Difference in Changes of Surface EMG during Low-level Static Contraction between Monopolar and Bipolar Lead

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Abstract. The changes in surface electromyogram(EMG) during sustained contractions were compared between monopolar and bipolar lead. Ten subjects performed static elbow flexion at contraction levels of 10 and 40% of maximum voluntary contraction(MVC). Seven subjects performed static knee extension at contraction levels of 3, 8, and 15%MVC. Surface EMGs were recorded from the biceps brachii, vastus lateralis, rectus femoris and vastus medialis in monopolar and bipolar lead. Mean amplitude of EMG(AEMG) and frequency spectra of EMG were calculated.

AEMG was larger in monopolar EMG than in bipolar EMG except 3 cases. AEMG changed more abruptly in bipolar EMG than in monopolar EMG. Relative power in low frequencies increased(slowing) during contractions in all cases of monopolar EMG. But bipolar EMG sometimes did not show slowing. Decrease of relative power in low frequencies(quickening) was sometimes seen during contractions. This quickening was not marked in monopolar EMG because of its large slowing.

This study shows that obscure EMG slowing during a low-level fatiguing contraction is related to the lead methods. Led amplitude was compared between the lead methods using a simple model. The sphere that EMG reflects(pickup sphere) is larger in monopolar EMG than in bipolar EMG. This difference in the pickup sphere is thought not to be the main reason for the difference in the slowing. (*APPL Human Sci*,14(2) : 79-88,1995)

Keywords: surface EMG, fatigue, spectra, monopolar lead, low-level contractions

Introduction

Surface electromyogram(EMG) has been studied for the estimation of muscle load and/or fatigue. In most studies EMG was usually measured in bipolar lead. Bipolar lead has the following advantages compared with monopolar lead.

1) Noise from artifacts and AC power supply of equipment is low in bipolar lead. This advantage is impor-

tant for the measurement of actual working condition where many noise sources exist.

2) Besides noise from equipment, electrocardio-potential and EMG of other muscles are easier to be detected in monopolar lead than in bipolar lead.

3) Muscles at many parts participate in actual works. In monopolar lead caution is needed to decide the reference position where electric potential is stable.

Some factors that influence EMG have been studied, such as inter-electrode distance, size of electrode, directions of an electrode pair relative to muscle fibers, and relations between the endplate and electrodes(Sato, 1964; Lynn et al., 1978; Kaneko et al., 1990). The effects of relative position of electrodes to the endplates are large and must be cautioned. In bipolar EMG led potential is diminished by simultaneous firing of other muscle fibers. Bipolar EMG is not appropriate to the examination of the shape of action potential, since the potential in bipolar EMG is a relative one. Though the information in bipolar EMG is less than that in monopolar EMG such as non-conductive component(Morimoto and Kamo, 1990), the restriction of lead signal is also an advantage of the bipolar lead. In surface EMG it is difficult to identify motor units as clear as needle or wire EMG. Frequency spectra are used to analyze surface EMG that is composed of action potential of many motor units. Inter-electrode distance influences the sphere that EMG reflects(pickup sphere) in bipolar lead(Lynn et al., 1978). In short inter-electrode distance the duration of led potential is short and the identification of motor units becomes possible in some conditions(Akazawa et al., 1985). In monopolar EMG pickup sphere is broader than in bipolar EMG, and the state of potential at the reference point can influence the EMG. These imply that monopolar EMG is more complicated than bipolar EMG. This less complicated nature of bipolar EMG is probably the main reason that bipolar EMG is used in most studies.

Increase in EMG amplitude and compression of EMG spectrum towards lower frequencies(slowing) during fatiguing contractions are noted for the evaluation of muscle fatigue. Though a reduction in conduction veloc-

ity of muscle action potential is one factor that causes the slowing of EMG, this reduction in conduction velocity cannot explain the slowing of EMG completely (Krogh-Lund and Jørgensen, 1993; Zwarts et al., 1987). Synchronization and/or grouping of motor units discharges are thought to be the other reasons for the slowing (Bigland-Ritchie et al., 1981; Naeije and Zorn, 1982). Though strict synchronization of motor units discharges is not the main reason of the EMG slowing in low level contractions (Ohashi, 1993), the grouping discharges of motor units are not denied as the reason for the slowing. If grouping discharges are composed of motor units that disperse in a whole muscle, EMG in which the pickup sphere is broad would be sensitive to detect the grouping discharges. Monopolar EMG would be easier to detect the slowing than bipolar EMG.

