International Journal of Strength and Conditioning

Decker, A., Fagley, J., McNally, S., Regalski, Z., Rhinehart, K., Urda, A., Holmstrup, M., & Dicus, J. (2026). A Comparative Biomechanical Analysis of the Farmers and Zercher Carry Exercises. International Journal of Strength and Conditioning
https://doi.org/10.47206/ijsc.v6i1.566

A Comparative Biomechanical Analysis of the Farmers and Zercher Carry Exercises

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ABSTRACT

The loaded carry is a movement pattern characterized by walking for a set time or distance with an externally applied resistance. The application of an external load alters the body's biomechanics, creating an environment for potential adaptation and performance enhancement. Further, the location of the load may have additional adaptation implications as a result of center of mass (CoM) changes which may impact body stability. Specifically, a resistance carried anterior to the body as done in the Zercher Carry (ZC), causing a vertical and anterior translation of the CoM, may stimulate different compensatory mechanisms from the Farmers Carry (FC) where the load is held at the sides of the body, keeping the CoM in its relative anatomical location. To investigate this, healthy, college aged individuals were recruited and their anthropometric data was captured. Participants were then prepped for surface electromyography with electrodes placed on their dominant side rectus abdominis, external oblique, multifidus, longissimus, upper trapezius, and middle deltoid. Maximal voluntary isometric contractions were performed to standardize muscle activation. Participants were outfitted with inertial measurement units for the trunk and lower limb to capture gait and joint kinematics. One counterbalanced 5-meter walk loaded with 100% fat free mass was completed for each exercise with participants stepping on a force plate halfway to capture single step ground reaction forces (GRF). Data for all variables were compared utilizing a MANOVA. Comparisons between the FC and ZC revealed significant differences ($P < 0.05$) in muscle activation (e.g., multifidus: FC 12.0 ± 6.7 v. ZC 25.4 ± 13.0), joint kinematics (e.g., lumbar extension: FC -4.4 ± 4.1 v. ZC 2.0 ± 6.5), gait (e.g., stride length: FC 108.1 ± 9.4 v. ZC 101.4 ± 9.4) and GRFs (e.g., anterior GRF: FC 22.3 ± 6.8 v. ZC 17.5 ± 7.4). The differences observed between the ZC and FC as a result of load position may implicate unique adaptation and influence how these exercises should be implemented in resistance training programs.

INTRODUCTION

The loaded carry is a movement pattern commonly implemented in resistance training programs. Characterized by traversing over a surface for a time or distance with an externally applied resistance, the loaded carry offers the ability to elicit a stimulus that can affect both the muscles and mechanics of the body (1). Specifically, the application of an external load as the body engages in locomotion requires the activation of multiple muscles over multiple joints with notable activation observed in the core (3). These core muscles are utilized to create stability of the spine and pelvis as well as to create the transfer of torque and momentum for maximal force generation during activity (3,4). As such, the durability of these muscles may have implications on training and athletic performance as supported by previous literature which has observed increases in performance levels through core strengthening interventions (5,6). Provided chronic training and

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progressive overload are executed, the loaded carry may create an environment for specific adaptations as seen in other resistance training exercises, which may explain its implementation in resistance training programs (7,8).

Like other resistance training movement patterns, the loaded carry can be manipulated to alter its stimulus. One common method of manipulation is achieved by altering the position of the externally applied resistance in relation to the body. This gives rise to many specific exercise variations including the Farmers Carry (FC) and the Zercher Carry (ZC). In the FC, the implement is carried in both hands at the sides, placing the load directly in line with the body (Figure 1). Alternatively, in the ZC, the implement is carried in the bend of the arms, placing the load higher and more anterior relative to the FC (Figure 2). This load position difference between the FC and ZC may have implications on the effects these exercises will have on body biomechanics due to the shift in the body's center of mass (CoM). A shift in CoM from its anatomical position may alter the stability of the body as it changes how the mass is distributed within the base of support (BoS). As such, the higher, more anterior load of the ZC compared to the vertically aligned load of the FC may indicate the ZC to be a less stable exercise and therefore may implicate the biomechanical differences observed between the exercises.

