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Implications of Arm Restraint on Lower Extremity Kinetics During Gait

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Abstract

Background

Literature indicates the importance of the upper extremities in providing stability and propulsion for the body during ambulation. However, the kinetic implications of upper extremity restraint during gait are not as well documented.

Aim

The objective of this study was to examine the effect of arm restraint (unilateral and bilateral) on lower extremity joint kinetics during walking.

Methods

Twenty-three healthy young participants were instrumented for three dimensional motion analysis, and tested in four randomly ordered upper extremity restraint conditions (unrestrained, bilateral restraint, right side restraint, and left side restraint). Temporal spatial parameters and gait/phase-specific lower extremity kinetics and kinematics were measured. For each restraint condition, pointwise differences from the unrestrained condition were compared using a two-way ANOVA model of restraint condition ("Condition") and gait cycle phase ("Timing").

Results

Decreases in walking speed and stride length were observed for all restraint conditions. Differences in kinetic demands were also noted, primarily at the hip and knee.

Conclusion

Upper extremity restraint in healthy young adults leads to significant changes in temporal-spatial parameters and proximal joint kinetics, most prominently during periods of load accommodation and balance.

Key words

Gait, kinetics, restraint, stability, upper extremity

1. Introduction

The present body of literature dealing with the upper extremity during gait has demonstrated that the arms play a significant role in ambulation. Early work by Elftmann refuted the popular belief that arm swing was purely passive pendular motion, demonstrating active muscular action and equivalent angular momentae associated with the trunk and arms.[1] Subsequent work by Fernandos-Ballesteros used electromyography to demonstrate muscular control of the arms during walking, with the triceps and posterior deltoid especially active.[2] These findings were initially attributed to the minimization of thoracic rotation. Jackson's later studies of the trunk, however, found decreased thoracic rotation when the arm was restrained, leading to a counter argument that arm swing was only meant to minimize high acceleration movements during ambulation.[3]

Previous studies have found that several upper extremity motion components are required for smooth, efficient gait. Murray et al demonstrated that counter phase rotation of the pelvic and shoulder girdles is essential,[4, 5] and Ohsata measured the effect of contralateral leg acceleration on arm swing through counter rotation of the trunk, effectively arresting rotation of the shoulder girdle.[6] Investigations have not been limited to motion; principal components analysis of ground reaction forces by Li et al demonstrated that the absence of arm swing in fixed-arm walking has little effect on vertical and sagittal contact forces.[7] They concluded that the influence of arm swing on the kinetics of gait must be exerted primarily through its effects on transverse force and/or on the vertical free moment about the ground contact point of the foot. Subsequent work by this group found that during restricted arm swing, the vertical free moment produced by the foot is increased to compensate for the loss of the action of the moment of arm swing in balancing lower limb swing.[8]

The role of arm swing in augmenting or inhibiting performance has been studied extensively in vertical jumping[9, 10, 11, 12] and long jumping.[13] Fewer studies of this nature dealing with gait have been published, and recent work in this area has focused on the restriction of arm swing. Early reports demonstrated differences in kinematic behaviors and temporal-spatial parameters when walking under restraint.[14, 15] A case study by Marks reported on a single participant walking under a unilateral restraint condition.[16] Findings indicated alterations in motion patterns in both the upper and lower extremities, but did not extend to the joint loading changes underlying these changes in motion. A report by Ford et al described the coordination between segments during arm restraint while walking at different speeds on a treadmill.[17] The authors found increased movement amplitude in the unconstrained arm, and noted that healthy participants have a flexible coordination scheme which allowed adaptations in trunk motion and inter-limb coordination at higher walking velocities. However, because the Ford study used a treadmill to vary walking speeds, joint loads and powers could not be calculated, and study findings were limited to measures of motion. A similar study by Yizhar et al reported differences in energy expenditure associated with variations in walking speed while walking with restrained arms.[18] A treadmill was again used for controlling gait velocity. Progressive increases in cadence and energy expenditure were noted in conjunction with velocity increases and arm restraint, with the largest increases attributed to conditions combining high walking speeds and arm restraint.

