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Arm and leg coordination during treadmill walking in individuals with motor incomplete spinal cord injury: A preliminary study

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Abstract

Arm and leg coordination naturally emerges during walking, but can be affected by stroke or Parkinson's disease. The purpose of this preliminary study was to characterize arm and leg coordination during treadmill walking at self-selected comfortable walking speeds (CWSs) in individuals using arm swing with motor incomplete spinal cord injury (iSCI). Hip and shoulder angle cycle durations and amplitudes, strength of peak correlations between contralateral hip and shoulder joint angle time series, the time shifts at which these peak correlations occur, and associated variability were quantified. Outcomes in individuals with iSCI selecting fast CWSs (range, 1.0–1.3 m/s) and speed-matched individuals without neurological injuries are similar. Differences, however, are detected in individuals with iSCI selecting slow CWSs (range, 0.25–0.65 m/s) and may represent compensatory strategies to improve walking balance or forward propulsion. These individuals elicit a 1:1, arm: leg frequency ratio versus the 2:1 ratio observed in non-injured individuals. Shoulder and hip movement patterns, however, are highly reproducible (coordinated) in participants with iSCI, regardless of CWS. This high degree of inter-extremity

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coordination could reflect an inability to modify a single movement pattern post-iSCI. Combined, these data suggest inter-extremity walking coordination may be altered, but is present after iSCI, and therefore may be regulated, in part, by neural control.

1. Introduction

Arm swing is not required for walking [1, 2], but naturally emerges and is coordinated with the legs [3]. Amplitude and frequency of arm swing depends on walking speed, yet arm swing retains a phase relationship with the legs regardless of ambulatory velocity [3]. While many report changes in walking speed, coupled with altered spatiotemporal parameters and/or biomechanics following spinal cord injury (SCI) [4, 5]; few studies investigate the impact of SCI on walking-related arm swing. Such studies are important and timely as emerging evidence suggests arm integration during rhythmical and reciprocal activities may alter leg movement via spinal pathways connecting and coordinating the arms and legs [6–10].

The impact of SCI on inter-extremity walking coordination is unknown, yet effects of other neurological impairments are documented. Synchronization of leg and arm movements has been detected post-stroke [11, 12], however decreased synchronization also has been reported following stroke [13, 14] and onset of Parkinson's disease [15, 16]. To our knowledge, only our group is investigating arm swing during walking post-iSCI. We recently described varied arm swing presentation during walking on a treadmill without assistive devices post-injury [17]. The current study furthers this characterization by describing arm and leg coordination. For comparison, we first evaluated walking speed effects on inter-extremity coordination when the nervous system is intact. Our results are consistent with studies showing speed influences arm amplitude and frequency; and inter-extremity coordination, or coupling between the arms and legs becomes tighter with increasing speed [2, 3, 18, 19]. Because spinal circuitry can contribute to automatic, rhythmical, and reciprocal patterns like stepping, we hypothesized arm and leg coordination would be tight in individuals with iSCI selecting CWSs similar to CWSs seen in individuals without neurological injury (1.0–1.3 m/s) [20]. We further hypothesized arms and legs would be less coordinated in individuals with iSCI selecting slower CWSs (0.25–0.65 m/s), due to variability associated with slower walking and greater impairment.

2. Methods

2.1 Subjects

Eleven individuals demonstrating walking-related arm swing with iSCI (American Spinal Injury Association Impairment Scale grade C/D, International Standards for Neurological Classifications of SCI (ISNCSCI) [21] with neurological level of impairment at or below C4; 4 females, 7 males) and eleven controls (5 females, 6 males) (Table 1) were recruited from a sample of convenience (Brain Rehabilitation Research Center, Veterans Affairs Medical Center (VAMC), Gainesville, FL). Institutional and federal regulations concerning ethical use of human volunteers were followed using University of Florida and VAMC-approved protocols. Participants provided informed consent.

