

Design and evaluation of a new filter-based fatigue index

Jungyoon Kim

The Graduate School

Yonsei University

Department of Biomedical Engineering

Design and evaluation of a new filter-based fatigue index

A Dissertation

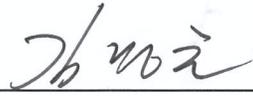
Submitted to the Department of Biomedical Engineering
and the Graduate School of Yonsei University

in partial fulfillment of the
the requirements for the degree of
Doctor of Philosophy

Jungyoon Kim

August 2014

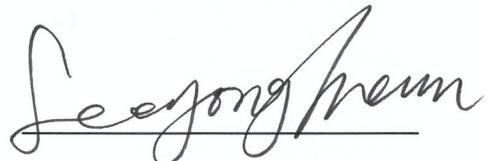
This certifies that the dissertation of Jungyoon Kim is approved.



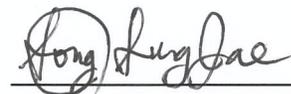
Thesis Supervisor: Prof. Youngho Kim



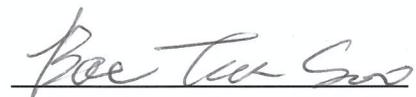
Thesis Committee Member: Prof. Kyoungjung Lee



Thesis Committee Member: Prof. Yongheum Lee



Thesis Committee Member: Prof. Sungjae Song



Thesis Committee Member: Prof. Taesoo Bae

The Graduate School

Yonsei University

August 2014

Acknowledgments

At first, I would like to thank my supervisor, Professor Youngho Kim for giving me the opportunity of studying this theme and for all kinds of support that I achieved. I am also thankful to my dissertation committee members, Professor Kyoungjoung Lee, Professor Yongheum Lee, Professor Sungjae Song, and Professor Taesoo Bae for thoughtful suggestions and advices. In addition, I thank over other professors of the Department of Biomedical Engineering in Yonsei University for their sincere teaching.

I would like to thank all members of Biomechanics & Rehabilitation Lab. for all the supports and motivation. They are as important as my family in my life.

I thank my colleagues in Yonsei University for providing motivation during the research of biomedical engineering. I also thank all of my friends

Finally and most of all, I would like to thank my family for continuous love and generous support. Nothing I can say can adequately express my gratitude for the adoration, support and encouragement they provided throughout my life.

Thank you!

Jungyoon Kim

Contents

Acknowledgements	i
Contents	ii
List of Figures	vi
List of tables	ix
Abstract	xi
1. Introduction	1
1.1 Muscle fatigue	1
1.2 Measurement of muscle fatigue	3
1.3 Previous EMG-based fatigue indices	4
1.4 Aim of the research	6
1.5 Outline of the dissertation	6
2. Design of filter-based fatigue index	7
2.1 Purpose	7
2.2 Experimental methods	8
2.2.1 Subjects	8

2.2.2	Experimental procedures	8
2.2.3	Data acquisition	9
2.2.4	Data analysis	9
2.2.5	Statistical analysis	12
2.3	Results	11
2.3.1	Changes in power output	11
2.3.2	Changes in EMG indices	12
2.3.3	Relationships between EMG indices and power output	13
2.4	Discussion	15

3.	Bandwidth optimization of filter-based fatigue index during isotonic exercise	19
3.1	Purpose	19
3.2	Experimental methods	20
3.2.1	Subjects	20
3.2.2	Experimental procedures	20
3.2.3	Data acquisition	21
3.2.4	Data analysis	21

3.2.5 Statistical analysis	23
3.3 Results	24
3.3.1 Joint power output	24
3.3.2 EMG indices of rectus femoris (RF) muscle	26
3.3.3 EMG indices of vastus muscle (VM, VL)	27
3.3.4 Relationship between EMG indices and power output	30
3.4 Discussion	32
4. Validation of filter-based fatigue index	36
4.1 Purpose	36
4.2 Experimental methods	37
4.2.1 Subjects	37
4.2.2 Experimental procedures	38
4.2.3 Data acquisition	40
4.2.4 Data analysis	41
4.2.5 Statistical analysis	42
4.3 Results	44
4.3.1 Bandwidth optimization in various muscles	44

4.3.2 Bandwidth optimization during different exercises	45
4.3.3 Bandwidth optimization at various inter-electrode distances	47
4.4 Discussion	49
5. Discussion and conclusion	53
5.1 Discussion	53
5.2 Conclusion	56
References	59
Appendices	70
Abstract (in Korean)	79

List of Figures

Fig. 2.1. Changes in peak power and sEMG parameters (mean \pm standard deviation) during the five sets of a maximum of ten repetitions. (a) Peak power and (b) root mean square (RMS) values of the rectus femoris; (c) median frequency (F_{med}) of the rectus femoris; (d) 400 Hz cut-off new fatigue index (FI_{hr400}) of the rectus femoris; (e) 320 Hz cut-off new fatigue index (FI_{hr320}) of the rectus femoris; (f) 240 Hz cut-off new fatigue index (FI_{hr240}) of the rectus femoris (*significant differences at $p < 0.05$).

Fig. 2.2. Linear representations of changes in peak power (normalized by the first two repetitions) versus changes in different sEMG parameters (normalized by the first two repetitions) for all subjects. (a) Changes in power output versus changes in RMS value. (b) Changes in power output versus changes in F_{med} . (c) Changes in power output versus changes in FI_{hr400} . (d) Changes in power output versus changes in FI_{hr380} . (e) Changes in power output versus changes in FI_{hr360} . (f) Changes in power output versus changes in FI_{hr340} .

Fig. 3.1. Peak power and EMG indices of rectus femoris muscle (mean \pm standard deviation) during five sets of ten repetitions. (a) Peak power; (b) RMS values of sEMG signal; (c) median frequency (F_{med}); (d) median frequency after continuous wavelet transform (CWT & F_{med}); (e) the logarithm of Dimitrov's index (FI_{nsm5}); (f) the logarithm of Gonzalez-Izal's index (WIRE51); (g) bandwidth-optimized fatigue index (FI_{hrOPT}) (*significant differences at $p < 0.05$).

Fig. 3.2. Peak power and EMG indices of average of vastus muscles (mean \pm standard deviation) during five sets of ten repetitions. (a) RMS values of sEMG signal; (b) median frequency (F_{med}); (c) median frequency after continuous wavelet transform (CWT & F_{med}); (d) the logarithm of Dimitrov's index (FI_{nsm5}); (e) the logarithm of Gonzalez-Izal's index (WIRE51); (f) bandwidth-optimized fatigue index (FI_{hrOPT}) (*significant differences at $p < 0.05$).

Fig. 3.3. Linear representations of peak power output (normalized by the first two repetitions) versus various EMG indices (normalized by the first two repetitions) for all subjects. (a) Peak power output versus RMS values of sEMG signal; (b) peak power output versus F_{med} ; (c) peak power output versus CWT & F_{med} ; (d) peak power output versus the logarithm of FI_{nsm5} ; (e) peak power output versus the logarithm of WIRE51; (f) peak power output versus FI_{hrOPT} .

Fig. 4.1. Photograph of various inter-electrode distances (2cm, 6cm and 10cm)

Fig. 4.2. Optimized cut-off frequencies for various muscles during isotonic exercises. Optimized cut-off frequencies were 353.3 ± 49.5 Hz in RF muscle, 343.9 ± 34.2 Hz in VM muscle, 353.7 ± 36.1 Hz in VL muscle, 362.3 ± 28.2 Hz in BB muscle, 357.5 ± 46.0 Hz in TC muscle, and 348.1 ± 38.2 Hz in TA muscle.

Fig. 4.3. Optimized cut-off frequencies for different exercises in the rectus femoris (RF) muscle. Optimized cut-off frequencies were 353.3 ± 49.5 Hz in isotonic exercise, 354.0 ± 46.3 Hz in isokinetic exercise, and 357.4 ± 45.4 Hz in isometric exercise.

List of Tables

Table 2.1. Correlation coefficients between changes in the EMG parameters and the corresponding changes in peak power output

Table 3.1. Optimized cut-off frequencies and correlation coefficients

Table 3.2. Correlation coefficients between EMG indices and peak power output

Table 4.1. Characteristic of six kinds of exercises

Table 4.2. Correlation coefficients between EMG parameters and peak power output in various muscles

Table 4.3. Correlation coefficients between EMG parameters and peak power output or joint torque for different exercises

Table 4.4. Optimized cut-off frequencies and correlation coefficients between FI_{hrOPT} and the peak power output or joint torque in various inter-electrode distances

Table 4.5. Correlation coefficients between FI_{hrOPT} and peak power output or joint torque at various inter-electrode distances

Abstract

Design and evaluation of a new filter-based fatigue index

Jungyoon Kim

Dept. of Biomedical Engineering

The Graduate School

Yonsei University

In this study, new filter-based fatigue index was designed to estimate muscle fatigue.

Muscle fatigue is a temporary loss of strength and energy resulting from hard physical work. It is a hindrance during exercise, as it causes dysfunction. Thus muscle fatigue plays an important role in prolonged contractile activity-induced skeletal muscle injury. Because of these reasons, the measurement of muscle fatigue is very important.

Filter-based fatigue index, which is the ratio of high-frequency to low-frequency components of EMG power, was designed to assessing muscle fatigue. To determine the frequency bandwidth of the designed fatigue index, cut-off frequencies of the high-pass-filter were optimized to maximize the

correlation coefficient between the peak power or joint torque and the designed filter-based fatigue index. To evaluate the designed filter-based fatigue index, accuracy of muscle fatigue estimation was compared with other fatigue indices. The designed fatigue index was validated in various muscles, different exercises, and different inter-electrode distances.

Results, The filter-based fatigue index showed the best correlation with biomechanical fatigue over other fatigue indices in all conditions. This implies the filter-based fatigue index could assess muscle fatigue accurately than other fatigue indices. In addition, optimized cut-off frequencies in various muscles, under different exercises and various inter-electrode distances were consistent about 350 Hz. Consistency of the optimized cut-off frequencies in all conditions implied that 350Hz HPF could be a general solution for the filter-based fatigue index and the filter-based fatigue index reflects firing rate of motor units. Overall, these findings may be useful for muscle fatigue assessment.

One of the main finding of this study was that filter-based fatigue index reflects the firing rate of motor units. However, firing rates of motor units were not measured during fatigue. In further studies, MUAP decomposition methods or intramuscular EMG methods would be applied to measure the firing rate of motor units.

Key words: surface electromyography, digital filter, fatigue index, frequency band, cut-off frequency, optimization, validation, various muscles, different exercises, inter-electrode distance

Chapter 1

1. Introduction

1.1 Muscle fatigue

Muscle fatigue is a temporary loss of strength and energy resulting from hard physical work. Muscle fatigue has been defined as “a response that is less than the expected or anticipated contractile response, for a given stimulation” (MacIntosh and Rassier, 2002). Hard physical exercise or prolonged physical work involving 30% to 40% of maximal aerobic capacity induces muscle fatigue (Astrand and Rodahl, 1986). Muscle fatigue is a hindrance during exercise, as it causes dysfunction (Ishii and Nichida, 2013).

In industries and transportation, frequent and repetitive sub-maximal contraction and incorrect posture leads to many disorders of muscles, tendons and nerves (<http://www.ccohs.ca/oshanswers/diseases/rmirsi.html>). Repetition of same task and prolonged abnormal posture contributes to upper limbs and neck disorders (Walker-Bone and Cooper, 2005; Van et al, 2009). In sports, muscle fatigue plays an important role in prolonged contractile activity-induced skeletal muscle injury (Hägglund et al, 2013). In transportation industry, fatigue related accidents are major killers (Sung et al., 2005). 25% of

single vehicle accidents are direct result of driver's fatigue. Because of these reasons, understanding of muscle fatigue is important.

With fatigue, muscular response declines because less calcium is released, limiting the number of attached actin-myosin bridge connections of actin-myosin (Sesboüé and Guincestre, 2006). On a metabolic level, reduced pH and increased intracellular lactate levels are the most frequent changes (Arabadzhev et al., 2010).

In terms of electrical response, electromyography (EMG) signals reflect both central and peripheral fatigue. During maximal isometric contraction, EMG signal amplitudes decrease due to reductions in motor output from the central nervous system (Bigland-Ritchie et al., 1983a). Due to peripheral factors, EMG signal amplitudes may increase (Masuda et al., 1999). Thomas et al. (2006) explained the elevated peak-to-peak amplitudes and areas of motor unit action potentials (MUAPs) by invoking increased Na^+/K^+ pump activity.

1.2 Measurement of muscle fatigue

Various methods have been used to assess muscle fatigue. First, joint torque or power can reflect muscle fatigue (Fuglevand et al., 1999; Thomas et al., 1991). The speed of joint movement slows down naturally as fatigue increases during a classical training session (Gonzalez-Izal et al., 2010a), and joint torque decreases as fatigue increases during isokinetic exercise. This method can measure the decreased contractile response directly. However, joint torque may not decrease during sub-maximal exercise even if the muscles are fatigued. Furthermore, joint motion is limited during the measurement of joint torque or power because of the measuring devices. Second, blood lactate concentrations are generally used to assess muscle fatigue (Amann et al., 2010; Coco et al., 2010). Muscular contractions can cause lactate and hydrogen ion concentrations to increase to very high levels during physical exercise (Gladden, 2004). A larger number of fast twitch fibers is employed with increasing exercise intensity and these fibers are more suited to the production of lactate (Facey et al., 2013). This method needs a blood sample to analyze lactate concentrations. It cannot target the muscle, and cannot be measured during exercise. Thirdly, surface electromyography (sEMG) is a non-invasive fatigue assessment method. Muscle fiber conduction velocity (MFCV) decreases and the EMG power spectrum shifts to lower frequencies during peripheral fatigue (Cifrek et al., 2009; Farina et

al., 2004; Farina, 2006). This method can assess the fatigue of individual superficial muscles. This research focused on sEMG-based fatigue assessment.

