Upper extremity dynamics during Lofstrand crutch-assisted gait in children with myelomeningocele

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ABSTRACT

The use of quantitative models for evaluating upper extremity (UE) dynamics in children with myelomeningocele (MM) is limited. A biomechanical model for assessment of UE dynamics during Lofstrand crutch-assisted gait in children with MM is presented. This pediatric model may be a valuable tool for clinicians to characterize crutch-assisted gait, which may advance treatment monitoring, crutch prescription, and rehabilitation planning for children with MM. Nine subjects with L3 or L4 level myelodysplasia (mean ± S.D. age: 11.1 ± 3.8 years) were analyzed during forearm crutch-assisted gait: (1) reciprocal gait and (2) swing-through gait. Three-dimensional (3D) dynamics of the UE were acquired and the Pediatric Outcomes Data Collection Instrument (PODCI) was administered. The goal of this study was to determine if meaningful differences occur between gait patterns in UE kinematics and kinetics, and if correlations exist between dynamics and functional outcomes.

Temporal-distance parameters showed significant differences between reciprocal and swing-through gait in stride length, and stance duration. All joint ranges of motion were greater during swing-through gait. Thorax, elbow and crutch ranges of motion were found to be significantly different between gait patterns. Kinetic results demonstrated significant differences between reciprocal and swing-through gait, bilaterally, at all joints for the force variables of mean superior/inferior force, range of force, and maximum inferior force. Functional outcomes were strongly correlated with joint dynamics. Accurate quantitative assessment is essential for preventing injury in long-term crutch users. This study has potential for improving clinical intervention strategies and therapeutic planning of ambulation for children with MM.

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1. Introduction

Myelomeningocele (MM) is the most common central nervous system birth defect in the United States [1]. It occurs when the neural tube fails to close, which results in a cystic dilatation of meninges and protuberance of the spinal cord through the vertebral defect [2]. Approximately 1340 infants are born with MM each year in the United States [3]. From 1999 to 2001, birth incidence of the disease was reported to be 3.7 cases per 10,000 live births [4]. Patients with MM have functional deficits, including lower limb paralysis and sensory loss [5].

Paraplegia from the myelodysplasia typically causes impairment of mobility, thus leading patients to depend on assistive devices, such as Lofstrand (forearm) crutches, for ambulation. Studies have shown that approximately 50–60% of young adult patients with MM are ambulatory, with around 23% of these patients using an assistive device [5,6]. These devices often require significant upper body strength. During crutch walking, peak axial loads are reported to be up to 35% of body weight [7].

Literature has shown that long-term crutch usage may result in upper limb pathologies, such as destructive shoulder arthropathy, degenerative arthritis of the shoulder and wrist, or carpal tunnel syndrome (CTS) [8,9]. Repetitive impulse loading combined with prolonged wrist extension and radial deviation are proposed risk factors associated with crutch use [10,11]. Klimaitis et al. found that bearing weight through the upper limbs may hasten the development of degenerative arthritis [12]. It was found that large, superiorly directed, weight-bearing forces may potentially threaten glenohumeral joint integrity [13]. Patients using forearm crutches have also reported symptoms associated with CTS [11].

Kinematics and kinetics of the lower extremity have been studied extensively in children with MM using three-dimensional motion analysis, but upper extremity evaluation is limited. Upper extremity motion, statics and kinetics have been studied in adults with MM, but the potential for improving clinical intervention strategies and therapeutic planning of ambulation for children with MM is limited. A biomechanical model for assessment of UE dynamics during Lofstrand crutch-assisted gait in children with MM is presented. This pediatric model may be a valuable tool for clinicians to characterize crutch-assisted gait, which may advance treatment monitoring, crutch prescription, and rehabilitation planning for children with MM.
2. Methods

2.1. Kinematic model

The model consists of seven body segments (i.e., thorax, upper arms, forearms, and hands) and two crutch segments (i.e., right and left crutches) defined using 26 passive markers [Fig. 1 and Table 1]. The trunk and upper arm segment are connected by a degree of freedom shoulderglenohumeral joint, which follows the ISB recommendations for segment rotation [23,27]. Two degrees of freedom elbow and wrist joints connect the upper arm, forearm, and hand segments. Elbow motion is expressed as dynamic flexion/extension and pronation/supination to control varus/valgus of the elbow [23,28,29]. The thorax coordinate system is based on the method of Nguyen and Baker [21]. The joint coordinate systems of the upper arm and forearm follow ISB suggested convention [27]. Global wrist motion was determined by modeling the third metacarpal of the hand with respect to the forearm. Vicon BodyBuilder V3.6 (Vicon Motion Systems, Ltd., Oxford, England) was used for model development. The kinematic model was previously evaluated for accuracy and precision [23].