2.2 Experimental Procedure and Analyses

Individuals were acclimated to the instrumented treadmill during practice walking trials. Individuals with iSCI identified their CWS and reconfirmed their speed choice during another trial, prior to recording 30 seconds of steady-state walking data. CWSs for participants with iSCI were categorized based on a defined speed range (0.25–0.65 m/s versus 1.0–1.3 m/s), as a slow CWS or fast CWS. The fast CWSs correspond to the CWS range for adults without neurologic injury [20]; whereas the slow CWSs fall in a range

comparable to household or limited community ambulation [22, 23]. One individual with iSCI required manual assistance at one leg. All others walked independently. Overhead supports and safety harnesses providing partial body weight support permitted walking without assistive devices and allowed natural arm swing. Thirty second trials during steady-state walking also were collected from controls at walking speeds ranging from 0.3–1.3 m/s for speed-matching (within ± 0.15 m/s). For comparison, only one speed (randomly-selected) for each control is reported (Table 1). Bilateral 3-dimensional ground reaction forces (GRFs) were recorded at 2000 Hz and 3-dimensional joint kinematics collected at 100 Hz using a modified Helen Hayes marker set and twelve-camera motion analysis system (Vicon; Oxford, UK).

Marker trajectories were labeled and data imported into Visual 3D (C-Motion; Germantown, MD). Raw kinematic and GRF data were low-pass filtered (4th-order zero-lag Butterworth with cut-off frequencies of 6 and 20 Hz); and joint angles and GRFs normalized by body weight were imported into Matlab (MathWorks; Natick, MA). Joint angles were normalized to 100% of the gait cycle and averaged for the number of steps taken per 30 second walking trial. Custom Matlab programming permitted identification of hip and shoulder movement cycles from local maxima in the sagittal plane angle time series to determine cycle durations and amplitudes [12] (Figure 1A). At speeds from 0.3–0.8 m/s, two oscillatory peaks in the shoulder angle occasionally were present for every one hip angle oscillation. Some second peaks were small relative to the first and/or seemed extensions of the first cycle rather than independent cycles. As a result, thresholds were defined to determine whether second peaks would be classified as individual arm cycles. Thresholds required successive shoulder joint cycles to have amplitude ranges of at least 50% of previous cycles, with minima comparable to minima of previous and subsequent cycles (Figure 1A). If thresholds were not met (Figure 1B), shoulder joint movements were not counted as individual cycles. The strength of the peak correlation (cross-covariance correlation, r) between contralateral hip and shoulder joint angle time series and the time shift at which this peak correlation occurred (cross-covariance lag, t) were determined using cross-covariance analysis [12].

Outcomes within each 30 second walking trial were averaged and standard deviations calculated. Left and right side values within individuals were averaged since paired t-tests revealed non-significant differences. Arm:leg frequency ratios were determined by dividing hip angle by shoulder angle cycle durations and a two-factor ANOVA (unequal replication, model degrees of freedom (df)=3, error df=18) was used to test group and speed interactions. Group and speed interactions for other outcomes were tested with two-factor ANCOVAs (unequal replication, model df=3, error df=18). If no significant interaction was identified, a standard ANCOVA model (model df=2, error df=19) was performed to investigate group and speed effects.

3. Results

3.1 iSCI alters kinematic patterns of spontaneous arm swing

Sagittal plane joint angles from 4 individuals with iSCI and 4 controls illustrate examples of observed responses in hip and shoulder angles in the plane of forward progression (Figure 2A). These data suggest the most notable differences observed between groups may be manifested in arm:leg frequency at slower CWSs. Individuals with iSCI produce 1 arm swing cycle (shoulder oscillation) per gait cycle, regardless of CWS. In contrast, controls produce 2 arm swing cycles per gait cycle at walking speeds ≤ 0.5 m/s and transition to 1 arm swing cycle at walking speeds ≥ 0.8 m/s (Figure 2B).

