# BIOMECHANICAL AND ELECTROMYOGRAPHIC ANALYSES OF MUSCLE STRENGTHENING

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## Declaration

The author declares that the content of this thesis is presented to the best of the author's knowledge and it is original, except as acknowledged in the text. The concept and framework of this study comes solely from the discussion between the author, Dr Kevin S.C. Kwong and Professor Christina W.Y. Hui-Chan. The data collected in this project have not been submitted for a degree at this or any other university.

Rainbow Ka-Yee Law, October 2001

## Acknowledgements

I wish to express my genuine appreciation to the continuous and patient guidance and advice offered by Dr Kevin SC Kwong and Professor Christina WY Hui-Chan. I would also give my hearty thanks to Professor Joseph Mizrahi who has given the invaluable advice on the research design, technical problems and data analysis during his stay in Hong Kong. Dr Oron Leven gave his expert advice on technical issues. Dr Gabriel Ng gave his invaluable advice on my research proposal and seminars. The laboratory technicians Mr Man and Mr Kan offered their efficient assistance during data collection. A group of physiotherapy students volunteered to be the test subjects. To all of these people, I would like to express my very sincere thanks.

Special thank is given to Alice Ho Miu Ling Nethersole Hospital. Part of the data collection was done in the Physiotherapy Department of the Hospital. The Hospital also kindly offered the equipment on diagnostic electromyography for data collection. I would like to express my heart-felt thanks to Dr Alex CP Chow for his kind assistance in the data collection using intra-muscular wire electromyography.

## Abstract

Eleven male subjects completed a six-week strengthening program on their right quadriceps muscles, which consisted of 30 maximal voluntary contractions (MVC) per day, 3 days a week, for 6 weeks. The maximal isometric strength of the quadriceps muscles was measured by a computerized dynamometer (Cybex "Norm"), with the hip and knee at about 60° flexion. Surface electromyography (EMG) of the rectus femoris muscle was acquired at MVC and at two pre-set submaximal torques of 100 Nm and 150 Nm. The EMG signal was band-pass filtered at 5 to 350 Hz, and sampled at 5 kHz. EMG was measured initially before the strengthening and subsequently at the 2$^{nd}$, 4$^{th}$ and 6$^{th}$ weeks of the strengthening program. Integrated EMG (IEMG) and root mean square (RMS) voltage corresponding to MVC and torque levels at 100 Nm and 150 Nm were analyzed. The EMG data at the pre-set submaximal contractions were normalized twice to facilitate comparison between sessions. Spectral analysis was performed by the use of Fast Fourier Transform and the median frequency (MF) was obtained from the power density spectrum.

There was a significant ($p = 0.001$) increase in the strength of the right quadriceps muscles (mean is 22%). However there was no significant change in the EMG at MVC. Therefore the role of neural adaptation in the process of muscle strengthening was not substantiated in this study.

Both IEMG and RMS voltage at 150 Nm, after double normalization, had significantly decreased after the strengthening program ($p = 0.024$). Moreover, there was a right shift of the torque-EMG relationship. It implied that the muscle became more efficient after the six-week strengthening program. It also showed that the decrease in EMG under a reasonably strong submaximal contraction at pre-determined torque level in the course of strengthening was an indication of improvement in the quadriceps' mechanical output. This result can be used to develop a test protocol that uses submaximal torque level to monitor the improvement of muscle strength in the quadriceps during strengthening. This protocol is safer for persons undergoing rehabilitation when compared with the commonly used test protocol using percentage of MVC because the test of MVC may be difficult or dangerous for persons with lower limb injury. This test protocol has the potential to be applied in rehabilitation especially in monitoring the progress of muscles whose strength cannot be easily measured by mechanical means. Furthermore, the double normalization technique can be applied in future studies if the test protocol with pre-determined torque level is used.

# Chapter 1 General Introduction

One of the main objectives of physical rehabilitation is to restore muscle strength. Clinical test of muscle strength is usually performed by either mechanical means or manual technique. However, these methods have limitations. Firstly, human muscle force in voluntary movement is usually contributed by a group of muscles rather than a single one. The performance of an individual muscle cannot be reflected just by biomechanical force measurement. Very often, there are situations when the clinician may want to know whether there is a progress in the strength of an individual muscle. These include testing for muscle strength after surgical procedures such as nerve repair and tendon transfer. It also applies to conditions like chrondromalacia patellae, for which reassessment of muscle strength is necessary after selective muscle strengthening of the vastus medialis muscle (especially the oblique portion). Electromyography (EMG) can supplement the mechanical means in the assessment of muscle strength because it directly reflects the activity within an individual muscle. Therefore the first objective of this study is to establish an assessment technique or test protocol of muscle strength by close examination of the changes in surface EMG during muscle strengthening.

The second objective of this study to investigate the role of neural adaptation in the process of muscle strengthening through the study of surface EMG. A better understanding of the mechanisms that lead to an increase in muscle strength can facilitate

the development of a more effective strengthening program. In summary, muscle strength is determined by three factors (Moritani et al 1993). The first one is neural factor. The second factor is the quantity of muscle fiber, ie. cross-sectional area (CSA). The third one is the quality of muscle fiber, eg. musculo-tendinous structure, type of muscle fiber (Moritani et al 1993) and sarcolemmal morphology (Kamen et al 1996). The muscle length and velocity of muscle contraction would certainly also influence the muscle strength (Kraemer et al 1998). Surface EMG was widely used in an effort to understand the neuromuscular adaptations accompanying motor learning and exercise (Kamen et al 1996). Numerous previous studies (Narici et al 1996, Guimaraes et al 1995, Moritani 1993, Garfinkel et al 1992, Sale 1988 & Howald 1982) attempted to investigate the physiological basis of the change in muscle strength by the means of surface EMG measurement. EMG change was often expected after strengthening exercise if neural adaptation exists (Cannon et al 1987). Section 2.2 will give a detailed account of these physiological mechanisms.

Surface EMG is investigated in this study. There are several advantages of using surface EMG to monitor muscle performance. It is non-invasive and real-time. Moreover, EMG acquired with advanced technology allows the study of individual muscle activity with high fidelity. The design of electrodes minimizes artifacts due to relative movements. The high input impedance of the pre-amplifier enables the recording of the myoelectric

signals without a very demanding skin preparation. Furthermore, force measurement device cannot detect very weak muscle contraction whereas EMG can. Besides, the decline of the mean or median frequency of the EMG power spectrum was directly related to increased subjective sensation of perceived exertion during a sustained muscle contraction performed at moderate levels (Basmajian et al 1985). Leung and associates (1996) also found that mean power frequency (MPF) and median frequency (MF) were more sensitive fatigue indicators than mechanical work output. It was also demonstrated that change in MF was readily observable in the absence of any decline in the muscle's mechanical output (Mannion et al, 1996). EMG required only submaximal contraction for a shorter period of time to measure muscle fatigue (Ng et al 1996). Besides, surface EMG was preferred to needle EMG when studying muscle strength because it picked up electrical signals from a larger volume of muscle fibers. Surface EMG seemed o be the most useful technique available to monitor motor unit recruitment by the measurement of MF. It was because of the larger signal collecting area than that available when using intramuscular wire electrodes (Bernardi et al 1995). All these advantages justify more in-depth study of the surface EMG in the context of rehabilitation.

This study tried to develop a protocol to monitor the progress of muscle strength in rehabilitation by studying the change of surface EMG in both maximal and submaximal contractions. Testing protocol with maximal voluntary contraction (MVC) or percentage

of MVC was commonly used in research. Instead, tests with pre-set torque levels at submaximal isometric contractions would be employed in this study. The change in EMG in the rectus femoris muscle at both maximal and submaximal isometric contractions would be investigated during a six-week strengthening program. Rectus femoris muscle was chosen because it is a superficial muscle and is easy to be monitored by surface EMG.

# Chapter 2 Literature Review

## 2.1 Introduction

Electromyography (EMG) has been extensively studied and applied in many clinical areas. These included electrodiagnosis using needle EMG, qualitative and quantitative analysis using surface or intra-muscular wire EMG. EMG analyses, which were employed in the area of applied physiology, were usually divided into three areas: surface EMG analyses; intramuscular EMG analyses; and evoked potential analyses (Moritani et al 1998). The main theme of the present study was to find out the relationship between the changes in surface EMG and the improvement in the muscle strength of the quadriceps femoris muscle after a strengthening program in normal subjects. Quantitative analysis of surface EMG was adopted. A comprehensive literature review on several aspects related to muscle strengthening and surface EMG would be presented in this chapter.

## 2.2 Physiological Mechanisms of Muscle Strengthening

Muscle strengthening is one of the most common treatment modalities in rehabilitation. Scientists are keen on finding out the exact mechanisms leading to an increase in muscle strength so that a more effective muscle strengthening program can be designed for both athletes or patients.

There were different findings in the research concerning the physiological mechanisms that lead to the increase in muscle strength after training. Table 2.1 summarizes these opinions. Adaptation in the neural system can lead to an increase in muscle strength after training. Muscle contraction can be improved through coordination of motor unit activation. When the neural input to the agonist and synergist muscles is increased, and there is greater inhibition of the antagonists as a consequence of muscle strengthening, the muscle strength will increase (Hamill 1995). On the contrary, some hold different opinions. It was found that there was no significant change in the integrated EMG (IEMG) of biceps femoris muscle during a six-month strengthening of isometric knee extension exercise in middle-aged men and women (Hakkinen et al 1998). It implied that there was no sign of increased inhibition of the antagonist (fig. 2.1). However, significant decrease in the coactivation of antagonist did happen in the elderly group. Therefore the effect of co-contraction of antagonist in the generation of torque during isometric knee extension would also be investigated in this study. The result is reported in section 3.4.3.

Table 2.1. Summary of research results concerning the physiological mechanisms of increase in muscle strength

| Author | Methodology (major differences are highlighted) | | Results | | Conclusion |
|---|---|---|---|---|---|
| | Subjects | Exercise program | Change in Force/ Torque | Change in surface EMG/ muscle-tendon structure | |
| Garfinkel et al (1992) | 8 females with 7 females control Age: 22 - 24 | 8 weeks of isometric unilateral knee extension ex. 3 x 10 MVC/ day, 3/wk | Force at MVC increased by 28 %. | CSA increased 14.6 %. Maximal IEMG of vastus lateralis muscle was unchanged. | There was no evidence of neural adaptation. The synthesis of additional contractile proteins explained the increase in muscle strength. |
| Weir et al (1994) | 3 males, 4 females with 3 males and 3 females controls Age: 22 – 24 | 6 weeks of isometric unilateral knee extension ex. 80 % of maximal isometric torque | There was angle-specific increase in torque (23 %). | There was significant increase in the maximal IEMG at almost all angles. | It supported the view that there was evidence of neural adaptation. |
| Hakkinen et al (1995) | Middle-aged (n=18) and elderly (n=21) males and females | 12 weeks heavy strength training with explosive types of dynamic exercises 2/wk | There was significant increase in CSA ($p < 0.05$–$0.001$) and in maximal force ($p < 0.01$-$0.001$). | There was significant increase in maximal IEMG especially during the first 8 weeks ($p < 0.05$-$0.001$). | Both adaptation in the nervous system and muscle hypertrophy accounted for the increase in muscle force. |
| Narici et al (1996) | 7 males Mean age: 29 | 6 months strength training of quadriceps 6 sets, 8 repetitions, use 80 % 1 RM Exercise on alternate days | Maximal isometric torque increased by 25 %. CSA increased by 13 – 19 %. | There was a sizeable increase in maximal IEMG at 2 months. There was no change in maximal IEMG at 2 to 6 months. | It supported neural adaptation as a mechanism for increase in muscle strength especially before 2 months. A constant contribution of hypertrophy towards the increase in muscle strength after the second month of training. |

| Author | Methodology (major differences are highlighted) | | Results | | Conclusion |
|---|---|---|---|---|---|
| | Subjects | Exercise program | Change in Force/ Torque | Change in surface EMG/ muscle-tendon structure | |
| Hakkinen et al (1998) | Middle aged (n = 21) and elderly (n = 21) males and females | 6 months of heavy resistance and explosive type of exercise Isometric and dynamic leg extension | There was 34 – 66 % increase in isometric leg extension strength and significant hypertrophy but the increase in CSA was minor when compared with the increase in muscle strength. | All groups showed large increase in the maximal IEMG of vastus lateralis and vastus medialis. | It suggested that the contributing role of the nervous system for strength development might have been more important than that of muscle hypertrophy. |
| Rabita et al (2000) | 9 subjects in training group<br>7 controls<br>Mean age: 24 | 4 weeks of bilateral isometric knee extension at 80 % MVC<br>5 sets, 5 repetition<br>3/wk | Torque at MVC increased by 38.7 %. | Only EMG (rms) of rectus femoris showed significant differences in maximal activation. Vastus lateralis and vastus medialis did not present any significant changes in maximal activation. | Parts of the quadriceps muscle tested present different adaptation capacities and demonstrate inter-individual variability in the strategies used to enhance muscle strength. |

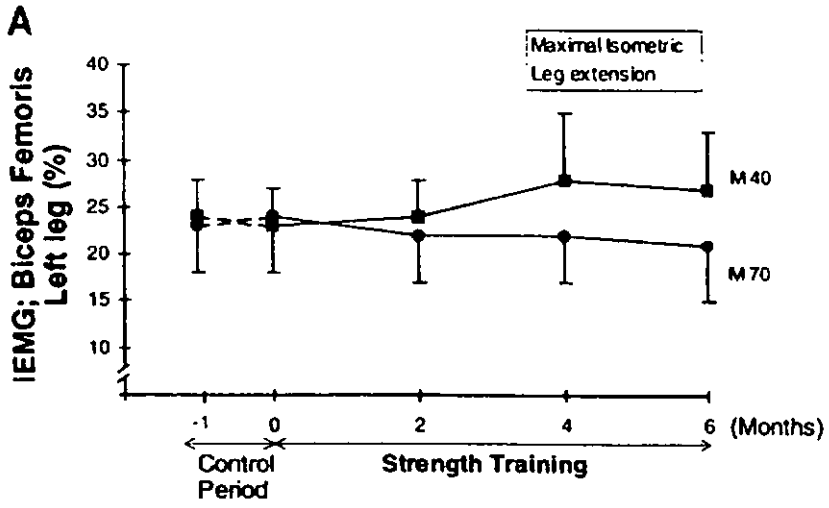



Figure 2.1. IEMG (in relation to the IEMG during maximal isometric voluntary contraction) of biceps femoris muscle during a six-month strengthening program of the quadriceps. There was no sign of inhibition of the antagonist in terms of neural activation in the middle age male subjects (M40) though significant decrease in the coactivation of antagonist did happen in the elderly (M70). (From Hakkinen et al 1998)

Neural adaptation could also be manifested by the increase in the amplitude of surface electromyography (EMG) at maximal voluntary contraction (MVC) with the increase in muscle strength (Weir et al 1994, Hakkinen et al 1995, 1998, Narici et at 1996, Molina et al 1997). This finding had been interpreted as an increase in central neural drive that enhanced maximal force production (Cannon et al 1987). The change in surface EMG with strengthening was because of change in several factors namely motor unit recruitment, excitation frequency, individual muscle fiber potential, degree of motor unit discharge synchronization and muscle fatigue. These factors could be group collectively into the "neural factors" (Moritani et al 1993). It had long been suggested that neural factors included the change in motor unit recruitment or firing rate (De Luca 1997) although recent study (Kamen et al 2001) found that resistance training seems to have little influence on submaximal motor unit discharge rate. It was suggested that a centrally located neural adaptation could explain the phenomenon of cross-training effect (Weir et al 1994). Hakkinen and associates (1998) opined that the increase in the cross-sectional area of the quadriceps muscles was minor compared with the increase in the maximum strength (34 – 66 %) after a six-month strengthening program. It suggested that the contributing role of the nervous system for strength development may have been more important that that of muscle hypertrophy.

The second physiological explanation to the increase in muscle strength was a change in the muscle-tendon structures. These included muscle hypertrophy and the change in sarcolemmal physiology (Kamen et al 1996). Garfinkel and associates (1992) refuted any contribution of neural element in the process of strengthening. Subjects were recruited for a strengthening program that consisted of an isometric resistance training of the knee extensors for eight weeks. The amplitude of the maximal IEMG of vastus lateralis was found unchanged in the trained leg whereas both the torque of knee extension and the cross-sectional area of the vastus lateralis did increase significantly. It was concluded that muscle hypertrophy was the reason for the increase in the torque produced, and that the increase in force-generating capacity of the muscle was due to the syntheses of additional contractile proteins. The extremely well-motivated subjects would usually display full motor unit activation even before training.

Besides muscle hypertrophy, Kamen and Associates (1996) suggested that change in sarcolemmal morphology might occur with intensive long-term exercise or activity. Sarcolemmal permeability to [Na+] or [K+] might be changed. There was evidence that the change in sarcolemmal permeability contributed to a change in muscle fiber conduction velocity. This was reflected in the spectral shift towards lower frequency of the EMG power spectrum during muscle fatigue. It was also found that conduction velocity increased with muscular force level. Therefore, a change in the sarcolemmal

permeability during muscle strengthening might be reflected in the power density spectrum of EMG because of the change in muscle fiber conduction velocity.

Several researchers have discussed about the timing of neural adaptation and muscle hypertrophy with muscle strengthening. Sale's study (1988) demonstrated significant gains in the muscle strength after approximately four weeks of strengthening. This gain in strength was not due to an increase in muscle fiber size, but rather a learning effect in which neural adaptation has occurred. Weir and associates (1994) found that the maximum IEMG of vastus lateralis at various joint angles increased after a six-week isometric strengthening program. Hakkinen (1995) also showed that the integrated EMG (IEMG) of vastus lateralis, vastus medialis and rectus femoris increased during the first eight weeks of strengthening program even in the elderly. There was also a significant increase in the cross-sectional area of the quadriceps femoris muscle appeared during a longer period (12-week) of strengthening program. It was concluded that progressive heavy strength training combined with explosive types of exercises led to considerable increase in the maximal muscle strength. This was accompanied by adaptations in the nervous system as well as by muscle hypertrophy. These findings were consistent with Hamill's (1995) suggestion that neural adaptation leveled off after about four to five weeks of training, and the increase in muscle strength beyond this point was usually due to muscle hypertrophy. It was further supported by research evidence (Narici et al 1996)

that there was a sizeable increase in the maximal IEMG of the vastus lateralis muscle after two months of strength training. It was also found that the torque increased at a much higher rate than the cross-sectional area (CSA) during the first two months and neural factors were the reason for the gain in the muscle torque. Integrated EMG at maximal voluntary contraction (IEMG max) of the vastus lateralis muscle in the second to the sixth months was not statistically different from that before strengthening. Moreover, there was a significant muscle hypertrophy. This implied that muscle hypertrophy was the main contributing factor towards the increase in muscle strength after the second month of strengthening. Phillips (2000) also commented that though the exact time-course for muscle fiber hypertrophy was not well-documented but it appeared that at least six to seven weeks of regular resistive training was required before the increase in the cross-sectional area was deemed significant.

Several other anatomical and physiological factors also contribute to muscle strength and endurance. These are muscle size, muscle composition, recruitment pattern, motivation, cardiopulmonary condition, active and passive components of the musculo-tendinous structure, muscle length and contraction speed (Hamill et al 1995, Astrand et el 1986).

It is recognized that the change in EMG during muscle strengthening and the physiological mechanism of muscle strengthening are not conclusive and are interesting

to pursue. A recent paper (Rabita et al 2000) studied the changes of EMG of the vastus medialis, rectus femoris and vastus lateralis during a four-week strengthening program. Only the EMG of rectus femoris muscle showed significant differences in maximal activation. Vastus lateralis and medialis did not present any significant changes in maximal activation. It was concluded those different parts of the quadriceps muscle presented different adaptation capacity and demonstrated inter-individual variability in the strategies used to enhance muscle strength. It is the purpose of this study to investigate the change of surface EMG during muscle strengthening. It is hoped that the results obtained in this study can enrich the evidence in the area of neural adaptation.

### 2.3 Electromyographic Changes During Immobilization

Changes in electromyography (EMG) after strengthening and the related physiological explanations are reported in the last section. It is also interesting to know about the EMG changes brought about by immobilization. Duchateau and associates (1991) mentioned that the observed reduction of contraction force after immobilization might be due to the changes of neural drive or changes of muscle membrane ionic processes, or both, besides muscle atrophy. The mechanical tension recorded during a maximal voluntary contraction was compared with electrically-evoked contractions at 100 Hz after 6 weeks of immobilization of the adductor pollicis muscle. It was found that there was a greater decrease of the torque during a maximal voluntary contraction. It

suggested that the neural drive to the muscle was changed after a period of reduced use of muscle in human. Duchateau (1991) also demonstrated that the force and integrated EMG (IEMG) were drastically reduced during a maximal voluntary contraction (by 55% and 45% respectively) after six weeks of immobilization of the adductor pollicis muscle. These findings suggest a possibility that IEMG is sensitive to change in condition of the muscle no matter it is an improvement or deterioration in muscle strength.

## 2.4 Electromyographic Parameters

There are numerous electromyographic (EMG) parameters. These include semi-quantitative parameters such as magnitude of raw signals, number of zero-crossings, number and magnitude of spikes. Quantitative parameters are voltage, wave rectification and envelope detection, integration, root mean square and force calculated from the EMG signal (Kumar et al 1996). Different researchers use different parameters depending on what they want to measure. Root mean square (RMS) voltage, integrated EMG (IEMG) and median frequency (MF) were used in this study. Details on the literature review of the change in the amplitude of EMG (ie. RMS value and IEMG) during muscle strengthening is presented in section 2.2. The possible change of MF during muscle strengthening is presented in section 2.4.3.

### 2.4.1 Root Mean Square Voltage

Root mean square (RMS) voltage was defined by Bouisset (1973) as a continuous mean voltage or average EMG. RMS voltage is equivalent to the standard deviation of a certain period of the EMG data in the time domain (equation 2.1, fig. 2.2) when the EMG is processed in digital form. The standard deviation of the EMG signals was also provided by the PC-based signal processing software (Global Lab, Data Translation) that was used in this study.

$$RMS = \left[ \frac{1}{T} \int_{t}^{t+T} EMG^2(t) \, . \, dt \right]^{1/2} \qquad (2.1)$$

The average value of the EMG potentials (average EMG) is equal to a constant coefficient times the RMS value. RMS is another representation of the electrical energy needed to activate the muscle besides integrated EMG. RMS voltage is a measure of the energy content of the signals, especially useful for complex signals (Kwong 1995). Basmajian and associates (1985) stated that RMS voltage depended on the number of active motor units, firing rate of motor units, motor unit action potential shape and cross-correlation of motor unit discharges. RMS value was recommended by Basmajian (1985) as the best parameter. Therefore the change in RMS value during strengthening will also be examined in this study.

### 2.4.2 Integrated Electromyography

Integrated EMG was defined by Bouisset (1973) as the total amount of electrical activity in which the value is proportional to the area under the EMG envelope, that is, the quantity of electricity. It is noted in its mathematical formula (equation 2.2, fig. 2.2) that the IEMG is obtained by the average rectified value. Therefore the integrated rectified value will increase continuously as a function of time. The IEMG shown in equation 2.2 is obtained by integrating the EMG signal over a fixed time period, T (Basmajian et al 1985). Previous measurements of integrated surface EMG during isometric contractions have demonstrated moderate to high reliability coefficients for the quadriceps muscles, ranging from r = 0.77 to r = 0.94 (Pincivero et al 2000)

$$IEMG = \int_{t}^{t+T} \left| EMG\,(t) \right| \,.\, dt \qquad (2.2)$$

Duchateau and associates (1990) demonstrated that even after 6 weeks of immobilization, the MF of adductor pollicis muscle was not significantly different from that of the control muscles although the force and integrated EMG were drastically reduced during a maximal voluntary contraction (by 55% and 45% respectively). Herzog and associates (1994) also agreed that the integrated EMG was related to muscular force more often than any other form of EMG parameters. Integration is equivalent to

calculating the area under the rectified EMG-time curve. Hakkinen (1995) and Weir (1994) found that the IEMG was increased after strengthening of the quadriceps. However, no change in the IEMG of the quadriceps femoris muscle after strengthening could be found in other study (Garfinkel et al 1992). Therefore further study is necessary.

### 2.4.3 Median Frequency

Median frequency (MF) is one of the most commonly used parameters in EMG studies. MF is derived by dividing the power density spectrum into two equal halves (fig. 2.3). The frequency in between the two halves is the MF. Power density spectrum is obtained by transforming the EMG data in the time domain into frequency domain. Fast Fourier Transform is a very common mathematical method to do the transform. Studies investigating the change in MF related to muscle strengthening were reviewed in this section.

Moritani and associates (1993) studied the change in the power density spectrum of the surface EMG of biceps brachii muscles with strengthening exercise. It was found that there was a marked shift towards lower frequency band after two weeks of strengthening exercise (fig. 2.4). It was hypothesized that a better synchronization of motor units after strengthening would lead to large and low frequency EMG oscillations.

Some researcher studied the change of MF with strengthening exercise using electrical stimulation. Molina and associates (1997) found no change in the MF obtained

from the rectus femoris muscle after ten days of strengthening exercise of the quadriceps. He attributed the finding to the use of electrical stimulation instead of voluntary exercise. This was supported by Cannon's comment (1987) that only voluntary exercise would induce an increase in the recruitment of motor units through central adaptation.

