Muscle Activity in the Spinal Cord–Injured During Wheelchair Ambulation

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Key Words: electromyography • motor activity • movement patterns • shoulder

Right shoulder complex muscles of nondisabled, paraplegic, and quadriplegic subjects were monitored with electromyography (EMG) during standardized wheelchair ambulation. It was shown that wheelchair ambulation required the recruitment of large amounts of available motor units in spinal cord–injured persons. Motor unit recruitments differed for the groups: recruitment was minimal for the nondisabled subjects, moderate for paraplegics, and often maximal for quadriplegics. In addition, large intra- and intergroup variabilities were found in the pattern of muscle recruitment during the standardized wheelchair ambulation movement.

The high variability shown in the muscle recruitment patterns of the normal individuals was unexpected, because the ambulation movement had been standardized as much as possible. The technique used to monitor muscle activity in this study reflects an example of how EMG can be employed to analyze activity during a movement. Using this technique one can objectively determine if assumptions about what is occurring in a muscle group during activity are correct.

Electromyography (EMG) is a technique used to record the activity of the electrical potentials of muscles during contraction. When this activity is correlated with limb position during movement tasks, one can determine both the muscle's function within the activity and its level of performance (Trombly, 1983, p. 245). Variability is thought to exist in the EMG of subjects required to do the same task. However, in one study a standardized, simple movement of elbow extension and flexion showed low variability in the level of muscle activity in the pattern of agonist and antagonist recruitment in normal individuals (Gandy, Johnson, Lynn, Reed, & Miller, 1980).

Our study sought to determine whether a repetitive upper extremity task produced similar patterns of muscle recruitment and levels of muscle activity in subjects when the movement was standardized. It was expected that the nondisabled subjects tested would show a stereotypical movement pattern during the standardized ambulation activity. The spinal cord–injured subjects were expected to show idiosyncratic movement patterns related to their residual muscle innervation. The clinical significance of the investigation is the demonstration of the use of EMG to determine muscle activity level and pattern of muscle recruitment within an upper extremity task.

A few studies examined upper extremity function during repetitive activities such as bilateral sanding (Newall, Robinson, & Spaulding, 1981; Spaulding & Robinson, 1984) elbow flexion and extension (Gandy et al., 1980), and stirring movements (Gandy, Johnson, Lynn, Miller, & Reed, 1977). However, literature on the muscle function of the upper extremities during wheelchair locomotion is scarce. This type of locomotion is common to patients seen in rehabilitation occupational therapy units. A study of the functional anatomical requirements of wheelchair ambulation would help occupational therapists in prescribing wheelchairs and in developing the motor patterns and strength for this occupational performance task.

Several studies have investigated wheelchair locomotion. Its ergonomic demands have been analyzed (Andersson, Brattgard, & Severinson, 1978; Andren, Brattgard, & Brax, 1970; Brattgard, Carlsoo, Lidberg, & Severinson, 1975; Brattgard & Dewin, 1969; Dreisinger & Londeree, 1982; Lehmann, Warren, Halar, Stonebridge, & Delateur, 1974; Pratthard, Lindstrom, Severinson, & Wihl, 1983; Sawka, Glaser, Laubach, Al-Samkari, & Suryaprasad, 1981; Sanderson & Sommer, 1985). The importance of the floor surface type has been explored (Cerquiglini, Figura, Marchetti, & Ricci, 1979; Grimby, 1983; Wolfe, 1978), and energy costs for wheelchair ambulation have been calculated (Cerny, Waters, Hislop, & Perry, 1980; Grimby, 1983; Glaser, Foley, Laubach, Sawka, & Sur-
Method

Subjects

Nine subjects (three nondisabled, three paraplegic, and three quadriplegic persons) participated in the study (see Table 1 for subject data), and all subjects were familiar with wheelchair mobility. Each subject received pretest training for the purposes of standardizing the phasic, timing, and movement patterns of the ambulation. All subjects used Everest and Jennings Premier model wheelchairs equipped with 20 cm casters in front, 50 cm wheels in back, inflated rubber tires, and hand rims for propulsion. Detachable desk-length arms, swing-away footrests, and a straight back were also used, and the wheelchair was fitted for each person. Throughout the testing, subjects sat on a cushion uniformly 7 cm thick. A large laboratory room with asphalt tile flooring was used for all testing.

