Tohoku J. Exp. Med., 1999, **187**, 339-351

Electromyographic Study Relating to Shoulder Motion: Control of Shoulder Joint by Functional Electrical Stimulation

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KAMEYAMA, J., HANDA, Y., HOSHIMIYA, N. and SAKURAI, M. *Electromyographic Study Relating to Shoulder Motion: Control of Shoulder Joint by Functional Electrical Stimulation*. Tohoku J. Exp. Med., 1999, **187** (4), 339-351 —— The purpose of this study is to create the standard stimulation patterns of shoulder motion from electromyographic (EMG) data in 13 healthy human volunteers in order to control the movement of the paralyzed shoulder in quadriplegic and hemiplegic patients by functional electrical stimulation (FES). Simultaneous EMG measurement was made at 24 points of 17 major muscles relating to shoulder motion. Since the number of the output channels in the portable FES apparatus is limited, 12 major muscles were selected from statistically processing these EMG data and stimulation patterns were created based on the EMG data of these muscles. Thus three standard stimulation patterns were created to move the shoulder, i.e., (i) 90° flexion to 90° horizontal abduction, (ii) 90° flexion to 20° horizontal adduction, and (iii) 90° abduction to 90° horizontal adduction. With the created stimulation patterns, the restoration of the shoulder motion in plegic patients was successful and it will be reported in the next paper. ——— electromyogram (EMG); shoulder motion; quadriplegic patient; hemiplegic patient

Functional electrical stimulation (FES) has been of therapeutic interest in restoring the motor functions in paralyzed extremities caused by upper motor neuron disorders. There are reports that electrical stimulation to lower motor neurons provokes muscle contraction in the paralyzed muscles of patients with upper motor neuron disorders (Peckham et al. 1980a, b, 1988; Nathan 1984). The use of suitable stimulation patterns should allow the restoration of functional

Received July 28, 1998; revision accepted for publication April 16, 1999.

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movement of the extremities.

Long and Masciarelli (1963) first used FES to perform prehension and release of the paralyzed hand. Since then, several investigators (Peckham et al. 1980a; Nathan 1984; Handa et al. 1989; Hoshimiya et al. 1989; Keith et al. 1989) have applied FES to the finger, wrist, and elbow.

C4 quadriplegic and some hemiplegic patients cannot use their hands and arms in three dimensional space because of the loss of shoulder control. Electromyographic (EMG) analysis of shoulder motion was made (Inmann and Saunders 1944; Saha et al. 1956; Saha 1973). Our EMG analysis on healthy subjects has confirmed Inmann's and Saha's findings (Kameyama et al. 1990, 1991). The EMG data were integrated and normalized to create standard stimulation patterns for FES, which were applied to stimulate the muscles of the finger, wrist and elbow (Handa et al. 1989; Hoshimiya et al. 1989).

By applying implantable FES to paralyzed upper extremities in C4 quardiriplegics who were wearing the shoulder orthosis to aid their intact trapezius, they restored the motion of their hands and arms which allowed the activity of daily livings (ADL) for eating, brushing the teeth and using the lipstick (Handa et al. 1989, 1992; Hoshimiya and Handa 1989; Hoshimiya et al. 1989).

The implantable electrode is superior to the surface electrode when applying FES to restore shoulder motion, because the muscles can be selectively stimulated. From our experience (Kameyama et al. 1992), simultaneous EMG recording of all the relevant muscles for shoulder motion is needed to create FES patterns, although there was no report dealing with this. As a preliminary step, we investigated EMG at 24 points of the 17 muscles which participate in standard shoulder motion. Statistical analysis of EMG data is expected to advance creation of the standard stimulation patterns, but they had not been brought in our previous studies.

In this paper, we statistically analyzed EMG of the muscles relating to shoulder motion. We also kinesiologically analyzed shoulder motion. From the data, we created some basic stimulation patterns for shoulder movement. Some of the data included in this paper have been reported in preliminary forms (Kameyama et al. 1990, 1991, 1992). The clinical application is described elsewhere (Kameyama et al. 1999).