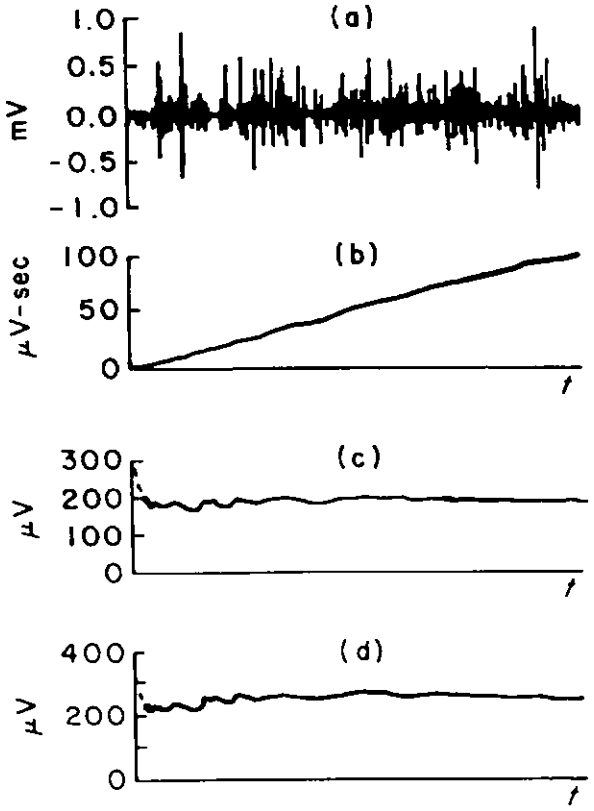



Figure 2.2. Comparison of data processing techniques: (a) raw EMG signal obtained from biceps brachii during a constant-force isometric contraction; (b) the IEMG; (c) the average rectified signal; (d) the RMS signal. (From Basmajian & De Luca 1985)

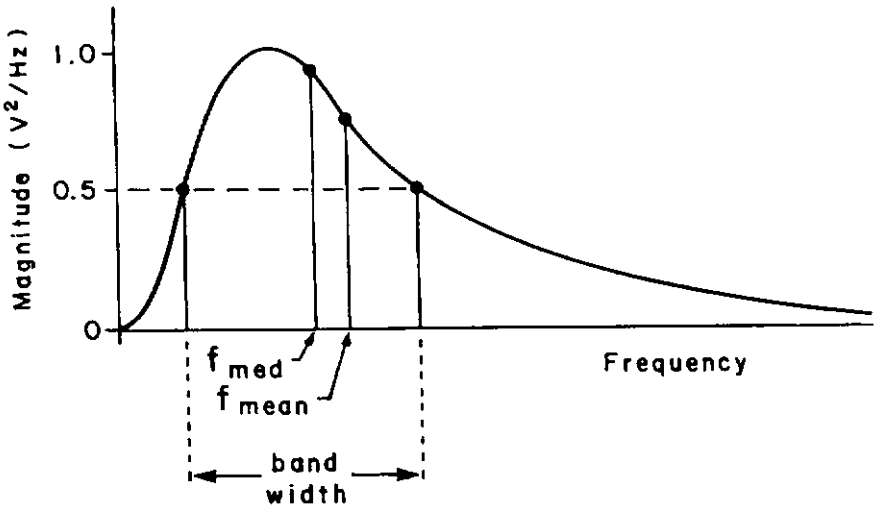
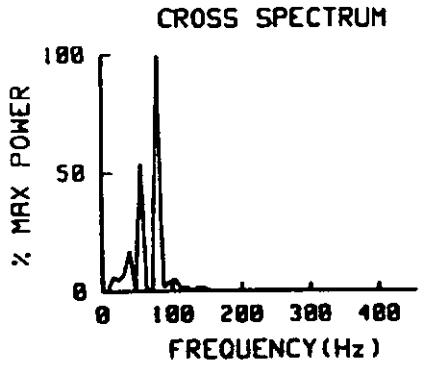
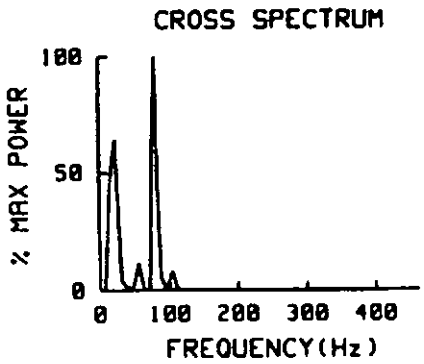



Figure 2.3. An idealized version of the frequency spectrum of the EMG signals. Three convenient and useful parameters: the median frequency, the mean frequency and the bandwidth are indicated. (From Basmajian & De Luca 1985)




Before training




After training

Figure 2.4. Cross spectrum of EMG recordings from the short and long heads of biceps brachii muscles before and after two weeks of training. Note the shift towards lower frequency after training. (From Moritani et al 1993)

Another possible reason of the change in the MF with strengthening exercises is due to changes in the muscle compositions or muscle types. Generally speaking, there are three types of motor units and muscle fibers. Each motor unit is innervated by a motor neuron and it consists of only one type of muscle fibers. They are the slow-twitch oxidative (Type I), fast-twitch oxidative (Type IIa), and fast-twitch glycolytic (Type IIb) muscle fibers (Basmajian et al 1985, Astrand et al 1986, Kupa 1995). Muscle composition is genetically determined, but they can be altered through training (Hamill 1995). Larger muscle fibers (type IIb) possess an inherently greater number of [Na+] and [K+] ion channels, thereby increasing the muscle fiber conduction velocity. This in turn will produce a larger MF value in the EMG (Pincivero et al 2000). Kupa (1995) studied the relation between the muscle types and median frequency of the EMG recorded in rats. He hypothesized that the fiber type can be estimated on the basis of EMG spectral parameters. He found that the initial MF and the rate of decline of MF during a sustained isometric muscle contraction, which was induced by a 20-second supramaximal stimulation, was greater in Type IIb fibers. It had been shown that Type IIa muscle fibers could be converted to Type IIb fibers through specific training which emphasized power and speed. It was not clear whether a transformation can be made from Type I to Type II through specific training (Hamill 1995). Basmajian (1985) also suggested that the characteristic frequency might be different in atrophied muscles although Duchateau and

associates (1991) found that the MF was not significantly different in disused and control muscles. It was found that rectus femoris muscle displayed the highest percentage of type II muscle fibers among all components of the quadriceps muscles (Pincivero et al 2000). It is interesting to know if the fiber composition of rectus femoris will be changed after strengthening exercise. The change in MF during the strengthening will be observed in this study and hope that this can give further insight in this aspect.

Furthermore, an increase in EMG frequency content was associated with an increase in relative force level (Basmajian & De Luca 1985, Gander & Hudgins 1985, Moritani & Muro 1987) although differences in skinfold thickness could be sufficiently potent to mask such a relationship (Bilodeau et al 1990). Mannion and associates (1996) studied the change of the MF in rectus femoris and vastus lateralis muscles during sustained isometric contraction of the quadriceps at various levels (20%, 40%, 60%) of maximum voluntary contraction (MVC) in a single session. It was demonstrated that there was a significant relationship between the rate of decline of MF and the rate of decline of the torque produced during sustained MVC in a single session. It was concluded that MF could be used to monitor muscle fatigue. In view of the discrepancy in the definition of muscle fatigue, Mannion and Associates defined it as "the inability of a muscle to generate the maximum force that can be produced by it in its fresh state". It was found that the rate of reduction in the MF and the rate of reduction in the torque at MVC were

linearly dependent on the % MVC at which the isometric contraction was sustained. The higher the submaximal force (% MVC) being maintained, the greater the rate of decline in MF and the strength at MVC. Mannion's study brought out a fact that the ability of a muscle to generate maximum force during a single session could be reflected by "the rate of decline in MF". A faster decline in MF implied that the ability of the muscle to generate force was compromised. Muscle strengthening is often the aim in rehabilitation. The ability of a muscle to generate maximum force will increase if the strengthening program is effective. It is interesting to know if MF can reflect this improvement in muscle strength during the rehabilitation process. It may be a useful monitoring parameter for muscle performance.

Median frequency per se is also a good indicator of muscle fatigue. The physiological mechanisms behind the change of MF during muscle fatigue have been extensively studied. Hagg (1992) listed several major physiological factors that affect the MF shift. These included nerve conduction velocity, temperature that affects the nerve conduction, action potential modification, firing rate and synchronization etc. Several cellular and molecular studies provided considerable insight into the various factors such as fatigue and temperature that affect the muscle fiber conduction velocity. As an excitable tissue, the muscle sarcolemma was sensitive to changes in either [K+] or [Na+] (Kamen et al 1996). Any change in sarcolemmal morphology, as what might occur with

intensive long-term exercise or activity resulting in muscle soreness, contributed to a change in muscle fibre conduction velocity. Because active metabolic processing was required to maintain the [Na+-K+] pump, any decline in adenosine tri-phosphatase (ATP) availability as what might occur with muscle fatigue could alter the conduction velocity (Kamen et al 1996). Muscle fiber conduction velocity was well correlated with central oxygen tension and conduction velocity failure in hypoxic conditions could also be due to a deleterious change in the [Na+-K+] pump (Kamen et al 1996). Kupa andassociates (1995) showed that MF is a well-documented fatigue index for Type I muscle under constant resistance in isometric contraction. Mannion and associates (1996) also agreed that change in MF was readily observable in the absence of any decline in the mechanical output. Leung (1996) further proved that mean power frequency (MPF) and MF were more sensitive fatigue indicators than mechanical work. Furthermore, it was reported that the frequency content of motor unit action potential trains was related mainly to the shape of the individual action potentials themselves. Conduction velocity was well related to the shape of the muscle fibre action potential. The well-known orderly recruitment of larger motor units with faster muscle-fibre conduction velocities was consistent with the tendency for more forceful contractions to result in higher-frequency content of the EMG. Motor-unit firing rate might contribute little to the frequency spectrum, and the shift towards higher frequencies observed with increasing force was likely attributable to the

recruitment of faster-conducting muscle fibers (Kamen et al 1996). It is possible that strengthening may change the recruitment pattern of the muscle resulting in a more efficient muscle contraction. If it is the case, MF may be changed after strengthening. There was evidence that light or moderate exercise might increase muscle fiber conduction velocity, possibly because of the changes in membrane characteristics, local increases in muscle fibre diameter, or increases in intramuscular temperature (Kamen et al 1996). The change in muscle fiber conduction velocity would also possibly induce a change in the MF.

There are a few EMG parameters in the frequency domain. The most commonly used parameters are MF and MPF. MPF is the average frequency of the power density spectrum. Bilodeau's study (1995) showed that there was a right shift of MF with increasing force levels despite the skinfold thickness whereas the trend of MPF is less predictable when the thickness of skinfold increased. MF is also preferred to MPF or mode frequency by De Luca (1997) because it is less sensitive to noise and signal aliasing. In most cases, it is more sensitive to the biochemical and physiological processes that occur within the muscles during sustained contractions. However, instability of the EMG signal spectrum at lower frequencies, which will distort the estimation of the MF, should be controlled. Bernardi and associates (1995) also pointed out that MF was more immune to noise than other spectral parameters. Therefore MF was chosen as one of the

parameters used in this study.

As a conclusion, it is worthwhile to study the change in MF during voluntary muscle strengthening. One of the purposes of this study is to find out whether a test protocol, which uses MF obtained from the power density spectrum, can be developed to assess muscle strength, in the process of rehabilitation.

## 2.5 Techniques for the Measurement of Torque and Electromyography

It was encouraging and fascinating in the 50's that electrical activity could be recorded from the muscle during its contraction. A lot of experiments were done to investigate whether and how this electrical activity (later called EMG) was related to the force produced by the muscle. It was found that raw EMG was not useful except as a qualitative indicator for muscle activity unless further data processing was done. The way that EMG was acquired was also crucial if EMG of high fidelity was expected. It was why the technical guidelines evolved. The format suggested in the "Standards for Reporting EMG Data" of the Journal of EMG and Kinaesiology, 1996" will be followed in this research. Based on these guidelines, the methodology of this study was developed. Information related to the techniques of surface EMG recording is outlined in the following sections.

### 2.5.1 Electrodes

*Types of surface electrodes*

Bipolar silver-silver chloride surface electrodes were recommended for the recording of surface EMG because it helped to minimize motion artifact. The size of electrode depends on the size of the muscle being recorded. It must be big enough so as to record EMG from a representative sample of the muscle fibers. However the size should not be too big because cross-talk from neighboring muscles may contaminate the signals (Kamen et al 1996).

***Placement of electrodes***

The active electrodes have to be placed over the muscle belly for surface EMG recording. Caution has to be taken to attach the electrodes away from the motor point because of the instability of EMG signal at the motor points. One suggestion is that the positive electrode is placed at the mid-point of the muscle. The negative electrode is then placed with an inter-electrode distance of 20 mm (Basmajian 1985, Fuglevand 1992). The electrode placement was in parallel with the orientation of the underlying muscle fibres.

### 2.5.2 Skin Impedance

In order to reduce the noise as much as possible, the impedance of the tissue-electrode junction should be kept as low as possible. The size of the electrode and the adequacy of preparation of the skin determine the impedance of surface EMG recording.

A thorough preparation of the skin is very important in order to reduce the noise as well as the sensitivity to other disturbances. The impedance should always be checked before any recordings are made and it is measured by using an alternating current impedance meter with a frequency in the range of 100 to 200 Hz. Impedance above a few kilo-ohms should not be accepted (Ortengren et al 1996). It was suggested that the general rule of thumb was to make certain that the resistance at the electrode site was 10 times lower than the input impedance of the preamplifier of the instrument. Before 1980's, almost all of the preamplifiers had input impedance of 100 $k\Omega$. This meant that an aggressive abrasion of the skin was needed to reduce the resistance at the electrode site to 10 $k\Omega$ or less. With the advent of new technology and better preamplifiers, the input impedance of current EMG instruments has risen to the 100 $M\Omega$ and $G\Omega$ range. This greatly reduces the need for "vigorous" skin preparation. This is not to say that skin preparation is not necessary. In fact, anyone who proclaims that skin preparation is a thing of the past and no longer necessary should be subjected to question. In theory and practice, impedance at the skin can easily climb up to the infinite range. No matter how high the input impedance of an EMG amplifier, it will never be able to handle the case of infinite impedance. The bottom line is that adequate attention needs to be given to skin preparation to ensure that the infinite resistance case does not happen in the experiment (Cram 1991). It seems that different authors have different opinion concerning the vigor

of skin preparation for surface EMG measurements. The claim made by Cram (1991) was reasonable. It was decided to do a thorough skin preparation in this research to ensure that skin impedance will not go to the infinite range. A detailed description of skin preparation and how skin impedance was controlled in this study would be given in chapter 3.

### 2.5.3 Signal-to-Noise Ratio

The signal-to-noise ratio is defined as the ratio between the maximum root mean square (RMS) voltage of an input sine wave signal that does not saturate the amplifier to that of the input noise voltage. It is expressed in decibel (dB), i.e. 20 times the logarithm of the voltage ratio. The noise voltage of an amplifier is measured at the output with the input short-circuited and the gain set to maximum. This measured voltage divided by the gain is the amplifier input noise voltage that is added to all signals connected to the input (Kumar et al 1996). The signal-to-noise ratio of this research was measured by the above method. Basmajian et al (1985) said that noise should be less than 5 μV (RMS voltage). The noise in this research was 2 μV (RMS voltage). The maximal undistorted level was 5 V. Therefore the signal-to-noise ratio of the set-up in this study was 128 dB as calculated by the method suggested by Kumar et al (1996).

### 2.5.4 Cross-talk

EMG cross-talk can be defined as the contribution of muscles, other than the muscle(s) of interest, to the detected myoelectric signal (Fuglevand et al 1992). For too small or deep-lying muscles, the appropriate strategy is to report the muscle group from which surface EMG recordings are made, rather than risk false identification of a muscle lying nearby. Large electrodes may be appropriate for larger or broader muscles when latency detection is important, because the larger electrodes can detect signals from a broader area. The surface-recorded signal may be less representative of activity in large muscles than in small ones (Kamen et al 1996). Therefore, the choice of appropriate size of electrodes has to balance between these two factors. Furthermore, Kamen and associates suggested that cross-talk could be minimized by placing pairs of electrodes at least 30 cm apart to reduce the volume-conducted signal. However, Fuglevand & associates (1992) found that the cross-talk from surface EMG was not severe enough to prevent it from examining individual muscle activity. It was found that only the largest motor units were detectable at depths more than 35 mm. In terms of the breadth of detection, the recording zone is typically restricted to those motor units within 10 to 12 mm of the surface electrodes. He pointed out two general misconceptions that may have led to an overestimation of the detection range of surface electrodes. First, it has been assumed that surface electrodes detect activity from greater distances because of their

large leading-off areas. His study suggested that the detection depth was influenced slightly only by the size of electrodes. And second, detection distances have been estimated for the volume conduction arising from synchronous activation of the entire muscle mass. Motor units primarily discharge independently from one another during voluntary activity. Assessment of cross-talk from synchronous activation of all motor units, therefore, would seem to severely overestimate the detection distances expected in normal, voluntary muscle activity. It was found that although the surface signal is widely considered as representative of the EMG activity in the underlying muscle, the amount of area represented in the signal is rather small. Only the largest motor units are detectable at depths of about 35 mm. In terms of the breadth of detection, the recording zone is typically restricted to those motor units within 10-12 mm of the surface electrodes, and the signal from the remaining muscle is received at the electrodes with about the same signal strength as the ambient noise (Fuglevand et al 1992). Interelectrode spacing should also be controlled if only the EMG of the superficial muscle eg. rectus femoris instead of vastus intermedius, is of interest. It is because an increased inter-electrode distance will moderately increase the detection depth (Fuglevand et al 1992). Solomonow and associates (1990) carried out a study on cat's calves and tibialis anterior muscles. It was found that the EMG signal of the M waves recorded from the lateral gastrocnemius and tibialis anterior after surgical severance of their nerves (ie. only the medial gastrocnemius

was intact and being stimulated) did not exceed 5 % of their maximal value (when the nerve was intact and stimulated with a supramaximal electrical stimulus) for surface electrodes. It also did not exceed 2.5 % of their maximal value for intra-muscular wire electrodes. Since this cross-talk (5 %) was obtained in maximally activated muscle, an over-estimation of the cross-talk that can be expected in voluntary contraction. He concluded that the cross-talk problem in surface recording was negligible for most biomechanical studies in which standard EMG recording protocol was employed. Yet a warning was issued against the indiscriminate recording of surface EMG from muscles covered by a substantial amount of adipose tissue. Surface EMG cross-talk values were significantly higher in preparations in which a substantial amount of subcutaneous fat covered the muscles, ie. about 20 %. Typical muscles from which surface recordings should be avoided are the gluteus, abdominals and most muscles of individuals with a noticeable overlay of adipose tissue. On the contrary, it could be concluded that for surface recording of the EMG with the appropriate size of electrodes, correct placement over the muscle and inter-electrode distance one should disregard the effect of cross-talk in most skeletal muscles of the extremities, spine and upper trunk. Combining the opposing effects of larger distances for the EMG to travel in human muscles and the increased signal intensity because of the larger muscle size, the net effect is that the

proportions are similar. Therefore the cross-talk found in the cat was expected to be valid for human studies as well.

Kamen and associates (1996) suggested that cross-talk could be minimized by using the double-differential or branched-electrode techniques. They also suggested two methods to examine cross-talk:

1. Perform functional resistance tests that isolate specific muscle groups and examine the activity in non-active muscles.
2. Pairs of EMG signals in which the existence of cross-talk is suspected can be cross correlated to examine the interrelationship between the two muscles. Signals that are highly correlated either simultaneously or with a time lag attributable to volume conduction delays are frequently assumed to represent an appreciable degree of cross-talk. This method may be used to differentiate cross-talk from neighboring quadriceps muscles.

Nevertheless, De Luca (1997) has discussed about the limitation of the cross-correlation method. Cross-correlation of the EMG signals from two neighboring muscles might not be high just because the tissues between and within the various muscles are anisotropic and inhomogeneous. This yielded a false-negative error. On the other hand, the firing rates of motor units in different muscles contracting to perform a specific task could be considerably cross-correlated. Thus, the surface EMG signals from such

simultaneously contracting muscles could be substantially cross-correlated without the presence of cross-talk, yielding a false-positive error. Moritani and associates (1993) also found that the cross-correlation coefficient between the long and short heads of biceps brachii muscles changed from 0.4 to 0.91 after two weeks of power training. They proposed that this might be due to a more synchronous motor unit activities. This supported the viewpoint of De Luca that the cross-correlation coefficients between two muscles might be high just because of the similarity in their motor recruitment pattern instead of cross-talk. De Luca also criticized the muscle test method because one could not know definitely if the subject was activating the nearby muscles during the tests, thus yielding a false-negative error. De Luca commented further that there was only one way to reduce and possibly eliminate cross-talk in the EMG signal detected with surface electrodes. This was the double differential technique. This could be applied when cross-talk from antagonistic signal was suspected.

Different methods of determining the cross-talk were used by Draganich and associates (1989). They examined the EMG of all knee flexors and extensors during voluntary isometric contraction of the quadriceps and checked the ratio of the EMG acquired in the hamstrings to that in the quadriceps. The EMG of the two muscle groups were obtained simultaneously. The cross-talk measured by this method was less than 3 %. The second method used by Draganich was to compare the EMG recorded at the vastus

lateralis and biceps femoris muscles during supra-maximal stimulation of the femoral nerve. The cross-talk were 2.7 and 2.8 % in two subjects.

The cross-talk to the rectus femoris muscle from the adjacent vastus lateralis muscle and the antagonist biceps femoris muscle was investigated in this study using the cross-correlation method although the limitation of this method was noted. Therefore the EMG measured was cross-correlated with the surface EMG obtained from vastus lateralis and biceps femoris. More details is presented in section 3.4.2.

### 2.5.5 Co-contraction of the Hamstrings

When a muscle contracts vigorously, it is very common that the other muscles around the joint will also contract. This is called co-contraction or coactivation. Co-contraction of the antagonist (ie. hamstrings in this study) can affect the measurement of torque of the quadriceps. Therefore, the significance of its effect must be determined before the torque measured by dynamometer can be used for further analysis. It is known that simultaneous activation of the agonist and antagonist is more apparent during the early stages of motor learning and during muscle fatigue. Coactivation is different in trained athletes compared with control individuals (Kamen et al 1996). However, Hakkinen and associates (1998) found that coactivation was not significant in middle aged male subjects who underwent six months of intensive strengthening and power training (fig. 2.1) whereas it was much more influential in elderly male subjects.

Sufficient warm-up and training were given in order to reduce co-contraction in this study. More details on the preparation procedures are presented in section 3.2. The method and results to determine the significance of hamstrings co-contraction are reported in section 3.4.3.

### 2.5.6 Normalization of Data

The amplitude and frequency characteristics of the raw EMG detected by surface electrodes have been shown to be sensitive to many intrinsic and extrinsic factors. Section 2.2 has already given a detailed description of these factors. Some of the intrinsic factors cannot be perfectly controlled. The amount of tissue between the muscle and the electrode, the exact number of muscle fibers under the electrodes are some examples of these factors. Therefore in order to allow comparison of EMG activity between different muscles, across time and between individuals, the EMG is often normalized with respect to force (De Luca 1997). Normalization means that the EMG value is expressed in relation to a reference value obtained during standardized and reproducible conditions (Burden et al 1999). Various methods of normalization have been used in EMG studies. EMG from an isometric maximal voluntary contraction (MVC) was often used as the normalization reference value (Draganich et al 1989, Salzman et al 1993, Narici et al 1996, Hakkinen et al 1998, Pincivero et al 2000). However, other researcher (De Luca 1997) suggested to normalize the EMG at values less than 80 % MVC. It was because

above this level, the EMG signal and the torque are exceptionally unstable and do not provide a suitable reference point. The normalization method depends on how the data will be analyzed. Pincivero and associates (2000) decided to use the IEMG obtained at the beginning of the lunge performance to normalize all subsequent IEMG values instead of using the EMG at MVC. It was because the specificity of the lunge performance and the isometric MVC would not yield a valid interpretation. De Luca also raised an important reservation to the use of normalization. Its tendency to render similar the data from different subjects tends to suppress distinctions in the data that would be associated with abnormal or pathological cases. This is a major concern when the EMG signal is used for analysis of clinical data.

Double normalization of EMG data was performed in this study because pre-determined torque level instead of a fixed percentage of the MVC was used in this study. The double normalization procedure facilitates the between-session comparison of EMG data at submaximal contractions. Since both the tests and the strengthening exercise involved isometric knee extension, the EMG at the MVC of knee extension was chosen for the first step of normalization. The EMG value would then be normalized with the relative change in the torque at MVC. Section 3.5.3 will give a detailed description of this normalization technique.

## 2.6 Summary

The theme of the present study was to find out the relationship between the change in EMG and the improvement in muscle strength of the quadriceps femoris muscle with a strengthening program. The study was carried out in normal subjects. A comprehensive literature review on several aspects related to strengthening of muscle and surface EMG was presented in this chapter. The possible physiological mechanisms of muscle strengthening are neural adaptation and change in the muscle-tendon structures. Neural adaptation includes increase in neural input by changing motor recruitment pattern or firing rate, improved coordination of agonist, antagonist and synergistic muscle action. Change in EMG at MVC is a good indicator for neural adaptation. The changes of muscle-tendon structures are muscle hypertrophy, change in the sarcolemmal physiology or muscle fiber type. Changes in surface EMG during muscle strengthening and immobilization of muscle was also discussed. The characteristics of three common parameters used in surface EMG recording (root mean square voltage, integrated EMG and median frequency) have been reviewed. Their suitability for this research was discussed. A thorough description of the technical aspects of surface EMG recording has also been given. This included types of electrode, placement of electrode, skin preparation, signal-to-noise ratio, cross-talk, hamstrings co-contraction and normalization

of data. The information in this chapter supports the development of methodology of this study. Details of the methodology are reported in chapter 3.

# Chapter 3 Methodology and Instrumentation

## 3.1 Introduction

The objective of the present study was to find out the change of EMG of the rectus femoris muscle (part of the quadriceps femoris) during a period of strengthening exercise of the quadriceps femoris muscles in normal healthy subjects. The electromyography (EMG) parameters used in this research were median frequency (MF), integrated EMG (IEMG) and root mean square voltage (RMS). The change of EMG during maximal and submaximal contraction of the quadriceps femoris muscles was investigated. This study attempted to examine whether MF, IEMG and RMS were useful indicators of muscle strength besides biomechanical measurement (eg. torque). It is expected that new information about the relation between the change in EMG and the improvement in muscle strength will improve the understanding of the physiological mechanism of increase in muscle strength. It may also facilitate the development of a better method to monitor the progress in muscle strength during rehabilitation.

## 3.2 Methodology

Twelve normal, healthy and young (20 to 35 years old) male subjects were recruited for the study. The study was approved by the Ethics Committee of The Hong Kong Polytechnic University and Alice Ho Miu Ling Nethersole Hospital. Obese

subjects, ie. those with Body Mass Index (BMI = body weight, in Kg/ body height$^2$, in m) above 27 (ACSM Resource Manual 1998), were excluded because a significant layer of adipose tissue will increase the cross-talk (Bilodeau et al 1992, Solomonow 1990). The strengthening program and test protocols are described below.

### 3.2.1 Strengthening Program

The strengthening protocol involved isometric resisted exercise of the knee extensors of the right leg. The subjects were engaged in a scheduled strengthening program of 30 maximal voluntary contractions (MVC) per day, 3 days a week, for 6 weeks (Garfinkel et al 1992). Warm-up exercise including stretching of the hip, knee and ankle muscles was done before the strengthening. The subjects were then seated with arms crossed in front of the chest. The hip and knee were fixed at about 60 degrees of flexion. Resistance was applied at the distal part of the lower leg, just above the ankle joint. The subject could easily be positioned again at subsequent exercise session because all the positional information was recorded in the software installed in the computer of the dynamometer (Cybex NORM, USA). Another warm-up exercise was performed using submaximal right knee extension. The subjects then performed the strengthening program using maximal voluntary contraction of the right knee extensors with the dynamometer. During the exercise, verbal encouragement was given in order to maximize motor recruitment.

### 3.2.2 Test Protocol

The testing position was exactly the same as that of strengthening (section 3.2.1). The following data were collected.

1. The extension torque of the knee was measured at maximal voluntary contraction (MVC) of the right quadriceps muscles;
2. Surface electromyography (EMG) was recorded on the right rectus femoris muscle at MVC and two submaximal torque levels of isometric contraction, at 100 Nm and 150 Nm. The raw EMG was further processed with a PC-based signal processing software (Global Lab, Data Translation). MF, IEMG and RMS voltage were calculated from the EMG data for subsequent data analysis.

   The test was repeated every two weeks by the same operator with the same set-up. Figure 3.1 shows the schematic diagram of the set-up.

| Electrode | Data Acquisition (EMG Amplification) | ...Filter | Analogue-to-Digital Convertor | Signal Processing |
|---|---|---|---|---|

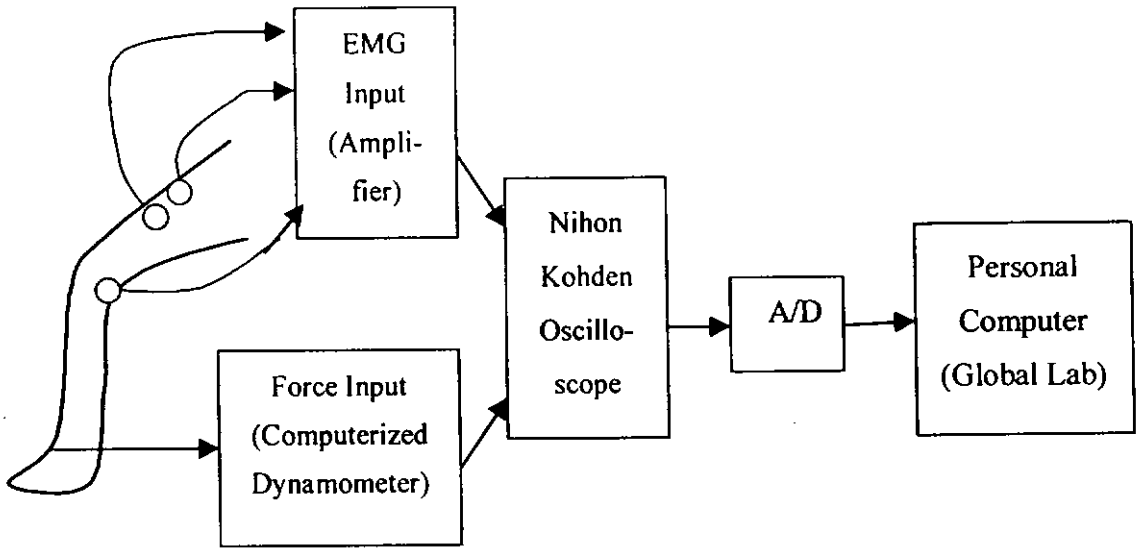



Figure 3.1. Schematic diagram of the experimental set-up

***Measurement of torque and surface EMG at MVC***

There was a fifteen-minute warm-up session of stretching exercise, submaximal isotonic and isometric exercises of the right quadriceps muscles. The subjects then practiced submaximal isometric contractions of right knee extension at 100 Nm and 150 Nm with visual feedback from the visual display (14"), which helped the subject to maintain a pre-set torque level. The position of the subjects was crucial for reliable measurement of torque. Therefore the subjects were secured on the chair by thigh, pelvic, and torso straps in order to minimize body movements. The same position was

used in subsequent tests. The subjects were asked to exert a maximum torque of the knee extension and sustain for ten seconds until they were told to relax. They were allowed to stop if they could no longer sustain the knee extension at full effort. Verbal encouragement was provided to facilitate the generation of maximum torque. Three trials with five minutes rest between each one were performed. The greatest value of the three trials was taken as the torque at MVC (De Luca 1997, Narici et al 1996, Hakkinen et al 1995, 1998) although there was study using the average peak torque of the three MVCs (Pincivero et al 2000). Nonetheless, Pincivero and associates found that the intra-class correlation coefficient (ICC) of the three MVCs within the same session was very high (ICC = 0.98). Automatic gravity correction was obtained by measuring the torque exerted on the dynamometer resistance adapter with the knee in a relaxed state at 60° flexion. Position calibration and weight calibration of the dynamometer were also performed according to the manufacturer's specifications prior to every testing session. Surface EMG of the rectus femoris muscle was also recorded simultaneously. The choice of EMG of the rectus femoris muscle to infer their myoelectrical activity to all heads of the quadriceps was based on the fact that all heads work equally during the isometric contraction of the quadriceps muscles (Molina et al 2000, Pincivero et al 2000, Salzman et al 1993).

