Twitch and Tetanic Properties of Human Thenar Motor Units Paralyzed by Chronic Spinal Cord Injury

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Häger-Ross, C. K., C. S. Klein, and C. K. Thomas. Twitch and tetanic properties of human thenar motor units paralyzed by chronic spinal cord injury. J Neurophysiol 96: 165–174, 2006. First published April 12, 2006; doi:10.1152/jn.01339.2005. Little is known about how human motor units respond to chronic paralysis. Our aim was to record surface electromyographic (EMG) signals, twitch forces, and tetanic forces from paralyzed motor units in the thenar muscles of individuals (n = 12) with chronic (1.5–19 yr) cervical spinal cord injury (SCI). Each motor unit was activated by intraneural stimulation of its motor axon using single pulses and trains of pulses at frequencies between 5 and 100 Hz. Paralyzed motor units (n = 48) had small EMGs and weak tetanic forces (n = 32 units) but strong twitch forces, resulting in half-maximal force being achieved at a median of only 8 Hz. The distributions for cumulative twitch and tetanic forces also separated less for paralyzed units than for control units, indicating that increases in stimulation frequency made a smaller relative contribution to the total force output in paralyzed muscles. Paralysis also induced slowing of conduction velocities, twitch contraction times and EMG durations. However, the elevated ratios between the twitch and the tetanic forces, but not contractile speed, correlated significantly with the extent to which unit force summed in response to different frequencies of stimulation. Despite changes in the absolute values of many electrical and mechanical properties of paralyzed motor units, most of the distributions shifted uniformly relative to those of thenar units obtained from control subjects. Thus human thenar muscles paralyzed by SCI retain a population of motor units with heterogeneous contractile properties because chronic paralysis influenced all of the motor units similarly.

INTRODUCTION

Human muscles show diverse responses to spinal cord injury (SCI) in that some muscles become weak, whereas others remain as strong as control muscles despite years of paralysis (Peckham et al. 1976; Shields 1995; Stein et al. 1992; Thomas 1997a,b). Twitch contraction times of human muscles can either be shorter, longer, or unchanged by paralysis (Thomas 1997a). The extent to which different paralyzed muscles fatigue also varies (Thomas and Zijdewind 2006). The results suggest that the motor units within a given muscle may be influenced by chronic paralysis in different ways.

Most mammalian muscles contain motor units with a broad range of force capacity, contractile speed, and fatigability, characteristics that reflect the function of the muscle (Bigland-Ritchie et al. 1998). A few months after spinal transection in animals, the properties of the motor units within cat hind limb muscles still differ widely even though the units become atrophied, weak, fast contracting, and fatigable (Cope et al. 1986; Mayer et al. 1984; Munson et al. 1986; Pierotti et al. 1991). Whether motor units in a human muscle retain this diversity when all functional inputs from higher centers to the spinal cord have been disrupted for many years by injury is unknown. Only the electromyography activity (EMG) and twitch force of paralyzed human thenar motor units have been recorded previously using either percutaneous nerve stimulation, intramuscular stimulation or spike triggered averaging (Thomas 1997a; Thomas et al. 2002; Yang et al. 1990). Because twitch forces are easily influenced by muscle mechanics and brief periods of activity (Bawa et al. 1976; Burke 1967), tetanic force data would provide a more valid measure of paralyzed motor-unit strength. While the tetanic forces of human motor units have been recorded from various healthy muscles (Fuglevand et al. 1999; Maciefield et al. 1996; McNulty et al. 2000; Thomas et al. 1990), they have never been evaluated after any neuromuscular disorder.

The aim of the present study was to evaluate the electrical and mechanical properties of human thenar motor units that have been paralyzed chronically (>1 yr) by cervical SCI. Each motor unit was excited by intraneural stimulation of its motor axon (Westling et al. 1990), a technique that allowed assessment of the unit EMG, twitch force, tetanic force and axon conduction velocity. We also documented the potentiation of twitch forces following a series of pulse trains at various frequencies as well as unit fatigue (Klein et al. 2006). The present data reveal that despite years of paralysis, human thenar muscles retain a population of motor units with heterogeneous contractile properties. Even though the distributions of conduction velocities, twitch contraction times, twitch/tetanic force ratios, and the frequency to produce half-maximal force were shifted after paralysis, they remained aligned with the corresponding distributions for control units (Thomas et al. 1990, 1991; Westling et al. 1990). Thus chronic paralysis imposed similar influences on all of the paralyzed motor units.