Subjects and Methods

EMG analysis during motion

EMG analysis was performed on 12 right shoulders and one left shoulder of the 13 healthy human volunteers aged 18 to 30 years (all male subjects, mean age 22.6 years). The examined 17 muscles with major functions are shown in Table 1. The subclavius was excluded because of its minor functions. Since larger muscles such as the trapezius, pectoralis major and deltoid, provide different actions depending on the origin and/or insertion of their muscle fibers, electrodes

TABLE 1. *Muscles studied for electromyograms*

I. Shoulder girdle to forearm (SF group)	III. Trunk to humerus (TH group)
1. Biceps brachii	1. Pectoralis major
1) Short head	1) Clavicular head
2) Long head	2) Sternocostal head
2. Triceps brachii (long head)	3) Abdominal head
	2. Latissimus dorsi
II. Shoulder girdle to humerus (SH group)	IV. Trunk to shoulder girdle (TS group)
1. Coracobrachialis	1. Trapezius
2. Deltoid	1) Upper part
1) Anterior part	2) Middle part
2) Middle part	3) Lower part
3) Posterior part	2. Serratus anterior
3. Supraspinatus	3. Rhomboideus major
4. Infraspinatus	4. Rhomboideus minor
5. Teres minor	5. Levator scapulae
6. Subscapularis	6. Pectoralis minor
7. Teres major	

were inserted separately into the functionally different parts of the muscle. Thus, the number of EMG recording channels was increased to 24. Bipolar intramuscular electrodes made of Teflon-coated SUS 316 stainless steel wire, 75 μm in diameter, (A-M systems, Carlsborg, WA, USA) were implanted percutaneously into the middle of the muscle bellies. The interelectrode distance was 5 mm. Wire insertion was assured by checking the EMG of the muscle during manual muscle testing and/or electrical stimulation.

The following are the shoulder joint movements examined. The reason for selecting these movements is that they control major activities of daily living.

1) Shoulder flexion of 45° and 90° from the neutral position.
2) 45° and 90° abduction from the neutral position.
3) 45° and 90° horizontal abduction from the position of 90° flexion.
4) 20° horizontal adduction from the position of 90° flexion.

While measuring these motions, the extension of the elbow, wrist and finger was almost 0°. The time allowed to move the shoulder joint from 0° to 45° and from 0° to 90° was set to be 2 and 3 seconds (the phasic state in Fig. 1), respectively. When the angle of the shoulder joint reached 45° or 90°, the joint was kept at this position for 5 seconds (the tonic state in Fig. 1).

EMG signals were amplified with our original differential amplifiers and rectified with our original full-wave rectifiers. They were integrated with a time constant of 0.5 seconds (integrated EMG). The data was fed into a data recorder

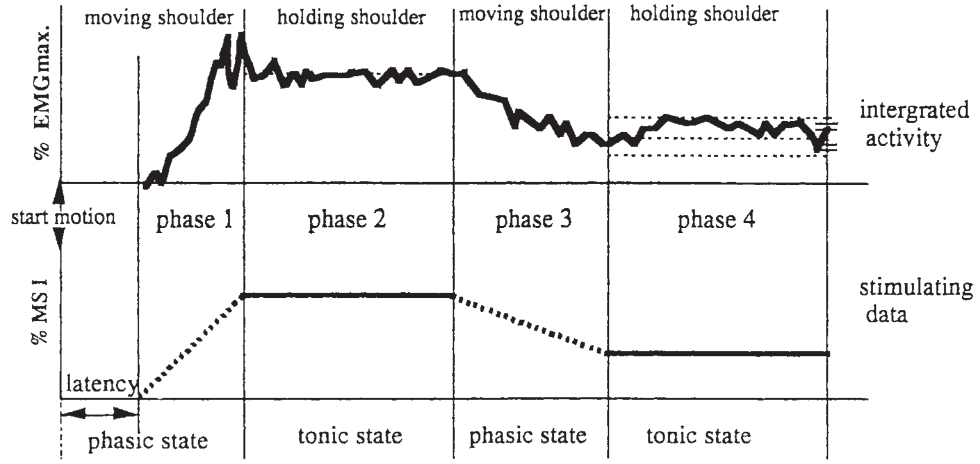

Fig. 1. Diagram of method to creating standard stimulation patterns: From the neutral position, activity induction at 90° shoulder flexion or abduction is assigned as "phase 1", and the position maintaining part as "phase 2", and then horizontal abduction or adduction "phase 3" and the maintaining duration as "phase 4". Stimulus intensity to each paralyzed muscle was determined by the % of maximum stimulus intensity (% MSI) and the rate was adjusted to the mean % EMG max. t. Stimulation patterns was obtained by trapezoidal approximation of the EMG data from phase 1 to 4. The mean latent time from the beginning of 90° flexion or 90° abduction to the onset of the muscle discharge was also taken into account for determining the onset of the stimulation for individual muscles.

with 28 input channels (SR 90, TEAC Co., Ltd., Tokyo) and a pen recorder (RECTI-HORIZ-16K, NEC San-ei, Tokyo).