Though differences of monopolar and bipolar EMG were reported as a fundamental study (Gydikov and Kosarov, 1972), differences of the EMG slowing between the lead methods have not been studied during low-level contractions. Sato and Tsuruma (1967) proposed to apply frequency analysis to monopolar EMG. The main reason for the preference of monopolar EMG, however, is not concerned with the EMG slowing but to avoid the effects of placements of electrodes. Chaffin (1973) showed that the slowing of surface EMG is useful to evaluate the muscle fatigue, and reported using monopolar EMG. However, no reason was given for choosing monopolar lead. Though this report was referred in many reports, none noticed that monopolar EMG was being used. The differences of EMG changes during fatiguing contractions between monopolar and bipolar EMG would not be considered as an important problem.

The slowing of surface EMG during contractions was expected to be useful as an objective index of muscle fatigue (Kogi and Hakamada, 1962). The slowing, however, is small in low-level contractions (Ohashi, 1993), and even quickening of EMG is sometimes seen with muscle fatigue (Hagberg, 1981; Malmqvist et al., 1981; Hägg, 1991). The small and uncertain slowing poses a serious problem in the application of EMG for the evaluation of muscle fatigue. If the slowing of EMG is clearer in monopolar lead than in bipolar lead, EMG in monopolar lead will be useful for the application of EMG in spite of its demerit as mentioned above.

In this paper the differences in EMG changes during fatiguing contractions are shown between monopolar and bipolar EMG. Static contractions mainly in low contraction levels where the estimation of muscle fatigue is important in light works such as office work are being investigated. It should be noted that the conduction velocity of muscle action potential is known not to be the main factor in EMG slowing in low-level contraction (Krogh-Lund and Jørgensen, 1993; Ohashi, 1993).

Methods

Subjects were seated on a chair of a multipurpose muscle strength measuring apparatus (Takei Kiki Kogyo 1281). Static contractions of the elbow flexors (EF) and the knee extensors (KE) were performed by holding weights. The elbow angle was kept at 90 degrees with the forearm supinated and with the upper arm being on a horizontal arm rest. The knee angle was kept at 90 degrees. The weights pulled the wrist and the ankle horizontally with a pulley.

Contraction levels were set at 10 and 40% of maximum voluntary contraction (MVC) in EF and 3, 8 and 15% MVC in KE. Subjects were asked to continue the contractions until the set time; 40min for EF at 10% MVC, 2min for EF at 40% MVC, 60min for KE at 3% MVC, 30min for KE at 8% MVC and 5min for KE at 15% MVC. In EF at 40% MVC subjects were asked to continue the contraction as long as they could. In KE at 8 and 15% MVC subjects were asked to continue the contractions, if they did not wish to stop the contraction at the set time. Intervals of the experiments were longer than a day.

MVCs were measured with strain gauge dynamometer. The body was stabilized by three belts secured around the trunk. The subjects were told to keep that they kept contraction maximum for 2 to 3 seconds, and then contracted more strongly momentarily during the maximum voluntary contraction. The peak force of the contraction was used as MVC. MVCs were measured for 2 or 3 times, and the largest value was used as MVC. MVCs ranged between 191N and 311N with its average of 263N in EF and between 580N and 727N with its average of 664N in KE.

Subjects were male volunteers whose ages ranged from 22 to 32 years. The number of subjects was 10 (subj A-J) in EF and 7 (subj A-G) in KE. Subject J did not perform the 40% MVC condition of EF. Subject G of KE was the author.

Surface EMG was recorded from the long head of biceps brachii (BB) with an array electrode in EF. The electrode consisted of 9 parallel stainless steel wires parallel to each other with the inter-electrode interval of 5mm. The length of the wire was 10mm and its diameter was 0.5mm. These wires were fixed on a rubber sheet with a thickness of 2mm. This electrode was attached between the center and the distal part of the muscle with adhesive tapes. Electrodes for the ground and the reference were placed over the dorsum of the forearm at the points which was 2 to 6cm apart from the elbow. Bipolar EMGs were led from two successive electrodes of wires. EMG for analysis was selected so that no motor point existed between the electrodes. The electrode for monopolar EMG was common to one of the electrodes

for bipolar EMG.