Rotations were described using Euler angles (Z–X–Y). The sequence of Euler angles was chosen such that the first two rotations defined the orientation of the longitudinal axis of the bone or trunk, and the third rotation was about this axis (axial rotation). The Euler rotation sequence corresponds to flexion/extension (Z), adduction/abduction (X), and internal/external rotation (Y). This sequence helps to minimize the potential gimbal lock. The thorax and crutch rotations were described with reference to the global coordinate system, while all other rotations were described with respect to the proximal coordinate system.

2.2. Kinetic model

Kinetic equations were formulated according to the inverse dynamics Newton–Euler approach and programmed using BodyBuilder. The model calculates 3D joint forces and moments for the crutch (crutch/hand interface), wrist, elbow, and shoulder. The joint forces and moments were expressed in the proximal segmental coordinate frame.

The accuracy of the force transducers was examined prior to subject testing. Force data was simultaneously collected from a calibrated force plate (AMTI; Watertown, MA) and crutch transducers during reciprocal and swing-through gait by having the crutch tip contact the force plate for each walking trial completed by one subject. Superior/inferior force data from 10 trials each of reciprocal and swing-through gait were analyzed for the right and left crutch transducers. The mean error of the difference between the force plate and transducer was calculated to determine accuracy.

2.2.1. Crutch hardware

Lofstrand crutches (Walk Easy, Inc.; Delray Beach, FL) were instrumented with MCW-6-500 walker sensors (AMTI; Watertown, MA) to measure applied reaction forces and moments along the X-, Y-, and Z-axes. AMTI MSA-6 high gain amplifiers provided excitation and amplification of the transducers. The sampling frequency of the force transducers was 1800 Hz.

2.3. Patient population

Nine subjects, aged 11.1 ± 3.8 years, participated in the research study. Written parental consent and subject assent was obtained in compliance with the Institutional Review Board.

Table 1

<table>
<thead>
<tr>
<th>Marker</th>
<th>Location</th>
<th>Segment</th>
</tr>
</thead>
<tbody>
<tr>
<td>mp/c7</td>
<td>Spinous process of C7 vertebra</td>
<td>Thorax</td>
</tr>
<tr>
<td>mp/ask</td>
<td>Xiphoid process</td>
<td>Thorax</td>
</tr>
<tr>
<td>m/lclav</td>
<td>R/L clavicle</td>
<td>Thorax</td>
</tr>
<tr>
<td>m/latcr</td>
<td>R/L acromion</td>
<td>Upper arm</td>
</tr>
<tr>
<td>m/lmed</td>
<td>R/L medial epicondy</td>
<td>Upper arm and forearm</td>
</tr>
<tr>
<td>m/lile</td>
<td>R/L lateral epicondy</td>
<td>Upper arm and forearm</td>
</tr>
<tr>
<td>m/radl</td>
<td>R/L radial styloid</td>
<td>Forearm and hand</td>
</tr>
<tr>
<td>m/unl</td>
<td>R/L ulnar styloid</td>
<td>Forearm and hand</td>
</tr>
<tr>
<td>m/lm3</td>
<td>R/L 3rd metacarpal</td>
<td>Hand</td>
</tr>
<tr>
<td>m/lm5</td>
<td>R/L 5th metacarpal</td>
<td>Hand</td>
</tr>
<tr>
<td>m/acr</td>
<td>R/L anterior crutch</td>
<td>Crutch</td>
</tr>
<tr>
<td>m/post</td>
<td>R/L posterior crutch</td>
<td>Crutch</td>
</tr>
<tr>
<td>m/sup</td>
<td>R/L superior crutch</td>
<td>Crutch</td>
</tr>
<tr>
<td>m/infer</td>
<td>R/L inferior crutch</td>
<td>Crutch</td>
</tr>
</tbody>
</table>

3. Results

3.1. Temporal-distance parameters

Cadence, walking speed, stride length, and stance duration were compared during reciprocal gait (RG) and swing-through gait (STG). Cadence (RG: 70 steps/min, STG: 77 steps/min), walking speed (RG: 0.5 m/s, STG: 0.7 m/s), and stride length (RG: 0.8 m, STG: 1.0 m) were greater during swing-through gait than reciprocal gait, while stance duration decreased (RG: 60%, STG: 50%). Significant differences between reciprocal and swing-through gait were found for stride length ($p = 0.035$) and stance duration ($p = 0.016$).