***Measurement of surface EMG at submaximal contractions***

Besides the maximal contraction of right knee extension, the subjects were also instructed to do isometric knee extension at two pre-determined submaximal levels (100 Nm and 150 Nm). The subjects were asked to maintain the torque at the pre-set level for twenty seconds in each trial. Two minutes break was given in between the trials. The subjects were required to exert an extension torque of the knee within 20% of the required level. Subjects were asked to match a line on the Cybex computer monitor that corresponded to the pre-set torque level (fig. 3.2). It was considered as fatigue when there was a decrease in muscle strength of more than 20% (Mannion et al 1996). Surface EMG of the rectus femoris muscle was recorded simultaneously. The measurement was repeated every two weeks under the same condition. Details of the instrumentation for measurements of torque and surface EMG is described in the next section. The torque levels of 100 Nm and 150 Nm were chosen because the isometric torque at MVC of all subjects was above 160 Nm in this study. Therefore it was realistic to use these two levels of torque for submaximal tests. The very low torque level, eg. 50 Nm was not chosen because the EMG signal was weak relative to the noise in the environment. This would probably result in a low signal-to-noise ratio. Another characteristic of this test protocol was to use two fixed levels of torque

instead of using a percentage of the torque at MVC during submaximal test. The characteristics of this test protocol will be discussed in section 5.1.

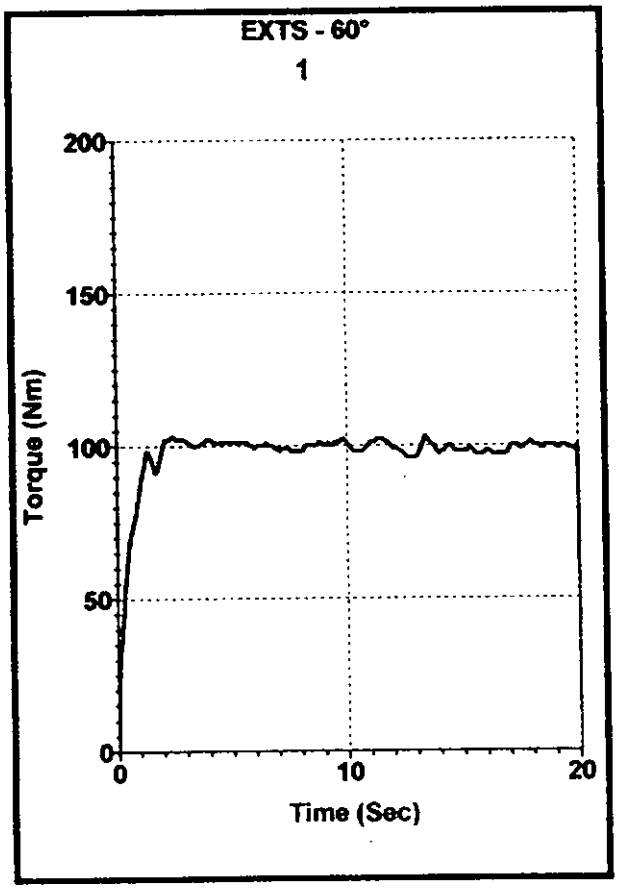



Figure 3.2. Visual display of the computerized dynamometer. Isometric contraction at 100 Nm torque level is facilitated by visual feedback from the display.

## 3.3 Instrumentation

The details of the instrument used for the measurement of torque and EMG will be described in this section. The method used to synchronize the onset of the EMG signals and the torque produced at the extension of the leg will also be described.

### 3.3.1 Measurement of Torque

The torque at maximal voluntary contraction generated by isometric extension of the right knee was measured by a dynamometer (Cybex NORM, USA). The position was recorded in the software installed in the computer of Cybex dynamometer and therefore the subjects could be accurately repositioned every time. Details of the set-up were reported in sections 3.2.1 and 3.2.2.

### 3.3.2 Measurement of Electromyography

The format described in the "Standards for Reporting EMG Data" of the Journal of EMG and Kinaesiology, 1996 was followed in this study.

Electrode types : Bipolar silver-silver chloride surface electrodes

Electrode size : 8 mm in diameter

Electrode placement : Electrodes were placed over the rectus femoris muscle. A line was drawn between the anterior superior iliac spine (ASIS) and the tip of the patella when the subjects lied flat. The positive electrode was placed in the mid-point of this

line. The negative electrode was fixed distally with 20 mm centre-to-centre distance. The ground electrode was placed over the fibular head of the same leg. The positions of the electrodes were marked by silver nitrate (Weir et al 1994) so that the electrodes could be placed at the same location at subsequent sessions.

Skin preparation : The skin was cleaned and scrubbed with detergent and then cleaned with alcohol. Elefix gel (Nihon Kohden) was used as the conducting medium. Skin impedance was kept below 5 kΩ.

EMG acquisition : EMG was captured and amplified by the Nihon Kohden oscilloscope. The EMG was then acquired by a PC-based analogue-to-digital system, sampled at 5 kHz.

Amplification : Differential amplifier with common mode rejection ratio (CMRR) of 80 dB and 180 MΩ input impedance was used.

Filtering : High-pass cut-off frequency was 5 Hz and low-pass cut-off frequency was 1 kHz. The signal was then filtered at low-pass frequency of 350 Hz using the infinite impulse response (Butterworth) digital filter in Global Lab (Data Translation).

Gain : 1000

Signal-to-noise ratio : 128 dB (please refer to section 2.5.3 for the derivation of the signal-to-noise ratio)

### 3.3.3 Synchronization of Signals

A segment of EMG signals and the torque corresponding to the same period of time have to be extracted for further data processing. Therefore, the onset of the torque, which was acquired by the dynamometer, and the EMG recording, which was acquired by the amplifier, have to be synchronized. An interface cable, which connected the dynamometer with the amplifier, will serve the purpose of synchronization. Both signals were then recorded and processed using the PC-based signal processing software (Global Lab, Data Translation).

## 3.4 Control of Variables

There are several variables that should be controlled if surface EMG and muscle torque could be acquired with high fidelity. These are skin impedance, cross-talk and hamstrings co-contraction.

### 3.4.1 Skin Impedance

Electrical signal from a muscle has to pass through the skin before it reaches the surface electrodes. The skin layer and the greasy material on the skin create high impedance to the electrical signal and thus affect its fidelity. This is particularly

important for quantitative measurement and analysis. In order to keep the skin impedance below 5 kΩ, three kinds of gel and two skin preparation methods have been tested. The method and result are presented.

***Method***

Resistance of three types of gel was measured to see whether it affected the outcome. All these gel were commonly used in diagnostic EMG. They were "Redux Paste" (Hewlett Packard), "Synapse conductive electrode cream" (Med Tek Corporation) and "Elefix" (Nihon Kohden).

The two skin preparation methods are described as follows. Firstly, the skin was cleaned with soap, scrubbed with the electrodiagnostic gel - skin rub (Nihon Kohden) and then cleaned with alcohol. Secondly, the skin was prepared by scrubbing with a detergent that was efficient in removing the grease first and then cleaned with alcohol.

***Results***

Table 3.1 shows the results. It is obvious that the resistance of gel is negligible. All gels tested had very low resistance (below 1 kΩ) though Elefix gel was finally used in the main study because of its low viscosity. Skin impedance was measured with the Hewlett Packard 4194A Impedance/ Gain-Phase Analyzer at the Industrial Centre of The Hong Kong Polytechnic University. Two measurements were taken with skin prepared by the two methods. In the first method, the skin was

prepared by soap, skin rub and alcohol, the impedance was about 15 kΩ at 500 Hz and about 20 kΩ at 200 Hz. In the second method, the skin was scrubbed by a good detergent and cleaned with alcohol, skin impedance dropped to about 2.4 kΩ at 200 Hz (Figure 3.1). This was within the desirable range of skin impedance.

Table 3.1 Resistance of three types of gel as measured by "Hewlett Packard 4332A" LCR meter. All measurements were taken at 1 kHz

| Resistance and capacitance / Conditions | Redux Paste | Synapse conductive electrode cream | Elefix |
|---|---|---|---|
| Direct measurement of the gel (2 Ag-AgCl surface electrodes sandwiched with the gel) | R = 50 Ω | R = 220 Ω | R = 46 Ω |

R: resistance, in Ohms

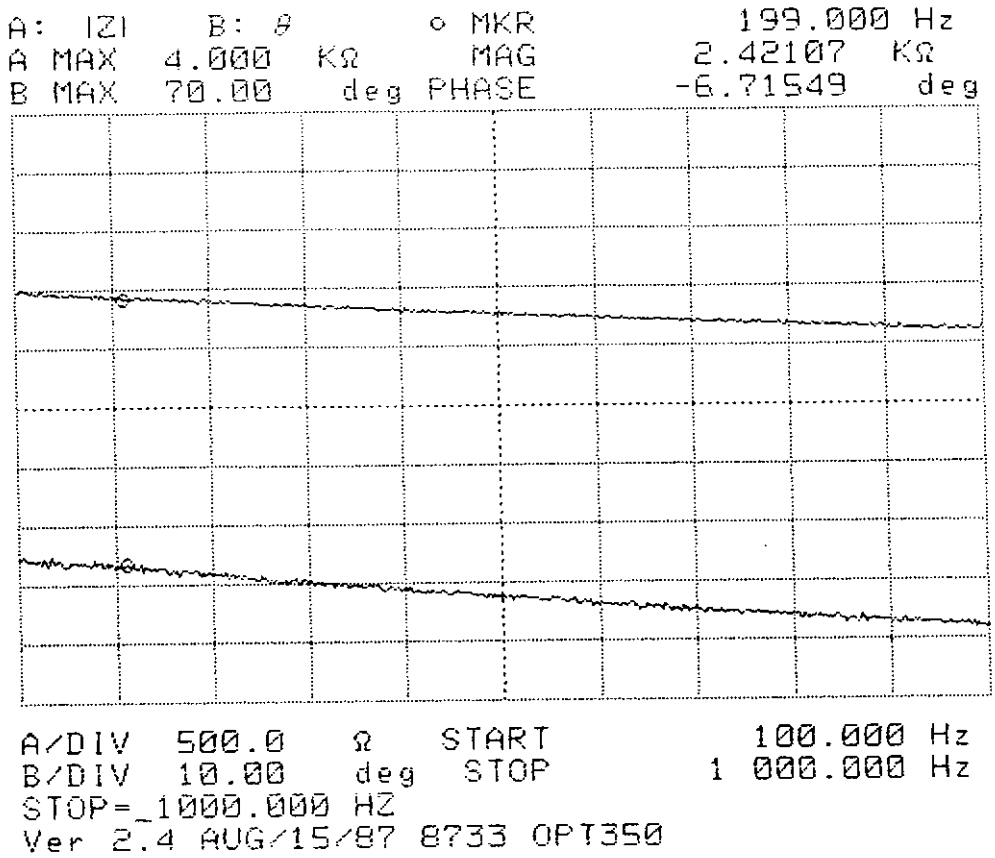



Figure 3.2. Skin impedance between the Ag-AgCl electrode and human skin as measured by the Hewlett Packard 4194A Impedance/ Gain-Phase Analyzer. The skin is prepared by gentle scrubbing with a strong detergent and then cleansed with alcohol. Elefix gel is used as the conduction medium.

***Conclusion***

The later method of skin preparations, ie. scrubbed with detergent and cleaned with alcohol, was adopted in the main study because the impedance was within the desirable range. Moreover, it was concluded that all the three types of gel tested had low resistance (below a few hundred ohms) and they were suitable for surface EMG measurements.

### 3.4.2 Cross-talk

The EMG of rectus femoris muscle would be measured in this study. It was possible that the electrical activity from neighboring muscles such as vastus lateralis and biceps femoris muscle would contaminate the EMG signal measured. Therefore it was crucial to understand the effect of cross-talk. Pairs of EMG signals, in which the existence of cross-talk was suspected, were cross-correlated to examine the interrelationship between the two muscles.

***Method***

Two young male subjects were tested. Three pairs of surface electrodes were put on the right rectus femoris (RF), vastus lateralis (VL) and biceps femoris (BF) muscles. The subjects were asked to do MVC of isometric knee extension and MVC of isometric knee flexion against the resistance applied at the lower leg by the Cybex dynamometer. Surface EMG of the RF, VL and BF were acquired during these contractions.

1. Measurement of the surface EMG of rectus femoris muscle

   The method described for the main study (section 3.2.2) was used.

2. Measurement of the surface EMG of the vastus lateralis muscle

Electrode types : A pair of bipolar silver-silver chloride surface electrodes was securely attached on the muscle belly of vastus lateralis muscle.

Electrode size : 8 mm in diameter

Electrode placement : The vastus lateralis muscle was located by identifying its muscle belly under resisted extension of the knee. The negative electrode was put distally with centre-to-centre distance of 20 mm (Basmajian 1985). The electrode placement was in parallel with the fibre orientation of the underlying muscle. The indifferent electrode was placed on the patella of the same leg.

Skin preparation : The skin was thoroughly scrubbed with detergent and cleaned with alcohol. Elefix gel (Nihon Kohden) was used as the conducting medium

EMG acquisition : EMG was captured and amplified by the Nihon Kohden oscilloscope. The EMG was then acquired by a PC-based analogue-to-digital system, sampled at 5 kHz.

Amplification : Differential amplifier, CMRR 80 dB, input impedance 180 MΩ

Filtering : High-pass cut-off frequency 5 Hz, low-pass cut-off frequency at 1 kHz. The signal was then filtered at low-pass frequency of 350 Hz using the infinite impulse response (Butterworth) digital filter

Full wave rectification

EMG data processing : Global Lab, a general-purpose signal processing software, was used, sampling frequency: 5 kHz.

Gain : 1000

3. Measurement of the surface EMG of the hamstrings

Electrode types : A pair of bipolar silver-silver chloride surface electrodes was secured over the muscle belly of biceps femoris muscle.

Electrode size : 8 mm in diameter

Electrode placement : The biceps femoris was located by resisted muscle work and palpation. The negative electrode was put distally with centre-to-centre distance of 20 mm to the positive electrode (Basmajian et al 1985). The electrode placement was in

parallel with the underlying muscle fiber orientation. The indifferent electrode was put on the patella of the same leg.

Skin preparation, EMG acquisition, amplification, filtering, EMG data processing were the same as that for the measurement of surface EMG of the vastus lateralis muscle.

The surface EMG signals were then cross-correlated with the software SPSS 7.5 (Statistic Package for Social Sciences). Graphical illustration of CCF is given in figure 3.3. The cross-talk (X-talk) was calculated from the cross-correlation coefficients (CCF). Equation 3.1 illustrates the derivation of the cross-talk.

$$\text{X-talk} = (\text{CCF})^2 \qquad (3.1)$$

***Results***

Table 3.2 illustrates the results of cross-talk between rectus femoris (RF) and vastus lateralis (VL) muscles at the maximal voluntary contraction (MVC) of the knee extensors, and the cross-talk between rectus femoris and biceps femoris (BF) muscles at MVC of the knee extensors and flexors. The test protocol of the main study required the subjects to perform MVC of the knee extensors. The cross-talk values ranged from 1.44 % to 17.64 % during the MVC of quadriceps muscles in this pilot study. This pilot study also illustrated that the cross-talk values between RF and BF muscles were much greater during MVC of the hamstrings muscles (9.61 % and 30.25

%) when compared with the values during MVC of the quadriceps muscle. Moreover, all the cross-talk values of subject B were greater than those of subject A.

***Conclusion***

The effect of cross-talk should not be neglected in the measurement of surface EMG although previous reports (Fuglevand et al 1992, Solomonow et al 1990) found that its effect was overestimated (section 2.5.4). Factors that may increase the cross-talk are electrode size, electrode spacing, and the amount of adipose tissue of the subjects (section 2.5.4). The electrode size (8 mm in diameter) chosen for this study was optimal for the size of the human rectus femoris muscle. It is because too big a size will increase the risk of cross-talk whereas too small electrodes may not be able to pick up adequate electrical signals that can represent the muscle as a whole. This size was also often used in other similar studies (Garfinkel et al 1992, Narici et al 1996). The inter-electrode spacing (20 mm centre-to-centre distance) was also similar to other EMG studies on human quadriceps muscle (Garfinkel et al 1992, Hakkinen et al 1995, Narici et al 1996, Mannion et al 1996). It is because a large inter-electrode distance will increase the detection depth of SEMG and may pick up electrical signals from underlying muscle. It was also reported that muscles covered by a significant layer of adipose tissue would increase the cross-talk (Solomonow 1990). Therefore obese subjects were excluded from this study. The average body mass index of the

subjects recruited in this study was 22.2. Despite all the precautions, the cross-talk between RF and BF was still high (0.3025) during the MVC of the hamstrings. Since only MVC of knee extension and not knee flexion would be performed in the main study, this would not influence the interpretation of EMG. It was interesting that the cross-talk between the same muscle groups (RF and BF) was very low (0.04 and 0.0625) during the MVC of the quadriceps. During the MVC of the quadriceps, all cross-talk values were considerably low except the cross-talk between the RF and VL of subject B (0.1764). Although signals that are highly correlated have been assumed to represent an appreciable degree of cross-talk (Kamen et al 1996) but a high cross-correlation between the RF and VL might be due to the synchronous activation of motor units (De Luca 1997). It is because RF and VL worked together to produce knee extension and these two muscles are supplied by the same motor nerve (femoral nerve). Therefore, the cross-talk values found in this pilot study were considered as acceptable and the same set-up for surface EMG measurement was adopted in the main study.

Table 3.2.Results of the maximum cross correlation coefficients (CCF) and cross-talk (X-talk) between RF and VL muscles at MVC of the knee extensors, and that between RF and BF muscles at MVC of the knee extensors and flexors. Graphical illustration of the CCF is given in figure 3.3.

| | Subject A | Subject B |
|---|---|---|
| CCF & X-talk between RF & VL at the first MVC of the quadriceps | CCF = 0.12<br>X-talk = 0.0144 | CCF = 0.42<br>X-talk = 0.1764 |
| CCF & X-talk between RF & BF at the second MVC of the quadriceps | CCF = 0.2<br>X-talk = 0.04 | CCF = 0.25<br>X-talk = 0.0625 |
| CCF & X-talk between RF & BF at the MVC of the hamstrings | CCF = 0.31<br>X-talk = 0.0961 | CCF = 0.55<br>X-talk = 0.3025 |

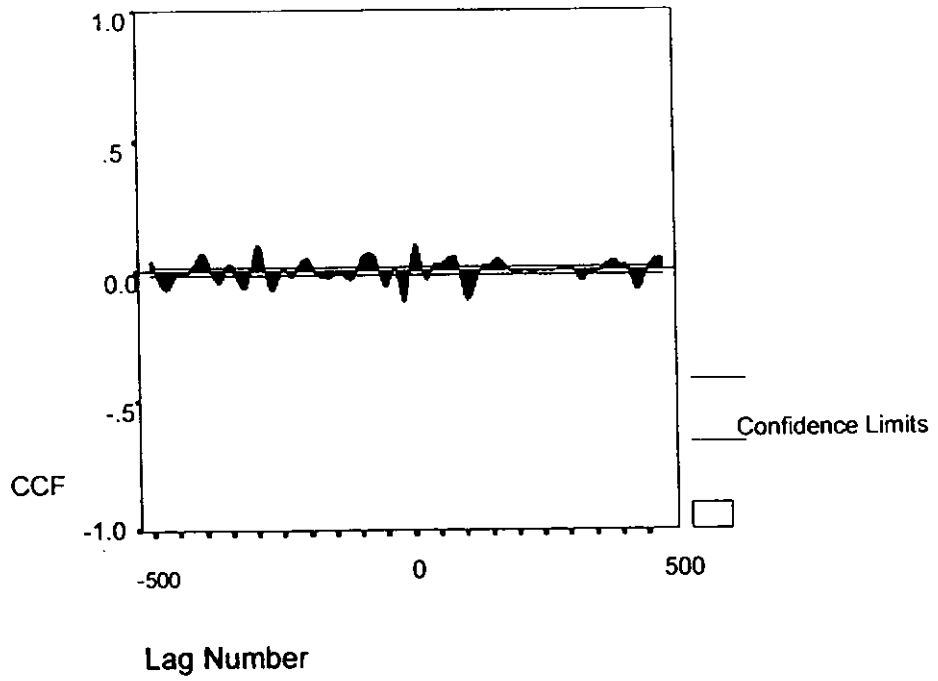



Figure 3.3 (a). Graphical illustration of the cross-correlation coefficients (CCF) between RF & VL muscles at MVC of the knee extensors (subject A).

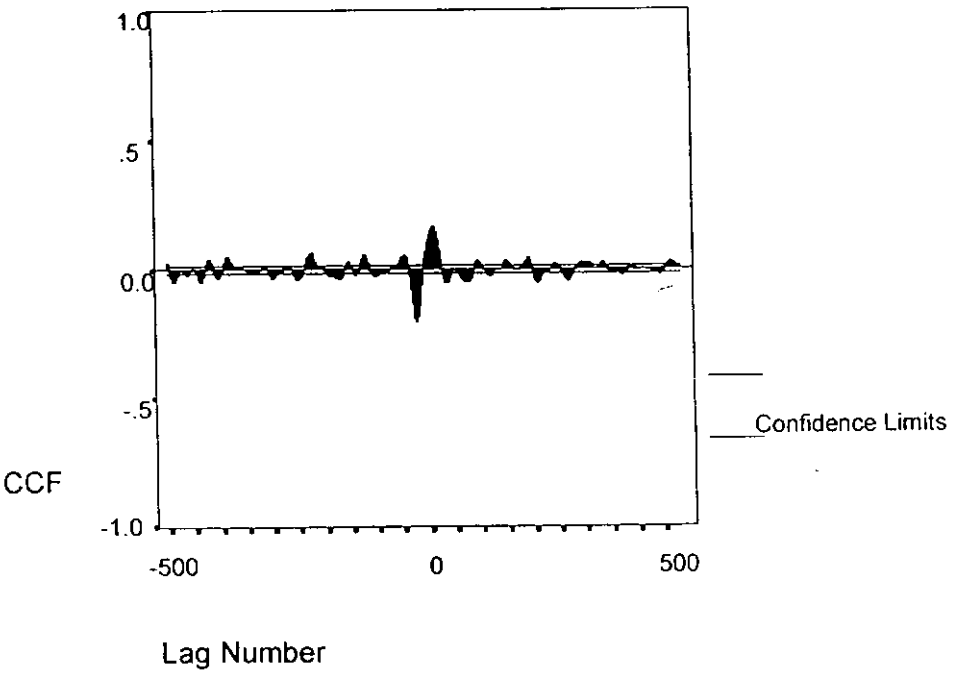



Figure 3.3 (b). Graphical illustration of the cross-correlation coefficients (CCF) between RF & BF muscles at MVC of the knee extensors (subject A).

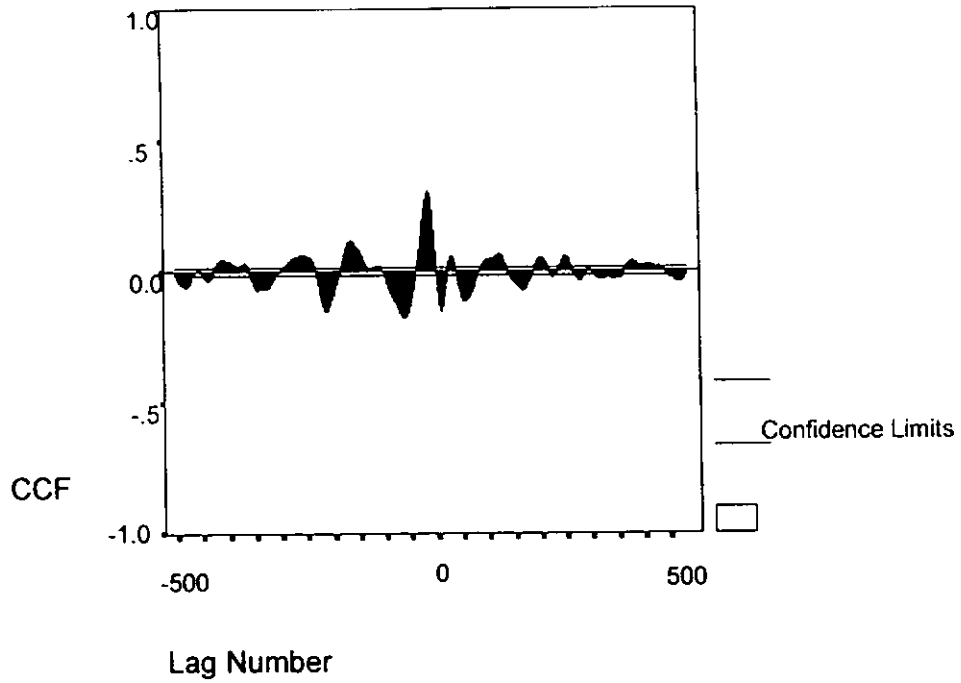



Figure 3.3 (c). Graphical illustration of the cross-correlation coefficients (CCF) between RF & BF muscles at MVC of the knee flexors (subject A).

### 3.4.3 Co-contraction of the Hamstrings

Co-contraction of antagonist (ie. hamstrings muscle is the antagonist of quadriceps muscle in this study) can affect the measurement of torque of the quadriceps. Therefore, the significance of its effect must be documented before the torque measured by dynamometer can be used for further analysis.

***Method***

In order to document the influence of hamstrings contraction on the measurement of the torque at MVC of the quadriceps, a pilot study was conducted. Intramuscular wire (IMW) EMG of the biceps femoris muscle was measured. IMW EMG was used because the effect of cross-talk in IMW EMG is less than that in surface EMG. Therefore, the IMW EMG activity could be considered as a true reflection of the activity of the biceps femoris muscle. Three young subjects (2 male and 1 female) were recruited. The subjects were trained for fifteen minutes so that they could exert 50 Nm, 100 Nm, 150 Nm and MVC of isometric knee extension. The subjects were also asked to perform MVC of isometric knee flexion. Fifteen minutes break was given before data collection. Intramuscular wire EMG of the biceps femoris muscle was recorded simultaneously during all tests.