Quantitative data in which hip angle cycle durations are divided by shoulder angle cycle durations to calculate arm:leg frequency ratios (Figure 2C) provide the same results. Leg

cycle durations appear approximately two times arm cycle durations in controls walking at 0.5 m/s, thereby resulting in the 2:1, arm:leg frequency ratio (Figure 2C). In contrast, cycle durations of the arms and legs are nearly equivalent, resulting in a 1:1 frequency ratio at walking speeds 0.8 m/s (Figure 2C). Interestingly, this 1:1 frequency ratio also is present in all individuals with iSCI, irrespective of CWS (Figure 2C). A two factor ANOVA with interaction illustrates a group (iSCI versus uninjured) and speed interaction ($F=3915.45$, $p<0.0001$). Differences between low (<0.8 m/s) and high (>0.8 m/s) speeds within groups suggest large differences within controls (Estimated difference=1.0224; $F=7903.07$, $p<0.0001$) and essentially no difference within individuals with iSCI (Estimated difference=0.0047; $F=0.17$, $p=0.6889$).

3.2 Walking coordination is present post-iSCI

CWSs in individuals with iSCI naturally divided into two groups, 0.65 m/s (range 0.25–0.65 m/s, slow CWS) and 1.0 m/s (range 1.0–1.3 m/s, fast CWS). This is consistent with divisions of 0.8 m/s [22] and 0.88 m/s [23] identifying household versus community ambulators, as well as speed-dependent differences in arm:leg frequency ratios in controls [3]. Irrespective of CWS, however, inter-extremity coordination is present following iSCI as demonstrated by similarities between arm and leg kinematic patterns and step-to-step consistency.

Correlations of cross-covariance between contralateral hip and shoulder joint angle time series gradually become closer to 1 as walking speeds increase in controls (see Figure 3). Surprisingly, these values are close to 1 for all individuals with iSCI, regardless of CWS (Figure 3A), and reflect the statistically significant interactions detected between group and speed ($F=35.24$, $P<0.0001$). Similarly, group and speed interactions for time shifts at which the peak correlations between contralateral hip and shoulder angles occur also are statistically significant ($F=9.73$, $P=0.0059$). These time shifts are closer to 0 for individuals with iSCI, compared to controls, particularly at speeds 0.65 m/s (Figure 3C). Interestingly, time shifts for individuals with iSCI commonly are negative (Figure 3C), demonstrating maximal shoulder flexion occurs before maximal hip flexion (Figure 1A).

The variability associated with joint angle correlations also reveal statistically significant group and speed interactions (Figure 3B, $F=15.87$, $P=0.0009$); while interactions between group and speed are marginally significant ($F=3.65$, $P=0.0722$) for the variability associated with time shifts (Figure 3D). Decreases in these two types of variability are observed with increases in walking speed for both groups but are most distinct for controls since these individuals have greater disparities between variability associated with slow versus fast walking speeds.

3.3 Post-iSCI, hip and arm swing amplitudes remain speed-dependent

ANCOVAs reveal insignificant group and speed interactions for shoulder and hip angle amplitudes ($F=2.96$, $P=0.1025$ and $F=0.23$, $P=0.6383$, respectively) and associated shoulder and hip angle variability ($F=0.93$, $P=0.3473$ and $F=0.03$, $P=0.8753$, respectively). However, ANCOVA models without interaction terms reveal significant associations exist for joint angle amplitudes and speed (shoulder: $F=6.72$, $P=0.0179$ and hip: $F=50.58$, $P<0.0001$, respectively), but not group (shoulder: $F=1.61$, $P=0.2201$ and hip: $F=0.67$, $P=0.4221$, respectively). Therefore, it appears that like controls, the amplitudes of shoulder and hip joints become larger with increases in walking speed among individuals with iSCI (Figure 4A,C). Shoulder and hip angle variability, however, does not appear to be associated with either group ($F=2.95$, $P=0.1019$ and $F=1.95$, $P=0.1786$, respectively) or speed ($F=2.94$, $P=0.1027$ and $F=1.65$, $P=0.2146$, respectively), suggesting differences in the amount of variability between groups and across speeds are insignificant (Figure 4B,D).

