A Comparison of Shoulder Joint Forces During Ambulation With Crutches Versus a Walker in Persons With Incomplete Spinal Cord Injury

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Objective: To compare 3-dimensional (3D) shoulder joint reaction forces and stride characteristics during bilateral fore-arm crutches and front-wheeled walker ambulation in persons with incomplete spinal cord injury (SCI).

Design: Cross-sectional cohort study.

Setting: Biomechanics laboratory.

Participants: Fourteen adult volunteers with incomplete SCI recruited from outpatient rehabilitation hospital services.

Interventions: Not applicable.

Main Outcome Measure: Peak force, rate of loading, and force-time integral were compared for each component of the net 3D shoulder joint reaction force during ambulation with crutches and a walker. Stride characteristics were also compared between assistive device conditions.

Results: The largest weight-bearing force was superiorly directed, followed by the posterior force. The superior joint force demonstrated a significantly higher peak and rate of loading during crutch walking (48.9N and 311.6N/s, respectively, vs 45.3N and 199.8N/s, respectively). The largest non-weight-bearing force was inferiorly directed with a significantly greater peak occurring during crutch ambulation (43.2N vs 23.6N during walker gait). Walking velocity and cadence were similar; however, stride length was significantly greater during crutch walking (62% vs 58% of normal).

Conclusions: Shoulder joint forces during assisted ambulation were large. Crutch use increased the superior force but did not increase walking velocity.

Key Words: Assistive devices; Gait; Kinetics; Rehabilitation; Shoulder joint; Spinal cord injuries.

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FUNCTIONAL INTEGRITY OF THE shoulder joint is vital for achieving and maintaining independent mobility in persons with spinal cord injury (SCI). Despite the fact that more than half of the new SCIs that occur in the United States are incomplete lesions,3 research has predominantly focused on the mobility demands associated with wheelchair propulsion in individuals with complete SCI.2-5 The prevalence of disabling shoulder pain in those with incomplete SCI (Frankel grade D), however, has been reported to be nearly as high as that in the complete population,1 despite the fact that their primary means of mobility is ambulation.6 Thus, documentation of the upper extremity (UE) demands associated with ambulation following SCI is imperative.

For the patient with incomplete SCI, partial preservation of lower-extremity function makes ambulation a realistic mobility goal. Waters et al7 reported that 76% of individuals with incomplete paraplegia and 46% of those with incomplete tetraplegia achieve community ambulation, although the majority of these individuals require lower-extremity orthoses to ambulate.8 While the use of ankle-foot and knee-ankle-foot orthoses compensate for weakness at the ankle and knee, a practical orthotic substitution for weakness of the hip extensors and abductors is lacking.9 Consequently, UE assistive devices are commonly prescribed to compensate for reduced strength in the hip and trunk extensor musculature to permit independent ambulation.10

The prescription of an appropriate assistive device is an essential component of ambulatory training following SCI. In the selection of an ambulatory aid, the goal of the therapist and patient is to maximize maneuverability and independence, while maintaining safety and stability.11 Walkers are traditionally thought to provide more stability, and thus maximize safety.10 Crutches are less cumbersome and therefore are believed to allow patients to more easily maneuver small spaces. Patient also may be concerned with decreasing the appearance of disability, and thus prefer crutches to the use of a walker. The long-term effects of crutch or walker use on the upper extremities, however, should also be considered in the selection of ambulatory aids.

The weight-bearing loads associated with assistive device use are substantial. Previously, investigators12,13 reported that the peak vertical loads exerted on a walker average between 39% and 48% of body weight in persons with incomplete SCI. During crutch walking, peak axial loads are reported to be slightly lower (22% to 35% of body weight). These studies, however, documented subjects using only their customary assistive device.12,13 A direct comparison of the demands associated with a single group of subjects using different assistive devices has not been performed. Further, the forces transmitted to the shoulder joint through assistive device use, taking into account the effects of gravity and inertia, have not currently been determined. The purpose of this study, therefore, was to describe and compare 3-dimensional (3D) glenohumeral joint reactions during ambulation with crutches versus a walker in persons with incomplete SCI.
(GHJ) reaction forces and stride characteristics during ambulation with crutches and a front-wheeled walker in persons with incomplete SCI. We hypothesized that GHJ reaction forces and walking velocity would be greater during ambulation with the crutches.