In KE surface EMG was led from vastus lateralis(VL), rectus femoris(RF) and vastus medialis(VM) with disc electrodes. The electrodes were 10mm in diameter and were attached parallel to the muscle fibers. The centers of the two electrodes were 30 to 40mm apart. Electrodes for the ground and the reference were placed over the anterior surface of the tibia. The electrode for monopolar EMG was common to one of the electrodes for bipolar EMG.

EMG was amplified at a time constant of 0.1sec and with high cut at 3kHz. EMG was recorded with a PCM magnetic tape recorder(TEAC RD-110T).

EMG was sampled at 1024Hz. Mean amplitude of EMG(AEMG) was calculated as the mean value of amplitude. AEMG was used as an absolute value when compared between lead methods. When AEMG was used as a relative value, the AEMG from 1/20 to 1/10 of contraction time was used as the base value and was set to 1. EMG power spectrum was calculated by FFT method. Relative power to the total power in 8-400Hz was obtained for 8-22, 22-42, 42-62, 62-82, 82-122, 122-202, 202-400Hz. The relative power in 8-22 and in 22-42Hz was used as indices of EMG slowing, and was called RPWL20 and RPW22-42. The relative power and AEMG was calculated continuously for the analysis time of 16sec in KE at 3 and 8 %MVC and EF at 10%MVC, 8sec in KE at 15%MVC, and 4sec in EF at 40%MVC.

Subjective sensations were asked to be reported free-

ly.

Wilcoxon matched-pairs signed-ranks test and linear correlation coefficient were used for statistical analysis. The level of significance was chosen at $p < 0.05$.

Fatigue levels were not evaluated quantitatively in this experiment. I assumed that fatigue would develop roughly with contracting time though synergist's help sometimes lightens fatigue level of a muscle during a contraction. Correlation coefficients of AEMG and RPWL to contracting time were used to evaluate the relations of AEMG and RPWL to muscle fatigue.

Results

Contraction time and fatigue sensation

All subjects performed the contraction for 40min in EF of 10%MVC. All subjects except subject I felt considerably fatigued. They, however, were not exhausted. Subject I felt a slight pain after 30min of the contraction. At 40%MVC of EF all subjects were exhausted at the end of the contraction. All subjects performed the contraction for 60min in KE of 3%MVC. Though occurrence of tremor, apparent fatigue sensation and slight pain were reported, the subjects did not need much effort to continue the contraction. Subject D reported that fatigue sensation was more severe in the middle of the contraction than at the end. In KE of 8%MVC most subjects felt pain before 10min in the contraction. Subject C continued it for 40min. Subject G ended it at

Fig. 1. Examples of the changes of amplitude of EMG(AEMG) during contractions. AEMG was represented in relative value to the start.

52min because of the end of the recording tape. Though the parts of the muscle in pain were altered during the contraction, they needed a greater effort to continue for 30min. In KE of 15%MVC muscle pain occurred before 2min in the contraction in most cases. Subject G endured for 8min 30sec. Most subjects ended the contraction at 5min. All subjects were almost exhausted at the end of the contraction. In KE of 8 and 15%MVC subject F ended the contraction because he felt sick. Since he felt severe fatigue at the end of the contractions, the results of subject F were used.

Change of EMG during the contraction

AEMG

AEMG increased during the contractions. The increasing manner was different between contraction levels. In low-level contractions, such as KE of 3%MVC and EF of 10%MVC, the increases in AEMG were larger in the early part of the contraction than in the latter half of the

contraction (Fig.1). In stronger contractions, such as KE of 15%MVC and EF of 40%MVC, the rate of increase in AEMG increased with the contraction. Especially in EF of 40%MVC AEMG increased steeply in the latter half of the contraction in 5 cases. In KE of 3%MVC abrupt increases and decreases of AEMG were observed in all subjects. The abrupt increase or decrease in AEMG of VL and VM occurred with the opposite changes in AEMG of RF. These abrupt changes show the migration of activity among the synergists.