METHODS

The contractile properties of paralyzed thenar motor units were examined in the right hand of 10 men and 2 women [33 ± 9 (SD) yr] with chronic (>1 yr) cervical SCI (Table 1). Injury level was defined...
TABLE 1. Subject history

<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
<th>Age, yr</th>
<th>Injury Level</th>
<th>Injury Cause</th>
<th>Injury Duration, yr</th>
</tr>
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<tbody>
<tr>
<td>1</td>
<td>M</td>
<td>33</td>
<td>C4</td>
<td>Gunshot</td>
<td>13</td>
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<td>2</td>
<td>M</td>
<td>25</td>
<td>C4</td>
<td>Diving</td>
<td>8</td>
</tr>
<tr>
<td>3</td>
<td>F</td>
<td>47</td>
<td>C4</td>
<td>Horseback riding</td>
<td>7</td>
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<td>4</td>
<td>M</td>
<td>44</td>
<td>C4</td>
<td>Motor vehicle</td>
<td>6</td>
</tr>
<tr>
<td>5</td>
<td>M</td>
<td>37</td>
<td>C5</td>
<td>Diving</td>
<td>9</td>
</tr>
<tr>
<td>6</td>
<td>M</td>
<td>20</td>
<td>C5</td>
<td>Motor vehicle</td>
<td>1.5</td>
</tr>
<tr>
<td>7</td>
<td>M</td>
<td>35</td>
<td>C5</td>
<td>Diving</td>
<td>14</td>
</tr>
<tr>
<td>8</td>
<td>F</td>
<td>39</td>
<td>C6</td>
<td>Motor vehicle</td>
<td>19</td>
</tr>
<tr>
<td>9</td>
<td>M</td>
<td>42</td>
<td>C6</td>
<td>Fall</td>
<td>14</td>
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<td>10</td>
<td>M</td>
<td>33</td>
<td>C6</td>
<td>Motor vehicle</td>
<td>13</td>
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<tr>
<td>11</td>
<td>M</td>
<td>23</td>
<td>C6</td>
<td>Horseback riding</td>
<td>2</td>
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<tr>
<td>12</td>
<td>M</td>
<td>23</td>
<td>C6</td>
<td>Motor vehicle</td>
<td>10</td>
</tr>
</tbody>
</table>

Intraneural stimulation of single thenar motor units

The path of the median nerve above the elbow was mapped by applying single stimuli to the skin while looking for contractions of median-innervated muscles. To locate the median nerve, an uninsulated tungsten electrode (0.2 mm diam, FHC, Bowdoinham, ME) was inserted into the arm ~10 cm proximal to the elbow. Its position was adjusted until weak current pulses evoked thenar muscle contractions. As described by Thomas and Westling (1995), subjects with cervical SCI often have impaired sensation so they are unable to guide the experimenter to the nerve by reporting sensations (e.g., paresthesias from electrical stimulation). In the present study, only two subjects were able to assist with nerve location by reporting sensations in the arm or hand. The position and depth of the uninsulated electrode was used as a guide to insert an insulated tungsten electrode (0.2 mm diam, ±1 MΩ, impedance, FHC) into the median nerve. Afferent activity was monitored while stroking median innervated regions of skin to verify that the electrode was in the nerve. The position of the insulated electrode was adjusted in minute steps within the nerve until weak negative current pulses (200-µs duration) excited only one thenar motor axon, judged using classical all-or-none criteria (Fig. 1). The stimulus current was increased slowly and then decreased while monitoring both the EMG signals and the abduction and flexion forces on oscilloscopes (Fig. 1A). At 11 µA, no force or EMG was evoked because the current was below threshold for excitation of a motor unit (Fig. 1, A and B, left). As the current was raised from 11.1 to 12.8 µA, both abduction and flexion forces as well as proximal and distal EMG of consistent amplitude and shape were evoked, indicating that each pulse activated the same motor unit (B, middle, left). This was also the case for the magnitude and direction of the resultant force (Fig. 1C), which were monitored on-line on another oscilloscope. Beyond 12.8 µA, the forces and EMG waveforms increased as other units were activated (B, middle, right). The force vector also increased. However, when the current was reduced, all the signals returned to those that were characteristic of the motor unit excited first (B, right). This procedure was repeated several times to establish the range of current that would excite only the single unit under study. A current near the middle of this range was used to stimulate the motor axon in all of the subsequent protocols.

