Connecting the dots: Musculoskeletal adaptation in cerebral palsy

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Spasticity inflicts substantial torsional adaptations in ulna and radius of patients with cerebral palsy

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Abstract

The objective of this study is to evaluate the influence of longstanding wrist flexion, ulnar deviation, and forearm pronation due to spasticity on the bone geometries of radius and ulna. Furthermore, the hypothetical influence of these deformities on potential maximal moment balance for forearm rotation was modeled. Bone volume, length and geometrical measures were determined in hemiplegic cerebral palsy patients (n=5) and healthy controls (n=5). Bilateral differences between the spastic arm and the unaffected side were compared to bilateral differences between the dominant and non-dominant side in the healthy controls. Patients showed significantly smaller (radius: 41.6%; ulna: 32.9%) and shorter (radius: 9.1%; ulna: 8.4%) forearm bones in the non-dominant arm than in the dominant arm compared to controls (radius: 2.4%; ulna 2.5% and radius: 1.5%; ulna: 1.0% respectively). Furthermore, patients showed a significantly higher torsion angle difference (radius: 24.1°; ulna: 26.2°) in both forearm bones between arms than controls (radius: 2.0°; ulna 1.0°). The decreased and unbalanced loading causes the bones of the spastic forearm to be substantially smaller and to have a torsion that is approximately 25 degrees larger compared to the contralateral unaffected arm. Torsion in the bones of the spastic forearm is likely to influence potential maximal moment balance and thus forearm rotation function. In clinical practice, bone torsion should be considered when evaluating movement limitations in the upper extremity of especially children with longstanding spasticity of the upper extremity. The presence of significant forearm bone torsion might affect planning and evaluation of treatment regimes in these patients.
Introduction

Hemiplegic cerebral palsy of the spastic type (CP) presents with a developmental disorder of movement and posture causing activity limitations that are attributed to non-progressive disturbances that occurred in the developing fetal or infant brain (Mutch et al., 1992; Bax et al., 2005). These patients typically present with awkward movement patterns that highly affect arm-hand function during functional tasks (Donkervoort et al., 2007; Livingston et al., 2011). Although the exact cause of movement limitations in the spastic arm is unknown, adaptations in soft tissue due to constant pathological loads on the muscles seem to play a role (de Bruin et al., 2012).

According to Wolff’s Law, bone adapts to mechanical loading (Daly et al., 2004; Whiteley et al., 2009). Bilateral morphological differences between bones of the dominant and non-preferred arm in tennis players (Bass et al., 2002; Ducher et al., 2006), baseball pitchers (Sabick et al., 2005; Warden et al., 2009), brachial plexus palsy patients (Hoeksma et al., 2003) and CP patients (Demir et al., 2006) confirm this law.

Recently, a new technique has been developed to describe bilateral symmetry of radius and ulna based on 3D imaging with computed tomography (CT) (Vroemen et al., 2012). By adapting this technique, we aimed to study the influence of longstanding wrist flexion, ulnar deviation and forearm pronation due to CP on forearm bone growth and on the development of torsion in the radius and ulna. For this, we compared CT scan-based three-dimensional reconstructions of the spastic forearm to the contralateral, unaffected forearm in patients with CP. It was expected that the radius, but not the ulna, will show a pronated orientation relative to the contralateral forearm and that the affected forearm will have a smaller volume than the contralateral unaffected forearm. These differences were expected to be significantly larger in patients than in healthy controls, for which no, or only minimal, differences are expected. A change in the geometry of the bones will cause shifts of relative muscle attachment sites resulting in changes in muscle moment arms. Consequently, the potential maximal moment (PMM) of several muscles and thus the PMM balance for each joint angle will change (Ettema et al., 1998; Veeger et al., 2004). Using these muscle moment arms and the torsion angle of the forearm bones,
we modeled the PMM for forearm rotation in the pathological situation. This information will help understanding movement disorders in this patient group and potentially improve treatment.