While the extant body of literature indicates the importance of the upper extremities in providing stability and propulsion for the body during ambulation, the kinetic implications of upper extremity restraint during gait are not as well documented.

The purpose of this study was to investigate the effect of arm swing restraint on lower extremity joint loading. We hypothesized that arm swing restriction would result in measurable differences in lower extremity kinetics, with increasing levels of restraint leading to increased kinetic demands. We specifically expected to see the greatest differences at the most distal articulation (i.e., at the ankles), concurrent with anticipated changes in temporo-spatial parameters (e.g., walking speed, stride length, cadence and stance/swing ratio).

2. Methods

This study was approved by the Institutional Review Board of the Medical College of Wisconsin, and all participants provided informed consent prior to testing. Twenty-three healthy adults (12 M, 11 F; Table 1) were recruited via public advertisement. Population size was determined via power analysis; pilot walking speed data from a single participant found this population size necessary to achieve 90% power. Participants were included if they were between the ages of 20 years and 50 years and were without any current neurological or musculoskeletal disorders (subject to clinical screen).

Table 1. Participant demographics

	All participants
n	23 (12 M/11 F)
Age (y)	32.5 ± 10.2
Mass (kg)	76.3 ± 13.4
Height (m)	1.7 ± 0.1
BMI (kg/m^2)	25.3 ± 4.0

Participants were measured and instrumented for gait analysis following the biomechanical model described by Kadaba et al.[19] Four restraint conditions were tested in this study: both arms free, right arm restrained, left arm restrained, and both arms restrained. Testing order of conditions was randomized to minimize any learning effect. Restraint involved tethering the indicated arm(s) to the torso in a position of slight shoulder flexion and adduction, with the elbow flexed to approximately 90° and the shoulder internally rotated so the arm fell across the chest. A soft Velcro strap was used to tether each arm, and both the arm and strap were placed such that they did not obscure the anatomical markers.

During gait testing sessions, participants were asked to walk at a freely selected speed down a 6-m walkway. Each testing condition subsession ended when a minimum of five acceptable force plate strikes were obtained. To avoid targeting, participants were not instructed to strike the force plates, and their starting positions on the walkway were altered as necessary to achieve acceptable force plate strikes.

2.1. Data collection

Kinematic and kinetic data were collected with a 15 camera Vicon 524 Motion Analysis System (Vicon Motion Systems Inc., Lake Forest, CA, USA) synchronized with two 6-DOF force plates (AMTI, Waterton, MA, USA) embedded in the laboratory walkway. Kinematic (video) data were collected at 120 Hz and kinetic (force plate) data were collected at 1080 Hz. Three dimensional joint kinematics and kinetics were calculated using the PlugInGait model.[20] Classical methods describing distal segment orientation relative to the next proximal segment were employed; Euler rotation angles were calculated with an YXZ (sagittal-coronal-transverse) order. All joint moments were reported as internal (demand) moments. Temporal-spatial parameters (walking speed, cadence, stride length, stance/swing ratio) were also calculated.

2.2. Statistical analysis

Because all participants were healthy young adults, symmetry of lower extremity biomechanics was assumed, limiting analysis to the right lower extremity. The four test conditions were then denoted as unrestrained ("UNRES"), bilateral restraint ("BIRES"), ipsilateral restraint ("IRES", for restraint on the right side), and contralateral restraint ("CRES", for restraint on the left side). Average joint kinematics and kinetics were calculated for each participant during each of the seven phases of gait. These seven phases, based on Perry's definition,[21] were designated as load response (LR, 0–10% gait cycle), midstance (MSt, 10–30% cycle), terminal stance (TSt, 30–50% cycle), pre-swing (PSw, 50–62% cycle), initial swing (ISw, 62–73% swing), mid-swing (MSw, 73–87% swing), and terminal swing (TSw, 87–100% swing).

The large volume of data produced in a gait analysis generally requires some simplifications of the data in order to allow statistical comparisons. In this analysis, the first simplification was to average trials within each participant, restraint condition, and phase of the gait cycle to compute a single average cycle for each. Within each phase of the gait cycle the averaged data was analyzed with a repeated measures model for restraint condition. Unstructured covariance was used to model the within-participant correlations. To help control the number of false positive results due to multiple testing, the positive False Discovery Rate[22] was calculated at a 5% level. This may be generally interpreted as the expected proportion of false positive results among all significant results. The analysis was performed using proc MIXED and proc MULTTEST in SAS version 9.2 (SAS Institute, Cary, NC, USA).