1.3 Previous EMG-based fatigue indices

In isometric fatiguing conditions, the EMG power spectrum shifted to the lower frequencies, and muscle fiber conduction velocity (MFCV) decreased (Masuda et al., 1999; Cifrek et al., 2009). Mean and median frequencies (F_{mean} and F_{med}) are generally used to detect these effects. However, MFCV did not decrease during submaximal dynamic exercises (Masuda et al., 1999), and thus F_{mean} and F_{med} have a relatively low sensitivity under dynamic exercise conditions (Farina et al., 2004; Farina, 2006). Various methods have been attempted to overcome this problem (Karlsson et al., 2000; Dimitrov et al., 2006; Gonzalez-Izal et al., 2010b). Karlsson et al. (2000) compared various time-frequency analysis methods (i.e. the short-time Fourier transform, the Wigner-Ville distribution, the Choi-Williams distribution and the wavelet transform) during dynamic exercises, and found that the estimations provided by the continuous wavelet transform (CWT) showed the best accuracy. Dimitrov et al. (2006) proposed the high-sensitivity spectral index as the ratio between the EMG power spectral moment of order (-1), and the EMG power spectral moment of order 5. They showed that their index $FI_{\text{nsm}5}$ increased eightfold from the initial value and F_{med} decreased only 32% from the initial value during cycling exercises. Gonzalez-Izal et al. (2010b) combined the

discrete wavelet transforms (DWT) and the ratios between different spectral moments. They confirmed that their index WIRE51 showed a good correlation with power output and the robustness against noise during leg press exercises. However, large amounts of computational load and time-frequency resolution were the drawbacks of those methods. Ratios of the EMG power content in high and low frequency ranges have been adopted as indices of peripheral muscle fatigue (Lee et al., 2009; Chaffin, 1973; Kogi and Hakamada, 1962; Sato, 1982; Moxham et al., 1982; Bellemare and Grassino, 1982). However, there have been difficulties in selecting bandwidth in boundary frequency, high-frequency, and low-frequency, and these difficulties are evident in the widely varying bandwidth choices across studies (30 Hz (Chaffin, 1973), 40 Hz (Kogi and Hakamada, 1962), or 64 Hz (Sato, 1982); 130–238 vs. 20–40 Hz (Moxham et al., 1982), 150–350 vs. 20–46.7 Hz (Bellemare and Grassino, 1982)). The difficulties arise from causes such as the EMG power spectrum shifting not only due to muscle fatigue, but also due to the distance between muscle fibers and electrode distances, and the use of inaccurate electrode attachment sites (Dimitrov and Dimitrova, 1998).

1.4 Aim of the study

The objective of this research is design of a new filter-based fatigue index which has the lower computational load and high time resolution. To improve the filter-based fatigue index, optimization method is used to determine the bandwidth of high-frequency components. Then, filter-based fatigue index is tested in various conditions for validation.

1.5 Outline of the dissertation

This dissertation is composed of 5 chapters, including the present one.

Chapter 2 presents the design of a new filter-based fatigue index, which is the ratio of high-frequency to low-frequency components of EMG power, for assessing muscle fatigue. In the Chapter 3, cut-off frequencies of the high-pass-filter were optimized to maximize the correlation coefficient between the peak power and the designed filter-based fatigue index. The frequency bandwidth is determined as optimized cut-off frequencies. Chapter 4 presents the validation of designed filter-based fatigue index in various muscles, different exercises, and different inter-electrode distances. The designed filter-based fatigue index was compared with other fatigue indices. Chapter 5 presents the discussions and the conclusions.

Chapter 2

2. Design of a filter-based fatigue index

2.1 Purpose

In contrast to isokinetic fatigue assessment (i.e. with constant velocity adjustments by the isokinetic device) during muscle daily function or in a classical training session (i.e. 3–5 sets of 10 repetition maximum) the speed of the repetitions slows down naturally as fatigue increases (i.e. with variable velocity adjustments). Indeed, hypothesis was that EMG fatigue quantification of peripheral impairments might provide useful information of the loss of power output during dynamic contractions.

In this study, new filter-based fatigue index, which is the ratio of high-frequency to low-frequency components of EMG power, was designed and evaluated to assess muscle fatigue. The designed fatigue index was compared with the conventional F_{med} and EMG amplitude.

2.2 Experimental methods

2.2.1 Subjects

Twenty healthy males (age: 25.6 ± 3.1 yrs, height: 172.8 ± 6.1 cm, mass: 73.3 ± 10.1 kg) were recruited for this study. None of the subjects participated in any systemic physical training. Subjects were informed about the experimental procedure and the purpose of the study. All subjects provided written informed consent to participate in the experiment. To familiarize the subjects with the study procedures, three to four voluntary submaximal isometric and isotonic contraction practice sessions were completed before the actual experiments.

2.2.2 Experimental procedures

The subjects performed the knee extension/flexion exercises on the isokinetic dynamometer (Biodex System 3, Biodex Medical Systems, USA) to measure the maximum torque and the maximal voluntary contraction (MVC). Full extension of the knee joint was set to 0° , and the maximum knee flexion was 115° . EMG signals and the maximum torque during MVC were measured at a knee angle of 90° , and each subject performed the maximal knee extension ten times to determine the ten repetition maximum (10RM) torque. After a sufficient break (about ten minutes), the measured 10RM was $61.5 \pm 4.6\%$ of the maximum torque and this value was used for fatigue exercises. The experimental protocol was designed according to the acute heavy-

resistance exercise protocol (AHREP) and consisted of five sets of ten knee extensions with two minutes of rest between sets. During the fatigue exercises, the subjects were asked to extend their knee as quickly as possible.

2.2.3 Data acquisition

An EMG signal was obtained from the right rectus femoris (RF) muscle using a Noraxon EMG System (MyoSystem 1200, Noraxon Inc., USA). The EMG signal and biomechanical signals (angle, angular velocity, and torque) acquired from the isokinetic dynamometer were recorded simultaneously at a sampling rate of 1 kHz. The EMG signal was recorded by a standard active bipolar surface electrode (14 mm in diameter, 18 mm inter-electrode distance; Noraxon, USA). After careful preparation of the skin (shaving, abrasion, and cleaning with alcohol), electrode pairs were placed longitudinally on the middle portion of the muscle, based on SENIAM (Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles) recommendations.

2.2.4 Data analysis

Data analysis was performed off-line using MATLAB R2010a software (MathWorks Inc., Natick, Massachusetts, USA). MVC torque was calculated as the average torque over a 1-s period during the torque-plateau level of each MVC attempt. The peak power output was calculated for each exercise repetition as the product of joint torque and angular velocity. Ten segments

from the raw EMG signal for each set were extracted, with each segment corresponding to a knee extension repetition. The parameters analyzed in this study corresponded to each extracted segment.

The following EMG parameters were calculated:

- (1) Mean RMS was calculated using a 50-ms moving root-mean-squared filter (Roy et al., 1986), as the integrated EMG divided by integration time.
- (2) F_{med} was calculated using the following equation:

$$\int_{f_1}^{F_{med}} PS(f) \cdot df = \int_{F_{med}}^{f_2} PS(f) \cdot df \quad (2.1)$$

where $PS(f)$ is the spectral power for the current frequency f , and f_1 (=10 Hz) and f_2 (=500 Hz) are the determined bandwidths of the EMG signals.

- (3) The proposed filter-based fatigue index was calculated as follows:

$$FI_{hlrx} = \int_{t_{start}}^{t_{end}} \frac{LE_{xHzHPF}(t)}{LE_{120HzLPF}(t)} dt / (t_{end} - t_{start}) \quad (2.2)$$

where t_{start} and t_{end} are the start and end time of each exercise repetition, respectively. $LE_{120HzLPF}(t)$ is the second-order Butterworth 3 Hz linear-enveloped signal of the 120 Hz low-pass-filtered EMG signal, and $LE_{xHzHPF}(t)$ is the second-order Butterworth 3 Hz linear-enveloped signal of the repeated high-pass-filtered EMG signal varied from 200 Hz to 400 Hz in

increments of 20 Hz. FI_{hlrx} was calculated as the average knee extension time to compare with other fatigue indices.

2.2.5 Statistical analysis

All the variables (peak power output and EMG-based parameters) were normalized with the first two contractions of the first set of exercises. Pearson's correlation analysis was used to analyze the relationship between the peak power and the different EMG parameters. One-way analysis of variance (ANOVA) was used to calculate significant differences in the parameters recorded for five consecutive repetitions. $P < 0.05$ was considered as the indicative of statistical significance.

2.3 Results

2.3.1 Changes in power output

Fig. 2.1(a) shows joint power outputs during the experiment. For each subject, joint power outputs decreased significantly with the number of sets as well as of the repetitions in each set. Joint power outputs from the last five repetitions of the fifth set was 47% lower than those from the first set ($p < 0.05$, Fig. 2.1(a)). Joint power outputs of the last five repetitions in each set were significantly smaller than those from the first five repetitions ($p < 0.05$). Joint power outputs from the first five repetitions of the third, fourth, and fifth sets

was significantly smaller than those from the first five repetitions of the first set ($P < 0.05$). In addition, joint power outputs from the first five repetitions of the fourth and fifth sets were significantly smaller than those from the first five repetitions of the second set ($p < 0.05$) and first five repetitions of the fifth set were significantly smaller ($p < 0.01$) than those from the first five repetitions of the third set.

2.3.2 Changes in EMG indices

Fig. 2.1(b) shows the RMS values of EMG signals in this study. From the first to third set, the last five repetitions had larger RMS values of EMG signals than the first five repetitions ($p < 0.05$, Fig. 2.1(b)). In the fourth and fifth sets, the RMS values of the last five repetitions were larger, but the difference was not statistically significant.

F_{med} was significantly smaller in the last five repetitions than in the first five repetitions for every set ($p < 0.01$, Fig. 2.1(c)). There was no statistically significant difference between F_{med} from the first five repetitions of the first set and the first five repetitions of other sets.

FI_{hrx} was significantly smaller for last five repetitions than first five repetitions in all cut-off frequencies for every set ($p < 0.05$, Fig. 2.1(d-f)). For cut-off frequency of 240 Hz or above, FI_{hrx} from the first five repetitions of the first set and the first five repetitions of the last set showed a statistically significant difference ($p < 0.05$). For cut-off frequency of 320 Hz or above,

FI_{hrx} from the first five repetitions of the first set and the first five repetitions of third, fourth, and fifth sets showed a statistically significant difference ($p < 0.05$). In FI_{hr400} , the first five repetitions of the second set and the first five repetitions of the last set showed a statistically significant difference ($p < 0.05$).

2.3.3 Relationships between EMG indices and power output

The EMG parameter with the strongest correlation with mechanical power was FI_{hr360} ($R=0.749$ (± 0.081); $p < 0.01$, Fig. 2.2(c)). For a cut-off frequency of 320 Hz or above, the correlation coefficient was 0.74 (Fig. 2.2(c-f)) or greater, and the decrease in frequency was accompanied by decreases in the correlation coefficient. F_{med} and RMS had correlation coefficients of 0.632 (± 0.164) and -0.135 (± 0.358), respectively (Table 2.1 and Fig. 2.2(a-b)).

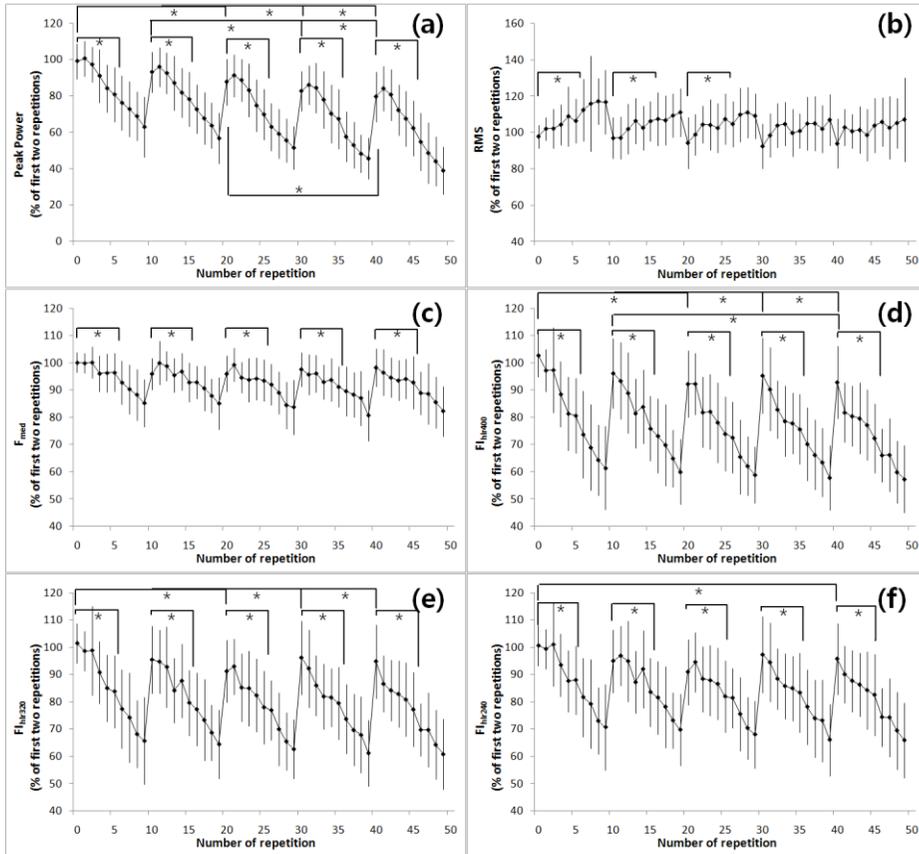


Fig. 2.1. Changes in peak power and sEMG parameters (mean \pm standard deviation) during the five sets of a maximum of ten repetitions. (a) Peak power and (b) root mean square (RMS) values of the rectus femoris; (c) median frequency (F_{med}) of the rectus femoris; (d) 400 Hz cut-off new fatigue index (FI_{hlr400}) of the rectus femoris; (e) 320 Hz cut-off new fatigue index (FI_{hlr320}) of the rectus femoris; (f) 240 Hz cut-off new fatigue index (FI_{hlr240}) of the rectus femoris (*significant differences at $p < 0.05$).

