the Acceleration-Deceleration Assessment (ADA) and 505 change-of-direction (COD) task while wearing GPS units and being recorded with motion capture. GPS derived maximum

decelerations provided by the manufacturer's software were compared to the calculated average deceleration derived from motion capture. Motion capture data were collected as continuous signals, whereas GPS-derived deceleration events were ultimately classified into predefined deceleration intensity zones. Sensitivity and specificity were used to determine how accurately GPS units classified events into the same zones identified by motion capture. Bland-Altman analysis was utilized to assess the agreement between the systems. For aims 2 and 3, participants completed an exercise intervention to accumulate 30 high-intensity deceleration movements before repeating the ADA and 505 COD tasks. Physical exertion was measured using time to complete the exercise intervention and reported RPE scores. Lower extremity kinematics were analyzed at initial contact for consecutive deceleration steps during the ADA, at the initial contact of the 505 COD task and full weight acceptance (zero velocity) the 505 COD task. Of 266 movement trials, only 187 trials were analyzed for GPS sensitivity and specificity. Trials without a GPS detected deceleration action (marked as "N/A") were excluded. GPS sensitivity (0.0–70.6%) decreased with increasing deceleration intensity, while specificity ranged from 69.9–98.9% across the pre-defined absolute deceleration intensity zones. Bland-Altman analysis demonstrated a mean systematic bias of 0.74 m/s² between the two systems. GPS units demonstrated low to moderate sensitivity for classifying deceleration actions into the correct deceleration intensity zone indicating caution should be used when interpreting decelerations for load monitoring when wearing GPS units indoors. Following the exercise intervention, RPE scores increased by 1 to 7 points across participants ($p < .05$), however, only frontal-plane hip abduction angle decreased significantly during the 505 COD task ($p < .05$). Accumulated high intensity decelerations did not significantly alter horizontal deceleration ability and only produced small reductions in hip abduction angle during the 505 COD task.

Monitoring Decelerations: GPS Sensitivity, Specificity, and Biomechanical Responses

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CHAPTER 1: Introduction

Various collegiate and professional-level sports teams are utilizing global positioning systems (GPS) to monitor player demands during training and competitive matches. GPS is a satellite-based navigation system that tracks player movement and monitors external load metrics, including total distance, running speed, and changes in velocity based on signals sent between GPSs and satellites (Cummins et al., 2013; Beato & Keijzer, 2019; Scott, Scott, & Kelly, 2016). Over the last two decades, wearable GPS technology has gained popularity as it provides real-time data on player external loading (player movement across the field and activity profiles) and sports performance (Cummins et al., 2013). Recent advancements in GPS technology have provided the ability to track high velocity movements due to increased sampling frequencies (Scott, Scott, & Kelly, 2016; Jennings et al., 2010; Varley et al., 2012). Validation studies utilizing pre-measured distance and radar guns have compared measures of velocity and distances tracked by GPS to verify the enhancement in accurate data provided by newer model units. Improved measurement accuracy allows coaches and sports scientists to better monitor daily and weekly loads, to optimize performance and reduce injury risks.

Of the various external load metrics, one that has gained particular attention recently is decelerations. Deceleration actions, defined as the ability to slow whole-body momentum quickly and efficiently (Wolfe et al., 2023), occur frequently in multidirectional sports during cutting, changing direction, getting prepared to score a goal, or attempting to beat opponents to the ball. Decelerations have become an important external load metric not only for their frequency, but because of the internal loading (physiological responses to the external load) they impose on the lower extremities. Rapid deceleration movements require high eccentric muscle contractions, and repeated bouts of high eccentric muscle loading lead to increased muscle tissue

microtrauma (Harper & Kiely, 2018; Wolfe et al., 2023; Prudholme et al., 2023). While these internal loads in the musculoskeletal system contribute to increased stress demands in the lower extremities, GPS units are only able to track that a deceleration has occurred based on the body's position and change in velocity. GPS units cannot assess any of the body's biomechanical changes that are associated with accumulated decelerations.

Traditionally, marker-based motion capture has been used to measure lower extremity mechanics during deceleration and change of direction (COD) tasks. The lower extremity acts as a kinetic chain, where each joint has a role in the loading of other joints. For example, during horizontal decelerations, it has been observed that immediate hip, knee, and ankle flexion must occur at initial ground contact to dissipate the forces across all joints and surrounding soft tissue with each foot-ground contact—reducing injury risks (Harper et al., 2024; Hewit et al., 2011). COD tasks further highlight how a wide range of strategies and lower extremity postures are used to cut at varying degrees, which can lead to bodily postures that increase the risk of injury, particularly ACL tears or ankle sprains (Frank et al., 2013; Gollhofer & Gehring, 2017; Imwalle et al., 2009). More extensive research has been done related to decelerations during COD tasks rather than straight-line decelerations alone. Thus, a gap within the literature of mechanics during rapid straight-line deceleration is present.

While marker-based motion capture is useful for examining lower extremity mechanics, incorrect marker placement to establish joint centers or soft-tissue artifact during movements such as sprinting (Antognini et al., 2025) are sources of error. The introduction of markerless motion capture has changed how whole-body kinematics are captured, as no skin markers are needed on the participant. Without skin-based markers, participants can move more naturally. Additionally, not having to utilize skin-based markers allows for non-laboratory use of motion

capture, which makes markerless motion capture more applicable in real-world sports settings rather than a traditional laboratory-based environment. A study by Song et al. (2023) compared traditional marker-based and marker-less motion capture systems across tri-planar movements. They found that these systems were able to track treadmill and overground walking, running, and squatting similarly, while there was higher variation in tracking during countermovement jumps and run and cuts. With markerless motion capture still being relatively new, there are limited published journal articles examining the use of markerless motion capture during high deceleration-specific sporting maneuvers. Decelerations represent a critical metric due to the generation of high mechanical demands on the lower extremities and potential link to injury risk. While research has described lower extremity joint mechanics during straight line deceleration and change of direction tasks, little is known about how GPS-derived deceleration measures correspond to the actual biomechanical postures experienced. Addressing this gap is essential for advancing both performance monitoring and injury prevention.

The aim of the present study is threefold: (1) Evaluate GPS unit's ability to correctly classify deceleration actions into predefined absolute intensity zones using motion capture as the reference, (2) examine if accumulated high-intensity decelerations influence horizontal deceleration ability, and (3) examine whether accumulated decelerations influence lower extremity kinematics during standardized deceleration tasks. By addressing these aims, this study seeks to improve the interpretation of deceleration metrics for athlete monitoring and provide insight into the biomechanical demands that may contribute to performance decrements and injury risk. The first hypothesis of this study is GPS units will demonstrate high sensitivity and specificity in detecting deceleration actions and correctly classifying them into deceleration intensity thresholds relative to motion capture. The second hypothesis is that after accumulation

of high intensity decelerations, there will be a change in the ability to decelerate, including change in pelvis position and number of steps to decelerate during the ADA task.

The third hypothesis is that after the accumulation of high deceleration efforts, there will be a change in lower extremity mechanics utilized to perform deceleration tasks.

Operational Definitions:

External Load: objective quantification of work done during training/matches such as total distance covered, number of accelerations and decelerations, maximum speed, etc. Note that “work” is not traditionally calculated ($W = F \cdot d$). In this instance, work is referred to as the load imposed on a player based on various external metrics, including but not limited to total distance covered, number of accelerations/decelerations, number of sprints, etc.

Internal load: The response of the sum of psychophysiological stress derived from external load.

Total Player Load: the total work on a player over an entire session based on accelerometer data alone. The magnitude of instantaneous changes is combined across three axes (XYZ), to reflect the physical exertion experienced during running or high intensity movements, and activity that is more static, including short distance acceleration/deceleration. The resultant value is scaled by 1000.

Deceleration: A decrease in velocity that takes place for at least half a second for an activity.

Horizontal Deceleration: slowing down horizontal velocity by generating high braking forces to reduce whole body momentum.

Horizontal Deceleration Ability: Ability to rapidly reduce horizontal whole-body momentum in a controlled manner (i.e. number of braking steps, lowered-pelvis position)

Limitations: Did not collect kinetic data. This study did not measure ground reaction forces during braking demands nor the actual internal loads that correspond to these movements.

Current literature provides information regarding the loads corresponding to varying positional changes, however we focused on postural changes. Secondly, previous literature has mainly used markerless motion capture for low acceleration movements such as walking. The cameras being

utilized at a capture frequency of 120Hz, which is a lower frequency than typically seen in marker-based motion capture studies with higher speed movements.

Delimitations: Using only female recreationally active individuals and/or collegiate female athletes. Performing data collection indoors to avoid weather constraints.

CHAPTER 2: Review of the Literature

2.1 Global Positioning Systems in Sports

Wearable technologies such as global navigation satellite systems (GNSS), global positioning systems (GPS), and accelerometers have been utilized in various sports that require multi-directional movements, such as soccer, field hockey, and lacrosse. An increased use of GPS technology in team sport settings over the last two decades has provided coaches, athletic trainers, and sport scientists with a better understanding of physical demands during training and competition based on real-time analysis of player performance (Cummins et al., 2013). GPS is a satellite-based navigation system that tracks player movement patterns and monitors external load metrics, including (but not limited to) total distance, running speed, peak velocity, and the number of accelerations and decelerations based on signals sent between the GPSs and various satellites (Cummins et al., 2013; Beato & Keijzer, 2019; Scott, Scott, & Kelly, 2016). Positional data, as well as running metrics, are gathered by the GPSs and can be used to construct performance profiles of athletes. These profiles allow for further analysis of speed and impact characteristics throughout training and game settings. Speed profiles can be built to understand band widths (how much an athlete runs at varying degrees including walk, light jog, etc.) an athlete is running in compared to their maximum speed, while impact characteristics give more details into the number of times the body experiences a tackle or collision. Coaches can quickly assess exertion levels and physical stress in real time based on the accumulation of external loading (i.e., high total distances, high number of sprints, etc.), or they can compare several training and competitions over a period of time to ensure athletes get adequate recovery mixed in with high training intensities.

2.2 Unit Differences and Validation

A variety of GPS units are in circulation, which have different sampling frequencies, most commonly 1-Hz, 5-Hz, 10-Hz, 15-Hz, or 18-Hz. Sampling frequency is one factor to consider when examining data output from GPS units, as the quality of collected data can be influenced. To effectively adapt training and match tactics based on the metrics provided by GPS units, it is vital that precise measurements are captured. In a brief review by Scott, Scott, & Kelly (2016) it was concluded that 1-15Hz GPS units are all able to track an athlete's distance with adequate intra-unit reliability; however, limitations exist for 1 and 5Hz GPS units when measuring high-intensity running, velocity changes, and short linear running. As the distance of locomotion (jogging, sprinting) decreases, the standard error of estimate of GPS estimations increases in both 1-Hz and 5-Hz. During a 40m, 20m, and 10m straight line sprint, the standard error for 1-Hz and 5-Hz increased from 12.2% to 32.4% and 11.9% to 30.9% respectively (Jennings et al., 2010). Standard error of the estimate was calculated as the standard deviation ($\pm 90\%$ CI) of the percent difference between the known distance (i.e., 40m measured out for a sprint) and the GPS recorded distance for each trial. Additionally, as speed increased from a walk to a jog to a sprint, these lower frequency GPS units showed decreasing accuracy in measurements. Conversely, both 10-Hz and 15-Hz GPS units had better results when measuring velocity changes from standing to acceleration or running with an increase in acceleration.

During straight-line running tasks, increased sampling rate (10 Hz) has shown increased reliability during higher constant and starting velocity, acceleration phase, and deceleration phase compared to 5-Hz units. For example, Varley et al. (2012) found the coefficient of variation for reliability decreased from 12.4% (± 1.18) with the 5 Hz unit to 5.3% (± 0.22) with the 10 Hz unit when maintaining a constant velocity from starting velocities of 1-3 m/s. To

validate the measured instantaneous velocity collected by GPS units, a Laveg laser was used as a comparison tool. Varley et al. (2012) were able to determine that the 10Hz GPS units exhibited higher Pearson correlation values to the Laveg laser during constant velocity, acceleration, and deceleration phases ($r^2 = 0.92$ - 0.96 , 0.92 - 0.98 , and 0.98 , respectively) than the 5Hz units ($r^2 = 0.28$ - 0.91 , 0.5 - 0.90 , and 0.83 , respectively). Although the 10 Hz units displayed high correlations to the laser measurements, the 10 Hz units still showed variable validity, with a coefficient of variation as high as 11.3% during the deceleration phase of sprints. Nevertheless, Varley et al. (2012) concluded that despite this variation, 10 Hz GPS units can accurately determine the number of accelerations and deceleration occurrences over the course of a match due to their ability to detect the smallest worthwhile changes in velocity.

The company STATSports is one of the leading GPS unit companies partnered with both professional and collegiate league sports. Their newer model, the Apex Series, had sampling frequencies at both 10 and 18Hz. The Apex 10Hz is capable of obtaining and tracking multiple satellite systems (i.e., GPS, GLONASS, and Galileo), which in turn provides more accurate positional information (Beato et al., 2018). Meanwhile, the Apex 18Hz unit can only access the GPS satellite system, so increasing the collection frequency is believed to aid in increased accuracy.

Previous studies examining external load measurement validity and reliability of the Apex 10-Hz unit have suggested that these units have high reliability when evaluating both peak velocity and speeds at short distances (<15m) and larger distances (>30m) (Beato & Keijzer, 2019; Crang et al., 2024). A study by Beato et al. (2018) had participants perform a 400-m dash, a 128.5-m circuit replicating differing movement demands (straight running, walking, jogging, zig-zag movements, etc.) of sports, and a 20-m sprint while wearing both GPS units. This study

found that the Apex 10Hz unit measured the average distance covered during the 400-m trial, 128.5m circuit, and 20-m trial to be 398.7m, 131.4m, and 20.07m, respectively. Similarly, the Apex 18Hz measured the average distance covered for the same maneuvers, resulting in 396.8m, 131.2m, and 20.19m, respectively. Furthermore, peak velocity during the 20-m sprint was measured to be 26.5 km/hr by both Apex 10Hz and 18Hz, which was compared to a radar gun that measured the peak velocity as 26.3km/hr. These results indicate that using the Apex 10Hz can provide valid measures for maximum velocity testing and evaluating distance and peak velocities during training, as its measurements are equal to those provided by the even higher sampling rate 18Hz unit.

Of particular interest, the Apex 10Hz units have been examined during intermittent running activities to better understand their capabilities to measure change of direction movements similar to those used in multi-directional sports. A team sports test commonly used for aerobic fitness testing is the Yo-Yo intermittent running activity, which requires participants to perform a 20m+20m course, with a 180° change of direction, and a 10-second active break after each 40m, with a speed increase at set intervals (Beato, Wren & Keijzer, 2024). All participants start from standing still and run through the first 20m before doing a 180° turn and running 20m back to the start. With the speed of the test increasing incrementally, players must accelerate and decelerate quickly at the turning point and end of the 40m. With the advanced technology provided by the Apex 10Hz units, accelerations (events with intensity $>3 \text{ m/s}^2$), decelerations (events with intensity $<3 \text{ m/s}^2$), and peak speeds were shown to have good and excellent interunit reliability (Beato, Wren & Keijzer, 2024). While GPS technology has come a long way since the first units, its capacity to provide valuable insights across multiple external load measures has proven to be a useful tool for coaches and sports scientists. The most recent

improvements in GPS units have had a particularly valuable impact on the capability to examine accelerations and decelerations, given their high frequency in multi-directional sports and mechanical demands within team sports.

2.3 Current Applications of GPS Based Data in Sport

GPS systems are valuable in providing information related to individual physical demands in sport, but it has also been useful in understanding differences among playing positions. In soccer, some of the most notable physical demand differences among playing positions are total distance covered, number of accelerations, number of decelerations, player load, and distances covered in various speed zones. When examining the total distance covered by the different positions in soccer matches, it was found that the average distance attackers cover is 7303m, whereas midfielders and defenders covered 8865m and 8362m respectively (Kuhlman et al., 2025). Additionally, this study found that midfielders have significantly more acceleration efforts ($> 3 \text{ m/s}^2$) and deceleration efforts ($> -3 \text{ m/s}^2$) than attackers. Another study conducted by Choice et al. (2023) found that in a soccer match, on average, midfielders cover a significantly larger total distance (10,509m) than forwards (8,320m) and defenders (9,109m). Positional differences were also captured in lacrosse matches as female collegiate lacrosse athletes who play in midfield traveled a greater distance on average at high-intensity speeds than defenders during matches (Devine et al., 2022). There were also findings that suggested on average, midfielders and defenders had significantly more acceleration efforts ($> 3 \text{ m/s}^2$) than attackers, and defenders had the most deceleration efforts ($> -3 \text{ m/s}^2$) (Kuhlman et al., 2025). Understanding key differences among playing positions has made the incorporation of GPS technology into team sports a critical tool. A coach's ability to optimize training protocols at the

individual and positional level and facilitate on-field performance can improve based on the external load metrics provided by GPS.

Furthermore, coaches and sports scientists can use key metrics such as movement patterns (positional changes), speed/velocity, distance travelled, accelerations, and decelerations, to quantify cumulative loading undergone by players (Prudholme et al., 2023). Based on the amount of loading on players, drills in training can be manipulated to ensure that the total player loading does not exceed the player's capabilities. To regulate training loads of individuals throughout a competitive season, a measure referred to as 'acute: chronic workload ratio (ACWR)' serves as a component to evaluate the relationship between injury risk and player loading provided by GPS. The ACWR is a calculation that indicates whether an individual player's acute loading is greater than, less than, or equal to the preceding chronic loading they have been exposed to (Bowen et al., 2017). To calculate ACWR, Bowen et al. (2017) took acute loading (1-week load) divided by chronic loading (4-week rolling average) – where loading was the combination of total distance covered at all speeds, high speed (above 20 km/h) distance covered, and number of accelerations during trainings and competitive matches. Interpretation of this metric is designed to inform coaches and sports scientists about potential increases in injury risks for players based on whether their acute total loading is greater than or vastly less than their chronic total loading. There is no standard ACWR to use for all players; therefore, it comes down to a range of ideal values that best fit the player or individual teams. It is important to keep in mind that players can either be undertrained or overtrained, which both increase the potential risk of injury.

Another measure that can be used to assess injury risks in combination with ACWR is known as 'relative risks' (RR), which is used to determine injury risks above and below given

loading accumulated in trainings. For example, one can measure the relative risks of injury from one week to the next, depending on whether the player's total loading increases and/or decreases. Bowen et al. (2017) utilized RR to discuss how total loading, including accelerations, high-speed running, and total distance, were significantly related to injury risks. When an RR was greater than 1, an increased risk of injury was reported (i.e., $RR = 1.50$ is indicative of a 50% increased risk). Across two seasons, Bowen et al. (2017) observed there were a total of 138 injuries (12.1 injuries/1000 hours 'on-legs' exposure time) – both contact and non-contact – in elite youth soccer players (mean age = 17.3, SD = 0.9). On average, the most common non-contact injuries per 1000 hours were at the ankle/foot (2.1), knee (1.1), and hip/groin (1.3), whereas the most common contact injuries were located at the ankle/foot (2.6), quadriceps (0.7), and knee (0.6). The two most common types of non-contact injuries included ligament sprains and muscle strains while the two most common contact injuries were hematoma/contusions and ligament sprains.

A few take aways from Bowen et al. (2017)'s study were: 1) the risks of contact injury was significantly increased when the acute:chronic total distance ratio was greater than 1.76, 2) a low chronic and high acute high-speed distance covered increased non-contact injury risks if the acute:chronic ratio was moderate-high (0.91-1.34), and 3) if the ratio between acute:chronic total loading was moderate-high (0.88-1.32) there was a significantly increased risk of non-contact injury. The most important finding from this study though was that accumulating >9245 accelerations in three weeks was the strongest of overall and non-contact injury risks, with an increased RR of 3.84 and 5.11, respectively. Although the authors did not explicitly state the exact computation for RR, the magnitude of injury risks associated with accumulated loading over 1-4 weeks should be closely monitored to ensure players are achieving a certain level of

training, not only to prepare players to withstand match demands but also to train them progressively and safely.

2.4 Deceleration Metrics and Change of Direction Movements

In recent years, there has been a rising interest in deceleration metrics quantified by GPS and the contribution decelerations have on player loading and injury risks. Deceleration is defined as the ability to slow whole-body momentum quickly and efficiently from one activity or movement to another (Wolfe et al., 2023). Accelerations and decelerations have distinct differences when comparing muscle contraction types and the specific intensity demands required to perform each motion. In an editorial by Harper and Kiely (2018), GPS data reveal that a higher frequency of accelerations occurs in low to moderate intensity ranges, whereas decelerations occur more frequently in high intensity ranges. They believe that this difference is due to players' gradually increasing running velocity when accelerating during match play, while decelerations require sudden decreases in velocity in a short time span. Thus, decelerations can be higher in magnitude compared to accelerations.

Deceleration demands require high eccentric muscle contractions, which amplify biomechanical loads on the lower extremities, leading to increased muscle tissue microtrauma (Harper & Kiely, 2018; Wolfe et al., 2023; Prudholme et al., 2023). Wolfe et al. (2023) referenced existing research on reports of indirect muscle damage biomarkers such as creatine kinase (CK), which is shown for up to three days following repeated bouts of sprinting and intense deceleration actions. CK is a widely used indicator of high physical exertion and muscle sarcomere damage (Varley et al., 2017). In a study on elite soccer players (27.4 ± 3.1 years), it was observed that CK levels were significantly higher than baseline values (pre-match) immediately following a match, 40-hrs post-match, and still 60 hours post-match (Varley et al.,

2017). This increase in CK was correlated to the total number of acceleration and decelerations during the matches.

Eccentric braking forces required to decelerate in sports can lead to high muscular tension, which in turn can potentially lead to excessive loading on the muscles, causing muscle strain or tearing. Given that decelerations require a high magnitude of eccentric contractions, examining the loading demands from these decelerations is necessary. It has been proposed that increasing the volume or intensity of deceleration tasks decreases the capability of the muscles to produce force to withstand these movements, as well as decreases coordinative proficiency (limb positioning) to properly perform deceleration maneuvers (Harper & Kiely, 2018). With decreased coordinative proficiency, it is assumed that a decrease in performance and an increase in injury risks will result.