**Materials and methods**

**Subjects**

Five adult patients (mean age 28, range 21 – 35 years) with spastic hemiplegic CP were included in the study. They had either a Zancolli type IIb or III grasp and release pattern, which means that the patients were not able to actively extend the wrist (Zancolli *et al.*, 1987). All patients had an ulnar deviation in the wrist. Two patients had a pronation deformity type 4 and three patients had a pronation deformity type 3 according to Gschwind’s classification, which means that none of the patients was able to actively supinate the forearm (Gschwind & Tonkin, 1992). The radius of one patient had to be excluded from analysis of torsion and bending angles due to a previous fracture. Five adult control subjects (mean age 25, range 23 – 31 years) were included for comparison of bilateral differences between groups. All subjects gave written informed consent before the start of the study, which was approved by the local Medical Ethics Committee. The study adhered to the ethical guidelines of the 1975 Declaration of Helsinki.

**CT scans/Imaging**

The spastic arm will from now on be referred to as the non-dominant arm. Regular-dose, high-resolution CT scans of both forearms were obtained using standardized clinical methods (Philips Brilliance 64 CT scanner, Cleveland, OH; voxel size 0.33 × 0.33 × 0.33 mm, 120 kV, 150 mAs, pitch 0.6). The original voxel sizes were kept unchanged between scans. Both forearms were scanned individually with patients lying in prone position with the forearm in full pronation and extended above the head.

Scans were segmented semi-automatically using in-house developed software that uses a region-growing algorithm to extract the bone surfaces. In each subject, the radii and ulnae were segmented by threshold-connected region growing, followed by a binary closing algorithm for filling residual holes and closing
of the outline (Dobbe et al., 2011). We derived a three-dimensional polygon from the segmented data that served as a virtual three-dimensional model of the bone. Surfaces of radius and ulna were obtained using Marching Cubes (Lorensen & Cline, 1987) as implemented in MATLAB® (The Mathworks, Natick, MA). Volumes were calculated by counting the number of voxels within the bone segmentations, multiplied by the voxel volume.

**Torsion and bending estimates**

Left side bones were mirrored to right side bones. Torsions of the radii and ulnae were determined with respect to the principal axes of the radius and ulna. These axes were estimated using principal component analysis (Webb, 2002) on the points that constitute the triangulated surface as extracted by the Marching Cubes method. Subsequently, the radii and ulnae were aligned by alternatively estimating the most likely point-to-point correspondences between the dominant and non-dominant bone models and rigidly aligning these until convergence while allowing for scaling (EM-ICP method; (Granger & Pennec, 2002)). Subsequently, the proximal 20% and distal 20% of the non-dominant bone models were registered to the contralateral side using the EM-ICP method with scaling. The torsion angle was then determined for the distal end with respect to the proximal end around the principal bone axis of the unaffected bone model (Supplementary Figure 5.1). Positive angles indicate torsion towards pronation and negative angles torsion towards supination. The bending angle was determined as the maximal rotation perpendicular to the principal bone axis. Torsion and bending angles are reported as a shift in angles of non-dominant arm relative to dominant arm.

**Modeling**

The average of bilateral differences in torsion estimates over all patients was used to model changes in potential maximal moment balance at different forearm rotation angles. Changes in moment arm and corresponding maximal moment for each muscle were modeled corresponding to the distribution of the torsion that was found along the length of the radius. Muscles that contribute to forearm rotation mainly attach to the radius (Supplementary Figure 5.2). The model calculates the
PMM as the product of a muscle’s cross sectional area and its moment arm, multiplied by a constant (100 N.cm\(^{-2}\)) and assumes that the torsion is divided linearly along the length of the bone. Furthermore, the model does not consider the possibility of force transmission between the different muscles (Huijing & Baan, 2001).

**Statistical Methods**

To investigate the influence of longstanding positional deformities, differences between forearms on morphological parameters (length and volume of both radius and ulna) were compared between groups with separate ANOVAs for each parameter, including group (patient vs. control) as between factor and forearm (dominant vs. non-dominant) as within factor, respectively. Data are presented as mean ± standard deviation (SD) unless stated otherwise. Bilateral differences in torsion and bending angles of both radius and ulna were compared between groups using Students’ independent t-test.