2.4 Discussion

In this study, ratios of EMG power content in the high and low frequency ranges were calculated using a FIR filter, not by a frequency analysis. This technique has the advantage of addressing time-frequency resolution issues and also reducing the computational load.

Results revealed that new fatigue indices, FI_{hlrx} , showed better correlation with changes in joint power in multiple high-intensity dynamic exercises. RMS values of EMG signals have increasing tendency with the muscle fatigue, as the previous studies (Masuda et al., 1999; Potvin and Bent, 1997), which is originated from the additional recruitment of motor unit (moritani et al., 1982), rise in the motor unit firing frequency (Bigland-Ritchie et al., 1983b), and the lengthening of intracellular action potential typical for fatiguing contractions (Komi and Tesch, 1979).

Many studies reported that the reduction in joint power is accompanied by the reduction in median frequency (Masuda et al., 1999; Gonzalez-Izal et al., 2010a; Dimitrov et al., 2006), which is due to the lengthening effect of motor unit action potential (Bigland-Ritchie et al., 1981) and reduced activation of fast twitch muscle fibers (Komi and Tesch, 1979).

Joint power and median frequency had the correlation coefficient of 0.632, implying that the median frequency would reflect the muscle fatigue well. The present study also supported many previous studies, stating that the median frequency could be used to estimate the muscle fatigue in high-

intensity dynamic exercises (Potvin and Bent, 1997; Ament et al., 1996; Gerdle et al., 2000). In FI_{hrx} values, FI_{hr360} was found to have the highest correlation (0.749) with joint power, and the correlation coefficient was 0.74 or greater for 320 Hz or above. This implied that FI_{hrx} could evaluate the muscle fatigue at high-intensity exercises well.

As in the results of previous reports, there were no significant differences in F_{med} for the first five repetitions of every set (Gonzalez-Izal et al., 2010a). However, FI_{hrx} differed significantly in the first five repetitions of each set, reflecting the reduction in joint power from the first five repetitions of every set. These findings also indicate that FI_{hrx} could serve as a means to assess muscle fatigue even under a low-fatigue state. FI_{hr400} showed the best results for identifying differences in the first five repetitions of each set.

The proposed filter-based fatigue index, FI_{hrx} , was applied to the rectus femoris muscle alone and did not take into account electrode sizes, inter-electrode distance, or electrode type. All of these parameters should be considered to determine the cut-off frequency of HPF.

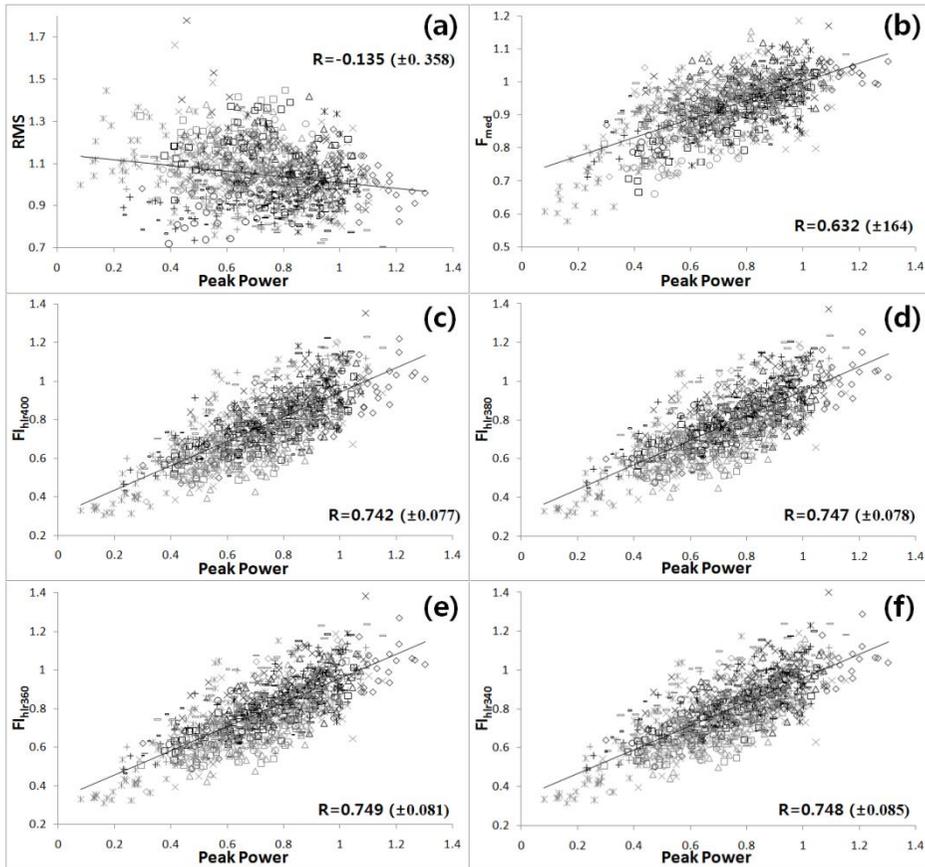


Fig. 2.2. Linear representations of changes in peak power (normalized by the first two repetitions) versus changes in different sEMG parameters (normalized by the first two repetitions) for all subjects. (a) Changes in power output versus changes in RMS value. (b) Changes in power output versus changes in F_{med} . (c) Changes in power output versus changes in FI_{hlr400} . (d) Changes in power output versus changes in FI_{hlr380} . (e) Changes in power output versus changes in FI_{hlr360} . (f) Changes in power output versus changes in FI_{hlr340} .

Table 2.1. Correlation coefficients between changes in the EMG parameters and the corresponding changes in peak power output

	F_{med}	FI_{hr400}	FI_{hr380}	FI_{hr360}	FI_{hr340}	FI_{hr320}	FI_{hr300}	FI_{hr280}	FI_{hr260}	FI_{hr240}	FI_{hr220}	FI_{hr200}	Peak power
RMS	-0.350 ± 0.235	-0.540 ± 0.300	-0.538 ± 0.300	-0.534 ± 0.301	-0.528 ± 0.301	-0.522 ± 0.300	-0.517 ± 0.298	-0.511 ± 0.297	-0.505 ± 0.295	-0.497 ± 0.292	-0.489 ± 0.288	-0.480 ± 0.284	-0.135 ± 0.358
F_{med}	-	0.745 ± 0.096	0.761 ± 0.092	0.773 ± 0.090	0.783 ± 0.086	0.793 ± 0.082	0.800 ± 0.080	0.807 ± 0.079	0.813 ± 0.078	0.820 ± 0.077	0.827 ± 0.075	0.834 ± 0.074	0.632 ± 0.164
FI_{hr400}		-	0.997 ± 0.002	0.991 ± 0.005	0.982 ± 0.010	0.973 ± 0.014	0.964 ± 0.018	0.954 ± 0.023	0.945 ± 0.028	0.934 ± 0.033	0.922 ± 0.040	0.908 ± 0.046	0.742 ± 0.077
FI_{hr380}			-	0.998 ± 0.001	0.992 ± 0.004	0.985 ± 0.008	0.977 ± 0.011	0.969 ± 0.017	0.960 ± 0.022	0.950 ± 0.028	0.938 ± 0.034	0.925 ± 0.041	0.747 ± 0.078
FI_{hr360}				-	0.998 ± 0.001	0.993 ± 0.003	0.987 ± 0.007	0.980 ± 0.011	0.972 ± 0.017	0.963 ± 0.022	0.952 ± 0.029	0.939 ± 0.035	0.749 ± 0.081
FI_{hr340}					-	0.998 ± 0.001	0.994 ± 0.003	0.989 ± 0.007	0.982 ± 0.012	0.973 ± 0.017	0.964 ± 0.024	0.952 ± 0.030	0.748 ± 0.085
FI_{hr320}						-	0.999 ± 0.001	0.995 ± 0.004	0.989 ± 0.008	0.982 ± 0.013	0.973 ± 0.019	0.963 ± 0.024	0.744 ± 0.092
FI_{hr300}							-	0.999 ± 0.001	0.995 ± 0.004	0.989 ± 0.008	0.982 ± 0.013	0.972 ± 0.018	0.738 ± 0.101
FI_{hr280}								-	0.999 ± 0.001	0.995 ± 0.004	0.989 ± 0.007	0.980 ± 0.012	0.731 ± 0.112
FI_{hr260}									-	0.999 ± 0.001	0.994 ± 0.003	0.988 ± 0.007	0.723 ± 0.121
FI_{hr240}										-	0.998 ± 0.001	0.994 ± 0.003	0.714 ± 0.132
FI_{hr220}											-	0.998 ± 0.001	0.703 ± 0.143
FI_{hr200}												-	0.690 ± 0.153

Chapter 3

3. Bandwidth optimization of the new filter-based fatigue index during isotonic exercise

3.1 Purpose

In this study, cut-off frequencies of the high-pass-filter was optimized to maximize the correlation coefficient between the peak power and the proposed filter-based fatigue index then determined the frequency bandwidth of the new fatigue index. The new fatigue index was compared with the conventional median frequency, the median frequency after the CWT, the spectral parameter proposed by Dimitrov (2006) and the wavelet index proposed by Gonzalez-Izal (2010b).

3.2 Experimental methods

3.2.1 Subjects

Thirty-two healthy males (age: 26.4 ± 4.4 years, height: 171.4 ± 6.3 cm, weight: 76.5 ± 12.4 kg) volunteered as subjects. Five subjects were excluded because of fatigue exercise failure. None of subjects had been involved in any regular strength training program. All subjects were informed of the purpose, the experimental procedure and the potential risk of this study. Before actual experiments, three or four voluntary isometric and isotonic contraction practices were performed to familiarize the subjects with the experimental procedures. All subjects provided written informed consent before participating in the study. Yonsei University research ethics committee reviewed and approved the proposed experimental design (1041849-201405-BM-022-02).

3.2.2 Experimental procedures

Subjects performed knee extension exercises on an isokinetic dynamometer (Biodex System 3, Biodex Medical Systems, USA) to measure the maximal voluntary contraction (MVC). Each subject's body was stabilized on the chair in the seated position with a hip flexion angle of 80° . The range of motion of the knee joint was from 115° to full extension 0° . The fatigue exercise protocol for this study was based on the previous study with one acute heavy-resistance exercise protocol (AHREP). The protocol consisted of

5 sets of 10 repetition maximum ($62.3 \pm 4.4\%$ of the maximum torque in this study) knee extensions with 2 minutes of rest between sets. Before fatigue exercises, subjects performed 20 extensions with 40% of the maximum torque as a warm-up. Fatigue exercise started after a sufficient break (about 10 minutes).

3.2.3 Data acquisition

Surface electromyography (sEMG) signals during knee extension were recorded from right rectus femoris (RF), vastus medialis (VM), and vastus lateralis (VL) muscles, using Noraxon EMG system (MyoSystem 1200, Noraxon Inc., USA) and standard active bipolar surface electrodes (14mm in diameter, 18mm inter-electrode distance). EMG electrode pairs were placed on the muscle belly after shaving, abrasion and skin cleaning according to the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) recommendations. Knee joint kinematics and joint torque were obtained by the isokinetic dynamometer at the sampling rate of 1 kHz.

3.2.4 Data analysis

Data analysis was performed using the MATLAB R2010a software (MathWorks Inc., USA). Corresponding to a knee extension repetition, 10 segments were extracted from raw signals for each set. EMG indices of VM and VL muscles were averaged (Gonzalez-Izal et al., 2010a).

The following EMG indices were calculated for each segment:

- (1) Peak joint power output, mean root-mean-square (RMS) and F_{med} were calculated based on the previous studies (Kim et al., 2012).
- (2) Median frequency after the continuous wavelet transform (CWT & F_{med}) was calculated by

$$\int_{f_1}^{F_{med}} PS_{CWT}(f) \cdot df = \int_{F_{med}}^{f_2} PS_{CWT}(f) \cdot df \quad (3.1)$$

where $PS_{CWT}(f)$ was the spectral power for the current frequency. $f_1 = 10\text{Hz}$ and $f_2 = 500\text{Hz}$ were the determined bandwidth of the sEMG signals.

- (3) The spectral parameter proposed by Dimitrov et al. (2006) (FI_{nsm5}) was calculated by

$$FI_{nsm5} = \frac{\int_{f_1}^{f_2} f^{-1} PS(f) \cdot df}{\int_{f_1}^{f_2} f^5 PS(f) \cdot df} \quad (3.2)$$

where $PS(f)$ was the spectral power for the current frequency. $f_1 = 10\text{Hz}$ and $f_2 = 500\text{Hz}$ were the determined bandwidth of sEMG signals.

- (4) The wavelet index proposed by Gonzalez-Izal et al. (2010b) (WIRE51) was calculated by

$$WIRE51 = \frac{\sum_{i=1}^N D_5^2[i]}{\sum_{i=1}^N D_1^2[i]} \quad (3.3)$$

where $D_1[i]$ and $D_5[i]$ were the detailed DWT signals at scales five and one, respectively.

- (5) The proposed filter-based fatigue index in this study was obtained by

$$FI_{hrOPT} = \int_{t_{start}}^{t_{end}} \frac{LE_{OPTHzHPF}(t)}{LE_{200HzLPF}(t)} dt / (t_{end} - t_{start}) \quad (3.4)$$

where OPT was the optimized cut-off frequency of the high-pass-filter. t_{start} and t_{end} were the start and end time of each exercise repetition. $LE_{200HzLPF}(t)$ and $LE_{OPTHzHPF}(t)$ were linear-envelopes (the 2nd-order, Butterworth low-pass filter, 3Hz cut-off) of the 200Hz low-pass-filtered sEMG signal, and of the OPT high-pass-filtered sEMG signal, respectively. Cut-off frequency of the high-pass-filter was optimized using the interior-point algorithm (Wright, 2005) to maximize the correlation coefficient between peak power and FI_{hrOPT} . Cut-off frequency of the low-pass filter was determined to retain more than 95% of the power of raw EMG signals.