Stimulation protocol

The stimulation protocols used to excite paralyzed thenar motor units were identical to those used previously for thenar units of able-bodied control subjects so that the two sets of data could be compared (Thomas et al. 1990, 1991; Westling et al. 1990). After establishing the level of current that would be used to stimulate a single thenar motor axon selectively, each unit was subjected to the following stimuli: 20 single pulses, delivered in relation to the pulse pressure wave, to measure twitches; pulse trains at different frequencies to characterize the frequency-force relation (5, 8, and 10 Hz, each for 2 s, 15, 20, 30, 40, and 50 Hz, each for 1 s; 100 Hz for 0.5 s); and 20 single pulses to provide posttetanic twitch data. The effects of fatigue on the electrical and mechanical properties of paralyzed motor units were also evaluated (Klein et al. 2006).

Data collection and analysis

Proximal and distal surface EMG signals were sampled on-line at 3,000 Hz, abduction and flexion forces at 375 Hz, pulse pressure at 375 Hz, and stimulus current at 94 Hz using a SC/Zoom system (Physiology Section, Umeå University, Umeå, Sweden). All data analyses were performed off-line. The integral of the EMG signals, the resultant force, and the force differential were calculated using Zoom software.

Ten twitches were averaged before and after delivery of trains of stimuli at different frequencies, termed initial and posttetanic.
twitches, respectively. Measurements included EMG latency (time from the stimulus to EMG onset), peak-to-peak amplitude, area and duration of the first two phases of EMG for both the proximal and distal signals; and peak abduction, flexion and resultant force, contraction time, (CT; time from resultant force baseline to peak), half-relaxation time (hRT; time for the resultant force to fall to half-maximum force), peak force differential during contraction and relaxation (each was normalized to peak resultant force to provide maximum contraction and relaxation rate; MCR and MRR, respectively). The same measurements were made for the forces evoked by trains of stimuli between 5 and 100 Hz except for CT and MCR because of their dependence on the duration of the train of pulses and stimulation frequency, respectively.

The direction in which force was generated was computed from the abduction and flexion forces. The twitch to maximal tetanic force ratio was calculated for initial and posttetanic twitches. The absolute forces evoked in response to various stimulation frequencies were expressed relative to maximal tetanic force. Correlations between parameters were analyzed using Spearmann rank correlation. Regression lines in illustrations are only shown when the correlations were statistically significant.

There were no differences between distal and proximal EMG latency, duration, peak-to-peak amplitude, or area for paralyzed units (Table 2). Therefore in the results, only distal EMG data for paralyzed units are compared with the corresponding control data.

RESULTS

All or none EMG and twitch force were recorded from 48 paralyzed motor units, as shown for one unit in Fig. 1. In 32 of these units, twitches were also evoked after the delivery of brief trains of pulses at frequencies between 5 and 100 Hz.

Conduction velocity

Figure 2 illustrates how the EMG latency varied from 8.6 to 14.3 ms in four paralyzed units (A and B) taken from two individuals, resulting in conduction velocities that ranged from...
TABLE 2.  
Twitch and tetanic parameters of paralyzed motor units

<table>
<thead>
<tr>
<th>Distal EMG</th>
<th>Proximal EMG</th>
</tr>
</thead>
<tbody>
<tr>
<td>Initial</td>
<td>Post-tetanic</td>
</tr>
<tr>
<td>(n = 48)</td>
<td>(n = 32)</td>
</tr>
<tr>
<td>Latency, ms</td>
<td>10.3 (8.6–15.3)</td>
</tr>
<tr>
<td>Duration, ms</td>
<td>11.7 ± 3.6</td>
</tr>
<tr>
<td>Amplitude, μV</td>
<td>48 (4–327)</td>
</tr>
<tr>
<td>Area, μVs</td>
<td>0.12 (0.01–1.10)</td>
</tr>
<tr>
<td>Conduction velocity, m/s</td>
<td>46.2 (25.8–65.1)</td>
</tr>
</tbody>
</table>