For quantitative standardization of EMG activities, amplitudes of the integrated EMG during shoulder motion were expressed as a percentage of the maximum voluntary contraction (% EMG max.). As shown in the upper right part of the Fig. 1, fluctuation of % EMG max. was observed even in phase 4 (holding shoulder). Therefore, % EMG max. t, which was the value of the middle point between the highest and lowest points in the phase, was derived for later calculation. The mean (mean % EMG max. t) and the standard deviations of % EMG max. t were calculated from the data of all subjects. The paired *t*-test was used to test for significant differences in muscluar activity with the shoulder joint at 45° compared to that with the shoulder joint at 90°.

Shoulder joint angles were measured with an apparatus developed by Takahashi et al. (1987). This apparatus displays three dimensional shoulder movement in the frontal, sagittal and horizontal direction with the change in angle when the upper arm is rotated.

Creation of standard stimulation patterns

Creation of the basic stimulation patterns requires the data of the waveforms of the integrated EMG changes. Fig. 1, as an example, indicates that the integrated EMG changes are divided into 4 phases:

Phase 1: moving from a neutral position to 90° flexion or abduction.
Phase 2: maintaining the positions in phase 1.
Phase 3: moving from 90° flexion to 20° horizontal adduction or 90° horizontal abduction, or from 90° abduction to 90° horizontal adduction
Phase 4: maintaining the positions in phase 3.

Fig. 1 also shows the time lag (latency) which indicates the gap between the time when the shoulder motion is started and when the muscle dischage appears. This latency value differs among muscles, because the time of discharge is not the same. That is, the latency determines the starting order of activation of the relevant muscles.

Stimulation patterns were created using the EMG data from phase 1 to 4. The % of maximum stimulus intensity (% MSI) used for stimulating a paralyzed muscle was set equal to the mean % EMG max. t.

Results

EMG analysis of the shoulder movement

% EMG max. t indicates the percentage of the integrated EMG at the maximum voluntary contraction. We compared the obtained % EMG max. t of 24 points of 17 muscles relating to specific shoulder motions.

Fig. 2 shows the results of the EMG analysis on shoulder flexion. Among the muscles which connect the shoulder girdle to the humerus (the SH group), the coracobrachialis and anterior part of the deltoid show the highest mean % EMG max. t (Fig. 2A), followed by the supraspinatus and infraspinatus. Among the muscles which connect the shoulder girdle to the trunk (the TS group), the serratus anterior indicates the highest mean % EMG max. t, followed by the levator scapulae, and the lower and upper parts of the trapezius. Among the muscles which connect the trunk to the humerus (the TH group), the clavicular head of the pectoralis major, which is a prime mover of shoulder flexion together with the coracobrachialis and anterior part of the deltoid, shows a high value of mean % EMG max. t. This value was only 10% even at 90° shoulder flexion, and was lower than that of other prime movers. Comparing the mean % EMG max. t of 45° flexion with that of 90° flexion, the latter was always higher except for the supraspinatus. The increment of mean % EMG max. t from 45° flexion to 90° flexion was significantly high (paired *t*-test) in the coracobrachialais ($p<0.05$), anterior part of the deltoid ($p<0.05$), middle part of the deltoid ($p<0.01$), the infraspinatus ($p<0.05$), upper trapezius ($p<0.01$), serratus anterior ($p<0.01$), and levator scapulae ($p<0.05$). This suggests that these muscles contribute greatly to shoulder flexion.

The mean latency from the initiation of shoulder flexion to the appearance of discharge of the individual muscles (Fig. 2B) shows that, when shoulder flexion started, the coracobrachialis and anterior part of the deltoid contracted followed by the supraspinatus, infraspinatus and serratus anterior. The latency of the