3.2.5 Statistical analysis

All variables (power output and EMG-based indices) were normalized by the first two contractions of the first set. In cases where the percentage changes did not follow the normal distribution (FI_{nsm5} and WIRE), the

corresponding variables were log-transformed. Pearson's correlation was used to analyze the relationship between the peak power and the different EMG indices. Differences between mean correlation coefficients for FI_{hrIOPT} and mean correlation coefficients for other EMG indices were determined by the unpaired student's t test. Power output and EMG indices were compared in relative terms using the Friedman analysis of variance (ANOVA) by ranks. When a significant difference was shown, Tukey post hoc procedures were performed to determine the pairwise differences between the means.

3.3 Results

3.3.1 Joint power output

Fig. 3.1(a) shows the peak joint power outputs during five sets of ten repetitions. Joint power output during the last five repetitions for every set was significantly smaller than the first five repetitions for every set ($p < 0.05$, Fig. 3.1(a)). The first five repetitions for each set differed significantly in joint power output (the first vs the third, the fourth and the fifth; the second vs the fourth and the fifth; the third vs the fifth; $p < 0.05$). Joint power output in the last repetition for the fifth set was 38% of those in the initial two repetitions of the first set.

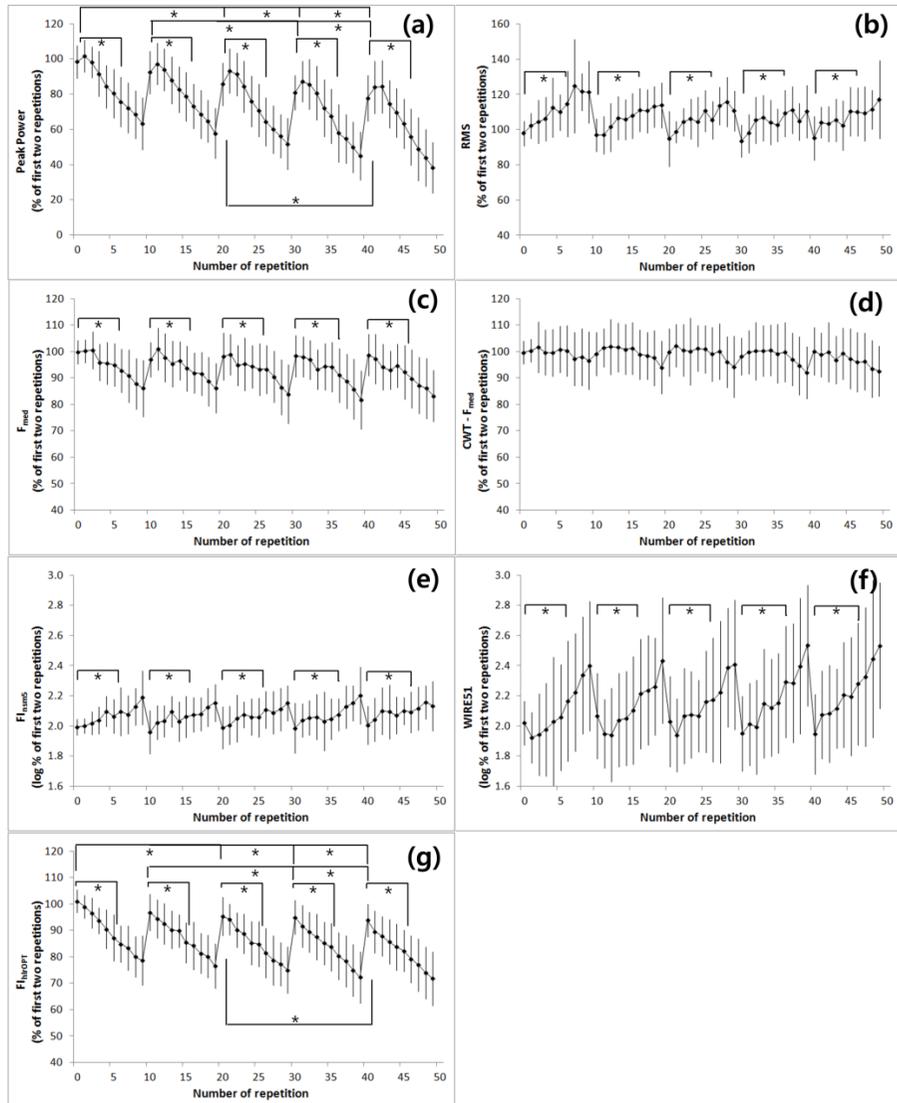


Fig. 3.1. Peak power and EMG indices of rectus femoris muscle (mean \pm standard deviation) during five sets of ten repetitions. (a) Peak power; (b) RMS values of sEMG signal; (c) median frequency (F_{med}); (d) median frequency after continuous wavelet transform (CWT & F_{med}); (e) the logarithm of Dimitrov's index (FI_{nsmS}); (f) the logarithm of Gonzalez-Izal's index (WIRE51); (g) bandwidth-optimized fatigue index (FI_{hlrOPT}) (*significant differences at $p < 0.05$).

3.3.2 EMG indices of rectus femoris (RF) muscle

Table 3.1 shows the optimized cut-off frequencies of RF muscle. The average cut-off frequency was $353.30 \pm 49.45\text{Hz}$ and the average correlation coefficient was 0.81 ± 0.08 .

Fig. 3.1(b) shows RMS values of sEMG signals for RF muscle. The last five repetitions had significantly greater RMS values of EMG signals than the first five repetitions for every set ($p < 0.05$, Fig. 3.1(b)).

Fig 3.1(c) shows F_{med} of sEMG signals for RF muscle. F_{med} of RF muscle significantly decreased during the last five repetitions for each set ($p < 0.05$, Fig. 3.1(c)). The first five repetitions for each set showed no significant differences in F_{med} .

CWT & F_{med} did not show the significant differences between the first five repetitions and the last five repetitions for all sets of data (Fig. 3.1(d)).

The logarithm of $FI_{\text{ns}5}$ and the logarithm of WIRE51 for RF muscle significantly increased during the last five repetitions for each set ($p < 0.05$, Fig. 3.1(e-f)). In logarithms of $FI_{\text{ns}5}$ and WIRE51, there were no statistically significant differences between the first five repetitions of the first set and the first five repetitions of other sets.

FI_{hrOPT} of RF muscle was significantly smaller for the last five repetitions than the first five repetitions for every set ($p < 0.05$, Fig. 3.1(g)). FI_{hrOPT} from the first five repetitions of the first set was significantly larger than those from the first five repetitions of the third, the fourth and the fifth

sets ($p < 0.05$). In the second set, FI_{hrOPT} from the first five repetitions was significantly larger than those from the first five repetitions of the fourth and the fifth sets ($p < 0.05$). In addition, the first five repetitions of the fifth set were significantly smaller ($p < 0.05$) than those of the third set.

3.3.3 EMG indices of vastus muscles (VM, VL)

Fig. 3.2(a) shows RMS values of sEMG signals for VM and VL muscles. The last five repetitions had larger RMS values of sEMG signals than the first five repetitions for the first and the second sets ($p < 0.05$, Fig. 3.2(a)). Table 3.1 shows the optimized cut-off frequencies for VM and VL muscles. The average cut-off frequency was $348.71 \pm 34.28\text{Hz}$ and the average correlation coefficient was 0.58 ± 0.25 . FI_{hrOPT} of VM and VL muscles were significantly smaller for the last five repetitions than for the first five repetitions for all sets ($p < 0.05$, Fig. 3.2(f)). FI_{hrOPT} from the first five repetitions of the first set was significantly different from those from the first five repetitions of the fourth and the fifth sets ($p < 0.05$). There were no statistically significant differences between other indices from the first five repetitions and the last five repetitions for all sets.

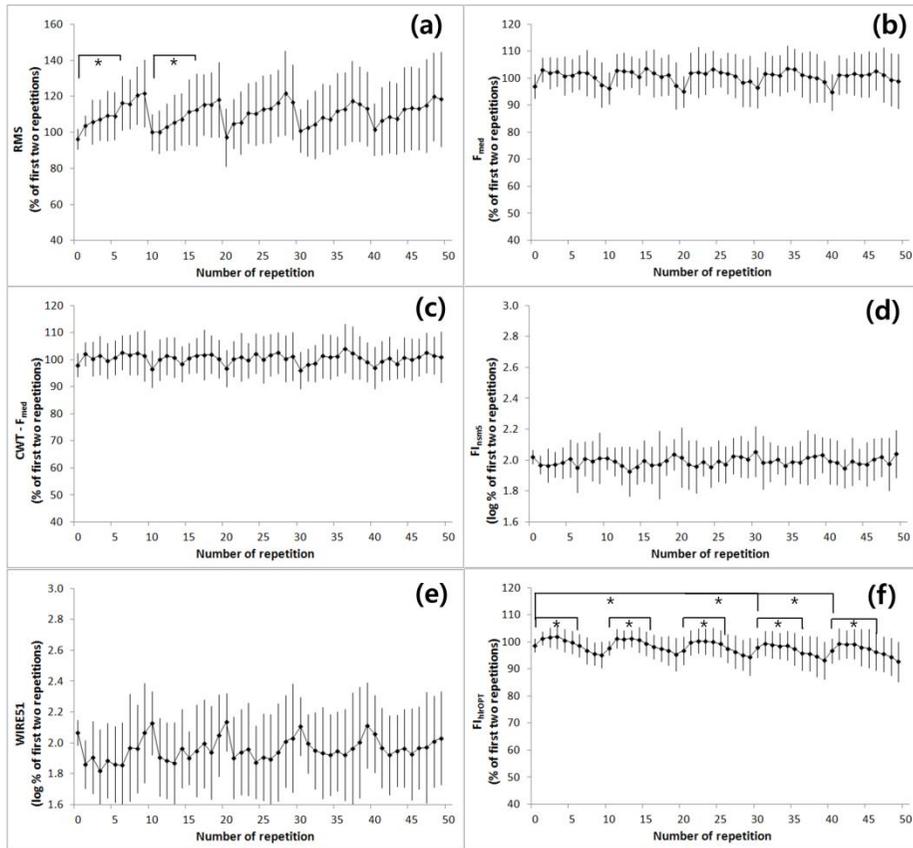


Fig. 3.2. Peak power and EMG indices of average of vastus muscles (mean \pm standard deviation) during five sets of ten repetitions. (a) RMS values of sEMG signal; (b) median frequency (F_{med}); (c) median frequency after continuous wavelet transform (CWT & F_{med}); (d) the logarithm of Dimitrov's index (FI_{nsm5}); (e) the logarithm of Gonzalez-Izal's index (WIRE51); (f) bandwidth-optimized fatigue index (FI_{hlrOPT}) (*significant differences at $p < 0.05$).

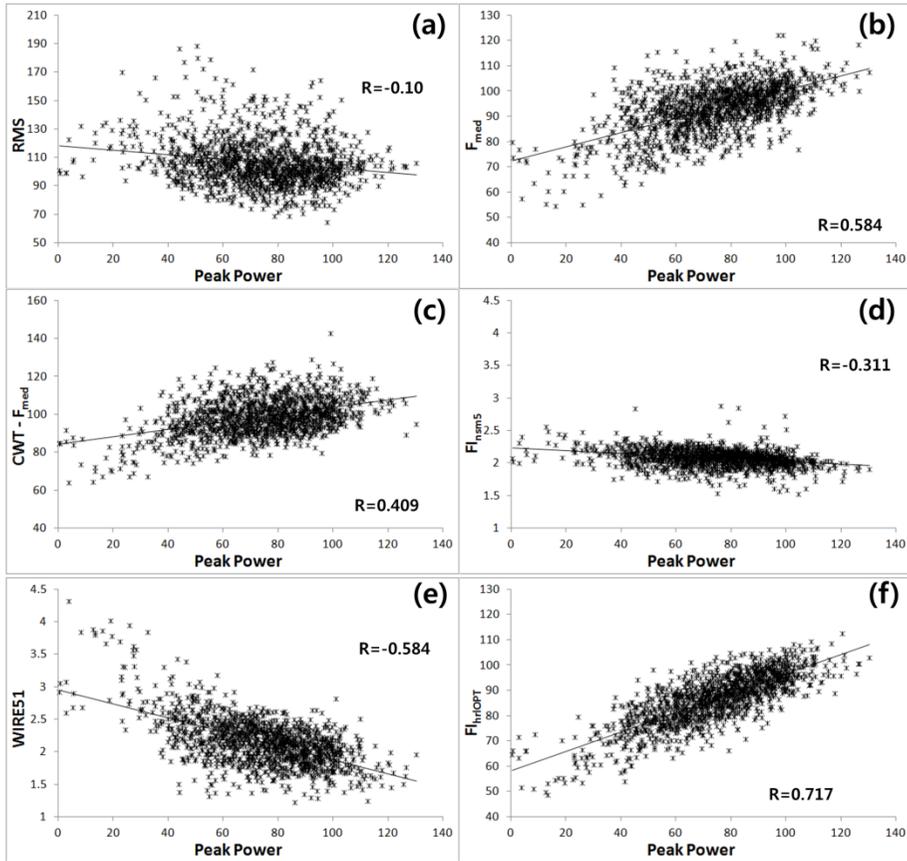


Fig. 3.3. Linear representations of peak power output (normalized by the first two repetitions) versus various EMG indices (normalized by the first two repetitions) for all subjects. (a) Peak power output versus RMS values of sEMG signal; (b) peak power output versus F_{med} ; (c) peak power output versus CWT & F_{med} ; (d) peak power output versus the logarithm of FI_{nsms} ; (e) peak power output versus the logarithm of WIRE51; (f) peak power output versus FI_{hrIOPT} .

