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Walking mechanics of persons who use reciprocating gait orthoses

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Abstract—Although ambulation with a reciprocating gait orthosis (RGO) may provide physical benefits to people with lower-limb paralysis, the high metabolic energy cost associated with ambulation limits orthosis use. The purpose of this case series was to investigate the dynamics of ambulation with RGOs to identify and better understand the potential causes of the high energy cost. Data were acquired from five regular users of RGOs. Kinematics and kinetics were measured, and the moments and powers acting at the hips and shoulders calculated. All RGO users walked with a flexed trunk and bore a large proportion of body weight through the arms during single support. Moments at the shoulder encouraged trunk extension, while moments at the hip encouraged trunk flexion. An extension moment acted on the hip at the beginning of swing, which was antagonistic to the goal of swing and contradicted the intent of the reciprocal link: to advance the swing leg. These results suggest that characteristics of RGO ambulation are consistent across users. The relationship between posture, forces acting on the walking aids, and the action of the RGO reciprocal link should be further explored because these factors likely contribute to the high metabolic cost of ambulation with an RGO.

Key words: biomechanics, gait, gait analysis, hip-knee-ankle-foot orthosis, kinematics, kinetics, orthotic devices, lower-limb paralysis, rehabilitation, reciprocating gait orthosis, RGO.

INTRODUCTION

Studies have shown that an upright posture and walking can benefit people with lower-limb paralysis. These benefits include reduced incidence of joint contractures, bone fractures, and pressure sores [1]; prevention of osteoporosis [2]; improved bowel function, urinary drainage, and peripheral circulation; and stimulation of leg growth for growing children [3]. Other studies have also claimed that upright ambulation has psychosocial advantages [4–6], such as improved interaction with peers [3] and the environment [5], more positive perception of the body, [6] and greater self-respect [4].

Many assistive devices have been developed to help people with lower-limb paralysis stand and walk. These devices range from the relatively simple, conventional hip-knee-ankle-foot orthosis (HKAFO) [7] to the more complex reciprocating gait orthosis (RGO) [8]. Both orthoses immobilize the knees and ankles in an appropriate standing alignment and allow ambulation with crutches

Abbreviations: 3-D = three-dimensional, COM = center of mass, GRF = ground reaction force, HGO = hip guidance orthosis, HKAFO = hip-knee-ankle-foot orthosis, IRGO = isocentric reciprocating gait orthosis, RGO = reciprocating gait orthosis, ROM = range of motion, VA = Department of Veterans Affairs, VACMARL = VA Chicago Motion Analysis Research Laboratory.

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or a walker. The conventional HKAFO also immobilizes the hip joint; hence, users usually adopt a swing-to or swing-through gait pattern. Other orthoses such as RGOs, the hip guidance orthoses (HGOs) [9], and the Walkabout orthosis (Polymedic; Ashmore, Gold Coast, Australia) [10] allow hip motion in the sagittal plane, which enables users to adopt a reciprocal gait pattern that is more similar to nondisabled walking than a swing-through gait pattern. While the HGO and Walkabout are designed with the intent of gravity advancing the swing leg [9–10], a special mechanism in the RGO, referred to as the reciprocal link, is intended to facilitate the reciprocal gait pattern by coupling motion of the two hip joints so that flexing one hip joint extends the contralateral hip, and vice versa.

Despite the perceived benefits of upright ambulation and the more cosmetic gait provided by RGOs, discontinuation of RGO use is high. Discontinuation of orthosis use ranges from 61 to 90 percent for children with myelomeningocele [11–12] and from 46 to 54 percent in adults with spinal cord injury [13–16]. Sykes et al. reported that adults who persevered in using RGOs considered the orthosis to be exercise equipment [12], which they used on average three times a week for approximately 2 hours. Many of the RGO users surveyed by Sykes et al. cited the effort needed to ambulate with the orthosis as the main reason for limiting RGO use [12].

Walking with RGOs is slow and exhausting [17–18]. Studies have reported the oxygen cost of RGO walking to be 1.0 mL/kg/m at user-selected walking speeds ranging from 0.2 to 0.3 m/s [17–19] compared with 0.176 mL/kg/m at 1.28 m/s for nondisabled persons [20]. Bernardi et al. measured metabolic energy and calculated the mechanical energies for each body segment of 10 RGO users and reported that the mechanical work associated with RGO ambulation was significantly greater than that for nondisabled walking [18]. They concluded that the mechanical power required for walking with an RGO was a major cause of the high metabolic energy expenditure. However, the aspects of RGO gait dynamics that contributed to these mechanical inefficiencies remain unclear. Further understanding of the gait dynamics of RGO ambulation is needed for the high metabolic energy expenditure required of users to be reduced. This reduction would likely make the RGO a more useful device for people with paraplegia.

Unfortunately, only a few quantitative analyses of RGO gait dynamics are reported in the literature [7]. Although Bernardi et al. measured the trajectory of each body segment to calculate mechanical energy, they did not report these trajectories [18]. Whittle and Cochrane undertook one of the first dynamic analyses of people walking with RGOs with a group of 12 adults with thoracic-level lesions [21], reporting the range of rotational motion at the hip in the coronal and sagittal planes. They also reported the subjects' stride characteristics and the peak crutch force experienced over the gait cycle. Kawashima et al. reported the hip angles in the sagittal plane for the entire gait cycle [19], as well as the vertical ground reaction forces (GRFs) acting on the feet and crutches of 10 RGO users. Probably the most comprehensive study published to date was by Tashman et al. [22]. They measured the three-dimensional (3-D) kinetics and kinematics for a single RGO user during the single-support phase of stance and reported motion of the legs relative to a global frame and the pelvis, as well as the motion of the upper body relative to the pelvis. They also reported the GRFs acting on the subject's foot and crutches in three dimensions and calculated the forces and moments acting on the subject's hips and shoulders in three dimensions. These studies illustrate that although some dynamic analyses are available in the literature, they are generally limited to a single subject, dimension, or phase of the gait cycle.