3.2. Upper extremity kinematics

Joint motion in the sagittal plane during reciprocal and swing-through gait was analyzed (Fig. 2). Right and left sides were found to be similar for all joint ranges of motion.

3.2.1. Crutches

The crutches demonstrated fore and aft tilt throughout the gait cycles. The mean crutch ranges of motion were significantly different between reciprocal (39°) and swing-through (47°) gait on right ($p = 0.016$) and left ($p = 0.008$) sides.

3.2.2. Wrists

The wrists displayed extension throughout the gait cycles. The mean wrist joint ranges of motion were similar (right: 17°, left: 15°) between gait patterns.

3.2.3. Elbows

The elbows remained in flexion during the gait cycles. The mean elbow joint ranges of motion were significantly different between reciprocal (29°) and swing-through (38°) gait on right ($p = 0.008$) and left ($p = 0.039$) sides.

3.2.4. Shoulders

The shoulders moved between flexion and extension during the gait cycles. Mean shoulder joint ranges of motion were not significantly different between gait patterns (RG: 41°, STG: 47°).

3.2.5. Thorax

The thorax remained in flexion throughout the gait cycles. The thorax range of motion (RG: 13°, STG: 21°) was significantly different between gait patterns ($p = 0.023$).

3.3. Upper extremity kinetics

The superior/inferior force of the transducers was compared to the force plate to determine accuracy. The mean error ranged from 2.9 N (RG) to 6.8 N (STG). This represented an average difference of 0.4% BW (RG) and 1% BW (STG).

The mean joint forces (superior/inferior) and moments (flexion/extension) during reciprocal and swing-through gait were analyzed (Figs. 3 and 4). Forces were greater during swing-through gait than reciprocal gait for all joints bilaterally.

3.3.1. Crutches

The mean compressive (inferior) and tensile (superior) peak crutch forces during swing-through gait exceeded those of

### Table 2: Force and moment variable names and descriptions.

<table>
<thead>
<tr>
<th>Variable name</th>
<th>Acronym</th>
<th>Measure (units)</th>
<th>Abbreviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean force</td>
<td>MF</td>
<td>% Body weight (N/N)</td>
<td>% BW</td>
</tr>
<tr>
<td>Maximum force</td>
<td>MAXF</td>
<td>% Body weight (N/N)</td>
<td>% BW</td>
</tr>
<tr>
<td>Percent where maximum force occurred</td>
<td>PMF</td>
<td>% Gait cycle</td>
<td>N/A</td>
</tr>
<tr>
<td>Range of force</td>
<td>RF</td>
<td>% Body weight (N/N)</td>
<td>% BW</td>
</tr>
<tr>
<td>Impulse</td>
<td>I</td>
<td>Newton-second (N/s)</td>
<td>N/s</td>
</tr>
<tr>
<td>Maximum rate of loading</td>
<td>MRL</td>
<td>% Gait cycle</td>
<td>N/A</td>
</tr>
<tr>
<td>Percent where maximum rate of loading occurred</td>
<td>PMRL</td>
<td>% Gait cycle</td>
<td>N/A</td>
</tr>
<tr>
<td>Force threshold index</td>
<td>FITL</td>
<td>% Gait cycle</td>
<td>N/A</td>
</tr>
<tr>
<td>Mean moment</td>
<td>MM</td>
<td>% Body weight × height (N/m/Nm)</td>
<td>% BWH</td>
</tr>
<tr>
<td>Maximum moment</td>
<td>MAXM</td>
<td>% Body weight × height (N/m/Nm)</td>
<td>% BWH</td>
</tr>
<tr>
<td>Range of moment</td>
<td>PMM</td>
<td>% Gait cycle</td>
<td>N/A</td>
</tr>
<tr>
<td>Moment threshold index</td>
<td>MTL</td>
<td>% Gait cycle</td>
<td>N/A</td>
</tr>
</tbody>
</table>

reciprocal gait. Significant differences between gait patterns at the crutch forces included the mean (right: $p = 0.004$; left: $p = 0.004$), range (right: $p = 0.004$; left: $p = 0.008$), impulse (left: $p = 0.039$), maximum inferior force (right: $p = 0.004$; left: $p = 0.008$), percent where maximum inferior (right: $p = 0.004$; left: $p = 0.008$) and superior force occurred (left: $p = 0.039$), and threshold index of inferior force (left: $p = 0.008$).