3.3.4 Relationships between EMG indices and power output

The EMG index of the strongest correlation with mechanical power was FI_{hrOPT} of RF muscle ($r = 0.81 \pm 0.08$, $p < 0.01$, Fig. 3.3(f)). RMS, F_{med} , CWT & F_{med} , the logarithm of FI_{nsm5} and the logarithm of WIRE51 of RF muscle had correlation coefficients of -0.10 ± 0.34 , 0.60 ± 0.17 , 0.34 ± 0.23 , -0.38 ± 0.21 and -0.50 ± 0.22 , respectively (Table 3.2 and Fig. 3.3(a-e)). RMS, F_{med} , CWT & F_{med} , the logarithm of FI_{nsm5} , the logarithm of WIRE51, FI_{hrOPT} of VM and VL muscles had correlation coefficients of -0.25 ± 0.25 , 0.16 ± 0.26 , -0.03 ± 0.24 , -0.14 ± 0.22 , -0.06 ± 0.24 and 0.58 ± 0.25 , respectively (Table 3.2).

Table 3.1. Optimized cut-off frequencies and correlation coefficients

	Cut-off frequencies		Correlation coefficient	
	RF	VM & VL	RF	VM & VL
Subject 1	415.07	308.28	0.95**	0.93**
Subject 2	373.54	335.11	0.74**	0.17
Subject 3	354.20	339.90	0.77**	0.46**
Subject 4	289.66	312.54	0.83**	0.52**
Subject 5	363.25	342.39	0.88**	0.35*
Subject 6	389.57	345.80	0.81**	0.38**
Subject 7	377.49	387.46	0.82**	0.77**
Subject 8	258.98	357.32	0.89**	0.15
Subject 9	268.47	355.20	0.77**	0.65**
Subject 10	306.44	320.41	0.80**	0.52**
Subject 11	347.39	344.03	0.81**	0.79**
Subject 12	417.11	356.35	0.91**	0.88**
Subject 13	355.62	442.60	0.89**	0.85**
Subject 14	348.14	321.45	0.68**	0.33*
Subject 15	310.06	344.43	0.84**	0.51**
Subject 16	291.68	322.39	0.79**	0.43**
Subject 17	396.52	345.39	0.76**	0.85**
Subject 18	421.83	405.84	0.90**	0.66**
Subject 19	322.66	341.04	0.81**	0.76**
Subject 20	281.72	364.99	0.71**	0.09
Subject 21	364.30	272.95	0.67**	0.28*
Subject 22	392.78	362.07	0.89**	0.74**
Subject 23	426.61	364.80	0.83**	0.78**
Subject 24	357.72	309.13	0.63**	0.58**
Subject 25	333.32	368.34	0.92**	0.85**
Subject 26	350.28	343.53	0.76**	0.45**
Subject 27	424.73	401.42	0.85**	0.86**
Average	353.30	348.71	0.81	0.58
SD	49.45	34.28	0.08	0.25

** p<0.01; *p<0.05.

Table 3.2. Correlation coefficients between EMG indices and peak power output

	RMS	F _{med}	CWT & F _{med}	FI _{nsm5}	WIRE	FI _{hloPT}
RF	-0.10± 0.34**	0.60± 0.17**	0.34± 0.23**	-0.38± 0.21**	-0.50± 0.22**	0.81± 0.08
VM & VL	-0.25± 0.25**	0.16± 0.26**	-0.03± 0.24**	-0.14± 0.22**	-0.06± 0.24**	0.58± 0.25

** p<0.01; *p<0.05. Differences from FI_{hloPT}

3.4 Discussion

According to the previous study, CWT & F_{med} showed the best fatigue estimation accuracy (Karlsson et al., 2000). In addition, FI_{nsm5} (Dimitrov et al., 2006) showed the high-sensitivity and WIRE51 (Gonzalez-Izal et al., 2010b) showed the good correlation with power output and the robustness against noise. However, CWT & F_{med}, FI_{nsm5} and WIRE51 had the lower correlation with joint power than F_{med} in this study. Some subjects showed the positive correlation with joint power but others the negative correlation. FI_{hloPT} showed the best correlation (0.81 ±0.08 in RF, 0.58 ±0.25 in VM & VL) with joint power, as shown in the previous study (Kim et al., 2012). This suggested that FI_{hloPT} had the best accuracy to estimate muscle fatigue during high-intensity exercises. Similar to the previous studies, F_{med} did not show the significant differences between the first five repetitions for all sets. In this study, there were no significant differences in FI_{nsm5} and WIRE51 for the first

five repetitions of every set. However, the first five repetitions of each set in FI_{hrOPT} differed significantly, which reflects the significant reduction of joint power. These findings also indicated that FI_{hrOPT} could assess muscle fatigue even in the less-fatigued state during multiple sets of dynamic fatigue exercises.

None of the EMG indices for VM and VL muscles, except for RMS and FI_{hrOPT} , were significantly different for every set. This indicates a relatively low-fatigue state. According to the previous study (Jacobs and van Ingen Schenau, 1992), mono-articular muscles rapidly lose their tension and decrease their activation in the quick movement of the limbs. In this study, subjects performed knee extensions as quickly as possible. Thus, VM and VL muscles, mono-articular muscles, would have been under submaximal exercises in this study.

The primary aim of this study was to determine the frequency bandwidth of new fatigue index (FI_{hrOPT}). The uniqueness of the present study was the optimization of cut-off frequency to overcome bandwidth selection problem in the previous studies. The shift in the power spectrum of EMG to lower frequencies was caused by the lengthening effect of MUAP (Bigland-Ritchie et al., 1981), decreased MFCV and the reduced activation of fast twitch muscle fibers (Komi and Tesch, 1979). The lengthening effect of MUAP was caused primarily by decreased MFCV (Dimitrova and Dimitrov, 2002). Masuda et al. (1999) reported that MFCV did not decrease during dynamic contractions. In addition, Mills (1982) showed that the compound muscle

action potential spectrum did not change beyond 200Hz during fatigue. Thus, the reduced activation of fast twitch muscle fibers was the main reason of decreased high-frequency components in fatigue state. The 200Hz low-pass filter cut-off frequency was selected to eliminate the effect of decreased high-frequency component by change of MUAPs shape. Low-pass-filtered EMG signals retained more than 95% of the power of raw EMG signals. High-pass-filtered EMG signals reflected the reduced activation of fast twitch muscle fibers because determined high-pass filter cut-off frequency was over 200Hz. In addition, 350 Hz high-pass filtered EMG signals retained less than 5% of the power of raw EMG signals.

The sEMG decomposition analysis can estimate the reduced activation of fast twitch muscle fiber in fatigue state. Some sEMG decomposition researches reported that the average motor unit firing rate was about 20 to 40Hz (26.4 ± 7.6 Hz (Roos et al., 1999), 32.1 ± 10.7 Hz (Bigland-Ritchie et al., 1992) and 41.9 ± 8.2 Hz (Connelly et al., 1999)). DeLuca and Hostage (2010) showed that the first dorsal interosseous, tibialis anterior and VL had about a 30Hz average firing rate and about 20 motor units in 100% MVC contraction. Stock et al. (2012) showed that VM and VL had about a 30Hz average firing rate and about 25 motor units. These results showed the similar firing rate and motor unit numbers in sEMG of different muscles. In a fatigued state, the average firing rate decreased from 30Hz to 25Hz (Stock et al., 2012), 30Hz to 22Hz (De Luca and Hostage, 2010) and 30Hz to 17Hz (DeLuca and Forrest, 1973). Based on these results, in the perfectly independent motor units, the

maximum feasible frequency of the MUAP train in a non-fatigued state is about 600Hz ($30\text{Hz} \times 20$ motor units) and the maximum feasible frequency in a fatigued state is about 400Hz ($20\text{Hz} \times 20$ motor units). Since there would be synchronized MUAP trains for each motor unit in the sEMG signal, the MUAP train has a lower frequency than the calculated. The maximum feasible frequency in a fatigued state was close to the optimized cut-off frequency calculated in this study (353.30Hz and 348.71Hz). The optimized cut-off frequencies of RF and vastus muscles were also similar to those from the previous study (353.30Hz, 348.71Hz, and 360Hz (Kim et al., 2012)). These results support the hypothesis that the optimized cut-off frequency appropriate for fatigue assessment.

A limitation of this study is that the proposed filter-based fatigue index, FI_{hrOPT} , was applied only to knee extensor muscles and did not take inter-electrode distance into account. The proposed filter-based fatigue index will be extended to various joint muscles and inter-electrode distances to determine general high-pass cut-off frequencies.

Chapter 4

4. Validation of the new filter-based fatigue index

4.1 Purpose

A filter-based fatigue index was proposed as ratios between high-frequency components and low-frequency components of EMG power (Kim et al., 2012). This technique has the same order of time resolution and required less computational load. However, there have been difficulties in selecting a bandwidth of the high- and low-frequency components of EMG power (Dimitrov and Dimitrova, 1998). To select a bandwidth, the cut-off frequencies of the high-pass-filter was optimized to maximize the correlation with peak power and proposed filter-based fatigue index (FI_{hrOPT}) during isotonic knee extension exercise. However, various types of muscles, different exercises, and different inter-electrode distances did not tested. It should be examined to determine the general bandwidth.

In this study, the filter-based fatigue index was validated through a comparison of the optimized cut-off frequencies in various muscles, using different inter-electrode distances and a variety of exercises. Hypothesis was that the optimized cut-off frequencies in various muscles and various inter-

electrode distances during different exercises are consistent. The filter-based fatigue index was compared with the conventional median frequency, the spectral parameter proposed by Dimitrov (2006), and the wavelet index proposed by Gonzalez-Izal (2010b).

4.2 Experimental methods

4.2.1 Subjects

Sixty one healthy males with no history of upper or lower limb disorders participated in this study. They were categorized into three groups performing: (1) isotonic knee extension exercises (27 subjects, age: 26.4 ± 4.4 years, height: 171.4 ± 6.3 cm, weight: 76.5 ± 12.4 kg); (2) isokinetic knee extension and isotonic elbow flexion (14 subjects, age: 25.7 ± 3.2 years, height: 173.3 ± 5.3 cm, weight: 73.1 ± 10.5 kg); and (3) isometric knee extension, isotonic elbow extension and isotonic ankle dorsiflexion (20 subjects, age: 22.3 ± 3.1 years, height: 171.9 ± 9.3 cm, weight: 84.3 ± 9.3 kg). Subjects were informed about the purpose of this study and the experimental procedure. Before the actual experiments, the subjects were required to practice three to four voluntary submaximal isometric and isotonic contractions to familiarize with the experiments. They provided written informed consent before participating in the study. The Yonsei University research ethics committee reviewed and approved the proposed experimental design (1041849-201405-BM-022-02).

4.2.2 Experimental Procedures

Subjects performed joint flexion/extension exercises within pre-determined range of motion (ROM) on an isokinetic dynamometer (Biodex System 3, Biodex Medical Systems, NY, USA). Six kinds of exercises were chosen to evaluate the filter-based fatigue index. In the isotonic exercises, each subject performed several maximal contractions to determine the 10 repetition maximum (10RM) load. The determined 10RM was used for the fatigue exercises. The experimental protocol consisted of five sets of 10 joint movements with 2-minute break. During fatigue exercises, the subjects were asked to perform joint movements with their own maximum efforts. Characteristic of the exercises were explained in Table 4.1.

Table 4.1. Characteristic of six kinds of exercises

Isotonic	Knee extension	ROM: from 115° to 0° (full extension) MVC measurement at 90° 10RM: 62.3 ±4.4% of the MVC torque
	Elbow flexion	ROM: from 40° to 100° MVC measurement at 90° 10RM: 64.7 ±3.8% of the MVC torque
	Elbow extension	ROM: from 100° to 40° MVC measurement at 80° 10RM: 60.1 ±5.6% of the MVC torque
	Ankle dorsiflexion	ROM: from -25° (plantarflexion) to 5° MVC measurement at -10° 10RM: 60.3 ±5.4% of the MVC torque
Isokinetic	Knee extension	ROM: from 115° to 0° (full extension) Angular velocity: 120°/s Fifty knee extensions
Isometric	Knee extension	MVC measurement at 90° Maximal isometric knee extension until exhaustion.

4.2.3 Data Acquisition

A EMG measurement system (MyoSystem 1200, Noraxon Inc., AZ, USA) was used to measure EMG signals from the right rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), biceps brachii (BB), triceps brachii (TC) and tibialis anterior (TA) muscles during exercise. Joint angle, angular velocity and torque were acquired from the isokinetic dynamometer. These signals and EMG signals were simultaneously recorded at a sampling rate of 1 kHz.

EMG signals were recorded by a standard bipolar surface electrode (14 mm in diameter, 18 mm inter-electrode distance; Noraxon, AZ, USA). The inter-electrode distance was set to 2cm, 6cm and 10 cm during isotonic elbow extension, isotonic ankle dorsiflexion and isometric knee extension (Fig. 4.1). After the careful skin preparation (shaving, abrasion and cleaning with alcohol), electrode pairs were placed longitudinally on the middle portion of the muscle, based on the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) recommendations.