EMG Signals

Seven sets of 16 mm (8 mm contact surface) bipolar, silver/silver chloride surface electrodes (Beckman Instruments Inc., California) were placed in the midline, and along the longitudinal axis, of superficial muscles of the right upper extremity of each subject. Each set of two electrodes was placed edge-to-edge over the muscle bellies of the sternal portion of pectoralis major (PM), the biceps brachii (BB), the anterior deltoid (AD), the middle deltoid (MD), the posterior deltoid (PD), and the medial triceps brachii (MT), and lateral triceps brachii (LT). The ground electrodes were placed on adjacent bony prominences. The muscles were chosen as representative of shoulder function during wheelchair ambulation.

It was recognized that other muscles (e.g., scapular depressors) would probably play a major role in wheelchair ambulation. However, the surface electrodes employed, when placed over the lower fibres of trapezius, tended to move from the friction created by the wheelchair back. Since we wished to use only noninvasive techniques, it was impossible to monitor activity from dorsal muscles. Other muscles, such as the deep humeral rotators and hand and forearm muscles, were also not monitored but would be expected to have important functions within wheelchair mobility.

We determined electrode positions by using manual muscle testing techniques, palpation, and/or knowledge of surface anatomy. We positioned the subject's limbs using standardized manual muscle testing techniques (Colhurst & Falconer, 1973) while observing and palpating the contracting muscle; this allowed the best determination of correct electrode placement over the muscle in question.

Before electrodes were mounted, the skin was rubbed with alcohol and abraded slightly until it turned pink. The same area of the skin was then lightly rubbed with electrode paste. Electrodes were taped to the subject's skin, and leads were fixed to skin or clothing to reduce artifact and to allow freedom of movement. The leads employed were heavy and insulated to minimize movement artifact. As a further measure to reduce noise in the monitoring system, raw EMG signals were initially recorded for each subject. Electrodes were adjusted if problems were discovered (e.g., if adjacent muscles were being monitored inadvertently).

We amplified and recorded signals of EMG activity using an eight-channel amplifier and recording system (Beckman RM Dynograph Recorder; Beckman preamplifier 9852A; Beckman amplifier 482; EMG

Table 1

<table>
<thead>
<tr>
<th>Subject</th>
<th>Sex</th>
<th>Age (years)</th>
<th>Level of Lesion</th>
<th>Months Since Injury</th>
<th>Months of Wheelchair Use</th>
</tr>
</thead>
<tbody>
<tr>
<td>DG</td>
<td>F</td>
<td>21</td>
<td>C6 spared complete</td>
<td>27</td>
<td>25</td>
</tr>
<tr>
<td>BR</td>
<td>M</td>
<td>23</td>
<td>C5 spared incomplete</td>
<td>12</td>
<td>10</td>
</tr>
<tr>
<td>BB</td>
<td>M</td>
<td>31</td>
<td>C5 spared incomplete</td>
<td>18</td>
<td>16</td>
</tr>
<tr>
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<td>F</td>
<td>19</td>
<td>T5 spared complete</td>
<td>12</td>
<td>10</td>
</tr>
<tr>
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<td>M</td>
<td>54</td>
<td>T1 spared complete</td>
<td>11</td>
<td>9</td>
</tr>
<tr>
<td>JW</td>
<td>M</td>
<td>42</td>
<td>T10 spared complete</td>
<td>9</td>
<td>7</td>
</tr>
<tr>
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<td>F</td>
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<td>—</td>
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<td>M</td>
<td>26</td>
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<tr>
<td>SS</td>
<td>F</td>
<td>27</td>
<td>—</td>
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</table>

Table 1: Subject Data

All nondisabled subjects had the same prior experience in wheelchair ambulation, i.e., 1 hour per day on 3 consecutive days.
electrode and input impedance were compatible) designed to monitor electromyographic data. Input impedance for the entire system was 2 Mohms. Amplifier capability ranged from 10 mV/cm to 50 mV/cm; the frequency range was from 5 to 5,000 Hz. The lower cut-off frequency also helped to reduce cable artifact. The amplifier sensitivity was individualized for each muscle and set at the most sensitive level possible.