We desire to develop an improved understanding of the gait dynamics of RGO users and to identify factors contributing to the high metabolic cost of RGO ambulation. Although ambulation may provide physical and psychological benefits, the high energy cost associated with orthotic ambulation currently prevents RGOs from being useful in daily living. The purpose of this case series was to investigate the dynamics of ambulation with RGOs, including characteristics that have not been previously presented for multiple subjects, to identify and better understand the potential causes of the high metabolic energy cost of RGO ambulation.

METHODS

Motion data from five subjects (**Table 1**) were acquired at the Department of Veterans Affairs (VA) Chicago Motion Analysis Research Laboratory (VAC-MARL), which is equipped with an 8-camera digital RealTime motion capture system (Motion Analysis Corporation; Santa Rosa, California) for recording kinematic data and six force plates (Advanced Mechanical

Table 1.
Descriptive information of five study subjects.

Subject No.	Sex	Age	Height (cm)	Weight (kg)	Pathology	Lesion Level	RGO	Walking Aid
1	F	28	152	66	SCI	C7*	IRGO	Parallel bars
2	M	7	101	16	Spina bifida	T12*	IRGO	Crutches
3	M	21	173	71	SCI	T11*	ARGO	Anterior walker
4	F	12	129	36	Spina bifida	L3†	IRGO	Crutches
5	F	8	92	18	Spina bifida	L3†	IRGO	Reverse walker

*Reported by subject.
†Based on evaluation by investigator.
ARGO = advanced reciprocating gait orthosis, C = cervical, F = female, IRGO = isocentric reciprocating gait orthosis, L = lumbar, M = male, RGO = reciprocating gait orthosis, SCI = spinal cord injury, T = thoracic.

Technology, Inc; Watertown, Massachusetts) embedded flush in the floor of a 10 m walkway for measuring GRFs. Kinematic and kinetic data were sampled at 120 Hz. Subjects were included in the study if they regularly used an RGO because of lower-limb paralysis. Subjects under the age of 6 were excluded because they were not considered mature enough to follow the protocol. The Northwestern University Institutional Review Board approved this study, and informed consent was obtained from each individual before participation. Data were collected from subjects walking in their customary orthosis and using their customary walking aids (**Table 1**). Each subject underwent a clinical examination by a qualified orthotist who measured the passive range of motion (ROM) of the hip, knee, and ankle joints. The orthotist also measured hip flexion contractures and performed manual muscle tests on the lower-limb muscles as described by Kendall et al. (**Table 2**) [23]. Subjects voluntarily reported lesion level because access to their medical history was unavailable. In instances when subjects were uncertain about their lesion level, it was estimated from the results of their clinical examination.

We quantified motion of the body segments using passive reflective markers that were taped to the body and orthosis in the Helen Hayes configuration [24]. Markers were placed on the orthosis for the lower-limb segments and on the subject for the torso and upper-limb segments (**Figure 1**). Four markers were also placed on each walking aid. For dynamic trials, markers were located on the shoe over the dorsum of the foot between the second and third metatarsals immediately proximal to the metatarsal heads; on the shoe over the posterior calcaneus at the same height as the toe marker, lateral ankle joint, lateral knee joint, and right and left anterior superior iliac spines; and on the RGO over the sacrum. Thigh and shank markers were placed on the RGO. On the upper body, markers were placed on the acromion processes, lateral humeral condyles, and posterior wrist between the styloid processes. For static trials, markers were added to the medial ankle and knee joints. For consistency, the same laboratory personnel placed all markers on all subjects.

Conventionally, force-plate measurements are only recorded when a foot or walking aid is exclusively and completely located on the force plate. Because of small step lengths or the use of walking aids, some of our subjects could not meet this criterion. Hence, force data were recorded when a foot or walking aid was completely and exclusively on a force plate during the single-support phase of stance. Force measurements were recorded from a minimum of five different foot strikes for each foot, unless the subject became too fatigued to continue the study. Anterior and posterior walkers were too large to fit on a single force plate. In those instances, force data for the anterior pair of walker legs were measured independently from those of the posterior pair. Force data from each pair of legs were averaged separately, and we then summed the means to estimate the total GRF acting on the walker. For the subject who used parallel bars, force data were not measured.

EVa Real-Time software (Motion Analysis Corporation; Santa Rosa, California) was used to determine the 3-D position of each marker relative to the laboratory coordinate system during each frame of each trial. The raw coordinate data were filtered with a second-order bidirectional Butterworth low-pass filter with an effective cutoff frequency of 6 Hz, as suggested by Winter [25]. We used Orthotrak software (Motion Analysis Corporation, Santa Rosa, California) to calculate the joint angles,

Table 2.
Clinical measurements of study subjects' right and left hip, knee, and ankle joints and lower-limb muscle strengths.