Twitch Data

| Force, mN | 14.4 (2.0–69.7) | 13.6 (4.4–69.7) | 15.0 (4.2–86.6) | 40.5 (9.3–269.9) |
| Contraction time, ms | 57.3 (33.0–102.0) | 58.0 (36.0–102.0) | 61.8 (44.9–101.8) | 63.8 ± 60.7 |
| Half relaxation time, ms | 59.3 ± 12.0 | 60.2 ± 12.4 | 64.5 ± 12.5 | 8.0 (1.0–18.3) |
| Angle, degrees | 61 (55–143) | 57 (55–143) | 50 (56–140) | 138.2 (89.7–485.7) |

Tetanic Data

| Force, mN | 14.4 (2.0–69.7) | 13.6 (4.4–69.7) | 15.0 (4.2–86.6) | 40.5 (9.3–269.9) |
| Contraction time, ms | 57.3 (33.0–102.0) | 58.0 (36.0–102.0) | 61.8 (44.9–101.8) | 63.8 ± 60.7 |
| Half relaxation time, ms | 59.3 ± 12.0 | 60.2 ± 12.4 | 64.5 ± 12.5 | 8.0 (1.0–18.3) |
| Angle, degrees | 61 (55–143) | 57 (55–143) | 50 (56–140) | 138.2 (89.7–485.7) |

Medians, with the range in parentheses, are shown first for each parameter. Mean ± SD values are given below.

30 to 65 m/s. Thus units with slow and fast conduction velocities were recorded from the same subject. The median EMG latency was longer for paralyzed units (10.3 ms) than control units (9.6 ms, P < 0.01). The distribution of conduction velocities was thus shifted toward lower values for paralyzed units compared with control units (Fig. 2C, P < 0.001, Table 2) as conduction distances were similar for both populations. Similar currents were also used to stimulate paralyzed and control thenar units (Westling et al. 1990).

Twitch responses

A diverse range of EMG and twitch force responses were recorded from paralyzed units (Table 2). Some units had small EMG amplitudes, whereas others were large. Similarly, twitch forces were weak or strong. Contraction times and half relaxation times also varied across units. There was no significant correlation between EMG amplitude and force (n = 48), consistent with results from control units (Thomas et al. 1990).

Figure 3 compares the distributions of the initial EMG and twitch force parameters for paralyzed and control units. The duration of the EMG was longer for paralyzed units (11.0 ms) than control units (9.1 ms, Fig. 3A, P < 0.001), but both the EMG amplitude and area were smaller for paralyzed units compared with control units (Fig. 3B and C, both P < 0.001).

The median twitch force of paralyzed units was stronger (14.4 mN) than that of control units (7.7 mN, Fig. 3D, P < 0.01), but most paralyzed units had longer twitch contraction times (57.3 vs. 48.1 ms, respectively). The contraction time distributions thus differed significantly (Fig. 3E, P < 0.001). However, the distributions of twitch half relaxation times were similar for paralyzed and control units (53.8 vs. 54.4 ms, Fig. 3F, P > 0.5) as were the distributions for maximum contraction rates (23.0 vs. 21.2 mNs, P > 0.2) and maximum relaxation rates (13.5 vs. 12.0 mNs, P > 0.3).

**Tetanic forces**

Maximum tetanic forces were obtained from a total of 32 units in 11 of 12 subjects, usually by stimulating with a train of pulses at 40 Hz (median). Paralyzed units had weaker maximal tetanic forces than control units (n = 30 units, Fig. 4A, P < 0.001), but the median direction of force production was similar for both populations (P = 0.4). Only 7 of the 32 paralyzed units (22%) generated maximal tetanic forces greater than the median force produced by control units (96 mN). The higher twitch forces (both initial and posttetanic) and lower maximal tetanic forces of paralyzed units resulted in significantly higher twitch/tetanic force ratios than obtained from control units (Fig. 4B, both P < 0.001). Maximal relaxation rates and half relaxation times were similar for paralyzed and control units (P = 0.3 and P = 0.5, respectively).