**METHODS**

**Participants**

We recruited individuals with incomplete SCI from the outpatient services at Rancho Los Amigos National Rehabilitation Center (RLANRC). Eleven men and 3 women (5 with tetraplegia, 9 with paraplegia) agreed to participate. Subjects had a mean age of 37 years and average time since SCI of 7 years (table 1). Participants were able to walk a minimum of 15m (50ft) with both a front-wheeled walker and forearm crutches. Individuals who required more than stand-by assistance to ambulate safely or who were unable to ambulate with a reciprocal gait pattern were excluded. Additionally, persons who reported UE pain requiring medical intervention or alteration of daily function also were excluded. Experimental procedures were approved by the institutional review board at RLANRC. All subjects read and signed a statement of informed consent prior to participation.

**Materials and Procedures**

We performed quantitative gait testing with subjects walking reciprocally with forearm crutches and a front-wheeled walker over a 10-m walkway at a self-selected velocity. The central 6m of the walkway were designated by photoelectric light switches to trigger data collection thus avoiding transitions of foot-floor contact. Footswitch data were recorded without an external load. For this measurement, each subject walked on a 6-point manual muscle test (0 to 5) to assess lower-extremity muscle strength. Lower-extremity ASIA motor scores were used only to determine the UE

**Stride analysis.** We measured walking speed, cadence, and stride length using the Stride Analyzer system.\(^a\) This system uses insoles taped to the bottom of each shoe. Insoles contain compression-closing switches located under the heel, first metatarsal head, fifth metatarsal head, and great toe for determining periods of foot-floor contact. Footswitch data were relayed via cables to a lightweight telemetry package (2.3kg) suspended from a belt around the subject’s waist. Signals were transmitted by FM-FM telemetry\(^b\) and digitized for collection on a PDP-11/73 computer\(^c\) at a sampling rate of 2500Hz.

**Force analysis.** 3D forces applied to crutches and a walker during ambulation were determined by fitting the assistive devices with custom-designed instrumentation.\(^f\) Two crutches were instrumented with a 6-component strain gauge load cell\(^d\) mounted along the crutch shaft, just below the handle (fig 1A). Additionally, foil strain gauges were mounted to the crutch shaft above the handle in order to measure forces applied through the forearm cuff. A front-wheeled walker was instrumented with similar 6-component load cells placed below the walker handles bilaterally (fig 1B).

We adjusted the height of the instrumented crutches and walker so that the handle of the assistive device was at the level of the subject’s wrist joint while the subject was standing, with the arms resting at the side of the body and the elbows flexed approximately 15°.\(^e\)\(^,\)\(^f\) Immediately prior to data collection, a baseline measurement of the assistive device forces was recorded without an external load. For this measurement, each crutch was supported in a stand, constructed to maintain the devices in a vertical position. Force data signals were relayed via cables to a multichannel patient-interface box\(^a\) (=1.0kg) suspended from a second belt around the subject’s waist. A thin coaxial cable transmitted force data from the connector box to the acquisition system. Similar to the footswitch data, force data were digitized and recorded at 2500Hz.