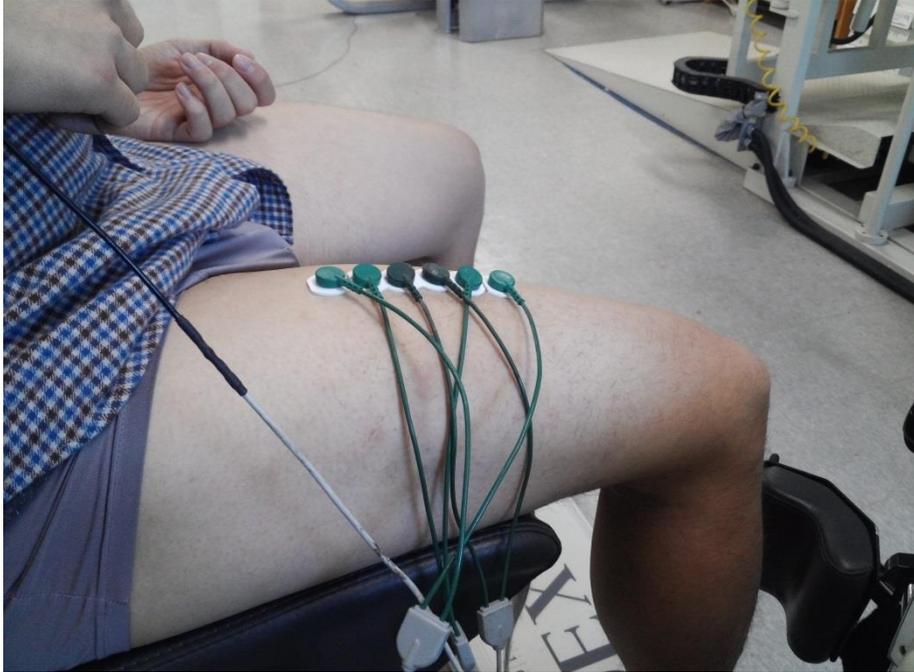


Fig. 4.1. Different inter-electrode distances (2cm, 6cm and 10cm)

2.4 Data analysis

Data analysis was performed using MATLAB R2010a (MathWorks Inc., Natick, MA, USA). In case of isotonic or isokinetic exercise, a segment was defined as one exercise repetition (i.e., one flexion and extension), and then all data were divided into 50 segments. The time-series data obtained during isometric exercise were equally divided into 50 segments. MVC torque was calculated as the mean value over a 1-s period with relatively torque-plateau levels of each MVC attempt. Peak power outputs of each segment were

calculated as the product of joint torque and angular velocity (isotonic and isokinetic exercises only).

Peak joint power output, mean root-mean-square (RMS), F_{med} , FI_{nsm5} and FI_{hrOPT} were calculated based on the chapter 3.

2.5 Statistical analysis

All variables (power output and EMG-based parameters) were normalized by the mean value of the first two segments. In cases where the percentage changes did not follow a normal distribution (FI_{nsm5} and WIRE51), the corresponding variables were log-transformed. Pearson's correlation was used to analyze the relationship between peak power (for isotonic and isokinetic exercises) or joint torque (for isometric exercises) and different EMG parameters. Differences between mean correlation coefficients for FI_{hrOPT} and other EMG indices were statistically determined by the unpaired t -test. One-way analysis of variance (ANOVA) was used to calculate any significant differences in the optimized cut-off frequencies in the optimized cut-off frequencies for the six different muscles, three different exercise methods and three different inter-electrode distances. Significant differences in the correlation coefficients for different inter-electrode distances were also calculated by ANOVA analysis. Tukey's studentized range test was performed to identify homogeneous subsets of means that are not different from each other.

4.3 Results

4.3.1 Bandwidth optimization in various muscles

Figure 4.2 shows the optimized cut-off frequencies for various muscles during isotonic exercises. The optimized cut-off frequencies for all muscles were identified as the same subset ($p = .811$). FI_{hrOPT} showed the significantly higher correlation with the mechanical power than the other fatigue indices in all muscles except FI_{nsm5} for the TA (Table 4.2).

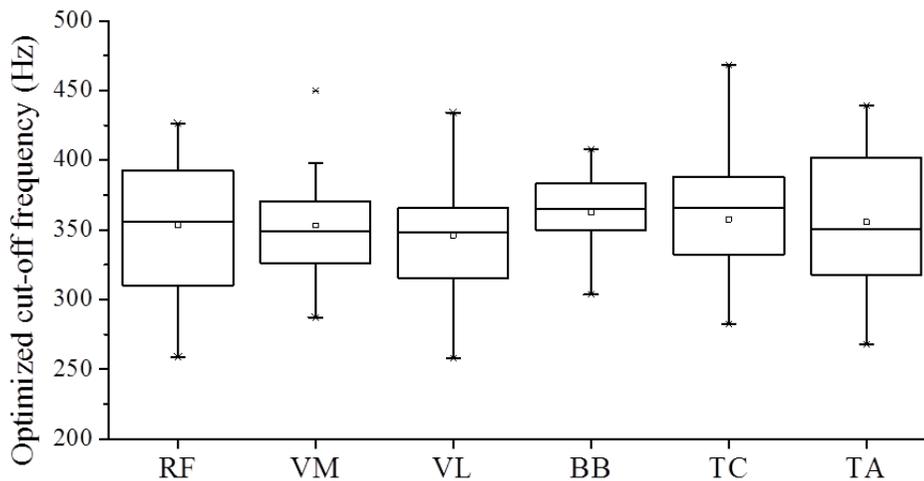


Fig. 4.2. Optimized cut-off frequencies for various muscles during isotonic exercises. Optimized cut-off frequencies were 353.3 ± 49.5 Hz in RF muscle, 343.9 ± 34.2 Hz in VM muscle, 353.7 ± 36.1 Hz in VL muscle, 362.3 ± 28.2 Hz in BB muscle, 357.5 ± 46.0 Hz in TC muscle, and 348.1 ± 38.2 Hz in TA muscle.

Table 4.2. Correlation coefficients between EMG parameters and peak power output in various muscles

	RMS	F _{med}	FI _{nsm5}	WIRE51	FI _{hlrOPT}
RF	-0.10 ±0.34*	0.60 ±0.17*	-0.38 ±0.21*	-0.50 ±0.22*	0.81 ±0.08
VM	-0.23 ±0.25*	0.08 ±0.23*	-0.07 ±0.22*	-0.02 ±0.25*	0.48 ±0.25
VL	-0.22 ±0.22*	0.17 ±0.28*	-0.20 ±0.25*	-0.07 ±0.23*	0.52 ±0.24
BB	-0.14 ±0.37*	0.48 ±0.12*	-0.56 ±0.13*	-0.36 ±0.19*	0.72 ±0.08
TC	-0.06 ±0.54*	0.58 ±0.13*	-0.52 ±0.25*	-0.55 ±0.07*	0.71 ±0.12
TA	-0.48 ±0.19*	0.43 ±0.21*	-0.54 ±0.18	-0.29 ±0.24*	0.58 ±0.13

*p<0.05. Differences from FI_{hlrOPT}

4.3.2 Bandwidth optimization during different exercises

Figure 4.3 shows the optimized cut-off frequencies for different exercises (isotonic, isokinetic and isometric) in the RF muscle. The optimized cut-off frequencies for all exercises were classified into the same group (p = .962). In all exercises, FI_{hlrOPT} showed significantly higher correlation with biomechanical fatigue (peak power output or joint torque) than other fatigue indices (Table 4.3).

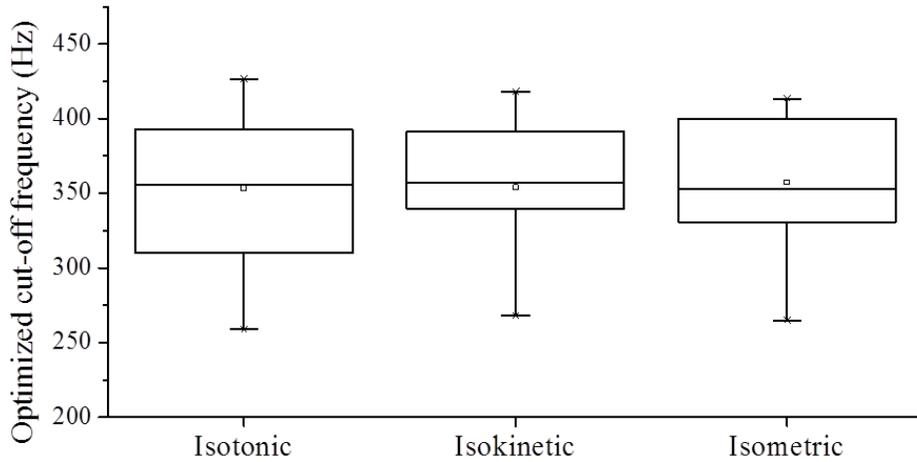


Fig. 4.3. Optimized cut-off frequencies for different exercises in the rectus femoris (RF) muscle. Optimized cut-off frequencies were 353.3 ± 49.5 Hz in isotonic exercise, 354.0 ± 46.3 Hz in isokinetic exercise, and 357.4 ± 45.4 Hz in isometric exercise.

Table 4.3. Correlation coefficients between EMG parameters and peak power output or joint torque for isotonic, isokinetic and isometric exercises

	RMS	F_{med}	FI_{nsm5}	WIRE51	FI_{hlrOPT}
Isotonic	-0.10 $\pm 0.34^*$	0.60 $\pm 0.17^*$	-0.38 $\pm 0.21^*$	-0.50 $\pm 0.22^*$	0.81 ± 0.08
Isokinetic	0.08 $\pm 0.51^*$	0.77 $\pm 0.15^*$	-0.64 $\pm 0.26^*$	-0.67 $\pm 0.12^*$	0.88 ± 0.08
Isometric	0.70 $\pm 0.34^*$	0.79 $\pm 0.10^*$	-0.68 $\pm 0.09^*$	-0.75 $\pm 0.09^*$	0.92 ± 0.05

* $p < 0.05$. Differences from FI_{hlrOPT}

4.3.3 Bandwidth optimization at various inter-electrode distances

Table 4.4 shows the optimized cut-off frequencies of various inter-electrode distances during isotonic ankle dorsiflexion, isotonic elbow extension, and isometric knee extension. The optimized cut-off frequencies for all muscles and all inter-electrodes were identified as the same group ($p = .969$). Table 4.5 shows the correlation coefficients between FI_{hrOPT} and biomechanical fatigue at different inter-electrode distances. Correlation coefficients for the same muscles were classified as the same subset ($p = .823$ for TC, $p = .998$ for TA and $p = 1.00$ for RF).

Table 4.4. Optimized cut-off frequencies and correlation coefficients between FI_{hrOPT} and the peak power output or joint torque in various inter-electrode distances

	Inter-electrode distance (cm)	Optimized cut-off frequency (Hz)	Correlation coefficient
TC	2	357.5 ± 46.0	0.71 ± 0.12
	6	351.4 ± 65.4	0.69 ± 0.14
	10	357.6 ± 54.6	0.75 ± 0.08
TA	2	355.3 ± 51.3	0.58 ± 0.13
	6	348.1 ± 38.2	0.58 ± 0.16
	10	345.1 ± 44.7	0.60 ± 0.12
RF (isometric)	2	357.4 ± 45.4	0.92 ± 0.05
	6	361.4 ± 61.2	0.92 ± 0.04
	10	363.9 ± 61.7	0.92 ± 0.05

Boxes indicate the same subset.

Table 4.5. Correlation coefficients between FI_{hrOPT} and peak power output or joint torque at various inter-electrode distances

	Inter-electrode distance	Correlation coefficient
TC	2cm	0.71 ± 0.12
	6cm	0.69 ± 0.14
	10cm	0.75 ± 0.08
TA	2cm	0.58 ± 0.13
	6cm	0.58 ± 0.16
	10cm	0.60 ± 0.12
RF (isometric)	2cm	0.92 ± 0.05
	6cm	0.92 ± 0.04
	10cm	0.92 ± 0.05

4.4 Discussion

The hypothesis of this study was that the optimized cut-off frequencies would be the same for different muscles, different exercises or different inter-electrode distances. RF and BB are multi-articular muscles, while VM, VL, TC and TA are mono-articular muscles. In addition, the composition ratios of slow- to fast-twitch fibers were different. The average fiber type distribution is 52% type I and 48% type II in RF muscle (Scelsi, 2001), 59% type I and 41% type II in VM muscle (Travnik et al., 1995), 52% type I and 48% type II in VL muscle (Schantz et al., 1983), 42% type I and 58% type II in BB muscle (Kukulka and Clamann, 1981), 43% type I and 57% type II in TC muscle

(Schantz et al., 1982), and 77% type I and 23% type II in TA muscle (Jakobsson et al., 1990). In spite of this multiplicity, optimized cut-off frequencies for all muscles were identified in the same subset. Similar to previous studies (Kim et al., 2012; Kim et al., 2013), the optimized cut-off frequency was about 350Hz. This result supports our hypothesis that the optimized cut-off frequency does not differ in muscle type, which implies that a 350 Hz high-pass filter (HPF) could be a solution for skeletal muscle fatigue assessment with FI_{hrOPT} during isotonic exercises.

Different knee extension exercise protocols were included in this study. Most previous fatigue index design studies selected only one exercise protocol (Karlsson et al., 2000; Dimitrov et al., 2006; Gonzalez-Izal et al., 2010b; Kim et al., 2012; van Dieen et al., 1996). However, different fatigue patterns appear with different types of exercise. Masuda et al. (1999) showed that MFCV did not decrease during isotonic exercise, although it was a major factor in EMG spectrum shift during fatigue (Cifrek et al., 2009; Dimitrova and Dimitrov, 2002). During isometric and isokinetic exercises, joint torque naturally decreases. Joint angular velocity naturally decreases during isotonic exercises. Similar to previous studies (Gonzalez-Izal et al., 2010a; Gonzalez-Izal et al., 2010b; Kim et al., 2012; Kim et al., 2013), RMS values of the EMG signals had a negative correlation with the peak power output during isotonic exercises. However, RMS values of the EMG signals showed a positive correlation with joint torque during isometric exercises. This result supports the previous findings that motor output from the central nervous system is

reduced during maximal isometric contractions (Bigland-Ritchie et al., 1983a). However, optimized cut-off frequencies were identified as the same group, and FI_{hrOPT} showed the best correlation with peak power or joint torque. Similar to previous studies (Kim et al., 2012; Kim et al., 2013) and various muscle tests, optimized cut-off value was about 350Hz. This result supports the hypothesis that the optimized cut-off frequencies for various exercises are consistent.

According to the previous study, EMG amplitude increased with increasing inter-electrode distance due to the larger detection volume (Castroflorio et al., 2005). EMG spectrum was also affected by inter-electrode distance. EMG power spectrum shifts to the lower frequency due to the lengthening effect of MUAP with increased inter-electrode distance (Castroflorio et al., 2005). Three different inter-electrode distances were used to verify these effects. As the inter-electrode distance increased, EMG amplitude was increased but F_{med} was decreased. However, optimized cut-off frequencies in three different exercises and in three different inter-electrode distances were classified into the same group. This supports our hypothesis that the optimized cut-off frequencies in various inter-electrode distances are not different. This also supports the previous result that the FI_{hrOPT} reflects firing rate of fast twitch muscle fibers (Kim et al., 2013). Correlations with biomechanical fatigue from the same muscle were also classified into the same subset. This indicates that the normalized FI_{hrOPT} was not affected by the inter-electrode distance.