Temporal-spatial parameters were compared between all test conditions using paired Student *t* tests. To maintain a family-wise error rate of 5% for each parameter, we used a Bonferroni correction for six paired comparisons to set significance at $p < 0.008$.

3. Results

Temporal-spatial parameters demonstrated several differences between restraint conditions (Table 2). Average walking speed was the fastest in the UNRES condition (1.19 m/s), slowest in the BIRES condition (1.11 m/s; 6.7% decrease), and intermediate in the unilateral conditions (IRES 1.14 m/s, 4.2% decrease from UNRES; CRES 1.15 m/s, 3.4% decrease from UNRES). Differences between the UNRES condition and all restraint conditions were significant. Similar significant differences between the UNRES condition and the restraint conditions were seen for measures of stride length. In addition, significant walking speed and stride length differences were measured between CRES and BIRES conditions. No significant differences were seen in stance duration.

Table 2. Temporal-spatial parameters for all walking conditions (unrestrained UNRES, bilateral restraint BIRES, ipsilateral restraint IRES, and contralateral restraint CRES). Data are presented as mean ± SD. † indicates significant difference from UNRES condition; ‡ indicates significant difference from BIRES condition. Significance set at $p < 0.008$

	UNRES	**BIRES**	**IRES**	**CRES**
Walking speed (m/s)	1.19 ± 0.14	1.11 ± 0.14†	1.14 ± 0.14†	1.15 ± 0.14†‡
Stride length (m)	1.30 ± 0.11	1.24 ± 0.10†	1.26 ± 0.11†	1.27 ± 0.10†‡
Cadence (steps/min)	109.84 ± 6.21	106.54 ± 7.24†	108.55 ± 5.78	108.65 ± 6.62
Stance duration (%)	61.50 ± 1.59	61.68 ± 1.33	61.53 ± 1.49	61.52 ± 1.32

Differences in gait biomechanics were observed at multiple joints and in multiple planes. At the hip, differences were observed in the sagittal plane during portions of swing phase between the UNRES condition and both BIRES and CRES (maximum difference 0.06 Nm/kg; Figure 1). Additional differences were observed in the coronal plane during early stance, with the BIRES condition differing from all other restraint conditions (maximum difference 0.07 Nm/kg; Figure 2). No differences in the transverse plane were observed between conditions at the hip.

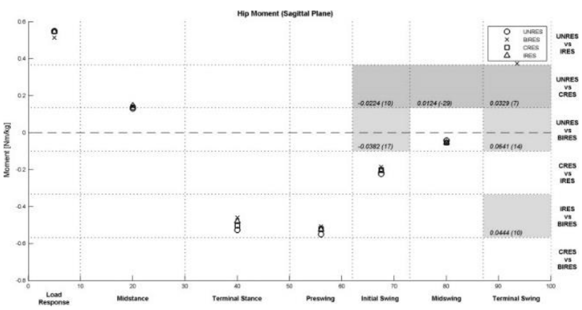

Figure 1. Mean sagittal plane hip moment values for each restraint condition at each phase of the gait cycle. Data are plotted at phase midpoint for clarity. Shaded portions of the grid indicate significant differences between two restraint conditions, interpreted via scale at right of plot. Magnitude of each significant difference is indicated in the lower left portion of the shaded square as [*magnitude of difference (% difference)*]. Example: During terminal swing, a significant difference was measured between the UNRES and CRES conditions (magnitude 0.0329 Nm/kg, or 7% of UNRES).