4. Discussion

To our knowledge, this is the first study to characterize arm and leg walking coordination post-iSCI. Although preliminary, trends in outcomes observed in individuals with iSCI appear similar to trends observed from individuals without injury during treadmill walking at different speeds. However, distinct differences in coordination patterns are identified.

4.1 Coordination is present, but not speed-dependent post-iSCI

In individuals without neurological injury, arm and leg coordination patterns depend on walking speed. At relatively slow speeds (~ 0.7 – 0.8 m/s) both arms typically swing minimally and in-phase at twice the ipsilateral step frequency (producing a 2:1 frequency ratio) [3, 18, 24]. At speeds ~ 0.7 – 0.8 m/s, this pattern transitions to a 1:1 frequency ratio with out-of-phase arm swing. Each arm is paired with its contralateral leg and synchronized with stride frequency. In this study, all individuals with iSCI produce a 1:1, arm:leg frequency ratio where each arm swings in-phase with the contralateral leg, regardless of CWS. As a result, the speed-dependent transition from a 2:1 to 1:1 frequency ratio observed in controls is absent. Differences in arm:leg frequency ratios only were detected between individuals with iSCI with slow CWSs (0.65 m/s) and speed-matched controls. Differences were not observed between individuals with iSCI with fast CWSs and speed-matched controls. Individuals with iSCI were matched to controls based on speed, however they were not matched on attributes such as age, height, and weight. Therefore, it is possible that influences other than injury confounded results and contributed to group differences [25]. Evaluating individuals with iSCI at multiple walking speeds and matching other attributes would help determine whether differences in coordination patterns are attributable to SCI, post-iSCI changes in CWS, and/or other gait-influencing factors. Because many individuals fatigue easily and/or have difficulty walking at faster speeds, testing at multiple walking speeds in this population is difficult. One individual with iSCI, however, was tested at multiple walking speeds: 0.30 m/s (slow), 0.65 m/s (CWS), and 0.85 m/s (fast). In addition, two other individuals with iSCI were tested at their CWS (0.4 m/s, a slow speed) versus a fast speed (0.8 m/s or 1.1 m/s). It is noteworthy that at all speeds, each of these individuals produced a 1:1 arm:leg frequency ratio.

Irrespective of differences detected between individuals with slow CWSs and controls, walking coordination was present in all individuals with iSCI evaluated. This coordination, however, is manifested in a different pattern between the arms and legs at slow CWSs. Despite this, the consistency of the walking pattern appears relatively unaffected by iSCI. Step-by-step outcome variability remains low, illustrating the high degree of reproducible arm and leg patterns, regardless of CWS post-iSCI. Interestingly, individuals with iSCI walking at slow CWSs demonstrate tighter arm and leg coupling than speed-matched controls. In controls, the correlations of cross-covariance between contralateral hip and shoulder joint angle time series and the time shifts at which these peak correlations occur slowly approach 1 and 0, respectively, as speed increases. However, the correlations and time shifts are close to 1 and 0, respectively, at all CWSs in individuals with iSCI. These highly coordinated arm and leg movement patterns may represent the nervous system's "solution" to walking post-iSCI. Alternatively, the patterned consistency may indicate an inability of individuals with iSCI to make adaptations outside of their reproducible, and almost stereotypical, movement patterns.