Of the loaded carry variations, the FC is more



A B

Figure 1. Sagittal (A) and oblique (B) view of the Farmers Carry



A B

Figure 2. Sagittal (A) and oblique (B) view of the Zercher Carry

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commonly analyzed, with several studies examining the biomechanics of the exercise. In one study, Koegh et. al. examined gait and joint kinematic differences at different walking speeds across multiple FC walks and at three different stages (i.e., 0-3m, 8.5-11.5m, 17-20m) within a single FC walk. Analysis of these variables across different speeds revealed greater stride lengths and rates as well as shorter ground contact times in a between subject analysis, although these were not significant within subjects. Analysis of the different stages across participants revealed significantly greater average velocities, stride lengths and stride rates, reduced ground contact times, as well as greater range of motion at the thigh and knee at the middle and final stages of the walk, in comparison to the first stage. In another study, McGill et. al. performed an analysis of common Strongman events, including the FC, utilizing surface electromyography (EMG) on several core and pelvic muscles (e.g., rectus abdominis, internal and external obliques, erector spinae, rectus femoris, etc.) as well as 3-dimensional spine motion for the purpose of capturing spine loading and stiffness (10). EMG data revealed increased abdominal muscle activation (i.e., rectus abdominis, internal and external obliques) during the walking phase of the FC. Further, a direct comparison of the FC to the Suitcase Carry (i.e., implement held unilaterally at the side of the body) and the Yoke Walk (i.e., implement held bilaterally on the posterior aspect of the shoulders) revealed similar spine stiffness in the bilateral carries, although, interestingly, no differences were reported between the FC and Yoke Walk. A comparison of the FC to an unloaded walk was additionally performed by Winwood et. al. who collected and compared data on gait and joint kinematics as well as kinetics for both conditions (11). Kinematic findings of the FC from this study included greater anterior tilt of the trunk as well as greater flexion angles of the ankle, knee, and hip at foot strike while kinetic findings of the FC included increased peak and mean vertical, anterior, posterior, and medial ground reaction forces.

To our knowledge, no analyses of the kinematics or kinetics of the ZC as an exercise have been executed. However, several studies investigating the effects of anterior loading on body biomechanics have been performed. Anderson et. al performed an analysis of anterior barbell and bucket load carriage, comparing the muscle activation and sagittal plane trunk kinematics between different anterior loading heights in a static versus dynamic condition (12). EMG data revealed increases in muscle activation of all muscles during locomotion with both implements, however, height changes did not elicit significant increases in anterior core muscle activity. Further, both conditions saw increases in trunk extension with increasing load height. Similar kinematic findings were observed by Alamoudi et. al. who examined the differences in locomotor stability, gait spatiotemporal measures, and spine biomechanics in different carrying positions with changing load magnitudes (13). Comparisons between each carry showed the front-loaded position had the lowest gait stability, the highest cadence increase, and a stride length that decreased at the greatest load magnitude. Analysis of the spine biomechanics further found the front load position elicited the highest amount of trunk extension that increased with increasing load as well as significant lateral bending differences from the bilateral and posterior conditions. The bilateral condition, which was performed similarly to the FC, was noted to be the most stable condition, implicating stability differences between the loaded conditions.

Although examinations of the loading conditions of the FC and ZC have been performed, much is still unknown regarding the effects of these exercises on the biomechanics of the body. Further, a comparison with the concurrent reporting of muscle activation, gait and joint kinematics, and kinetics between the FC and ZC has yet to be examined in regard to resistance training and performance. As such, the purpose of this study was to quantify the biomechanical differences between the FC and ZC, allowing us to accentuate the potential implications these exercises may have on the body in a resistance training setting. This analysis may provide exercise professionals with a deeper understanding of these loaded carry movement patterns, equipping them with the ability to implement them into training programs for the purpose of inducing specific biomechanical and physiological adaptations.