The maximum ratio of AEMG relative to the start of the contraction was compared between contraction levels with Wilcoxon's test. The ratio was significantly larger in 8%MVC than in 3%MVC in KE. The ratio was larger in 8%MVC than in KE of 15%MVC in all subjects except subject G who ended the contraction because of exhaustion in 15%MVC and because of limitation of recording tape in 8%MVC.

Fig. 2. Examples of the changes of relative power in 8-22Hz (RPWL20) during contractions. RPWL20 was represented in differences from the start.

Fig. 3. Examples of the changes of AEMG(upper graph) and frequency spectrum(lower graph) during contractions. Frequency spectrum is shown in cumulative power from 8Hz relative to total power. The lines represent cumulative power of frequencies 22, 42, 62, 82, 122, 202 and 400Hz from the lowest to the highest(100%).

Frequency spectrum

RPWL20 increased during the contractions in most cases(Fig.2). A very large peak was seen between 12-20Hz in some EMG frequency spectra. The increase of RPWL20(slowing) was more apparent in the early part of the contraction than in the latter. The increase of relative power above 80Hz was often seen in the case where the slowing was not seen. Changes of power spectra are shown as cumulative power that was calculated by accumulating the relative power from low frequencies(Fig.3). The increase in relative power in high and low frequencies was seen simultaneously in a few cases.

The correlation coefficients of RPWL20 and RPW22-42 to the time were calculated(Table 1). In knee extension the positive significant cases were most in 8%MVC and least in 15%MVC. In the correlation of RPW22-42 negative cases were more in 3 and 8%MVC, and positive cases were more in KE of 15%MVC and EF of 40% MVC. The frequencies where relative power increases during contractions were higher in higher contraction levels than in lower ones. The maximum increases of relative power from the start of the contraction was compared between contraction levels with Wilcoxon's test. The increases of RPWL20 was larger in 3 and 8% MVC than in 15%MVC. The increases of RPW22-42 were larger in 15%MVC than in 3 and 8%MVC.

Difference between monopolar and bipolar EMG AEMG

Amplitudes of EMG were larger in monopolar EMG than in bipolar EMG except three cases (RF at 3 and 8% MVC and VM at 15%MVC). The ratio of AEMG of

Table 1. Number of significant correlations($p < 0.05$) of AEMG, RPWL20 and RPW22-42 to contracting time.

Cor. sign Lead method	AEMG		RPWL20		RPW22-42	
	Positive Mon. Bi.	Negative Mon. Bi.	Positive Mon. Bi.	Negative Mon. Bi.	Positive Mon. Bi.	Negative Mon. Bi.
Elbow flexion						
10%MVC BB	10	10	•	•	6	9
40%MVC BB	9	9	•	•	9	8
Knee extension						
3%MVC VL	6	6	1	1	7	7
RF	5	3	•	1	7	7
VM	6	6	1	1	7	7
8%MVC VL	7	7	•	•	7	6
RF	7	7	•	•	7	7
VM	7	7	•	•	7	6
15%MVC VL	7	7	•	•	7	3
RF	7	7	•	•	7	3
VM	7	7	•	•	7	6

Cor: correlation, Cnt: contraction,
Mon: monopole, Bi: bipole,
'•' means 0

bipolar EMG relative to that of monopolar EMG was smaller in BB than in VL, RF and VM(Table 2). This ratio was compared between contraction levels with Wilcoxon's test. The ratio was larger in 3%MVC than in 15%MVC in VL and RF. Changes of the ratio during the contraction were examined by correlation coefficients between the ratio and contracting time(Table 2). Though the correlations were negative in all subjects in 2 cases of KE, a consistent tendency was not observed as a whole. The pattern changes of EMG during the contraction was not greatly different between monopolar and bipolar EMG. The changes of AEMG, however, tended to be more abrupt in bipolar EMG than in monopolar EMG. In EF of 40%MVC increases in AEMG changed to be steeper near the end of the contraction. The change near the end of the contraction was often larger in bipolar EMG than in monopolar EMG. The correlation coefficients between AEMG and contracting time were compared between monopolar and bipolar EMG. Though the correlation coefficients tended to be larger in monopolar EMG than in bipolar EMG, this difference was not consistent in all cases.