4.2 Altered coordination patterns post-iSCI may emerge as compensatory strategies to improve walking function

Though walking is possible without arm swing [1, 2], it may reduce metabolic cost by enhancing stability [26] and/or assist in transforming energy during walking [27]. Arm

swing generates a horizontal torque at the upper trunk which may (1) counteract pelvis rotation and leg progression, (2) minimize angular momentum, and (3) reduce the reaction moment about the foot [1, 26, 28, 29]. Therefore, the 1:1, arm:leg frequency pattern used post-iSCI with slow CWSs may be a compensatory strategy to enhance balance. Experience post-iSCI [17], attributes other than spinal injury (e.g. age, height, weight) known to influence gait [25], and/or the testing environment, however, also should be given equal consideration as factors potentially influencing the results of this study.

While substantial literature supports 2:1 coupling between the arms and legs at slow walking speeds, this relationship is not detected among all (neurologically-intact) individuals in the treadmill environment [19]. Caprinella et al. (2009) suggest this may be attributed to differences in temporal relationships between cadence and resonant arm frequency, since studies have demonstrated treadmill walking, compared to overground walking, can result in increased cadence [19, 30]. However, because all controls walking at slow treadmill speeds in our study elicit the 2:1 pattern and none of the individuals with iSCI demonstrate this same pattern, it is unlikely that differences detected are due to our testing environment alone. Regardless, to rule out the differing influences treadmill versus overground walking may have on arm-to-leg coordination, cadence and resonant arm frequencies in controls and individuals with iSCI should be evaluated and compared in both environments for future studies.

In controls, arm swing has been described as movement powered predominately by the legs, where force is transferred to the arms between trunk and shoulder ligaments and muscles during walking [31]. Once leg movement is impaired, the reverse may become true. Enhancing arm swing amplitude or altering its pattern may help generate forward propulsion at the foot. Interestingly, time shifts at which peak correlations between contralateral hip and shoulder angles occur are negative, indicating maximal shoulder flexion occurs before maximal hip flexion. Since arm swing counteracts rotation at the pelvis and progression of the legs, this may be one example of a strategy employed by individuals with iSCI to compensate for limited extension at the hip and assist with forward progression.

5. Conclusion

A high degree of coordination occurs during treadmill walking post-iSCI, regardless of CWS. Despite this, alterations in arm swing expression and inter-extremity coordination are observed, particularly in individuals with iSCI with slow CWSs. Some neural components, therefore, appear involved with the regulation of arm swing and inter-extremity walking coordination. Focusing subsequent investigations on the portion of the slower walking iSCI population may be most beneficial, since this is where greater alterations are detected. In addition, treadmill versus overground environments can impact gait patterns, and therefore this influence on inter-extremity coordination should be investigated. To elucidate associated mechanisms, future investigations also should consider the effects of restricting walking-related arm swing on head, pelvis, and trunk movement; movement of the hip and shoulder joints outside the plane of progression; and walking balance and propulsion. Furthermore, studies investigating arm swing and modifications made at the arms during more challenging walking tasks requiring adaptations or greater neural demand could provide insight into the neural control regulating inter-extremity coordination. Data from this study and future studies will continue to elucidate spinal connections between the arms and legs and their role during walking. As a result, we may gain insight into a useful neural substrate for rehabilitation to promote recovery following iSCI.