Technical aspect of this pilot study is as follows:

1. Procedures of the test

The positioning was the same as that described for the main study (section 3.2.2). Subjects were seated with arms crossed in front of the chest and with the hip and knee flexed at 60 degrees. An adjustable seat belt was fastened across the hips to avoid pelvic and hip movements when the knee extended. Resistance was applied at the distal part of the lower leg, just above the ankle joint. The subjects were asked to extend their knees with maximal effort isometrically and sustain for ten seconds until the operator told them to relax. They were allowed to stop if they could no longer hold the contraction. Verbal encouragement was provided to facilitate the generation of maximum muscle torque. The second test was to extend the knee isometrically at either 50 Nm, 100 Nm or 150 Nm and maintain this torque level for 20 second. The last test was to flex the knee maximally and sustain for ten seconds. Verbal encouragement was provided to facilitate the generation of maximum muscle torque.

2. Intramuscular wire (IMW) EMG measurement of the hamstrings muscles

Types of electrode : IMW electrodes were used as the active electrode, surface electrode was used as the ground electrode.

Electrodes placement : A pair of wire electrode was inserted in the muscle belly of the biceps femoris muscle. The site of muscle belly was located by the method recommended by the book "Anatomical Guide for EMG" (Perotto 1994). It was approximately 50 % of the distance from the ischial tuberosity to the lateral femoral condyle (Pincivero et al 2000).

Skin preparation : The skin was scrubbed with detergent and cleaned with alcohol.

EMG acquisition : Nicolet Viking IV system was used to acquire the intramuscular wire EMG.

EMG data processing : The root mean square voltage of the recorded EMG was calculated by the Nicolet Viking IV system.

***Results***

Hamstrings were composed of three muscles. They were the semi-membranosus, semi-tendinosis and biceps femoris. The biceps femoris muscle was selected to represent the hamstrings in this pilot study. A ratio – the hamstrings co-contraction ratio (HC) (equation 3.2) was proposed to document the co-contraction of hamstrings (Draganich et al 1989). The extent of co-contraction of the hamstrings was

represented by the amount of electrical activity (IMW EMG) in the biceps femoris muscle (long head) during various levels of active isometric contraction of the quadriceps muscles.

$$\textbf{HC (Hamstring co-contraction)} = \frac{\text{IMW EMG of biceps femoris at knee extension (50 Nm/ 100 Nm/ 150 Nm and MVC)}}{\text{IMW EMG of biceps femoris at MVC of knee flexion}} \qquad (3.2)$$

The extent of hamstrings co-contraction in biceps femoris muscle of the three subjects was examined. The results were illustrated in table 3.3. The IMW EMG (in $\mu$V) was acquired during the $6^{th}$ to $8^{th}$ seconds of every muscle contraction. It is shown that hamstrings co-contraction (HC) did not exceed 6 % even at vigorous quadriceps contraction.

Table 3.3. Summary of IMW EMG (6th to 8th second) of the right biceps femoris muscle during different levels of contraction. HC is the ratio between IMW EMG of the BF muscle acquired at various torque levels to the IMW EMG acquired at MVC. Subjects A & B are male, subject C is female.

| | Subject A | Subject B | Subject C |
|---|---|---|---|
| IMW EMG at MVC of Quadriceps (HC) | 12 μV (2.8 %) | 11 μV (4.6 %) | 19 μV (5.1 %) |
| IMW EMG at 50Nm contraction of quadriceps (HC) | NA | NA | 7 μV (1.9%) |
| IMW EMG at 100Nm contraction of quadriceps (HC) | 13 μV (3.0 %) | 10 μV (4.2 %) | 16 μV (4.3 %) |
| IMW EMG at 150Nm contraction of quadriceps (HC) | 9 μV (2.1 %) | 14 μV (5.9 %) | NA |
| IMW EMG at MVC of hamstrings | 435 μV | 237 μV | 375 μV |

- NA – non-applicable

***Conclusion***

A pilot study was carried out to document the influence of hamstrings co-contraction on the measurement of torque at MVC. The biceps femoris muscle was taken to be representative of the hamstrings in this pilot study. Intramuscular wire EMG, which indicates the amount of muscle activity, was acquired. Results showed that the biceps femoris muscle was not activated for more than 6 % of its maximal activity during maximal isometric contraction of the quadriceps muscles even in the

untrained subjects that participated in this pilot study. It was concluded that the co-contraction of the hamstrings had a negligible effect on the measurement of the extension torque of the knee at maximal isometric contraction. It was consistent with the findings in Draganich's study (1989). Draganich and associates found that coactivation of the three hamstrings muscles occur only during the terminal phase of extension (0° to 9° of knee flexion). The EMG activity acquired in the hamstrings at the other range of knee flexion was not more than the expected cross-talk value (3 %). Draganich further commented that coactivation occurred usually during simultaneous movements of both the knee and hip and during high-velocity movements. This study only involved isolated isometric extension of the knee. It might be the reason why co-contraction was not significant with present test protocols.

Although the result of this pilot study showed that co-contraction of hamstrings was low (less than 6 %) initially, it may be more informative if the changes of co-contraction among subjects can be monitored longitudinally with the strengthening program. The results may be useful in explaining the possible contribution of inhibition antagonist in the improvement of muscle strength. Hakkinen and associates (1998) did a study in this area. They traced the activity of the biceps femoris (long head) muscle during a 6-month resistance-training program of knee extensors in middle-aged and elderly men and women. It was found that there was no significant

change in the IEMG of biceps femoris in the middle-aged group although there was significant decrease in the IEMG of biceps femoris in the elderly (fig. 2.1). The result of this pilot study agrees with that of Hakkinen's study.

## 3.5 Signal Processing

High quality signals on torque and surface electromyography (EMG) can be acquired if the suitable methodology and reliable instrumentation are employed. Moreover the confounding variables have to be controlled. These raw signals have to be further processed in a standard way so that data collected in different sessions can be compared.

### 3.5.1 Signal Processing of the Torque

Digital data of 10 points per second were recorded by the dynamometer. It was suggested by De Luca (1997) that the mean of the peak torque (in one second) should be taken as the torque at MVC. However this protocol was modified in this study. It was because both the torque and the corresponding EMG at MVC would be measured and analyzed in this study. The EMG data in time domain would be transformed to power density spectrum by Fast Fourier Transform (FFT). In order to reduce the zero padding effect during FFT, the number of data points for FFT should be equal to the power of two. The sampling frequency was 5 kHz. Therefore 0.81 second, which was equivalent to 4096 data points, were used in this study.

### 3.5.2 Signal Processing of the Electromyography

Surface EMG was recorded on the right rectus femoris muscle at MVC and during sustained contraction at two submaximal levels of contraction, 100 Nm and 150 Nm. The raw EMG was acquired and processed using a PC-based signal processing software (Global Lab, Data Translation) to calculate the MF, IEMG and RMS value. The EMG was full wave rectified before IEMG was derived. Fast Fourier Transform (FFT) was used for spectral analysis. The FFT size was 2048.

A segment of EMG signals had to be selected for subsequent data processing and analysis. Different methods were used in previous studies (Table 3.4). Kamen and associates (1996) commented that "IEMG is sensitive to the time constant. The shorter the time constant, the greater the variability in the output, but shorter time constants are more sensitive to rapid changes in the signal. It has been recommended that the minimum sampling time necessary to obtain stable amplitude measures is 50 – 75 ms". It was decided that two methods would be used to select the appropriate segment of EMG in this study. It would be interesting to know if the selection of the time period of EMG signals did make any difference in the final results. The results in the change of EMG based on these two methods were compared in chapter 4.

The possible advantages and disadvantages of both method are described here. **Method A** used 0.81-second (4096 data points, FFT size = 2048) EMG segment

corresponding to the most stable segment of torque during submaximal contractions (ie. at 100 Nm and 150 Nm) to calculate the IEMG, RMS and MF. **Method B** used an 8-second (7 s to 15 s, 20 FFT) segment of EMG during these submaximal contractions to calculate the IEMG, RMS and MF. These two methods had their advantages and disadvantages. Method A used 0.81-second EMG segment. The advantage of using this method was that there was less variation (less than 5 %) in the torque during this relatively short period of time. However there were several disadvantages. Firstly, stable torque did not necessarily correspond to stable EMG (fig. 3.4) though it is not a common phenomenon. One of the twelve subjects recruited in this study displayed unstable EMG during sustained submaximal isometric contraction. Error might be introduced if a short segment of EMG, either at the peak or at the trough, was chosen. One possible explanation to the discrepancy between the peak torque and maximal EMG is the co-contraction of the hamstrings although it was shown that the effect of the co-contraction of the hamstrings muscles was negligible in the set-up of this study. This has been discussed in details in sections 2.5.5 and 3.4.3. Another explanation to the instability of EMG signal is that the activation pattern of the motor units may be unstable (De Luca 1997). The second disadvantage is that the EMG signals selected did not come from a fixed period of time, fatigue might be an uncontrolled variable.

Method B used a longer (7 s to 15 s) segment of EMG. The advantage of this method was that a longer segment of EMG compensated for the instability of EMG throughout the 20 seconds of submaximal contraction. Another advantage is that the EMG signals were selected from a fixed time period. The effect of fatigue was similar in all cases. However, although the subjects tried to maintain the stability in torque, the variation in torque might still be more than the torque used in method A. The variation in torque was between 5 to 10% when method B was used. This amount of variation was still considered acceptable in other study (Mannion et al 1996).

Table 3.4. Summary of the data processing method of EMG

| Author | Muscle studied | EMG parameters | Time segment chosen |
|---|---|---|---|
| Weir et al 1994 | Vastus lateralis | Maximal IEMG | 5-second integration period |
| Narici et al 1996 | Vastus lateralis | Maximal IEMG | 1-second integration period |
| Mannion et al 1996 | Vastus lateralis, rectus femoris | MF | 1-second sampling period |
| Hakkinen et al 1998 | Vastus lateralis, vastus medialis, rectus femoris | Maximal IEMG | EMG was integrated for the periods of 100 ms to 500 ms and then normalized for one second. |
| Rabita et al 2000 | Vastus lateralis, vastus medialis, rectus femoris | Maximal RMS | 2-second window |
| Pincivero et al 2000 | Vastus lateralis, vastus medialis, rectus femoris | IEMG | Middle 3 second of the 5 second contraction |

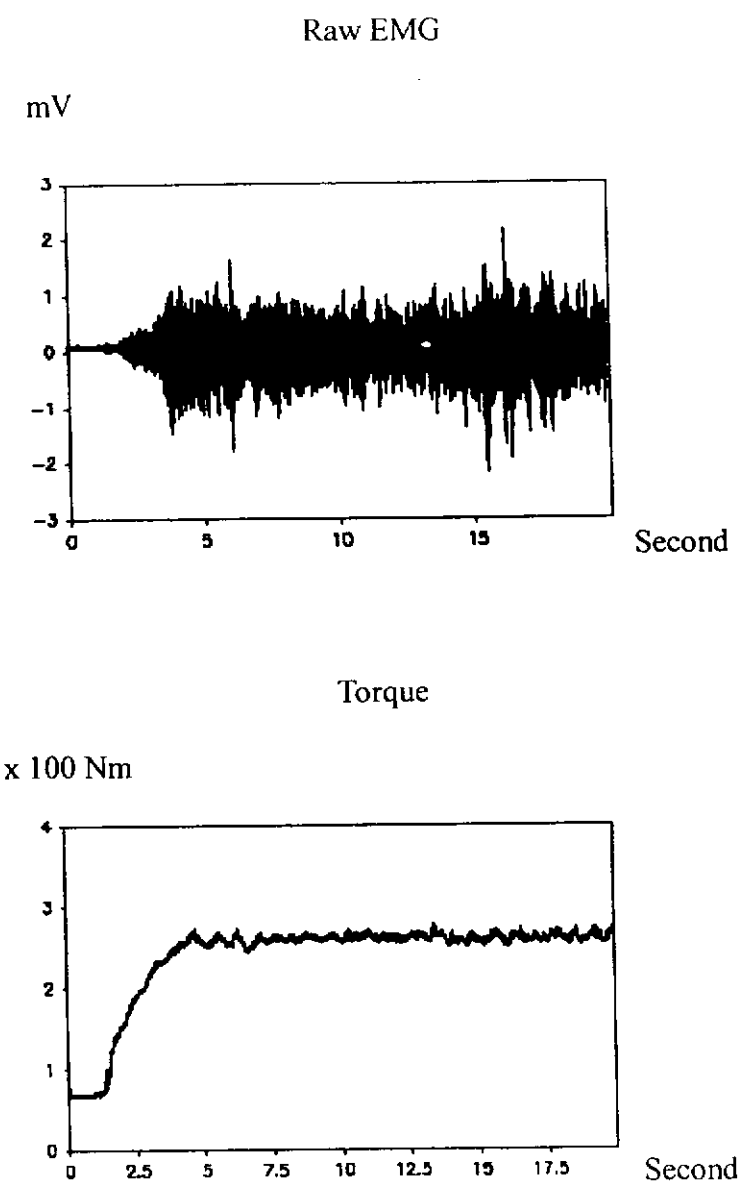



Figure 3.4. An example to illustrate that a stable segment of torque did not mean that the electromyography was stable in a 20-second isometric contraction of the right quadriceps. The EMG was acquired on the rectus femoris muscle. The torque of knee extension was acquired simultaneously in another channel.

### 3.5.3 Double Normalization

In order to compare the EMG parameters in the time-domain (IEMG, RMS value) between sessions, the data should be normalized. The EMG data was normalized twice (double normalization) in this study. It is because surface EMG was acquired during the tests at two pre-determined submaximal levels of torque, 100 and 150 Nm, of isometric knee extension. There are two steps in the normalization process of this study. Firstly, the EMG data was normalized with the EMG obtained at MVC in the same session. Secondly, since the muscle gained its strength during the six weeks of strengthening, the EMG data was then further normalized by a factor, which is the ratio of the torque at MVC at the current session to that in the initial session. This double normalization procedure allows comparison of EMG data between two days even with a pre-determined fixed torque level test protocol. Section 3.5.2 has already described the two methods (Method A and Method B) of EMG signal processing in details. Examples of how the raw EMG signals were processed using double normalization procedures and these two signal processing methods in this study will be given below.

***EMG data processing using method A and double normalization***

In method A, the EMG signal (0.81 second) corresponding to the maximum torque of knee extension at MVC was used. 0.81-second EMG segment during the most stable period of the torque was also taken at submaximal levels (100 Nm and 150 Nm) of the knee extension. In order to compare the integrated EMG (IEMG) and RMS voltage at 100 Nm and 150 Nm levels of contraction during the initial, $2^{nd}$, $4^{th}$ and $6^{th}$ weeks, these EMG data were doubly normalized. First, the raw EMG data (table 3.4 (a)) was normalized with the EMG obtained at MVC within the same session. This is shown in table 3.4 (b). This normalized data were then further normalized with the relative change in the torque produced at MVC between consecutive sessions. This is shown in table 3.4 (c). The normalized data at the initial, $2^{nd}$, $4^{th}$ and $6^{th}$ weeks of the strengthening program will be used for comparison and for subsequent statistical analysis.

Table 3.5 (a). Raw data of subject A. Data were processed using data processing method A. It includes the torque at maximal voluntary contraction (MVC), root mean square (RMS), integrated EMG (IEMG) and median frequency (MF) at maximal voluntary contraction, 100 Nm and 150 Nm levels of contraction.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Torque (Nm) at MVC | 233.2 | 242.1 | 270.8 | 260.4 |
| RMS (mV) at MVC | 0.9615 | 0.7235 | 0.877 | 1.5353 |
| IEMG (mVs) at MVC | 0.6293 | 0.465 | 0.5741 | 0.9876 |
| RMS (mV) at 100 Nm | 0.2735 | 0.1899 | 0.2118 | 0.2365 |
| RMS (mV) at 150 Nm | 0.5536 | 0.4111 | 0.3142 | 0.4464 |
| IEMG (mVs) at 100 Nm | 0.1787 | 0.1392 | 0.1506 | 0.163 |
| IEMG (mVs) at 150 Nm | 0.3291 | 0.2412 | 0.2039 | 0.2665 |

Table 3.5 (b). The raw EMG (IEMG and RMS) data was firstly normalized with the EMG obtained at MVC within the same session. The value shown were normalized root mean square (RMS) and integrated EMG (IEMG) at 100 Nm and 150 Nm levels of contraction.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Torque (Nm) at MVC | 233.2 | 242.1 | 270.8 | 260.4 |
| RMS (mV) at 100 Nm | 0.2845 | 0.2625 | 0.2415 | 0.154 |
| RMS (mV) at 150 Nm | 0.5758 | 0.5682 | 0.3583 | 0.2908 |
| IEMG (mVs) at 100 Nm | 0.284 | 0.2994 | 0.2623 | 0.165 |
| IEMG (mVs) at 150 Nm | 0.523 | 0.5187 | 0.3552 | 0.2698 |

Table 3.5 (c). Values shown in figure 3.4 (b) were further normalized with the relative change in the torque at MVC (double normalization). RMS and IEMG at 150 Nm contraction show a decreasing trend.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Torque ratio at MVC between sessions | 1 | 1.04 | 1.16 | 1.12 |
| RMS (mV) at 100 Nm | 0.2845 | 0.273 | 0.2801 | 0.1725 |
| RMS (mV) at 150 Nm | 0.5758 | 0.5909 | 0.4156 | 0.3257 |
| IEMG (mVs) at 100 Nm | 0.284 | 0.3114 | 0.3043 | 0.1848 |
| IEMG (mVs) at 150 Nm | 0.523 | 0.5394 | 0.412 | 0.3022 |

***EMG data processing using method B and double normalization***

As mentioned in section 3.5.2, data processing method B involved the use of EMG corresponding to the maximum torque at MVC. For 100 Nm and 150 Nm contractions, 8 seconds (7 s to 15 s) of EMG segment was used. The double normalization process was exactly the same. The only difference is that data processing method B was used in managing the EMG data obtained at 100 Nm and 150 Nm levels of contraction. The raw EMG data was first normalized with the EMG obtained at MVC of that session. This is shown in table 3.5 (b). This normalized data was then further normalized with the change in ratio of the MVC. This is shown in table 3.5 (c) and these data at the initial, $2^{nd}$, $4^{th}$ and $6^{th}$ weeks of the strengthening program were used for comparison and for subsequent statistical analysis.

Table 3.6 (a). Raw data of subject A. However, data were processed using data processing method B. It includes the torque at maximal voluntary contraction (MVC), root mean square (RMS) and integrated EMG (IEMG) at maximal voluntary contraction, 100 Nm and 150 Nm levels of contraction

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Torque (Nm) at MVC | 233.2 | 242.1 | 270.8 | 260.4 |
| RMS (mV) at MVC | 0.9615 | 0.7235 | 0.877 | 1.5353 |
| IEMG (mVs) at MVC | 0.6293 | 0.465 | 0.5741 | 0.9876 |
| RMS (mV) at 100 Nm | 0.2518 | 0.2295 | 0.1965 | 0.2464 |
| RMS (mV) at 150 Nm | 0.5383 | 0.3544 | 0.3463 | 0.414 |
| IEMG (mVs) at 100 Nm | 1.6098 | 1.5107 | 1.3521 | 1.6124 |
| IEMG (mVs) at 150 Nm | 2.6298 | 2.1942 | 2.1772 | 2.4801 |

Table 3.6 (b). The raw EMG data (IEMG and RMS) were first normalized with the EMG obtained at MVC within the same session. The values shown were normalized root mean square (RMS) and integrated EMG (IEMG) at 100 Nm and 150 Nm levels of contraction

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Torque (Nm) at MVC | 233.2 | 242.1 | 270.8 | 260.4 |
| RMS (mV) at 100 Nm | 0.2619 | 0.3172 | 0.2241 | 0.1605 |
| RMS (mV) at 150 Nm | 0.5599 | 0.4898 | 0.3949 | 0.2697 |
| IEMG (mVs) at 100 Nm | 2.5581 | 3.2488 | 2.3552 | 1.6326 |
| IEMG (mVs) at 150 Nm | 4.1789 | 4.7187 | 3.7924 | 2.5112 |

Table 3.6 (c). Data in figure 3.5 (b) were further normalized with the change in ratio of the MVC (double normalization). RMS and IEMG at 150 Nm also show a decreasing trend.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Torque ratio at MVC between sessions | 1 | 1.04 | 1.16 | 1.12 |
| RMS (mV) at 100 Nm | 0.2619 | 0.3299 | 0.26 | 0.1798 |
| RMS (mV) at 150 Nm | 0.5599 | 0.5094 | 0.4581 | 0.3021 |
| IEMG (mVs) at 100 Nm | 2.5581 | 3.3779 | 2.732 | 1.8285 |
| IEMG (mVs) at 150 Nm | 4.1789 | 4.9074 | 4.3992 | 2.8125 |

### 3.6 Statistical Analysis

"Friedman two-way analysis of variance by ranks" (Friedman test) was used to test whether there was any significant difference in various EMG parameters during the six-week strengthening program. Although all data were interval/ ratio data, non-parametric test was used for statistical analysis. It was because the sample size (n = 11) was relatively small. The significance level was set at 95 %. These EMG parameters were IEMG, RMS and MF at torque levels of 100 and 150 Nm. If Friedman test revealed significant difference, the "Wilcoxon signed ranks test" (Wilcoxon test) with Bonferroni's Correction would be used to test the EMG data between the initial and 2$^{nd}$ weeks, between the initial and 4$^{th}$ weeks, and between the initial and 6$^{th}$ weeks. This helped to identify when the results showed statistically significant difference.

## 3.7 Summary

The objective of the present study was to find out the change of EMG of the rectus femoris of the quadriceps during the strengthening period of the quadriceps muscle in normal healthy subjects. Twelve normal, healthy, lean and young male subjects were recruited for this study. The subjects were engaged in a scheduled strengthening program of 30 MVC per day, 3 days a week, for 6 weeks. The torque and surface EMG at MVC and the surface EMG at two pre-determined torque levels of submaximal contraction, 100 Nm and 150 Nm, were recorded. The techniques used to control several confounding variables and the results were presented. These confounding variables were skin impedance, cross-talk and hamstrings co-contraction. Special techniques on signal processing including double normalization of the EMG data were reported. This normalization procedure allowed the comparison of EMG data between sessions. The results of this study are presented in chapter 4.

# Chapter 4 Results

The subjects recruited in the main study were to be involved in a scheduled strengthening program of the right quadriceps muscles. The exercise and the test were done at 60° knee flexion. The program consisted of 30 maximal voluntary contractions (MVC) per day, 3 days a week, for 6 weeks. They were then tested of the torque generated and electromyography (EMG) acquired at the MVC of the right quadriceps. Therefore an accurate measurement of the torque at MVC was very important. A reliability test on the measurement of torque was conducted before the main study started. Its result is presented first in the following sessions.

## 4.1 Reliability of the Measurement of Torque at Maximal Voluntary Contraction

The method described by de Luca (1997) was adopted to measure the torque at MVC. A reliability test of this method using the set-up in this study was conducted. Seven male subjects were recruited to test whether the torque at MVC of the right quadriceps measured with the method described in section 3.3.1 was reliable. Two consecutive measurements with three days in between were taken. The demographic characteristics of the subjects in this pilot study were comparable to the subjects of the main study (table 4.1). Table 4.2 shows the result of the measurement of torque. The ntra-class correlation coefficient (ICC) was 0.86. It was concluded that the method

used for measuring the torque at MVC in this study was reliable.

Table 4.1 Demographic data of the subjects in the reliability test of MVC

| | |
|---|---|
| Number of subjects | 6 |
| Gender | Male |
| Age | 19.3 (Range: 19-20) |
| Dominant side | Right |
| Occupation | Students |

Table 4.2 Results of the measurement of torque at MVC

| Subject | Torque at MVC on the first day (Nm) | Torque at MVC three days later (Nm) |
|---|---|---|
| 1 | 225.1 | 215.8 |
| 2 | 237.8 | 259.8 |
| 3 | 214.0 | 216.8 |
| 4 | 307.4 | 321.1 |
| 5 | 240.0 | 288.0 |
| 6 | 249.4 | 276.2 |
| 7 | 263.9 | 272.2 |

## 4.2 Demographic Data

Twelve male subjects completed the six-week strengthening program. The data of one of the subjects was not included in data analysis. The reason is explained in section 5.2. None of them had past history of knee injury or chronic knee problem. None of them had knee pain at the time of this study. The detailed demographic data are shown in table 4.3. The results of the change in muscle strength and the changes in surface EMG parameters are reported in this chapter.

Table 4.3. The demographic data of 11 subjects

| Subject characteristics | Subjects included in the analysis (n = 11) | The subject excluded from the data analysis |
|---|---|---|
| Gender | Male | Male |
| Age | Mean: 23 (Range: 19 to 35) | 25 |
| Occupation | Physiotherapy students and physiotherapists | Physiotherapy student |
| Skin impedance | Below 5 kΩ | Below 5 kΩ |
| Body mass index<br>$BMI = \frac{\text{Body weight (Kg)}}{\text{Body height}^2 (m^2)}$ | Mean: 22.2 (Range: 17.9 to 26.1)<br>Body weight is stable (< 2.5 kg difference) during study | 20.3<br>Body weight is stable (< 2.5 kg difference) during study |
| Previous injury on right leg/ right knee pain | Nil | Nil |

## 4.3 Effects of the Strengthening Program

All eleven subjects showed an increase in the strength of their right quadriceps femoris muscle. The increase ranged from 11 % to 40 % with a mean of 22 % (table 4.4 and fig. 4.1). As mentioned in chapter 3, all EMG data were extracted when the torque was at its maximal level. The duration of EMG signals was 0.81 second, which corresponded to 4096 data points. Friedman two-way analysis of variance by ranks (Friedman) was carried out. There was significant difference in the torque at MVC of the right quadriceps muscles after the strengthening program ($p = 0.001$). It is interesting to see whether the changes in EMG have any relationship with the change in the muscle strength at MVC.

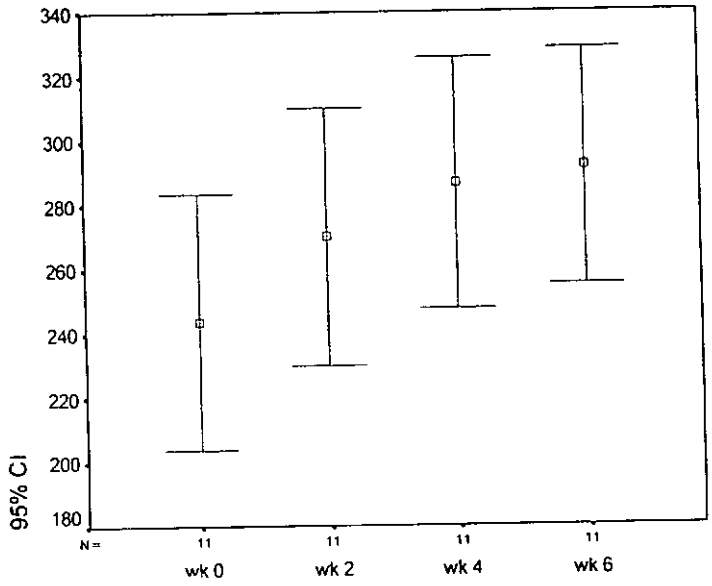



Figure 4.1. Change in the torque at maximal voluntary contraction of the right quadriceps muscles in a six-week strengthening program of 11 subjects.

Table 4.4 (a). The change in the torque at maximal voluntary contraction of the right quadriceps muscles in 11 subjects during the six-week strengthening program.

| Week | N | Minimum | Maximum | Mean | S.D. |
|---|---|---|---|---|---|
| 0 | 11 | 159 | 344 | 243.7 | 59.3 |
| 2 | 11 | 177 | 372 | 270.1 | 59.9 |
| 4 | 11 | 200 | 371 | 287.0 | 58.3 |
| 6 | 11 | 221 | 389 | 292.0 | 54.6 |

Table 4.4 (b). The relative change in the torque at maximal voluntary contraction of the right quadriceps muscles in 11 subjects during the six-week strengthening program.

| Week | N | Minimum | Maximum | Mean | S.D. |
|---|---|---|---|---|---|
| 0 | 11 | 1.00 | 1.00 | 1.0000 | 0.0000 |
| 2 | 11 | 1.04 | 1.32 | 1.1164 | 0.08 |
| 4 | 11 | 1.08 | 1.38 | 1.1873 | 0.09 |
| 6 | 11 | 1.11 | 1.40 | 1.2155 | 0.1147 |

Table 4.4 (c). The baseline muscle strength and the changes of muscle strength after the 6-week strengthening program with respect to different age group

| Age group | N | Baseline muscle strength | S.D. | Relative change in muscle power |
|---|---|---|---|---|
| 20 – 24 | 8 | 255 | 62 | 21 % |
| 25 – 29 | 2 | 205 | 65 | 28 % |
| 30 – 35 | 1 | 233 | Non-applicable | 12 % |
| 25 | The subject was excluded from data analysis | 231 | Non-applicable | 22 % |

Figure 4.2 shows the change in integrated EMG (IEMG), root mean square (RMS) and median frequency (MF) at maximal voluntary contraction of the right quadriceps during the six-week strengthening program. There was a slight increase in IEMG and RMS but the result was not statistically significant. MF tended to decline when muscle strength was increased. However the decline was not statistically significant neither.