For the purpose of this investigation, we performed analysis of shoulder joint forces only for the UE opposite the weaker lower extremity. Lower-extremity strength was defined by the American Spinal Injury Association (ASIA) motor score, which utilizes a 6-point manual muscle test (0 to 5) to assess the L2 to S1 myotomes: hip flexors (L2, iliopsoas), knee extensors (L3, quadriceps), ankle dorsiflexors (L4, tibialis anterior), long toe extensors (L5, extensor hallucis longus), and ankle plantarflexors (S1, gastrocnemius, soleus).\(^f\) Lower-extremity ASIA motor scores were used only to determine the UE

**Table 1: Subject Demographic Data**

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>Sex</th>
<th>Age (y)</th>
<th>Body Mass (kg)</th>
<th>Years Since SCI</th>
<th>Level of Injury</th>
<th>Weak LE ASIA</th>
<th>Strong LE ASIA</th>
<th>Total LE ASIA</th>
<th>Customary Assistive Device</th>
<th>Customary Orthoses</th>
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<tr>
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<td>86</td>
<td>1.6</td>
<td>C7</td>
<td>20</td>
<td>23</td>
<td>43</td>
<td>2 CR</td>
<td>AFO</td>
</tr>
<tr>
<td>2</td>
<td>M</td>
<td>21</td>
<td>60</td>
<td>1.3</td>
<td>L1</td>
<td>7</td>
<td>10</td>
<td>17</td>
<td>2 CR</td>
<td>AFO</td>
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<tr>
<td>3</td>
<td>F</td>
<td>59</td>
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<td>9</td>
<td>12</td>
<td>21</td>
<td>2 CR</td>
<td>KAFO</td>
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<td>96</td>
<td>1.7</td>
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<td>13</td>
<td>23</td>
<td>36</td>
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<td>6</td>
<td>16</td>
<td>22</td>
<td>2 CR</td>
<td>AFO</td>
</tr>
</tbody>
</table>

**NOTE.** Level and completeness of injury based on American Spinal Injury Association (ASIA) neurologic classification.\(^g\)

Abbreviations: AFO, ankle-foot orthosis; CR, crutch; F, female; KAFO, knee-ankle-foot orthosis; LE, lower extremity; M, male; WKR, walker.
that would be most likely to experience the largest weight-bearing loads during ambulation.

**Motion analysis.** For the purpose of determining shoulder joint reaction forces, we measured the motion of the UE, trunk, and assistive device using a Vicon motion analysis system. Six video-based infrared cameras tracked 26 lightweight, reflective markers (diameter, 17mm) taped to the assistive devices and over specific bony landmarks of the bilateral upper extremities and trunk.

To identify motion of the trunk, reflective markers were taped over the manubrium, xiphoid process, and the spinous processes of the T3 and T10 vertebrae. Markers also were taped over the following UE landmarks, bilaterally: the greater tuberosity of the humerus, mid humerus, medial and lateral epicondyle of the humerus, radial and ulnar styloid processes, and the third and fifth metacarpal heads. Assistive device motion was tracked by placing 3 markers on each crutch, and 6 markers on the walker. Crutch markers were placed at the tip of the crutch handle, at the proximal shaft, approximately 7.6cm (3in) below the handle, and on the distal shaft. Walker markers were placed bilaterally, just distal to both the anterior and posterior portion of the handles, as well as on the distal portion of the anterior shaft. Prior to beginning data collection, the Vicon motion acquisition system was calibrated over a 4×1.2×1.8m volume within the center of the walkway. Motion data were sampled at a rate of 50Hz and stored on a separate PDP-11/83 computer which was time synchronized with the force and stride data acquisition computer.

**Data Management**

Footswitch and force data were processed on a MicroVAX 3200 computer and were averaged over .01-second intervals. The foot-floor contact pattern of the weaker leg was used to determine gait cycle (GC) timing, where initial contact to the next initial contact defined 1 stride (0%–100% of GC). To facilitate comparison of forces among subjects and between assistive device conditions, stance phase duration was normalized to 62% of GC. Walking velocity, cadence, and stride length were averaged for each subject over 2 ambulation trials with each assistive device. Stride characteristics were ex-

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pressed relative to a laboratory normative database (% of normal) to account for age and gender variations.\textsuperscript{17}