METHODS

Participants

Data were collected on twenty-five participants. Due to technical issues with hardware, nineteen (12 female, 7 male; age: 19.21 ± 0.85, height: 167.03 ± 7.5 cm, weight: 72.36 ± 10.97 kg) participants yielded usable data across all variables used for statistical analysis. Participants were required to be at least 18 years of age and were excluded if

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they had any current musculoskeletal injuries or disorders.

Procedures

All participants attended one data collection session and were asked to maintain their normal eating and sleeping habits and to refrain from engaging in exercise on the day of their session. After participants were formally consented, their anthropometric data were acquired. Height (cm) was measured using a stadiometer and body mass (kg) and fat mass (FM) were measured using the Inbody 970. Fat free mass (FFM) was calculated (body mass - FM) and used to determine carry load.

Following a 3-minute warm-up and unweighted practice carries, surface Ag/AgCl dual electrodes with a 2 cm interelectrode distance were placed on the dominant side rectus abdominis (RA), external oblique (EO), multifidus (MT), longissimus (LS), upper trapezius (UT), and middle deltoid (MD). The skin for each site was shaved, cleaned, and abraded following standard preparation procedures and all sensors were placed according to previous published study methods (14,15,16,17). Specifically, the RA sensor was placed vertically, 3 cm lateral and 3 cm superior to the center of the umbilicus (14). The EO sensor was placed diagonally midway between the bottom of the rib cage and the anterior superior iliac spine (14). The MT sensor was placed diagonally, 3 cm lateral to L5 in a diagonal line with the posterior-superior iliac spine and L1 (15). The LS sensor was placed vertically, 3 cm lateral of L1 (15). The UT sensor was placed midway between C7 and the acromion process of the scapula (16). The MD sensor was placed midway between the acromion process of the scapula and the deltoid tuberosity of the humerus over the most prominent portion of the muscle belly (17). Sensors were placed by the same researcher for all participants.

Maximal voluntary isometric contractions (MVICs) for each muscle were performed immediately following sensor placement with each test mirroring previously recommended protocols (18,19,20,17). Testing for RA and EO was performed utilizing a foot elevated position and a standard position (18,19). The foot elevated test was performed with the knees and hips flexed to 90° while the standard position was performed with the feet flat on the ground. Participants laid supine on a mat with their feet stabilized for both tests. For RA, participants were instructed to place their hands on their chest and come into a crunch position in which a manual resistance was applied bilaterally to the anterior aspects of the shoulders. For EO, participants were instructed to perform a crunch with a rotation of the dominant shoulder to the opposite knee. A unilateral manual resistance to the anterior aspect of the dominant shoulder was applied. MVIC testing for MT and LS were performed utilizing a back extension (19). Participants laid prone on a mat with their feet stabilized and their hands placed together over their sacrum. They were then instructed to lift their chest off the ground against a manual resistance that was applied bilaterally to the upper back. UT and MD were performed using 90° shoulder abduction at 0° and 30° of horizontal flexion (20,17). Participants sat upright in a chair and were instructed to abduct their arm to the 0° and 30° positions. Stabilization of the shoulder was provided while manual resistance was applied proximal to the elbow. To maximize the potential of obtaining an MVIC, each test was performed for three sets of five seconds with consistent verbal encouragement. A thirty-second rest was provided between each set. The MVICs obtained were used for the normalization of muscle activation data across participants during data analysis.

Following completion of the MVICs, participants were outfitted with nine inertial measurement units (IMU) to capture body kinematics. IMU sensors were placed on the upper and lower thoracic regions of the trunk and bilaterally over the thighs, shanks, and feet using the Noraxon rigid-body model (21). All IMU straps were fastened tightly against the limbs to prevent sensor displacement when walking, and all sensors were checked for proper orientation prior to commencement of the carries. Once sensors were in their proper placement, they were calibrated to the participant's upright anatomical position.

Participants then completed the FC and ZC exercises in a randomized counterbalanced order with a load at 100% FFM using an Oppen Bar (Elieko, Halmstead Sweden). Each walk was 5 meters in length with a force plate located in the center of the walk to catch single step ground reaction forces with a 5-minute rest between. Prior to the first walk, participants were outfitted with weightlifting straps to eliminate grip strength barriers.