**Ergonomic Variables**

Each subject was required to propel his or her wheelchair forward at a comfortable set speed of one cycle/sec or approximately 40 cm/sec, which is within the range of comfortable speeds found previously in the spinal cord-injured population (Lehman et al., 1974; Cerquiglini et al., 1979). These movements were timed by a metronome while shoulder EMG was recorded. A 1-second cycle involved two basic movements or phases. The propulsion phase (down phase) ranged from 0 to 0.5 seconds and involved movements of the hands on the wheels from the 12 o'clock to the 3 o'clock position. From 0.5 to 1 seconds, subjects returned their hands to the 12 o'clock starting position (up phase).

Patient leads 5 m long allowed a movement excursion distance of approximately 8 m. The EMG monitor was located in the center of the room to allow approximately 10 full movement cycles. Twenty trials of 10 cycles were produced by each subject, with a 2-minute rest period occurring between each trial. A trial was considered to be one forward movement excursion of about 8 m.

The equipment setup was an umbilical system and not a telemetered system; therefore, the investigation was constrained by the size of the room and the number of trials possible within the room. However, the testing situation did allow translational movement of the subject's center of gravity under controlled speed. These translational movements represent important variables to control and have been recognized by others who used wheelchair ergometers to study work load and energy expenditure (Glaser, Sawka, et al., 1979; Wicks et al., 1977–1978).

**Videotaping**

Subjects were videotaped while they were performing wheelchair ambulation during the EMG recording. Selection of the EMG cycles for analyses depended on how well subjects were seen to synchronize their movements with the sound of the metronome which, in turn, was synchronized with the timing marker on the EMG recording machine. To eliminate investigator bias, a faculty member who was uninvolved with the project selected the EMG cycles for analyses.

**EMG Analyses**

EMG data were processed as the rectified linear envelope expressed in millivolts of activity and represented on graph paper. It has been shown that EMG amplitude closely represents the relative firing frequency and number of recruited motor units (Bigland & Lippold, 1954; Lippold, 1952).

When the full-wave rectified signal is filtered with a low-pass filter, a linear envelope results. This is not to be confused with a true integrated EMG. The rectified linear envelope can best be described as a moving average following the trend of the EMG, and it closely resembles the muscle tension curve. A true integrated EMG, on the other hand, represents the area under the full-wave rectified signal expressed with respect to time (Winter, 1979).

We digitized three EMG linear envelopes representing three consecutive cycles (measured by 1-second time markers on the recorded EMG data that had been synchronized with the metronome) from one representative trial using a microcomputer with plotting capabilities (Hewlett-Packard 9830A; Hewlett-Packard Digitizer 9864A; Hewlett-Packard Plotter 9862A). The time marker began with the start of the propulsion phase. Three cycles were selected from one of the trials that best represented (as determined by observing the videotaped data) the subject's ability to synchronize his or her propulsion with the timing device.

The digital sampling rate allowed 200 sample points to be measured from each cycle. These samples were then averaged over the three chosen cycles. Standard deviations at each point were expressed as vertical bars, but were not plotted in the final results since the average inrasubject variability was found to be minimal (i.e., less than 10%) for any averaged sample point over all subjects. Results were then plotted. The data were expressed at each point in the trace as a percentage of the MVC (Winter, 1979).

We elicited the MVC isometrically, prior to testing, for each muscle using manual muscle testing techniques and with the muscle close to midlength. This technique allowed comparison of each muscle's activity with its own 100% MVC value and facilitated intra- and intersubject comparisons. It is a reliable method of EMG quantification (Antti, 1977; Ekholm, Arborelius, Hillerød, & Orthqvist, 1978; Jonsson, 1978).