Max IEMG
(mVs)

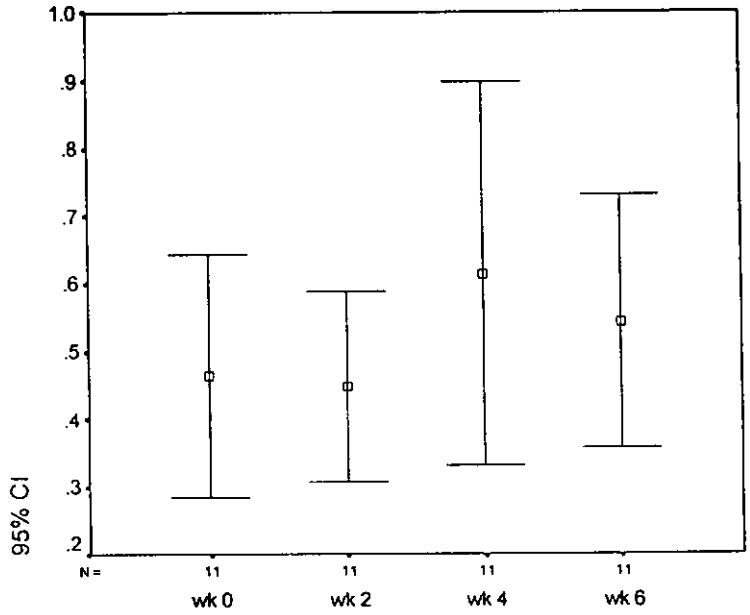



Figure 4.2 (a). Change in IEMG at maximal voluntary contraction of the right quadriceps in the six-week strengthening program of 11 subjects.

RMS (mV)

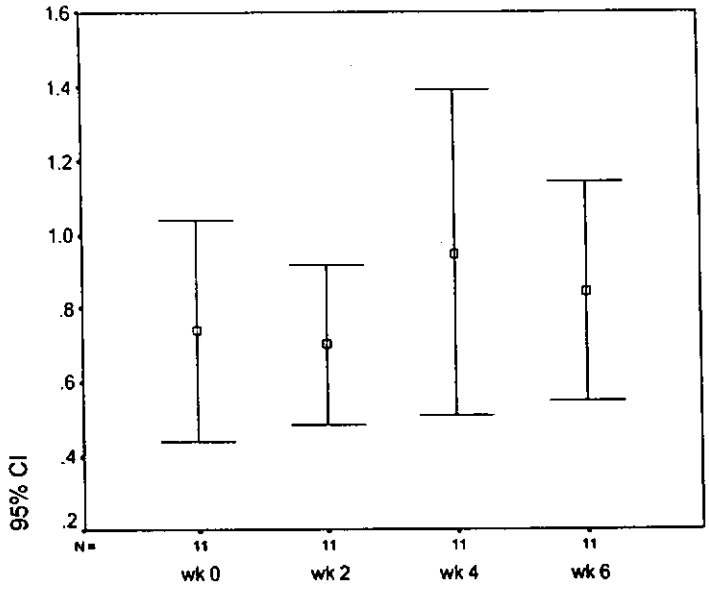



Figure 4.2 (b). Change in RMS voltage at maximal voluntary contraction of the right quadriceps in the six-week strengthening program of 11 subjects.

MF (Hz)

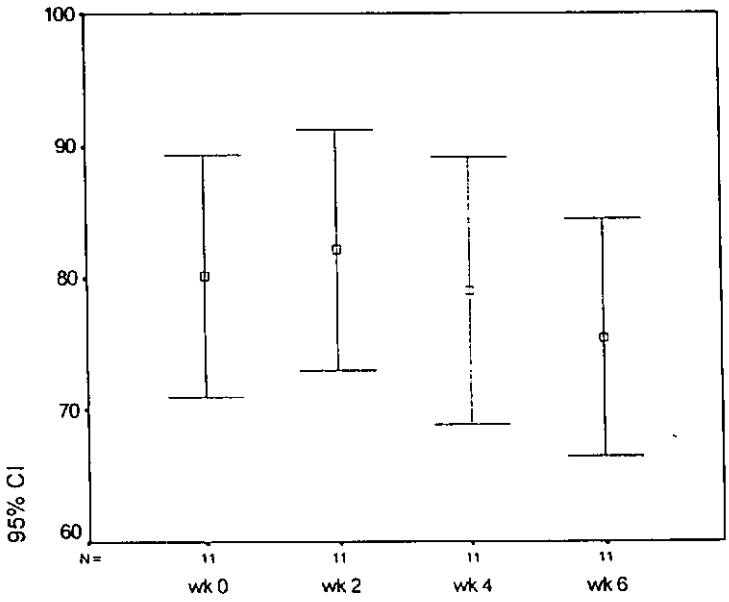



Figure 4.2 (c). Change in MF at maximal voluntary contraction of the right quadriceps muscles in the six-week strengthening program of 11 subjects.

## 4.4 Change of Electromyography During Strengthening

As mentioned in section 3.5.2, a segment of the EMG signals at 100 Nm and 150 Nm levels of contraction were selected using two different methods so that further data processing could be done. The advantages and disadvantages of these two methods have also been discussed. **Method A** used 0.81-second EMG segment corresponding to a stable segment of torque during submaximal levels of contraction, ie. at 100 Nm and 150 Nm. **Method B** used an 8-second segment of EMG during these submaximal levels of contraction to calculate the IEMG, RMS and MF. Therefore the results of the changes in EMG, which was processed using both methods, would be presented in this chapter. The difference in the results would be further discussed in section 4.4.

***Results using data processing method A***

There was no obvious trend in the change of IEMG and RMS at MVC as shown in section 4.3. However, the IEMG and RMS at torque level of 150 Nm showed a consistently decreasing trend in most of the subjects when the muscle strength increased during the six-week strengthening program. The changes in IEMG and RMS at 100 Nm torque level were not so obvious. The EMG data, which were processed using method A, of the subjects X and Y were shown in tables 4.5 (a) and (b). Double normalization of the data had been done. These two subjects behaved in a

similar way to all other subjects as shown in table 4.7. No consistent trend was shown in the MF.

Table 4.5 (a). Doubly normalized data of subject X during the six-week strengthening program. The EMG data were processed using method A.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Relative change of the torque at MVC between sessions | 1 | 1.11 | 1.26 | 1.39 |
| RMS (mV) at 100 Nm | 0.6018 | 0.5624 | 0.546 | 0.6034 |
| RMS (mV) at 150 Nm | 1.206 | 0.9444 | 0.8216 | 0.8407 |
| IEMG (mVs) at 100 Nm | 0.7474 | 0.8265 | 0.7288 | 0.7738 |
| IEMG (mVs) at 150 Nm | 1.1486 | 1.0018 | 0.901 | 0.9242 |

Table 4.5 (b). Doubly normalized data of subject Y during the six-week strengthening program. The EMG data were processed using method A.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Relative change of the torque at MVC between sessions | 1 | 1.09 | 1.12 | 1.11 |
| RMS (mV) at 100 Nm | 0.2926 | 0.179 | 0.1863 | 0.1764 |
| RMS (mV) at 150 Nm | 0.457 | 0.3063 | 0.3401 | 0.2714 |
| IEMG (mVs) at 100 Nm | 0.3395 | 0.1865 | 0.1934 | 0.1854 |
| IEMG (mVs) at 150 Nm | 0.4742 | 0.2918 | 0.3261 | 0.2559 |

***Results using data processing method B***

Tables 4.6 (a) and (b) showed the doubly normalized data of the two subjects using data processing method B. IEMG and RMS at the torque level of 150 Nm again showed consistently decreasing trends in most of the subjects. Changes in IEMG and RMS at 100 Nm torque level were not so obvious. It seemed that the trend in EMG changes did not change significantly no matter which method for selecting the EMG segment was used. EMG changes of all the subjects during the six-week strengthening program are presented in detail in section 4.4.1. The results of statistical analysis are reported in section 4.4.2.

Table 4.6 (a). Doubly normalized data of the subject X during the six-week strengthening program. The EMG data were processed using method B.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Relative change of the torque at MVC between sessions | 1 | 1.11 | 1.26 | 1.39 |
| RMS (mV) at 100 Nm | 0.5829 | 0.549 | 0.5097 | 0.5557 |
| RMS (mV) at 150 Nm | 1.1052 | 0.963 | 0.8655 | 0.866 |
| IEMG (mVs) at 100 Nm | 7.2238 | 8.0365 | 6.8095 | 7.0581 |
| IEMG (mVs) at 150 Nm | 10.5306 | 10.0311 | 8.9572 | 8.9883 |

Table 4.6 (b). Doubly normalized data of the subject Y during the six-week strengthening program. The EMG data were processed using method B.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Relative change of the torque at MVC between sessions | 1 | 1.09 | 1.12 | 1.11 |
| RMS (mV) at 100 Nm | 0.2776 | 0.173 | 0.1943 | 0.1739 |
| RMS (mV) at 150 Nm | 0.4491 | 0.3162 | 0.3345 | 0.2566 |
| IEMG (mVs) at 100 Nm | 3.1635 | 1.7869 | 1.9256 | 1.7858 |
| IEMG (mVs) at 150 Nm | 4.5322 | 2.975 | 3.1251 | 2.4602 |

### 4.4.1 Descriptive Statistics

***Results using data processing method A***

The descriptive statistics of 11 subjects with the six-week strengthening program is shown in table 4.7. The EMG data were processed with method A. The changes in the mean of IEMG and RMS are illustrated in figure 4.3. Both IEMG and RMS showed decreasing trends at 100 Nm and 150 Nm torque levels. This decreasing trend was more obvious in 150 Nm. However there was no obvious change in the MF at both 100 Nm and 150 Nm with the six-week strengthening.

Table 4.7 (a). Descriptive statistics of 11 subjects showing the relative change of torque at MVC of the right quadriceps muscles.

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 1.00 | 1.00 | 1.0000 | 0 |
| 2 | 1.04 | 1.32 | 1.1164 | 0.082 |
| 4 | 1.08 | 1.38 | 1.1873 | 0.091 |
| 6 | 1.11 | 1.40 | 1.2155 | 0.1147 |

Table 4.7 (b). Descriptive statistics of 11 subjects showing the change of IEMG (in mVs) at 100 Nm and 150 Nm torque levels of isometric extension of the right knee. The EMG data were processed using method A.

100 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 0.1128 | 0.7474 | 0.3714 | 0.1819 |
| 2 | 0.1865 | 0.8265 | 0.3489 | 0.1852 |
| 4 | 0.1069 | 0.7288 | 0.3199 | 0.1784 |
| 6 | 0.1653 | 0.7738 | 0.3348 | 0.1900 |

150 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 0.2035 | 1.1486 | 0.5957 | 0.2896 |
| 2 | 0.2918 | 1.0018 | 0.5222 | 0.2307 |
| 4 | 0.1869 | 09010 | 0.4742 | 0.2211 |
| 6 | 0.2559 | 09242 | 0.4733 | 0.2304 |

Table 4.7 (c). Descriptive statistics of 11 subjects showing the change of RMS voltage (in mV) at 100 Nm and 150 Nm torque levels of isometric extension of the right knee. The EMG data were processed using method A.

100 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 0.0960 | 0.6018 | 0.3237 | 0.1607 |
| 2 | 0.1333 | 0.5624 | 0.2620 | 0.1169 |
| 4 | 0.0870 | 0.5460 | 0.2733 | 0.1353 |
| 6 | 0.1396 | 0.6043 | 0.2827 | 0.1518 |

150 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 0.1847 | 1.2060 | 0.5888 | 0.3108 |
| 2 | 0.2520 | 0.9444 | 0.4947 | 0.2174 |
| 4 | 0.1771 | 0.8216 | 0.4582 | 0.2061 |
| 6 | 0.2374 | 0.8407 | 0.4485 | 0.2089 |

Table 4.7 (d). Descriptive statistics of 11 subjects showing the change of MF (in Hz) at 100 Nm and 150 Nm torque levels of isometric extension of the right knee. The EMG data were processed using method A.

100 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 63.4766 | 100.0977 | 77.9031 | 11.1588 |
| 2 | 58.5938 | 97.6563 | 72.7983 | 11.3371 |
| 4 | 63.4766 | 97.6563 | 76.5714 | 10.9332 |
| 6 | 63.4766 | 95.2148 | 76.5714 | 10.8236 |

150 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 61.0352 | 104.9805 | 80.3445 | 16.0263 |
| 2 | 65.9180 | 100.0977 | 81.6761 | 11.4701 |
| 4 | 65.9180 | 107.4219 | 78.5689 | 12.7243 |
| 6 | 68.3594 | 114.7461 | 81.0103 | 14.0595 |

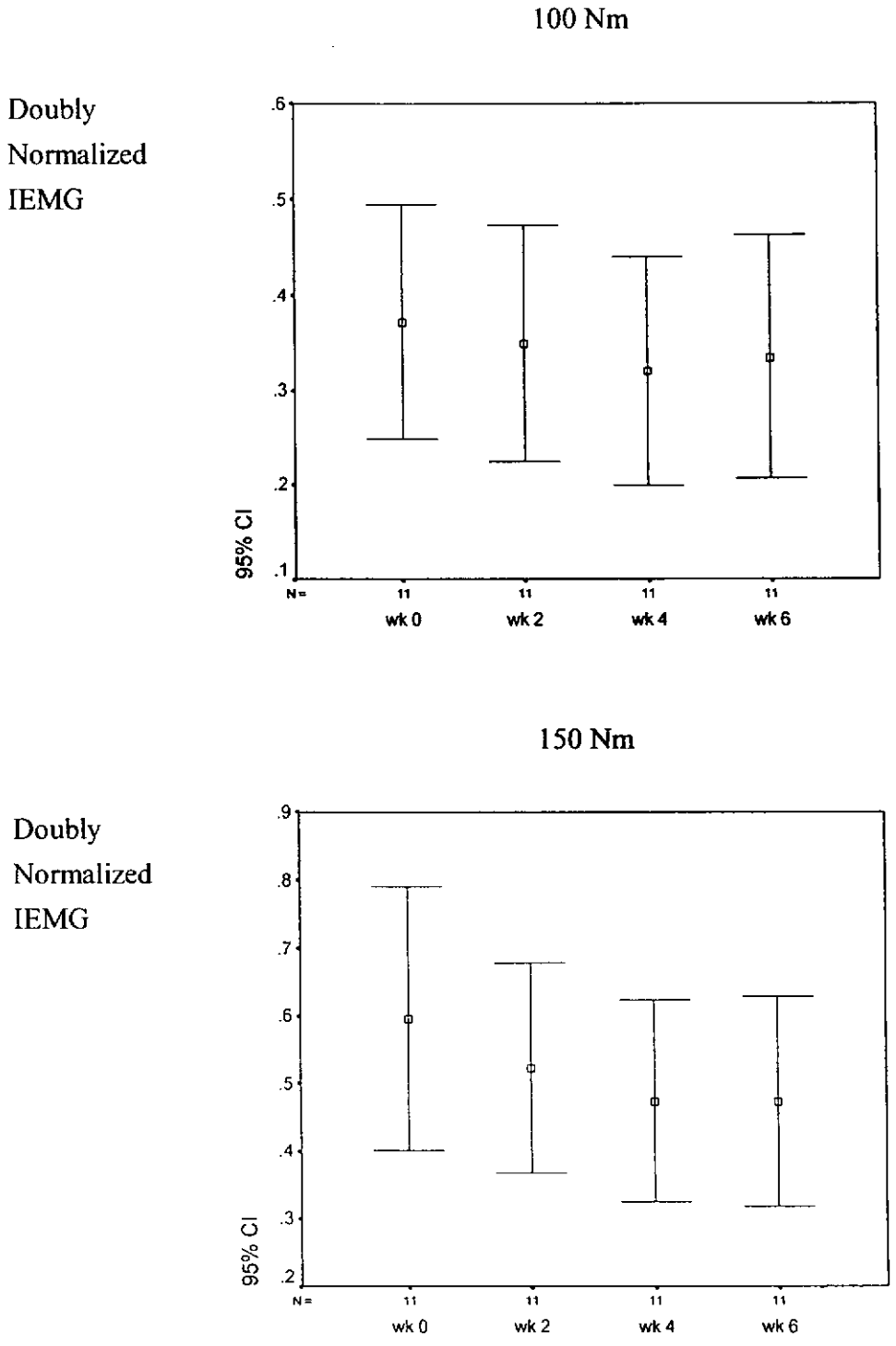
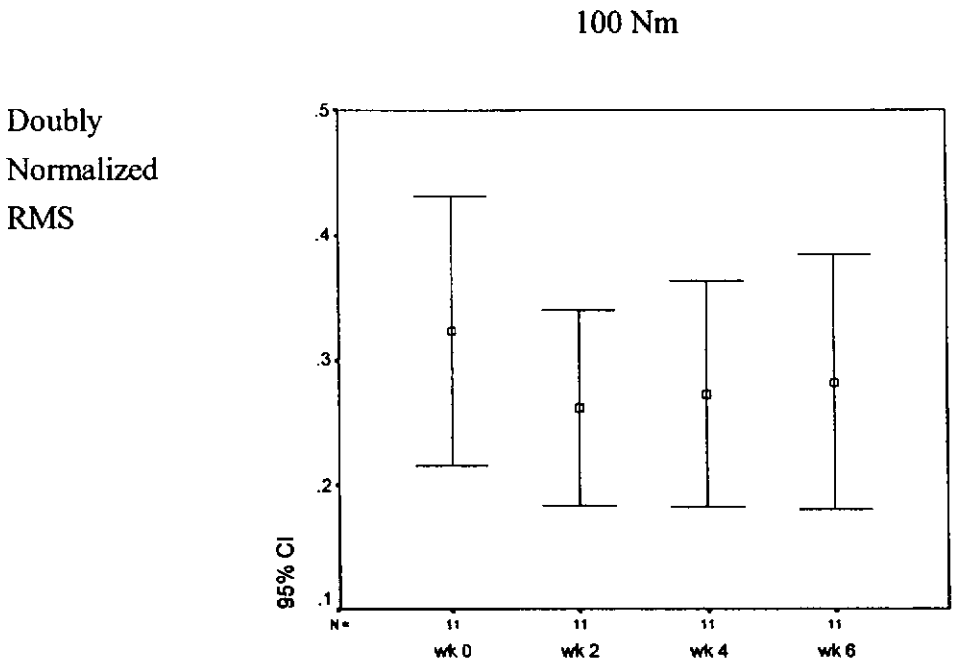



Figure 4.3 (a). Change of IEMG at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening program in 11 subjects. The EMG data were processed using method A.



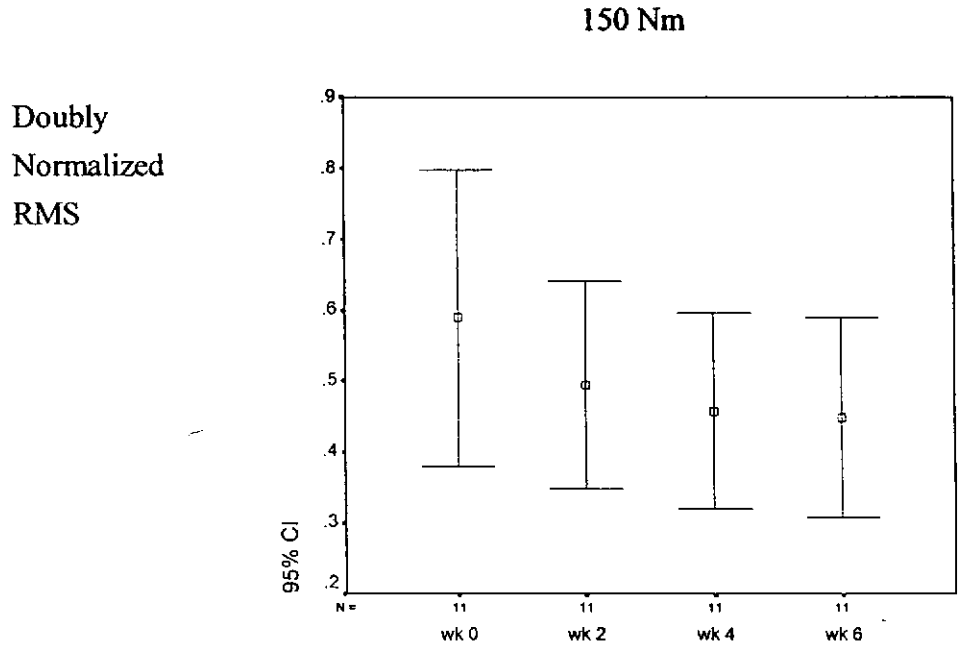




Figure 4.3 (b). Change of RMS voltage at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening program in 11 subjects. The EMG data were processed using method A.

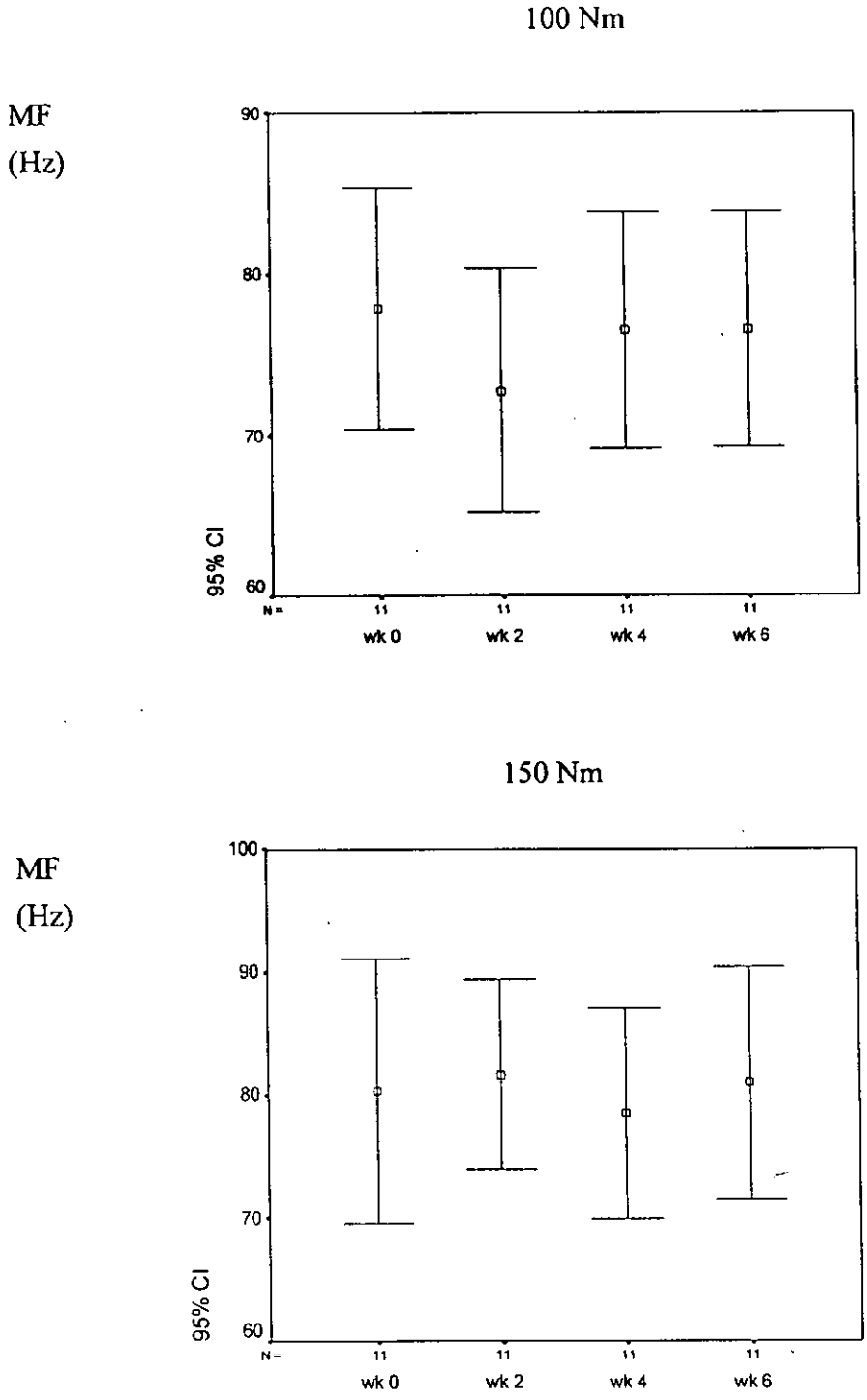



Figure 4.3 (c). Change of MF at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening program in 11 subjects. The EMG data were processed using method A.

***Results using data processing method B***

This section presents the results of the changes in EMG during the strengthening program. The only difference with the previous section was that the EMG were processed using data processing method B. Table 4.8 shows the descriptive statistics of 11 subjects with the six-week strengthening program. Figure 4.4 shows clearly the change in the mean of IEMG and RMS. Both IEMG and RMS showed decreasing trends during 100 Nm and 150 Nm torque levels. This decreasing trend was more obvious at 150 Nm. No significant difference was observed in the trends shown in figures 4.3 and 4.4. It implied that different data processing method (method A and B) concerning the selection of time segment of EMG data did not make significant difference in the results of this study. Method A and B are both acceptable for studies of this nature. This conclusion would be further supported by the results of statistical analysis reported in the next section.

Table 4.8 (a). Descriptive statistics of 11 subjects showing the change of IEMG (in mV.s) at 100 Nm and 150 Nm torque levels of isometric extension of the right knee. The EMG data were processed using method B.

100 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 0.9141 | 7.2238 | 3.4993 | 1.7171 |
| 2 | 1.7869 | 8.0365 | 3.3547 | 1.8256 |
| 4 | 0.9795 | 6.8095 | 3.1120 | 1.6614 |
| 6 | 1.5248 | 7.0581 | 3.1424 | 1.6685 |

150 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 2.1260 | 10.8731 | 5.8006 | 2.8780 |
| 2 | 2.8677 | 10.0311 | 5.1214 | 2.4280 |
| 4 | 1.7821 | 8.9572 | 4.7511 | 2.0909 |
| 6 | 2.4602 | 8.9883 | 4.5680 | 2.1493 |

Table 4.8 (b). Descriptive statistics of 11 subjects showing the change of RMS voltage (in mV) at 100 Nm and 150 Nm torque levels of isometric extension of the right knee. The EMG data were processed using method B.