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References

1. Umberger BR. Effects of suppressing arm swing on kinematics, kinetics, and energetics of human walking. *J Biomech.* 2008; 41(11):2575–2580. [PubMed: 18621376]
2. Ford MP, Wagenaar RC, Newell KM. Arm constraint and walking in healthy adults. *Gait Posture.* 2007; 26(1):135–141. [PubMed: 16997561]
3. Craik, R.; Herman, R.; Finley, F. Advances in behavioral biology. In: Herman, R.; Grillner, S.; Stein, P.; Stuart, D., editors. Human solutions for locomotion: Interlimb coordination. New York: Plenum Press; 1976. p. 51-64.
4. Pepin A, Norman KE, Barbeau H. Treadmill walking in incomplete spinal-cord-injured subjects: 1. Adaptation to changes in speed. *Spinal Cord.* 2003; 41(5):257–270. [PubMed: 12714988]
5. Leroux A, Fung J, Barbeau H. Adaptation of the walking pattern to uphill walking in normal and spinal-cord injured subjects. *Experimental Brain Research.* 1999; 126(3):359–368.
6. Visintin M, Barbeau H. The effects of parallel bars, body weight support and speed on the modulation of the locomotor pattern of spastic paretic gait. A preliminary communication. *Paraplegia.* 1994; 32(8):540–553. [PubMed: 7970859]
7. Behrman AL, Teitelbaum P, Cauraugh JH. Verbal instructional sets to normalise the temporal and spatial gait variables in parkinson's disease. *J Neurol Neurosurg Psychiatry.* 1998; 65(4):580–582. [PubMed: 9771792]
8. Huang HJ, Ferris DP. Neural coupling between upper and lower limbs during recumbent stepping. *J Appl Physiol.* 2004; 97(4):1299–1308. [PubMed: 15180979]
9. Kawashima N, Nozaki D, Abe MO, Nakazawa K. Shaping appropriate locomotive motor output through interlimb neural pathway within spinal cord in humans. *J Neurophysiol.* 2008; 99(6):2946–2955. [PubMed: 18450579]
10. Stephenson JL, De Serres SJ, Lamontagne A. The effect of arm movements on the lower limb during gait after a stroke. *Gait Posture.* 2010; 31(1):109–115. [PubMed: 19854654]
11. Ford MP, Wagenaar RC, Newell KM. Phase manipulation and walking in stroke. *J Neurol Phys Ther.* 2007; 31(2):85–91. [PubMed: 17558362]
12. Stephenson JL, Lamontagne A, De Serres SJ. The coordination of upper and lower limb movements during gait in healthy and stroke individuals. *Gait Posture.* 2009; 29(1):11–16. [PubMed: 18620861]
13. Ford MP, Wagenaar RC, Newell KM. The effects of auditory rhythms and instruction on walking patterns in individuals post stroke. *Gait Posture.* 2007; 26(1):150–155. [PubMed: 16996270]
14. Wagenaar RC, van Emmerik RE. Dynamics of pathological gait. *Hum Mov Sci.* 1994; 13(3–4): 441–471.
15. Dietz V, Michel J. Locomotion in parkinson's disease: Neuronal coupling of upper and lower limbs. *Brain.* 2008; 131(Pt 12):3421–3431. [PubMed: 18930965]
16. Carpinella I, Crenna P, Marzegan A, Rabuffetti M, Rizzone M, Lopiano L, et al. Effect of l-dopa and subthalamic nucleus stimulation on arm and leg swing during gait in parkinson's disease. *Conf Proc IEEE Eng Med Biol Soc.* 2007; 2007:6665–6668. [PubMed: 18003554]
17. Tester NJ, Howland DR, Day KV, Suter SP, Cantrell A, Behrman AL. Device use, locomotor training and the presence of arm swing during treadmill walking after spinal cord injury. *Spinal Cord.* 2010
18. Wagenaar RC, van Emmerik RE. Resonant frequencies of arms and legs identify different walking patterns. *J Biomech.* 2000; 33(7):853–861. [PubMed: 10831760]