*Supplementary figure 5.1. Schematic drawing of torsion, where the distal end moves towards pronation and the proximal end moves towards supination.*
Results

Volumes

Radius. The differences in volume and length between the dominant and non-dominant radii varied significantly between patients and controls ((P<0.01); Table 5.1). Patients’ non-dominant radii had a significantly smaller bone volume (mean relative difference 41.6 ± 3.5%; P<0.01) and were significantly shorter (mean relative difference 9.1 ± 0.9 %; P<0.01) than the dominant side, whereas differences between sides in controls were not significant for neither volume (mean relative difference 2.4 ± 3.3 %; P>0.05) nor length (mean relative difference 1.5 ± 1.1 %; P>0.05).

Ulna. Patients’ non-dominant ulnae also had significantly smaller bone volume (mean relative difference 32.9 ± 4.2 %; P<0.01) and were shorter (mean relative difference 8.4 ± 1.6 %; P<0.01) than the dominant side. Non-dominant ulnae were not significantly shorter than the dominant side in controls (mean relative difference SD 1.0 ± 1.0 %; P>0.05), but non-dominant ulnae of controls did have a significantly smaller volume than dominant side (mean relative difference 2.5 ± 0.9 %; P<0.05). However, differences in both volume and length were significantly larger in patients than in controls (P<0.01; Table 5.1).

Table 5.1. Mean volumes (mm$^3$) and lengths (cm) of radius and ulna in controls and patients. Per characteristic, the bilateral differences, Δarm, are calculated as non-dominant relative to dominant arm. P-values indicate difference between groups.

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Arm</th>
<th>Control</th>
<th>Patient</th>
<th>Δ arm (%)</th>
<th>Between groups</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volume (cm$^3$)</td>
<td>Radius</td>
<td>Control</td>
<td>41.0 ± 13.2</td>
<td>39.8 ± 12.0</td>
<td>-2.4 ± 3.3</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Patient</td>
<td>48.7 ± 13.8</td>
<td>28.2 ± 7.6</td>
<td>-41.6 ± 3.5</td>
</tr>
<tr>
<td></td>
<td>Ulna</td>
<td>Control</td>
<td>45.6 ± 13.9</td>
<td>44.5 ± 13.9</td>
<td>-2.5 ± 0.9</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Patient</td>
<td>56.7 ± 13.8</td>
<td>37.7 ± 8.9</td>
<td>-32.9 ± 4.2</td>
</tr>
<tr>
<td>Length (cm)</td>
<td>Radius</td>
<td>Control</td>
<td>24.5 ± 2.7</td>
<td>24.1 ± 2.4</td>
<td>-1.5 ± 1.1</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Patient</td>
<td>25.3 ± 2.6</td>
<td>23.0 ± 2.5</td>
<td>-9.1 ± 0.9</td>
</tr>
<tr>
<td></td>
<td>Ulna</td>
<td>Control</td>
<td>26.3 ± 2.6</td>
<td>26.0 ± 2.6</td>
<td>-1.0 ± 1.0</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Patient</td>
<td>27.1 ± 2.6</td>
<td>24.9 ± 2.7</td>
<td>-8.4 ± 1.6</td>
</tr>
</tbody>
</table>

Values are mean ± standard deviation.

Torsion and bending

Radius. The distal end of the non-dominant radii in patients were pronated relative to the proximal end in comparison to the dominant radii (mean difference -24.1 ±
14.4°), whereas in the control group this mean difference was 2.0° (± 3.2°). The
torsion angle was significantly more pronounced in patients than in controls (*P<0.01;*
Table 5.2; Figure 5.1A). The gradual shift in color, shown in Figure 5.2A, suggests that
torsion is inflicted both distally and proximally in our patients. However, towards the
distal end there seems to be a more rapid accumulation of torsion (Figure 5.2B).
Bending difference (Figure 5.1B) between the dominant and non-dominant radii of
the CP patients (mean 3.7 ± 3.0°) was not significantly different from the bending
difference in the controls (mean 1.3 ± 0.8°).