Similar to a previous study (Kim et al., 2013), FI_{hrOPT} showed a significantly higher correlation with biomechanical fatigue than the other fatigue indices in all conditions except for FI_{nsm5} in the TA. This suggests that FI_{hrOPT} provides the most accurate estimate of muscle fatigue in different muscles during different exercises.

Chapter 5

5. Discussion and conclusion

5.1 Discussion

In this study, a filter-based fatigue index was designed as the ratio of EMG power components in the high and low frequency bands. It was calculated using a FIR filter, not by a frequency analysis method. This technique has the advantage of the less computational load and the same order of time resolution with EMG signal (Kim et al., 2012).

Similar to previous studies, median frequency and the designed filter-based fatigue index decreased with fatigue because of the EMG power spectrum shifted to the lower frequencies (Cifrek et al., 2009; Bigland-Ritchie et al., 1983; Jang and Kim, 2004; Masuda et al., 1999; Gonzalez-Izal et al., 2010a; Dimitrov et al., 2006; Kim et al., 2012). These changes could be caused by the lengthening effect of motor unit action potential (Bigland-Ritchie et al., 1981) and reduced activation of fast twitch muscle fibers (Komi and Tesch, 1979).

Cut-off frequency of low-pass-filter was determined. In Chapter 2, 120Hz low-pass-filter was used to eliminate the effect of decreased high-

frequency component by change of MUAPs shape. However, Mills (1982) studied changes in shape of muscle action potentials and found that the compound muscle action potential spectrum did not change beyond 200Hz during fatigue. In addition, 200Hz low-pass-filtered EMG signals retained more than 95% of the power of raw EMG signals. Thus, the 200Hz cut-off frequency was selected to reflect the full muscle activation in low-frequency components terms.

The optimization method was applied to determine the frequency band of the filter-based fatigue index. The optimized cut-off frequencies were about 350Hz. The filter-based fatigue index showed the significantly higher correlation with the biomechanical fatigue than the other fatigue index in all conditions except FI_{nsm5} of TA during isotonic exercise. This suggested that the filter-based fatigue index had the best accuracy to estimate muscle fatigue during high-intensity exercises. During isotonic exercises, the first five repetitions of each set in the filter-based fatigue index differed significantly, which reflects the significant reduction of joint power. This finding indicated that the filter-based fatigue index could assess muscle fatigue even in the less-fatigued state during multiple sets of dynamic fatigue exercises. However, the optimized cut-off frequency, about 350Hz, has no physiological basis itself, just simply a curve-fitting.

Several previous studies were referred to assign the physiological characteristic of muscle to the filter-based fatigue index. According to previous studies (Roos et al., 1999; Bigland-Ritchie et al., 1992; Connelly et

al., 1999; DeLuca and Hostage, 2010; Stock et al., 2012; DeLuca and Forrest, 1973), the maximum feasible frequencies of motor unit firing rate in non-fatigued and fatigued states are 600 Hz and 400 Hz respectively. Since there would be synchronized MUAPs for each motor unit in the sEMG signal, the motor unit firing rate has a lower frequency than these frequencies. The maximum feasible frequency in a fatigued state was close to the 350Hz. Thus, optimized cut-off frequency could reflect the physiological characteristic of muscle such as these changes of motor unit firing rates. The filter-based fatigue index could be used for all skeletal muscles and all exercises, because of the optimized cut-off frequencies in various muscles and different exercises were consistent. In addition, the optimized cut-off frequencies in various inter-electrode distances were similar despite the length of M-wave changed by inter-electrode distance (Castroflorio et al., 2005). These results supported the filter-based fatigue index reflects the changes of motor unit firing rate.

Although ratios of the EMG power content in high- and low-frequency ranges showed good accuracy of fatigue estimation, there have been difficulties in selection of high- and low-frequency bandwidth (Bellemare and Grassino, 1982; Chaffin, 1973; Kim et al., 2012; Kogi and Hakamada, 1962; Moxham et al., 1982; Sato, 1982). To overcome this limitation, cut-off frequencies of the high-pass filter were optimized by maximizing the correlation coefficient between peak power and their filter-based fatigue index in Chapter 3. The determined frequencies were consistent (i.e., approximately 350 Hz), showing 353 and 349 Hz in RF and vastii muscles, respectively. In

Chapter 4, FI_{hrOPT} showed the significantly higher correlation with peak power or joint torque than other fatigue indices in all conditions with consistent optimized cut-off frequency (about 350 Hz). This suggested that FI_{hrOPT} had the best accuracy to estimate muscle fatigue. The results of this study support that around 350 Hz high-pass filter could be useful for general purposes.

One of the main finding of this study was that filter-based fatigue index reflects the firing rate of motor units. However, firing rates of motor units were not measured during fatigue. In further studies, MUAP decomposition methods or intramuscular EMG methods would be applied to measure the firing rate of motor units.

5.2 Conclusion

A filter-based fatigue index was designed as the ratio of high-frequency to low-frequency components of EMG power to assess the peripheral muscle fatigue. It has the same order of time resolution with EMG signal and the less computational load. It showed the greater accuracy than traditional fatigue indices (-0.135 ± 0.358 for RMS, 0.632 ± 0.164 for median frequency, 0.749 ± 0.081 for filter-based fatigue index with 360Hz of HPF) and 360Hz of HPF showed the best accuracy during isotonic knee extension exercise. Cut-off frequencies of HPF were optimized to maximize the correlation coefficient between the peak power and the filter-based fatigue index to determine the bandwidth of filter-based fatigue index. The optimized cut-off frequencies were about 350Hz and the filter-based fatigue index with this cut-off frequency showed the greater correlation with the peak power than the other fatigue indices. The filter-based fatigue index was validated by the comparison of the optimized cut-off frequencies in various muscles and various inter-electrode distances during different exercises. Optimized cut-off frequencies in different muscles, under different exercises and various inter-electrode distances were consistent. Thus around 350 Hz high-pass filter could be useful for general purposes. From this study, the following conclusions can be made:

- [1] The filter-based fatigue index showed the best correlation with joint power during isotonic knee extension exercise. Thus, the filter-based

fatigue index could evaluate the muscle fatigue at high-intensity exercises well.

- [2] The filter-based fatigue index differed significantly in the first five repetitions of each set, reflecting the reduction in joint power from the first five repetitions of every set. These findings indicate that the filter-based fatigue index could assess the muscle fatigue even under a low-fatigue state during multiple sets of dynamic fatigue exercises.
- [3] The EMG-based fatigue indices showed relatively lower correlation with joint power in vasti muscles during isotonic knee extension exercise. These implied that vasti muscles would have been under submaximal exercises in this study.
- [4] The optimized cut-off frequencies of HPF were about 350Hz during isotonic knee extension exercise. The maximum feasible frequencies of motor unit firing rate in non-fatigued and fatigued states were calculated as 600 Hz and 400 Hz respectively. These results indicated that the optimized cut-off frequency could reflect the physiological characteristic of muscle such as these changes of motor unit firing rates.
- [5] Different types of muscles were tested in this study. Despite the multiplicity of muscles, optimized cut-off frequencies of all muscles were consistent at about 350Hz. This result indicated that filter-based

fatigue index could assess the muscle fatigue of various skeletal muscles during isotonic exercises with 350Hz HPF.

[6] Although the various type of exercises were included in this study, optimized cut-off frequencies were classified into the same group. This implied that filter-based fatigue index could assess the muscle fatigue during different exercises with 350Hz HPF.

[7] Three different inter-electrode distances were used to verify the influence of inter-electrode distance. Although the EMG spectrum was affected by inter-electrode distance, optimized cut-off frequencies were identified as the same subset. This result indicated that 350Hz HPF could be a general solution for the filter-based fatigue index and the filter-based fatigue index reflects firing rate of motor units.

References

- [1] Amann, M., Blain, G. M., Proctor, L. T., Sebranek, J.J., Pegelow, D. F. and Dempsey, J. A., "Group III and IV muscle afferents contribute to ventilatory and cardiovascular response to rhythmic exercise in humans," *J Appl Physiol*, Vol. 109, No. 4, pp. 966–976, 2010.
- [2] Ament, W., Bonga, G. J., Hof, A. L. and Verkerke, G. J., "Electromyogram median power frequency in dynamic exercise at medium exercise intensities," *Eur J Appl Physiol Occup Physiol*, Vol. 74, No. 1-2, pp. 180-186, 1996.
- [3] Arabadzhiev, T. I., Dimitrov, V. G., Dimitrova, N. A. and Dimitrov, G. V., "Interpretation of EMG integral or RMS and estimates of "neuromuscular efficiency" can be misleading in fatiguing contraction," *J Electromyogr Kinesiol.*, Vol. 20, No. 2, pp. 223-232, 2010.
- [4] Astrand, P. and Rodhal, K. "Textbook of Work Physiology. Physiological Bases of Exercise," McGraw-Hill Book Company, 1986.
- [5] Bellemare, F. and Grassino. A., "Evaluation of human diaphragm fatigue," *J. Appl. Physiol*, Vol. 53, No. 5, pp. 1196–1206, 1982.

- [6] Bigland-Ritchie, B., Donovan, E. F. and Roussos, C. S., "Conduction velocity and EMG power spectrum changes in fatigue of sustained maximal efforts," *J Appl Physiol*, Vol. 51, No. 5, pp. 1300-1305, 1981.
- [7] Bigland-Ritchie, B., Johansson, R., Lippold, O. and Woods, J., "Contractile speed and EMG changes during fatigue of sustained maximal voluntary contractions," *J Neurophysiol.*, Vol. 50, pp. 313-324, 1983a.
- [8] Bigland-Ritchie, B., Johansson, R., Lippold, O. C., Smith, S. and Woods, J. J., "Changes in motoneurone firing rates during sustained maximal voluntary contractions," *J Physiol*, Vol. 340, pp. 335-346, 1983b.
- [9] Bigland-Ritchie, B. R., Furbush, F. H., Gandevia, S. C. and Thomas, C. K., "Voluntary discharge frequencies of human motoneurons at different muscle lengths," *Muscle Nerve*, Vol. 15, pp. 130-137, 1992.
- [10] Castroflorio, T., Farina, D., Botin, A., Piacino, M. G., Bracco, P. and Merletti, R., "Surface EMG of the jaw elevator muscles: effect of electrode locations and inter-electrode distance," *J Oral Rehabil.*, Vol. 32, pp. 411-417, 2005.
- [11] Chaffin, D. B., "Localized muscle fatigue, definition and measurements," *J. Occup. Med.*, Vol. 15, No. 4, pp. 346-354, 1973.

- [12] Cifrek, M., Medved, V., Tonkovic, S. and Ostojic, S., "Surface EMG based Muscle fatigue evaluation in biomechanics," *Clin. Biomech.*, Vol. 24, No 4., pp. 327-340, 2009.
- [13] Coco, M., Alagona, G., Rapisarda, G., Costanzo, E., Calogero, R. A., Perciavalle, V. and Perciavalle, V., "Elevated blood lactate is associated with increased motor cortex excitability," *Somatosens Mot Res.*, Vol. 27, No. 1, pp. 1–8, 2010.
- [14] Connelly, D. M., Rice, C. L., Roos, M. R. and Vandervoort, A. A., "Motor unit firing rates and contractile properties in tibialis anterior of young and old men," *J. Appl. Physiol.*, Vol. 87, pp. 843-852, 1999.
- [15] De Luca, C. J. and Forrest, W. J., "Some properties of motor unit action potential trains recorded during constant force isometric contractions in man," *Kybernetik*, Vol. 12, pp. 160-168, 1973.
- [16] De Luca, C. J. and Hostage, E. C., "Relationship between firing rate and recruitment threshold of motoneurons in voluntary isometric contractions," *J. Neurophysiol.*, Vol. 104, pp. 1034-1046, 2010.
- [17] Dimitrov, G. V. and Dimitrova, N. A., "Fundamentals of power spectra extracellular potentials produced by skeletal muscle fibre of finite length. Part I: effect of fibre anatomy," *Med. Eng. Phys.* Vol. 20, No. 9, pp. 580–587, 1998.

- [18] Dimitrov, G. V., Arabadzhiev, T. I., Mileva, K. N., Bowtell, J. L., Crichton, N. and Dimitrova, N. A., "Muscle fatigue during dynamic contractions assessed by new spectral indices," *Med. Sci. Sports Exerc.*, Vol. 38, No. 11, pp. 1971-1979, 2006.
- [19] Dimitrova, N. A. and Dimitrov, G. V., "Amplitude-related characteristics of motor unit and Mwave potentials during fatigue. A simulation study using literature data on intracellular potential changes found in vitro," *J. Electromyogr. Kinesiol.*, Vol. 12, No. 5, pp. 339-349, 2002
- [20] Facey, A., Irving, R. and Dilworth, L., "Overview of Lactate Metabolism and the Implications for Athletes," *Am. J. Sports Med.*, Vol. 1, No. 3, 42-46. 2013.
- [21] Farina, D., Merletti, R. and Enoka, R. M., "The extraction of neural strategies from the surface EMG," *J Appl Physiol* Vol. 96, No. 4, pp. 1486-1495, 2004.
- [22] Farina, D., "Interpretation of the surface electromyogram in dynamic contractions," *Exerc Sport Sci Rev*, Vol. 34, No. 3, pp. 121-127, 2006.
- [23] Fuglevand, A. J., Macefield, V. G. and Bigland-Ritchie, B., "Force-frequency and fatigue properties of motor units in muscles that control digits of the human hand," *J. Neurophysiol.*, Vol, 81, pp. 1718–1729, 1999.