Levels of muscle activity were defined as follows: If, during the activity, the muscle was mainly active between 0% and 10% MVC, this was considered to be minimum activity; a range from 10% to 50% MVC was considered moderate activity; maximal activity ranged from 50% to 100% MVC.

**Results**

Computer-plotted results are shown in Figures 1 to 3. Each muscle curve represents the average of three trials performed consecutively on one person. This section discusses these data with respect to the pattern and level of muscle activity. The level of
muscle activity is represented as a percentage of the MVC, of 100%.

Pattern and Level of Muscle Activity

Pectoralis Major (sternal portion). Muscle activity approached 50% MVC in only one quadriplegic subject (BB). In the other eight subjects, the muscle was either inactive or minimally active during wheelchair mobility.

Biceps Brachii. Muscle activity was minimal in two paraplegic (CW, BN) and two nondisabled subjects (TN, KR). In the remaining five subjects (SS, JW, BB, BR, DG), who demonstrated a moderate activity level, the activity was greatest during the first 0.4 seconds. This corresponded to the hand exerting pressure on the wheel to move the wheelchair forward. This was followed by a period of about 0.6 seconds of no activity in this muscle as the hand moved back up to the 12 o'clock position. A smaller burst was evident at the end of the cycle when the hand resumed its grasp of the wheel. Biceps brachii tended to be more active than pectoralis major, but this activity was noted in only five of the nine subjects tested and was evident for less than half the activity time.

Anterior Deltoid. In five subjects (SS, KR, CW, BN, DG), minimal activity was seen in this muscle, whereas moderate activity was noted in the other four (TN, JW, BB, BR). In one quadriplegic (BR) and one paraplegic (JW), the activity pattern was similar to that of biceps brachii in these persons. There were two bursts of activity, one at the beginning and one at the end of the cycle; the activity level was moderate. A second pattern was noted in three subjects (KR, CW, BB), in whom the activity was noted predominantly in the second half of the cycle (when the hand was returning to the 12 o'clock position). Thus, the anterior deltoid showed pattern variation among subjects with the two notable points being minimal activity seen in five subjects and a moderate activity level in four subjects.

Middle Deltoid. The middle deltoid was active in all subjects throughout the entire cycle. In all three normals and in one quadriplegic (DG), the activity was minimal while in all other subjects, moderate activity was seen. In two paraplegics (JW, CW) and one quadriplegic (BR), who demonstrated moderate activity, the activity showed no consistent pattern, being present in either the down or the up phase. In the remaining paraplegic (BN) and quadriplegic (BB) subjects, the activity was maximal, especially during the up phase. Of all the muscles monitored, the middle deltoid was the most consistently active.

Posterior Deltoid. In eight subjects, the posterior deltoid was moderately active; the remaining subject (KR) had minimal activity. The posterior deltoid appeared to have an important role during wheelchair ambulation for all subjects.

Triceps Brachii Lateral Head. The lateral head was only minimally active in most subjects (SS, TN, KR, CW, BN, BB, BR). It exhibited moderate activity in one paraplegic subject (JW). In one quadriplegic

Figure 1
EMG Data for Nondisabled Subjects SS (a), TN (b), and KR (c)

Note. Muscles represented are as follows: 1-PM; 2-BB; 3-AD; 4-MD; 5-PD; 6-LT; 7-MT. Inactive muscles are represented by blanks beside their respective numbers. Graphed data represent an average of EMG for each muscle over three complete cycles. Data are expressed over time with respect to the percentage of maximal voluntary contraction (MVC). Zero seconds marks the beginning of the cycle; 1.0 seconds marks the end. Zero to 0.5 seconds is the propulsive phase (down); 0.5 to 1.0 seconds, the return to starting position phase (up). MVC, along the Y-axis, is not represented by the same scale since plotting space did not allow this.
subject (DG), the triceps lateralis showed 70% MVC activity, but this was 70% of the MVC represented by trace (grade 1) muscle activity as indicated by prior manual muscle testing.