Passive ROM (°)	Subject 1		Subject 2		Subject 3		Subject 4		Subject 5	
	Right	Left	Right	Left	Right	Left	Right	Left	Right	Left
Thomas Test	0	0	0	0	0	0	35	35	20	25
Hip Flexion	120	120	130	130	130	110	100	110	90	25
Hip Abduction (hip 0°)	30	30	50	50	25	25	20	20	0	10
Knee Flexion	WNL	WNL	WNL	WNL	130	130	NA	NA	WNL	WNL
Knee Extension (hip 0°)	5 HE	5 HE	0	0	5 HE	5 HE	25	20	5	10
Ankle Dorsiflexion (knee extended)	5 PF	5 PF	15	15	5	0	0	15	10	0
Muscle Strength										
Hip Flexion (hip 0°)	0	0	1	1	0	0	4	4	3	3
Hip Extension (knee 0°)	0	0	0	0	0	0	NA	NA	0	0
Hip Extension (knee 90°)	0	0	0	0	0	0	NA	NA	0	0
Hip Abduction (hip 0°)	0	0	1	1	0	0	1	1	2	2
Knee Flexion (prone)	0	0	0	0	0	0	2	1	1	0
Knee Extension (sitting)	0	0	0	0	0	0	3+	4	5	1

HE = hyperextension, NA = not available, PF = plantar flexion, ROM = range of motion, WNL = within normal limits.

joint centers, and centers of mass (COMs) of body segments. Gait events were determined within Orthotrak.

We used custom programs (MATLAB, The MathWorks, Inc; Natick, Massachusetts) to calculate the velocities and accelerations of body segments and the forces, moments, and powers acting on the hips and shoulders. These calculations were based on a link segment model where each body segment, (thigh, foot, shank, etc.) was represented as a rigid link between two adjacent joints with a given mass and moment of inertia. The masses and radii of gyration for body segments were estimated from anthropomorphic data for nondisabled people [25] because corresponding data for people with lower-limb paralysis were not available. The motion of the segments and GRFs were used to calculate the minimum joint reaction forces. Forces and moments at the hip joints were calculated with traditional inverse dynamic equations. The sum of the forces acting on both shoulders was calculated from the measured motion of the trunk and the calculated forces acting at both hips as

$$\sum \vec{F}_{\text{shoulders}} = {}^{-}\vec{F}_{\text{stancehip}} + {}^{-}\vec{F}_{\text{swinghip}} + m_{\text{trunk}}\left(\vec{a}_{\text{trunk}} - \vec{g}\right), \quad \textbf{(1)}$$

where $\vec{F}$ = forces, m = mass, a = acceleration, and g = the acceleration due to gravity. Both hip and shoulder forces created moments (M) about the trunk COM that were calculated in the sagittal plane according to

$$M = \vec{r} \times \vec{F}, \quad \textbf{(2)}$$

where $\vec{r}$ = the distance from the trunk COM to the stance hip joint or the midpoint between the shoulder joints and $\vec{F}$ is the sum of the forces acting at the stance hip or both shoulders, respectively. We then used velocities to calculate the power generated at the hip joints during swing phase. The translational power generated at the hip was calculated as the dot product of the force vector acting at the stance hip joint and the linear velocity of the hip-joint center. Similarly, we calculated the rotational power

Figure 1.
Example of marker placement used for all subjects.

generated at the hip by taking the dot product of the hip moment and the angular velocity of the hip joint.

Finally, all variables were ensemble averaged for each subject. For each variable, we normalized time by the duration of the gait cycle. Interpolated values from the variable were taken at fixed intervals on the normalized time scale. These values were then averaged at each interval over all the gait cycles. Data from RGO users were compared with those from a database of nondisabled ambulators.

RESULTS

Temporal spatial data for each subject are shown in **Table 3**, along with the number of trials collected for each subject. None of the subjects asked to end the data collection prematurely because of fatigue. All subjects walked at freely selected speeds that were only about one-third of that typically adopted by nondisabled individuals. For all subjects, step length and cadence were found to be less than two-thirds of the mean value from nondisabled ambulators. RGO users tended to have substantially longer double-support phases compared with nondisabled persons (**Figure 2**).

Table 3.
Temporal spatial data and number of trials analyzed for each subject.

Subject No.	Step Length (cm)	Cadence (steps/min)	Freely Selected Speed (m/s)	Trials Analyzed
1	24.6	38.6	0.19	6
2	40.7	56.9	0.39	5
3	27.0	17.4	0.08	11
4	33.2	60.4	0.34	9
5	23.1	72.3	0.29	10
Nondisabled	64.9	109.5	1.18	NA

NA = not available.

Sagittal plane kinematics for the trunk, shoulders, pelvis, and hips are shown in **Figure 2**. Sagittal plane motion of the pelvis closely followed trunk motion, varying sinusoidally over time with a period equal to the step cycle (i.e., half a gait cycle). The RGO users walked with a continuously flexed trunk and moved their trunk through a larger flexion range than nondisabled ambulators. Generally, motion of the trunk and pelvis was out of phase with that exhibited by nondisabled individuals. Hip flexion occurred predominantly during single-support phases, with negligible motion during double-support phases. At the shoulder, subjects 2, 3, and 5 flexed their arms in a sinusoidal fashion like nondisabled individuals, except at twice the frequency.