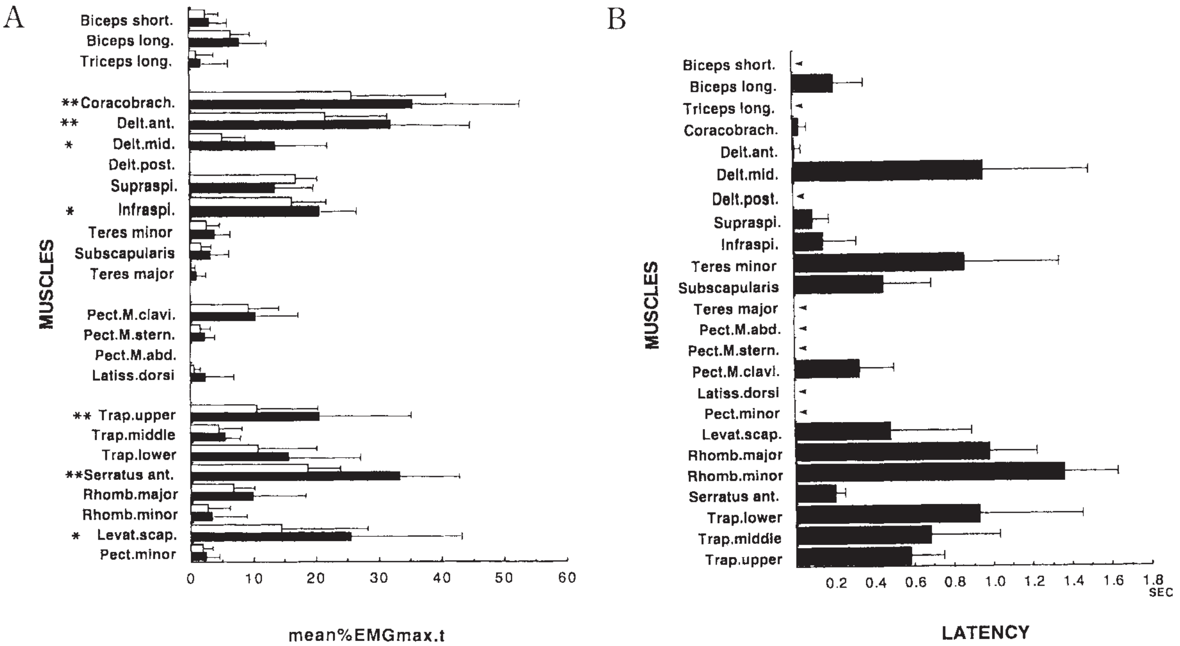

Fig. 2. Kinesiological analysis of shoulder flexion.
A: Histogram of the mean % of EMG during maximum voluntary contraction: (tonic) (mean % EMG max. t) when the shoulder flexion is 40° and 90°. Comparing the mean % EMG max. t's at 45° flexion with that of 90° flexion, they were always higher at 90° flexion except in the supraspinatus. Increments of muscle activities in the coracobrachialais, anterior part of deltoid, middle part of deltoid, the infraspinatus, upper part of trapezius, serratus anterior and levator scapulae were significantly high (paired-t, $^{*}p<0.05$, $^{**}p<0.01$). □, flex. 45 deg.; ■, flex. 90 deg.
B: Latency from initiation of the shoulder flexion to onset of muscle discharge is shown: Coracobrachialis and anterior part of deltoid contracted without delay from the movement, and then the supraspinatus, infraspinatus and serratus anterior followed in turn. ◀, it indicates that impossible to measure latent time.

clavicular head of the pectoralis major was 0.3 seconds, and that of the upper and lower parts of the trapezius, which relates to upward rotation of the scapula (the anatomical textbook), was 0.58 seconds and 0.92 seconds, respectively.

As for shoulder abduction, the mean % EMG max. t's of the middle part of the deltoid ($p<0.01$) and the supraspinatus (the prime movers) ($p<0.05$) were significantly high. Comparing the % EMG max. t between 45° and 90° shoulder abduction in the individual muscles, the mean % EMG max. t of 90° shoulder abduction exceeded that of 45° abduction in most of the muscles tested (Fig. 3A). Mean % EMG max. t of almost all muscles belonging to the TS group was above 10%. The middle part of the trapezius and rhomboideus minor, which act as the scapular adductors, showed a high mean % EMG max. t, while a comparison of the values between 45° and 90° abduction showed no significant differences. The mean % EMG max. t values of the muscles (the SF group) which connect the shoulder girdle to the forearm and the TH group were below 5%.