Instrumentation

Data were synced via the Noraxon Portable Lab (Scottsdale, AZ, USA) and recorded with MR3 software (Version 3.20). EMG data were recorded at a sampling rate of 1500 Hz using Noraxon

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mini DTS sensors. An IIR Butterworth band width frequency with a highpass of 20 Hz and a lowpass of 450 Hz was applied to raw data. Kinematic data were recorded with the IMU sensors at a sampling frequency of 100 Hz. A Bertec Force Plate sampling at 1000 Hz was used to record ground reaction forces.

Data Analysis

A multivariate analysis of variance (MANOVA) was utilized to identify potential differences between the ZC and FC regarding the EMG, kinematic, and kinetic variables. Prior to analysis, EMG was normalized to %MVIC and kinetic data was normalized to %bodyweight (BW). EMG and kinematic gait and joint angle data were analyzed for three strides in the middle of the walk. The EMG variables analyzed included average (AVG) activation of RA, EO, MT, LS, UT, and MD. Gait variables included stride length (cm), step width (cm), velocity (km/h), stride time (ms), and cadence (step/min). Peak joint angles (°) were analyzed for upper thoracic and the lumbar spine as well as the right and left hips, knees, and ankles. Upper thoracic and lumbar spine movements included flexion/extension (FLX/EXT), right and left lateral flexion (RLFLX/LLFLX), and right and left axial rotation (RROT/LROT). Hip movements included FLX/EXT, abduction/adduction (ABD/ADD), and internal/external rotation (IROT/EROT). Knee movements included FLX/EXT, and ankle movements included dorsiflexion/plantarflexion (DF/PF) and inversion/eversion (INV/EV). Kinetic data were recorded with the force plate from heel strike to toe off. Variables included peak anterior (GRFA), posterior (GRFP), medial (GRFM), lateral (GRFL), and vertical (GRFV) ground reaction forces. Statistical significance was set a priori to 0.05. Statistical analysis was performed in R statistical software (4.4.1).

RESULTS

No main effect between the ZC and FC (F (1,52) = 0.325, p = 0.572) was observed. However, statistically significant findings were observed across individual EMG, kinematic, and kinetic variables.

EMG

Significant differences were found in the MT Avg (F = 15.832, $p < 0.001$) and the LS Avg (F = 58.163, $p < 0.001$) with higher activation occurring in the ZC for both muscles (Table 1). No significant differences were found between the average muscle activation of RA, EO, UT, or MD.

Gait Kinematics

Of the kinematic gait variables analyzed, only stride length (F = 4.910, $p < 0.05$) was found to be significantly different between exercises with participants taking longer strides during the FC (Table 2).

Table 1. Mean muscle activation, expressed as %MVIC

	Zercher Mean ± SD	Farmer's Mean ± SD
Rectus Abdominis	4.4 ± 3.0	5.5 ± 4.6
External Oblique	12.1 ± 6.5	9.2 ± 5.4
Multifidus***	25.4 ± 13.0	12.0 ± 6.7
Longissimus***	20.6 ± 7.6	6.5 ± 2.8
Upper Trapezius	31.2 ± 19.8	32.4 ± 14.9
Middle Deltoid	9.6 ± 5.5	11.9 ± 7.9

*Note: *** $p < 0.001$*

Table 2. Gait kinematics averaged across three strides

	Zercher Mean ± SD	Farmer's Mean ± SD
Stride Length (cm)*	101.4 ± 9.4	108.1 ± 9.4
Step Width (cm)	6.1 ± 4.6	6.7 ± 3.7
Velocity (km/h)	3.04 ± 0.52	3.35 ± 0.49
Stride Time (ms)	1224.04 ± 164.78	1175.70 ± 123.06
Cadence (step/min)	99.93 ± 13.22	103.24 ± 10.64

*Note: * $p < 0.05$*

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Table 3. Maximum joint angles expressed in degrees