Anterior, medial, and vertical GRFs acting on the stance foot and walking aids during single support are depicted in **Figure 3**. With the exception of subject 2, anterior GRFs acting on the stance foot of RGO users were small compared with nondisabled persons, remaining below 5 percent of body weight for most of single support. Overall, GRFs acting on the stance foot and walking aids of most subjects opposed forward motion during single support, with only two subjects (subjects 3 and 5) having anteriorly directed GRFs on the walking aids for any portion of single support. In the medial-lateral direction, most subjects had more laterally directed GRFs acting on their stance foot than nondisabled persons, with small lateral forces acting on the walking aids. Only two subjects (subjects 3 and 5) had medial GRFs acting on the walking aids during single support. All RGO users had smaller minimum vertical GRFs than nondisabled persons and experienced periods of single support during

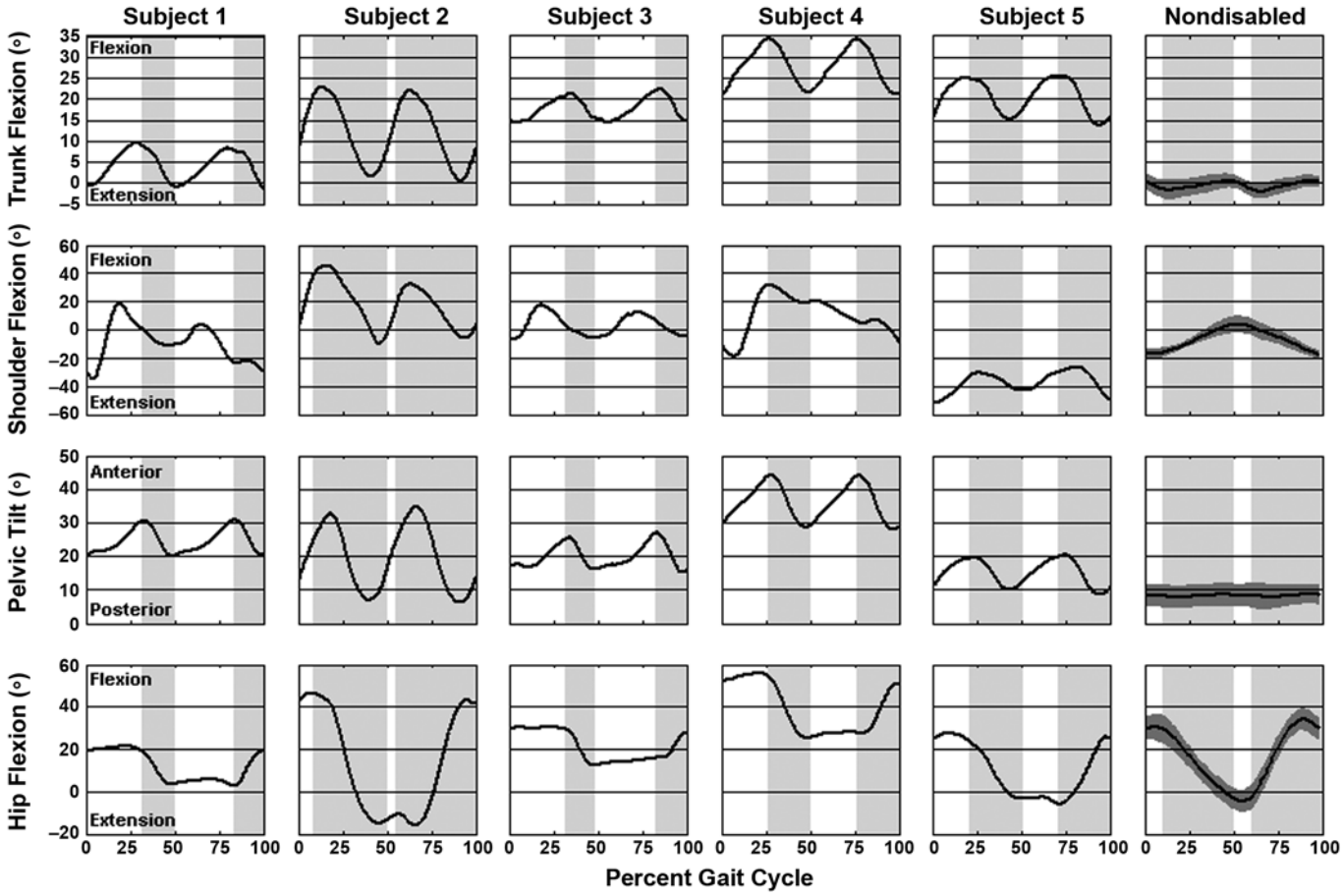

Figure 2.
Mean sagittal plane kinematics for the trunk, shoulder, pelvis, and hip for all subjects from right heel strike to subsequent right heel strike. Data from nondisabled persons represent mean ± 1 standard deviation. Periods of single support are shaded in gray.

which they had less than half their body weight on the stance foot. The vertical GRFs acting on the walking aids reached a maximum of at least 50 percent of body weight.

Figure 4 illustrates the moments about the trunk COM caused by forces acting at the shoulders and the hips during single support. Forces acting at the shoulders promoted trunk extension moments for all subjects, while forces acting at the hips promoted trunk flexion moments for all subjects with the exception of subject 2. **Figure 5** illustrates the moments generated at the hip joints during swing phase. While nondisabled persons experience hip flexion moments at the beginning of swing, all the RGO users experienced hip extension moments. Hip flexion moments tended to occur before midstance and were small than those of nondisabled persons.

Figure 6 depicts the power generated at the hip during swing phase and the rate of work done on the leg by hip-joint reaction forces during swing phase. The power generated from hip moments was small and dissipative compared with the power from hip forces for most subjects. The power from hip forces demonstrated a sinusoidal pattern.