Frequency spectrum

The difference of the changing manner of RPWL20 between monopolar and bipolar EMG was most marked in 8%MVC of VL(Fig.2). RPWL20 of monopolar EMG increased during the contraction in all subjects and muscles in KE of 8%MVC. RPWL20 of bipolar EMG in the same condition also increased in most cases. But in some cases RPWL20 of bipolar EMG did not increase but even decreased. In EF of 10 and 40%MVC and KE of 15%MVC the increases of RPWL20 during contraction were sometimes obscure in bipolar EMG while the increases were obvious in monopolar EMG. The correla-

Table 2. Rate of absolute amplitude of bipolar EMG to that of monopolar EMG, and number of significant correlation ($p < 0.05$) between the rate and contracting time.

	Rate(%)		Correlation	
	Bi	Mon	signifcant no	
	Mean	SD	Posi.	Neg.
Elbow flexion				
10%MVC BB	16	4	6	3
40%MVC BB	16	4	2	2
Knee extension				
3%MVC VL	73	13	4	1
RF	82	15	1	5
VM	61	20	2	4
8%MVC VL	70	11	1	4
RF	68	25	.	7
VM	58	14	4	3
15%MVC VL	64	12	.	4
RF	66	19	1	5
VM	63	25	.	7

Posi: positive, Neg: negative

SD: standard deviation

'.' means 0

tion coefficients between RPWL20 and contracting time were significant in all positive cases in monopolar EMG. In bipolar EMG these correlation coefficients were significant in all subjects only in KE of 3%MVC. These correlation coefficients were usually larger in monopolar EMG than in bipolar EMG. Mean and SD of these correlation coefficients calculated for each subject were shown in Table 3. In bipolar EMG the differences between cases were large. The correlation coefficients for RPW22-42 were often larger in bipolar EMG than in monopolar EMG.

When the increase of relative power in high frequencies(quickening) was seen in bipolar EMG, this quickening was also often observed partially in monopolar EMG. But in monopolar EMG, the quickening seldom canceled the EMG slowing out, since the increases in relative power in lower frequencies were large.

Discussion

Fatigue in KE of 3%MVC

MVC in this study would be larger than those in other studies(Sato and Ohashi, 1988; Søjgaard et al., 1988), since the peak value of momentary force exertion was used as MVC. Most subjects needed great efforts to continue KE of 8%MVC for 30min. Sato and Ohashi(1988) estimated that KE can be endured for 60min at 9.4%MVC. This difference in fatigability would be attributed to the measurement of MVC.

Apparent fatigue sensation and tremor were reported during KE of 3%MVC. Subjective fatigue levels often fluctuated during 3%MVC contractions. One subject

Table 3. Mean and standard deviation of correlation coefficients between RPW and contracting time calculated from the result of each subject

Lead method	RPWL20				RPW 22-42			
	Mean		SD		Mean		SD	
	Mon.	Bi.	Mon.	Bi.	Mon.	Bi.	Mon.	Bi.
Elbow flexion								
10%MVC BB	.847	.551	.107	.270	.405	.480	.349	.325
40%MVC BB	.887	.720	.071	.336	.888	.772	.069	.324
Knee extension								
3%MVC VL	.787	.588	.100	.229	-.136	.200	.503	.315
RF	.690	.713	.233	.086	-.213	.160	.399	.417
VM	.713	.603	.263	.235	-.542	.206	.210	.402
8%MVC VL	.942	.531	.040	.330	-.654	-.036	.329	.487
RF	.937	.794	.030	.197	-.574	-.214	.327	.549
VM	.945	.694	.016	.597	-.802	-.283	.124	.693
15%MVC VL	.827	.399	.172	.479	.450	.283	.159	.659
RF	.857	.684	.113	.262	.470	.481	.286	.193
VM	.823	.642	.083	.238	.351	.180	.251	.422

Mon: monopole , Bi: bipole , SD: standard deviation

reported that fatigue sensation was more severe in the middle of the contraction than at the end. AEMG often decreased during one third of contractions from the end in 3%MVC. The adaptation of muscle activities to the static contractions would lighten or keep fatigue level. The fluctuation of the fatigue state must be the main cause of the ambiguous changes of AEMG and RPWL20 during 3%MVC contraction.