**Triceps Brachii Medial Head.** In five subjects, two nondisabled subjects (TN, KR), one paraplegic (CW), and two quadriplegics (BR, DG), the triceps medial head was minimally active or totally inactive. The remaining subjects (SS, JW, BN, BB) showed moderate activity with no consistent pattern observable in the data.

**Intergroup Comparisons**

Trends within each group were not observed. This result, especially with respect to the nondisabled subjects' data, was contrary to expectation. However, a few pattern similarities were observed across the groups. Figures 1a, 2a and 3a, which represent graphed data for one nondisabled subject (SS), one paraplegic (JW), and one quadriplegic (BR), respectively, show a distinct pattern in biceps and anterior deltoid. These muscles work together at the beginning of the up phase, and the end of the down phase. In these three subjects, middle and posterior deltoid worked concomitantly at the end of the down phase (working after the biceps and anterior deltoid ceased activity and before activity resumed).

Another group exhibited trends. As Figures 1c, 2c and 3c illustrate, these subjects (Kh, BN, and DG) showed most of their muscle activity half way through the cycle rather than at the beginning or end. The cause for these across-group trends was not obvious from the comparative EMG data. To approach this problem, the video taped data were reviewed and it was revealed that the EMG data trends were not represented by similarities in movement patterns. Indeed, movement patterns, along with muscle recruitment patterns, appeared idiosyncratic. Even though all subjects were instructed to produce the same movement pattern in terms of hand placement during the phases, the strategies used by subjects differed. For example, some subjects chose to perform the movement with more shoulder abduction throughout than others.

To attempt an understanding of individual idiosyncrasies, the authors wondered whether an energy analysis of an individual's wheelchair ambulation would correlate with his or her generalized muscle EMG levels. Cinematography data and a resultant mathematical energy analysis on these digitized data were acquired for the three nondisabled subjects during wheelchair locomotion. Energy analysis can be performed by a frame-by-frame determination of limb mass displacement and the theoretical energy required to move that mass (Winter, 1979). These data were collected and analyzed in the Department of Kinesiology (University of Waterloo, Waterloo, Ontario, Canada). The cinematography data analysis showed that the nondisabled subject KR, who used only four muscles minimally during wheelchair ambulation (see Figure 1c), was the most energy efficient of the three subjects tested. The analysis showed distinct differences in the upper extremity energy values between the most efficient (KR) and the least efficient (SS) of the normal subjects. KR depressed the scapulas in the down phase, then elevated the scapulas to the original position in the up phase and kept the shoulders adducted throughout. SS exhibited...
a fairly constant shoulder position throughout the movement; the arms were abducted to about 70 degrees, and much of the push exerted on the wheel was initiated by wrist movement. Previous energy analyses on wheelchair-ambulating individuals (Grimby, 1983) lend support to our finding that certain movement strategies are less energy consuming than others in producing wheelchair propulsion.

Discussion

The data showed no intergroup trends and indicated wide intersubject variability of muscle recruitment patterns. Variable movement (Sanderson and Sommer, 1985) and muscle recruitment (Newall et al., 1981) patterns have previously been observed in spinal cord-injured persons, and this is probably related to neurological deficit. However, an unexpected finding was the lack of similar muscle recruitment patterns among nondisabled subjects.

The shoulder muscle complex offers a much wider range of movements and, concomitant with that, greater abilities to compensate than the hip and pelvic complex (Carlin, 1963). This may partially explain why normal gait is fairly consistent in muscle recruitment patterns (Winter, 1979) and why upper extremity repetitive movements have been found to be so variable in normal and spinal cord-injured persons (Newall et al., 1981; Spaulding & Robinson, 1984). The present investigation was limited by the small number of subjects tested, which precludes generalizing the results. However, the data suggest that, although intergroup and intersubject variances were high, the low intrasubject variability found from cycle to cycle points to the activity being a learned skill. Kamon and Gormley (1968) support this view and have shown low variability in muscle activity following the learning of a gymnastic exercise, although only peak EMG activity was analyzed.