**Ulna.** The distal end of non-dominant ulnae in patients were also more pronated
relative to the proximal end in comparison to the dominant ulnae (mean -26.4 ±
13.9°; Figure 5.1C), whereas in the control group this mean difference was -0.8°
(±6.4°). Again, the torsion angle was significantly more pronounced in patients than
in controls (*P<0.01, Table 5.2). Bending difference (Figure 5.1D) between the
dominant and non-dominant ulnae of the CP patients (mean 4.7 ± 3.1°) was not
significantly different from the bending difference in controls (mean 1.8 ± 0.9°).

| Table 5.2. Differences in torsion and non-directional bending of the distal end relative to the proximal end of both radius and ulna between the dominant and non-dominant forearm. *P*-values indicate difference between groups. |
|-------------------------------|------------------|------------------|
| **Δ Torsion (°)** | **Δ Bending (°)** | **Between groups** |
| **Radius** | **Control** | **Patient** | **Control** | **Patient** | **Control** | **Patient** | **Between groups** |
| **Δ arm** | -2.0 ± 3.2 | -24.1 ± 14.4 | -0.8 ± 6.4 | -26.2 ± 13.9 | 1.3 ± 0.8 | 3.7 ± 3.0 | **P<0.01** | **P>0.05** |
| **Ulna** | 1.8 ± 0.9 | 4.7 ± 3.1 | **P>0.05** | **Values are mean ± standard deviation.** |
Figure 5.1. A, distal view of a pathological radius (dark-gray) matched to the contralateral unaffected radius (light-gray). B, lateral view of a pathological radius (dark-gray) distally matched to the contralateral unaffected radius. C, distal view of a pathological ulna (dark-gray) matched to the contralateral unaffected ulna (light-gray). D, lateral view of a pathological ulna (dark-gray) distally matched to the contralateral unaffected ulna (light-gray).
Figure 5.2. A, gradual cumulative torsion of the distal radius relative to the proximal radius in the non-dominant arm compared to the dominant arm in a patient. B, mean cumulative torsion (SD) in the distal radius relative to the proximal radius plotted as a function of bone length fraction, in which 0 is completely proximal and 1 is completely distal.
Potential maximal moment

Because the torsion seems to be inflicted both distally and proximally (Figure 5.2A) with a somewhat more rapid accumulation of torsion at the distal end (Figure 5.2B), we modeled the pathological situation as if the total torsion of 25° is inflicted partially from proximal loading towards supination and partially from distal loading towards pronation (Figure 5.3). Negative moments are supination moments. In the pathological situation, the potential maximal moment balance for forearm rotation becomes less negative. This means that there is a smaller moment balance of all relevant forearm muscles towards supination, which would result in approximately a 30% decrease of maximal supination moment. The shaded part of Figure 5.3 represents the maximal forearm rotation range of motion (ROM) of CP patients. This part visualizes that the movement range in which the difference in supination moment between the normal and pathological situation becomes smaller, is outside of the maximal ROM of these patients.

Discussion

In this study we have shown that the radius and ulna in the spastic forearm in our group of CP patients are deformed. The spastic forearm bones were not only significantly decreased in length and volume compared to the dominant arm, spastic forearm bones also showed substantial torsion between their proximal and distal ends. These differences were significantly larger than the bilateral differences between dominant and non-dominant forearm bones in healthy controls.