DISCUSSION

For this case series, we investigated the gait dynamics of RGO users to identify potential causes for the high metabolic energy costs that have been so frequently reported in the literature [17–19]. The subjects tested for this study reflected the variability typically observed in

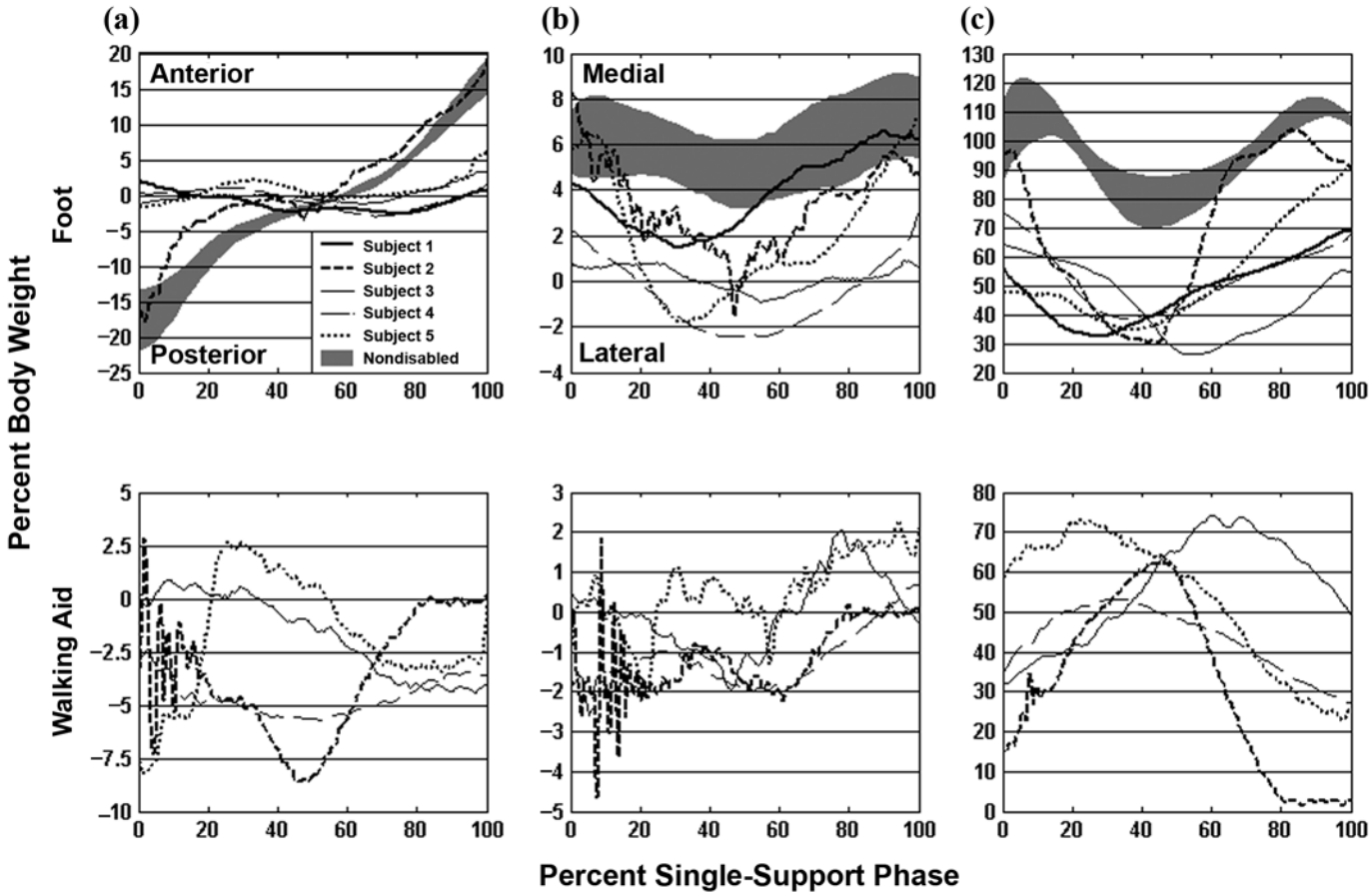

Figure 3.
Mean **(a)** anterior-posterior, **(b)** medial-lateral, and **(c)** vertical ground reaction forces (GRFs) acting on the stance foot and walking aids during single-support phase of stance for each subject. For walking aid data, sum of GRFs acting on both crutches or on all walker legs during single-support phase are shown. Walking aid GRF data are not shown for subject 1 because forces acting through parallel bars could not be measured. Gray band indicates foot forces (mean ± 1 standard deviation) from nondisabled persons.

the population of RGO users with respect to age, pathology, walking aids used, etc. The data included variables that have not been previously presented for multiple subjects, such as sagittal plane trunk kinematics and hip moments and powers.

Despite substantial heterogeneity in the study sample, many gait characteristics were consistent. It was demonstrated that all five RGO users walked with a flexed trunk. Pelvic motion was coupled to that of the trunk by the RGO, and hip flexion angle remained constant, indicating that both legs were rotating along with the pelvis while maintaining contact with the ground. At the beginning of single support, the RGO users transferred the majority of their weight from the stance leg to the walking aids and elevated the swing-side pelvis upward to facilitate foot clearance by tilting the trunk laterally over the stance leg [18–19,21–22]. The trunk and pelvis also extended during this phase, which pushed the pelvis forward and extended the stance leg hip. This motion produced flexion of the swing leg hip. The swing leg then contacted the ground and the cycle began again.