Multi-directional sports, including soccer, handball, and rugby, require athletes to perform rapid accelerations and decelerations a considerable number of times within a training session and/or a competitive match. Given these frequent changes in movement cause differing amount of loading on players, researchers have examined how specific sprint metrics contribute to overall player loading. For example, Prudholme et al. (2023) performed a study with female collegiate soccer players over 19 training sessions to assess which GPS derived sprint metrics were correlated with player loading. Training sessions for this team included dynamic warmups, ball drills or technical skill work (i.e. juggling or passing boxes), and tactical drills (i.e. small or large-size competitive play). Based on the average sprint metrics across the sample, moderate to large positive correlations ($r = 0.512$ to 0.861) were observed between player load and sprint count, maximum top speed, sprint distance, sprint volume, number of decelerations at -1 m/s^2 to -3 m/s^2 , and number of accelerations at 1 m/s^2 to 3 m/s^2 , and 5 m/s^2 (Prudholme et al., 2023).

Stepwise regression analysis indicated that sprint distance predicted player loading the most (72.5% explained variance), and when combined with the number of decelerations at -1 m/s^2 , explained variance increased to 85.1%.

Although it has not received a lot of attention, accelerations and decelerations play a significant role in predicting total player loading (Prudholme et al., 2023). Accelerations require an increase in momentum, while decelerations require a decrease in momentum and involve braking mechanisms that produce higher stress loads on soft tissues. The accumulation of both low- and high-intensity actions would therefore influence player total loading throughout matches and/or training. Even at relatively lower intensities (1 m/s^2), the significant velocity changes throughout match play increases player total loading due to the metabolic demands of the tasks, resulting in muscle damage and soreness (Prudholme et al., 2023).

When measuring deceleration efforts in collegiate female sports, female lacrosse players can average >50 high-intensity accelerations ($> 3 \text{ m/s}^2$) and high-intensity decelerations ($< -3 \text{ m/s}^2$) each during a 60-minute competitive match (Fields et al., 2023). Conversely, it has also been found that women's lacrosse players only accelerate and decelerate an average of 38 times per competitive match (Devine et al. 2022). Different playing tactics and team formations can have an influence on the amount of deceleration efforts within lacrosse, which can explain the range of acceleration and deceleration efforts performed. In women's soccer, the number of acceleration and deceleration efforts increases as competitions are 90 minutes compared to 60 minutes, and soccer consists of sprint bouts that occur approximately every 90 seconds (Stolen et al., 2005). At the collegiate level, it has been found that soccer players can accumulate greater than 100 low intensity ($< 3 \text{ m/s}^2$ and $> -3 \text{ m/s}^2$) and high intensity ($> 3 \text{ m/s}^2$ and $< -3 \text{ m/s}^2$) accelerations and decelerations per half in a competitive match (Trapp et al., 2021). Similar

deceleration efforts are seen at the elite/professional level, where professional Spanish club players accumulated an average of 180 moderate intensity ($< -2 \text{ m/s}^2$) and 30 high intensity decelerations ($< -3 \text{ m/s}^2$) in competitive matches (Casamichana et al., 2024). In soccer specifically, it has been observed that second-half performance in accelerations and decelerations decreases from first-half performance in collegiate soccer players (Trapp et al., 2021; Russel et al., 2016). Between-half differences have been attributed to muscular fatigue induced by repetitive acceleration and deceleration efforts throughout a competitive match. In an editorial by Harper & Kiely (2018), the idea that increasing volume or intensity of deceleration activities contributes to a continuous cycle of increasing fatigue, diminished coordination proficiency, and accumulation of tissue damage was emphasized. Since decelerations have been correlated with high mechanical loading (Prudholme et al., 2023) performing hundreds of deceleration efforts in a single competitive match (Trapp et al., 2021) may negatively influence performance and alter lower extremity coordination.

The ability to decelerate is essential to athletic performance for the reason that it requires an athlete to slow down their body relative to their ability to accelerate, which is typically observed in change of direction (COD) tasks. COD tasks are inevitable in multi-directional sports; therefore, how well an athlete can control their movement aids in the prevention of lower extremity injury (Wolfe et al., 2023). Previously, studies testing deceleration during COD tasks have been done in laboratory settings with the use of force plates. Thus, the ability to translate findings of deceleration metrics and lower extremity biomechanics from laboratory settings to real-world sports applications is limited.

Two field-based tests to assess deceleration abilities are the 505 COD test and the Acceleration-Deceleration-Assessment (ADA). The 505 COD test is performed by the individual

sprinting from a start line to a marker placed 5-20m away (depending on how much initial acceleration is wanted by the tester) doing a 180° turn at the marker, followed by a 5m sprint back towards the starting line (Beato et al., 2025). The benefits of the 505 COD test are assessing deceleration from a predetermined distance as well as understanding the player's proficiency to turn at sharp angles and reaccelerate appropriately. In addition to the ability to decelerate for a COD movement, it is equally important to understand horizontal deceleration abilities, as players have different strategies for slowing and stopping their body's momentum (Beato et al., 2025). The ADA test is utilized to assess horizontal deceleration abilities, specifically how quickly and effectively an individual can go from maximal acceleration to maximal deceleration (Beato et al., 2025; Harper, Jordan, & Kiely, 2021). To perform the ADA, the individual is required to sprint maximally 20m, followed by a maximal deceleration to come to a complete stop. It has been suggested that the ADA can be used as an athlete's absolute or maximal deceleration capacity under controlled circumstances, and the 505 COD test exemplifies decelerations under sport-specific maneuvers since it includes a sharp COD (Philipp et al., 2023). The use of these tests together is important since research suggests that sprint velocity may not be correlated to COD velocity. Freitas et al. (2023) found that players with faster linear sprint times were not efficient at COD, meaning they were either not able to effectively decelerate quickly, followed by an abrupt reacceleration. Comparably, Philipp et al. (2023) suggest that individuals with a high deceleration potential (seen in the ADA) may fail to achieve similar performance levels within the 505 COD test. Therefore, it is important for coaches and supporting staff to understand deceleration abilities as a whole and obtain accurate information for each player to adjust training as needed.

Performance metrics provided by GPS units have become a valuable tool in structuring training in preparation for competitive matches. Despite technological advances for GPS units, to the author's knowledge, no studies have reported the translation of GPS technology metrics to lower extremity biomechanical factors that occur during decelerations or COD tasks. Elements including acceleration, deceleration, COD angle, braking steps, and muscle physiology all play a role in what is happening mechanically in the lower extremity joints during deceleration tasks (McBurnie et al., 2022). GPS units are a viable way to characterize total loading on the whole body, but it is essential for researchers to understand what biomechanical postures are occurring throughout the lower extremities while players perform deceleration movements. In addition to the quantifiable deceleration metrics provided by GPS, understanding the mechanical postures and joint loading in the lower extremity may provide deeper insight into deceleration abilities, deceleration efficiency (controlling the braking action), and risks of injury during these multiplane movements.

2.5 Lower Extremity Mechanics

2.5.1 Straight-Line Deceleration Demands

Rapid horizontal decelerations are essential in team sports as they require slowing down whole-body momentum to prepare for tasks such as cutting maneuvers or scoring a goal. The ability to decelerate efficiently permits athletes to reduce their velocity in shorter distances and at quicker rates; however, decelerations have unique characteristics in the lower extremity joints. Unlike accelerations, decelerations require the trunk's center of mass (COM) to be more erect and have a posterior lean, which moves the body's COM posterior to the base of support (Hewitt et al., 2011; McBurnie et al., 2022). Upon initial ground contact, it is important for immediate hip and knee flexion and ankle dorsiflexion to occur as this is the way impact forces can be

dissipated by the lower extremities with each foot-ground contact– reducing injury risks (Harper et al., 2024; Hewit et al., 2011). During the first 30% of ground contact for the initial steps to decelerate, peak posterior ground reaction forces (GRF) are approximately four times the external forces experienced during other running tasks (Verheul, Harper, & Robinson, 2024). This indicates that external force demands during rapid decelerations have a large impact on loads experienced on the lower extremity.

During the first to second braking steps of horizontal deceleration, ankle plantarflexion occurs during the first 20% of ground contact (Mean peak = -15° - -20°), followed by gradual dorsiflexion (Mean peak = 4° - 10°) before the next plantarflexion leading into the following step. The knee goes from slight flexion ($M=12.8^{\circ}$ - 25°) at initial ground contact into large flexion ($M = 75^{\circ}$ - 81°) through ground contact to attenuate forces (Fitzwilliam et al., 2024; Verheul, Harper, & Robinson, 2024). Lastly, the hip also flexes throughout ground contact with peak angles ranging from 25° to 44° as momentum begins to slow (Verheul, Harper, & Robinson, 2024). These movement patterns provide the foundation for kinetic demands as the hip extensors, knee extensors, and ankle plantar flexors are all increasing their moments about their respective joints. Additionally, these muscles are doing negative mechanical work to decelerate the forward moving body which enables a more controlled deceleration.

During maximal deceleration, it has been found that across the lower extremity joints, the hip and the knee produce more negative joint work than during accelerations, with the ankle joint having comparably less negative work. Additionally, decelerations produce greater peak joint moments compared to accelerations during knee flexion, ankle dorsiflexion, hip extension, and knee extension (Fitzwilliam et al., 2024). These findings are important as negative joint work indicates energy absorption through eccentric contraction of muscles, such as the gluteus

muscles, hamstrings, quadriceps, tibialis anterior (Fitzwilliam et al., 2024; Verheul, Harper, & Robinson, 2024), and calf muscles. In contrast to these findings, work by Williams III, Cole, & Powell (2017) suggests that the knee and ankle contribute to more negative work followed by the hip joint during the deceleration phase of a straight-line run. Though it is agreed that the knee contributes the most to negative work during decelerations compared to accelerations, contradictions between the relative amount of negative work produced at the hip and ankle could be due to differences in protocols. Williams III, Cole, & Powell (2017) did not have participants decelerating from a maximal speed; rather, they had participants go from a self-selected comfortable pace to deceleration. Despite these differences, a key takeaway from the lower extremities during rapid changes in locomotion is that each joint has a distinct role in generating or absorbing forces to ensure whole-body control. Importantly, rapid decelerations precede COD tasks, which means the lower extremity joints are required to absorb braking forces, but are also required to switch into generating forces during directional transitions.

2.5.2 Change of Direction Deceleration Demands

Not only do straight-line horizontal decelerations cause the lower extremity joints to undergo excessive loading, but decelerations preceding COD tasks also place the hip, knee, and ankle under excessive stress. In sports situations, a quick reaction time as well as precise execution of COD tasks are important performance factors that help a player outperform the opposing player. In competitive match situations, time constraints to change direction alter lower extremity joint biomechanics. Consequently, altered lower-extremity biomechanics may increase the risk of injury (Fuerst, Gollhofer, & Gehring, 2017). The lower extremities act as a kinetic chain linking the trunk to the hip, through which associations to tri-planar knee loading (Frank et al., 2013) and ankle loading during COD are present. After a rapid deceleration in sports, a COD

typically follows, and the body is projected into a new direction. The kinetic chain of this movement trickles down the joints, starting with trunk movement, which has been observed to modulate frontal and transverse plane knee moments (associated with ACL injury) based on its link to the hip (Frank et al., 2013). During cutting maneuvers at 45°, 60° and 90°, frontal plane hip adduction and internal rotation have been shown to be a significant predictor of knee abduction (valgus) motion (Frank et al., 2013; Imwalle et al., 2009), indicating a higher risk of ACL loading.

For instance, a study performed by Frank et al. (2013) had participants perform a 60° cut away from the dominant leg after a double leg hop over a 17-cm wall placed 25% body height away from the participant. A key finding was that the combination of trunk flexion and rotation, as one's center of mass shifts from the initial direction of movement (landing from hop) towards the new direction of movement (cutting), loads the knee in the transverse and frontal plane. More specifically, Frank et al. (2013) found that less trunk rotation towards the new direction of motion, combined with greater hip adduction moment, explained the vast majority of variance in knee internal varus moment during cutting. Additionally, it was observed that compensatory knee external rotation moment increased with greater trunk flexion displacement paired with hip internal rotation moment. These results from Frank et al. (2013) indicate that trunk control during side-stepping maneuvers can impact knee movements known to increase ACL loading; therefore, training for better trunk control may be one key to reducing excessive loading on the knee – a concept which can also be applied to deceleration tasks.

While the trunk and hip have a proximal role in lower extremity mechanics, the knee has been a central focus as sudden COD movements makes the knee susceptible to ACL tears. During quick, unanticipated cuts, individuals do not consciously consider the proper postural

mechanics occurring at each joint (Besier et al., 2001). Compared to 45° and 90° cuts, a 180° COD shows greater preliminary deceleration due to dual support of both legs in the braking mechanics, which leads to distinct loading patterns in the knee (Dos'Santos, Thomas, & Jones, 2021). During a 180° COD, ground contact time during the cut increases while approach speed, final foot contact, and exit velocity (into new direction) decrease in relation to 90° and 45° cuts. Sagittal plane kinematic demands during 180° COD demonstrated that the peak knee flexion increased from 55.8° to 60° when compared to a 45° cut. At initial contact of the cutting movement, there was decreased knee flexion (27.4 to 20.6°) and ankle dorsiflexion angles (57.5 to 48.5°) between a 45° cut and the 180° COD (Dos'Santos, Thomas, & Jones, 2021). The differences in biomechanical demands of a 180° COD cause the knee to undergo higher loading during these tasks in comparison to less severe cuts. Further, Dos'Santos, Thomas, & Jones (2021), referenced previous literature which observed greater knee valgus and extended knee during sharp CODs. The combination of knee postures required by sharp COD magnify ACL strain as multi-plane loading occurs. Additionally, there is a greater angle-velocity tradeoff as increased angles are used for cutting in relation to knee abduction moments, knee abduction angle, and internal rotation moment (Dos'Santos, Thomas, & Jones, 2021).

While the knee has received much attention due to its susceptibility to ACL injury, the ankle also plays a critical role in decelerations during COD tasks by attenuating ground reaction forces and influencing stability during sharp directional changes. The middle phase of a deceleration is marked by high ankle plantarflexion as eccentric force requirements are utilized for slowing down (Jasper Verheul, 2025, blog commentary). For larger cutting angles, such as 90°, ankle plantarflexion angles increase 8.4° compared to a 45° cut (Havens & Sigward, 2015). The increase in ankle plantarflexion angle enhances the ability to attenuate ground reaction

forces to reduce lower extremity joint loading. Additionally, Fuerst, Gollhofer, & Gehring (2017) found significant influences on maximum ankle inversion velocity during a 180° COD task. When an individual has limited time to prepare for a quick deceleration/large COD, the average maximum ankle inversion velocity increased from 508°/s to 587°/s. Fuerst, Gollhofer, & Gehring (2017) concluded that having less time to prepare for a 180° turn can cause a high risk of ankle injury (i.e., ligament sprains) due to the high value of ankle inversion velocities.

2.5.3 Repetitive Loading Influence on Lower Extremity Mechanics

The accumulation of deceleration actions throughout multi-directional sport play may potentially lead to increased physical exertion levels in individuals. With increased physical exertion, research has looked into the lower extremity mechanical changes for increased risk of injury. Currently, research has used protocols to elicit physical exertion based on isolated muscle fatigue (measured with EMG), pre- and post-exercise regimens (i.e., drop jumps/vertical jump, step-up/step-down, squats, etc.), and athletic-specific tasks (i.e., 5-10-5 agility test, Yo-Yo intermittent recovery test) (Cortes et al., 2013; Kernozek, Torry, & Iwasaki, 2008; Taylor, 2024; Zago et al., 2021).

Both laboratory and sport-specific tasks to induce physical exertion have consistently shown alterations in lower extremity mechanics in ways that may heighten injury risks. Cortes et al. (2013) showed that short-term fatigue through a functional agility protocol elicited alterations in knee flexion angle and hip abduction angle at initial contact during a side-stepping maneuver (at 45°). Knee flexion and hip abduction angles decreased 3° and 5.0°, respectively, at initial ground contact from prefatigued to post-fatigue levels. At peak stance, internal hip adduction and knee adduction moments decreased 0.33 Nm/kgm and 0.12 Nm/kgm, respectively, from pre-fatigue to 100% fatigue. Interestingly, hip flexion at initial contact and peak stance, knee flexion

at peak stance, knee abduction at initial contact, and knee adduction moment at peak stance all increased from prefatigued to 50% fatigue, followed by a decrease in 50% fatigue to 100% fatigue. Specifically, hip flexion at initial contact and peak stance changed from 45.4° to 46.2° to 40.9° and 49.8° to 52.9° to 46.3°, respectively, while knee flexion at peak stance altered from 52.9° to 56.1° to 50.5°.

Similarly, Kernozek, Torry, & Iwasaki (2008) found neuromuscular fatigue in lower extremity kinematics and kinetics during single-leg drop landings. Based on repeated sets of squatting 60% of the participants' 1-rep max, significantly altered landing mechanics by increasing hip flexion and ankle dorsiflexion, as a means to potentially minimize knee loading during ground contact. Recreationally active females increased hip flexion by 7.3°, and increased ankle dorsiflexion by 2.41° from pre-fatigue to post-fatigue protocol. Though these kinematic differences presented by Cortes et al. (2013) and Kernozek, Torry, & Iwasaki (2008) may be 'small', these studies highlighted that there are alterations in lower extremity kinematics based on fatigue inducing protocols. Even minor deviations in lower extremity mechanics during dynamic tasks has the potential to change loading behaviors across the joints.

Moving into application in a more sports context, Zago et al. (2021) observed that repeated CODs in 20 professional/semi-professional Italian division female soccer players displayed decrements in lower extremity kinematics as the number of CODs increased. This study had participants perform continuous 5m shuttle runs with 180° CODs. The most notable changes occurring in the majority of players were a reduction in hip (70% of players) and knee (60% of players) flexion angles of 10-25° and 5-20°, respectively, during initial foot contact for the COD. Additionally, hip abduction angle decreased in 8 players by 5-10° while it did not change for the rest of the players. Lastly, pelvis rotation to the new running direction decreased

in 25% of players. This study found no significant frontal plane motion changes at the knee and ankle and no significant transverse plane motion at the hip, knee, or ankle in 70 and 75% of players, respectively, as the number of CODs increased. Though the number of CODs varied between players (15 – 68), 75% of players reported a rate of perceived exertion ≥ 6 , indicating that these players perceived their exercise intensity on the COD test to be “hard” to “maximum” exertion Zago et al., (2021). The findings of this study were important for sports scientists to reinforce individual biomechanical changes after accumulating training load, as individuals adopt varying movement strategies to cope with repeated CODs.

Building on these prior works, a master’s thesis by Taylor (2024) examined the influence of neuromuscular fatigue on hip and knee mechanics across movements in differing sport assessments, such as squatting, drop jump, and cutting (at 90°). The fatigue protocol Taylor (2024) used included having participants perform a circuit of a three second isometric wall sit where knees were at a 90° angle, five submaximal vertical jumps, and 30 seconds of high knees at a cadence of 150 beats per minute (based off an online metronome). Participants in this study repeated this circuit until they showed a 10% decrease in muscle activation (determined with EMG) in one of the quadriceps muscles during the wall sit from their baseline values (muscular activity during first wall sit). Once the neuromuscular fatigue threshold was met, during the wall sit, the participants finished that circuit and immediately went into their post-assessment of squatting, drop jump, and cutting (at 90°). This fatigue protocol produced simultaneous alterations in all movements; for example, on hip flexion angle, hip internal rotation angle, and knee abduction angle, where decreases were seen from 28.2° to 19.1°, 3.3° to 2.3°, and -8.0° to 6.0°, respectively, 200ms after ground contact of cutting. Together, these findings indicate consistency in fatigue, or physical exertion resulting in biomechanical changes throughout the

lower extremity. Given the consistent yet varied results in laboratory-controlled and sport-like settings, further investigation is needed to clarify the extent to which high physical exertion during repeated rapid decelerations (accumulated loading) influences lower extremity joint kinematics. This rationale is the basis for the third aim of the present study.

Overall, deceleration actions are an area of growing interest due to both the physiological and biomechanical demands placed on individuals. GPS units have allowed for the tracking of deceleration metrics based on frequency and intensity; however, limited research has been done on their ability to adequately classify deceleration intensities. Additionally, GPS units have not been used as a tool to assess biomechanical changes associated with the accumulation of deceleration actions. Motion capture has provided insight into the biomechanical demands of deceleration actions, mostly in change of direction task rather than horizontal deceleration task. Both change of direction and rapid horizontal deceleration actions are critical to understand as individuals use a variety of movement techniques to slow down whole-body momentum. Furthermore, understanding the effects of accumulated high-intensity decelerations will provide information into strategies used to cope with deceleration demands, which can potentially lead to increased risk of lower extremity injury. The aim of the present study is threefold: (1) Determine the sensitivity and specificity of GPS unit detected decelerations and the classification into deceleration thresholds using motion capture as the reference, (2) examine if accumulated high-intensity decelerations influence horizontal deceleration ability, and (3) examine whether accumulated decelerations influence lower extremity kinematics during standardized deceleration tasks. By addressing these aims, this study seeks to improve the interpretation of deceleration metrics for athlete monitoring and provide insight into the biomechanical demands that may contribute to performance decrements and injury risk. The first hypothesis of this study

is GPS units will demonstrate high sensitivity and specificity in detecting deceleration actions and correctly classifying them into deceleration intensity thresholds relative to motion capture. The second hypothesis is that after accumulation of high intensity decelerations, there will be a change in the ability to decelerate, including change in pelvis position and number of steps to decelerate during the ADA task. The third hypothesis is that after the accumulation of high deceleration efforts, there will be a change in lower extremity mechanics utilized to perform deceleration tasks.

CHAPTER 3: Methods

3.1 Research Design

A mixed study approach was utilized in this study where 1) concurrent validity design was used to assess the accuracy of GPS-derived deceleration classifications relative to motion capture (reference standard). Motion-capture-derived deceleration events were categorized into the same deceleration-intensity thresholds, allowing sensitivity and specificity analyses to evaluate the GPS unit's ability to correctly classify deceleration events, 2) within subject experimental design to examine if accumulated decelerations influenced deceleration ability during sport-related-task, and 3) within subject experimental design to assess if lower extremity kinematics were influenced by accumulated decelerations.