The latency of the middle part of the deltoid was almost zero, followed by

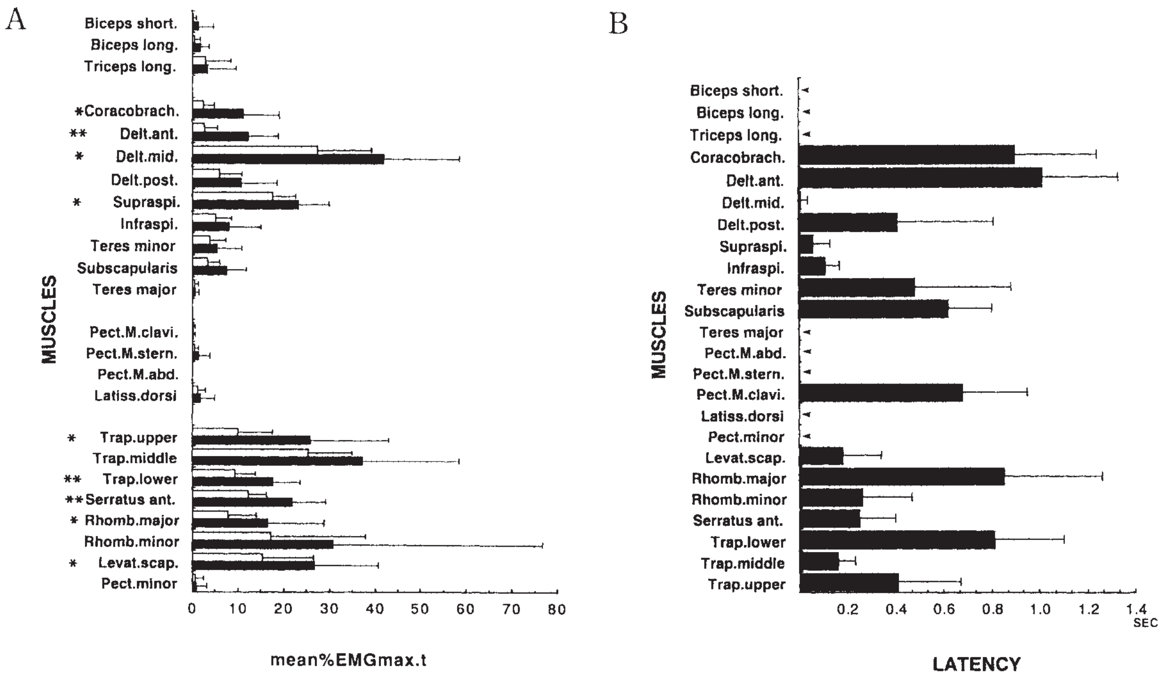

Fig. 3. Kinesiological analysis of shoulder abduction.
A: The mean % EMG max. t at the shoulder abduction 45° and 90° is shown: % EMG max. t of prime movers, that is, the middle part of the deltoid and supraspinatus are markedly high, and the value at abduction 90° is significantly higher than the value at abduction 45° (*$p<0.05$, **$p<0.01$). Muscles in the TS group showed mostly larger than 10% of % EMG max. t. The % EMG max. t of the shoulder girdle to forearm (SF) group and TH group showed lower than 5%. (paired-*t*, *$p<0.05$, **$p<0.01$). □, abd. 45 deg.; ■, abd. 90 deg.
B: The latency in the middle part of the deltoid was zero, and then the supraspinatus, infraspinatus and the middle part of the trapezius showed small values. ◀, it indicates impossible to measure latent time.

shorter latencies of the supraspinatus, infraspinatus and the middle part of the trapezius (Fig. 3B)(This result suggests that the above muscles apparently contributed to the motive power which initiates shoulder abduction.).

Fig. 4 shows the mean % EMG max. t of the movement of the horizontal abduction from the position of 90° shoulder flexion. In this movement, mean % EMG max. t of the middle part of the deltoid was the highest, and moreover the increment of the mean % EMG max. t from 45° to 90° horizontal abduction was the largest ($p<0.01$). The % EMG max. t of the posterior deltoid, which is the prime mover in horizontal abduction, was 2% at 45° and horizontal abduction and increased to 12% at 90° (six times the increment: $p<0.01$). Most of the muscles in the TS group showed a high mean % EMG max. t during 45° horizontal abduction. Among these muscles, the rhomboideus and the middle part of the trapezius, which participate in scapular adduction, showed a significant increase in % EMG max. t at 90° ($p<0.05$). By contrast, mean % EMG max. t of the serratus anterior, which acts as a scapular abductor, showed a significant decrease ($p<0.01$).