100 Nm

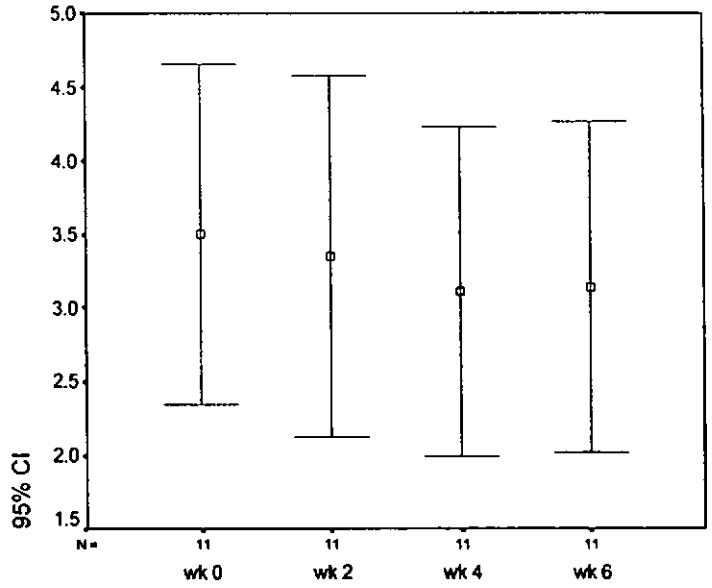
| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 0.0740 | 0.5829 | 0.3189 | 0.1580 |
| 2 | 0.1404 | 0.5490 | 0.2590 | 0.1145 |
| 4 | 0.0764 | 0.5097 | 0.2686 | 0.1265 |
| 6 | 0.1327 | 0.5557 | 0.2658 | 0.1323 |

150 Nm

| Week | Minimum | Maximum | Mean | Std. Deviation |
|---|---|---|---|---|
| 0 | 0.2071 | 1.1354 | 0.6002 | 0.3043 |
| 2 | 0.2453 | 0.9630 | 0.4908 | 0.2323 |
| 4 | 0.1696 | 0.8655 | 0.4749 | 0.2024 |
| 6 | 0.2520 | 0.8660 | 0.4479 | 0.2103 |

100 Nm

Doubly Normalized IEMG




150 Nm

Doubly Normalized IEMG

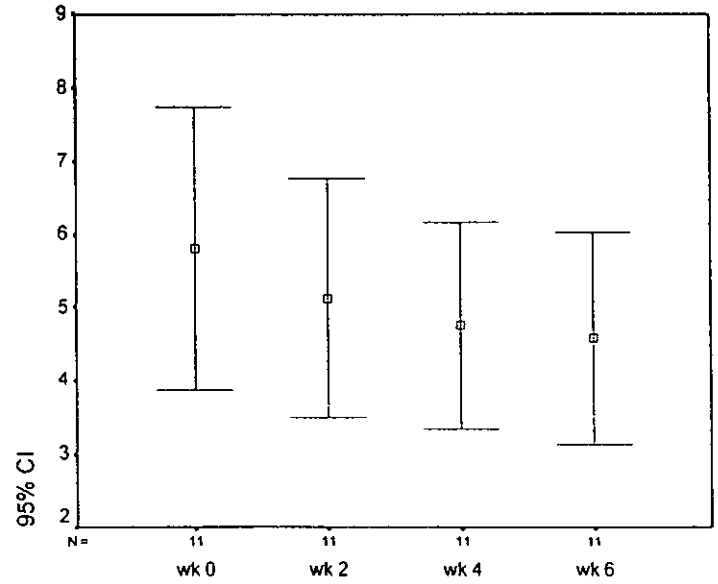



Figure 4.4 (a). Change of IEMG at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening program in 11 subjects. The EMG data were processed using method B.

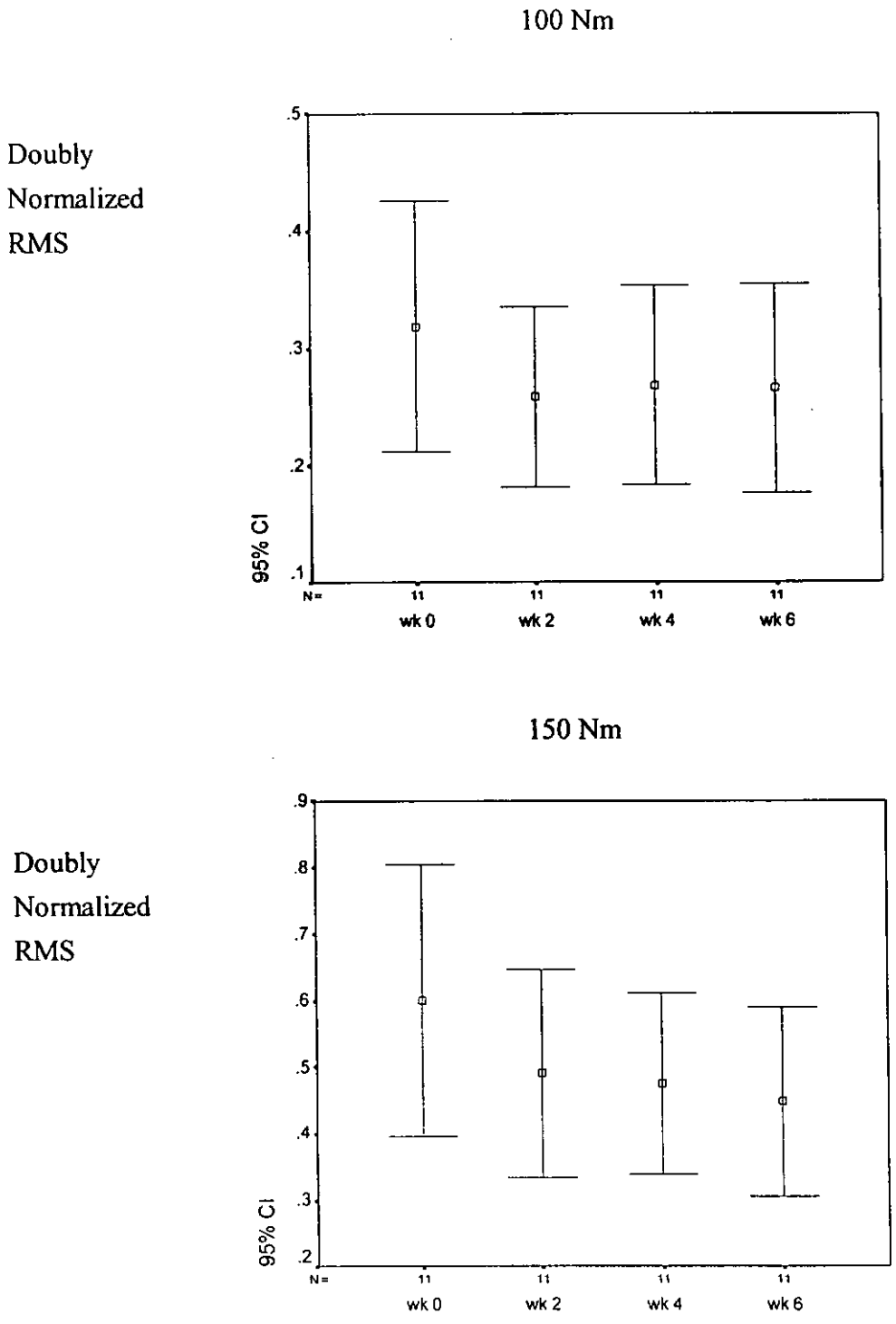



Figure 4.4 (b). Change of RMS voltage at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening program in 11 subjects. The EMG data were processed using method B.

### 4.4.2 The Torque-EMG Relationship During the Six-week Strengthening

If the EMG (in RMS value, processed using method A) is plotted against the corresponding torque level, the torque-EMG curves at week 0 to week 6 are shown. There is a right shift in the torque-EMG curves when the results after strengthening are compared with that before training (fig. 4.5).

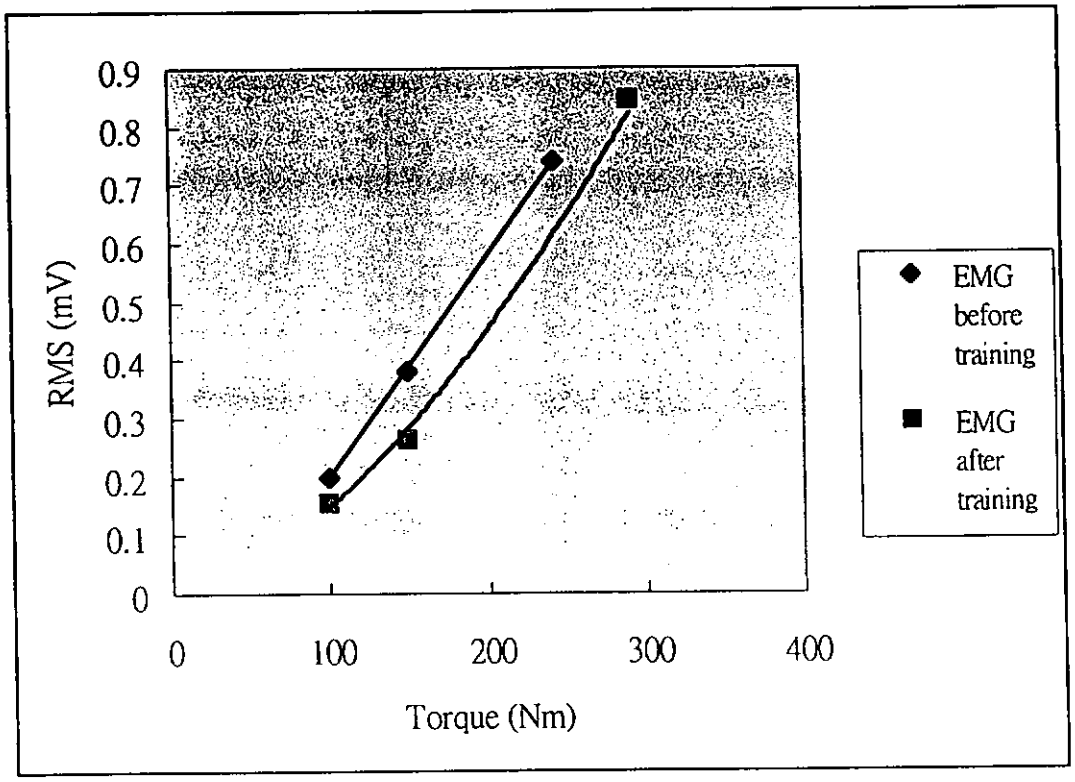



Figure 4.5. Change in the torque-EMG relationship of the right rectus femoris muscle in a 6-week strengthening program (n = 11). The EMG data were processed using method A.

### 4.4.3 Statistical Analysis

Friedman two-way analysis of variance by ranks (Friedman test) was used to test whether there was any significant difference in various EMG parameters during the six-week strengthening program. These EMG parameters were IEMG at 100 Nm and 150 Nm torque levels, RMS at 100 Nm and 150 Nm and MF at 100 Nm and 150 Nm torque levels. If Friedman test revealed significant difference, the Wilcoxon signed ranks test" (Wilcoxon test) with Bonferroni's correction would be used to test which time period showed significant difference. The EMG data between the initial and 2$^{nd}$ weeks, between the initial and 4$^{th}$ weeks and between the initial and 6$^{th}$ weeks would be tested.

***Results using data processing method A***

Tables 4.9 (a) and (b) show that only IEMG and RMS voltage at 150 Nm torque level had statistically significant difference ($p = 0.024$) during the six-week strengthening program. Tables 4.9 (c) and (d) show that there was difference between the initial and the third sessions (at 4$^{th}$ week) for both IEMG and RMS voltage, with $p = 0.008$ and 0.013 respectively.

Table 4.9 (a). IEMG at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening were tested using Friedman two-way analysis of variance by ranks. EMG data were processed using method A.

Ranks

| 100 Nm | | 150 Nm | |
|---|---|---|---|
| Week | Mean Rank | Week | Mean Rank |
| 0 | 2.73 | 0 | 3.36 |
| 2 | 2.73 | 2 | 2.73 |
| 4 | 2.00 | 4 | 2.09 |
| 6 | 2.55 | 6 | 1.82 |

Test Statistics

| | 100 Nm | 150 Nm |
|---|---|---|
| N | 11 | 11 |
| Chi-Square | 20345 | 9.436 |
| Df | 3 | 3 |
| Asymp. Sig. | 0.504 | 0.024* |

*$p < 0.05$, statistically significant

Table 4.9 (b). RMS voltage at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening were tested using Friedman two-way analysis of variance by ranks. EMG data were processed using method A.

Ranks

| 100 Nm | | 150 Nm | |
|---|---|---|---|
| Week | Mean Rank | Week | Mean Rank |
| 0 | 3.09 | 0 | 3.36 |
| 2 | 2.00 | 2 | 2.73 |
| 4 | 2.27 | 4 | 2.09 |
| 6 | 2.64 | 6 | 1.82 |

Test Statistics

| | 100 Nm | 150 Nm |
|---|---|---|
| N | 11 | 11 |
| Chi-Square | 4.418 | 9.436 |
| Df | 3 | 3 |
| Asymp. Sig. | 0.22 | 0.024* |

*$p < 0.05$, statistically significant

Table 4.9 (c). IEMG at 150 Nm torque level between the initial and 2nd weeks, between the initial and 4th weeks and between the initial and 6th weeks were tested using Wilcoxon signed ranks test. EMG data were processed using method A.

Test Statistics

| | wk 0 – wk 2 | wk 0 – wk 4 | wk 0 – wk 6 |
|---|---|---|---|
| Z | -1.334 | -2.667 | -2.312 |
| Asymp. Sig. (2-tailed) | 0.182 | 0.008* | 0.021 |

*$p < 0.013$ (0.05/3), statistically significant

Table 4.9 (d). RMS voltages at 150 Nm torque level between the initial and 2nd weeks, between the initial and 4th weeks and between the initial and 6th weeks were tested using Wilcoxon signed ranks test. EMG data were processed using method A.

Test Statistics

| | wk 0 – wk 2 | wk 0 – wk 4 | wk 0 – wk 6 |
|---|---|---|---|
| Z | -1.689 | -2.490 | -2.223 |
| Asymp. Sig. (2-tailed) | 0.091 | 0.013* | 0.026 |

*$p < 0.013$ (0.05/3), statistically significant

***Results using data processing method B***

Although the EMG were processed using another method (method B), the changes in IEMG and RMS voltage at 150 Nm torque level with the strengthening program were also statistically different (with p = 0.011 and 0.005 respectively). Tables 4.10 (c) and (d) show that the difference was between the initial and the last sessions ($6^{th}$ week) for both IEMG and RMS (with p = 0.013 and 0.01 respectively). There was also statistically significant difference between the RMS at the initial and third sessions (p = 0.01). The change in RMS voltage, which was calculated by data processing method B, showed a greater statistical difference at 150 Nm torque level compared with the result obtained with method A.

It was obvious that IEMG and RMS voltage at 150 Nm torque level decreased when the muscle strength, which was expressed as the torque at MVC, increased with the six-week strengthening. Further analysis was done to investigate whether there was a predictable relationship between the change in the torque at MVC and that in the EMG. The result is reported in section 4.4.5. Moreover, the changes in IEMG at 100 Nm and 150 Nm torque levels were very much similar to that of RMS voltage as shown in figures 4.3 and 4.4. This will be discussed in section 4.4.4.

Table 4.10 (a). IEMG at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening were tested using Friedman two-way analysis of variance by ranks. EMG data were processed using method B.

Ranks

| 100 Nm | | 150 Nm | |
|---|---|---|---|
| Week | Mean Rank | Week | Mean Rank |
| 0 | 2.73 | 0 | 3.36 |
| 2 | 2.73 | 2 | 2.82 |
| 4 | 2.27 | 4 | 2.18 |
| 6 | 2.27 | 6 | 1.64 |

Test Statistics

| | 100 Nm | 150 Nm |
|---|---|---|
| N | 11 | 11 |
| Chi-Square | 1.364 | 11.182 |
| Df | 3 | 3 |
| Asymp. Sig. | 0.714 | 0.011* |

*$p < 0.05$, statistically significant

Table 4.10 (b). RMS voltage at 100 Nm and 150 Nm torque levels of isometric extension of the right knee during the six-week strengthening were tested using Friedman two-way analysis of variance by ranks. EMG data were processed using method B.

Ranks

| 100 Nm | | 150 Nm | |
|---|---|---|---|
| Week | Mean Rank | Week | Mean Rank |
| 0 | 3.09 | 0 | 3.64 |
| 2 | 2.00 | 2 | 2.36 |
| 4 | 2.55 | 4 | 2.27 |
| 6 | 2.36 | 6 | 1.73 |

Test Statistics

| | 100 Nm | 150 Nm |
|---|---|---|
| N | 11 | 11 |
| Chi-Square | 4.091 | 12.927 |
| Df | 3 | 3 |
| Asymp. Sig. | 0.252 | 0.005* |

*$p < 0.05$, statistically significant

Table 4.10 (c). IEMG at 150 Nm torque level of isometric extension of the right knee between the initial and 2nd weeks, between the initial and 4th weeks and between the initial and 6th weeks were tested using Wilcoxon signed ranks test. EMG data were processed using method B.

Test Statistics

| | wk 0 – wk 2 | wk 0 – wk 4 | wk 0 – wk 6 |
|---|---|---|---|
| Z | -1.156 | -2.312 | -2.490 |
| Asymp. Sig. (2-tailed) | 0.248 | 0.021 | 0.013* |

*p < 0.013 (0.05/3), statistically significant

Table 4.10 (d). RMS voltage at 150 Nm torque level of isometric extension of the right knee between the initial and 2nd weeks, between the initial and 4th weeks and between the initial and 6th weeks were tested using Wilcoxon signed ranks test. EMG data were processed using method B.

Test Statistics

| | wk 0 – wk 2 | wk 0 – wk 4 | wk 0 – wk 6 |
|---|---|---|---|
| Z | -2.401 | -2.578 | -2.578 |
| Asymp. Sig. (2-tailed) | 0.016 | 0.010* | 0.010* |

*p < 0.013 (0.05/3), statistically significant

### 4.4.4 Relation Between Integrated Electromyography and Root Mean Square Voltage

The changes in IEMG and RMS voltage were very similar in this study no matter the EMG signals was selected using method A or B. Table 4.11 also shows the high correlation coefficients between the IEMG and RMS. This could probably be explained by their mathematical formulae (Herzog et al 1994). It is because the IEMG is always obtained by integrating the rectified EMG signals over a fixed time period (either 0.81-second or 8-second) in this study. Therefore the two parameters could be used interchangeably if the same segment of EMG signals was processed to derive the IEMG and RMS voltage. Physically, both parameters represent the electrical energy needed to activate the muscle. However, since RMS voltage is usually readily available in most of the data processing software, it is found to be more convenient in its use to represent the electrical activity. On the contrary, full wave rectification and then integration of the EMG signals have to be done to obtain the IEMG.

$$IEMG = \int_{t}^{t+T} \left| EMG\,(t) \right| \, . \, dt$$

$$RMS = \left[ \frac{1}{T} \int_{t}^{t+T} EMG^2\,(t) \, . \, dt \right]^{1/2}$$

### 4.4.5 Correlation between the Change in Torque at Maximal Voluntary Contraction and Electromyography

The results of this study were described in detail in sections 4.4.1 and 4.4.3. The IEMG and RMS voltage at 150 Nm torque level decreased when muscle strength, which was expressed as the torque at MVC, increased with the six-week strengthening. Further analysis was done to investigate whether there was a predictable relationship between the change in the torque at MVC and that in the EMG. The correlation between the relative change in MVC and the relative change in IEMG and RMS voltage at second week, fourth week and sixth week were tested by the Spearman rank correlation coefficient ($n = 11$). The results were similar no matter how the EMG signals were selected (with method A or B). The details about the advantages and disadvantages of the two methods (A and B) has been explained in section 3.5.2. Therefore, table 4.11 only shows the data obtained with method B. All the correlation coefficients between the relative change in torque at MVC and EMG were less than 0.5. This demonstrated poor correlation between the relative change in torque at MVC and EMG. Scatter plots also did not reveal any relationship.

As a conclusion, significant differences were found in both the torque at MVC and EMG at 100 Nm and 150 Nm torque levels with the six-week strengthening program. However, the relationship could not be predicted. This could be explained by the fact that there were more than one variable or mechanism leading to the

increase in the torque at MVC during the strengthening program. Moreover, the extent of influence of these mechanisms might vary from individual to individual. This idea will be discussed in detail in section 5.2.

Furthermore, there were high correlation between the IEMG and RMS value with two-week and four-week strengthening ($r = 0.764$ and 0.673 respectively). The correlation between these two EMG parameters with the six-week strengthening was very high ($r = 0.936$). This confirmed the observation as mentioned in sections 4.4.1 and 4.4.2. Section 4.4.3 has a detailed discussion on this point.

Table 4.11 Spearman rank correlation coefficient between the relative change in the torque at maximal voluntary contraction (MVC) and integrated electromyography (IEMG) or root mean square value (RMS) at 150 Nm with strengthening program. EMG data were processed using method B.

(a) At the second week

| | Torque at MVC | Change of IEMG at 150 Nm | Change of RMS at 150 Nm |
|---|---|---|---|
| Torque at MVC | | -0.415 (p = 0.205) | -0.378 (p = 0.252) |
| Change of IEMG at 150 Nm | -0.415 (p = 0.205) | | 0.764* (p = 0.006) |

n = 11

(b) At the fourth weeks

| | Torque at MVC | Change of IEMG at 150 Nm | Change of RMS at 150 Nm |
|---|---|---|---|
| Torque at MVC | | 0.165 (p = 0.627) | -0.138 (p = 0.687) |
| Change of IEMG at 150 Nm | 0.165 (p = 0.627) | | 0.673* (p = 0.023) |

n = 11

(c) At the sixth week

| | Torque at MVC | Change of IEMG at 150 Nm | Change of RMS at 150 Nm |
|---|---|---|---|
| Torque at MVC | | 0.110 (p = 0.748) | 0.110 (p = 0.748) |
| Change of IEMG at 150 Nm | 0.110 (p = 0.748) | | 0.936* (0.001) |

n = 11

## 4.5 Summary

The subjects recruited in this study were involved in a scheduled strengthening program of 30 maximal voluntary contractions (MVC) per day, 3 days a week, for 6 weeks. A between-day reliability test on the measurement of the torque at MVC was conducted. The intra-class correlation coefficient was 0.86. It was concluded that the method used for measuring the torque in this research was reliable. Eleven male subjects completed the six-week strengthening program. There was significant difference in MVC after the strengthening program ($p = 0.001$). Moreover, significant differences were found in both the torque at MVC and EMG at 100 Nm and 150 Nm torque levels with the six-week strengthening program. However, the trend could not be predicted. Furthermore, there were high correlation between the IEMG and RMS value with two-week, four-week and six-week of strengthening ($r = 0.764$, 0.673 and 0.936). The changes in IEMG and RMS value were very similar and this could be explained by their mathematical equivalence. Therefore, it is suggested that IEMG and RMS value of the same period of EMG could be used interchangeably if the same segment of EMG signals was processed to derive the IEMG and RMS voltage.

# Chapter 5 Discussion

## 5.1. Development of a Test Protocol

The extension torque level at maximal voluntary contraction (MVC) of the quadriceps femoris muscles increased significantly by an average of 22 % during the six-week strengthening program. In addition, lower amplitudes of EMG were recorded at the pre-set submaximal levels of contractions. As a result, there was a right shift of the torque-EMG curve after strengthening (fig. 4.5). This result agreed with those findings reported in other similar studies (table 5.1). It suggested that the decrease in EMG under a reasonably strong submaximal contraction at pre-determined torque level in the course of strengthening was an indication of improvement in the quadriceps' mechanical output. This finding supported the use of a test protocol with pre-determined torque level to monitor the improvement in muscle strength. This test protocol is safer when compared with the commonly used test protocol with percentage of MVC because the test for MVC is always difficult and sometimes dangerous in persons after injury. Therefore this new test protocol has the potential to be applied in rehabilitation.

Table 5.1. Summary of the change in the force-EMG (F-E) curves after strengthening

| Author | Muscle studied | Program duration | Exercise protocol | Change in the F-E relationship |
|---|---|---|---|---|
| Garfinkel et al 1992 | Vastus lateralis | 8 weeks<br>3/ week | Isometric knee extension exercise<br>3 sets of 10 MVC/ day | Significant right shift in the F-E curve<br>(fig. 5.1) |
| Narici et al 1996 | Vastus lateralis | 6 months<br>Exercise at alternate days | Concentric and eccentric exercise of the quadriceps<br>6 sets of 8<br>80 % 1 RM | Significant right shift in the F-E curve<br>(fig 5.2) |
| Rabita et al 2000 | Rectus femoris, vastus lateralis, vastus medialis | 4 weeks<br>3/ week | Bilateral isometric knee extension exercise<br>5 sets of 5<br>80 % MVC | Significant right shift in the F-E curves in all muscles<br>(fig. 5.3) |

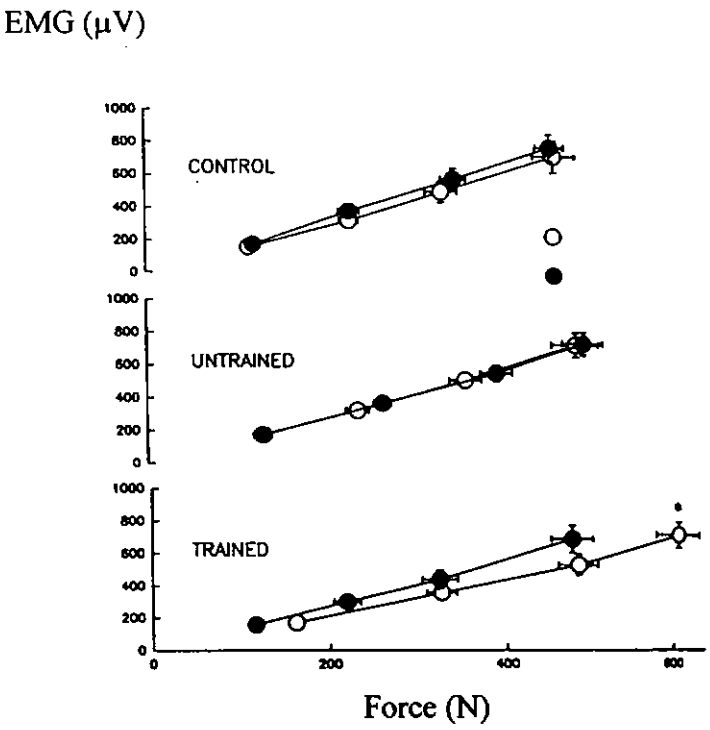



Figure 5.1. The force-EMG relationship in the right vastus lateralis muscle of the control subjects (n = 7), trained (n = 8) and untrained experimental subjects, before (solid circle) and after (open circle) 8 weeks of training. No changes in this relation were seen in the control and untrained legs but there is significant change in the trained leg. (From Garfinkel et al 1992)

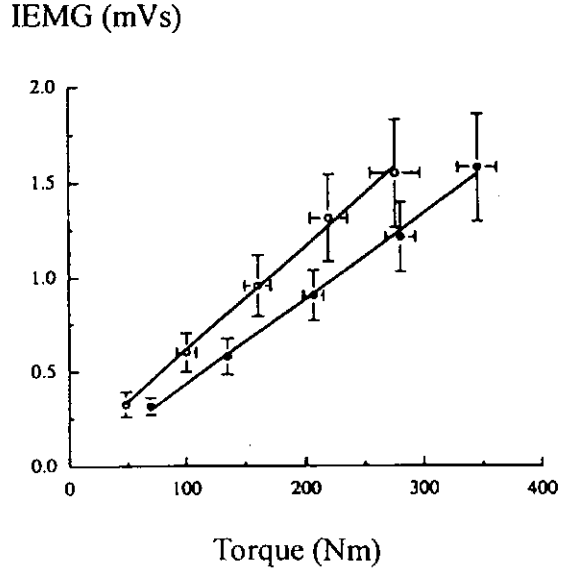



Figure 5.2. Torque-IEMG relation before (open circle) and after (solid circle) 6 months strengthening (n = 7). Experimental points are fitted with linear functions. (From Narici et al 1996)

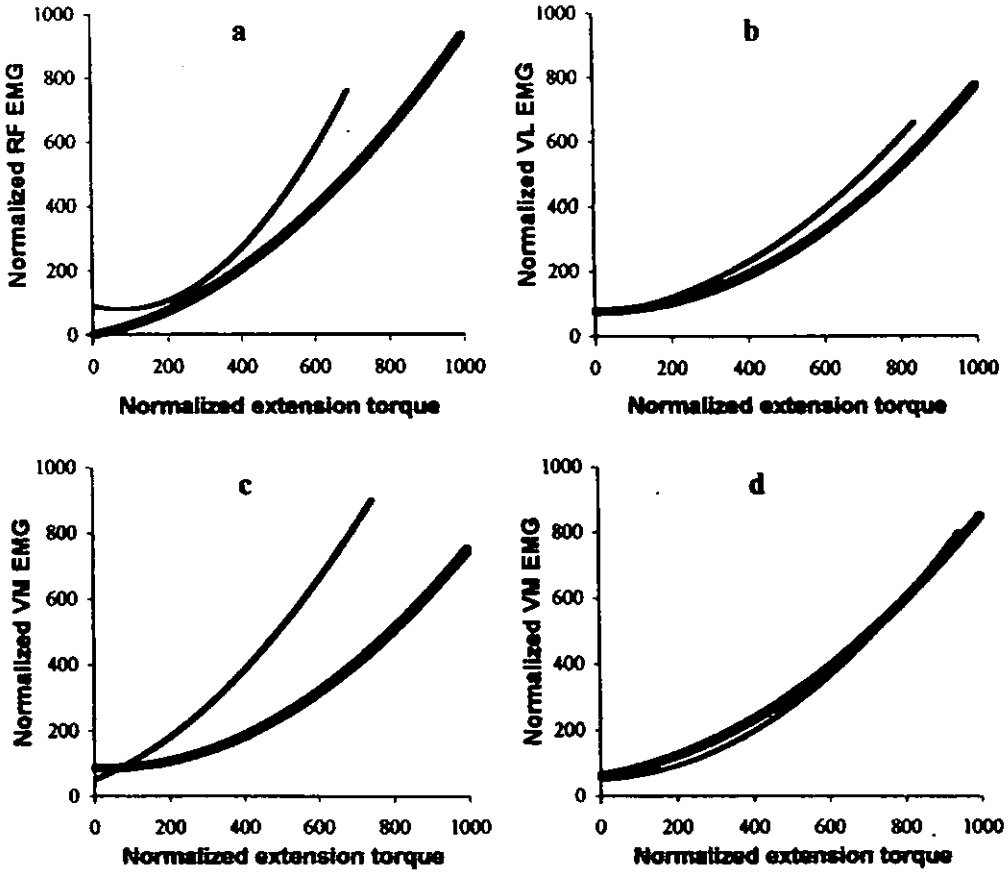



Figure 5.3. Torque-EMG relationships established before (thin line) and after (thick line) the 4 weeks strengthening period for three trained subjects (a, b, c) and one control subject (d). The relationships were obtained for the rectus femoris (RF), vastus lateralis (VL) and vastus medialis (VM) muscles. Torque and EMG were normalized to the highest values calculated during all the tests. (From Rabita et al 2000)

### 5.1.1. Pre-requisite for Using Surface EMG to Monitor Muscle Strength

Surface electromyography (EMG) is commonly used to study the muscle function. However several conditions have to be fulfilled in order to acquire signals of high fidelity. Firstly, the confounding variables must be reasonably controlled, otherwise the data will be contaminated with noise and become invalid. This has been discussed in sections 3.4 and 3.5. Moreover, the stability of EMG signals during the muscle contraction at the pre-determined torque level has to be observed.