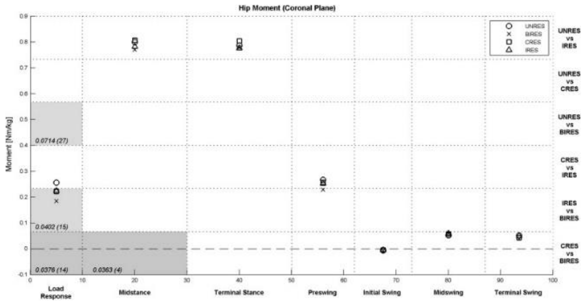

Figure 2. Mean coronal plane hip moment values for each restraint condition at each phase of the gait cycle. Data are plotted at phase midpoint for clarity. Shaded portions of the grid indicate significant differences between two restraint conditions, interpreted via scale at right of plot. Magnitude of each significant difference is indicated in the lower left portion of the shaded square as [*magnitude of difference (% difference)*]. Example: During load response, a significant difference was measured between the UNRES and BIRES conditions (magnitude 0.0714 Nm/kg, or 27% of UNRES).

At the knee, the majority of differences were observed in the coronal plane (maximum difference 0.05 Nm/kg; Figure 3). Differences were generally seen in late swing and early stance, and were most consistently observed between UNRES and BIRES. Additional differences were noted between BIRES and the unilateral restraint conditions (IRES and CRES). Significant transverse planes differences were noted in late swing and load response, but these were very low magnitude. Few differences were observed in the sagittal plane.

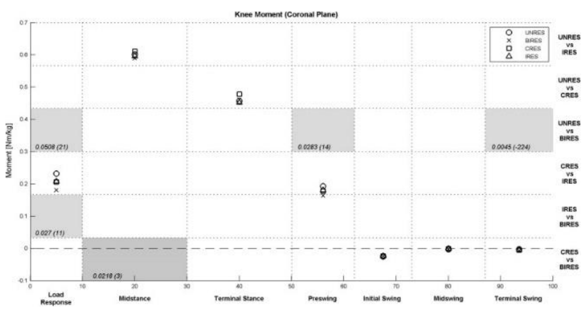

Figure 3. Mean coronal plane knee moment values for each restraint condition at each phase of the gait cycle. Data are plotted at phase midpoint for clarity. Shaded portions of the grid indicate significant differences between two restraint conditions, interpreted via scale at right of plot. Magnitude of each significant difference is indicated in the lower left portion of the shaded square as [*magnitude of difference (% difference)*]. Example: During load response, a significant difference was measured between the UNRES and BIRES conditions (magnitude 0.0508 Nm/kg, or 21% of UNRES).

The only significant differences observed at the ankle occurred in the transverse plane during late swing and early stance, and these were very low magnitude (maximum difference 0.01 Nm/kg). Additional differences were noted in transverse plane motion of the pelvis throughout most of the gait cycle, with the UNRES, BIRES, and unilateral restraint conditions each having a characteristic pattern of motion during the stride (maximum difference 2.7°; Figure 4).

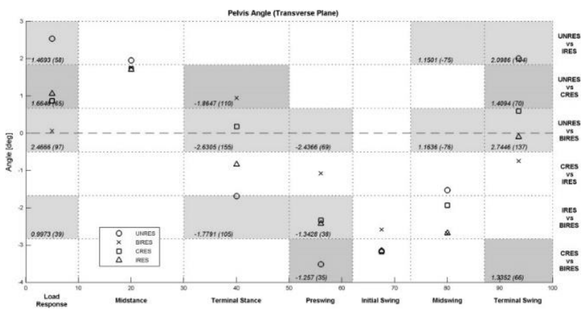

Figure 4. Mean transverse plane pelvic rotation values for each restraint condition at each phase of the gait cycle. Data are plotted at phase midpoint for clarity. Shaded portions of the grid indicate significant differences between two restraint conditions, interpreted via scale at right of plot. Magnitude of each significant difference is indicated in the lower left portion of the shaded square as [*magnitude of difference (% difference)*]. Example: During load response, a significant difference was measured between the IRES and BIRES conditions (magnitude 0.9973°, or 39% of UNRES).

Based on the coronal plane differences observed in early stance phase, a secondary analysis was conducted to study the temporal nature of restraint-based change during stance (Figure 5). These differences between each restraint condition and the UNRES condition demonstrate an “inverted N” pattern with two distinct curves. The first is a progressively larger negative difference between restraint conditions and the UNRES condition, peaking at ∼12% of the gait cycle (“negative curve”). The second is a progressively larger positive difference, peaking at ∼30% cycle (“positive curve”).