The analyses of the data were limited in relation to the small number of muscles monitored. The movement of scapular muscles, humeral rotators and adductors, and wrist and hand muscles were not recorded and analyzed, although these would have been expected to play important roles within the activity.

The major ramification of these findings is that there may be a lack of certainty in predicting which muscles are active in a repetitive upper extremity movement such as wheelchair ambulation. By assuming that a certain therapeutic activity is appropriate in taxing the required muscles and/or providing a carryover effect to activities of daily living, we may be doing our patients a disservice. Although the EMG recording techniques used in this study would be impractical for clinical use, other clinically appropriate EMG systems are available. We concur with Trombly (1983) and Trombly and Cole (1979) in suggesting that a simple EMG biofeedback system can be used to monitor one muscle or a muscle group during a therapeutic activity. Indeed, we have been successful in using this technique when treating patients.

An intriguing finding arising from these data is that the quadriplegic subjects employed the highest percentage of MVC for all muscles compared with the paraplegic and nondisabled subjects. The paraplegics were also less efficient, in terms of percentage of

Figure 3
EMG Data for Quadriplegic Subjects BB (a), BR (b), and DG (c)

Note: See comments under Figure 1.
muscle activity, to achieve the same goal as compared with the nondisabled subjects. This fact can be looked at from two perspectives. First, since only seven muscles were monitored during this activity, one can say that the quadriplegics used more available motor units from their muscles overall than the paraplegics, who used more available motor units than the nondisabled subjects in terms of the percentage of MVC. However, another approach is to look at the movement requirements for the task studied. Quadriplegics need to employ more motor units in each of their available muscles because they lack strong elbow extension, precise hand grip abilities, and trunk stability. In their efforts to compensate for the lack of these abilities, they tend to tax remaining muscles to complete the movement. This idea has similar support in the paraplegic group, but the percentage of MVC usage is less, and therefore the remaining muscles are stressed to a lesser extent.

Summary and Conclusions

Several shoulder muscles were monitored electromyographically in three nondisabled, three paraplegic, and three quadriplegic subjects during the activity of wheelchair ambulation. The study sought to answer two questions related to the activity: a) Was there a pattern of muscle activation during wheelchair ambulation? and b) What was the degree of this muscle activity during wheelchair ambulation?

The most active muscles monitored during the activity were the middle deltoid, posterior deltoid, and in some subjects, triceps brachii. Wheelchair ambulation was found to be an activity-producing idiosyncratic muscle recruitment pattern not only in spinal cord-injured subjects, but in nondisabled subjects as well. These idiosyncratic patterns for each person tested showed low variance from movement cycle to movement cycle, implying that we were observing a learned skill (Kamon & Gormley, 1968).

The data showed that quadriplegic subjects employed an extremely high percentage of voluntary muscle recruitment ability during wheelchair ambulation. The paraplegic subject group generally used less recruitment ability than the quadriplegic group, but more than the normal group. These findings are probably related both to the voluntary motor unit recruitment ability of the shoulder muscles available to each group and to the strength and voluntary recruitment availability of stabilizing and synergic muscles.

The data also indicate that the movements of spinal cord-injured persons demand high muscle activity. In decisions regarding wheelchair prescription, this factor could be kept in mind. Since abduction of the shoulder throughout the activity appears to be a factor in the high muscle activity requirement, consideration of the armrest height is important. For example, the higher the armrests, the more humeral abduction would be required to assure that the arms would clear the armrests during the wheel propulsion phase.

While this study employed only a small number of subjects and the generalization of the results should be tentative, we think it is appropriate to provide recommendations for occupational therapy. It is important for therapists to recognize that their assumptions about what may be going on in a task with respect to muscular activity might not necessarily be correct. Simple EMG monitoring systems can be employed in the clinical setting to assist the therapist in choosing an appropriate physical activity for a patient.

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