***Stability of EMG signals in submaximal contraction***

In quantitative analysis of EMG, the stability of the EMG signals will affect the amplitude of the EMG. A segment of EMG signals has to be selected for further data processing and analysis (table 3.4). Two methods were used to select the appropriate segment of EMG at 100 Nm and 150 Nm torque levels in this study. **Method A** used 0.81-second (equivalent to 4096 data points) EMG segment corresponding to a period of stable torque during submaximal contractions, ie. at 100 Nm and 150 Nm. **Method B** used an 8-second (7 to 15 second) segment of EMG during these submaximal contractions. Theoretically, a longer segment can compensate for the instability because the error due to accidental selection at either the peak or tough EMG segment is reduced. It was shown in section 4.4.3 that the results derived from the EMG of both methods showed significant decreases in the EMG (both RMS value and IEMG)

required to generate a 150 Nm torque after six weeks of strengthening exercise. The change in RMS voltage, which was calculated by data processing method B, showed a slightly greater statistical difference at 150 Nm torque level. Therefore it is opined that both methods are suitable for EMG data processing whereas method B is preferable to method A.

### 5.1.2. Potential Application in Rehabilitation

Although it is common in many studies to examine the EMG activity at a certain percentage of the maximal voluntary contraction (MVC) (Mannion et al 1996 and Narici et al 1996), pre-set torque levels (100 and 150 Nm) for submaximal tests were used in this study for the following reasons. Firstly, it is usually difficult or sometimes impossible for people who are injured or suffering from musculoskeletal problems of the limbs to perform MVC of the muscle over the affected limb. A test protocol, which must rely on MVC, is therefore unrealistic. It is also the intention of this study to establish a test protocol potentially applicable on these types of patients. A pre-determined level of torque at submaximal contraction is therefore safer and easier to be administered as a test protocol for these patients during rehabilitation. Secondly, feedback was facilitated from the excellent visual display of the dynamometer (Cybex "NORM"), which could be set at precise torque levels (fig. 3.2), such that the subjects under tests could follow easily by keeping the torque output stable (eg. at 100 Nm or

150 Nm). If required, the test protocol could be extended to other torque levels according to the strength of the subjects or to the joints to be tested in future studies. This makes the system more technically manageable by setting at fixed torque levels with a step-wise increment within a pre-determined range.

Another important consideration of the test protocol is the choice of torque level. The signal-to-noise ratio may be unsatisfactory if the torque level is too low. The EMG signals acquired may be of similar level to the noise of the system and to that coming from the test environment if the signal level is too low. On the other hand, the torque level cannot be set too high as it would be too close to or even higher than the torque at MVC. As a conclusion, the torque should be set at a reasonably strong level of submaximal contraction but less than that at MVC. The lowest torque at MVC was 160 Nm in this study. Therefore, the torque levels of 100 Nm and 150 Nm for sustained isometric tests were considered appropriate in this study.

Test protocol using pre-determined torque level can also be used in rehabilitation especially in monitoring the progress of muscle strength when in-vivo measurement of isolated muscle by mechanical means is not possible. This happens when the muscle is still very weak (Oxford grade 1 or 2) after injury. Moreover, it is also difficult to detect whether the muscle, which has been transferred surgically, is contracting after the surgery especially at the initial stage because the muscle is too

weak. Test protocol using surface EMG would be useful in these circumstances. A standardized methodology and instrumentation, a strict control of confounding variables and a careful selection of data processing technique are all very important for successful implementation of surface EMG as an assessment tool for monitoring the improvement of muscle strength. Some prerequisites, such as the matching of the time in occurrence of the maximal torque and electromyography, should also be satisfied (section 5.1.1) if EMG is to be used to monitor the progress in muscle strength. Further study on the change of EMG with strengthening in persons undergoing rehabilitation is necessary (section 5.4). This facilitates the development of a test protocol, eg. on the selection of optimal level of torque, that is suitable for a specific group of patient.

## 5.2. Double Normalization Technique

The double normalization technique as mentioned in section 3.5.3 was designed in this study to allow the comparison of EMG signal amplitude during submaximal contractions at different days even when a pre-set level (instead of % of MVC) of torque was used as the test protocol. However, whether or not the maximal EMG matches well in timing with the maximal torque does make a dramatic difference in the final EMG amplitude (ie. RMS value or IEMG). Special attention has to be paid on this when the double normalization technique is used for future research.

***Time in occurrence of maximal torque and electromyography***

Another important consideration before surface EMG was applied to monitor the progress in muscle strength was the time in occurrence of maximal torque and electromyography. It was once believed that EMG increased in proportion to muscle strength. However previous studies demonstrated that this was not the case due to several reasons. Firstly, though an increased force was always accompanied by an increased EMG in-vitro, it was not certain that whether the relation was linear or not (Vredenbregt 1973, Guimaraes et al 1995, Solomonow 1990 and 1987). Different explanations for this phenomenon have been proposed. Guimaraes and associates (1995) studied the cat soleus muscles. It was concluded that the relation between the mean force and integrated EMG (F-E) was linear for isometrically contracting cat

soleus muscle within the physiologically relevant (intermediate) region though the F-E relation was non-linear for the whole range of stimulations used in his test protocol. The F-E curves of different muscle lengths were also studied. It was found that all the intermediate regions simulated a straight line (fig. 5.4). However, Solomonow & associates attributed the difference in F-E relation to different control strategies during muscle contractions and different muscle fibre composition. It was found that cat soleus exhibited a linear F-E relation when it employed the 30% and 40% recruitment range control strategy, ie. using motor unit recruitment to produce the force equal to initial 30 % to 40 % of the maximum force. The further increase in force was achieved by increasing the firing rate of the same motor unit (fig. 5.5). Control strategies with more than 40 % recruitment range produced progressively non-linear curves. It happened in a similar way in gastrocnemius muscles. It was also concluded that not only different muscle has different control strategy and exhibit different curve shape, but the same muscle might also have different control strategies under different functional conditions.

The composition of muscle fibre was another major factor that affects the F-E model, ie, whether it is predominately fast or slow twitch muscle. Slow twitch muscle usually got large proportion of small and medium sized motor units. It was highly sensitive to recruitment in the low and medium force range, whereas fast twitch

muscle, with a large proportion of large motor units, depended on recruitment to generate most of its force.

The second possible reason to explain the mismatch between EMG and force measurements in-vivo was that the muscle torque was usually contributed by a group of agonistic muscles. In the present study, the torque measured was contributed by a combination of muscle work of the rectus femoris, vastus intermedius, vastus lateralis and vastus medialis muscles. However, surface EMG was only measured on the rectus femoris muscle. It is not certain about the proportion of the torque that was contributed by the rectus femoris muscle. It is also not sure whether the rectus femoris muscle will be activated to the same extent in every active contraction. Variation in the state of activation of the rectus femoris muscle will affect the EMG signal recorded. Weir and associates (1994) studied the change in maximum IEMG and muscle strength of the vastus lateralis muscle in human. It was reported that there was also dissociation between the change in IEMG and muscle strength. It was possibly due to differential responses to training in the four muscles of the quadriceps femoris. On the contrary, it was reported that all heads work equally during the isometric contraction of the quadriceps muscles (Salzman et al 1993). Therefore, the choice of the EMG acquired from the rectus femoris muscle was justified to infer their myoelectrical activity to all heads of the quadriceps muscles (Molina et al 2000).

Furthermore, the effect of co-contraction of the hamstrings on the resultant extension torque has been mentioned though it was found that this was not significant in the present study (section 3.4.3).

In this study, the EMG acquired at submaximal contraction was normalized with the EMG data that correspond to the 0.81-second peak torque at the MVC of the quadriceps. Therefore, a good match in the time of occurrence between the maximum EMG segment and the maximum torque segment was an important pre-requisite for further data analysis. Figure 5.6 illustrates the behavior of these two parameters of different subjects. Figures 5.6 (a) and (b) show the EMG and torque curves at MVC of two subjects who exhibited satisfactory correspondence of these two curves in all trials during the initial, $2^{nd}$, $4^{th}$ and $6^{th}$ weeks. Figure 5.6 (c) shows the maximum EMG and torque curves of a subject who showed poor correspondence in the time of occurrence of the maximal EMG and maximum torque.

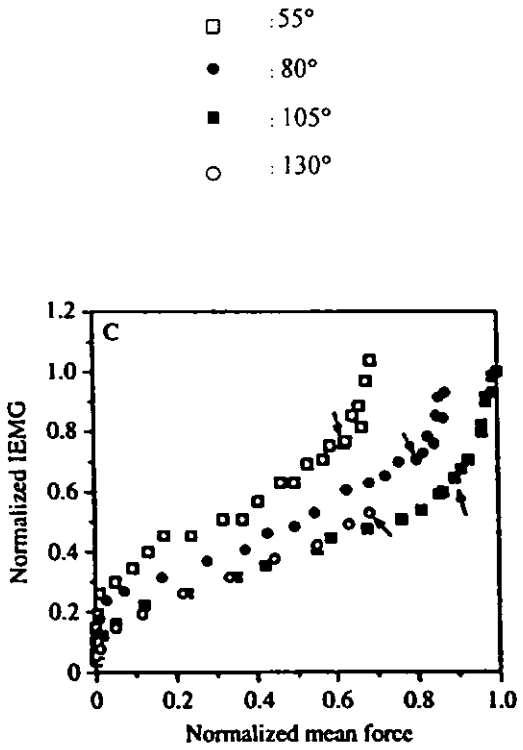



Figure 5.4. Effects of four different muscle lengths of the cat soleus muscle on the relationship between IEMG and mean force obtained in one experiment. The muscle length of the soleus was adjusted according to different joint angles, 55°, 80°, 105° and 130°. 55° corresponded to the shortest muscle length. 130° corresponded to the longest. The last data point of each linear region is indicated with an arrow. It is noted that all the intermediate region of the curves is approximately linear. The author considered this linear region as the physiological region of the muscle. (From Guimaraes et al 1995)

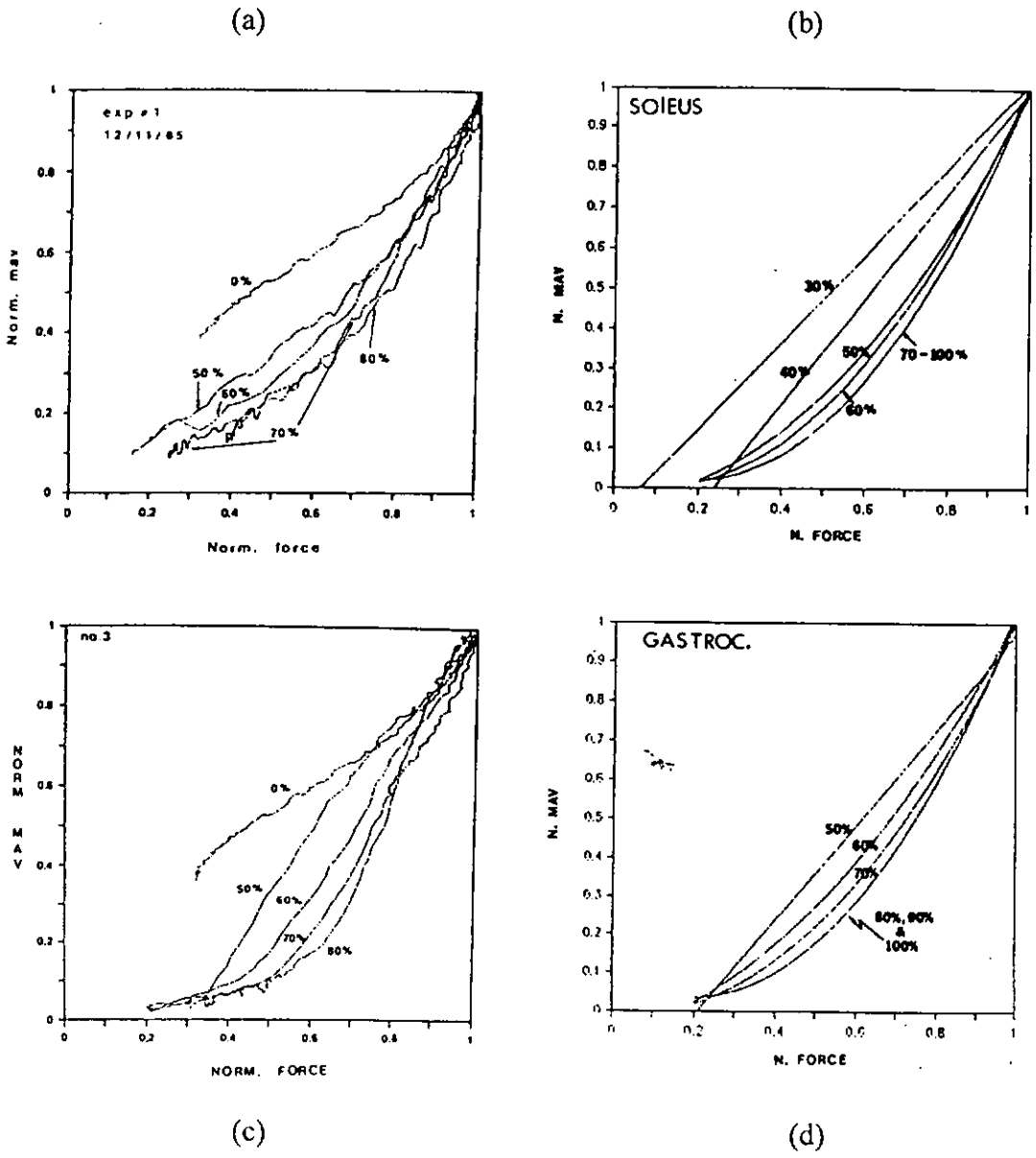



Figure 5.5. The Force-EMG relationship of the soleus (a and b) and medial gastrocnemius (c and d) muscles of the cat. Normalized EMG (mean absolute value, MAV) was plotted against normalized force (percentage of MVC). Several control strategies employing 0, 50, 60, 70, and 80 percent recruitment range obtained from the calf muscles of two cats. Note the quasilinear curves for 50 percent recruitment range and the gradual increase in nonlinearity for larger recruitment ranges. (From Solomonow et al 1990)

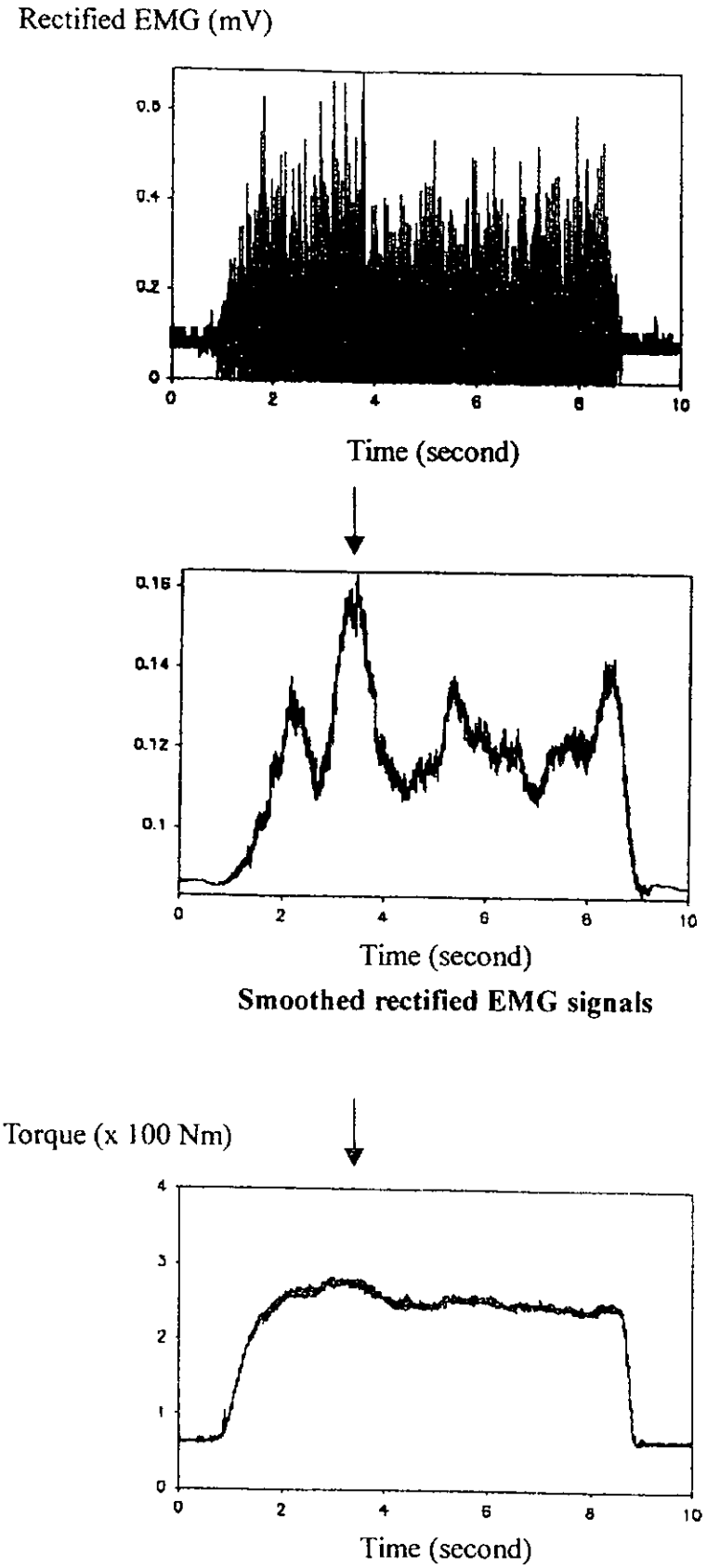
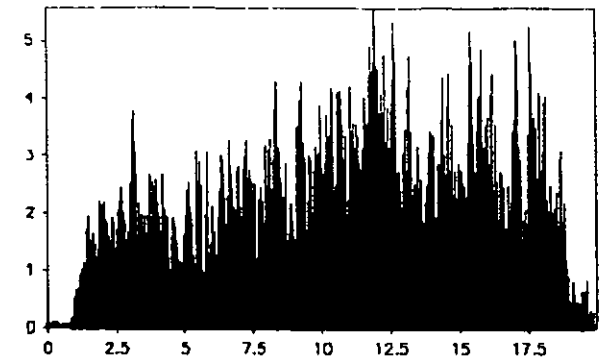
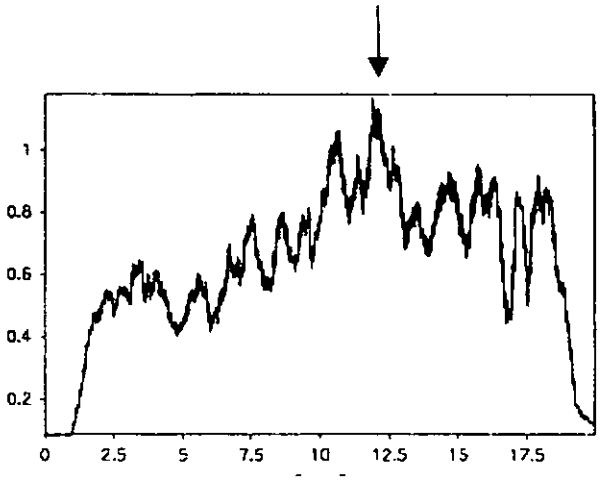



Figure 5.6 (a). EMG signals and torque produced by subject C. It shows a good correspondence between the maximum EMG and the maximum torque as indicated by the arrows.

Rectified EMG (mV)



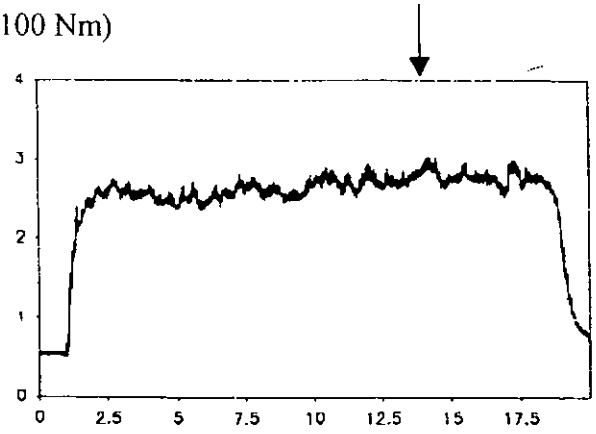

Time (second)




Time (second)

**Smoothed rectified EMG signals**

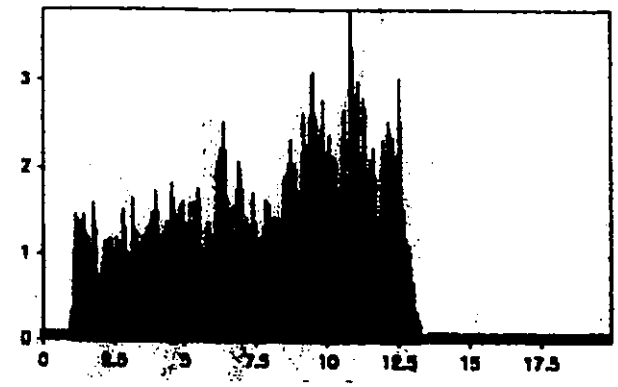
Torque (x 100 Nm)



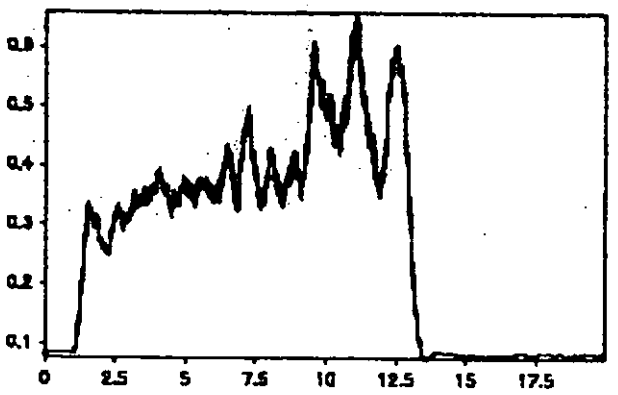

Time (second)

Figure 5.6 (b). EMG signals and torque produced by subject D. It shows a satisfactory correspondence between the maximum EMG and the maximum torque as indicated by the arrows.

Rectified EMG (mV)



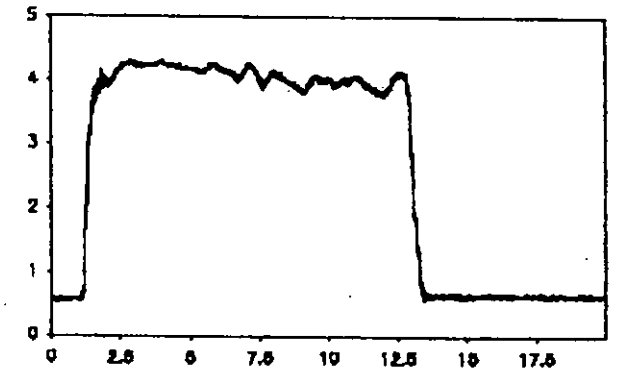

Time (second)




Time (second)

**Smoothed rectified EMG signals**

Torque (x 50 Nm)




Time (second)

Figure 5.6 (c). EMG signals and torque produced by subject E. It shows a poor correspondence between the maximum EMG and the maximum torque (at the 6$^{th}$ week) as indicated by the arrows.

In this study, all eleven subjects, who have completed the strengthening program, exhibited reasonably good correspondence between the maximum EMG and peak torque. A close correspondence between the time of occurrence of the maximum EMG and maximum peak torque was considered as an important prerequisite for using surface EMG data to monitor improvement in muscle strength. Therefore, data from the subject as illustrated in figure 5.6 (c) was excluded from the data analysis. However the data of this subject were processed separately and the result is presented as follows.

***Change in result of the 12th subject after adjusting the selection of maximum electromyography segment***

Twelve subjects participated in this study but only the data of eleven subjects were analyzed as reported in chapter 4. It is because one of the twelve subjects exhibited very poor correspondence between the maximum torque and maximum EMG. The data of this subject was therefore not included in data analysis. Table 5.2 (a) showed that the EMG (both RMS voltage and IEMG) did not exhibit the decreasing trend as shown in other 11 subjects. However if the maximum EMG segment instead of that corresponding to the maximum torque segment was selected for normalization, the change in IEMG and RMS voltage show a similar pattern as the other subjects. It is illustrated in table 5.2 (b). The IEMG and RMS voltage at 100 Nm and 150 Nm

torque levels showed a decreasing trend as the torque at MVC increased during the six-week strengthening.

Table 5.2 (a). EMG data (after double normalization) of the 12th subject using 0.81-second of EMG segment corresponding to maximum torque. The subject performed maximal isometric extension of the right knee. The data of this subject was excluded from analysis in the main study because of mis-matching in the occurrence of maximal torque and maximal EMG.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Relative change of torque at MVC | 1 | 1.09 | 1.18 | 1.22 |
| RMS (mV) at 100 Nm | 0.3627 | 0.2439 | 0.5491 | 0.4662 |
| RMS (mV) at 150 Nm | 0.5525 | 0.4099 | 0.8238 | 0.6787 |
| IEMG (mVs) at 100 Nm | 0.4614 | 0.3118 | 0.6298 | 0.5109 |
| IEMG (mVs) at 150 Nm | ***0.6469*** | ***0.4604*** | ***0.8231*** | ***0.6786*** |

Table 5.1 (b). EMG data (after double normalization) of the 12th subject using 0.81-second of maximum EMG segment at maximal isometric contraction of the right knee.

| Wk | 0 | 2 | 4 | 6 |
|---|---|---|---|---|
| Relative change of torque at MVC | 1 | 1.09 | 1.18 | 1.22 |
| RMS (mV) at 100 Nm | 0.3627 | 0.2439 | 0.3482 | 0.2241 |
| RMS (mV) at 150 Nm | 0.5525 | 0.4099 | 0.5225 | 0.3264 |
| IEMG (mVs) at 100 Nm | 0.4614 | 0.3118 | 0.4443 | 0.3071 |
| IEMG (mVs) at 150 Nm | ***0.6469*** | ***0.4604*** | ***0.5807*** | ***0.4077*** |

## 5.3. Physiological Explanation of Muscle Strengthening

The aim of this study was to investigate how the EMG changes when the muscle strength increases during a six-week strengthening program. It was hoped that the change in the amplitude of EMG would assist the understanding of the role of neural adaptation in the process of muscle strengthening. This was based on the theory that any improvement in the synchronization of motor unit recruitment and excitation frequency would induce an increase in the amplitude of EMG (section 2.2). It has also been reviewed that median frequency (MF) would change during muscle strengthening (section 2.4.3). The possible physiological explanation of the increase in muscle strength will be discussed in this section.