The slowing of EMG during low-level contraction

The slowing of EMG is small during low-level contractions, though the contractions cause fatigue(Ohashi, 1993). While the quickening of EMG is even seen during low-level contractions(Hagberg, 1981; Malmqvist et al., 1981; Hägg, 1991), the slowing is seen in the contraction even below 10%MVC(Kogi and Hakamada, 1962; Fallentin et al., 1985; Ohashi et al., 1987). The increases in relative power in high and low frequencies were seen simultaneously in a few cases. There are two factors which influence EMG spectra. One factor increases low frequency component and the other increases high frequency component.

In the present experiment the increases of RPWL20 during contractions were sometimes obscure in bipolar EMG but obvious in monopolar EMG. The followings points are noted. Motor unit activities change to the mode in which EMG shows slowing even during low-level contractions. But bipolar EMG cannot detect the changes of the activities sufficiently, unlike monopolar EMG. The EMG slowing was sometimes seen in monopolar EMG but not in bipolar EMG. Though bipolar EMG also detects the changes of the activity, the ability of detection is not stable in bipolar EMG. When the quickening was seen in bipolar EMG, the influence of this quickening was often also seen in monopolar EMG.

Fig. 4. Two dimensional model for the lead amplitude of action potential on surface electrodes.

SER: surface electrode, l : distance between electrodes, a : length of electrode, O : source of action potential, p : horizontal distance between action potential and electrode, d : vertical distance (depth) between action potential and electrode.

The cause of the quickening would be the recruitment of motor units whose muscle fiber has fast conduction velocity of action potential (Arendt-Nielsen et al., 1989; Krogh-Lund and Jørgensen, 1991; Ohashi, 1993). Since the change of the conduction velocity influences each wave of surface EMG, this quickening was seen in both monopolar and bipolar EMG. But in monopolar EMG the large slowing concealed the quickening factor.

Difference of amplitude of EMG between monopolar and bipolar EMG

A simple model (Fig. 4) is used to consider the difference of amplitude of EMG between monopolar and bipolar EMG. This model is the same with the 2-dimensional model of Lynn et al. (1978). The potential at the electrode is proportional to the inverse of the square of the distance between the electrode and the source of the action potential (Suzuki et al., 1986). Amplitude of the action potential is V and its position from the electrode is p in the horizontal and d in the depth. The lead potential $V_s(p)$ is calculated as integration of the potential on the electrode of which length is a .

$$V_s(p) = \int_{p-a/2}^{p+a/2} \frac{V}{(x^2 + d^2)} dx \quad \text{eq.1}$$

$$= (V/d) [\tan^{-1}(x/d)]_{p-a/2}^{p+a/2}$$

This equation is applied to monopolar lead. Lead potential V_s is integrated during the conduction of action potential for w on both sides from the center of the electrode. The integrated value I_{mon} is calculated as follows.

$$I_{\text{mon}} = \int_{-w}^w V_s(p) dp \quad \text{eq.2}$$

$$= (V/d) \int_{-w}^w [\tan^{-1}(x/d)]_{p-a/2}^{p+a/2} dp$$

$$= 2 \cdot V \{ A \cdot \tan^{-1}(A) - B \cdot \tan^{-1}(B) + \log(\cos(\tan^{-1}(A)) / \cos(\tan^{-1}(B))) \}$$

$$A = (w + a/2)/d, B = (w - a/2)/d$$

The potential in bipolar lead $V_{\text{sbi}}(p)$ is calculated as the difference in potential of paired electrodes that are

Fig. 5. Relations between amplitude of surface EMG and the depth of the origin of action potential. Thick lines represented the lead condition of quadriceps in this experiment and thin lines represented the condition of biceps brachii.

spaced l apart as follows.