		Zercher Mean ± SD	Farmer's Mean ± SD
Upper Thoracic	Flexion	6.6 ± 5.9	4.7 ± 6.3
	Extension	-4.0 ± 6.0	-2.1 ± 6.9
	Right Lateral Flexion	3.6 ± 2.5	2.3 ± 2.6
	Left Lateral Flexion	0.7 ± 2.8	1.7 ± 2.7
	Right Rotation	3.3 ± 2.1	3.2 ± 2.6
	Left Rotation	4.6 ± 1.9	5.2 ± 3.6
Lumbar	Flexion***	1.7 ± 6.1	8.3 ± 4.5
	Extension***	2.0 ± 6.5	-4.4 ± 4.1
	Right Lateral Flexion**	2.5 ± 2.8	4.9 ± 2.3
	Left Lateral Flexion*	5.7 ± 2.4	3.8 ± 2.0
	Right Rotation*	2.8 ± 2.5	5.2 ± 3.4
	Left Rotation	3.8 ± 2.2	2.4 ± 2.4
Right Hip	Flexion**	22.7 ± 4.8	26.4 ± 4.2
	Extension	14.3 ± 3.9	13.0 ± 3.0
	Adduction	8.1 ± 2.6	8.3 ± 2.7
	Abduction	3.9 ± 2.6	4.4 ± 1.9
	Internal Rotation	5.7 ± 2.8	5.6 ± 3.0
	External Rotation	4.7 ± 1.9	4.9 ± 2.0
Left Hip	Flexion*	22.5 ± 3.6	26.8 ± 4.6
	Extension	12.7 ± 2.1	13.3 ± 2.6
	Adduction	8.4 ± 3.5	8.0 ± 3.3
	Abduction	4.9 ± 4.1	6.9 ± 3.6
	Internal Rotation	8.3 ± 2.7	7.1 ± 3.0
	External Rotation	4.1 ± 1.9	5.0 ± 2.3
Right Knee	Flexion	67.8 ± 8.6	64.9 ± 8.7
	Extension	-2.6 ± 5.0	-1.4 ± 4.3
Left Knee	Flexion	66.6 ± 9.0	65.2 ± 5.5
	Extension	-2.0 ± 5.5	-0.4 ± 4.6
Right Ankle	Dorsiflexion*	14.6 ± 4.7	11.4 ± 4.0
	Plantar Flexion	10.4 ± 6.0	12.9 ± 6.4
	Inversion	7.6 ± 4.3	6.1 ± 3.5
	Eversion	8.3 ± 3.1	9.0 ± 3.4
Left Ankle	Dorsiflexion*	14.5 ± 3.9	12.0 ± 3.0
	Plantar Flexion	5.8 ± 6.1	8.5 ± 6.1
	Inversion	14.9 ± 7.5	12.9 ± 7.2
	Eversion	7.2 ± 3.3	7.2 ± 2.8

*Note: *** $p < 0.001$, ** $p < 0.01$, * $p < 0.05$*

Joint Kinematics

Significant differences were observed in the lumbar spine, the right and left hip, and the right and left ankle. Lumbar EXT (F = 13.175, $p < 0.001$) and LLFLX ($F = 6.778$, $p < 0.05$) were significantly greater in the ZC while lumbar FLX ($F = 14.691$, $p < 0.001$), RLFLX (F = 7.884, $p < 0.01$), and RROT ($F = 6.432$, $p < 0.05$) were significantly greater in the FC (Table 3). Significantly greater hip flexion was observed during the FC in both the right ($F = 6.188$, $p < 0.01$) and left ($F = 10.312$, $p < 0.05$) hip joints (Table 3). Further, DF was observed to be significantly greater in the ZC for both the right ($F = 5.226$, $p < 0.05$) and left ($F = 4.800$, $p < 0.05$) ankles (Table 3).