3.2 Participants

For this study, the participant group was recreationally active college-aged females (mean age $22\text{yrs}\pm 2\text{yrs}$, height: $1.68\text{m}\pm 0.5\text{m}$, mass: $69\text{kg}\pm 10\text{kg}$). A summary of participant demographics can be found in Table 1. To determine an acceptable level of physical activity, participants needed to score at least 8 out of 16 possible points on the Marx Activity Rating Scale (MARS) assessment. A score of 8 indicated that the individual had participated in high-demand physical activities in the last year, however they may not do so at the highest intensity. This score also reflects moderate participation in sporting and/or exercise routines. This set of participants had qualitatively differing skill levels to provide a wider range of deceleration abilities for data analysis. With a broad range of deceleration abilities, we also chose to use a modified MARS assessment to quantify how often participants performed the same movements within the last month as this gave insight into activity behaviors closer to the time of data collection. Only the traditional MARS scale was used to determine participant eligibility. All

participants were free of musculoskeletal injuries at the time of data collection and self reported healthy. All participants signed an informed consent to participate in study (Appendix B).

Table 1: Participant Demographics, including age, weight, height, mass, MARS scores for the last year and the last month.

Subject ID	Age	Height (m)	Mass (kg)	MARS Year Score	MARS Month Score
001	24	1.60	58.59	10	11
002	21	1.63	64.22	10	11
003	22	1.63	61.23	8	8
004	23	1.65	60.40	16	15
005	23	1.75	78.92	9	8
006	25	1.78	79.37	12	12
007	22	1.65	63.50	16	16
008	18	1.68	67.13	16	9
009	24	1.69	70.31	12	7
010	18	1.68	50.80	12	11
011	20	1.70	90.72	10	12
012	21	1.68	74.84	10	10
013	21	1.63	63.50	8	2
014	19	1.71	79.38	12	6
015	23	1.68	71.67	10	7

3.3 Materials and Instruments

3.3.1 Marx Activity Rating Scale

The Marx Activity Rating Scale (MARS) is a self-reported measure used to ask patients about their physical activity levels over the last 12 months related to 4 specific movements that are common in a variety of sports (Appendix C). The four activities include:

- Running: running while playing a sport or jogging
- Deceleration: coming to a quick stop while running
- Cutting: changing directions while running
- Pivoting: turning your body with your foot planted while playing a sport; For example, skiing, skating, kicking, throwing, or hitting a ball.

Item generation was performed by interviewing a combination of 35 orthopedic surgeons who specialize in sports medicine, physical therapists, and athletic trainers who specialize in sports medicine and athletic patients with disorders of the knee (Marx et al., 2001). These items assess a patient's activity level without the influence of seasonal changes (i.e., only playing winter sports), recent injury, illness, or surgery. Rather, the MARS focuses on patients' peak (or highest) activity level in the past 12 months to obtain more accurate estimates of their baseline activity when they are participating in sport (Marx et al., 2001). Activities are rated on a 5-point scale of frequency (participation less than 1 time a month = 0 and participation 4 or more times a week = 4) and scores are summed to a maximum of 16 points. For the current study, participants needed to score a minimum of 8 points to participate, which will be indicative of "recreationally active".

3.3.2 CR-10 Rating of Perceived Effort (RPE) Scale

The CR-10 RPE scale is a measure used for participants/athletes to report their perceived effort/intensity during an exercise session. This scale has been shown to provide a valid subjective estimate of training load during non-steady state exercise in relation to heart rate responses (Foster et al., 2001). This 10-point scale ranges from 0 = rest to 10 = maximal effort.

Global Positioning System Units

Commercially available GPS units (Apex, STATSports, Northern Ireland, Newry) were worn in a purpose designed vest (STATSport) with a pocket positioned in the center area of the upper back, slightly superior to the shoulder blades. The GPS units measure at a 10Hz frequency. GPS units mark a deceleration based on an individual decreasing velocity for a minimum of 0.5 seconds. Once a deceleration action has been detected, it is placed in a 'deceleration intensity zone', which ranges from Zone 1 to Zone 6 (Table 2). GPS data was downloaded in related software (Sonra v4.2.1). GPS units were turned on outdoors 20 minutes prior to the session start

time as to allow for a connection between units and multiple satellites. There have been no previous reports about discomfort of GPS units placed inside of vests as these units are utilized by numerous collegiate and professional sports teams.

Table 2: GPS Zones and Associated Absolute Deceleration Intensities

<i>Variable</i>	<i>Zone 1</i>	<i>Zone 2</i>	<i>Zone 3</i>	<i>Zone 4</i>	<i>Zone 5</i>	<i>Zone 6</i>
<i>Deceleration intensity</i>	<i>0.5 – 1.0 m/s²</i>	<i>1.0 – 2.0 m/s²</i>	<i>2.0 – 3.0 m/s²</i>	<i>3.0 – 4.0 m/s²</i>	<i>4.0 – 5.0 m/s²</i>	<i>5.0 – 10.0 m/s²</i>

3.3.4 Markerless Motion Capture

Markerless motion capture software (Theia Markerless Inc., Kingston, ON, Canada) was used for this study. This software uses synchronized video data and a deep learning tool to estimate three-dimensional (3D) human poses. A 16-segment model is scaled to fit the subject-specific landmarks positions in 3D space, and an optimization approach using inverse kinematics is used to measure spatiotemporal gait parameters (Kanko et al., 2021). This software allows for field-based assessments as the camera setup is portable, and skeletal models do not require the use of reflective markers to be placed on participants. Video data was captured using 13 Qualisys Motion Capture Miquis Video Plus (120Hz with 1080p resolution) synchronized cameras that were uploaded to Theia3D software. Cameras were placed in a horseshoe shape distributed equally around the capture volume and be placed approximately 1.8 meters above the ground and four meters apart. All data collected were performed indoors on a basketball court.

3.3.5 Visual3D Modeling

3D skeletal pose estimates from Theia3D software were exported as .c3d files and imported into Visual3D, where musculoskeletal models were constructed and integrated to generate lower limb joint angles during the deceleration tasks. The global coordinate system is established during camera calibration, which is performed before motion measurements.

Theia3D provides a raw and a filtered version of tracked movements. The filtered version of movement trails uses a software built-in filter to smooth the tracking and reduce any noise by applying a Generalized Cross-Validation Spline (GCVSPL) cutoff frequency of 20Hz (equivalent to a double Butterworth filter with additional advantages). Using the filtered version provided is recommended as it allows the model joint constraints to be maintained as the unfiltered skeletal pose data can potentially cause unnatural segment orientation or out of plane movement due to high-frequency noise or “estimation jitter”. A 4 x 4 pose matrix is provided by Theia3D software for kinematic modeling to be viewed in Visual3D.

Once motion files were opened in Visual3D, an automatic skeletal model was assigned to the motion file based on the segment lengths and other scaling factors that are subject-specific from the Theia3D model. Mass and height, however, needed to be adjusted per participant as Theia3D assigns a default value to the model. Seventeen rigid segments to track participants are utilized by Theia to track the movement of the body. Segments include head, thorax, left and right upper arm, lower arm, and hand, pelvis, right and left thigh, shank, foot, and toes. Joints made from these segments are modeled to have between two to six degrees of freedom (DOF) in the default (upper and lower body) model. For example, the pelvis and shoulder are “free joints” and given 6 DOF, while the thigh connecting to the pelvis to make the hip and the knee joint are given 3 DOF. Additionally, there are no built-in angle offsets used to normalize joint angle calculations, rather joint angles are calculated as the full, absolute angle between adjacent segments.

The current study analyzed kinematic data on trunk segment COM position, pelvis segment position, as well as twelve joint angles: three for the trunk segment: sagittal flexion/extension, frontal lateral-flexion, transverse rotation; three for the hip joint: sagittal

flexion/extension, transverse internal rotation, frontal abduction/adduction; three for the knee: sagittal flexion/extension, transverse internal/external rotation, frontal abduction(varus)/adduction(valgus); and three for the ankle: sagittal dorsiflexion/plantarflexion, frontal eversion/inversion and transverse internal/external rotation. Measurements were performed on both lower extremity limbs as well as on the trunk. To calculate joint angles throughout each motion trial, a pipeline was created in Visual3D utilizing the Compute Model-Based Data command and selecting the segments articulating to form the joints (i.e., the thigh and shank to determine knee angle). By utilizing the Cardan sequence XYZ, joint coordinate systems were defined relative to the global coordinate system. Additionally, deceleration metrics were obtained based on center of mass (trunk) position changes during each task to replicate how the GPS units track trunk movement. To calculate trunk acceleration in Visual3D, the first derivative of trunk position was calculated to obtain trunk velocity, and the average deceleration was calculated based on the change in velocity from peak to min (or zero velocity) over time in the anterior/posterior (horizontal) direction. Deceleration values were then placed in the same ‘absolute intensity threshold zone’s that are derived from GPS. Kinematics were examined during initial contact of all consecutive deceleration steps for the ADA, during initial contact of the cutting limb and full weight acceptance on the cutting limb during the change of direction for the 505 tests. Initial contact for both ADA and 505 COD task was determined as the frame to which the participant’s foot was in contact with the ground. Accelerations were examined for the deceleration phase of the ADA and 505 COD tests.

3.4 Procedures

Participants were walked through a 10-minute dynamic warm up consisting of various movements (lunges, high knees, hip openers, grapevine shuffle, etc.) and four brief 5m sprints

into quick decelerations to ensure that participants were adequately prepared to go through the testing procedures. For testing, all participants will complete a static calibration trial followed by 3 trials each of a double leg squat, self-selected pace walk, ADA, 505 COD. Participants performed the set of squats, walks, ADA, and 505 COD tasks during the pre-test phase. After completing the squatting and walking trials, participants were walked through the high intensity deceleration tasks to be performed and allowed one practice attempt to get comfortable with the tasks before data collection. Once the ADA and 505 COD tasks were completed, participants entered the exercise intervention: the 5-10-5 pro-agility test with an intermittent ‘active recovery’ of a 30 second jog. This test requires participants to perform a 5-meter sprint into a 180° COD, sprint 10-meters into another 180° COD, and finally sprint 5-meters to the finish line (Figure 1). Participants were prescribed to complete the 5-10-5 pro-agility test 15 times to accumulate 30 total high deceleration COD efforts, similar to the amount that are accumulated in elite/professional soccer matches (Casamichana et al., 2024). With each repetition of the 5-10-5 pro agility test, the participant would swap starting direction (i.e. start test to the right on first repetition and start test to the left on second repetition). After completion of the exercise circuit, participants immediately performed only the ADA and 505 COD assessments in a counterbalanced order. Exercise intensity/participant effort was collected for the participant immediately after each repetition of the exercise intervention by participants providing their RPE score in addition to each trial of the 5-10-5 pro-agility assessment being timed.

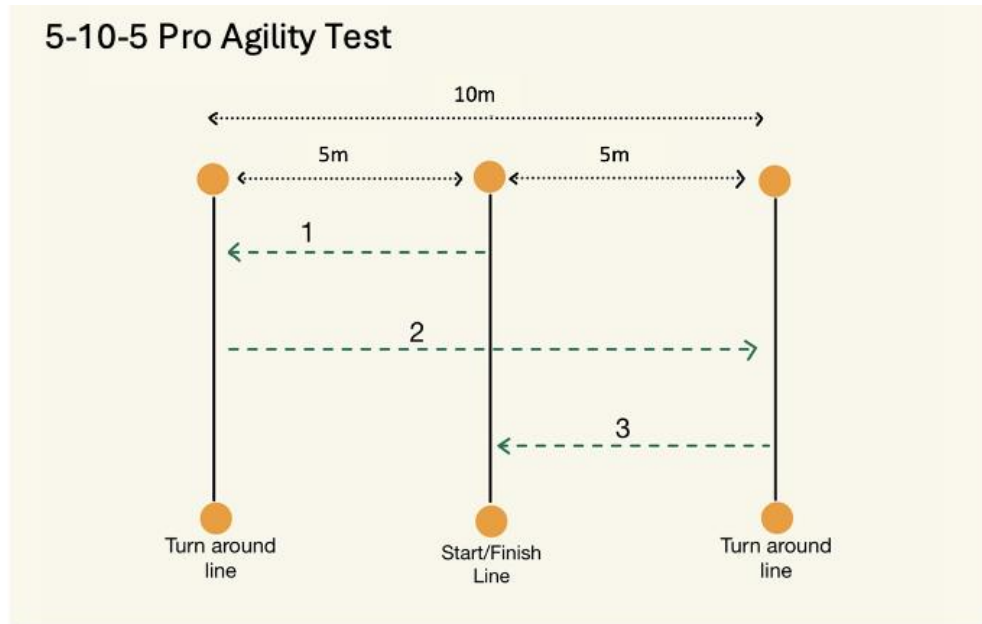


Figure 1: 5-10-5 Pro Agility Test Set Up

3.4.1 Acceleration-Deceleration Assessment (ADA)

Procedures for the ADA included having participants perform a 20m sprint through velocity gates followed by a rapid deceleration to a full stop within a 10m space. The starting position for the participants was to stand at the start line (marked at 0m) in a staggered stance position (preferred front leg is determined by participant). With a given auditory signal, the participants were instructed to sprint as fast as possible through the timing gates, and to decelerate as fast and controlled as possible once they hear the auditory signal of the timing gates. Participants are asked not to slow down before reaching the 10m mark and to aim to decelerate only once they have reached the 20m mark (Figure 2). If participants visually slowed down before reaching the 20m mark during the pre-testing phase, a 2-min rest period was provided, and the trial was repeated. This test was chosen based on its relevance and appropriateness to a sports setting, where sports scientists are typically given limited time to conduct data collection as to maintain continuity of the practice session. Horizontal deceleration was collected for the entire deceleration phase of the ADA.

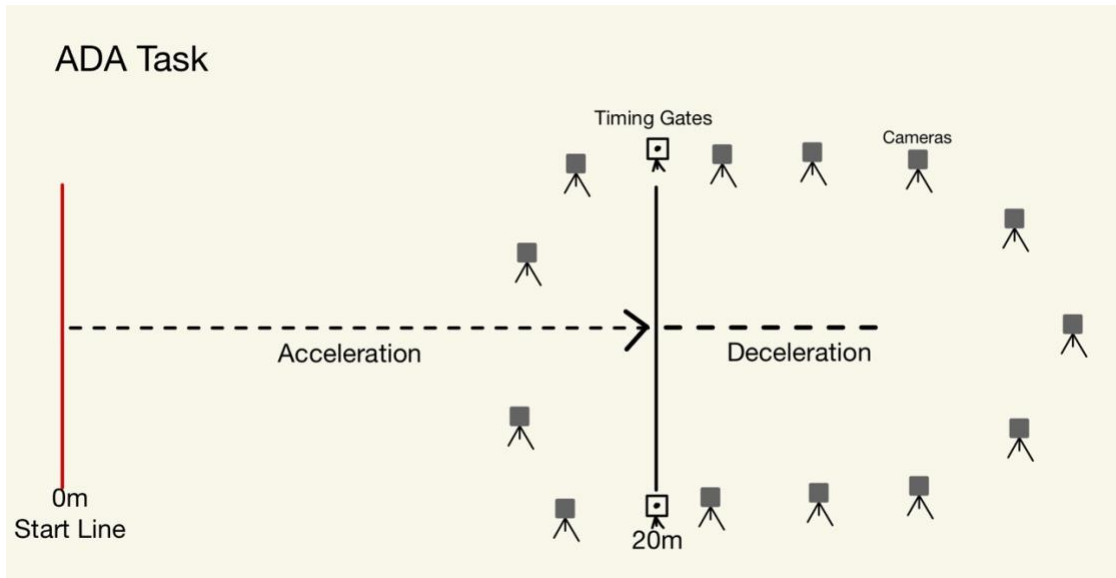


Figure 2: Acceleration - Deceleration Assessment (ADA) Set up with example of camera layout

3.4.2 505 Change of Direction Test

The 505 COD test is based on the methodology used in research by Philip et al. (2023). This test required athletes to stand at the start line (marked at 0m) in a staggered stance position. Prior to initiating the sprint, participants were instructed to sprint through velocity gates placed at a 10m mark as fast as they can, perform a 180° turn at another set of cones placed at a 15m mark, then rapidly reaccelerate back through the 10m mark (Figure 3). During each trial, the participants were able to change direction with their preferred leg but were asked to repeat the second and third trial using the same leg to change direction. Participants were asked not to slow down before reaching the 10m mark and to aim to decelerate only as they approach the 15m mark (giving 5m of deceleration space). If participants visually slowed down before reaching the 10m mark, a 2-min rest period was provided, and the trail was repeated. Horizontal velocity was recorded in the same manner as highlighted in the ADA assessment section.

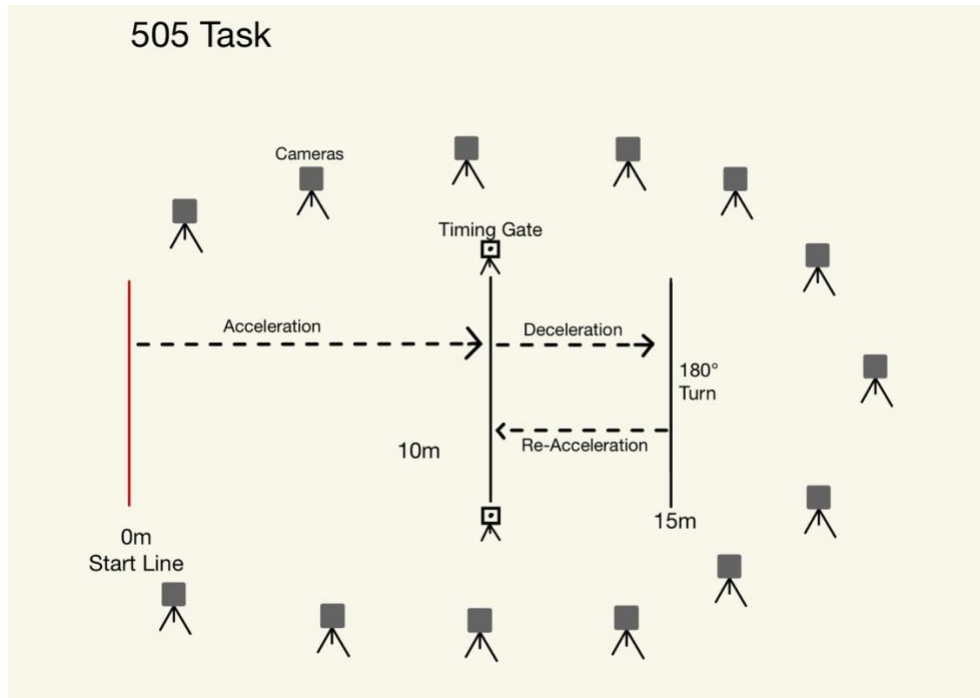


Figure 3: 505 COD Task Set Up

3.5 Statistical Analysis

All statistical analyses were conducted using SPSS (version 31.0.0.0), with significance set a priori at $\alpha = .05$. Descriptive statistics were calculated in Excel to summarize participant-reported RPE and time to complete each repetition of the exercise intervention. Individual movement variability in joint angle from pre- to post-intervention were also examined descriptively, without inferential statistics, to characterize variability in participant responses. Additionally, exploratory analysis using Pearson correlation was performed to determine if changes in RPE scores were correlated with kinematic strategy alterations.

To address the first aim for examining GPS derived decelerations in relation to calculated decelerations through motion capture, a sensitivity and specificity analysis was performed. This analysis determined if GPS could correctly classify whether a deceleration of a certain intensity happened (compared against motion capture). The maximum deceleration value provided by GPS derived deceleration during each movement was compared to the calculated average

deceleration from motion capture, which was determined from the point of peak velocity to the lowest velocity (close to 0.0 m/s). Specificity and sensitivity analyses were performed for each deceleration zone (zones 1-6) from the GPS as this provided information on the probability of GPS giving correct and incorrect identification of deceleration actions. Additionally, a Bland-Altman analysis was conducted to assess the level of agreement between GPS and motion capture derived deceleration values across all intensity zones and movements combined.

To address the second and third aims, this study employed a within-subjects design to evaluate changes in lower extremity kinematics following the exercise intervention. To examine if the accumulation of high-intensity decelerations influenced horizontal deceleration abilities (decrease in pelvis position and number of steps to decelerate), a paired samples t-test was utilized. The paired t-test determined if there was a significant difference in deceleration abilities within-subjects before and after accumulated decelerations were introduced. Pelvis displacement was calculated as the difference in vertical pelvis position at initial contact of the first deceleration step to the point the pelvis was the lowest during the deceleration phase of the ADA, as this metric represents a specific deceleration technique. Additionally, exploratory analysis using Pearson correlation was used to determine if the change in RPE scores were correlated with alterations in pelvis displacement after the exercise intervention.

Lastly, to examine whether an acute bout of accumulated decelerations influenced joint kinematics within-subjects, paired samples t-tests were utilized. For the 505 COD task, the average joint angle of the trunk, hip, knee, and ankle at initial contact of the 180° cut and the point of full weight acceptance (zero-velocity) were compared pre- to post- exercise intervention for all planes (sagittal, frontal, transverse). For the ADA task, kinematic strategy (joint angle progression) across all consecutive deceleration steps in the sagittal plane were compared from

pre- to post- exercise intervention. Kinematic strategy was calculated by taking the slope of a linear regression model of the sagittal plane joint angles for the deceleration phase (Appendix G). The slope of each linear regression model describes the average change in joint angle per deceleration step. Paired samples t-tests were then conducted to examine differences in the average change in trunk, hip, knee, and ankle angles across deceleration steps from pre- to post-exercise intervention. Descriptive statistics of individual movement variability in joint kinematics in the sagittal plane for both ADA and 505 COD task were examined to determine if there were patterns in lower extremity kinematic changes. Lastly, an exploratory analysis using Pearson correlation was applied to determine if the change in RPE score from the beginning to the end of the exercise intervention was correlated with changes in kinematic strategy of the knee during the ADA task.