According to Wolff’s law, bone mass and geometry are mainly regulated by mechanical loading (Bergmann et al., 2011). Bone mass can increase with exercise (Haapasalo et al., 2000) and decrease with disuse as in for instance brachial plexus palsy (Ibrahim et al., 2011). Moreover, the effect of exercise on bone growth has been shown to be greater if exercise has started before puberty (Kannus et al., 1995). Furthermore, arm length discrepancy in CP has been shown to increase with age and to be related to hand function, but not to spasticity (Demir et al., 2006). Although the influence of systemic factors on growth in CP is not ruled out with current observations, the difference in loading between both arms is a plausible explanation for the bilateral volume asymmetry within the patient group.
Figure 5.3. Potential maximal moment balance at different forearm rotation angles. The solid line represents PMM expected in a normal situation and the dashed line represents PMM when the forearm bones are deformed both distally and proximally. The shaded part of the graph represents the maximal forearm rotation ROM for CP patients. The border of the maximal pronation angle is faded because CP patients belong to a heterogenic population with several levels of disability. It is common within this population that patients cannot achieve forearm rotation towards supination beyond neutral (0°).

Whereas decreased loading of the bones is likely to have caused decreased growth of the radius and ulna in the spastic arm, unbalanced loading could have caused the bone to grow with increased torsion (Figure 5.1; Supplementary figure 5.1). This has previously been described in baseball pitchers (Sabick et al., 2005) and handball players (Pieper, 1998) as well as in the humerus of brachial plexus palsy patients (Hoeksma et al., 2003) and CP patients (Katthagen et al., 2009). Several muscles of the arm are thought to be spastic and thus more active than the other muscles of the affected arm in CP patients, i.e. the mm. biceps brachii, flexor carpi ulnaris, pronator teres and pronator quadratus. Together with elbow flexion, the biceps brachii imposes a supination moment on the proximal end of the radius. Pronator quadratus and pronator teres on the other hand, have a primary pronation function and pull the distal part of the radius towards pronation (Supplementary Figure 5.2).
These increased loads together with the decreased antagonistic loading of supinator muscle could result in torsion as illustrated in Supplementary Figure 5.1. As the radius is rotating around the ulna during forearm rotation and most muscles that rotate the forearm are mainly attached to radius, we did not expect the ulna to show the same amount of torsion as the radius. Our findings might, at least partially, be explained by the loading of muscles that have a pronation moment on the ulna (Supplementary Figure 2). For instance the supinator muscle with its proximal attachments to the ulna or the extensor carpi ulnaris muscle with its distal connections to the ulna could inflict such a pronation moment.

A change in the geometry of the bones will cause shifts of relative muscle attachment sites resulting in changes in muscle moment arms. Consequently, the PMM of several muscles and thus the PMM balance for each joint angle will change (Ettema et al., 1998; Veeger et al., 2004). As illustrated in Figure 5.3, PMM theoretically changes considerably compared to the normal situation when applying torsion to the model. The border of the maximal pronation angle is faded because CP patients belong to a heterogenic population with several levels of disability. Still, it is common within this population that patients cannot achieve forearm rotation towards supination beyond neutral (0°). Moreover, within the range of 0 to 60 degrees of pronation, the effect of torsion on PMM balance is most substantial.

As opposed to treatment of movement limitations in brachial plexus palsy (Abzug et al., 2010), studies on the etiology and treatment of movement limitations in the spastic forearm mainly focus on the soft tissues (Rameckers et al., 2009; Koman et al., 2010; de Bruin et al., 2012). However, the assumption that movement limitation is partly caused by forearm bone deformities implies that in some cases soft tissue procedures could be insufficient to correct these deformities.

**Conclusion**

The decreased and unbalanced loading of the bones in the spastic forearms in our group of patients with hemiplegic CP have caused these bones to be smaller and to have a torsion that is approximately 25 degrees larger compared to the contralateral unaffected arm. When compared to healthy controls, these differences between dominant and non-dominant arms are substantial. The torsion in the spastic forearm
is likely to influence potential maximal moment balance and thus forearm rotation function. This torsion should be considered when planning treatment and evaluating outcome of treatment regimes for the upper extremity in CP patients.

Supplementary Figure 5.2. Schematic view of ulna and radius, in which the origin and insertion of the muscles acting on the forearm are shown. The solid lines mark the area of the origin and the filled areas mark the area of insertion.