The mean crutch moment was significantly greater during swing-through gait than reciprocal gait (right: $p = 0.004$; left: $p = 0.008$). In addition, the maximum flexion moment was

Fig. 2. Mean upper extremity kinematics (sagittal plane). Mean joint angles of the thorax, shoulders, elbows, wrists, and crutches during reciprocal gait (solid) and swing-through gait (dashed) are plotted from 0% to 100% of the gait cycles. Right (black); Left (gray).

Fig. 3. Mean joint forces for the right (black) and left (gray) crutches, wrists, elbows, and shoulders. Reciprocal gait (solid); swing-through gait (dashed). Superior force (+); inferior force (−). Joint forces are positive if they align with the Z-axis (lateral right), X-axis (anterior), and Y-axis (superior) in the local segmental system. Positive joint forces along the Y-axis correspond to tension on the joint, whereas negative joint forces correspond to joint compression.
significantly different between reciprocal and swing-through gait (right: \( p = 0.008 \)).

### 3.3.2. Wrists

The mean peak wrist forces during swing-through gait were greater than during reciprocal gait. Significant differences between gait patterns at the wrist forces include the mean force (right: \( p = 0.004 \); left: \( p = 0.008 \)), range of force (right: \( p = 0.008 \); left: \( p = 0.008 \)), and maximum inferior force (right: \( p = 0.008 \); left: \( p = 0.008 \)). Significant differences at the left wrist were the impulse (\( p = 0.008 \)), threshold index of inferior force (\( p = 0.004 \)), and percent of gait cycle where maximum superior force occurred (\( p = 0.020 \)).

The mean wrist moment ranged between 0% BWH and \(-1.5\% BWH\) during the gait patterns. The mean left wrist flexion/extension moment was significantly different between reciprocal and swing-through gait (\( p = 0.004 \)).

### 3.3.3. Elbows

The mean peak elbow forces were greater during swing-through gait than during reciprocal gait. The mean elbow force (right: \( p = 0.004 \); left: \( p = 0.004 \)), range of force (right: \( p = 0.004 \); left: \( p = 0.008 \)), impulse (right: \( p = 0.039 \); left: \( p = 0.004 \)), maximum inferior force (right: \( p = 0.004 \); left: \( p = 0.008 \)) and percent of gait cycle where maximum inferior force occurred (right: \( p = 0.020 \); left: \( p = 0.004 \)) were found to be significantly different between reciprocal and swing-through gait. The mean threshold index of inferior force (left: \( p = 0.008 \)), percent of gait cycle where maximum superior force occurred (left: \( p = 0.008 \)), and threshold index of superior force (right: \( p = 0.020 \)) also demonstrated significance.

The mean elbow moment ranged between 0.5% BWH and \(-0.5\% BWH\) during the gait patterns. The elbow moment did not present significant differences.

### 3.3.4. Shoulders

Mean peak shoulder forces during swing-through gait exceeded those of reciprocal gait. Significant differences included the mean force (right: \( p = 0.004 \); left: \( p = 0.004 \)), range of force (right: \( p = 0.004 \); left: \( p = 0.008 \)), impulse (right: \( p = 0.039 \); left: \( p = 0.004 \)), maximum inferior force (right: \( p = 0.004 \); left: \( p = 0.008 \)), percent of gait cycle where maximum inferior force occurred (right: \( p = 0.012 \); left: \( p = 0.004 \)) and threshold index of inferior force (right: \( p = 0.020 \); left: \( p = 0.012 \)). Other unilateral significant differences were the percent of gait cycle where maximum inferior force rate of loading (left: \( p = 0.008 \)) and maximum superior force (left: \( p = 0.039 \)) occurred, and threshold index of superior force (right: \( p = 0.023 \)).

The mean shoulder moment ranged between 2% BWH and \(-1\% BWH\) during the gait patterns. No significant differences at the shoulder moment were found between gait patterns.

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**Table 3**

Summary of significant correlations between dynamic metrics and PODCI.

<table>
<thead>
<tr>
<th>PODCI</th>
<th>ROM</th>
<th>I</th>
<th>MAXF (inferior)</th>
<th>FTI</th>
<th>MAXM (flexion)</th>
<th>PMM (flexion)</th>
<th>MAXM (extension)</th>
<th>PMM (extension)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Transfers and basic mobility</td>
<td>S</td>
<td>S</td>
<td>W</td>
<td>S</td>
<td>S</td>
<td>E, W, E</td>
<td>E, W</td>
<td>E, W</td>
</tr>
<tr>
<td>Global function and symptoms</td>
<td>S</td>
<td>S</td>
<td>W</td>
<td>S</td>
<td>S</td>
<td>S, E, W</td>
<td>S, E</td>
<td>S, E</td>
</tr>
</tbody>
</table>

Significant correlations (\( R > 0.7 \) and \( p < 0.05 \)): ROM, range of motion; S, shoulder; E, elbow; W, wrist.