CHAPTER 4: Results

Across all participants, a total of 270 movement trials were collected, however four trials were excluded due to data transfer errors or non-compliant movement during trials. Data were analyzed using two different approaches: GPS-derived deceleration metrics compared to motion capture (reference standard) for all movements, and kinematic analysis of only the high-speed movements (ADA and 505 COD tasks). Due to the separate analyses, the number of trials included varied per analysis and is reported separately below.

4.1 GPS and Motion Capture Comparison

Comparison between GPS derived decelerations to motion capture calculated decelerations was performed on both slow (squat/walk) and fast movements (ADA/505 COD) to examine whether GPS units could correctly identify varied deceleration intensities. Of the 266 trials, 79 trials across slow and fast movements were excluded due to the GPS units reading 'N/A' for a deceleration. Therefore, the remaining movement trials (N=187, or 70% of trials) were utilized for data analysis.

Absolute deceleration zones were utilized as the default GPS unit settings which identify decelerations into six zones from low intensity (Zones 1-3, 0.5-3.0 m/s²) to high intensity (Zones 4-6, 3.0-10m/s²) and was evaluated against motion capture as the reference standard. Sensitivity ranged from 0.0% to 70.6% across zones, and specificity ranged from 69.9% to 98.9% across zones (Table 3). Sensitivity decreased progressively as deceleration intensity increased, from 70.6% in Zone 1 (0.5-1.0m/s²) to 0% in Zones 5 and 6 (4.0-5.0 m/s² and 5.0-10.0m/s², respectively). Specificity was lowest in Zone 2 (69.9%) and highest in Zone 6 (98.9%). Full confusion matrices can be found in Appendix D.

Table 3: Sensitivity and Specificity between GPS units and Motion Capture

Analysis Type	Zone 1 (0.5 – 1.0 m/s ²)	Zone 2 (1.0 – 2.0 m/s ²)	Zone 3 (2.0 – 3.0 m/s ²)	Zone 4 (3.0 – 4.0 m/s ²)	Zone 5 (4.0 – 5.0 m/s ²)	Zone 6 (5.0 – 10.0 m/s ²)
Sensitivity	70.6%	57.1%	46.2%	12.5%	0%	0%
Specificity	89.4%	69.9%	73%	88.5%	94.9%	98.9%

Bland-Altman analysis revealed the mean difference (systematic bias) between Motion Capture and GPS derived decelerations across all deceleration zones combined (N=187) with the difference calculated as Motion capture minus GPS. The mean difference (systematic bias) was 0.74m/s², with 95% limits of agreement (LOA) ranging from -1.95m/s² to 3.43m/s² (Figure 4). The coefficient of repeatability (COR) resulted at 2.74 m/s², representing 112.74% of the mean deceleration values across both methods.

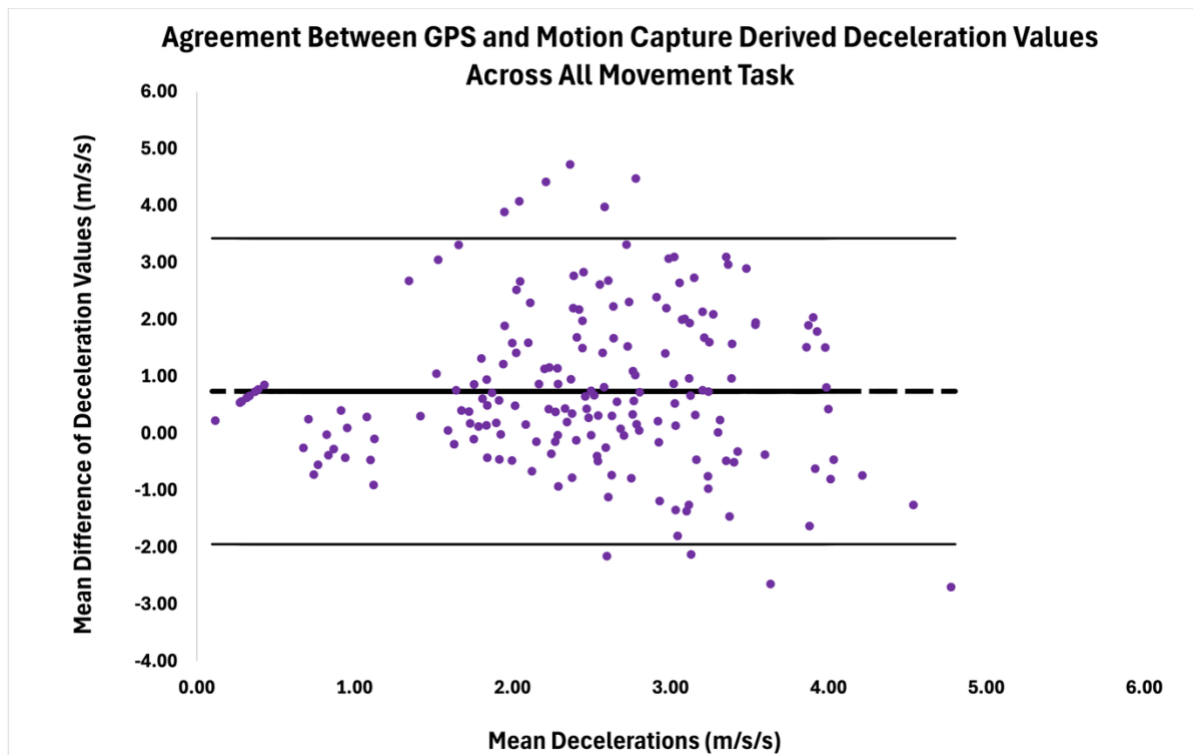


Figure 4: Bland-Altman Analysis: Motion Capture vs. GPS Derived Decelerations (Difference = Motion Capture - GPS)

4.2 Exercise Intervention Protocol

All participants completed 15 repetitions of the 5-10-5 Pro Agility test. RPE scores and time to complete each repetition were collected for all participants (N=15), average results can be found in Table 4. The mean RPE of first repetition was a score of 2.2 (SD = 1.3) and the mean RPE score on the final repetition was 6.1 (SD = 2.1), $t(14) = -7.987, p < .001, g = -1.949$. The mean completion time of the first repetition was 6.88 seconds (SD = 0.67s) and the mean time to complete the last repetition was 6.96 seconds (SD = 0.80s), $t(14) = -1.086, p = .296, g = -.265$. Individual RPE scores and times for each exercise intervention repetition are provided in Appendix E.

Table 4: Participant RPE Scores and Time to Complete Repetitions During Exercise Intervention, presented as mean (SD)

Variable	Rep 1	Rep 2	Rep 3	Rep 4	Rep 5	Rep 6	Rep 7	Rep 8	Rep 9	Rep 10	Rep 11	Rep 12	Rep 13	Rep 14	Rep 15
RPE Scores	2.2 (1.3)	2.7 (1.5)	3.2 (1.9)	3.7 (2.2)	4.2 (2.0)	4.6 (2.1)	4.7 (2.1)	4.7 (2.0)	5.1 (1.9)	5.1 (2.1)	5.5 (2.2)	5.6 (2.0)	5.8 (1.9)	5.7 (1.8)	6.1 (2.1)
Time to Complete Rep (s)	6.88 (0.67)	7.01 (0.78)	7.02 (0.85)	6.99 (0.79)	7.03 (0.81)	6.94 (0.77)	6.99 (0.72)	7.04 (0.87)	7.10 (0.96)	7.10 (0.91)	7.02 (0.90)	7.11 (0.81)	6.94 (0.75)	7.03 (0.83)	6.96 (0.80)

4.3 Cutting – 505 Change of Direction Task

Kinematic data at initial contact and full weight acceptance during the 180° cut on the plant leg were collected for all 15 participants. One participant was excluded from data analysis due to switching the limb used for the 180° cut from pre- to post- testing; therefore, analysis was conducted on the remaining participants (N=14).

4.3.1 Physical Exertion Effects on Lower Extremity Kinematic Changes

Paired samples t-tests to examine pre- to post-test kinematic changes in the lower extremity were conducted on 12 kinematic variables (trunk, hip, knee, and ankle angles across sagittal, frontal, and transverse planes) of the plant foot during the 505 Change of Direction task.

At the point of initial contact (IC) during the 180° cut, mean frontal plane hip joint kinematics resulted in a statistically significant decrease in abduction angle by 2.26° from pre- to post-exercise intervention ($M = -21.71^\circ$ and -19.45° respectively), $t(13) = -3.232$, $p = .007$, $g = -.813$, 95% CI $[-1.384, -.0220]$. Mean trunk angle in the transverse plane was not statistically significant during initial contact from pre- to post-exercise intervention (57.20° and 49.81° respectively), $t(13) = 2.064$, $p = .060$, $g = .519$, 95% CI $[-0.020, 1.042]$. All other joint angles did not reach statistical significance at the point of initial contact during the 180° cut. Results for all kinematic variables at initial contact are presented in Table 5. Pearson correlation between change in RPE scores and change in knee angle at initial contact during the 505 COD pre- and post-exercise intervention resulted with nonsignificant correlation ($r = -0.38$, $p = .18$). At the point of full weight acceptance (when the participant was moving at zero velocity before pushing off into the opposite direction), only mean frontal plane hip joint kinematics showed a statistically significant decrease in abduction angle by 3.20° from pre- to post-exercise intervention ($M = -20.78^\circ$ and -17.58° respectively), $t(13) = -2.655$, $p = .020$, $g = -.668$, 95% CI $[-1.212, -0.103]$. A summary of results for kinematic changes during full weight acceptance on cut leg can be found in Table 6.

Table 5: Paired Samples T-Test Results for all joint angles on plant leg at Initial Contact During 505 COD Task. Positive values indicate flexion/dorsiflexion, adduction/inversion, and internal rotation.

Joint/Plane	Pre Intervention Mean (SD)	Post Intervention Mean (SD)	95% CI of the Mean Difference	t-value	p-value	Hedges's Correction (g)
Trunk						
Flexion/Extension (°)	6.53 (8.17)	6.23 (6.70)	[-3.461, 4.061]	.172	.866	.043
Abduction/Adduction (°)	-24.12 (11.80)	-23.95 (14.66)	[-3.319, 2.966]	-.121	.905	-.030
Rotation away from plant leg (+)/ Rotation towards plant leg(-) (°)	57.20 (31.31)	49.81(35.87)	[-0.346, 15.124]	2.064	.060	.519
Hip						
Flexion/Extension (°)	42.76 (8.30)	43.52 (7.42)	[-2.359, 0.8442]	-1.022	.326	-.257
Abduction/Adduction (°)	-21.71(4.52)	-19.45 (5.32)	[-3.779, -0.751]	-3.232	.007*	-.813
Internal/External Rotation (°)	5.79 (5.74)	5.27 (5.45)	[-1.967, 3.018]	.456	.656	.115
Knee						
Flexion/Extension (°)	26.84 (5.58)	28.13 (6.13)	[-4.463, 1.879]	-.880	.395	-.221
Abduction/Adduction (°)	-7.20 (2.82)	-7.41 (2.57)	[-1.330, 1.755]	.298	.771	.075
Internal/External Rotation (°)	-11.62 (6.13)	-10.43 (5.23)	[-3.143, 0.745]	-1.332	.206	-.335
Ankle						
Dorsiflexion/ Plantarflexion (°)	-18.99 (5.97)	-17.02 (6.48)	[-5.870, 1.941]	-1.087	.297	-.273
Inversion/Eversion (°)	8.32 (5.44)	9.61 (2.64)	[-4.018, 1.436]	-1.023	.325	-.257
Internal/External Rotation (°)	9.17 (6.33)	7.14 (5.87)	[-1.115, 5.187]	1.396	.186	.351

* = $p < .05$

Table 6: Paired Samples T-Test Results for all joint angles on plant leg at Full Weight Acceptance (Zero Velocity) During 505 COD Task. Positive values indicate flexion/dorsiflexion, adduction/inversion, and internal rotation.

Joint/Plane	Pre Intervention Mean (SD)	Post Intervention Mean (SD)	95% CI of the Mean Difference	t-value	p-value	Hedges's Correction (g)
Trunk						
Flexion/Extension (°)	22.60 (24.17)	20.99 (24.82)	[-4.437, 7.661]	.576	.575	.145
Abduction/Adduction (°)	-39.27 (14.08)	-35.29 (11.07)	[-9.562, 1.606]	-1.539	.148	-.378
Rotation away from plant leg (+)/ Rotation towards plant leg(-) (°)	97.21 (58.03)	45.04 (107.02)	[-18.651, 122.994]	1.591	.136	.400
Hip						
Flexion/Extension (°)	45.75 (15.85)	42.66 (15.13)	[-5.209, 11.385]	.804	.436	.202
Abduction/Adduction (°)	-20.78 (7.53)	-17.58 (8.56)	[-5.801, -0.596]	-2.655	.020*	-.668
Internal/External Rotation (°)	8.00 (5.37)	6.15 (5.09)	[-2.819, 6.531]	.858	.407	.216
Knee						
Flexion/Extension (°)	55.58 (13.93)	51.87 (12.30)	[-3.646, 11.056]	1.089	.296	.274
Abduction/Adduction (°)	-2.94 (5.44)	-3.44 (4.76)	[-2.243, 3.235]	.391	.702	.098
Internal/External Rotation (°)	-3.49 (8.92)	-5.04 (7.25)	[-1.731, 4.825]	1.020	.327	.256
Ankle						
Dorsiflexion/Plantarflexion (°)	13.38 (11.60)	11.11 (11.18)	[-0.787, 5.325]	1.604	.133	.403
Inversion/Eversion (°)	17.97 (4.05)	19.07 (4.00)	[-2.449, 0.253]	-1.756	.103	-.442
Internal/External Rotation (°)	-2.02 (7.99)	-0.31 (8.18)	[-5.486, 2.061]	-0.981	.345	-.247

* = $p < .05$

4.3.2 Individual Variability

Individual participant responses were examined descriptively to characterize variability in responses to the exercise intervention at initial contact and at full weight acceptance during the 505 COD. No inferential statistics were applied to the individual movement variability data.

Descriptive analyses were limited to sagittal plane kinematics only.

During the initial contact period of the 180° cut, 8 of 14 participants had increased trunk and knee flexion angles while 6 of 14 had decreased joint angles; 6 of 14 participants had increased hip flexion, while 8 of 14 decreased hip flexion; and 5 of 14 participants increased in

plantarflexion, while 9 of 14 decreased plantarflexion angle (Figure 5). At the point of full weight acceptance, 4 of 14 participants increased in trunk and knee flexion, while 10 of 14 had a decrease in flexion angle; 5 of 14 had increased hip flexion, while 8 of 14 decreased hip flexion and 1 of 14 had no change; 8 of 14 had increased plantarflexion, while 6 of 14 demonstrated increased dorsiflexion (Figure 6).

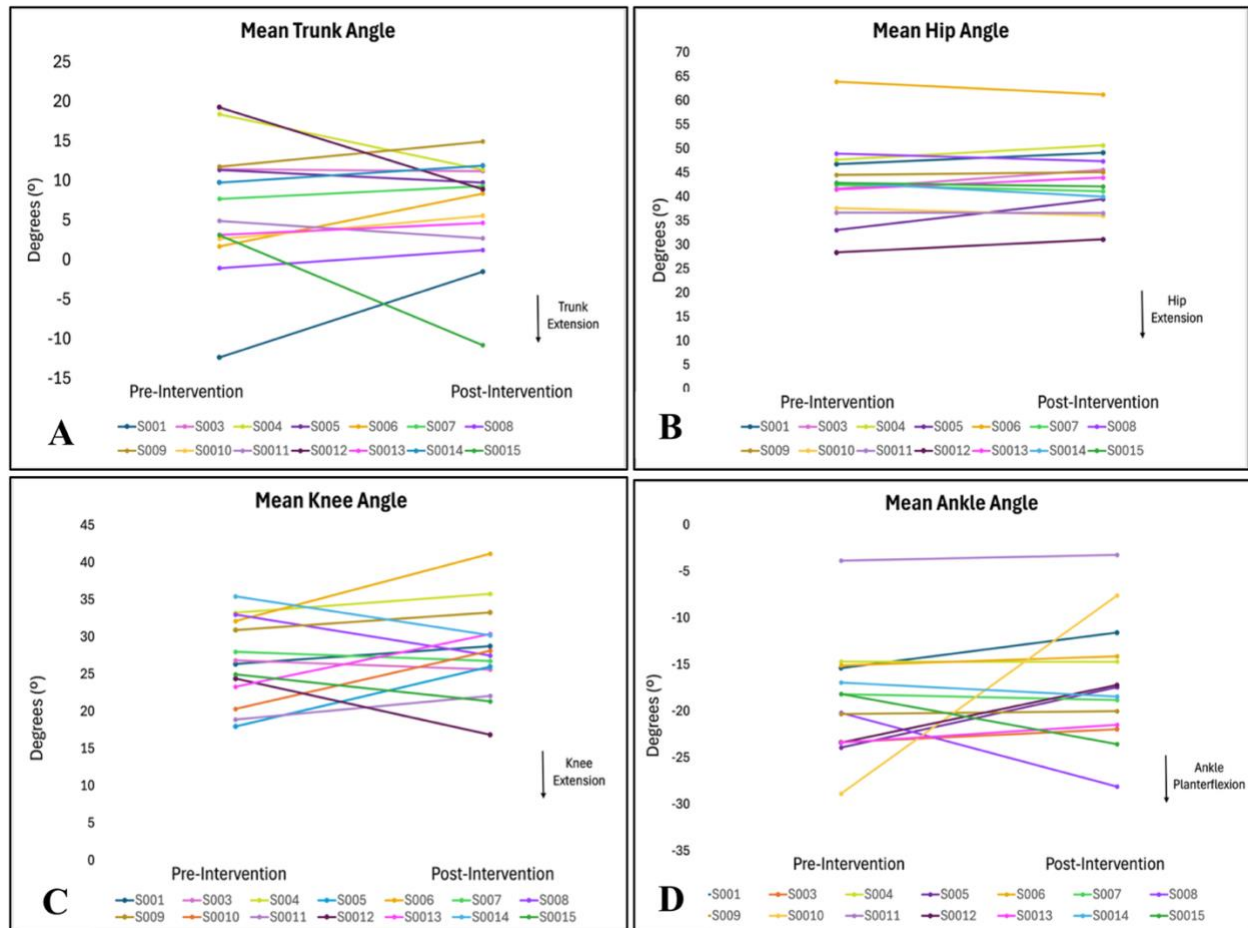


Figure 5: Comparison of Mean joint sagittal plane angles for (A)Trunk, (B)Hip, (C)Knee, and (D)Ankle at Initial Contact during the 505 COD task before and after completing the exercise intervention.

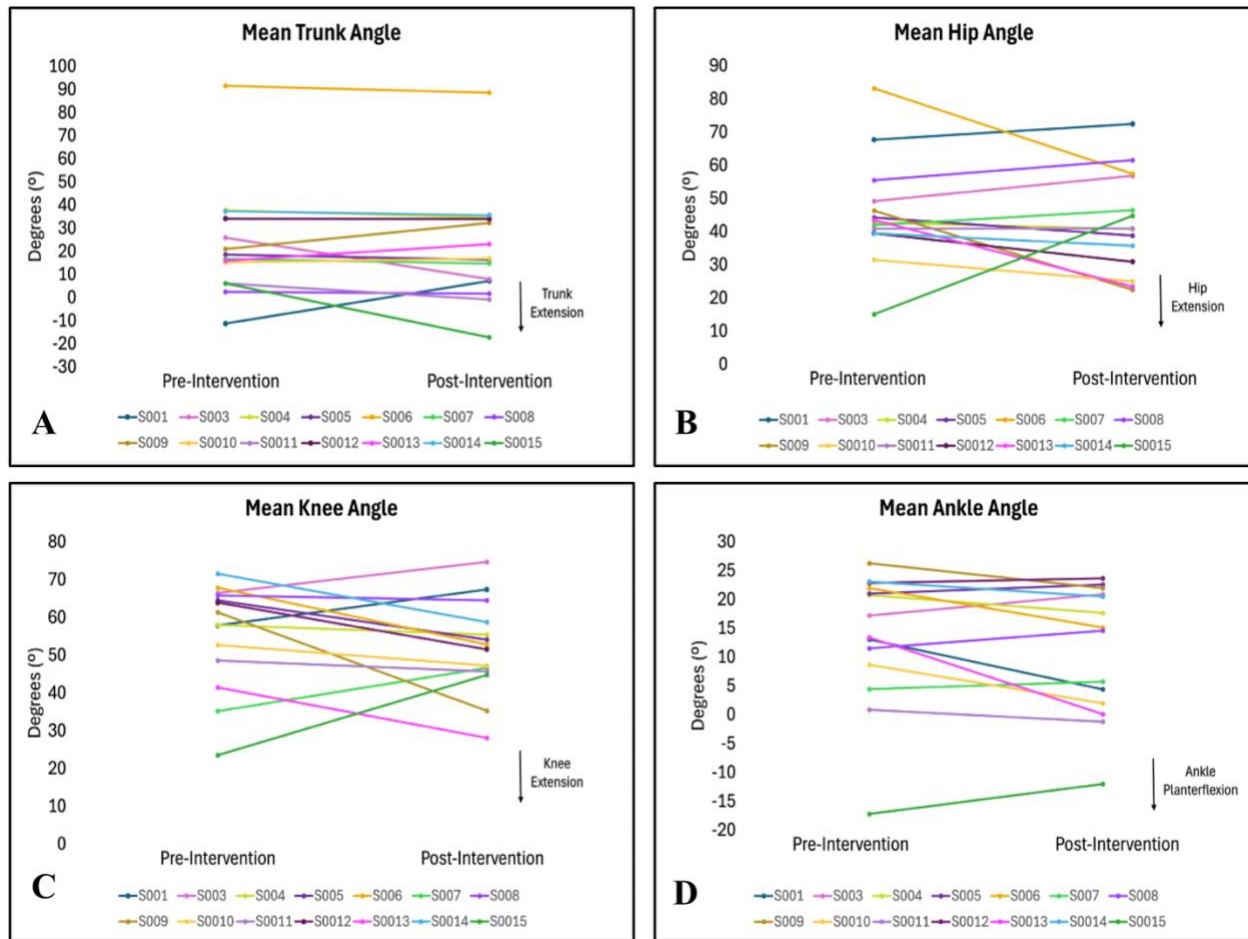


Figure 6: Comparison of Mean joint sagittal plane angles for (A) Trunk, (B) Hip, (C) Knee, and (D) Ankle at point of full weight acceptance (zero velocity) during the 505 COD task before and after completing the exercise intervention

4.4 Horizontal Deceleration: Acceleration Deceleration Assessment

4.4.1 Physical Exertion Effect on Lower Extremity Kinematic Changes

Statistical analysis for joint angle progression per deceleration step in the sagittal plane for the trunk, hip, knee, and ankle across all deceleration steps during the ADA were conducted for all participants (N=15). There were no statistically significant differences found in the joint angle progression per deceleration step in the four kinematic variables from pre- to post- exercise intervention (all $p > .05$, Table 7). Individual participant kinematic data and linear regression models are presented in Appendix G. Pearson correlation between change in RPE score and knee angle progression per deceleration step resulted in a nonsignificant correlation ($r = 0.05$, $p = .85$).