All RGO users in this study demonstrated a reciprocating gait when ambulating with the RGO. Although a reciprocating gait is considered distinct from a swing-through gait, they appear to share similar dynamic characteristics. The motions of the arms and trunk are similar if the swing phase of the reciprocating gait is likened to the flight phase of the swing-through gait. The trunk flexes forward during double support and extends backward during swing/flight. Walking aids are moved forward

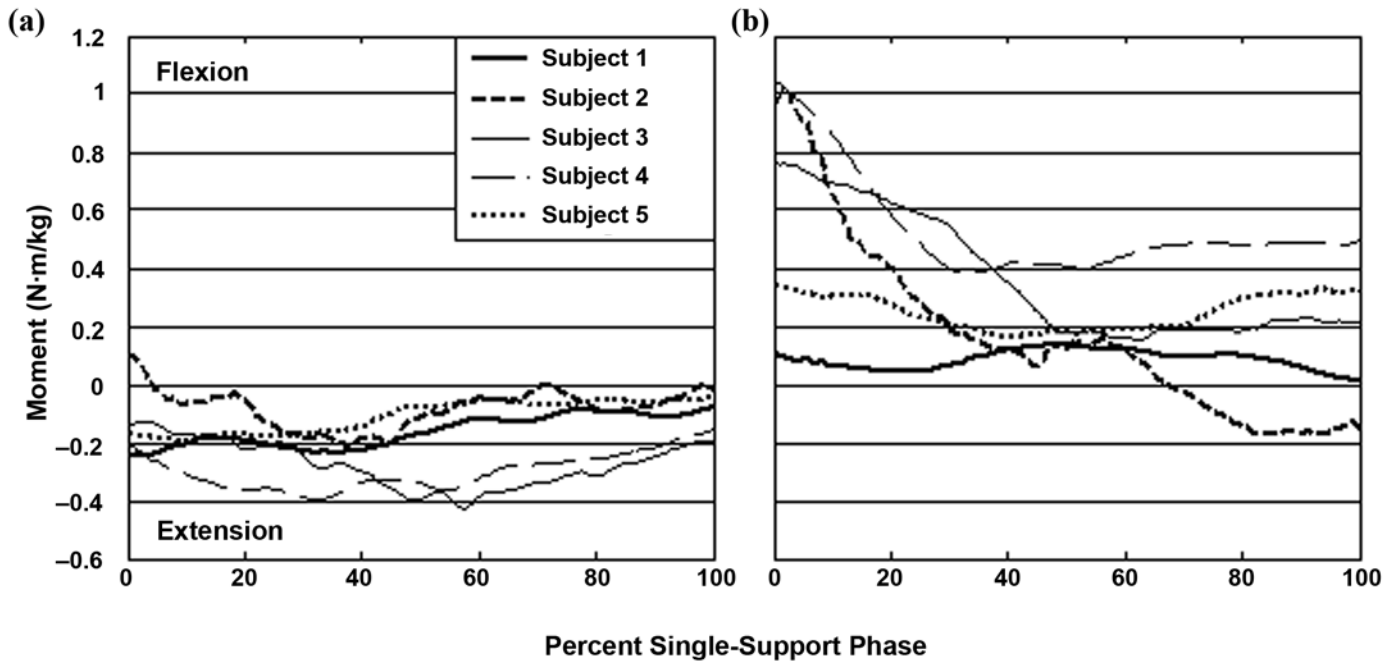

Figure 4.
Mean moments about trunk center of mass in sagittal plane caused by joint reaction forces at **(a)** shoulder and **(b)** hip during single-support phase for each subject.

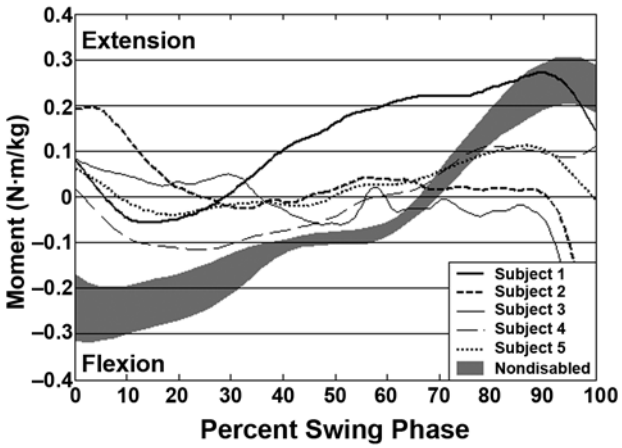

Figure 5.
Mean moments occurring at hip in sagittal plane during swing phase for each subject. Gray band indicates hip moments (mean ± 1 standard deviation) from nondisabled persons.

during double support and used to bear a large portion of body weight during swing/flight. In people with lower-limb paralysis, the upper body appears to produce substantial power for both gaits. This finding may explain why Thomas et al. did not find a significant difference in oxygen cost between children who used a swing-through gait pattern and children who used a reciprocating pattern with an RGO [26]. Based on these considerations, perhaps a reciprocal gait could be considered a modified form of swing-through gait that alternately drags one leg behind the other.