19. Carpinella I, Crenna P, Rabuffetti M, Ferrarin M. Coordination between upper- and lower-limb movements is different during overground and treadmill walking. *European Journal of Applied Physiology*. 2009; 108(1):71–82. [PubMed: 19756711]
20. Waters RL, Lunsford BR, Perry J, Byrd R. Energy-speed relationship of walking: Standard tables. *J Orthop Res*. 1988; 6(2):215–222. [PubMed: 3343627]
21. Ditunno JF Jr, Burns AS, Marino RJ. Neurological and functional capacity outcome measures: Essential to spinal cord injury clinical trials. *J Rehabil Res Dev*. 2005; 42(3 Suppl 1):35–41. [PubMed: 16195961]
22. Perry J, Garrett M, Gronley JK, Mulroy SJ. Classification of walking handicap in the stroke population. *Stroke*. 1995; 26(6):982–989. [PubMed: 7762050]
23. van Hedel HJ, Group ES. Gait speed in relation to categories of functional ambulation after spinal cord injury. *Neurorehabil Neural Repair*. 2009; 23(4):343–350. [PubMed: 19036717]
24. Donker SF, Beek PJ, Wagenaar RC, Mulder T. Coordination between arm and leg movements during locomotion. *J Mot Behav*. 2001; 33(1):86–102. [PubMed: 11303522]
25. Røislien J, Skare Ø, Gustavsen M, Broch NL, Rennie L, Opheim A. Simultaneous estimation of effects of gender, age and walking speed on kinematic gait data. *Gait & Posture*. 2009; 30(4):441–445. [PubMed: 19665379]
26. Ortega JD, Fehlman LA, Farley CT. Effects of aging and arm swing on the metabolic cost of stability in human walking. *J Biomech*. 2008; 41(16):3303–3308. [PubMed: 18814873]
27. Gracovetsky SA. The spinal engine. 2009
28. Li Y, Wang W, Crompton RH, Gunther MM. Free vertical moments and transverse forces in human walking and their role in relation to arm-swing. *J Exp Biol*. 2001; 204(Pt 1):47–58. [PubMed: 11104710]
29. Park J. Synthesis of natural arm swing motion in human bipedal walking. *Journal of Biomechanics*. 2008; 41(7):1417–1426. [PubMed: 18417138]
30. Nymark JR, Balmer SJ, Melis EH, Lemaire ED, Millar S. Electromyographic and kinematic nondisabled gait differences at extremely slow overground and treadmill walking speeds. *J Rehabil Res Dev*. 2005; 42(4):523–534. [PubMed: 16320147]
31. Pontzer H, Holloway JHt, Raichlen DA, Lieberman DE. Control and function of arm swing in human walking and running. *J Exp Biol*. 2009; 212(Pt 4):523–534. [PubMed: 19181900]

Figure 1.

Outcomes. Data presented are continuous data from one representative healthy control walking on the treadmill at 0.5 m/s (A) and 0.8 m/s (B). Hip and shoulder angle cycle durations (α and χ , respectively) are defined as the time between consecutive maxima in the sagittal plane hip (gray, right hip angle) and shoulder (black, left shoulder angle) angle time series. Hip and shoulder angle amplitudes (β and δ) are defined as the maximum range in sagittal hip and shoulder angles between consecutive maxima. In the data presented here, peak shoulder flexion (a) occurs before contralateral peak hip flexion (b), indicating that lag times for the movement cycles illustrated are negative. Conversely, when peak hip flexion occurs prior to contralateral peak shoulder flexion, lag times are positive (not shown). Lag

times of 0, therefore indicate peak flexion of the contralateral hip and shoulder occurs at the same time. Controls walking at speeds <0.8 m/s sometimes elicit two oscillatory peaks in the shoulder for every one movement cycle at the hip. Thresholds, defined in the methods, were used to determine independent cycles of arm swing. Examples of two independent cycles of movement at the shoulder (a and c) are illustrated in A, and contrasted with an example where only the first oscillation in shoulder movement (d, and not e) is defined as an independent arm cycle in B.

Figure 2.

Sagittal plane joint angles and related arm:leg frequency ratios. Average hip (A) and shoulder (B) angles normalized to 100% of the gait cycle for individuals with iSCI (right, Panels A and B) and speed-matched uninjured controls (left, Panels A and B) from 8 individuals (4 iSCI and 4 controls) are presented. Arm:leg frequency ratios for all controls and individuals with iSCI (C) were calculated by dividing hip angle cycle durations by shoulder angle cycle durations and illustrate the same ratio of movement patterns observed in the arms and legs from angular joint kinematics (A and B). A cubic spline regression model was fit to the data to illustrate trends in the two groups. Panels A and B: red – left side; blue – right side. Error bars denote standard deviation and arrows indicate directions of joint angle flexion and extension.