Table 7: Summary of Paired t-test on the joint angle progression per deceleration step across Each consecutive step required to decelerate during the ADA pre- and post- exercise intervention. Positive values indicate flexion/dorsiflexion, and negative values indicate extension/plantarflexion.

Joint/Plane	Pre Intervention Mean (SD)	Post Intervention Mean (SD)	95% CI of the Mean Difference	t-value	p-value	Hedges's Correction (g)
Trunk						
Flexion/Extension (°)	-1.62 (2.29)	-1.64 (2.89)	[-0.868, 0.904]	.044	.966	.011
Hip						
Flexion/Extension (°)	0.87 (3.84)	0.918 (3.62)	[-0.993, 0.900]	-.105	.918	-.026
Knee						
Flexion/Extension (°)	3.50 (2.77)	3.776 (3.18)	[-1.518, 0.961]	-.482	.637	-.118
Ankle						
Dorsiflexion/Plantarflexion (°)	-0.81 (2.50)	-0.639 (1.84)	[-0.960, 0.617]	-.466	.648	-.114

A paired samples t-test revealed no statistically significant differences in pelvis displacement from pre- to post- exercise intervention across all participants, $M_{diff} = 0.0004\text{m}$, $SD = 0.03\text{m}$, $t(14) = .062$, $p = .951$, 95% CI of the difference $[-0.014, 0.015]$. Pearson correlation between change in RPE score and change in vertical pelvis displacement revealed a nonsignificant correlation ($r = 0.46$, $p = .08$). Additionally, a paired-samples t-test of the number of steps it took participants to decelerate during the ADA from pre- ($M = 5.47$ steps) to post- ($M = 5.80$ steps) exercise intervention revealed no statistically significant differences, $t(14) = -1.581$, $p = .136$, 95% CI $[-0.785, 0.119]$. Lastly, the approach velocity or maximum velocity before taking the first deceleration step was also analyzed through paired samples t-test. There was no statistical difference in the maximum velocity entering into the ADA across participants, $M_{diff} = 0.095\text{m/s}$, $SD = 0.25\text{m/s}$, $t(14) = 1.447$, $p = .170$, 95% CI of the difference $[-0.047, 0.235]$.

4.4.2 Individual Variability

Individual participant responses were examined descriptively to examine individual variability in responses to the exercise intervention. No inferential statistics were applied to the

individual-level data. Descriptive analyses were limited to sagittal plane kinematics. The individual change in vertical displacement during the ADA from pre- to post- intervention can be seen in Figure 7. From pre- to post- intervention, 6 of 15 participants decreased their pelvis vertical displacement across deceleration steps, 8 of 15 participants increased the amount of pelvis vertical displacement, and 1 of 15 participants demonstrated no change in the amount of pelvis displacement across decelerations steps. Maximum velocity before the first deceleration step in the ADA increased from pre- to post- exercise intervention for 5 of 15 participants, while 10 of 15 participants exhibited a decrease in the maximum velocity at ADA approach. Individual maximum velocities from pre- to post- intervention can be found in Appendix F.

Across all deceleration steps during the ADA, the average rate of angular change per deceleration step in trunk angle decreased (toward extension) for 6 of 15 participants and increased (toward flexion) for 9 of 15 participants. The average rate of angular change per deceleration step in hip angle decreased (toward extension) for 8 of 15 participants and increased (toward flexion) for 7 of 15 participants. The average rate of angular change per deceleration step in knee angle decreased (towards extension) for 7 of 15 participants and increased (toward flexion) for 8 of 15 participants. The average rate of angular change per deceleration step in ankle angle decreased (toward plantarflexion) for 7 of 15 participants and increased (toward dorsiflexion) for 8 of 15 participants (Figure 8).

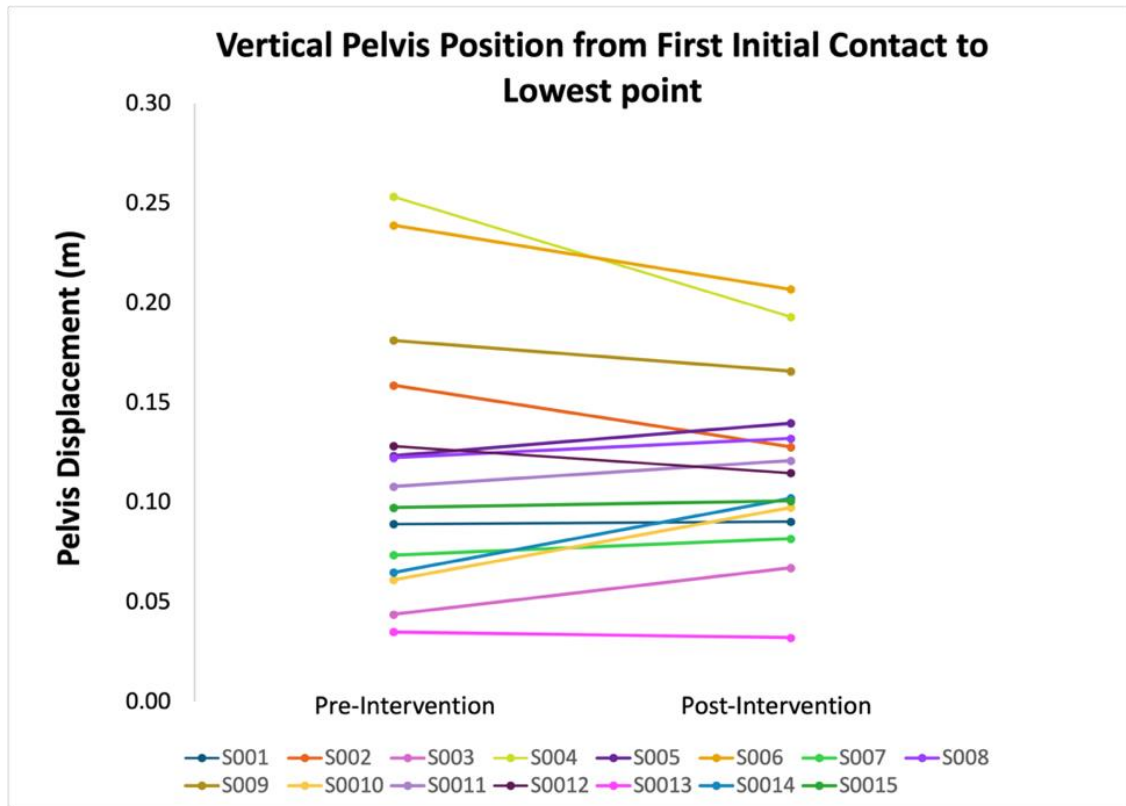


Figure 7: Comparison of Vertical Displacement of the Pelvis (meters) from first deceleration step initial contact to the point of lowest pelvis position.

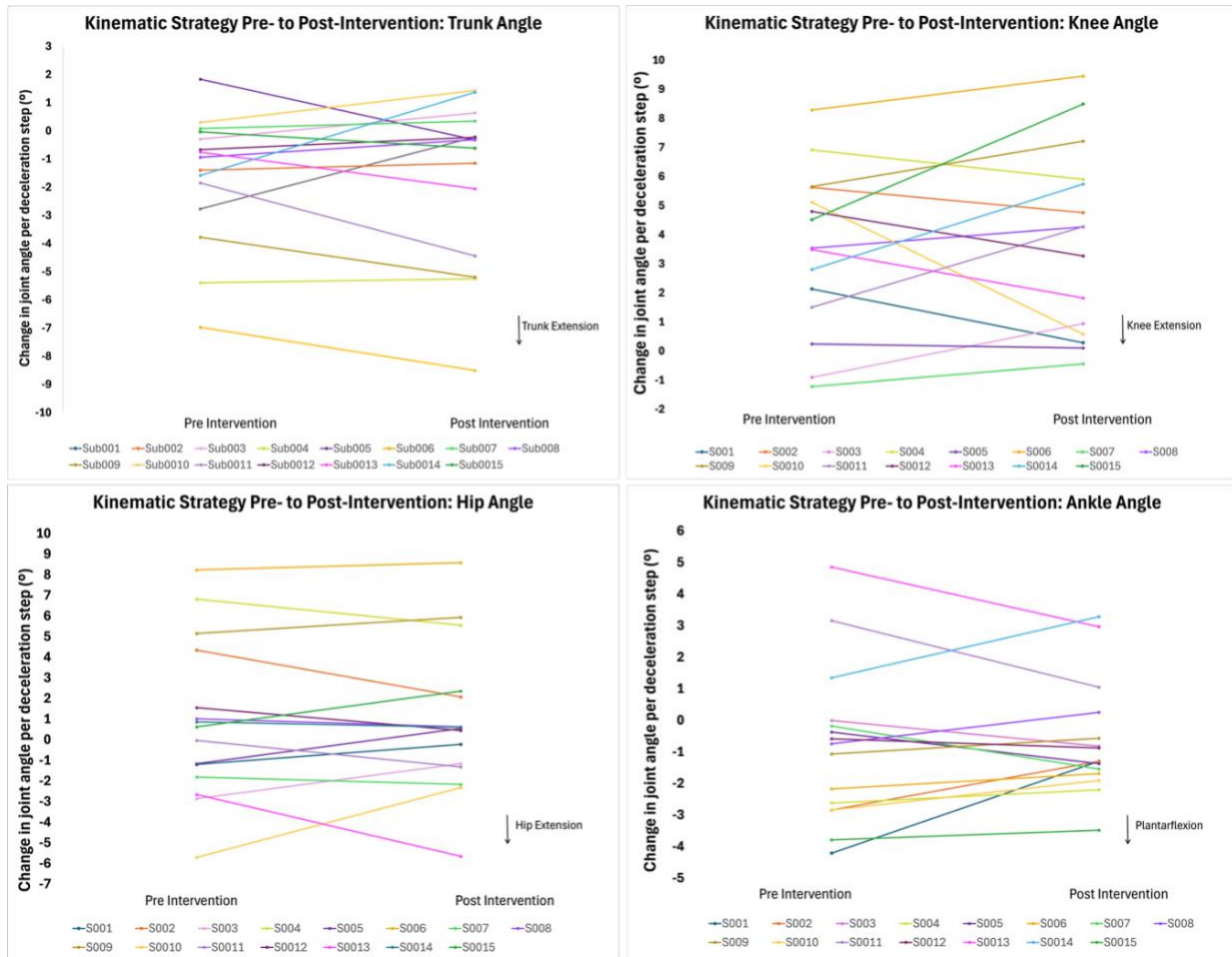


Figure 8: Average change in joint angle (sagittal plane) per deceleration step pre- and post-exercise intervention for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right). The change in joint angle explains movement strategy across deceleration steps.

CHAPTER 5: Discussion

The aim of the present study was threefold: (1) evaluate GPS unit's ability to correctly classify deceleration actions into predefined absolute intensity zones using motion capture as the reference, (2) examine if accumulated high-intensity decelerations influence horizontal deceleration ability, and (3) examine whether accumulated decelerations influence lower extremity kinematics during standardized deceleration tasks. Addressing these aims would improve the interpretation of deceleration metrics for athlete monitoring and provide insight into the biomechanical demands that may contribute to performance decrements and injury risk. Overall, GPS demonstrated moderate to high specificity, suggesting that GPS was able to identify when a deceleration did not belong in a specific intensity zone. However, GPS demonstrated low to moderate sensitivity, indicating that GPS frequently failed to correctly identify deceleration actions into the proper intensity zones compared to motion capture; therefore, partially supporting the first hypothesis.

The second hypothesis was not supported as physical exertion effects on deceleration ability changes (pelvis position height and number of steps to decelerate) during the ADA task resulted in non-significant differences. The third hypothesis was partially supported as hip abduction angle was found to significantly decrease during the initial contact and full weight acceptance periods of the 505 COD task after completing the exercise intervention.

5.1 GPS Sensitivity & Specificity of Deceleration Detection

The main findings for GPS derived maximum decelerations provided by the GPS software for athlete load monitoring revealed GPS has a poor to moderate ability to correctly identify a deceleration action into the correct absolute intensity zone. Sensitivity varied from 0.0% to 70.6% across deceleration intensity zones, indicating that the GPS unit was able to correctly

identify a deceleration action into the correct deceleration intensity zone as infrequently as never and as often as 70%. GPS sensitivity was highest in Zone 1 (0.5m/s^2 - 1.0m/s^2), however as participants increased their deceleration intensity, the GPS accuracy declined. Specificity ranged from 69.9% to 98.9% across deceleration intensity zones. Specificity initially decreased from Zone 1 (89.4%) to Zone 2 (69.9%) but steadily increased from Zone 3 (73%) through Zone 6 (98.9%). The higher specificity values indicated that the GPS was able to correctly identify when a deceleration did not belong in a specific intensity zone as determined by motion capture, reflecting a low rate of false positives across deceleration intensity zone.

The Bland-Altman analysis further indicated poor agreement between the GPS and 3D motion capture, where the GPS unit was more commonly underestimating deceleration intensity relative to motion capture across all movements. The wide LOA (range of 5.4m/s^2) between the two systems means there was considerable variability in agreement between the GPS unit and motion capture derived decelerations.

To the author's knowledge, there is one other study that has examined the validity and reliability of GPS (10Hz) deceleration derived values, but only during the ADA test with a 20-m sprint approach distance. In a study conducted by Jones et al. (2025) using 32 male youth soccer players (age: 18.1 ± 1.6 years), participants performed the ADA test and examined decelerations with a radar device collecting at 47Hz and both the raw velocity data and maximum deceleration directly obtained from GPS software from the 10Hz GPS units. They found that the GPS exported maximum decelerations had a LOA range of 2.26 m/s^2 with an upper and lower limit of -1.25m/s^2 and 1.01m/s^2 respectively, which is lower than the LOA range observed in the present study. There are three likely key methodological differences to account for this discrepancy. First, the present study pooled all movement types (squat, walk, ADA, and 505 COD) into a

single Bland-Altman analysis rather than isolating the ADA test as Jones et al. (2025) did, therefore direct comparison should be interpreted cautiously. Secondly, Jones et al. (2025) compared GPS derived maximum deceleration to radar-derived maximum deceleration, whereas the present study compared GPS derived maximum deceleration provided by GPS software to motion capture average deceleration in the horizontal direction. Third, the present study was performed indoors compared a Jones et al.'s collection outdoors. GPS units are designed to be utilized outdoors as they require GNSS satellite connectivity to use positional data (change in one's coordinates) to determine velocity and further calculate deceleration if there was a decrease in velocity for at least 0.5 seconds. A disadvantage to GPS devices is they require outdoor use (Harper et al., 2026). The original design of the present study was intended for outdoor use, however due to weather constraints, the study protocols were performed indoors.

The most likely reason for the larger LOA examined in this study is believed to be caused by using the GPS system indoors as previous literature reporting the accuracy, validity, and/or reliability of GPS units have used the system outdoors (Beato et al., 2018; Beato, Wren & Keijzer, 2024; Jones et al., 2025). When using GPS sensors indoors, their satellite connectivity can be interrupted by the walls, tall stadiums, etc. which lead to issues in GPS based metrics such as changes in position and velocity. The use of local positioning systems (LPS) has been tested over the last decade, where player's wear units similar to GPS units, however receivers (anchors) are placed around a gym and ultra-wideband or radio frequency identification waves are sent between the units and the anchors to track movement patterns, accelerations, jumps etc. Therefore, sensitivity and specificity analysis should be interpreted with caution as indoor use of the GPS units could have impeded signals to satellites to collect accurate positional data for

small-distance movements, and future studies indoors could use LPS to avoid the satellite dependency issue.

Separate from the comparison to Jones et al. (2025), the discrepancy in sampling frequency between the systems may contribute to the low to moderate sensitivity and poor agreement between systems. Motion capture data was collected at 120Hz whereas GPS data was collected at 10Hz. Given GPS units collect at a lower frequency than motion capture, fewer data points are available within any given time window to detect when a deceleration action starts and ends. As a result, the onset of deceleration actions may be less accurate which further influence the GPS's ability to calculate deceleration from changes in velocity, an effect likely negatively influenced more when performed indoors. The GPS units used in the present study were equipped with accelerometers that collect data at 100Hz. The maximum deceleration values are derived from the 10Hz position/velocity data, not the accelerometer data so the higher-frequency signal is not leveraged for this metric unless users extract the raw accelerometer data manually. The software designed by the GPS company does an excellent job at applying smoothing filters to accelerometer data to remove exaggerated noise from high-impact foot strikes, however the smoothing process is not well understood by users. Examining raw accelerometer data may be a possible source to find true deceleration peaks from the GPS units in comparison to motion capture, however, it is not a feasible method for coaching staff who have limited experience with raw accelerometer data. To examine raw accelerometer data, smoothing filters applied through coding programs would need to be utilized. Additionally, an individual would need to understand how the GPS unit tilts with movement, as this causes the deceleration to be described across multiple planes of movement rather than just one. Additionally, accelerometer data will show noise from high impact steps, which could result in values that are not descriptive of the

deceleration phase but of a more specific impact the player experienced. For example, a foot strike during deceleration can produce a spike in accelerometer data that exceeds 1g, however this is not a reflection of true sustained reduction in velocity. Motion capture however is capable of directly capturing the point of true maximum and minimum velocity in different directions to calculate deceleration during short movements such as the ADA and 505 COD.

As noted above, the present study compared motion capture derived average deceleration in the horizontal direction to GPS maximum deceleration through the GPS software (movement may be in multiple planes based on GPS unit orientation). Maximum deceleration is an easily accessible metric provided to coaches and training staff in the live tracking app or once the GPS data is exported onto a computer with the GPS software. If coaches wanted to access a true comparison of average deceleration derived with GPS, they would have to download the raw accelerometer data, apply a smoothing filter to the data, and use a specialized code (i.e. in R) to calculate decelerations, which takes time and experience that coaches/staff may not have. Collectively, these findings suggest that practitioners should exercise caution when relying on GPS software provided deceleration classifications, particularly at higher intensities when using GPS units indoors, as low to moderate sensitivity of GPS units may be under- or overestimation of player load when using absolute deceleration intensity zones.

5.2 Effect of Accumulated Decelerations on Deceleration Ability

The second aim of the study specifically examined if accumulated high-intensity decelerations influenced horizontal deceleration ability, including a decrease in pelvis position and the number of steps it took to decelerate during the ADA task. Lowering the center of mass during deceleration steps is a technique which aids in prevention of forward rotation about the lead braking foot and is key in stabilization which prolongs the time in which braking forces can

be applied (Harper et al., 2024; Harper et al., 2026). Additionally, lowering the body's center of mass will lead to an increase in flexion of lower extremity joints which further assist in attenuation of braking forces across deceleration steps. Although this is one technique into proper deceleration, to the author's knowledge, there is no literature examining the vertical displacement of the pelvis after an exercise intervention.

The present study used pelvis position displacement as a measure of whole-body center of mass vertical change from the first braking step during the ADA to the step of lowest pelvis position. The average vertical pelvis displacement across all participants from pre- to post-exercise intervention was 0.12m (± 0.07 m) and 0.12m (± 0.05 m) respectively. The minimal average change in vertical displacement in pelvis position could have been caused by comparing the average change in position across participants from pre- to post- exercise intervention, whereas individual descriptive analysis (Figure 7) displayed individualistic techniques in decelerating. The nature of the present study included recreationally active college females of all levels, from DI college athletes to participants who are part of non-competitive recreational sports teams. Additionally, participants are from a variety of sports including soccer, field hockey, volleyball, basketball, etc. With a range of physical activity levels and sport participation, individuals may be coached with different deceleration techniques (or never taught how to decelerate). For example, a participant who played on a DI soccer team had the greatest vertical pelvis displacement pre- and post- intervention (0.25m and 0.19m respectively), while a participant who participates in non-competitive basketball games had the least vertical pelvis position pre- and post- exercise intervention (0.04m and 0.03m respectively). For the participant with the greatest vertical displacement, it was noted they also had an increase in both hip and knee flexion, which is a result of lowering the body's center of mass. Increased flexion at the hip

and knee may allow eccentric muscle contraction of the hamstrings and gluteus maximus to absorb braking forces at both the hip and the knee. On the contrary, the participant with the least amount of lowering of the center of mass showed extension at the hip and a reduced angle of flexion at the knee.

It is possible that for this participant, eccentric muscle contractions could not absorb braking forces as efficiently as the highly trained participant. These kinematic differences across participants may possibly suggest differences in muscular force production, and future studies should investigate muscular force output as a result of kinematic changes in high deceleration tasks. Additionally, kinematic differences could be a result of general skill level with performing deceleration movements as some participants may be more familiar with these kinds of deceleration movements, or it could reflect differences in physical exertion experienced during the intervention. Participant RPE scores increased significantly, with scores increasing between 1 to 7 points from the first to last repetition of the exercise intervention, reflecting that some individuals had a substantial rise in perceived exertion. However, change in RPE score was not significantly correlated with pelvis vertical displacement before and after the exercise intervention ($p > .05$), suggesting that the magnitude of perceived effort was not a predictor for how much participants lowered their COM. With individual variability, deceleration technique can influence the lower extremity kinematic chain and needs to be trained based on the individuals' capabilities.