$$V_{\text{sbi}}(p) = V_s(p + l/2) - V_s(p - l/2) \quad \text{eq.3}$$

The integrated value I_{bi} for bipolar lead is calculated by integrating absolute value of $V_{\text{sbi}}(|V_{\text{sbi}}|)$ in the same way of getting I_{mon} . Integrated area of lead potential V_s is length w on both sides from the middle of the electrode pair. The differences in effects of attenuation at skin, subcutaneous fat and muscle are not considered since the effects of the difference are common for both lead methods. I_{mon} and I_{bi} were calculated in the conditions similar to those in the present experiment; the biceps brachii (0.5 mm of electrode length and 5 mm of electrode interval) and the quadriceps femoris (10 mm of electrode length and 35 mm of electrode interval). The w in eq. 2 and for bipolar lead was set at 40 mm. These integrated potentials are shown against the depth of the source of action potential as the percentage to I_{bi} in the condition of 10 mm of electrode length, 35 mm of electrode interval and 10 mm of the depth of action potential (Fig 5). The attenuation ratio of I with the increase in the depth is larger in bipolar EMG than in monopolar EMG. The I from deep parts of a muscle relative to that from shallow parts is smaller in bipolar EMG than in

Table 4. The depth(mm) where the I attenuates to 1/2, 1/4 and 1/8 of the reference potential and the depths to which the sum of I from the reference position are 50%,75% and 87.5% of the total sum.

Muscle	Lead method	Attenuated I			Percentile of sum of I		
		1/2	1/4	1/8	50%	75%	87.5%
Biceps brachii	Bi.	3.8	5.4	7.7	4.6	7.9	12.3
	Mon.	4.8	8.9	15.8	7.8	15.3	22.3
Quadriceps femoris	Bi.	8.3	12.8	18.5	9.1	13.6	17.6
	Mon.	9.3	16.4	27.0	10.3	15.6	19.5

Mon: monopole, Bi: bipole

monopolar EMG. The thickness of skin-fat layer was reported 5mm over the biceps brachii (Clamann, 1970) and 10mm(male) in the femur (Sato, 1988). Since these values seem to be measured with a caliper, the real thickness of skin-fat layer is considered as half of these values. The nearest point of muscle fibers to the electrode, that is just under the skin-fat layer, is set as the reference position. The I at the reference position is set as the reference potential. The depths where the I attenuates to 1/2, 1/4 and 1/8 of the reference potential were calculated(Table 4). I for a constant V was summed up from reference position to the deepest position of the muscles at 0.1 mm interval. The deepest positions are 35mm in biceps brachii (Clamann, 1970) and 25mm in quadriceps femoris(measured the figure of Kaneko, 1973). The depths to which the sum of I from the reference position are 50%,75% and 87.5% of the total sum were obtained(Table 4). The depths where I attenuate 1/4 are different less than 5mm between lead methods. The depths for the same accumulation of I are largely different between lead methods in the condition of biceps brachii but are not largely different in the condition of quadriceps femoris.

The values in Fig.5 were calculated as constant potentials. The tissue between the source and detecting electrode has filtering effects on the waves of action potential. The attenuation at 20, 50 and 100Hz relative to those at 5Hz was measured from the Figure 6 of De Luca(1985). The attenuation at 20Hz is 3, 4 and 7dB at 5, 10 and 20mm respectively of the distance between an electrode and an active muscle fiber. Those at 50Hz are 7, 10 and 17dB. Those at 100Hz are 11, 19 and 34dB.

Ratio of I_{mon} to I_{bi} increases with the increase in the depth of action potential. When the interval of electrodes is large in bipolar lead, I_{bi} is larger than I_{mon} near the surface of muscle and this relation reverses in the deep part of a muscle. Deep motor units tend to have a low recruitment threshold (Clamann, 1970; Kossev et al., 1991). Since contraction levels were mainly low in this study, the distribution of active motor units would be biased to the deep part of the muscle. AEMG was smaller in bipolar EMG than in monopolar EMG. This difference in AEMG between lead methods

was larger in the biceps brachii than in the quadriceps femoris, and the difference between I_{mon} and I_{bi} is larger in the condition of the narrower interval of electrodes. Though the led potential of an action potential decreases with the increase in the distance between a muscle fiber and an electrode, the number of muscle fibers increases with the increase in the distance. The relation between I_{mon} and I_{bi} explains partially the difference in AEMG between lead methods. An increase in contraction levels accompanies an increase in ratio of active motor units near the surface of muscle. This increase in the ratio is thought to decrease the difference in AEMG between lead methods. But the difference in AEMG even increased with the increase in contraction levels. The synchronization between motor units is higher in higher contraction levels (Person and Kudina, 1968). Simultaneous discharges at different muscle fibers usually diminish recording potential in differential (bipolar) lead. This will be one of the main causes of smaller AEMG in bipolar EMG than in monopolar EMG.