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3.4. Pediatric Outcomes Data Collection Instrument (PODCI)

Important clinical findings were revealed from the current study, which may prove useful during continued, longer-term applications. Numerous correlations were found between dynamic parameters and PODCI functional categories (Table 3). Cadence, walking speed, and stride length correlated to global function and symptoms, and transfers and basic mobility during swing-through gait.

4. Discussion

A unique UE dynamic model for assessment of crutch-assisted gait in children with MM is presented. The UE model has a three degree of freedom shoulder (glenohumeral) joint, and two degree of freedom elbow and wrist joints. The current model simplifies methods for joint center determination by using fixed centers of rotation. Regression techniques are another option for determining the centers of rotation [29,35–37]. The model was based on previous work and literature [23,27,32]. Data from this study expresses the relative magnitude of UE kinetics as percent body weight (% BW) and percent body weight times height (% BWH). For comparison purposes to the lower extremity, Noreau et al. investigated lower extremity forces during crutch-assisted gait. The early load may also be established sooner in the reciprocal gait cycle to prevent falling, which may lead to upper limb pathology, such as shoulder arthritis, due to the high compression forces at the joints. Significant differences between reciprocal and swing-through gait range of motion were detected at the thorax, elbows, and crutches. The ranges of motion agree with those reported by Requejo et al., Noreau et al., and Liggins et al. [26,38,41]. Model similarities include marker placement, segments, and the use of anthropometric data with the inverse dynamics method for calculating joint reaction forces and moments.

Large joint demands were placed on the UE during crutch-assisted gait. Long-term usage of the swing-through gait pattern may lead to upper limb pathology, such as shoulder arthritis, due to the high compression forces at the joints. Significant differences between reciprocal and swing-through gait were found at all joints (right and left sides) for the force variables of mean superior/inferior force, range of force, and maximum inferior force. Other force variables that were found to be significantly different at most joints include the impulse, percent of the gait cycle where the maximum inferior force occurred, and threshold index of inferior force. The earlier occurrence of peak forces during reciprocal gait than swing-through gait indicates that stability must be established sooner in the reciprocal gait cycle to prevent falling, especially at slow walking speeds. The early load may also be associated with an attempt to slow the walking speed, since the upper extremities largely control balance. Several significant differences between gait patterns were found when analyzing joint moment variables, including mean flexion/extension moment and maximum flexion moment.

The joint forces and moments agree with those reported by Haubert et al., Requejo et al., Noreau et al., and Liggins et al. [7,26,38,41]. Although Haubert et al. investigated UE dynamics during reciprocal gait in subjects with spinal cord injury, the joint force and moment results were similar [7]. Requejo et al. reported handle forces to be similar to the distal crutch reaction force, while cuff forces ranged from 7 N (0.74% BW) to 18 N (1.9% BW) during reciprocal gait in one adult subject [26]. Due to this report, cuff forces were assumed to be negligible and were not included in this model, while the superior/inferior transducer force was assumed to act in the line of the crutch shaft and third metacarpal of the hand. It is possible that future inclusion of handle and cuff forces may alter the dynamic patterns.

The model was shown to be effective for detecting significant differences between reciprocal and swing-through crutch-assisted gait in children with MM. It will be used for future studies involving further characterization of dynamic gait in children with MM. The information gained in this study may also be useful to develop an improved rehabilitation protocol and to gain a better understanding of UE dynamics during Lofstrand crutch-assisted gait.

Currently, no studies exist which quantify the relationship between clinical measures and UE dynamics during Lofstrand crutch-assisted gait in children with MM. Range of motion was correlated to upper extremity and physical function during reciprocal and swing-through gait at the right and left crutches. This demonstrates that crutch range of motion could be used to predict UE function. Main findings from the correlations studies showed strong relationships between the PODCI outcomes and the kinetic metrics of maximum inferior force, impulse, threshold index of inferior force, and maximum flexion and extension moments. The biomechanical metrics of the crutches, wrists, elbows, and shoulders can be used to predict upper extremity function, transfers and basic mobility, sports and physical function, pain/comfort, and global function and symptoms. These outcomes offer insight about a subject’s activity, participation, and quality of life. These tools also provide clinicians with additional information about a subject, which may be used to impact further treatment and rehabilitation strategies.

Conflict of interest

There are no conflicts of interest.

Acknowledgements

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