Figure 4C illustrates the force produced by a paralyzed motor unit in response to trains of stimuli at frequencies between 5 and 100 Hz. This unit generated a twitch force of 12.9 mN and a maximal tetanic force of 35.3 mN, resulting in a twitch/tetanic ratio of 0.36. It also produced half of its maximal tetanic force at only 5.2 Hz. The typical absolute forces produced by 19 paralyzed units (from 11 subjects) when they were stimulated at different frequencies are illustrated in Fig. 5A. Compared with control units, paralyzed units produced weaker absolute forces at all of the stimulation frequencies delivered between 10 and 100 Hz. When normalized to maximal tetanic forces, paralyzed units produced higher relative forces at all frequencies ≤30 Hz (Fig. 5B). Half-maximal force was also produced at significantly lower frequencies in paralyzed units (Fig. 5C, P < 0.007, 8.0 vs. 12.5 Hz).

Figure 5D compares the distributions of twitch and maximal tetanic forces for all of the paralyzed and control units expressed as a percentage of the total cumulative tetanic force for
et al. 2006), but there was no change in EMG latency (0.0001, respectively) as did the EMG of control units (Thomas et al. 1990). However, the posttetanic twitch contraction time for paralyzed units remained significantly longer than that for control units, whereas half relaxation time remained similar for each population.

**Relationships between twitch and tetanic properties**

Paralyzed units with strong initial twitch forces had strong tetanic forces (Fig. 6A, rs = 0.90, P < 0.001). Twitch contraction time and half relaxation time covaried (rs = 0.66, P < 0.001). However, there were no significant correlations between twitch or tetanic force and twitch contraction time, twitch half relaxation time or tetanic half relaxation time (rs ranging from −0.24 to −0.03). Conduction velocity was negatively correlated to initial twitch contraction time (Fig. 6B, rs = −0.32, P < 0.03 for paralyzed units) but not to any other twitch or tetanic force or speed parameter. Similar trends were observed in control data (Thomas et al. 1990, 1991).

Paralyzed units with stronger tetanic forces reached half-maximum force at higher stimulation frequencies (rs = 0.67, P < 0.001) as did paralyzed units with lower twitch/tetanic force ratios (Fig. 6C, rs = −0.89, P < 0.001). Only the latter relationship was significant in control units (rs = −0.72, P < 0.001). The stimulation frequency needed to evoke half-maximum force was not correlated to twitch contraction time, half relaxation time, or tetanic half relaxation time in paralyzed units, whereas control units with faster twitch contraction times and faster twitch or tetanic half relaxation times needed higher frequencies to reach half-maximum force (Thomas et al. 1991). For the paralyzed units, there were no significant correlations between any twitch or tetanic parameters and injury level or injury duration.

**DISCUSSION**

The present data show that thenar motor units have small EMGs and tetanic forces after years of paralysis due to SCI. Twitch forces were strong, however, resulting in substantial force production just by unit recruitment, and thus less force, relative to maximum, from increases in stimulation frequency. The conduction velocities, twitch contraction times, and EMG durations were shifted toward slower values, but the elevated twitch/tetanic force ratios, not speed, correlated strongly with half-maximal force being achieved at low frequencies. Nevertheless the distributions of many paralyzed motor unit properties shifted uniformly compared with those of control thenar units (Thomas et al. 1990, 1991; Westling et al. 1990), implying that chronic paralysis influenced all motor units similarly. Except for twitch force, these parameters have never been examined at the motor unit level after human SCI.
Conduction velocity