Another deceleration ability is the number of steps to decelerate. With a greater approach velocity, the number of braking steps will increase to reduce momentum because of lower braking and muscle force-generating capacities (Harper et al., 2026). In the present study, no statistically significant differences were observed in the number of steps it took to decelerate

during the ADA or in the maximum velocity in approaching the ADA task from pre- to post-exercise intervention. These finding may suggest that deceleration strategy was preserved after the exercise intervention. However, it is also possible the exercise intervention did not elicit changes as the intervention of an acute bout of high intensity decelerations was not a sufficient magnitude to disrupt horizontal deceleration abilities in all participants This further reflects on individual variability between movement strategies and perceived effort, and future research should examine training background and skill level in addition to exertion levels from an exercise protocol.

5.3 Effect of Accumulated Decelerations on Lower Extremity Kinematics

5.3.1 ADA Task

The third aim of the present study examined whether an acute bout of exercise with accumulated decelerations altered lower extremity kinematics during standardized deceleration tasks. During the ADA task, there were no significant within-subjects lower extremity kinematic changes in the joint angle progression per deceleration step, therefore not supporting the third hypothesis. The joint angle progression per deceleration step for the trunk, hip, knee, and ankle were examined as this would explain how much each lower extremity joint would increase or decrease in flexion (dorsiflexion) across all deceleration steps. As some participants took varying deceleration steps from pre- to post- exercise intervention, the joint angle progression per deceleration step provides a valuable understanding in participant techniques at each joint through the entire deceleration phase.

Despite within-subject differences in kinematic strategies not showing any significant differences after the exercise intervention, this does not mean there were not meaningful biomechanical responses due to individual movement variability. Having participants of varying

skill levels resulted in various movement techniques to slow whole body momentum down, which is shown in Figure 7. There was a roughly proportional split between participants who increased versus decreased their joint angle progression per deceleration step (i.e. 8 of 15 increased knee flexion angle vs. 7 of 15 progressed into more knee extension across deceleration steps). These differences between individuals mean that the kinematic strategies used (change in joint angle per deceleration step) at the individual level could have potentially cancelled out when averaging during the paired t-test, producing a null group difference. Individual movement variability could point to skill level differences in the ability to decelerate at a high intensity.

As part of the exploratory analysis and consistent with the nonsignificant correlation between RPE and vertical pelvis displacement, change in RPE was also not significantly correlated with knee progression angle per step during the deceleration phase ($p > .05$). It is possible that the exercise intervention itself may not have required sufficient physical demands for all participants to elicit changes in lower extremity kinematics or it could be that movement occurring in one direction could result in less compensatory movement behaviors during straight-line decelerations. However, given individual variability in RPE and biomechanical responses to the exercise intervention, some participants may have experienced meaningful changes in physical demands relative to their own abilities due to increasing RPE scores.

Firstly, looking at trunk angle relative to the ground during the ADA task. Across deceleration steps, trunk angle went into a more erect or posterior lean (hyperextension) for 6 of 15 participants, while 9 of 15 participants had an increase in trunk flexion. One proper technique of a deceleration is to position the body's COM behind the base of support or leading limb at every foot contact in order to control the body's momentum and increase braking impulse (Harper et al., 2026; Hewitt et al., 2011). Not only does a more erect/posterior leaned trunk assist

in better braking abilities, the 9 of 15 participants who had increasing forward trunk lean (flexion), may place the hamstrings at a heightened risk if they had an extended knee joint across deceleration steps, as this excessive lengthened state can possibly contribute to the risk of hamstring strain due to the requirement of controlling high eccentric braking loads at a non-optimal length.

Moving down the kinetic chain, Hewitt et al. (2011) also discussed that immediate hip and knee flexion and ankle dorsiflexion upon initial contact occur to dissipate impact forces with each foot ground contact. The present study only examined the initial contact during each consecutive step; however, we were able to examine trends of only 7 of 15 participants increasing in hip flexion across steps, 8 of 15 increased in knee flexion and 8 of 15 participants increased in dorsiflexion. Unlike Hewitt et al. (2011), the present study was not examining joint angle change for the first 20% of initial contact, however increased flexion angles at initial contact still exemplify superior deceleration mechanics. Increased dorsiflexion allows the tibia to produce the most negative mechanical work, increased knee flexion allows the quadriceps to produce more negative mechanical work, and increased hip flexion allows the gluteus and hamstrings to produce negative mechanical work all to absorb forces at impact with each step (Fitzwilliam et al., 2024; Verheul, Harper, & Robinson, 2024).

Focusing on the knee joint angle progression per deceleration step, the knee starting at slight flexion with the initial contact and increasing in flexion throughout consecutive steps is critical to decreasing knee loads such as shear and compressive forces caused by high braking forces. ACL injuries are noncontact injuries that are largely determined by these factors causing higher knee loading. During high intensity decelerations from the ADA task, peak braking GRF have been found to be on average 4.38 times body mass at the first braking foot contact (Lin et

at., 2025). Exceptional braking forces during horizontal decelerations demonstrate the importance of examining lower extremity biomechanics during consecutive deceleration steps, especially as decelerations are used in high amounts during multidirectional sports. These findings therefore require further investigation at the individual level as horizontal deceleration techniques are subject specific and individuals may have different areas they need to improve to reduce their risk of injury.

5.3.2 505 COD task

The third hypothesis that an acute bout of exercise including 30 high intensity decelerations would elicit changes in lower extremity kinematics was partially supported, as hip abduction angle at initial contact and at full weight acceptance during the 505 COD task were the only variables to show statistically significant changes. A decrease in hip abduction angle at initial contact of a 180° cut was similar to the findings presented by Zago et al. (2021) who examined the changes in lower extremity kinematics during continued 5-m shuttle tests. For Zago et al. (2021), 8 of 20 soccer players demonstrated a 5-10° decrease in hip abduction as they performed 5-m shuttle test until exhaustion, while the present study found a 2.26° decrease in hip abduction angle from pre to post exercise intervention. Slightly different methodologies were utilized as the present study utilized the 5-10-5 pro agility test compared to a 5-m shuttle in both directions (to not overexert one limb). These findings suggest that an acute bout of exercise including 30 high intensity deceleration movements for CODs will elicit small but meaningful changes in lower extremity kinematics in hip frontal plane movement. The main difference between the two studies is Zago et al. (2021) let participants perform shuttles until exhaustion which ranged from 23-67 completed shuttle test with a 180° turn on the dominant limb (non-dominant limb matched shuttle test number). The decrease in hip abduction and increase towards hip adduction is one

mechanical change that can possibly lead to increased loading on the knee, specifically frontal plane, which places an individual at risk of ACL injury hip adduction causes the knee to move into valgus during a cut (Burnham et al., 2026). Neither the present study nor Zago et al. (2021) found significant changes in frontal plane knee angles, suggesting that the hip kinematics were still insufficient to influence the rest of the kinematic chain. Additionally, unlike the present study, Zago et al. (2021) also found significant changes in hip and knee flexion angles with a reduction of 10-25° and 5-20°, respectively, during initial foot contact for the COD. This discrepancy could be primarily due to the differences in exercise protocols as the present study used a standardized design of a fixed 30 high intensity decelerations. The significant change in hip abduction during full weight acceptance during the 505 COD task also follows this pattern as this time point would coincide with peak ground reaction force before moving into the new direction. The reduced hip abduction at this time may also place the knee under great loads in the frontal plane when deceleration demands are highest.

With frontal plane hip angles being the only variable that reached statistical significance, the results suggest that like the ADA, individual variability in movement strategy may have possibly influenced the overall findings of mean joint angle at initial contact and full weight acceptance during a cut. During initial contact of the 505 COD there was a proportional mix of participants who increased or decreased sagittal plane joint angles from pre- to post- exercise intervention. Eight of 14 participants had increased trunk and knee flexion angles while 6 of 14 had decreased joint angles; 6 of 14 participants had increased hip flexion, while 8 of 14 decreased hip flexion; and 5 of 14 participants increased in plantarflexion, while 9 of 14 went towards more dorsiflexion. At the point of full weight acceptance, 4 of 14 participants increased in trunk and knee flexion, while 10 of 14 had a decrease in flexion angle; 5 of 14 had increased hip flexion,

while 8 of 14 decreased hip flexion and 1 of 14 had no change; 8 of 14 had increased plantarflexion, while 6 of 14 demonstrated increased dorsiflexion. Individual analysis can display the influence of the exercise intervention based on joint angles at both time points. For example, participants who decreased hip flexion angle at initial contact and full weight acceptance, showed poorer deceleration mechanics as a ‘stiffer’ hip does not allow the gluteus maximus or hamstrings to absorb braking forces. This reduced biomechanical technique in decelerating after the exercise intervention may be caused by the high level of physical exertion felt by participants which could lead to altered neuromuscular performance in controlling the body (Taylor, 2024). Consistent with the non-significant findings RPE correlations reported for the ADA task, the present study also did not find significant correlations between changes in RPE score and knee angle at initial contact pre- and post- exercise intervention ($p > .05$) through exploratory analysis. Individual variability in RPE scores and movement strategies still pose an important finding as they suggest that perceived exertion alone may not reflect a specific biomechanical adaptation in response to repeated bouts of decelerations. Individual movement variabilities are a result of the physical capacity participants have and should be monitored separately as skill levels, training background, and compensation strategies to perform deceleration tasks vary and should be considered separately from within-subject differences.

5.4 Limitations

The primary limitation regarding GPS derived decelerations was using the system indoors as it is recommended to always utilize the software outdoors for adequate GNSS satellite connection (Harper et al., 2026). The indoor facility possibly inhibited GPS tracking across all movements due to not traveling far distances before performing deceleration limiting positional data to be tracked properly. Collecting outdoors however presents its own issues with collection

motion capture data as those systems also require multiple power sources to run equipment. Thus, collecting kinematic data in an outdoor setting or “on the field” is not as common.

Additionally, although GPS more commonly underestimated maximum decelerations during movement tasks, it is possible that motion capture itself did not get the true average deceleration during all tasks. Given how high intensity decelerations occur in a very short time window, the sampling rate of 120Hz may have been insufficient to capture true deceleration. However, the GPS and motion capture discrepancy could be larger in the present study. Sensitivity and specificity analyses in the present study should therefore be interpreted with caution given that 1) GPS units were used indoors, deviating from the recommended outdoor use, and 2) the analysis compared maximum deceleration derived from GPS to average deceleration derived from motion capture. Not utilizing equivalent metrics between systems could likely explain some of the discrepancy found within the present study.

The two primary limitations related to kinematic analysis are that the present study 1) did not collect kinetic data and 2) only examined sagittal plane kinematics for the ADA task. Firstly, while kinematic data offers practical value for coaches as they can visually watch players perform, having kinetic data would provide deeper insight into the internal demands of both horizontal decelerations and COD actions before and after an exercise intervention. Lastly, the only significant findings were in frontal plane hip kinematics during the 505 COD task, but it should be noted that the ADA task only examined sagittal plane movement. Further analysis of frontal and transverse plane kinematics across deceleration steps during the ADA would provide information on whether the accumulation of deceleration actions influenced other movement patterns that can be related to injury (i.e. knee valgus increasing ACL loading).

5.5 Future Directions

Future research should aim to replicate these findings in an outdoor field-based setting in order to determine if GPS sensitivity will improve with clear connection to GNSS satellites. Sensitivity and specificity may prove to have higher accuracy in correctly identifying deceleration actions into intensity zones with outdoor use. Once the study is repeated, researchers can visit the concept of connecting GPS absolute decelerations to biomechanical responses to repeated decelerations. Additionally, including separate groups for direct comparisons, such as elite athletes versus recreationally active or by sport type, could provide deeper insight into if deceleration techniques based on sport or true athletic skills could make a difference in kinematic results. Furthermore, future research analyzing all three planes of motion across both the ADA and 505 COD tasks would allow for a more direct comparison between task types. Notably, both task have different deceleration phase distances where the ADA is not a predetermined deceleration distance, whereas the 505 COD task is an established 5m to decelerate. Having a full kinematic analysis in all planes would allow for more in-depth movement pattern comparisons as well as the examine the influence of distance to decelerate and the interaction between these factors based on a protocol to elicit increased physical exertion. This could overall strengthen the understanding of specific deceleration skills and movement strategies used to overcome/compensate physical exertion during movements used in multi-directional sports.

CHAPTER 6: Conclusion

The aim of the present study was threefold: (1) evaluate GPS unit's ability to correctly classify deceleration actions into predefined absolute intensity zones using motion capture as the reference, (2) examine if accumulated high-intensity decelerations influence horizontal deceleration ability, and (3) examine whether accumulated decelerations influence lower extremity kinematics during standardized deceleration tasks. It was determined that GPS units provide low to moderate sensitivity in correctly identifying deceleration actions into their absolute intensity zone. As deceleration intensity increased, sensitivity decreased. The GPS units demonstrated moderate to high specificity, with increasing specificity from zone 3 through zone 6, indicating the GPS units were able to identify if a deceleration did not belong in a specific intensity zone.

Secondly, the exercise intervention of performing 15 trials of the 5-10-5 pro agility test to accumulate several decelerations through COD maneuvers, did not cause significant changes in deceleration abilities related to vertical pelvis position or number of steps to decelerate during the ADA task. Lastly, there was only a significant difference in hip abduction angle during the 505 COD task at initial contact and full weight acceptance from pre- to post- exercise intervention. All other kinematic variables for the ADA and 505 COD were non-significant for within-subjects changes. The present study provided insight into athlete performance monitoring during high intensity deceleration actions common in multidirectional sports, however when GPS units are utilized indoors, caution should be used when interpreting deceleration metrics. Lower extremity kinematic results will help sport scientists and coaching staff understand potential injury risk based on individual movement technique to decelerate during both horizontal deceleration and COD tasks.

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Appendices

Appendix A: IRB

EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board
Willis Building · Mail Stop 682
Office 252-744-2914 · Fax 252-744-2284 · redc@ecu.edu · redc.ecu.edu/umcib/

Notification of Initial Approval: Expedited

From: Biomedical IRB
To: [Selina Banks](#)
CC: [Patrick Rider](#)
Date: 12/16/2025
Re: [UMCIRB 25-001785](#)
Influence of Accumulated Decelerations on Lower Extremity Mechanics

I am pleased to inform you that your Expedited Application was approved. Approval of the study and any consent form(s) occurred on 12/15/2025. The research study is eligible for review under expedited category # 4,6,7. The Chairperson (or designee) deemed this study no more than minimal risk.

As the Principal Investigator you are explicitly responsible for the conduct of all aspects of this study and must adhere to all reporting requirements for the study. Your responsibilities include but are not limited to:

1. Ensuring changes to the approved research (including the UMCIRB approved consent document) are initiated only after UMCIRB review and approval except when necessary to eliminate an apparent immediate hazard to the participant. All changes (e.g. a change in procedure, number of participants, personnel, study locations, new recruitment materials, study instruments, etc.) must be prospectively reviewed and approved by the UMCIRB before they are implemented;
2. Where informed consent has not been waived by the UMCIRB, ensuring that only valid versions of the UMCIRB approved, date-stamped informed consent document(s) are used for obtaining informed consent (consent documents with the IRB approval date stamp are found under the Documents tab in the ePIRATE study workspace);
3. Promptly reporting to the UMCIRB all unanticipated problems involving risks to participants and others;
4. Submission of a final report application to the UMCIRB prior to the expected end date provided in the IRB application in order to document human research activity has ended and to provide a timepoint in which to base document retention; and
5. Submission of an amendment to extend the expected end date if the study is not expected to be completed by that date. The amendment should be submitted 30 days prior to the UMCIRB approved expected end date or as soon as the Investigator is aware that the study will not be completed by that date.

The approval includes the following items:

Name	Description
Banks Thesis Proposal	Study Protocol or Grant Application
Marx Activity Rating Scale	Surveys and Questionnaires
Marx Activity Rating Scale Past Month	Surveys and Questionnaires
Recruitment Email	Recruitment Documents/Scripts
Recruitment Flyer	Recruitment Documents/Scripts
Revised Informed Consent.pdf	Consent Forms

For research studies where a waiver or alteration of HIPAA Authorization has been approved, the IRB states that each of the waiver criteria in 45 CFR 164.512(i)(1)(i)(A) and (2)(i) through (v) have been met. Additionally, the elements of PHI to be collected as described in items 1 and 2 of the Application for Waiver of Authorization have been determined to be the minimal necessary for the specified research.

The Chairperson (or designee) does not have a potential for conflict of interest on this study.

IRB00000705 East Carolina U IRB #1 (Biomedical) IORG0000418
IRB00003781 East Carolina U IRB #2 (Behavioral/SS) IORG0000418

Note. UMCIRB approval notification (UMCIRB 25-001785), East Carolina University & Medical Center Institutional Review Board, dated 12/16/2025.

Appendix B: IRB Consent Form

East Carolina University



Informed Consent to Participate in Research

Information to consider before taking part in research that has no more than minimal risk.

Title of Research Study: Influence of Accumulated Decelerations on Lower Extremity Mechanics.

Principal Investigator: Selina Banks

Institution, Department or Division: East Carolina University, Department of Kinesiology

Address: 150 Minges Coliseum

Telephone #: 252-737-2370

Researchers at East Carolina University (ECU) study issues related to society, health problems, environmental problems, behavior problems and the human condition. To do this, we need the help of volunteers who are willing to take part in research.

Why am I being invited to take part in this research?

The purpose of this research is to compare measured deceleration efforts between Global Positioning System (GPS) units and motion capture as well as examine lower limb postures before and after a prescribed exercise circuit in recreationally active college females. You are being invited to take part in this research because you are a healthy, full-time female college student who is recreationally active. The decision to take part in this research is yours to make. By doing this research, we hope to learn if GPS and motion capture measure high deceleration efforts similarly and if the accumulation of high deceleration efforts influence lower limb postures.

If you volunteer to take part in this research, you will be one of about 20 people to do so.

Are there reasons I should not take part in this research?

If you are not a female, not a full-time college student, you are under the age of 18, or you have a current injury in your lower limbs or have had a lower limb injury in the last two weeks.

What other choices do I have if I do not take part in this research?

You can choose not to participate. If you wish to learn more about research in this area you are free to ask the P.I. for more information on the overall purpose of the study or about research within the area.

Where is the research going to take place and how long will it last?

The research will be conducted at the Murphy Center on the Grady-White Boats Athletic Campus. You will need to come to Minges Coliseum, room 63, one time during the study. The total amount of time you will be asked to volunteer for this study is approximately one to 1.5 hours.

What will I be asked to do?

You will be asked to do the following:

- Complete the Marx Activity Rating Scale, which is a four-question survey asking how often you participate in movements similar to those seen in different exercises.
- Dynamic Warm-up: Complete a series of movements and four brief 5-meter sprints to prepare your body for deceleration tasks.

Consent Version # or Date: _____

- a. Protocol walkthrough: you will be walked through the deceleration tasks to be performed and allowed one practice attempt to get comfortable with the tasks before data collection.
- Pre-test Assessment: Three trials each of double leg squat, self-selected pace walk, Acceleration-Deceleration Assessment, and 505 Change of Direction Test.
 - a. Acceleration-Deceleration Assessment includes maximally sprinting 20 meters before coming to a rapid and controlled full stop.
 - b. 505 Change of Direction Test includes sprinting 10 meters into a complete (180°) change of direction, followed by another 5-meter sprint.
- Exercise Circuit (intervention): Fifteen trials of the 5-10-5 Agility test, which consist of both 5- and 10-meter sprints with complete (180°) change of direction movements. Between each trial of the 5-10-5 Agility test you will complete a 30 second jog.
- Post-test Assessment: Three trials each of Acceleration-Deceleration Assessment and 505 Change of Direction Test.

What might I experience if I take part in the research?

We don't know of any risks (the chance of harm) associated with this research. Any risks that may occur with this research are no more than what you would experience in everyday life. We don't know if you will benefit from taking part in this study. There may not be any personal benefit to you, but the information gained by doing this research may help others in the future.

Will I be paid for taking part in this research?

We will not be able to pay you for the time you volunteer while being in this study.

Will it cost me to take part in this research?

It will not cost you any money to be part of the research.

Who will know that I took part in this research and learn personal information about me?

ECU and the people and organizations listed below may know that you took part in this research and may see information about you that is normally kept private. With your permission, these people may use your private information to do this research:

- Any agency of the federal, state, or local government that regulates human research. This includes the Department of Health and Human Services (DHHS), the North Carolina Department of Health, and the Office for Human Research Protections.
- The University & Medical Center Institutional Review Board (UMCIRB) and its staff have responsibility for overseeing your welfare during this research and may need to see research records that identify you.

How will you keep the information you collect about me secure? How long will you keep it?

Your data will be stored electronically on a secured server. This data will not contain any information that could be used to identify you. You will be assigned an ID number, and this number will be used to identify any of your data files. The spreadsheet linking you to your ID number will be kept in a locked office by the principal investigator. Video data collected with motion capture will be secured in a password protected file, in a restricted folder behind the university firewall. All of your data will be kept for a minimum of 6 years after study completion.

What if I decide I don't want to continue in this research?

You can stop at any time after it has already started. There will be no consequences if you stop and you will not be criticized. You will not lose any benefits that you normally receive.

Title of Study: Influence of Accumulated Decelerations on Lower Extremity Mechanics

Who should I contact if I have questions?

The people conducting this study will be able to answer any questions concerning this research, now or in the future. You may contact the Principal Investigator through email (bankss20@students.ecu.edu) during normal business hours (M-F, between 9am and 5pm).

If you have questions about your rights as someone taking part in research, you may call the ECU University and Medical Center Institutional Review Board (UMCIRB) at phone number 252-744-2914 (days). If you would like to report a complaint or concern about this research study, you may call the Director for Human Research Protections, at 252-744-2914.

If you have questions about the sharing of PHI related to this research study, call Patrick Rider at 252-737-2370. If you have questions about your rights as someone taking part in research, you may call the ECU University and Medical Center Institutional Review Board (UMCIRB) at phone number 252-744-2914 (days). If you would like to report a complaint or concern about this research study, you may call the Director for Human Research Protections, at 252-744-2914. In addition, if you have concerns about confidentiality and privacy rights, you may phone the Privacy Officer at East Carolina University at 252-744-5200

Is there anything else I should know?