The 3D coordinate data and marker trajectories were identified and processed using Adtech Motion Analysis Software.\textsuperscript{6} Trajectory data were smoothed using a 4-Hz low-pass second-order recursive Butterworth digital filter with forward and backward passes to eliminate phase shift. Shoulder joint forces were subsequently calculated with an inverse dynamics algorithm using the individual subject’s anthropometric parameters, motion data, and 3D assistive device reaction forces. The individual components of the 3D net GHJ force were calculated: superior/inferior (+Fz/Fz), anterior/posterior (+Fx/Fx), and medial/lateral (+Fy/Fy).

An ensemble average of force data was calculated for each 1% of GC from approximately 5 strides for each assistive device condition. The peak force, rate of loading, and force-time integral (FTI) were calculated for each component of the shoulder force during the period of greatest weight-bearing support (loading response and single limb support, from 1% to 50% of GC). Peak force was determined as the greatest mean force (in newtons) occurring over a 5% of GC interval. Rate of loading (in newtons per second) was determined from the slope of the force over a 10% of GC interval in order to minimize the effect of brief transitions in force direction. FTI (newton seconds) was determined by calculating the area of the force from 1% to 50% of GC.

Statistical Analysis

Data were analyzed utilizing SPSS, version 11.0, software.\textsuperscript{b} The Shapiro-Wilk statistic determined that walking velocity, cadence, and stride length data for the 2 assistive devices were normally distributed, allowing for comparison of group means using paired t-tests. Repeatability of intercycle force profiles (Fx, Fz) was assessed using the coefficient of multiple correlation (CMC) statistic.\textsuperscript{18} Peak force, rate of loading, and FTI data were not normally distributed, therefore, force data medians were determined and compared using the nonparametric Wilcoxon signed-rank test. A significance level of .05 was selected. Inter-cycle force patterns were very consistent for both crutch and walker conditions (CMC: Fz crutch = .90; Fz walker = .81; Fx crutch = .92; Fx walker = .83).

RESULTS

Mean self-selected walking velocity did not differ significantly during ambulation with crutches or the walker (t = 1.4, 38\% ± 17\% of normal vs 35\% ± 17\% of normal, respectively). Stride length, however, was significantly greater during crutch walking than during ambulation with the walker (t = 3.5, 62\% ± 11\% of normal vs 58\% ± 12\% of normal). Cadence did not differ during ambulation with the crutches and the walker (t = 0.2, 58\% ± 20\% of normal vs 59\% ± 21\% of normal).

Peak Shoulder Joint Force

For both the crutches and walker, the dominant weight-bearing component of the GHJ reaction force was the superi-orly (+Fz) directed force (fig 2). A significantly greater peak force was recorded during crutch ambulation for most components of the GHJ reaction force (table 2). The median peak superior force was statistically greater during crutch ambulation (49N) (see fig 2A) compared with that recorded during ambulation with the walker (45N) (see fig 2B). Ten of 14 subjects had higher superior forces during ambulation with crutches. The median inferior distraction force (−Fz) also was significantly greater during the crutch condition (43N vs 24N). The next largest weight-bearing force was posteriorly (−Fx) directed, however, no significant difference between assistive device conditions (crutches = 21N vs walker = 25N) was observed. The median peak anterior force (+Fx) was low for both test conditions. Finally, the medially (+Fy) directed peak force was significantly greater during crutch walking (15N) than walker ambulation (7N).

Rate of Loading

Similar to the peak force data, a significantly higher rate of loading occurred for the superior shoulder joint force during crutch ambulation (312N/s) compared with that recorded during the walker condition (200N/s) (table 3). The rate of inferior shoulder force unloading (negative slope) also was significantly greater during crutch walking (304N/s vs 178N/s during walker ambulation). The rate of posterior joint loading was greater for the crutches than the walker (106N/s vs 85N/s, respectively), which approached statistical significance (P = .06). The rate of loading for the anterior force was similar for the 2 test conditions. Finally, both medial and lateral force rate of loading was significantly greater during crutch walking (see table 3).