Figure 3.

Arm and leg coordination during treadmill walking. The strength of the peak correlation between the hip and contralateral shoulder joint angle time series (cross covariance correlation, r , represented in "A") and the time shift at which this peak correlation occurs (cross covariance lag, t , represented in "C") are presented on the left; while associated variability for each outcome is presented on the right (B and D, respectively). Time shifts, or lags, of 0 indicate an in-phase pattern of coordination where maximal hip and shoulder flexion between the contralateral arm and leg occur at the same time; while time shifts, or lags, of 1 indicate an out-of-phase pattern of coordination. Negative time shifts indicate maximal shoulder flexion occurs before maximal hip flexion, and positive time shifts indicate maximal hip flexion occurs before maximal shoulder flexion. Each data point represents one individual; lines represent regressions calculated from ANCOVA models.

Figure 4.

Shoulder and hip angle amplitudes and associated variability during treadmill walking. Shoulder joint (A) and hip joint (C) angle amplitudes are presented on the left; while each outcome's associated variability is presented on the right (B and D, respectively). Each data point represents one individual; lines represent regressions calculated from ANCOVA models.

Table 1

iSCI (n=11)	Etiology	Time since iSCI (mos)	Neurological Level	ISNCSCI Classification	ISNCSCI LEMS max=50	ISNCSCI UEMS max=50	WISCIII Score	Overground Assistive Device	Gender	Age (yrs)	Height (m)	Weight (kg)	CWS (m/s)
SCI01	Fall from horse	7.5	C4-6	D	45	39	19	Cane	F	51	1.73	68	0.25
SCI02	Bicycle accident	12	C4	D	43	30	16	2 Crutches	F	47	1.57	53	0.35
SCI03	Transverse myelitis	43	C4-5	D	44	44	13	Walker	F	76	1.81	70	0.35
SCI04	MVA fusion, Paralysis after injection	13	T1	D	49	50	20	No Device	M	58	1.81	82	0.40
SCI05	Fall	92	C7	D	48	50	19	Cane	M	67	1.89	93	0.50
SCI06	Motorcycle accident	27	C6	D	45	37	13	Walker	M	23	1.85	77	0.65
SCI07	Stab wound	49	C4-5	D	46	44	20	No Device	F	50	1.63	48	1.00
SCI08	Epidural hematoma	33	C5-T9	D	45	50	19	Cane	M	54	1.86	99	1.10
SCI09	Surfing accident	63	C5	D	50	48	20	No Device	M	21	1.87	76	1.30
SCI10	Unknown	56	C4-6	D	49	43	20	No Device	M	26	1.80	70	1.30
SCI11	Tumor	30	T1-3	D	41	50	19	Cane	M	28	1.89	95	0.25
Mean ± Standard Deviation		39 ± 25	---	---	46 ± 3	44 ± 7	18 ± 3	---	---	46 ± 19	1.79 ± 0.10	75.5 ± 16.3	0.68 ± 0.42

Controls (n=11)	Speed (m/s)	Gender	Age (yrs)	Height (m)	Weight (kg)
C01	0.80	F	19	1.61	80
C02	1.30	M	18	1.79	80
C03	0.30	F	20	1.70	67
C04	1.30	F	47	1.79	61
C05	0.50	M	35	1.76	73
C06	1.30	F	60	1.79	52
C07	1.00	M	63	1.73	74
C08	1.30	M	42	1.76	93
C09	0.50	M	29	1.74	66
C10	1.30	F	21	1.67	58
C11	0.30	M	52	1.89	102
Mean ± Standard Deviation			37 ± 17	1.75 ± 0.1	73 ± 15