Identifiers might be removed from the identifiable private information or identifiable biospecimens and, after such removal, the information or biospecimens could be used for future research studies or distributed to another investigator for future research studies without additional informed consent from you or your Legally Authorized Representative (LAR). However, there still may be a chance that someone could figure out the information is about you.

I have decided I want to take part in this research. What should I do now?

The person obtaining informed consent will ask you to read the following and if you agree, you should sign this form:

- I have read (or had read to me) all of the above information.
- I have had an opportunity to ask questions about things in this research I did not understand and have received satisfactory answers.
- I know that I can stop taking part in this study at any time.
- By signing this informed consent form, I am not giving up any of my rights.
- I have been given a copy of this consent document, and it is mine to keep.

Participant's Name (PRINT)

Signature

Date

Person Obtaining Informed Consent: I have conducted the initial informed consent process. I have orally reviewed the contents of the consent document with the person who has signed above, and answered all of the person's questions about the research.

Person Obtaining Consent (PRINT)

Signature

Date

Page 3 of 3

Consent Version # or Date: _____

Appendix C: MARX Activity Scale

Marx Activity Rating Scale

Please indicate how often you performed each activity in your healthiest and most active state, in the past year

	Less than 1 time in a month	One time in a month	One time in a week	2-3 times in a week	4 or more times in a week
Running: running while playing a sport or jogging					
Cutting: changing directions while running					
Decelerating: coming to a quick stop while running					
Pivoting: turning your body with your foot planted while playing a sport; For example: skiing, skating, kicking, throwing, hitting a ball					

Appendix D: Sensitivity and Specificity calculations

Table 8: Zone 1 Sensitivity and Specificity Confusion Matrix

Variable	MOCAP Positive	MOCAP Negative	Total
GPS Positive	12	18	30
GPS Negative	5	152	157
Total	17	170	187
Sensitivity & Specificity Calculation	$12/17 = 0.706$	$152/170 = 0.894$	

Table 9: Zone 2 Sensitivity and Specificity Confusion Matrix

Variable	MOCAP Positive	MOCAP Negative	Total
GPS Positive	12	50	62
GPS Negative	9	116	125
Total	21	166	187
Sensitivity & Specificity Calculation	$12/21 = 0.571$	$116/166 = 0.699$	

Table 10: Zone 3 Sensitivity and Specificity Confusion Matrix

Variable	MOCAP Positive	MOCAP Negative	Total
GPS Positive	30	33	63
GPS Negative	35	89	124
Total	65	122	187
Sensitivity & Specificity Calculation	$30/65 = 0.462$	$89/122 = 0.730$	

Table 11: Zone 4 Sensitivity and Specificity Confusion Matrix

Variable	MOCAP Positive	MOCAP Negative	Total
GPS Positive	6	16	22
GPS Negative	42	123	165
Total	48	139	187
Sensitivity & Specificity Calculation	$6/48 = 0.125$	$123/139 = 0.885$	

Table 12: Zone 5 Sensitivity and Specificity Confusion Matrix

Variable	MOCAP Positive	MOCAP Negative	Total
GPS Positive	0	8	8
GPS Negative	31	148	179
Total	31	156	187
Sensitivity & Specificity Calculation	$0/31 = 0.00$	$148/156 = 0.949$	

Table 13: Zone 6 Sensitivity and Specificity Confusion Matrix

Variable	MOCAP Positive	MOCAP Negative	Total
GPS Positive	0	2	2
GPS Negative	1	184	185
Total	1	186	187
Sensitivity & Specificity Calculation	$0/1 = 0.00$	$184/186 = 0.989$	

Appendix E: Results from Exercise Intervention

Table 14: Participant's Rate of Perceived Exertion (RPE) scores after each repetition of the exercise intervention

Subject	Rep 1	Rep 2	Rep 3	Rep 4	Rep 5	Rep 6	Rep 7	Rep 8	Rep 9	Rep 10	Rep 11	Rep 12	Rep 13	Rep 14	Rep 15
001	5	5	5	6	5	6	6	6	6	6	6	6	7	6	6
002	2	3	3	3	4	4	4	4	4	5	5	5	5	6	6
003	1	2	3	3	4	4	5	5	6	6	7	7	7	7	7
004	1	2	2	2	3	3	3	3	3	3	3	3	4	3	3
005	2	2	2	2	3	3	3	3	3	3	3	3	4	4	4
006	1	1	2	2	2	3	3	3	4	4	4	4	5	5	5
007	2	2	2	3	3	3	3	4	4	4	5	5	5	5	6
008	3	3	3	3	4	4	4	4	5	5	5	6	7	8	9
009	3	3	3	4	4	4	4	3	3	3	3	4	4	4	4
010	2	3	4	5	5	6	6	7	7	7	7	8	8	8	9
011	4	5	6	7	7	8	8	7	8	8	8	8	8	7	7
012	3	4	6	8	9	9	9	9	8	9	10	9	8	8	8
013	0	0	0	0	1	2	2	3	3	2	3	4	2	3	3
014	1	1	1	2	3	3	3	3	4	4	5	4	5	5	6
015	3	5	6	5	6	7	7	7	8	7	8	8	8	7	9

Table 15: Participant Completion Times for each Repetition of the Exercise Intervention

Subject	Rep 1 (s)	Rep 2 (s)	Rep 3 (s)	Rep 4 (s)	Rep 5 (s)	Rep 6 (s)	Rep 7 (s)	Rep 8 (s)	Rep 9 (s)	Rep 10 (s)	Rep 11 (s)	Rep 12 (s)	Rep 13 (s)	Rep 14 (s)	Rep 15 (s)
001	6.74	6.49	6.61	6.8	6.68	6.84	6.74	6.53	6.92	6.75	6.80	6.78	6.93	6.81	6.61
002	6.5	6.75	6.59	6.39	6.58	6.81	7.22	6.89	7.09	7.15	6.9	7.01	6.93	7.03	6.97
003	6.38	6.32	6.53	6.5	6.22	6.4	6.31	6.46	6.51	6.39	6.53	6.59	6.64	6.47	6.47
004	6.09	6	6.01	6.02	6.02	5.84	5.97	5.77	5.82	5.9	5.83	5.89	5.95	5.88	5.84
005	6.46	6.5	6.69	6.4	6.34	6.6	6.5	6.63	6.44	6.63	6.45	6.69	6.44	6.46	6.58
006	5.97	6.15	5.96	5.91	6.14	6.16	6.07	6.5	5.77	6.16	6.02	6.4	5.75	6.01	6.07
007	7.83	7.82	7.62	7.94	8.34	7.2	7.45	7.64	7.85	7.45	8.07	8.21	7.59	8.09	8.03
008	6.9	7.07	6.76	7.15	7.07	6.56	6.9	6.88	7.28	6.72	6.7	6.71	6.88	6.97	6.95
009	6.09	6.15	6.08	6.21	6.02	6.07	6.1	5.96	6.06	5.97	6.0	6.2	6.07	6.14	6.01
010	7.56	7.53	8.21	7.28	7.75	7.59	7.82	8.21	8	7.89	7.89	7.83	7.56	8.1	7.64
011	7.38	7.95	8.25	7.83	7.77	7.71	7.66	7.88	7.7	7.91	7.68	8.03	7.83	7.82	7.33
012	6.95	6.95	7.44	7.9	7.64	7.89	7.38	7.46	7.47	7.25	7.45	7.25	7.43	7.22	7.15
013	7.84	8.14	8.09	8.14	8.14	8.34	8.27	7.95	7.82	8.14	7.33	7.96	7.45	7.53	7.46
014	6.56	6.88	6.39	6.44	7.08	6.25	6.7	6.2	6.46	7.03	6.52	6.58	6.33	6.38	6.52
015	7.9	8.39	8.14	7.89	7.69	7.82	7.71	8.65	9.28	9.21	9.09	8.56	8.31	8.51	8.75

Appendix F: Average ADA Entrance Velocity Comparisons

Table 16: Average maximum velocity entering the ADA task for each participant pre- and post-exercise intervention

Subject	Average Velocity Entering ADA Pre Intervention	Average Velocity Entering ADA Post Intervention
001	5.58	5.72
002	5.29	5.17
003	6.39	6.79
004	6.78	6.57
005	5.96	5.71
006	7.16	6.99
007	5.62	5.69
008	6.20	6.39
009	6.52	6.44
010	4.56	4.27
011	5.05	4.92
012	5.97	5.64
013	5.19	4.71
014	5.31	5.55
015	5.66	5.26