Force-Time Integral

The FTI of the superior shoulder joint forces during the primary weight-bearing period of ambulation was 47\% greater during crutch ambulation (1442N·s) than in the walker condition (984N·s, P = .12), although not statistically significant (table 4). The next largest weight-bearing FTI occurred for the posterior force, which was statistically similar between assistive device conditions (crutches = 442N·s vs walker = 598N·s). Finally, the total medial shoulder joint FTI was significantly greater during ambulation with crutches (396N·s vs 146N·s).

DISCUSSION

The current study provides the first comparison of the loads experienced by the shoulder joint during ambulation with both crutches and a walker in a patient population. Weight-bearing shoulder joint forces were predominantly in the superior and posterior directions, with the superior force being the greatest. The peak superior forces recorded during assisted ambulation in this investigation were considerably larger than those experienced during manual wheelchair propulsion and were more comparable with forces reported while propelling a wheelchair up an 8% grade.\textsuperscript{3} This large superiority directed weight-bearing force may potentially threaten GHJ integrity as translation of the humeral head and subsequent impingement of subacromial structures may occur if forces are not matched by an appropriate response of the rotator cuff and thoracohumeral depressor musculature.\textsuperscript{19,20} Additionally, the repetitive nature of these large net joint forces may contribute to shoulder joint degeneration in persons with incomplete SCI.\textsuperscript{21}

A comprehensive description of the shoulder joint demands of ambulation with an assistive device was provided by analysis of 3 different quantitative measures of force. Quantification of peak force levels demonstrates the magnitude of demand that the shoulder structure must counteract in order to preserve joint integrity and avoid injury. Rate of loading is essential in the consideration of shoulder joint demands because it describes how quickly the muscles must respond to a particular magnitude of force. Quantification of FTI reflects the total force experienced over a distinct time period (from 1% to 50% of GC) during which the shoulder musculature must respond to counteract the external force impacting the GHJ. This may provide information regarding the potential for over-
Fig 2. Shoulder joint forces throughout the gait cycle during ambulation with crutches and walker (N=14). Thick lines represent mean profile and thin lines represent + and – 1 standard deviation. Dashed vertical lines represent division between stance and swing phase of gait (62% of GC). NOTE: Mean force profiles represent a group mean over the entire gait cycle and do not correspond exactly with median peak force data presented in table 2, owing to intersubject variability of the timing of peak forces.
use or repetitive strain injuries when muscles are required to counteract the magnitudes of the applied forces for an extended period of time. Together, these 3 quantitative measures of the shoulder joint forces experienced during each stride represent the total joint demand and the potential for injury.

Consistent with our hypothesis, 2 of the 3 measures of the superior shoulder joint force, peak and rate of loading, were significantly greater during ambulation with crutches than with a walker. The third measure, FTI, was 47% higher during crutch walking; however, it was not statistically significantly different from that recorded during the walker condition. While we hypothesized that subjects would walk faster with crutches, walking velocity was similar between the 2 assistive device conditions. The finding that walking with crutches resulted in significantly higher superior shoulder joint loading, without improving walking velocity, is critical in the selection of ambulatory aids for individuals with shoulder weakness or pain.

Higher shoulder forces associated with crutch use was not consistent with the vertical assistive device loads described by Melis and Waters and colleagues, who reported similar loads for both crutches and walker use. Disparity between the superior shoulder joint force and previous reports of vertical assistive device loads may be explained by 2 factors. First, the current study compared each subject walking with both crutches and a walker. Previous investigations, by contrast, tested subjects only with their customary ambulatory aid, which was likely selected based on walking ability. Thus, those with weaker lower extremities likely ambulated with a walker to improve stability, resulting in the need to exert a greater percentage of body weight on the assistive device. Second, the instrumentation employed in the current study allowed separate measurement of loads exerted through each UE; whereas previous investigators reported the sum of the loads imposed by both UEs when using a walker and crutches. As limb strength is often disparate following SCI, the ability to measure each extremity separately is essential for documenting demands. Bachschmidt et al described a similar method for documenting internal UE joint moments associated with walker ambulation. Their study, however, omitted calculation of 3D shoulder joint forces, in addition to testing nondisabled individuals.