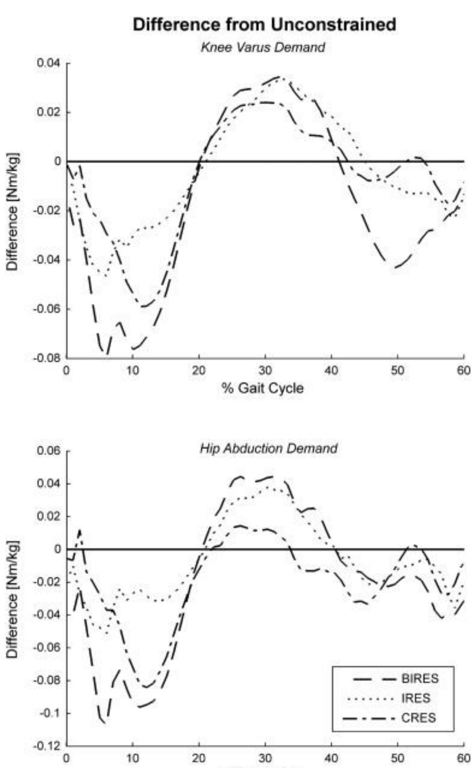

Figure 5. Difference between demand moment for each restraint condition and UNRES demand moment. Data are plotted during stance phase (0–60% gait cycle) for the coronal plane moments at the hip (top) and knee (bottom).

4. Conclusion

This study investigated the effects of arm swing restraint on temporal-spatial parameters and lower extremity kinetics. We hypothesized that alterations in arm swing would slow walking speed and shorten stride length. We also hypothesize that restraint of the arms would result in changes to lower extremity joint moments, with greater kinetic changes at more distal articulations.

Our hypotheses related to temporal-spatial changes were clearly confirmed. All restraint conditions demonstrated significant reductions in walking speed. In the unilateral conditions (IRES and CRES), these changes could be directly linked to significant reductions in stride length; in the bilateral condition (BIRES), a significant reduction in cadence also played a role in the reduced walking speed. These findings agree with those of previous studies.[14, 15, 16]

Kinetic changes due to restraint demonstrated significant changes at all levels, with changes of the largest magnitude taking place at the hip. These results refuted our hypothesis that changes would be limited to more distal articulations. Differences were investigated on a per-condition basis at key time points in the gait cycle, allowing us to associated changes with the biomechanical characteristics of the phase of the gait cycle in which they occurred. The largest number and most consistent differences were observed between the UNRES and BIRES conditions. Additional differences were noted between these conditions and the unilateral restraint conditions, but these were less consistent, and differed depending on which arm was restrained (ipsilateral vs. contralateral). Overall magnitudes of change were small.

The period of the gait cycle from 0% to 20% represents initial contact and load response, followed by opposite toe-off and early midstance. This portion of the cycle is remarkable for limb control and increasing load accommodation as the foot is placed on the floor and descends to a plantigrade position, and the opposite limb moves into swing phase.[21] Study findings indicate that with increasing levels of restraint, the hip demonstrates reductions in coronal plane demand from 0% to 20% gait cycle (load response and early midstance). These changes are significant at the level of bilateral restraint; the BIRES condition demonstrates a significant 27% reduction from the UNRES condition and 14–15% reduction from the unilateral restraint conditions. The knee demonstrates similar reductions in varus demand, with a significant 21% reduction between BIRES and UNRES, and an 11% reduction between BIRES and IRES.

Given the involvement of the hip abductors in stabilizing pelvic position and the varus/valgus involvement of the knee in accommodating weight, these results suggest an alteration of stabilization during periods of load accommodation (i.e., from initial contact through load response and early midstance). The significant differences between restrained conditions and the UNRES condition suggest that our tested population of healthy young adults could not fully adapt to upper extremity restraint, despite decreases in temporal-spatial parameters. It seems likely that aged or impaired populations would have further difficulty adapting. These findings have significant implications for the rehabilitation process, especially in situations in which impaired arm motion is expected, such as those involving upper extremity injury or stroke.