Paralyzed units had significantly slower conduction velocities than control units despite being recorded at similar temperatures (31–33°C, Fig. 2C) (Dengler et al. 1988; Kakuda et al. 1992; Westling et al. 1990), whereas axon conduction velocity was unchanged or increased in cats after cord transection (Cope et al. 1986; Mayer et al. 1984; Munson et al. 1986). The slow conduction velocity in paralyzed thenar units may reflect deficits in the axon, neuromuscular junction, or muscle. Interestingly, conduction from the stimulus site to the spinal cord in seven paralyzed units with F waves was similar to control data, so conduction along these axons may not be slow. However, ventral roots often contain large-diameter axons with thin myelin after human SCI (Thomas and Grumbles 2005), a feature expected to slow axon conduction (Waxman 1989). The decline in axon conduction velocity with paralysis may also reflect decreases in muscle use (Robinson et al. 1991), chronic motor-unit activation at low frequencies (Munson et al. 1997; Zijdewind and Thomas 2001), nerve damage due to mechanical pressure at the elbow or wrist (Boninger et al. 1999; Clark et al. 2003), or the consequences of axon sprouting and muscle reinnervation after the motoneuron death that can accompany SCI (Thomas and Zijdewind 2006; Thomas et al. 2002; Yang et al. 1990). Following partial nerve severance uninjured axons can conduct slowly (Havton et al. 2001; cf. Huizar et al. 1977). Axon sprouts may also be remyelinated incompletely and form small terminals resulting in slow neuromuscular transmission (Rochel and Robbins 1988). Reduced conduction would also be expected along atrophic muscle fibers (Cruz-Martinez and Arpa 1999; Farina et al. 2005; Stålberg 1966), consistent with our prolonged EMG durations (Fig. 3A). If our surface electrodes were 5 mm from the endplates, the muscle conduction latency would be 2.5 ms in paralyzed fibers conducting at \( \sim 2 \) m/s but 1.0–1.7 ms in control fibers conducting at \( \sim 3–5 \) m/s, similar to the 0.7-ms latency difference between the groups.

Unit forces

Most paralyzed units had weak tetanic forces and small EMGs (Figs. 4A and 3B), consistent with the tetanic force deficits and atrophy in cat units after spinal transection (Cope et al. 1986; Mayer et al. 1984; Munson et al. 1986). Whether specific tension and innervation ratios were also changed is
unclear. Unit specific tension declined in the cat hind limb months after spinal transection (Cope et al. 1986; Munson et al. 1986) but increased in whole rat soleus after a year of paralysis (Lieber et al. 1986), a time closer to our data. Because thenar muscles often have reduced numbers of motor units after SCI (Thomas et al. 2002; Yang et al. 1990), innervation of more fibers by surviving axons may explain how some paralyzed thenar units, and muscles, have forces similar to controls. Still, lower tetanic forces for most paralyzed units suggest that fiber atrophy was severe enough (Martin et al. 1992; Rochester et al. 1995a) to counteract any force enhancement from increases in innervation ratios and/or specific tension.

FIG. 4. Tetanic properties of paralyzed ($n = 32$) and control ($n = 30$) motor units. Distributions of maximal tetanic force (A) and twitch/tetanic force ratios (initial and posttetanic, B). C: force produced by a paralyzed unit in response to stimuli ranging from 5 to 100 Hz.

FIG. 5. Force frequency relations. A: absolute forces for 19 paralyzed units from 11 spinal cord injury (SCI) subjects evoked by stimulation with trains of pulses at different frequencies (thin lines). Symbols and thick lines show median data for 32 paralyzed and 30 control units. Stimuli were not delivered to control units at 40 Hz. B: median (5th and 95th percentiles) forces evoked by stimuli at different frequencies with data for each unit expressed relative to its maximal tetanic force. C: distributions of the stimulus frequencies needed to evoke half-maximum force. D: cumulative twitch and tetanic forces expressed relative to the sum of the tetanic forces generated by 32 paralyzed units and 30 control units (Thomas et al. 1991).
Despite weak tetanic forces, the initial twitch forces of paralyzed thenar units were stronger than control data (Fig. 4D) as found by others (Thomas et al. 2002; Yang et al. 1990). This divergent behavior of twitch and tetanic forces suggests that twitch forces may not reflect unit strength after SCI, other neurological conditions and aging (Doherty and Brown 1997; McComas et al. 1971; Yang et al. 1990). Thus recording tetanic forces is important if we are to optimize stimulation parameters to control force and movement in paralyzed limbs.

Contractile speed

Twitch contraction times of thenar motor units were prolonged by chronic paralysis (Fig. 3E) as found by Yang et al. (1990), whereas half relaxation times were unchanged (Fig. 3F). In contrast, reductions in both contraction and half-relaxation times occurred in all cat unit types after spinal transection (Cope et al. 1986; Mayer et al. 1984) as expected from the prevalence of fast type fibers and from increases in muscle stiffness due to more connective tissue (Bawa et al. 1976; Lieber et al. 1986; Williams et al. 1988). Thus thenar unit twitch contraction and half relaxation times may not accurately reflect paralyzed muscle fiber type, consistent with the insignificant relationships found between twitch speed and the stimulation frequencies needed to generate half-maximal force. These same discrepancies have been reported for whole paralyzed human muscles (Gerrits et al. 1999; Rochester et al. 1995b; Thomas 1997a). Yet twitch speed strongly influences force fusion in control units (Botterman et al. 1986; Kernell et al. 1983; Thomas et al. 1991).