The posteriorly directed shoulder joint force has not been described previously in a patient population. Although the magnitude of the posterior force was as much as 50% less than the superior force, the resultant force in the superior-posterior direction may lead to compromise of the posterior capsule, as well as the supraspinatus, infraspinatus, and teres minor tendons. The magnitude of the posterior joint forces did not differ significantly between the 2 walking aids. Finally, each of the 3 measures of the medial shoulder joint force (peak, rate of loading, FTI) was significantly greater during crutch walking, although the magnitude was 3 to 6 times less than the corresponding measure of the superior force.

Of the non-weight-bearing forces influencing the shoulder joint during assistive device ambulation, the inferior force was the most substantial. The inferior force represents the effect of gravity and inertia on the arm and the assistive device. At the shoulder joint, the weight of the arm creates an inferior distraction force of approximately 35N. Crutch use exerted an additional distraction force on the UE (total, 43N) whereas pushing the front-wheeled walker over a tile floor provided support for the weight of the arms (total, 24N). Most people with incomplete SCI will have adequate UE muscle strength to accommodate an increased inferior distraction force during crutch walking. For those with C5 or C6 incomplete tetraplegia, however, muscle groups critical for shoulder joint protection may be weakened. For example, in the presence of deltoid weakness, compensation with shoulder girdle elevation and trunk rotation can be used to lift and advance the crutch, yet no compensatory strategy is available to substitute for rotator cuff weakness.

Despite the clinical assumption that people ambulate faster with crutches than with a walker, velocity did not differ significantly between the 2 assistive devices. Even though stride length demonstrated a statistically significant difference between the crutches and the walker, the difference was relatively small (4% of normal). The decreased stride length recorded during ambulation with a walker may be attributable to differences in structural design, with the walker providing an anterior barrier to limit forward progression of the lower extremities. While some subjects may have adjusted for a decreased stride length by increasing cadence, on average, there was no significant difference in cadence during crutch and walker assisted ambulation.

The limitations of this study relate to variability in both the subject population and individual performance. First, the clinical presentation of subjects with incomplete injury to the spinal cord varies greatly, depending on the location and severity of insult. A more homogeneous subject pool may demonstrate differences in GHJ reaction forces during crutch and walker ambulation associated with tetraplegia or paraplegia. Next, the walking ability of the individuals tested was severely limited, as demonstrated by the average self-selected walking velocity of 35% to 37% of normal. Thus, the period of adaptation to the novel assistive device, as well as number of walking trials, was limited by subject fatigue. These factors likely contributed to some variability in the force data within subjects. While more subjects (11/14) customarily walked with crutches, shoulder joint forces were higher during crutch walking despite the fact that walking velocity was similar during the crutch and walker conditions. This finding suggests that device novelty did not have an effect on velocity or shoulder joint forces. Finally, van Hedel et al recently compared walking velocity during a 10-m and 6-minute walk test (6MWT) in 75

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**Table 2: Median Peak Shoulder Joint Force (25th–75th Quartile) During Ambulation With Crutches and Walker**