Appendix G: Individual Movement Strategy Linear Regression Plots

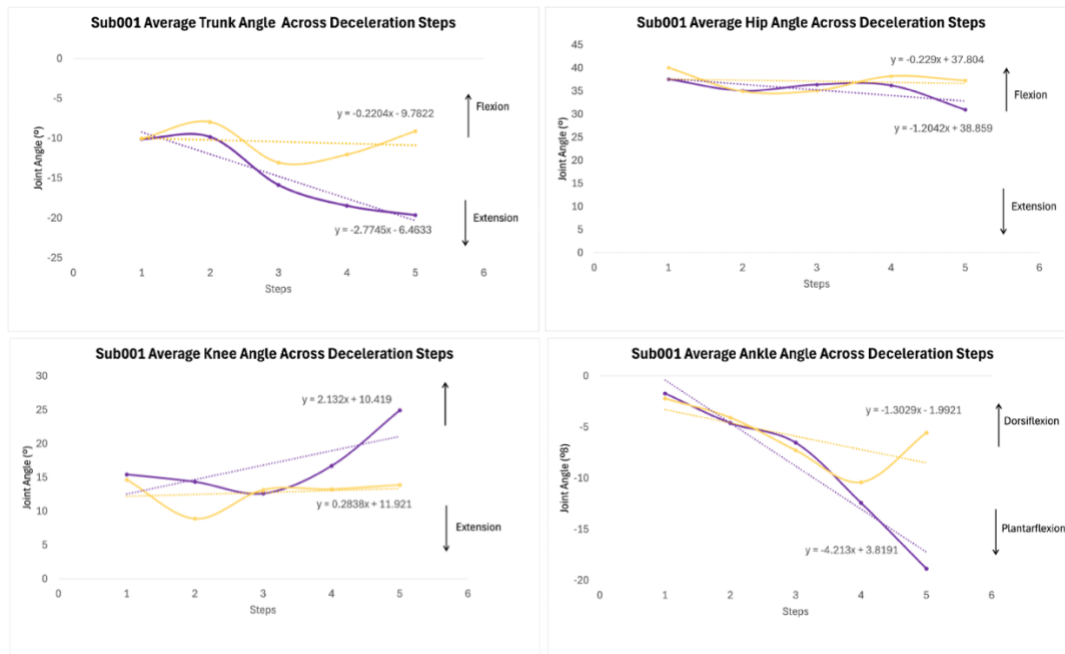


Figure 9: Subject 1 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

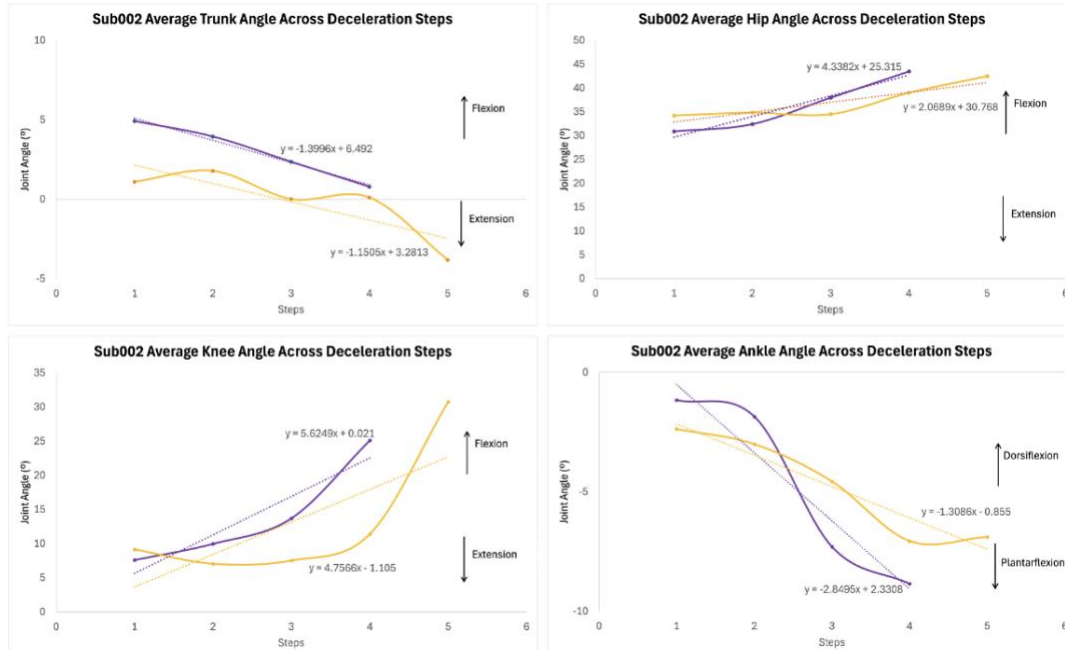


Figure 10: Subject 2 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

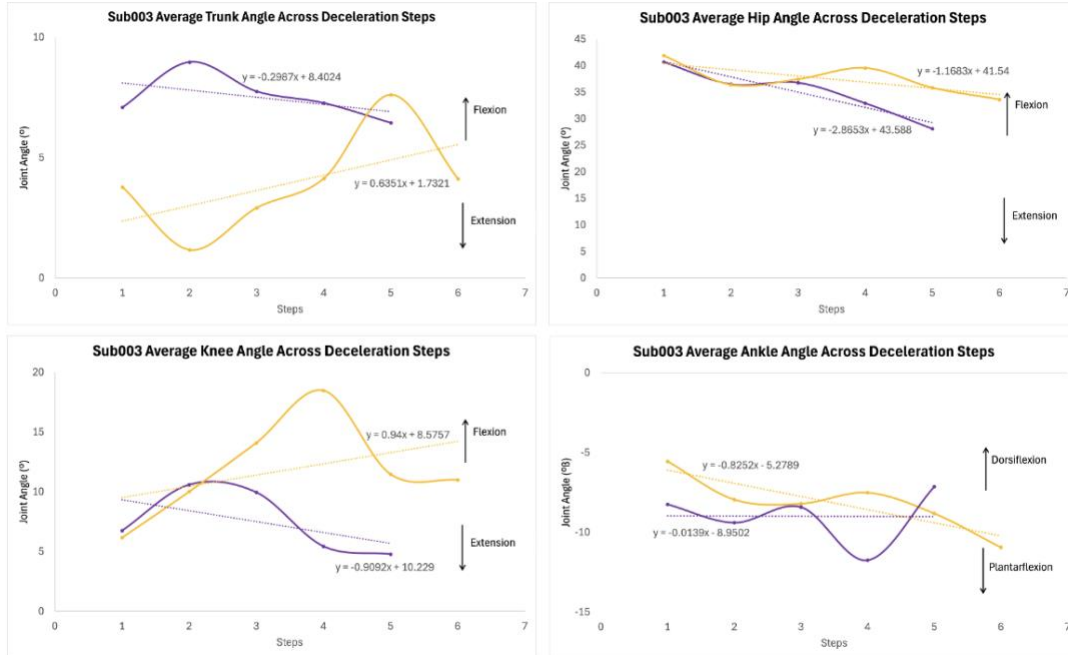


Figure 11: Subject 3 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

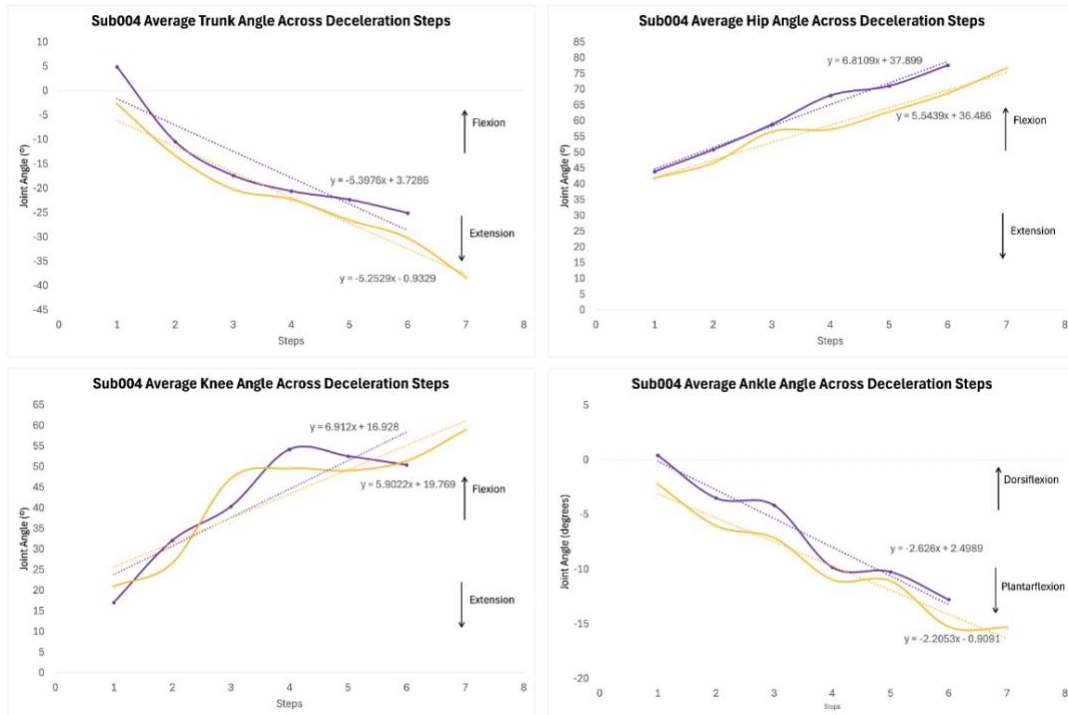


Figure 12: Subject 4 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

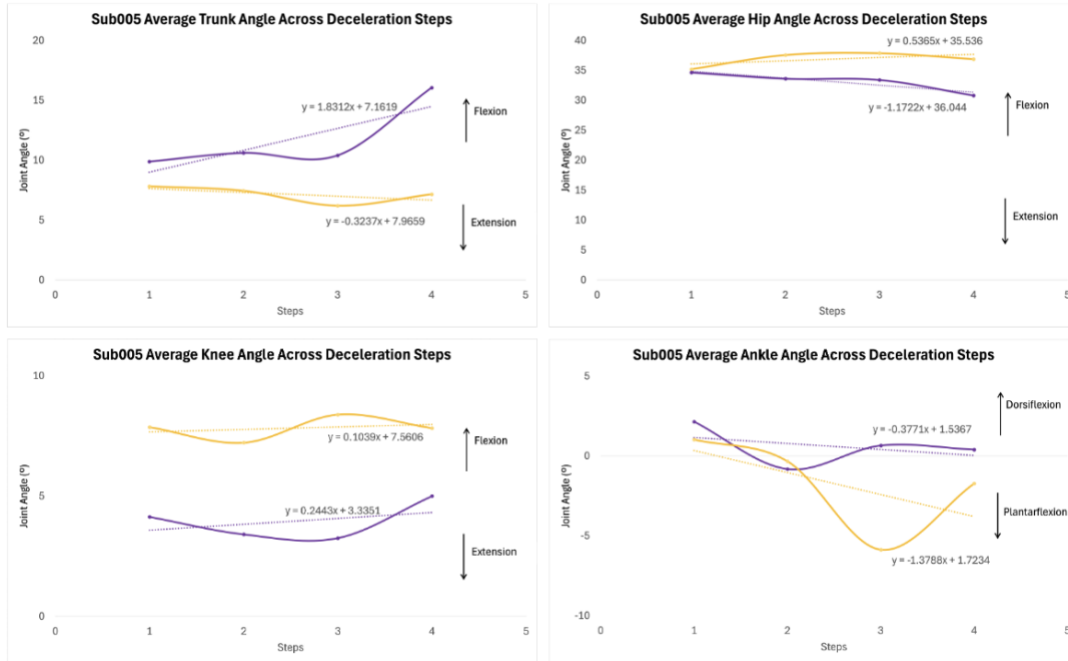


Figure 13: Subject 5 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

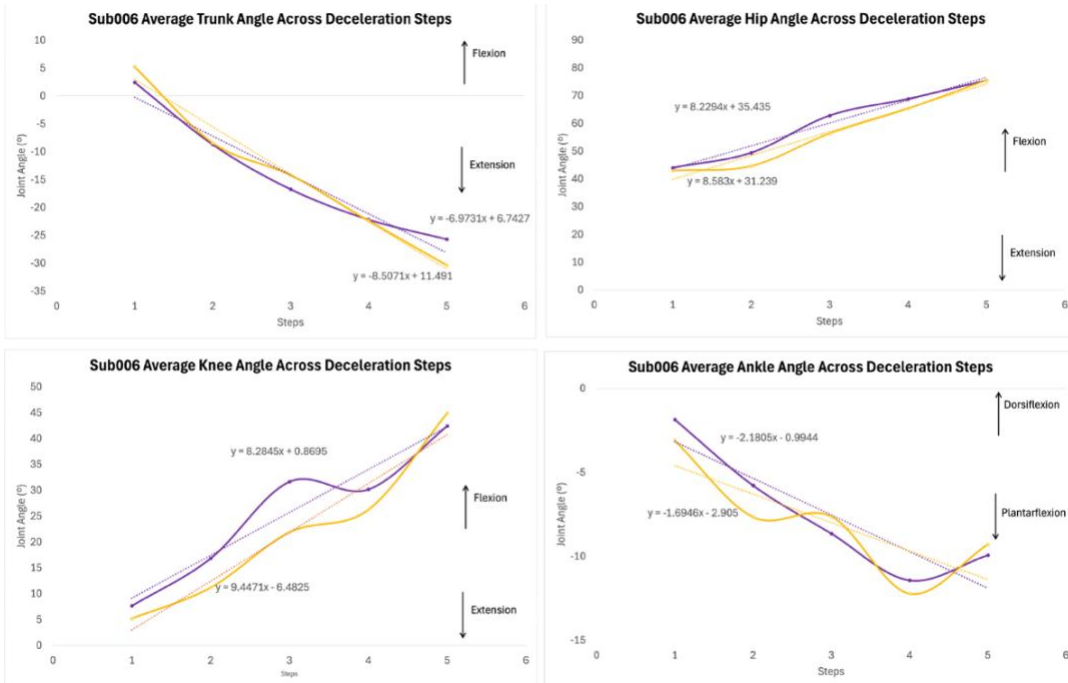


Figure 14: Subject 6 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

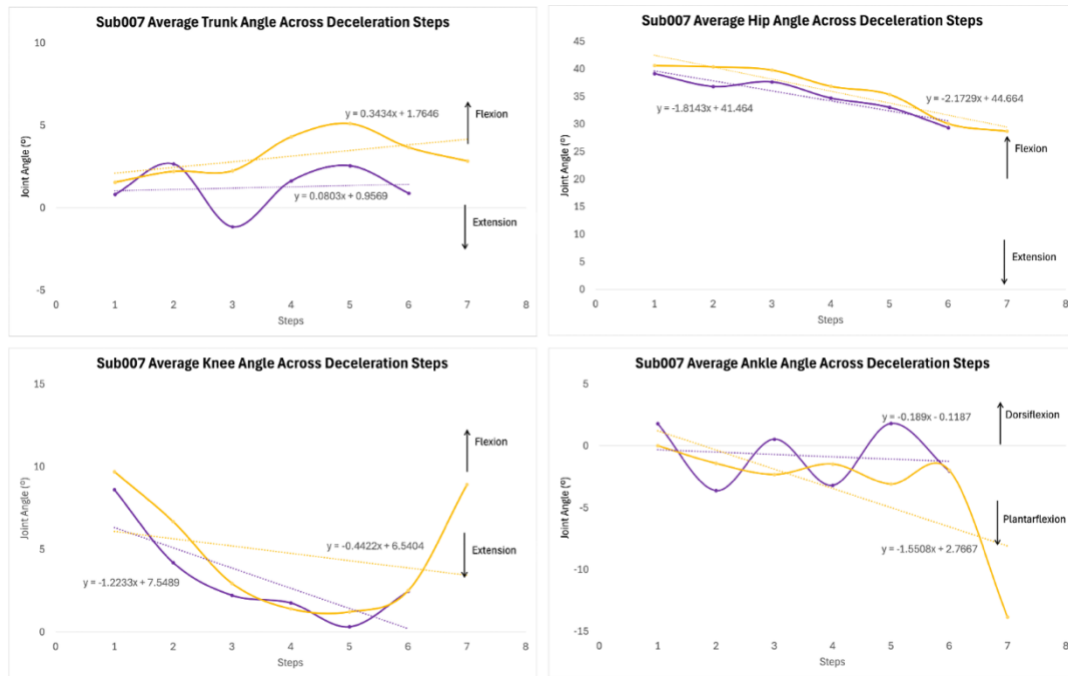


Figure 15: Subject 7 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

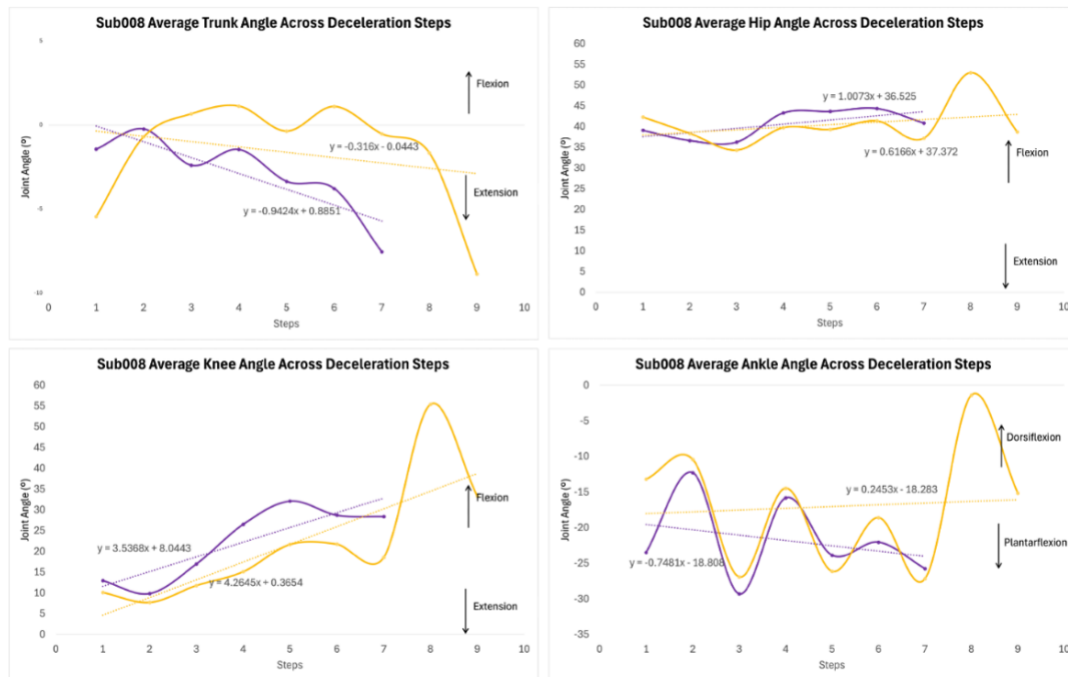


Figure 16: Subject 8 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

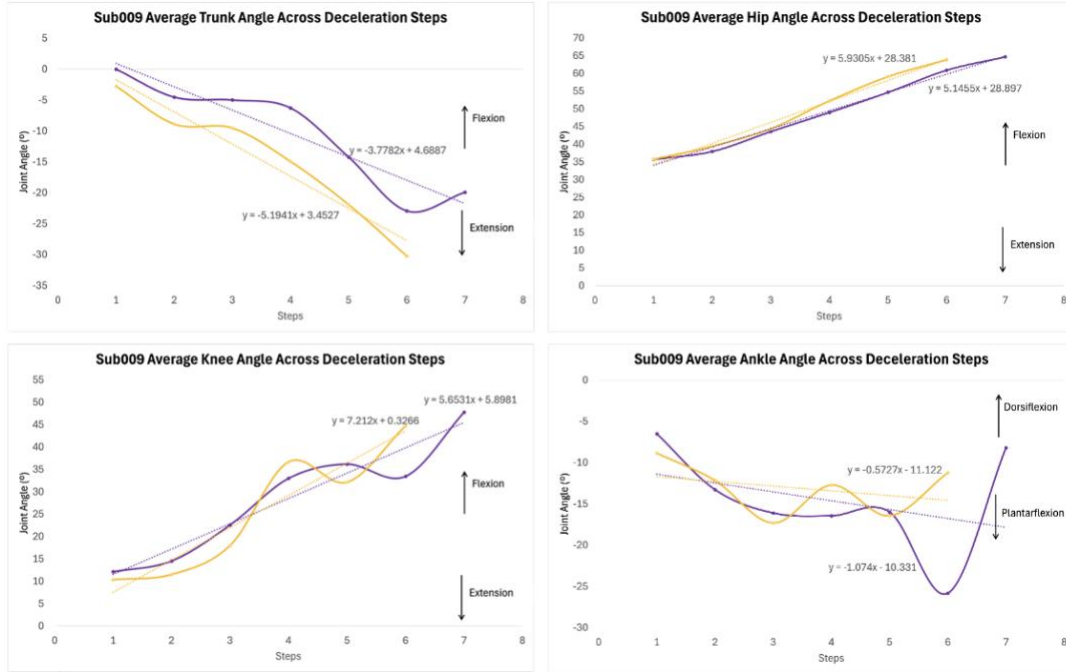


Figure 17: Subject 9 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

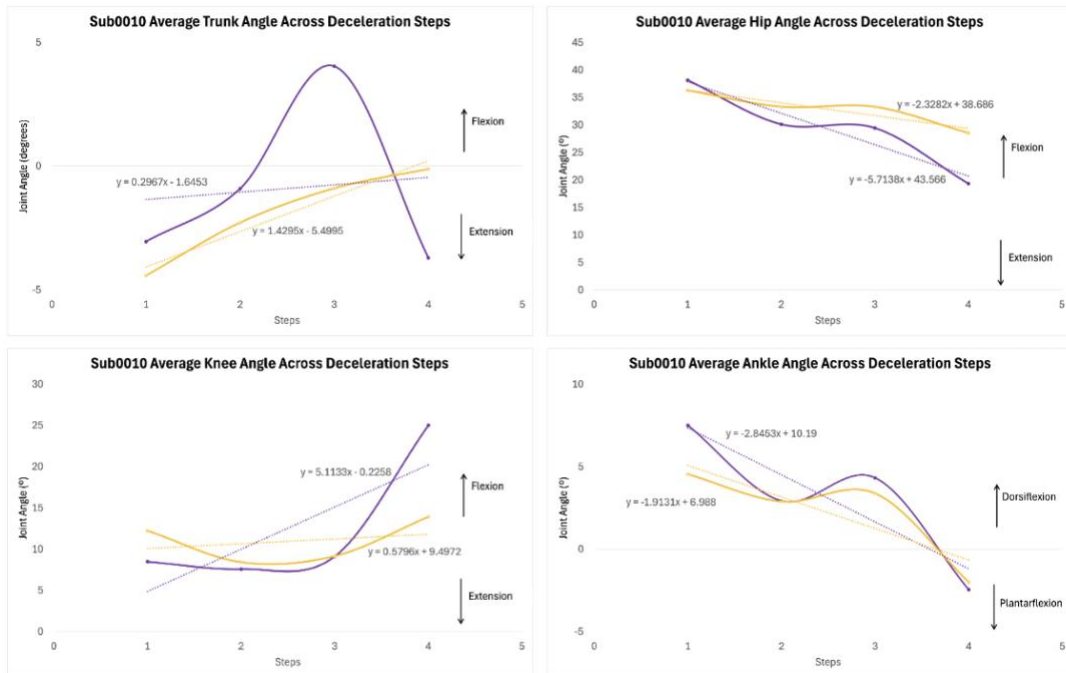


Figure 18: Subject 10 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

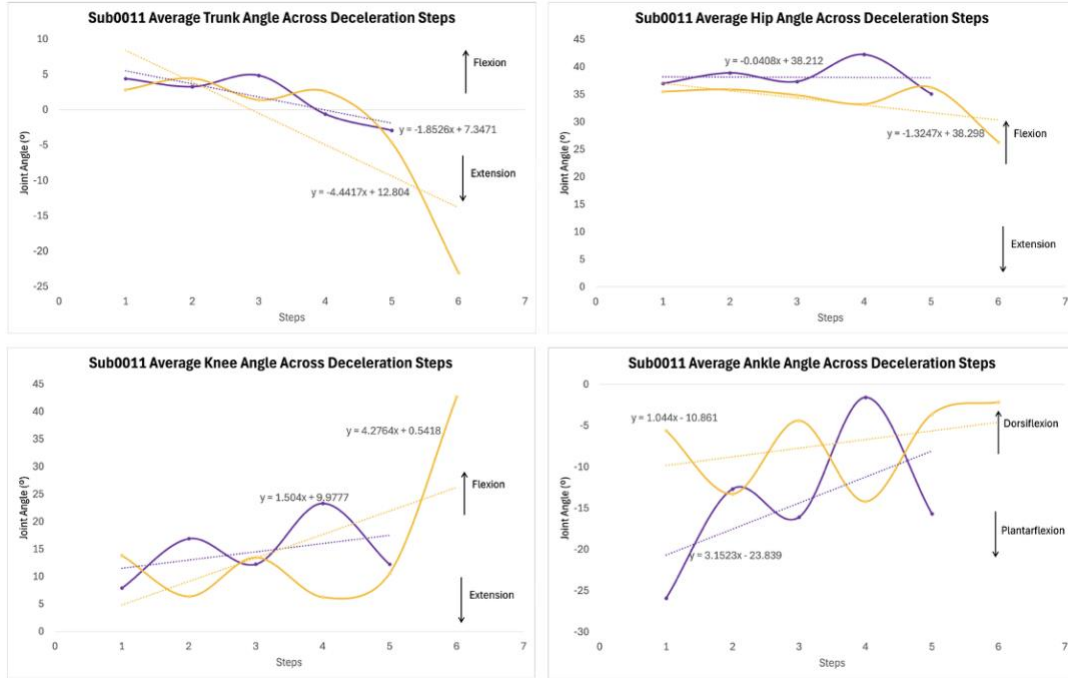


Figure 19: Subject 11 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

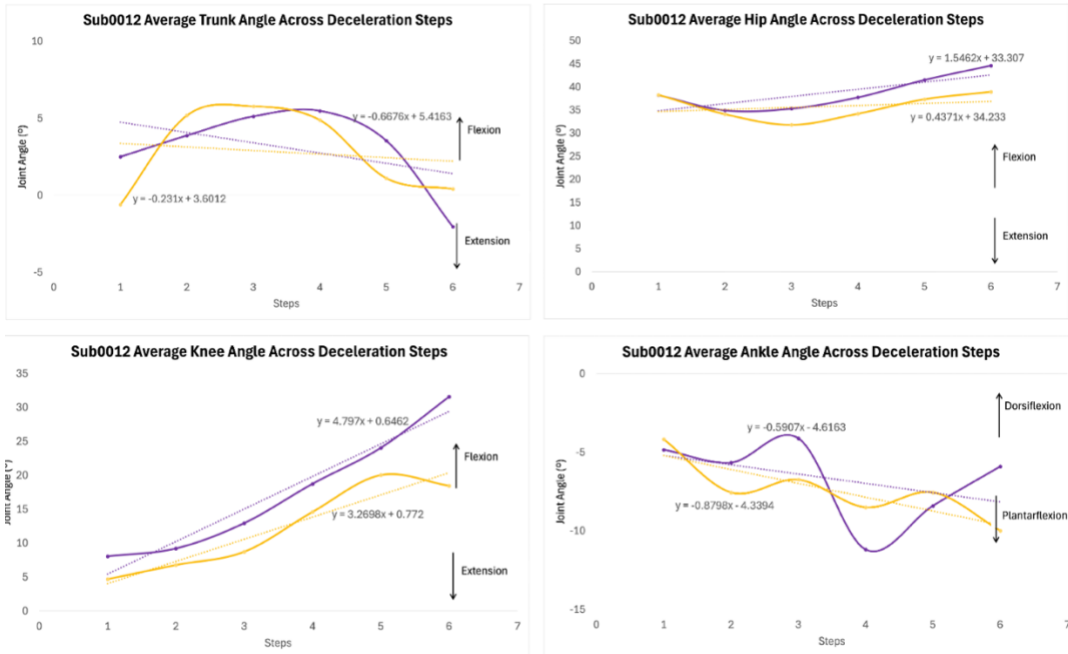


Figure 20: Subject 12 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step

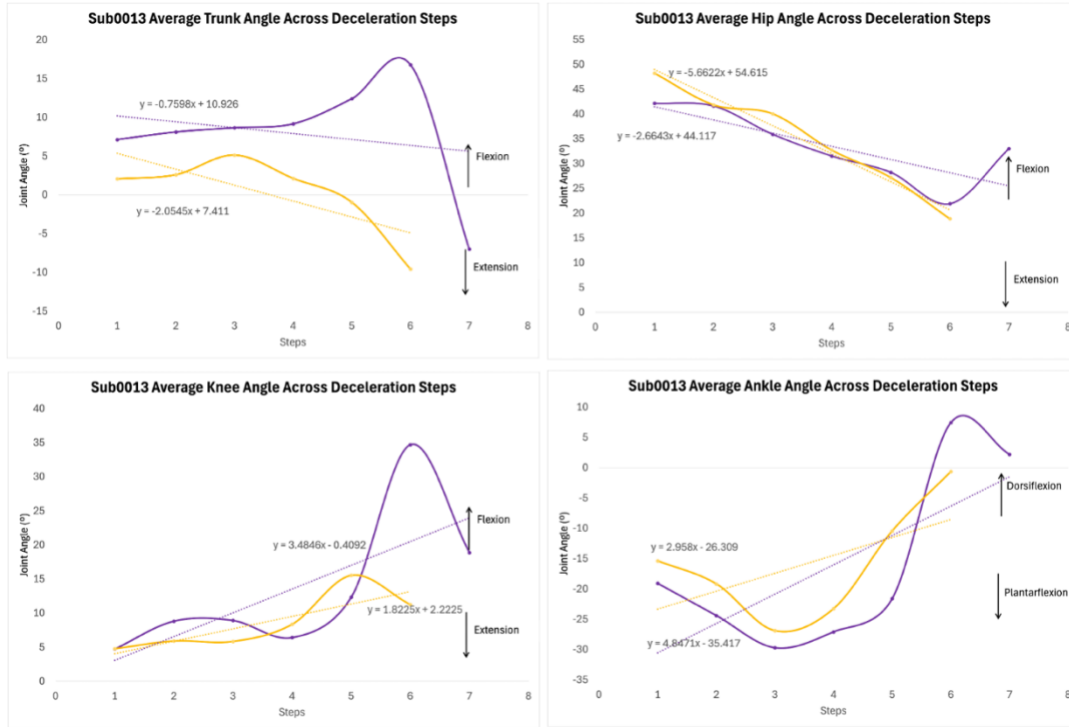


Figure 21: Subject 13 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy.

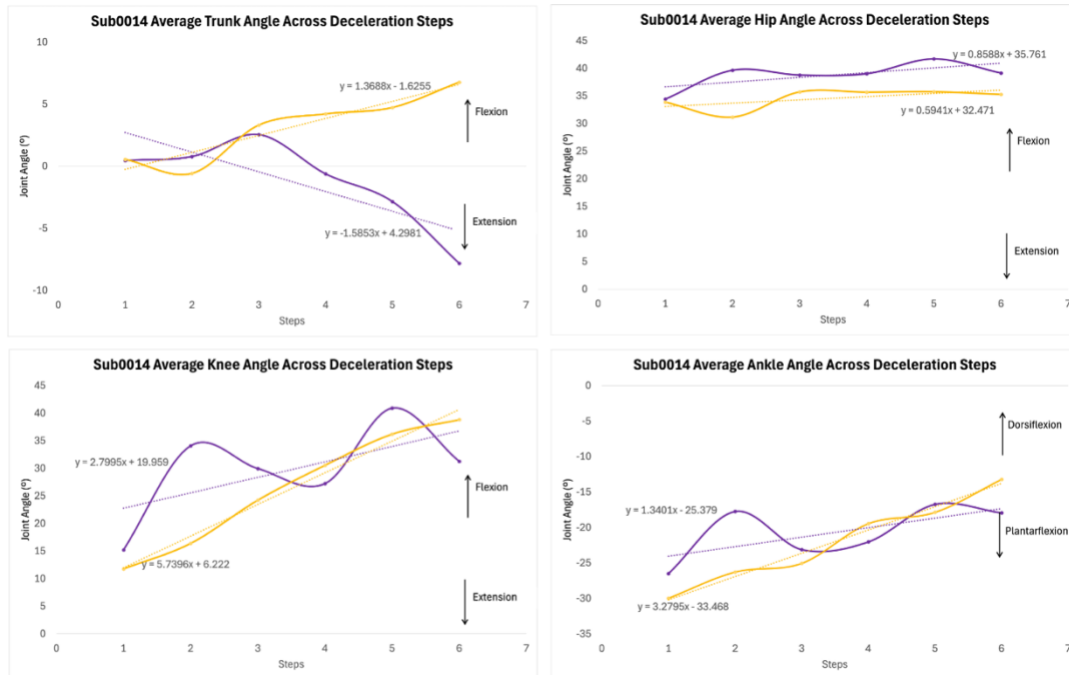


Figure 22: Subject 14 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step.

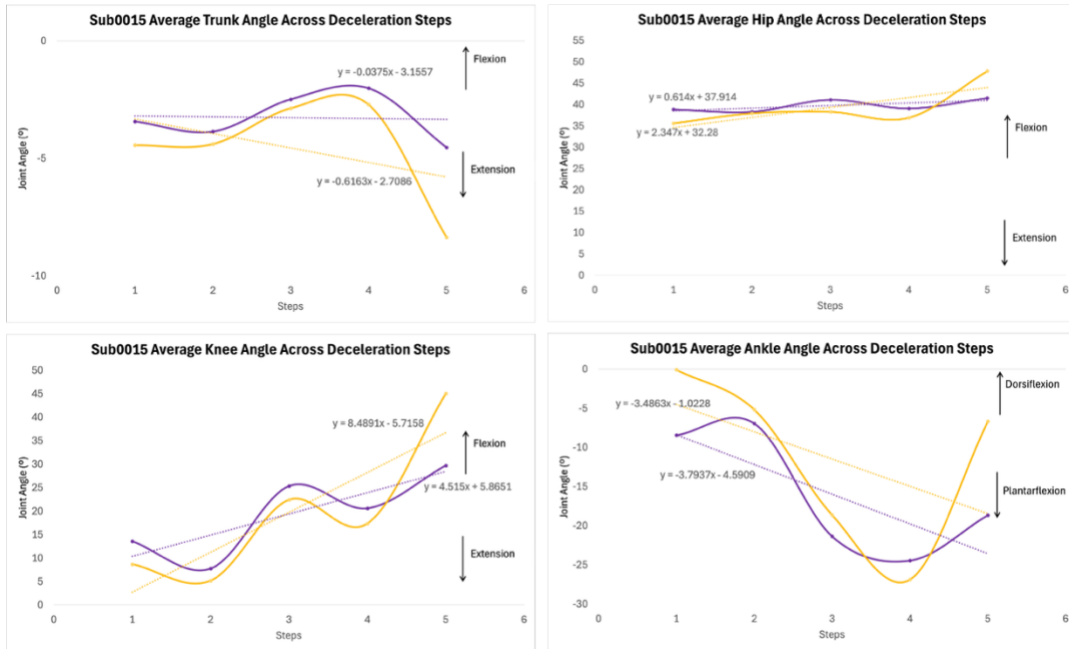


Figure 23: Subject 15 sagittal plane joint angles for trunk (top left), hip (top right), knee (bottom left), and ankle (bottom right) at initial contact of consecutive deceleration steps during the ADA task. Purple solid line shows pre-intervention and yellow solid line shows post-intervention movement strategy. Dotted lines are linear regressions showing the change in joint angle (slope) per deceleration step