Changes in calcium release with chronic paralysis, and/or the sensitivity of actin-myosin to calcium, may prolong thenar twitch contraction times independently of half relaxation time, possibly via activation of calcineurin (Baar 2005). These suggestions are consistent with the stronger twitch forces of paralyzed units, their reduced posttetanic twitch potentiation (Fig. 3D), and with stronger evoked forces (relative to maximum) in paralyzed compared with control thenar muscles after differences in muscle stiffness have been minimized (Baudry and Duchateau 2004; Griffin et al. 2002; Howell et al. 1997). Chronic reductions in use also slow twitches (Duchateau and Hainaut 1990). Whether the spontaneous firing of paralyzed thenar units at low rates (Zijdewind and Thomas 2001) reflects a decline in use is unclear. Nevertheless, this activity seems insufficient to slow the entire twitch as occurs with chronic low-frequency stimulation (Gordon et al. 1997; Kernell et al. 1987). Thus the relationships typically observed between thenar unit twitch speed, force gradation and fiber types are masked after an average of 10 yr of paralysis. Time-dependent changes, and recording from motor pools close to an injury that probably involves contusion, demyelination and/or motoneuron death (Bunge et al. 1993), likely contributes to the differences between our data and that obtained from cat hind limb units supplied from segments well below a cord transection.

General considerations

The usefulness of these results depends on whether the units are representative of paralyzed thenar muscles. The similar
range of conduction velocities for paralyzed and control units (Fig. 2C) suggests that this is the case because the control axons were characteristic of the entire range of thick myelinated fibers within the median nerve (Westling et al. 1990). Axons with slow and fast conduction velocities were also stimulated within a subject (Fig. 2, A and B). Furthermore, the conduction velocity, contraction time, twitch/tetanic force ratio, and F50 distributions for paralyzed units all shifted uniformly relative to the control data (Figs. 2C, 3E, 4B, and 5C), implying that chronic paralysis changed the electrical and mechanical properties of all units similarly but not the range of data. Thus changes in motor-unit use (Zijdewind and Thomas 2001) do not seem to be a primary determinant of the range of paralyzed motor-unit properties or the interrelationships between axon and muscle unit parameters, consistent with studies that have imposed either chronic low frequency stimulation (Gordon et al. 1997) or reduced activity on paralyzed units (Cope et al. 1986; Pierotti et al. 1991).

From a functional perspective, the strong twitch and weak tetanic forces of paralyzed units ensured that half-maximal force was achieved by stimulating at only 8 Hz, a typical frequency for recruitment of units during voluntary contractions (Person and Kudina 1972). The difference between the twitch and tetanic force distributions for paralyzed and control units (Fig. 5D) provides an estimate of the force gradation that may result from delivering pulses at higher frequencies, assuming that units are recruited in order of size (force) and are representative of thenar muscles (Botterman et al. 1986). The force distributions for paralyzed units separate less than those for control units, suggesting that increases in stimulation frequency make a smaller relative contribution to the total force output in paralyzed muscles than control muscles. For example, when 50% of the available motor-unit population was activated, the cumulative twitch and maximal tetanic forces differed by only 13% for paralyzed units but by 31% for control units. Thenar units influenced by chronic spinal injury also show limited increases in firing rate during voluntary contractions (Zijdewind and Thomas 2003). Thus similar alterations in unit force gradation strategies occur after SCI irrespective of whether their voluntary drive is interrupted or only reduced. Our data also suggest that relatively smooth contractions can be produced in paralyzed units by low frequency stimulation. However, most paralyzed units (80%) would need to be activated to produce half their cumulative force (60% of control units, Fig. 5D). Reductions in tetanic force, twitch potentiation (Fig. 3D), and fatigue resistance (Klein et al. 2006) therefore all limit force generation in chronically paralyzed units.

ACKNOWLEDGMENTS

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GRANTS

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