Kinetics

Significant differences were found in GRFA ($F =$

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Table 4. Peak kinetic data from one step expressed as % body weight (BW)

	Zercher Mean ± SD	Farmer's Mean ± SD
Anterior GRF*	17.5 ± 7.4	22.3 ± 6.8
Posterior GRF*	25.4 ± 6.0	29.6 ± 5.8
Medial GRF	4.2 ± 2.7	4.9 ± 2.8
Lateral GRF	8.4 ± 4.0	8.8 ± 4.7
Vertical GRF	185.6 ± 14.4	179.9 ± 12.5

*Note: GRF = Ground reaction force; *p < 0.05*

4.332, $p < 0.05$) and GRFP ($F = 4.901$, $p < 0.05$) with the FC eliciting greater peaks for both forces (Table 4).

DISCUSSION

Multiple analyses of locomotion in unloaded and loaded conditions have determined that the addition of an external load alters the body's stability and causes changes in muscle activation, gait and joint kinematics, and kinetics (10,12). Further, previous observations regarding load placement have determined that the location of the load may alter the compensatory actions of the body by altering stability through a change in CoM (13). Specifically, moving the CoM higher and more anterior creates a more unstable condition relative to its anatomical position (13). As such, the purpose of this study was to examine the impact on body biomechanics during the ZC and FC exercises as a result of the load position differences. To achieve this, EMG, kinematic, and kinetic data of each exercise were quantified and compared. Observations of statistical significance amongst individual variables revealed several differences, providing insight into the unique characteristics of each exercise.

Analysis of trunk kinematics in this study found that the lumbar spine was subject to marked variability between the ZC and FC, showing significant differences in all three planes of movement. An examination of the sagittal plane specifically revealed that the ZC elicited greater peak lumbar EXT (+ 6.4°) while the FC elicited greater peak lumbar FLX (+ 6.6°). The greater EXT of the ZC is a common finding of anterior loading and is most likely attributed to the anterior horizontal shift in the body's CoM (12,13). This anterior shift moves the CoM forward in the BoS, causing instability in which the body compensates by initiating lumbar EXT. This EXT realigns the CoM over the BoS, allowing the body to regain stability. However, this spinal compensation results in vertebral column instability as noted by Studinickav et. al. who observed that spinal displacement from anatomical position disrupts the integrity of the spinal column (22). Further observation determined the body may compensate for this instability by initiating increased activation of the surrounding musculature, allowing for the stabilization of the spine while in a displaced position. This may explain the significant heightened activation observed in the MT (+ 13.4%) and LS (+ 14.2%) muscles in the ZC. These muscles have previously been characterized as powerful spine stabilizers and their heightened activation in the exercise where the lumbar spine is in an unstable position aligns with this notion (3,23). Interestingly, RA and EO exhibited similar activation between exercises (RA ± 1.1%, EO ± 2.9%), a finding in line with Anderson et. al. who found no significant change in the activation of these muscles during a change in anterior load bar height (12). This suggests the ventral core muscles are not significant contributors to the core stabilization needed in the ZC and denotes the importance of the dorsal trunk muscles in this specific exercise.

Although the ZC elicited higher peak lumbar EXT (2.0° ± 6.5), a peak lumbar FLX (1.7° ± 6.1) was still observed during the walk. In contrast, the FC produced greater peak FLX (8.3° ± 4.5) but never moved into an extended position, maintaining a minimum flexed position of 4.4° ± 4.1. Interestingly, a similar range of motion was exhibited during both carries with a range of 3.7° in the ZC and 3.9° in the FC. This range of motion is likely attributed to natural CoM changes during locomotion. To engage in this form of movement, the body disrupts its stability by shifting its CoM outside of its BoS while simultaneously activating the lower limbs to catch the CoM so as to avoid falling (24). In the case of forward locomotion, this CoM shift is translated anteriorly and can be achieved through flexing the lumbar spine, potentially explaining the FLX observed in both carries. However, the EXT differences are most likely a result of the load placement effects on CoM, further denoting the

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significance of this on potential compensatory mechanisms.