### 5.3.1. Change in the Amplitude of EMG

The muscle strength as reflected by the increase in the torque at MVC (22 % in average) was definitely increased after the 6-week strengthening program. According to the literature, several physiological mechanisms may cause the increase in the muscle strength (Moritani et al 1993). These are:

1. Neural factors;

2. Quantity of muscle fiber, ie. muscle hypertrophy or hyperplasia; and

3. Quality of muscle fiber, ie. type of muscle fiber, sarcolemmal morphology.

The results of the present study showed that the change in RMS voltage and

IEMG at MVC did not differ significantly. There were contradicting findings in the literature concerning the change of EMG at MVC (table 2.1). However, measurements of muscle bulk and microscopic change in the muscle fibers were not included in present study. Therefore, it is tried to explain the results found in this study by the information given in the literature. Since there is fairly good consensus in the literature that at least six to seven weeks of regular resistive training is required before muscle hypertrophy is significant, the relatively short (six weeks) strengthening program adopted in present study is not likely to cause a significant muscle hypertrophy. Moreover, if muscle hypertrophy does happen, there should be a more remarkable decrease in the EMG at 150 Nm in the latter part of the training period because of the obvious increase in specific tension of the muscle. However, most of the decrease in EMG in this study happened at the first four weeks. The EMG did not change much in the fourth to sixth weeks. Whether the improvement in muscle strength can be explained by the other two factors demands a deeper understanding on the concept of "neural adaptation" and its relation to the structure of the neuromuscular system.

Enoka (1994) gave a more detailed description of "neural adaptation". He summarized "neural factors" into motor unit recruitment, discharge rate of motor units and discharge pattern (synchronization of motor units). According to Enoka (1994),

motor unit recruitment involves the concepts of orderly recruitment according to the size principle. It further depends on the morphological, biophysical and input characteristics that vary with motor neuron size such that the smallest motor neurons can be excited most easily. Besides motor neuron size, motor unit recruitment is also influenced by other characteristics of the motor neuron (intrinsic factors) and by the organization of synaptic input (extrinsic factors) on the dendrites and soma of the motor neurons in the pool. The "intrinsic factors" include motor neuron size, the sensitivity of the neurotransmitter receptors of the motor neuron, and the electrotonic characteristics of the motor neuron. Electrotonic refers to the electrical responses of an excitable membrane due to changes in conductance. The "extrinsic factors" include the number of synaptic terminals on a motor neuron from a given input system, the average amount of neurotransmitter liberated at each synapse and the spatial distribution of the synapses over the soma and dendrites. Changes in these intrinsic or extrinsic factors (ie. neural adaptation) after muscle strengthening may improve the excitation of muscles and result in an increase in the maximal surface EMG. However, obvious increase in the maximal EMG could not be demonstrated in this study.

There are other mechanisms such as metabolic and ultrastructural adaptations that have been mentioned by Howald (1982). Muscle strengthening can induce a decrease in oxidative enzymes, increase in myosin ATPase activity and transformation

of myosin and tropomyosin with 6 weeks of training. Muscle strengthening exercise as short as six weeks would possibly induce a change in the types of muscle fiber, eg. from type I to IIc. This can probably account for the increase in muscle strength even when there is no change in maximum IEMG and also explain the results of the present study. Other research showed that a rapid (within days) exercise-induced increase in new force-producing myofibrillar proteins appeared to be a real possibility. This casts again doubt on the idea that strength gains that occur early in a strengthening program are due exclusively to neural mechanisms (Phillips 2000). This may also explain the increase in force-generating capability without a change in fiber diameter, ie. muscle hypertrophy.

### 5.3.2 Change in the Power Density Spectrum

It has been reported that there was a change in the power density spectrum (PDS) during muscle strengthening (Moritani et al 1993). The change in PDS is often represented by a change in the median frequency (MF). There was no obvious change in the MF at both maximal and submaximal contractions in the present study. The results reported in other literatures (Moritani 1993, Pincivero 2000; Kupa 1995, Hamill 1995) could not be reproduced in this study. In fact, the possible changes in MF after strengthening as suggested by Moritani and Pincivero are contradicting (table 5.3). The lack of obvious trend in the MF as shown in the present study can be

explained by the discrepancy in physiological explanations proposed in the literature. Further study is therefore necessary before a definite trend in the change of PDS after strengthening can be predicted.

Table 5.3. Summary of results in the change of PDS after strengthening

| Author | Muscle studied | Methodology | Change in PDS | Physiological explanation |
|---|---|---|---|---|
| Moritani et al 1993 | Long and short heads of biceps brachii | Subjects underwent 2 weeks of 30 % MVC power training. The change in PDS was monitored. | Mean power frequency shifted toward **lower frequency** bands. | The possible reason is a better synchronization of motor units. This produced EMG oscillations with larger and lower frequency. |
| Pincivero et al 2000 | Rectus femoris, vastus lateralis and vastus medialis | Subjects performed 5-second isometric contraction from 10 to 90 % MVC in random order. | The rectus femoris muscle had a **higher median frequency** than the vastus medialis. | The possible reason was that the rectus femoris had the highest percentage of type II muscle fibers. Type II muscle fibers have higher frequency EMG. |

## 5.4. Limitations and Further Studies

One major assumption of this study is that the knee extension torque measured by Cybex dynamometer is equivalent to the torque of rectus femoris muscle. In fact, this is contributed by all four components of the quadriceps femoris muscle. They are the rectus femoris, vastus intermedius, vastus lateralis and vastus medialis muscles. However, it is inevitable to adopt such assumption in in-vivo study. Many groups

studying the change in EMG with muscle strengthening (Narici et al 1996, Weir et al 1994, Garfinkel et al 1992) have adopted this assumption. This assumption is also supported by Pincivero's study (2000). It was found that there was parallel activation of superficial quadriceps muscles within the middle range of contraction intensities (40 – 70 % MVC). In addition to this, it was found that all heads of the quadriceps muscles worked equally during the isometric contraction of the quadriceps (Salzman et al 1993).

Furthermore, this study was conducted in normal healthy subjects. It is not certain whether the change in EMG will be the same in subjects after injury. Further study on patients is necessary. The test protocol using pre-determined torque level has the potential to be applied in clinical situation when the test of isolated muscle strength is desirable. A common scenario is anterior knee pain due to mal-alignment of the patella. There is hypothesis claiming that muscle imbalance between the vastus medialis (especially the obliquus) muscle and the vastus lateralis muscle is causing the pain. Specific strengthening of the vastus medialis muscle is a common part of the rehabilitation program. It has been very difficult to assess specifically the improvement in the muscle strength of the vastus medialis muscle except by observation of the muscle bulk in clinical situation. Measurement of muscle hypertrophy of the muscle by magnetic resonance imaging is a good method but is too

costly to be applied in daily clinical situation. As a result, it is difficult to determine which exercise is effective in selective strengthening of the muscle. Previous study (Vaatainen et al 1995) on the EMG and muscle strength of the quadriceps muscles showed that they were significantly decreased in chondromalacia patients. Therefore, further study to monitor the change in the amplitude of surface EMG of the vastus medialis muscle during a period of selective strengthening exercise would be useful to monitor if there is any improvement in the muscle strength. An important point is the determination of the relevant torque level in this group of patients. 150 Nm is the appropriate torque level for normal male subjects. Further trial of lower torque levels is necessary in patients with anterior knee pain. Female patients may require a lower torque level than male subjects. However, too low a torque level (eg. 50 Nm in this study) should be avoided because this will decrease the signal-to-noise ratio. Moreover, the change in the amplitude of EMG is smaller in lower level of torque. Therefore, a higher level of torque is preferred as soon as it does not exceed the torque level at maximal voluntary contraction.

Lastly, the strengthening program lasted for only six weeks in this study. A longer period of training may be able to confirm whether maximum IEMG or RMS voltage will increase with further strengthening. It is also interesting to know whether the decrease in MF will become significant with a longer training period.

# Chapter 6 Conclusions

Muscle strengthening is commonly used in rehabilitation. One of the objectives of this study was to develop a test protocol to monitor the improvement of muscle strength by the use of surface electromyography (EMG). It was hoped that this could supplement the muscle-testing method by biomechanical means. The second objective was to investigate the role of neural adaptation in muscle strengthening through the study of the change in EMG at maximal voluntary contraction (MVC).

Eleven normal, healthy, lean and young (20 to 40 years old) male subjects were recruited for this study. The subjects were engaged in a scheduled strengthening program of the right quadriceps muscles. The program required 30 maximal voluntary contractions (MVC) per day, 3 days a week, for 6 weeks. The knee extension torque and surface electromyography (EMG) of the rectus femoris muscle at MVC and at two pre-determined levels of submaximal contraction of 100 Nm and 150 Nm, were recorded. The techniques used to control several confounding variables and the results were presented. These confounding variables included skin impedance, cross-talk and co-contraction of the hamstrings muscles. It was found that thorough cleaning of the human skin by detergent and alcohol was sufficient to lower the skin impedance to below 5 k$\Omega$. This method could replace aggressive abrasion of the skin, which was sometimes not preferred because of the scarring problem when frequent consecutive

recording was necessary. The effect of cross-talk ( 4 % to 17 %) and co-contraction of the hamstrings ( < 6 % of the maximal EMG activity) were considered acceptable and would not affect the fidelity of the EMG recorded in the rectus femoris muscle. Special techniques on signal processing including a careful selection of the EMG data segment and double normalization of the EMG data were reported. This normalization procedure was designed to facilitate comparison of EMG at submaximal torque levels between sessions when a test protocol using pre-determined torque level was used.

A between-day reliability test on the measurement of the torque at MVC was conducted. The intra-class correlation coefficient was 0.86. It was concluded that the method used for measuring the torque at MVC in this study was reliable. There was a significant increase (22 %) in the torque at MVC after the strengthening program ($p = 0.001$). Moreover, a significant decline in the amplitude of EMG (integrated EMG and root mean square voltage) was found at 100 Nm and 150 Nm torque levels with the six-week strengthening program. However, the exact relationship could not be predicted as shown by the statistical test (Spearman rank correlation coefficient). It was also found that there was a right shift of the torque-EMG relationship after strengthening.

One of the important achievements of this study was the application of the double normalization technique to facilitate comparison of EMG data at pre-

determined submaximal levels of muscle contraction. The importance in the time in occurrence of the maximal torque and EMG was emphasized when the double normalization technique was used.

Another important achievement was the establishment of a test protocol at pre-determined submaximal level of torque to monitor the improvement of muscle strength after strengthening exercise. The feasibility of this test protocol was supported by the evidence of a right shift of the torque-EMG relationship after strengthening. This evidence showed that the decrease in the amplitude of EMG under a reasonably strong submaximal contraction at pre-determined torque level in the course of strengthening was an indication of improvement in the quadriceps' mechanical output. This new protocol is safer than that uses percentage of MVC as test protocol. It is potentially applicable in rehabilitation because the test for MVC is usually difficult or sometimes impossible in patients. The results of this study may be used as a reference for future studies involving patients who are suffering from conditions in which the muscle strength is difficult to be measured with mechanical means. The importance of the prerequisite, the stability of EMG at submaximal contraction, for the use of surface EMG was also emphasized when the same methodology was applied.

## References


1. Astrand P & Rodahl K (1986). Textbook of work physiology. Third edition. McGraw-Hill, New York.

2. Basmajian JV & de Luca (1985). Muscle alive. Their functions revealed by electromyography. 5th edition. Williams & Wilkins, Baltimore.

3. Bernardi M, Solomonow M, Sanchez JH, Baratta RV & Nguyen G (1995). Motor unit recruitment strategy of knee antagonist muscles in a step-wise, increasing isometric contraction. European Journal of Applied Physiology. 70:493-501.

4. Bilodeau M, Cincera M, Gervais S, Arsenault AB, Gravel D, Lepage Y & Mckinley P (1995). Changes in the electromyographic spectrum power distribution caused by a progressive increase in the force level. European Journal of Applied Physiology. 71:113-123.

5. Bilodeau M, Arsenault B, Gravel D & Bourbonnais D (1992). Influence of gender on the EMG power spectrum during an increasing force level. Journal of Electromyography and Kinesiology. 2:3:121-129.

6. Bouisset S & Maton B (1973). Comparison between surface and intramuscular EMG during voluntary movement, in "New developments in electromyography and clinical neurophysiology". Edited by Desmedt JE. Volume B. Karger, Basel.

7. Burden A & Bartlett R (1999). Normalisation of EMG amplitude: an evaluation and comparison of old and new methods. Medical Engineering and Physics. 21:247-257.

8. Cram JR (1990). Clinical EMG for surface recordings. Volume 2. Clinical Resources, Nevada City.

9. De Luca CJ (1997). The use of surface electromyography in biomechanics. Journal of EMG and Kinaesiology. 7:135-163.

10. Draganich LF, Jaeger RJ & Kralj AR (1989). Coactivation of the hamstrings and quadriceps during extension of the knee. The Journal of Bone and Joint Surgery. 71-A:1075-1081.

11. Duchateau J & Hainaut K (1990). Effects of immobilization on contractile properties, recruitment and firing rates of human motor units. Journal of Physiology. (Lond.) 422:55-65.

12. Enoka RM (1994). Neuromechanical Basis of Kinesiology. Human Kinetics, Champaign.

13. Fuglevand AJ, Winter DA, Patla AE & Stashuk D (1992). Detection of motor unit action potentials with surface electrodes : influence of electrode size and spacing. Biological Cybernetics. 67:143-153.

14. Gander RE & Hudgins BS (1985). Power spectral density of the surface myoelectric signal of the biceps brachii as a function of static load. Electromyography and Clinical Neurophysiology. 25:469-478.

15. Garfinkel S & Cafarelli E (1992). Relative changes in maximal force, EMG, and muscle cross-sectional area after isometric training. Medical Science Sports Exercise. 24:11:1220-7.

16. Guimaraes AC, Herzog W, Allinger TL & Zhang YT (1995). The EMG-force relationship of the cat soleus muscle and its association with contractile conditions during locomotion. Journal of Experimental Biology. 198:975-87.

17. Hägg GM (1992). Interpretation of EMG spectral alterations and alteration indexes at sustained contraction. Journal of Applied Physiology 73: 1211-1217.

18. Hakkinen K & Hakkinen A (1995). Neuromuscular adaptations during intensive strength training in middle-aged and elderly males and females. Electromyography Clinical Neurophysiology. 35:3:137-47.

19. Hamill J & Knutzen KM (1995). Biomechanical basis of human movement. Williams and Wilkins, Baltimore.

20. Howald H (1982). Training-induced morphological and functional changes in skeletal muscle. International Journal of Sports Medicine. 3:1-12.

21. Kamen G & Caldwell GE (1996). Physiology and interpretation of the electromyogram. Journal of Clinical Neurophysiology. 13:366-384.

22. Keating JL & Matyas TA (1996). The influence of subject and test design on dynamometric measurements of extremity muscles. Physical Therapy. 76: 866-889.

23. Kraemer WJ, Duncan ND & Volek JS (1998). Resistance training and elite athletes: adaptations and program considerations. Journal of Orthopaedic and Sports Physical Therapy. 28:2:110-119.

24. Kumar S & Mital A (1996). Electromyography in ergonomics. Taylor & Francis, London.

25. Kupa EJ, Roy SH, Kandarian SC & De Luca CJ (1995). Effects of muscle fiber type and size on EMG median frequency and conduction velocity. Journal of Applied Physiology. 79: 23-32.

26. Kwong KSC (1995). Vibration response analysis in orthopaedics and its application at the lumbar spine. PhD thesis. University of Strathclyde, Glasgow.

27. Leung SCS, Xiao S & Chan KM (1996). Estimating EMG median and mean power frequency to monitor localized muscle fatigue during isokinetic exercise. Proceeding of the International Conference on Biomedical Engineering. June 3-5:210-213.

28. Mannion AF & Dolan P (1996). Relationship between myoelectric and mechanical manifestations of fatigue in the quadriceps femoris muscle group. European Journal of Applied Physiology. 74:411-419.

29. Molina MR, Garcia MSM & Mayoral MLG (2000). Effect of muscular ultrasound stimulation on power spectrum electromyography during a strengthening training. Electromyography Clinical Neurophysiology. 40:163-168.

30. Molina MR, Galen AT & Garcia MSM (1997). Spectral electromyographic changes during a muscular strengthening training based on electrical stimulation. Electromyography Clinical Neurophysiology. 37:287-295.

31. Moritani T (1993). Neuromuscular adaptations during the acquisition of muscle strength, power and motor tasks. Journal of Biomechanics. 26:Suppl. 1:95-107.

32. Moritani T & Masuo M (1987). Motor unit activity and surface electromyogram power spectrum during increasing force of contraction. European Journal of Applied Physiology. 56:260-265.

33. Narici MV, Hoppeler H, Kayser B, Landoni L, Claasen H, Gavarde C, Conti M & Cerretelli P (1996). Human quadriceps cross-sectional area, torque and neural activation during 6 months strength training. Acta Physiology Scandinavia. 157: 175-86.

34. Ng JKF & Richardson CA (1996). Reliability of electromyographic power spectral analysis of back Muscle endurance in healthy subjects. Archives of Physical Medical Rehabilitation. 77:259-264.

35. Ortengren R (1996). Noise and artefacts in "Electromyography in ergonomics". Edited by Kumar S. Taylor & Francis, London.

36. Perotto AO (1994). Anatomical Guide for electromyography. The limbs and trunk. Third edition. Charles C Thomas Publisher, Illinois.

37. Rabita G, Perot C & Lensel-Corbeil G (2000). Differential effect of knee extension isometric training on the different muscles of the quadriceps femoris in humans. European Journal of Applied Physiology. 83:531-538.

38. Roitman JL, Kelsey M, Lafontaine TP, Southard DR, Williams MA & York T (1998). ACSM's Resource Manual for Guidelines for Exercise Testing and Prescription. p.381. 3$^{RD}$ edition. Williams & Wilkins. Baltimore etc.

39. Sale DG (1988). Neural adaptation to resistance training. Medicine and Science in Sports and Exercise. 20:S135-145.

40. Salzman A, Torburn L & Perry J (1993). Contribution of rectus femoris and vasti to knee extension. Clinical Orthopaedics. 290:236-243.

41. Solomonow M, Moritani T & Maton B (1996). Standards for reporting EMG data. Journal of Electromyography & Kinesiology. 6:3:III - IV.

42. Solomonow M, Baratta R, Shoji H & D'Ambrosia R (1990). The EMG-force relationships of skeletal muscle: dependence on contraction rate, and motor units control strategy. Electromyography Clinical Neurophysiology. 30:141-52.

43. Solomonow M, Baratta R, Zhou BH, Shoji H & D'Ambrosia RD (1987). The EMG-force model of electrically stimulated muscles: dependence on control strategy and predominant fiber composition. IEEE Transactions on Biomedical Engineering. 34: 692-703.

44. Weir JP, Housh TJ & Weir LL (1994). Electromyographic evaluation of joint angle specificity and cross-training after isometric training. Journal of Applied Physiology. 77: 197-201.

45. Vredenbregt J & Rau G (1973). Surface electromyography in relation to force, muscle length and endurance, in "New developments in electromyography and clinical neurophysiology. Edited by Desmedt JE. Karger, Basel.

46. Vaatainen U, Airaksinen O, Jaroma H & Kiviranta I (1995). Decreased torque and electromyographic activity in the extensor thigh muscles in chondromalacia patellae. International Journal of Sports Medicine. 16:1:45-50.

# Appendices

Appendix I: Technical specifications

The Nihon Kohden Memory Oscilloscope VC-11 is a specialized medical oscilloscope developed to meet various needs in research and experimentation in biological phenomena.

**Features of the VC-11**

1. Built-in 4-channel memory
- Stored data can be displayed on the CRT as a frozen waveform.
- A maximum of four waveforms can be stored, allowing for comparison between waveforms.
- High-fidelity waveforms are reproducible by a 10 bit A/D conversion.

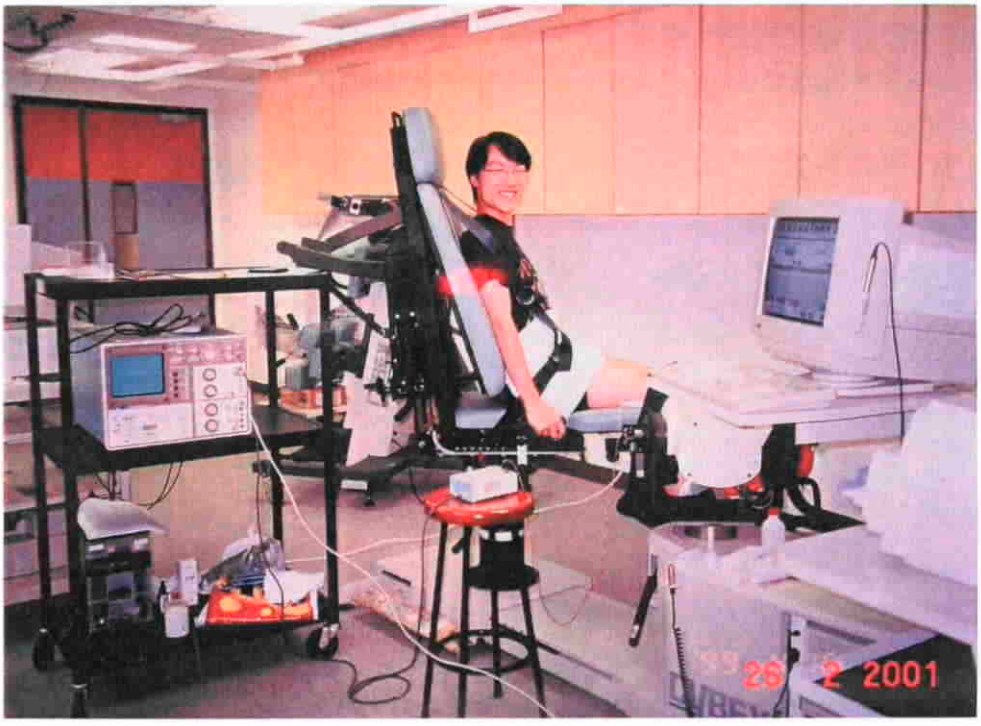
2. Advanced plug-in units available
- The Dual-channel Biophysical Amplifier AVB-21 is composed of two channels of the AC amplifiers provided in the AVB-11 module.
- The Biophysical Amplifier AVB-11 is a preamplifier designed to measure a variety of electrophysiological phenomena as well as DC signals.

Features of the Dual-channel Biophysical Preamplifier (AVB-21)

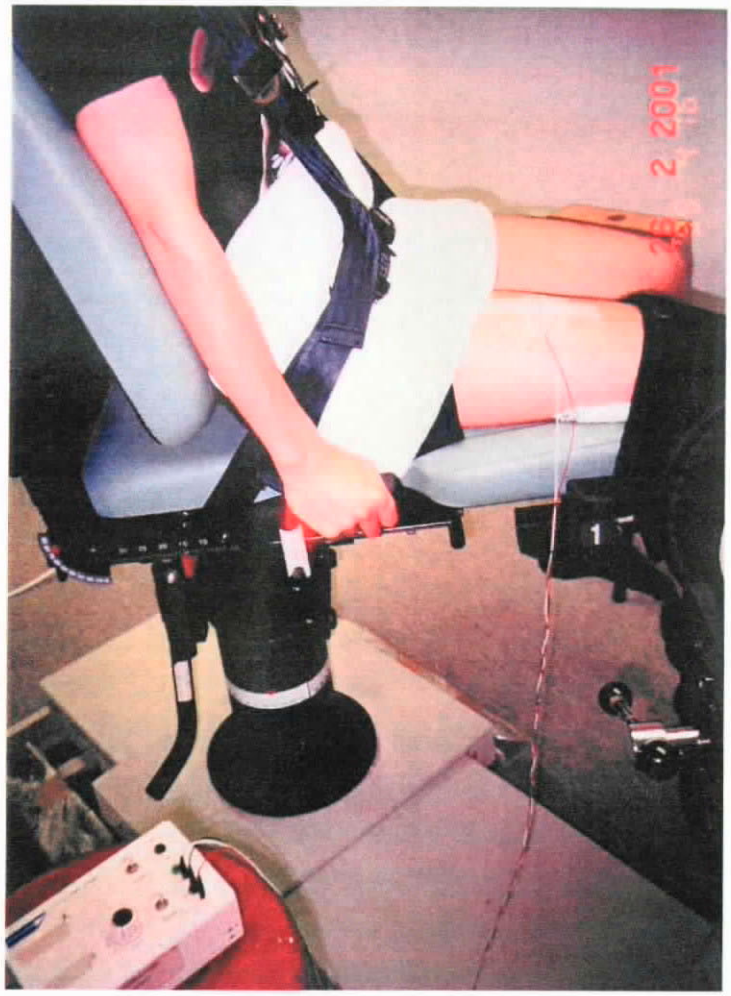
| Functions | Options | Features |
|---|---|---|
| Input Circuitry | AC | Differential (via JB-210J)<br>180 MΩ – E – 180 MΩ |
| Input Circuit Current | — | 2 x 10 A or less |
| Sensitivity | AC | 10 μV – 10 mV/ DIV<br>± 3 % |
| Common-Mode-Rejection Ratio | AC | 80 dB or greater |
| Frequency Response | LO-CUT | DC, 0.08, 0.5, 1.5, 5, 15, and 50 Hz |
| | HI-CUT | 30, 100, 300, 1k, 3k, and 10 k Hz |
| Internal Noise Level | — | 5 μV p-p or less |
| Input Signal Off-set Capability | — | None |

Appendix II: Experimental set-up

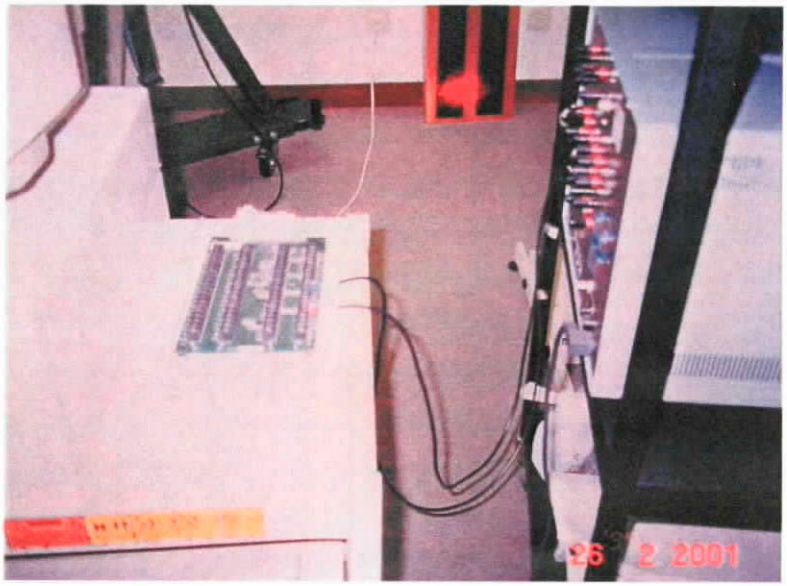

The complete setup consists of the computerized dynamometer (most right), the oscilloscope (middle) and the personal computer (left).




Another view of the computerized dynamometer and the oscilloscope




A close-up view of the position of the electrodes and the junction box (AVB-21) that contains the preamplifier




The EMG data acquired in the oscilloscope is transferred to the PC-based analogue-to-digital system.