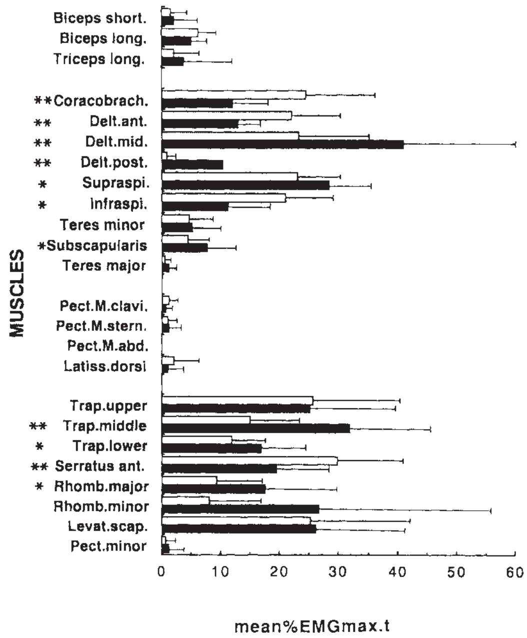

Fig. 4. The result of analysis of EMG in the movement of the horizontal abduction from the position of 90° shoulder flexion is shown: In this movement, % EMG max. t of the middle part of the deltoid muscle was the largest. The increment rate of % EMG max. t from 45° horizontal abduction to 90° was also the largest ($p < 0.01$). The % EMG max. t of the posterior part of the deltoid, which is the prime mover in the horizontal abduction, was as low as 2% at 45° of horizontal abduction and increased to 12% at 90° (six times increment: $p < 0.01$) (paired-t, $^*p < 0.05$, $^{**}p < 0.01$). □, horiz. abd. 45 deg.; ■, horiz. abd. 90 deg.

The mean % EMG max. t at a horizontal adduction of 20° from the position of 90° shoulder flexion was remarkably high in the coracobrachialis, anterior part of the deltoid, clavicular head of the pectoralis major, and serratus anterior in Fig. 5. The mean % EMG max. t of these muscles at 20° horizontal adduction were significantly high ($p < 0.05$), when compared with those at 0° horizontal adduction (flex. 90°). The mean % EMG max. t of the sternocostal head of the pectoralis major increased. On the other hand, the % EMG max. t of the scapular abductors at 20° horizontal adduction was below 5%, and this value was smaller than that at the horizontal adduction of 0°.

The result of the analysis on the upper arm movement using computerized shoulder motion measuring system developed by Takahashi et al. (1987) showed that shoulder flexion of 45° and 90° induced a humerus internal rotation of 6.3° and 7.3° on average, respectively, while 45° and 90° abduction elicited a humerus external rotation of 6.3° and 31.7° on average, respectively (Table 2).

Creation of standard stimulation patterns

Since the number of the output channels in the portable FES apparatus is 30 (two patterns of 14 and 16 channels each), twelve muscles were selected and stimulation patterns for muscles were created based on the EMG data. The

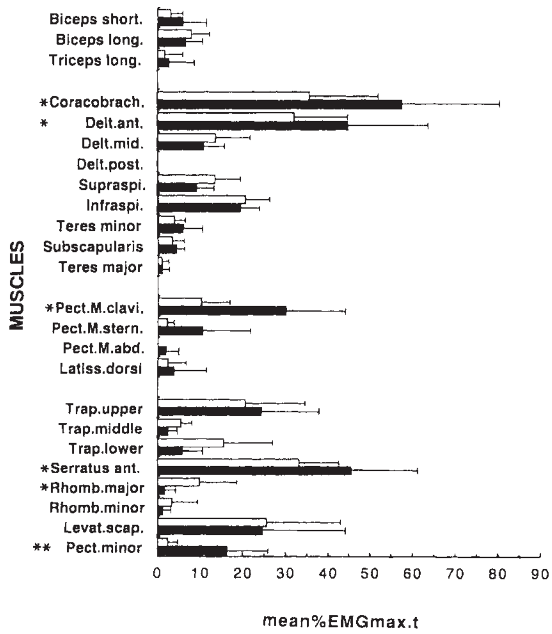

Fig. 5. % EMG max. t at horizontal adduction of 20° from the position of 90° flexion is shown: It is markedly higher in the anterior part of the deltoid, coracobrachialis, clavicular head of the pectoralis major and serratus anterior, and it increased significantly from the % EMG max. t at flexion 90° and horizontal adduction 0° ($p<0.05$) (paired-t, *$p<0.05$, **$p<0.01$). □, horiz. add. 0 (flex. 90); ■, horiz. add. 20 deg.

TABLE 2. *Analysis of movements of the upper arm by using the apparatus for measuring shoulder motion*

Motion	Subject 1	Subject 2	Subject 3	Mean	S.D.
Flexion 45°	−7°	−2°	−10°	−6.3°	3.2°
Flexion 90°	−7°	−4°	−11°	−7.3°	2.8°
Abduction 45°	+4°	+11°	+4°	+6.3°	3.2°
Abduction 90°	+39°	+33°	+23°	+31.7°	6.6°

The result of analysis showed that shoulder flexion of 45° and 90° induced internal rotation of the humerus of mean 6.3° and 7.3°, respectively, while 45° and 90° abduction elicited the mean 6.3° and 31.7° external rotation of the humerus, respectively.