Gait analysis findings from the present study further contribute to the differences between these exercises and may indicate significant stability differences. Gait analysis revealed a significant difference in stride length and a velocity difference approaching significance (p = 0.0658) with the ZC producing shorter strides (- 6.7 cm) and slower speeds (- 0.31 km/h). This finding aligns with those of Alamoudi et. al. who observed shorter strides taken with anterior load carriage in comparison to bilateral load carriage. Further, both gait characteristics have been observed by Hak et. al. who found individuals use shorter strides and slower speeds as a mechanism to increase their stability through locomotion (25). As such, the gait differences between the ZC and FC may further denote the impact of load position on the stability of these exercises.

Additionally, the speed difference (9.7%) between the exercises aligns with the kinetic findings from our study. Kinetic analysis revealed the FC produced greater anterior (+ 4.8%) and posterior (+ 4.2%) peak GRFs. Examinations of GRFs have previously determined that speed has a direct influence on braking and propulsive forces with braking commonly exceeding propulsive, a finding that aligns directly with data from the present study (26).

Kinematic findings of the lower limbs additionally align with our gait analysis observations as discussed in a meta-analysis examining the effect of speed on gait parameters and lower limb joint kinematics in varying populations (27). This analysis found that in young adults, as speed and stride length increased, hip FLX and ankle PF increased with no reportable differences in the knee (27). Similarly, as these values decreased, hip and knee FLX and ankle PF decreased. This relationship of the hip to speed and stride length directly aligns with the observations in our study which found significantly greater bilateral hip FLX (right + 3.7°; left + 4.3°) in the FC. Further, the FC also produced greater PF values which were approaching significance in both ankles (right + 2.5°; left + 2.7°). Like our data from the hip, this aligns with the results of the meta-analysis.

Our study additionally observed significant differences of the lumbar spine in the frontal and transverse planes. Greater peak LLFLX (+ 1.7°) was observed in the ZC and greater peak RLFLX (+ 2.4°), and RROT (+ 2.4°) was observed in the FC. Further, although not statistically significant (p = 0.0783), peak LROT (+ 1.4°) was found to be higher in the ZC as well. Significant peak ankle DF was also found with both ankles (right + 3.2°; left + 2.5°) exhibiting higher peaks in the ZC. These findings may be a result of the CoM differences between the ZC and FC, however, further research into these specific variables should be done to better understand these results in relation to these loaded carry movement patterns.

CONCLUSION & FUTURE RECOMMENDATIONS

The purpose of the present study was to compare an equally loaded ZC to a FC to determine how the change in load placement of these exercises affects body biomechanics. Specifically, we aimed to analyze how the body compensates for the alteration in stability as a result of the CoM changes between the carries. Generally, participants compensated for the higher, more anterior load of the ZC by shifting their lumbar spine into EXT and altering their gait to move slower and take shorter steps. Conversely, in the FC, participants exhibited sustained lumbar FLX while engaging in faster speeds and larger steps. This may have further led to the higher AGRFs and PGRFs as well as greater bilateral hip FLX and ankle PF observed in this exercise. These findings may be the result of the stability differences between these exercises with the ZC presenting as less stable than the FC. This highlights the potential implications these exercises may have on the body as a result of performing them.

Our study does present with some limitations which should be considered when interpreting our results. First, our study sampled healthy, college aged individuals which means results are not generalizable to populations that fall out of this category. Further, resistance training experience level was not considered in the data analysis. Future studies may wish to focus on a sample of novice participants and/or participants who are experienced with the FC and ZC to determine how training may affect potential differences between these exercises. Finally, muscles analyzed in the present study focused on the core and the upper body. Future studies may wish to take a more comprehensive approach by analyzing muscles not included in the present study, as it is plausible that the observed kinematic differences could have impacted muscle activation in other body regions.

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CONFLICTS OF INTEREST

The authors report no potential conflict of interest.

FUNDING

This project was financially supported by Slippery Rock University through the Faculty Student Research Grant.

ETHICAL APPROVAL

Ethics for this study were approved in line with the Slippery Rock University Institutional Review Board (IRB). Participants received email communication with a copy of the informed consent prior to their sessions. Participants were additionally asked to read and sign a paper copy of the document at the beginning of their session.

DATES OF REFERENCE

Submission - 06/05/2025
Acceptance - 10/07/2025
Publication - 29/05/2026

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