<table>
<thead>
<tr>
<th>Force Direction</th>
<th>Crutches (N)</th>
<th>Walker (N)</th>
<th>z Statistic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior</td>
<td>311.6* (184.8–741.5)</td>
<td>199.8 (119.1–291.3)</td>
<td>3.11</td>
</tr>
<tr>
<td>Inferior</td>
<td>101.0 (65.4–163.8)</td>
<td>100.0 (55.4–124.6)</td>
<td>1.22</td>
</tr>
<tr>
<td>Anterior</td>
<td>106.2 (79.7–196.2)</td>
<td>84.5 (52.0–128.2)</td>
<td>1.85</td>
</tr>
<tr>
<td>Medial</td>
<td>62.7* (54.6–116.6)</td>
<td>33.9 (24.7–70.5)</td>
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</tr>
<tr>
<td>Lateral</td>
<td>64.4* (49.5–94.5)</td>
<td>42.2 (32.1–58.9)</td>
<td>2.20</td>
</tr>
</tbody>
</table>

*Crutch condition significantly greater than walker (P<.05).

**Table 3: Median Rate of Loading for Shoulder Joint Force (25th–75th Quartile) During Ambulation With Crutches and Walker**

<table>
<thead>
<tr>
<th>Force Direction</th>
<th>Crutches (N/s)</th>
<th>Walker (N/s)</th>
<th>z Statistic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior</td>
<td>304.1* (246.9–612.6)</td>
<td>178.2 (115.4–243.9)</td>
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<td>100.0 (55.4–124.6)</td>
<td>100.0 (55.4–124.6)</td>
<td>1.22</td>
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<tr>
<td>Anterior</td>
<td>106.2 (79.7–196.2)</td>
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<td>2.67</td>
</tr>
<tr>
<td>Lateral</td>
<td>64.4* (49.5–94.5)</td>
<td>42.2 (32.1–58.9)</td>
<td>2.20</td>
</tr>
</tbody>
</table>

*Crutch condition significantly greater than walker (P<.05).
participants with incomplete SCI and reported increased velocity during the 6MWT (34m/min vs 20m/min for the 10-m walk). These findings suggest that walking velocity is likely greater in a nonlaboratory, community environment, potentially producing larger GHJ forces than those recorded in the current investigation.

As there was no difference in walking velocity, differences in the shoulder joint forces recorded between the 2 conditions must relate to variations in mechanics of assistive device use. Shared UE weight bearing is likely 1 factor that contributes to the decreased vertical shoulder joint forces recorded with walker use, versus the almost unilateral weight bearing that occurs during crutch walking. Comparison of shoulder kinematics and net joint moments for subjects ambulating with both crutches and a walker, as well as associated muscular activity, is also necessary to delineate differences in shoulder joint demands.

The current investigation demonstrated significantly greater loads impacting the GHJ during ambulation with crutches. Thus, clinicians may want to consider the ability of the shoulder musculature to meet these high demands when prescribing crutches as an ambulatory device, in addition to stability, maneuverability, and aesthetic considerations. For subjects with UE weakness or shoulder pain, the current data provides a rationale for use of a walker for ambulating longer distances in the community. Crutch use could be limited to the house- hold, where greater maneuverability may be required.

**CONCLUSIONS**

The weight-bearing forces incurred at the shoulder joint during crutch and walker ambulation were primarily in the superior and posterior directions. The superior shoulder joint force was significantly higher during ambulation with crutches. The posterior shoulder joint force did not differ significantly between the crutch and walker conditions. Velocity did not differ significantly during reciprocal ambulation with crutches and a walker.

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**References**


Suppliers
a. B&L Engineering, 3002 Dow Ave, Ste 416, Tustin, CA 92780.
b. Bio Sentry Telemetry Inc, 20720 Earl St, #4, Torrance, CA 90503.
c. Digital Equipment Corp, 111 Powdermill Rd, Maynard, MA 01754-1499.
d. Bertec Corp, 819 Loch Lomond Ln, Columbus, OH 43085.
e. MA300; Motion Lab Systems, 15045 Old Hammond Hwy, Baton Rouge, LA 70816.
g. Adtech Motion Analysis Software System, 2002 Ruatan St, Adelphi, MD 20783.
h. SPSS Inc, 233 S Wacker Dr, 11th Fl, Chicago, IL 60606.