−, Internal rotation; +, External rotation.

following shows the criteria for selecting the muscles:

(a) The muscles whose % EMG max. t is below 10% in the shoulder motion were excluded.

(b) The downward rotators of the scapula were excluded, since they are less important to the relevant shoulder movement.

(c) The muscles with a % EMG max. t slightly over 10%, whose function can be compensated by other muscles, were excluded.

According to criteria (b) and (c), the levator scapulae (downward rotators), the

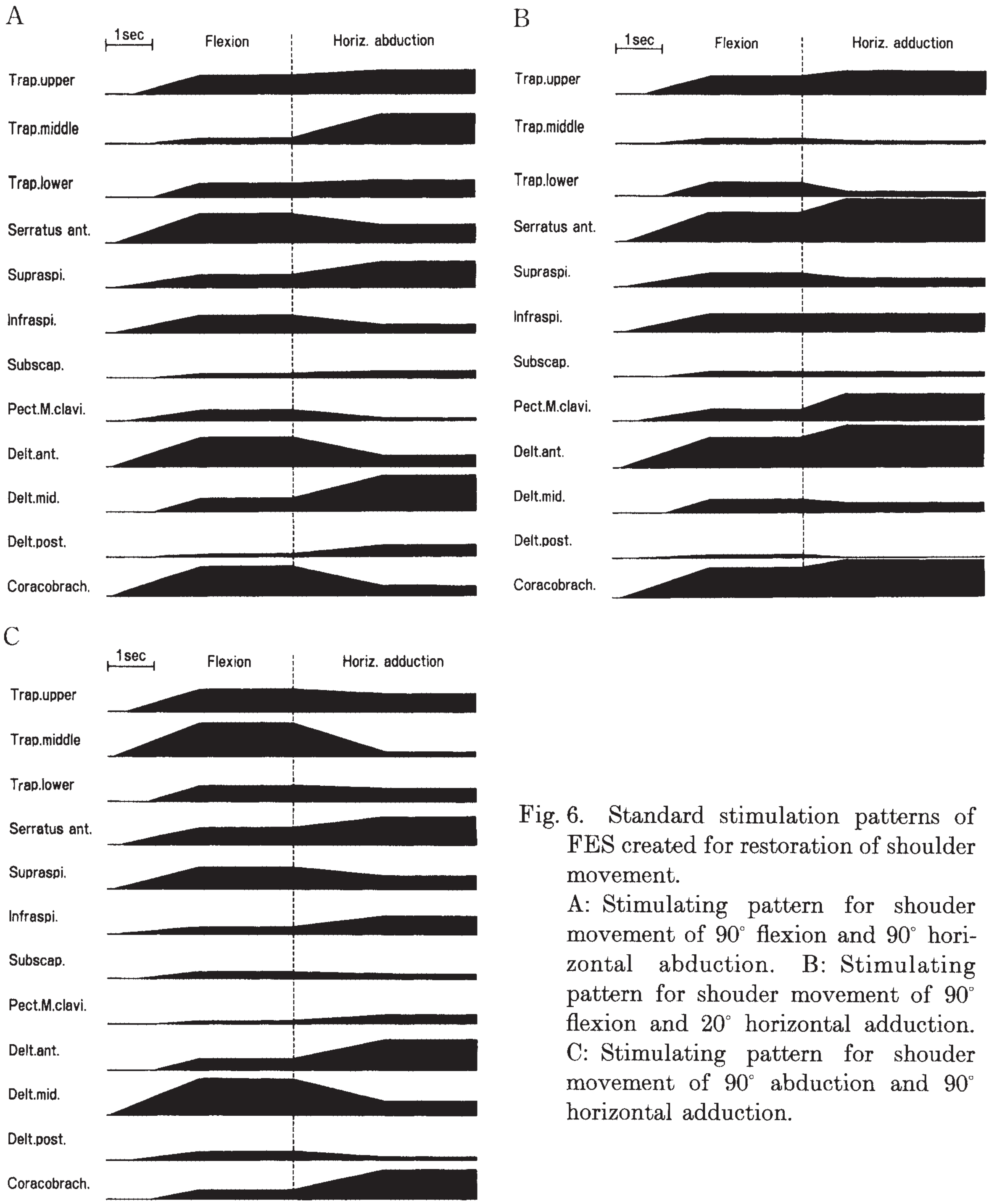

Fig. 6. Standard stimulation patterns of FES created for restoration of shoulder movement.
A: Stimulating pattern for shouder movement of 90° flexion and 90° horizontal abduction. B: Stimulating pattern for shouder movement of 90° flexion and 20° horizontal adduction. C: Stimulating pattern for shouder movement of 90° abduction and 90° horizontal adduction.