The increase in the attenuation of potential with the increase in depth is larger in bipolar lead than in monopolar lead. This means that far action potential from the electrodes influences EMG more in monopolar lead than in bipolar lead. In other words, monopolar EMG reflects a larger sphere of motor unit activities than bipolar EMG. The changes of AEMG tended to be more abrupt in bipolar EMG than in monopolar EMG. This abrupt change of AEMG was derived from the smaller pickup sphere in bipolar EMG. For the high frequency components this difference of pickup sphere, however, is not very large since the attenuation at high frequency components from deep parts of a muscle is equally large for both the lead methods.

I thought that monopolar EMG would be easier to show the slowing than bipolar EMG, because of the larger pickup sphere of monopolar EMG. The slowing of surface EMG was shown to be more regular in monopolar EMG than in bipolar EMG. The differences of the pickup spheres evaluated by the simple model between lead methods were large in the condition of biceps brachii but small in the condition of quadriceps femoris. But the difference of the EMG slowing between lead methods was sometimes very large in quadriceps femoris. The difference of the slowing between lead methods is not explained completely by the difference of the pickup sphere.

The relation between contraction levels and the EMG slowing

The slowing of EMG was thought to be smaller in lower contraction levels than in higher contraction levels(Krogh-Lund and Jørgensen, 1991; Ohashi, 1993). In this experiment, however, the correlation coefficients between RPWL20 of bipolar EMG and contracting time were significant positively in all subjects in KE of 3%

MVC, though they were not significant in some cases in 8 and 15%MVC. Furthermore, fatigue sensation at the end of the contractions was higher in 8 and 15%MVC than in 3%MVC. One reason that RPWL20 increased more regularly in lower contraction levels than in higher ones is the effect of the relation between changing frequency band and contraction levels. The frequencies where relative power increases with a fatiguing contraction were higher in higher level contractions than in lower level ones. This tendency was seen in other experiments (Ohashi, 1993). This relation, however, could not explain the regular increase in RPWL20 in 3%MVC in bipolar EMG since the increases in RPWL20 in 8%MVC were large in monopolar EMG.

Large increases of AEMG in 8 and 15%MVC indicate the large recruitment of motor units, which have muscle fibers with faster conduction velocity of action potential. Larger recruitment increases high frequency components more. Motor units recruited below 15%MVC are non-fatigable and those recruited above 25%MVC are highly fatigable (Stephens and Usherwood, 1977), though this result was obtained in the first dorsal interosseous of the hand. This shows that the changes in the motor units type recruited are larger above 15%MVC than below 15%MVC. In very low contraction levels the influence of the motor units recruitment on frequency spectrum of EMG would be small because of small changes of motor unit types. Steep increase in AEMG, however, often accompanied steep decrease in RPWL20 in KE of 3%MVC. These effects of the recruitment of motor units would not be the only cause of the regular increase in RPWL20 in 3%MVC in bipolar EMG.

If the distance between an electrode and a muscle fiber changes, the detection performance of the electrode changes. When the detection performance of an electrode of paired electrodes in bipolar lead is largely lower than that of the other electrode, the bipolar EMG resembles monopolar EMG. The ratios of positive and negative parts of bipolar lead potential were calculated with the model (Fig.4) in the electrode condition of the quadriceps femoris. When the detection performance of an electrode was 80% of the other, the ratios of the action potentials at the depth of 5, 15 and 25mm are .79, .69 and .61 respectively. Muscle fibers with low recruitment thresholds tend to be distributed in the deep parts of a muscle (Clamann, 1970; Kossev et al., 1991). Though the bipolar EMG for the deep parts of a muscle would resemble the monopolar EMG at the same position, the effect of the depth is not large.

The three factors mentioned above would contribute to the regular increase in RPWL20 of bipolar EMG in 3%MVC. But the degree of the contributions cannot be confirmed.

Acknowledgment. The author wishes to thank Profes-

sor H.Sato for his advice, and the subjects for their assistance in the experiments.

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Received : August 1, 1994

Accepted : January 27, 1995

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