Although the RGO renders the stance leg suitable for weight bearing by passively holding the lower limbs and torso upright, RGO users continue to bear a large portion of their body weight through the arms during single-limb support. Weight bearing through the arms likely contributes to the high energy expenditure associated with RGO ambulation, since studies have shown that upper-body musculature produces power less efficiently than lower-body musculature [27–29]. Our data suggest that the trunk posture of RGO users may encourage arm loading. All subjects in this study walked with a flexed trunk, extending during single support presumably to flex the swing leg. With the trunk flexed, forces at the hip promoted trunk flexion moments, but forces at the shoulder promoted trunk extension moments. Therefore, for RGO users to extend their trunk, forces at the shoulder must be

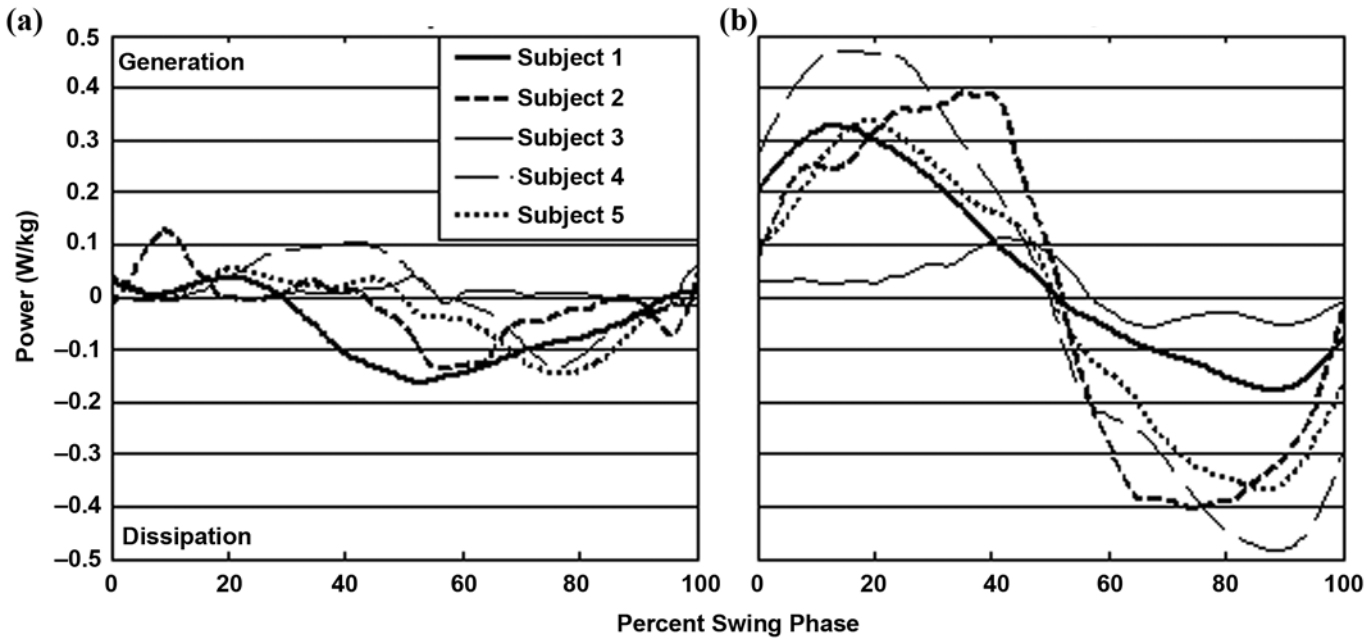

Figure 6.
Mean sagittal plane power from **(a)** hip moments and **(b)** hip forces during swing phase for each subject.

increased to counteract the trunk flexion moments created by forces at the hip. This trunk extension is accomplished by users shifting their weight onto the walking aids. The net effect of this arm loading is that moments about the trunk are approximately balanced (**Figure 4**). Although, the moments created about the trunk COM by shoulder joint reaction forces are not on their own large enough to extend the trunk, they do contribute. Upper-body muscle action, which was not reported in this study, may also contribute to trunk extension.

Trunk orientation also alters the moment arm between the trunk COM and the hip and shoulder joints. If the trunk were in an extended position, the moment arm would change so that an upward force from the hip joint would lead to the desired trunk extension motion. This orientation may encourage weight-bearing through the stance leg rather than the walking aids, which in turn could decrease metabolic energy expenditure. Though not presently clear, RGO users may adopt a flexed trunk to ensure stability by increasing the base of support created by their feet and walking aids. The choice of walking aid and user technique should be further explored with respect to metabolic energy expenditure.

Trunk posture may also affect function of the reciprocal link, which is intended to harness flexion at one hip to facilitate extension of the other hip, and vice versa. However, since the trunk and pelvis are rigidly linked by the RGO, trunk flexion produces stance hip flexion, which in turn encourages swing hip extension through engagement of the reciprocal link. Our data demonstrate that for all subjects, the stance leg hip was flexed for most of single support and that an extension moment acted on the hip at the beginning of swing phase. This extension moment is antagonistic to the goal of swing, which is to advance the swing leg with hip flexion, and contradicts the intended purpose of the reciprocal link.

Based on an analysis of the tension generated in the cables that form the reciprocal link of an RGO, Dall et al. have also argued that the link does not assist in leg swing as traditionally thought [30]. Our results indicated that all subjects experienced small hip flexion moments compared with nondisabled persons. Without active hip flexors (or the presence of weak hip flexors—subjects 4 and 5), these hip flexion moments are assumed to be the result of the reciprocal link. However, the moment is too small to contribute substantially to swing, with the power generated by moments at the hip being much smaller than the power generated by hip-joint forces. Apparently, trunk and pelvis motion contributes more to leg swing than elements of the orthosis such as the reciprocal link. Hence,

our results concur with those of Dall et al. in suggesting that the reciprocal link is minimally important in leg swing [30].