rhomboideus (compensated by the middle part of the trapezius), and the sternocostal head of the pectoralis major and the pectoralis minor (compensated by the clavicular head of the pectoralis major and anterior part of the deltoid, respectively) were eliminated.

Three standard stimulation patterns were created based on Fig. 1. Figs. 6A, 6B and 6C show three kinds of the standard stimulation patterns for the 12 muscles, (i) 90° flexion to 90° horizontal abduction, (ii) 90° flexion to 20° horizontal adduction, and (iii) 90° abduction to 90° horizontal adduction. The threshold and maximum stimulation voltage data of individual muscles were input into the portable FES apparatus (Handa et al. 1989, 1992; Hoshimiya and Handa 1989; Hoshimiya et al. 1989).

Discussion

The anatomy of the shoulder is complicated and provides sophisticated upper extremity functions. In particular, the muscles allow a wide range of motion in the scapulohumeral joint by co-contracting of the muscles in a cooperative manner, and this makes shoulder control by FES especially difficult.

The range of shoulder movement is large, and three dimensional movement of the scapula and clavicle could not be detected in the experimental set up. Using computerized measuring system developed by Takahashi et al. (1987), however, we successfully analyzed shoulder motion (e.g., shoulder flexion of 45° and 90° induced a humerus internal rotation of 6.3° and 7.3° on average, respectively, while 45° and 90° abduction elicited a humerus external rotation of 6.3° and 31.7° on average, respectively) and muscular activities relating to the shoulder joint. Flexion and abduction of the shoulder joint were followed by internal and external rotations of the upper arm, respectively, and this suggests that activities of the rotator cuff muscles are closely related to the movement of the upper arm. From the data, the musclular activities of the shoulder joint in the spatio-temporal contraction were more precisely analyzed. On the basis of the EMG analysis of the shoulder movement in normal subjects, we created standard stimulation patterns of FES for controlling shoulder motion. The results confirming coordinated and reproducible control of the shoulder by FES are described elsewhere (Kameyama 1999).

Keith et al. (1989) and Kilgore et al. (1987) stated that the EMG recording of dynamic motion is useful for analyzing physical and mechanical functions of hand movements, but the obtained data cannot be directly used to create the FES stimulation patterns. They also reported that the contraction properties of the paralyzed muscle induced by electrical stimulation are supposed to be different from those of healthy muscles. It is true that the activation process of the muscle by the electrical stimulation is different from the orderly activation process (orderly recruitment) of the muscle controlled by the normal central nerve system (CNS). Basmajian et al. (1967) have mentioned, however, that muscluar activity in integrated EMGs using intramuscular electrodes is proportional to muscle strength (Takahashi et al. 1987).

Our research team found that coactivation of the antagonist with the synergist worked well to smooth the curve of muscle strength vs. stimulus intensity of the synergist, and made the slope of the curve smaller (Watanabe et al. 1992).

There are some problems to solve, in improving the application of FES to the shoulder. The % EMG max. t is not the only factor of muscle contraction, and the cross-sectional area of the muscle, which represents exertion power, possibly should be considered. For example, in the flexion of the shoulder joints, the coracobrachialis showed a % EMG max. t value which belongs to the largest % EMG max. t group, while % EMG max. t of the clavicular head of the pectoralis major was relatively low. In fact, the muscular activities of the clavicular head of the pectoralis major is much stronger than the coracobrachialis because of the larger cross-sectional area of the muscle, and this suggests that the clavicular head of the pectoralis major may play a more important role. The prime movers of shoulder flexion are the anterior part of the deltoid and clavicular head of the pectoralis major, with the coracobrachiales as one of the synergists.

We had to specifically select shoulder muscles to perform FES control, since the number of the output channels of our FES system is limited. If we could have simultaneously stimulated more muscles, more smoothly coordinated shoulder joint movements would have been produced.

Acknowledgment

The authors thank Ichie, M., M.D. (Director of the Hokuryo Clinic) for permitting us to see the patients (our cases) in his hospital.

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