The patterns observed in the secondary analysis of temporal changes point to a timing lag in demand moments for the restraint conditions, as peak demand during load response is reached later in the cycle than during the UNRES condition. The subsequent positive curve represents prolonged kinetic demand at a magnitude greater than the UNRES condition. These findings suggest that in adapting to a restraint condition, participants attenuate proximal moment demands to accommodate weightbearing. However, once the demand moment has peaked, it falls off much slower than in the UNRES condition, implying a prolonged demand for stability. The relationship of this pattern to temporal-spatial parameters is not entirely clear; as no significant differences were observed in stance duration (Table 2), these changes cannot be attributed to differences in periods of weightbearing. However, significant differences in walking speed may play a role in these findings.

In addition to the coronal plane effects of arm restraint, sagittal plane differences were also observed at the hip during swing phase. As in the coronal plane changes, increasing levels of restraint led to more neutral moment patterns, with the BIRES condition demonstrating moments closest to zero. The largest magnitudes of difference were observed in initial swing (UNRES vs. CRES, 10% decrease; UNRES vs. BIRES, 17% decrease) and in terminal swing (UNRES vs. CRES, 7% decrease; UNRES vs. BIRES, 14% decrease). These reductions in flexor demand in early swing and extensor demand in late swing may be linked to reductions in walking speed, as demand to swing the leg forward and subsequently control and slow its progress is diminished. Despite the statistical significance of these differences, their small magnitude was also noted; the clinical significance of these minimal changes in hip motion is unclear.

Kinematic changes were also observed in measures of pelvis rotation; distinct differences were noted between all levels of restraint, and were persistent through much of the gait cycle. In general, higher levels of restraint led to more neutral orientations of the pelvis, with the UNRES condition demonstrating the most motion and the BIRES condition demonstrating the least. These findings seem to be in agreement with the original reports from Murray regarding counter rotations of the pelvic and shoulder girdle.[4, 5] These findings are also in agreement with our measures of reduced stride length, as increased pelvic rotation is a direct contributor to increased stride length.[23]

A number of the kinetic changes observed in this study might be explained by joint positions remaining closer to neutral, reducing the overall range of motion (ROM). These reduced ROMs tie into the relationship between walking speed and ROM which has been demonstrated by a number of investigators.[24, 25] As walking speed decreases, joint motion decreases and joint positions remain closer to neutral. The maintenance of more neutral positioning leads to shorter moment arms and reduced kinetic demands. This strategy may have allowed the able-bodied participants in this study to compensate for feelings of instability by reducing the stability demands placed on each joint. A slower walking speed, while not necessarily desired, was the necessary outcome of this strategy.

Contrary to our hypothesis, few significant differences were noted at the ankle. Overall differences between the restraint conditions and the UNRES condition seemed to be most prominent proximally, and were related to the degree of impairment (i.e., bilateral restraint had a larger effect than unilateral restraint). This minimal distal compensation suggests that rehabilitation strategies which focus on core stability and proximal joints may be more appropriate for patients faced with an impairment which mimics these restraint conditions. Ford et al have measured alterations in frequency and phase between upper and lower extremity coordination under conditions of constraint, suggesting a compensatory strategy involving not just the lower extremity but also the unconstrained upper extremity.[17] Further modeling of alterations in upper extremity kinetics may help clarify the kinetic nature of these compensations, and the interaction between upper extremity inertial factors and lower extremity load-bearing factors.

The overall results of this study suggest that in healthy young adults, any level of upper extremity restraint will have a kinetic effect during periods of weight accommodation and limb propulsion. These effects are increased by increasing the level of restraint (i.e., unilateral vs. bilateral restraint). The small magnitudes of these changes makes their clinical significance somewhat unclear, but their statistical significance points clearly toward some type of gait alteration imposed by arm restraint. The response of a population with orthopedic or neuromuscular impairment to such restraints deserves further attention, as their ability to acclimate and adjust to the restrictions will be impacted by their impairments. In patients with upper extremity impairment who demonstrate reduced ability to compensate for arm swing deficiencies, these findings may have important implications in regaining and maintaining ambulatory ability.

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