As with many mechanical systems, a certain degree of backlash is present in the reciprocal link. Ballistic movement of the swing leg within the backlash of the reciprocal link may help explain the smaller moments calculated at the hip and the larger power generated by the hip-joint reaction forces than by hip moments. The motion of the pelvis and the force of gravity may flex the leg before the reciprocal link can fully engage. Trunk flexion may predispose the link to hip extension so that the link has to overcome the entire extent of the backlash before it can promote hip flexion. If the trunk were extended at the beginning of swing phase, the reciprocal link may be engaged in flexion and contribute more to leg swing.

We do not advocate that the reciprocal link is completely without merit; our current knowledge about the function of the reciprocal link is too limited to assess definitively. Several authors have hypothesized that the link may have some utility during the double-support phase of walking [17,30]. For example, Dall et al. suggested that the link may help keep the trunk upright by preventing bilateral hip flexion [30]. Unfortunately, this study cannot contribute further to exploring the action of the reciprocal link during double support because of insufficient data during this gait cycle period. In some of our subjects, because of small step lengths or the use of walking aids, force-plate measurements could only be recorded when a foot or walking aid was exclusively and completely located on the force plate, i.e., during single support. Additional studies are required that specifically investigate the role of the reciprocal link in RGO ambulation.

Like many previous studies of RGO ambulation, a limitation of this case series was the small and heterogeneous sample population. Having such a diverse population introduces a multitude of uncontrolled variables, including the different strength to mass ratios between adults and children, different passive ROMs of the joints, varying degrees of spasticity, different levels of muscle control and sensation, and different bases of support and maneuverability provided by the various walking aides. While little research has been conducted to quantify the effects of these different variables on the gait mechanics of RGO users, one particular study reported a significant relationship between lesion level and walking speed, hip ROM, and peak crutch force [17]. Therefore, we recognize that these uncontrolled variables can interfere with the interpretation of the results from this investigation. However, we believe that the simplest explanation for the observed consistency in certain gait characteristics across this study's sample population comes from the traits the subjects have in common and not from their differences. Indeed, with such a diverse population, the list of common traits among the subjects is small but vital, consisting of the RGO itself. In this way, the diversity in the subject sample along with the consistency of gait characteristics increases this study's generalizability because it suggests that aspects of ambulation exist with an RGO that are consistent across RGO users regardless of age, lesion level, type of RGO, and walking aid used.

Sources of measurement error included marker placement and assumptions regarding anthropometry. Marker placement on the pelvis was particularly difficult for the subjects who used an isocentric RGO (IRGO), because the corset and rocker bar prevented direct placement of a marker on the sacrum as required by the Helen Hayes marker set [24]. Since the pivot point of the IRGO rocker bar was located over the sacrum, the sacral marker was placed on the pivot, separating the sacral marker from the sacrum by about 3 cm. As a result, relative motion between the orthosis and the subject could affect measurements of pelvic tilt and hip flexion and, consequently, introduce error in the calculations of hip flexion moments and rotational hip-joint power. However, the average relative motion between the sacral marker on the orthosis and the pelvic markers on the subjects was found to be <1 cm, which is comparable with the relative motion found between two markers placed directly on the skin of the lower limbs of nondisabled people [31]. Therefore, the error introduced by the marker placement used in this study is no greater than that in gait analyses that use the traditional Helen Hayes marker set.

We used anthropomorphic approximations for nondisabled people to estimate the masses, COMs, and radii of gyration of the lower limbs because such information could not be found for people with lower-limb paralysis. Atrophy of the lower limbs due to disuse renders these estimates prone to error, particularly overestimation of segment masses. Overestimating the leg mass inflates the hip-joint forces and moments and, consequently, the power generated at the hip. However, the possibility that hip moments may actually be smaller than those calculated strengthens the argument that small moments at the hip

indicate limited action of the reciprocal link in RGO ambulation. The errors associated with inaccurate COM and radii of gyration are more difficult to predict. If accelerations are small, errors are probably not as significant.

Although none of our subjects requested a rest period during the data collection trials, the known exhaustive nature of ambulating with RGOs prevents the dismissal of fatigue as a possible source of error. However, little is known about how fatigue affects the gait pattern of RGO users.

The results of this case series suggest a number of additional investigations are required to further increase our understanding of RGO gait. Studies should further investigate the effects of posture on arm loading, energy expenditure, and the reciprocal link to determine if trunk flexion contributes to arm loading and increases metabolic energy expenditure. Additionally, we need to determine whether users who ambulate with an extended trunk posture may increase hip flexion moments during swing through action of the reciprocal link. Furthermore, backlash within the reciprocal link and its effect on hip moments should also be investigated. This additional information may improve RGO design or user technique and therefore result in more functional and efficient ambulation with RGOs.

CONCLUSIONS

Analyses of the dynamics of ambulation with RGOs indicate consistent gait characteristics across users, such as persistent trunk flexion and the pattern of load bearing during the single-support phase of stance. The results of this case series suggest that the relationship between posture, forces acting on the walking aids, and the action of the reciprocal link should be explored further because these factors likely contribute to the high metabolic cost of ambulation with an RGO.

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