- [24] Gerdle, B., Larsson, B. and Karlsson, S., "Criterion validation of surface EMG variables as fatigue indicators using peak torque: a study of repetitive maximum isokinetic knee extensions," *J Electromyogr Kinesiol*, Vol. 10, No. 4, pp. 225-232, 2000.
- [25] Gladden, L. B., "Lactate metabolism: a new paradigm for the third millennium," *J Physiol*, Vol. 558, No. 1, pp. 5-30, 2004.
- [26] Gonzalez-Izal, M., Malanda, A., Navarro-Amezqueta, I., Gorostiaga, E. M., Mallor, F., Ibanez J. and Izquierdo M., "EMG spectral indices and muscle power fatigue during dynamic contractions," *J Electromyogr Kinesiol*, Vol. 20, No. 2, pp. 233-240, 2010a.
- [27] Gonzalez-Izal, M., Rodríguez-Carreño, I., Malanda, A., Mallor, F., Navarro-Amezqueta, I., Gorostiaga, E. M. and Izquierdo, M., "sEMG wavelet-based indices predicts muscle power loss during dynamic contractions," *J. Electromyogr. Kinesiol.*, Vol. 20, pp. 1097-106, 2010b.
- [28] Hägglund, M., Waldén, M. and Ekstrand, J., "Risk factors for lower extremity muscle injury in professional soccer the UEFA injury study," *Am J Sports Med.*, Vol. 41, No. 2, pp. 327-335, 2013.
- [29] Ishii, H. and Nichida, Y., "Effect of lactate accumulation during exercise-induced muscle fatigue on the sensorimotor cortex," *J Phys Ther Sci.*, Vol. 25, No. 12, pp. 1637-11642, 2013.

- [30] Jacobs, R. and van Ingen Schenau, G. J., "Control of an external force in leg extensions in humans," *J. Physiol.*, Vol. 457, pp. 611-626, 1992.
- [31] Jakobsson, F., Borg, K. and Edstrom, L., "Fibre-type composition, structure and cytoskeletal protein location of fibres in anterior tibial muscle," *Acta Neuropathol.*, Vol. 80, pp. 459–468, 1990.
- [32] Jang, K. and Kim, Y., "Muscle fatigue analysis by median frequency and wavelet transform during lumbar extension exercises," *J. Biomed. Eng. Res.*, Vol. 25, No. 5, pp. 377-382, 2004.
- [33] Karlsson, S., Yu, J. and Akay, M., "Time–frequency analysis of myoelectric signals during dynamic contractions: a comparative study," *IEEE Trans. Biomed. Eng.*, Vol. 47, No. 2, pp. 228-238, 2000.
- [34] Kim, J., Park, S., Ahn, S. and Kim, Y., "A Novel Approach of Defining Fatigue Indices with sEMG Power during Isotonic Contractions," *J. Precis. Eng. Manuf.*, Vol. 13, No. 6, pp. 977-983, 2012.
- [35] Kim, J., Kim, S., Park, S., Ahn, S., Lee, Y. and Kim, Y., "Bandwidth optimization of the fatigue index to estimate muscle fatigue during dynamic contractions," *J. Precis. Eng. Manuf.* Vol. 14, No. 7, pp. 1185-1191, 2013.

- [36] Kogi, K. and Hakamada, T., "Slowing of surface electromyogram and muscle strength in muscle fatigue," Rep. Inst. Sci. Labour., Vol. 60, pp. 27-41, 1962.
- [37] Komi, P.V. and Tesch, P., "EMG frequency spectrum, muscle structure, and fatigue during dynamic contractions in man," Eur J Appl Physiol Occup Physiol, Vol. 42, No. 1, pp. 41-50, 1979.
- [38] Kukulka, C. G. and Clamann, H. P., "Comparison of the recruitment and discharge properties of motor units in human brachial biceps and adductor pollicis during isometric contractions," Brain Res., 219: 45-55, 1981.
- [39] Lee, K. Y., Shin, K. Y., Kim, H. S. and Mun, J. H., "Estimating Muscle Fatigue of the Biceps Brachii using High to Low Band Ratio in EMG during Isotonic Exercise," Int. J. Precis. Eng. Manuf., Vol. 10, No. 3, pp. 147-153, 2009.
- [40] MacIntosh, B. R. and Rassier, D. E., "What is fatigue?," Can J Appl Physiol Vol. 27, No. 1, pp. 42-55, 2002.
- [41] Masuda, K., Masuda, T., Sadoyama, T., Inaki, M. and Katsuta, S., "Changes in surface EMG parameters during static and dynamic fatiguing contractions," J. Electromyogr Kinesiol., Vol. 9, No. 1, pp. 39-46, 1999.

- [42] Mills, K. R., "Power spectral analysis of electromyogram and compounded muscle action potential during muscle fatigue and recovery," *J. Physiol.*, Vol. 326, pp. 401-409, 1982.
- [43] Moritani, T., Nagata, A. and Muro, M., "Electromyographic manifestations of muscular fatigue," *Med Sci Sports Exerc*, Vol. 14, No. 3, pp. 198-202, 1982.
- [44] Moxham, J., Edwards, R. H. T., Aubier, M., De Troyer, A., Farkas, G., Macklem, P. T. and Roussos, C., "Changes in EMG power spectrum (high-to-low ratio) with force fatigue in humans," *J. Appl. Physiol*, Vol. 53, No. 5, pp. 1094–1099, 1982.
- [45] Potvin, J. R. and Bent, L. R., "Validation of techniques using surface EMG signals from dynamic contractions to quantify muscle fatigue during repetitive tasks," *J Electromyogr Kinesiol*, Vol. 7, No. 2, pp. 131-139, 1997.
- [46] Roos, M. R., Rice, C. L., Connelly, D. M. and Vandervoort, A. A., "Quadriceps muscle strength, contractile properties, and motor unit firing rates in young and old men," *Muscle Nerve*, Vol. 22, pp. 1094-1103, 1999.
- [47] Roy, S. H., De Luca, C. J. and Schneider, J., "Effects of electrode location on myoelectric conduction velocity and median frequency estimates," *J. Appl. Physiol*, Vol. 61, No. 4, pp. 1510–1517, 1986.

- [48] Sato, H., "Functional characteristics of human skeletal muscle revealed by spectral analysis of the surface electromyogram," *Electromyogr. Clin. Neurophysiol.*, Vol. 22, pp. 459–516, 1982.
- [49] Scelsi, R., "Skeletal muscle pathology after spinal cord injury. Our 20-year experience and results on skeletal muscle changes in paraplegics, related to functional rehabilitation," *Basic Appl Myol.*, Vol. 11, No. 2, pp. 75-86, 2001.
- [50] Schantz, P., Randall-Fox, E., Hutchison, W., Tydén, A. and Astrand. P. O., "Muscle fibre type distribution, muscle cross-sectional area and maximal voluntary strength in humans," *Acta Physiol Scand.*, Vol. 117, No. 2, pp. 219-226, 1983.
- [51] Schantz, P.G., Billeter, R., Henriksson, J. and Jansson, E., "Training-induced increase in myofibrillar ATPase intermediate fibres in human skeletal muscle," *Muscle Nerve.*, Vol. 5, pp. 628–636, 1982.
- [52] Sesboüé, B. and Guincestre, J.-Y., "Muscular fatigue," *Ann Phys Rehabil Med.*, Vol. 49, No. 6, pp. 348-354, 2006.
- [53] Stock, M. S., Beck, T. W. and Defreitas, J. M., "Effects of fatigue on motor unit firing rate versus recruitment threshold relationships," *Muscle Nerve*, Vol. 45, pp. 100-109, 2012.

- [54] Sung, E. J., Min, B. C., Kim, S. C., and Kim, C. J., "Effects of oxygen concentrations on driver fatigue during simulated driving," *Appl Ergon*, Vol. 36, pp. 25-31, 2005.
- [55] Thomas, C. K., Johansson, R. S. and Bigland-Ritchie, B., "EMG changes in human thenar motor units with force potentiation and fatigue," *J Neurophysiol* Vol. 95, No.3, pp.1518–1526, 2006.
- [56] Thomas, C. K., Johansson, R. S. and Bigland-Ritchie, B., "Attempts to physiologically classify human thenar motor units," *J. Neurophysiol.*, Vol. 65, pp. 1501–1508, 1991.
- [57] Travnik, L., Pernus, F. and Erzen, I., "Histochemical and morphometric characteristics of the normal human vastus medialis longus and vastus medialis obliquus muscles," *J Anat.*, Vol. 187, pp. 403-411, 1995.
- [58] van Dieen, J. H., Boke, B., Oosterhuis, W. and Toussaint, H. M., "The influence of torque and velocity on erector spinae muscle fatigue and its relationship to changes of electromyogram spectrum density," *Eur J Appl Physiol Occup Physiol*, Vol. 72, No. 4, pp. 310-315, 1996.
- [59] Van Rijn, R. M., Huisstede, B. M., Koes, B. W. and Burdorf, A., "Associations between work-related factors and specific disorders at the elbow: a systematic literature review," *Rheumatology (Oxford)*, Vol. 48, 528-536, 2009.

[60] Walker-Bone, K. and Cooper, C., "Hard work never hurt anyone: or did it? A review of occupational associations with soft tissue musculoskeletal disorders of the neck and upper limb," *Annals of Rheumatic Diseases*, Vol. 64, 1391-1396, 2005.

[61] Work-related Musculoskeletal Disorders (WMSDs)
"http://www.ccohs.ca/oshanswers/diseases/rmirsi.html," (last updated on January 8, 2014).

[62] Wright, M. H., " The interior-point revolution in optimization: History, recent developments, and lasting consequences," *Bull. Am. Math. Soc.*, Vol. 42, No. 1, pp. 39-56, 2005.

Appendices

Appendix A. MATLAB code to compute EMG parameters

```
function [ cc, results ] = Fdataanl( fcut, emg, force, velocity, force_ex_MVC )

fs = 1000;

Power = force.*velocity;

[low_b,low_a]=butter(2,3/fs*2);%3Hz

[low2_b,low2_a]=butter(1,200/fs*2);%200Hz

[high_b,high_a]=butter(2,fcut/fs*2,'high');

[high2_b,high2_a]=butter(2,10/fs*2,'high');

LPF = filter(high2_b,high2_a,emg);

LPFF = filter(low2_b,low2_a,LPF);

%linear envelop

LPF_REC = abs(LPFF);

LPF_LE = filter(low_b,low_a,LPF_REC);

HPF = filter(high_b,high_a,emg);

%linear envelop

HPF_REC = abs(HPF);
```

```
HPF_LE = filter(low_b,low_a,HPF_REC);
```

```
HPF_LE_1 = HPF_LE;
```

```
LPF_LE_ = LPF_LE;
```

```
NEW_HPF1 = HPF_LE_1./(LPF_LE_+1);
```

```
maxvel=0;
```

```
maxpow=0;
```

```
emgon=0;
```

```
k=1;
```

```
%cwt
```

```
cwtsum=zeros(length(LPF),3);
```

```
ccfs=cwt(LPF(:,1),1:32,'sym5','plot');
```

```
cwtsum(:,1)=sum(ccfs);
```

```
for i=1:length(force)
```

```
    if(emgon==0 && force(i)>force_ex_MVC*0.2)
```

```
        emgon=1;
```

```
        results(k,1)=i;
```

```
    end
```

```
    if(emgon==1)
```

```

if(maxvel<Power(i))
    maxvel=Power(i);
    maxpow=Power(i);
end

if(force(i)<force_ex_MVC*0.1)
    emgon=0;
    if (i-results(k,1)>200)
        results(k,2)=i;
        results(k,3)=maxpow;% maxvel*maxfor;
        maxvel=0;
        maxpow=0;
        results(k,4) = MedianFreq(LPF(results(k,1):results(k,2), 1),fs);

        results(k,5) = MedianFreq(cwtsum(results(k,1):results (k,2),1),fs);

        tmp = fft(LPF(results(k,1):results(k,1)+1024,1));
        [numRows, numCols] = size(tmp);
        RAW_tmp1 = zeros(3);
        RAW_tmp2 = zeros(3);
        for j = 8:numRows/2
            RAW_tmp1(1) = RAW_tmp1(1) + abs(tmp(j))/(j*500/numRows*2);
            RAW_tmp2(1) = RAW_tmp2(1) + abs(tmp(j))*((j*500/numRows*2)
^5);
        end
end

```

```

results(k,6) = RAW_tmp1(1)/RAW_tmp2(1);

%DWT

len=length(emg(results(k,1):results(k,1)+1024,1));

[ce,1]=wavedec(emg(results(k,1):results(k,1)+1024,1), 5,'sym5');
[ce,1]=wavedec(LPF(results(k,1):results(k,1)+1024,m), 5,'sym5');

% Compute and reshape DWT to compare with CWT.

cfd=zeros(5,len);

for z=1:5

    d=detcoef(ce,1,z);

    d=d(:,ones(1,2^z))';

    cfd(z,:)=wkeep(d(:,len));

end

cfd=cfd(:);

I=find(abs(cfd) < sqrt(eps));

cfd(I)=zeros(size(I));

cfd=reshape(cfd,5,len);

WIREsum1=0;

WIREsum5=0;

for x=1:len

    WIREsum1=WIREsum1+cfd(1,x)^2;

    WIREsum5=WIREsum5+cfd(5,x)^2;

end

results(k,7)=WIREsum5/WIREsum1;

```

```
results(k,8) = mean(NEW_HPF1(results(k,1):results(k, 2),1));

results(k,9) = mean(emg(results(k,1):results(k,2),1). ^2)^0.5;

k=k+1;

end

end

end

end

cc_tmp = corrcoef(results);

cc = cc_tmp;

end
```

Appendix B. MATLAB code to optimize the cut-off frequency of high-pass filter

```
clear all;

% 피험자 구분파일명 입력
fnindex = ['HSJ';'KBJ';'KHN';'LHY';'SIS'];

% 최적화 옵션
options = optimset('fmincon');
options.Display = 'iter';
options.MaxIter = 1000;
options.TolFun = 1.0000e-020;
options.TolX = 1.0000e-010;
options.Algorithm = 'interior-point';

k=1;

[numRows numCols] = size(fnindex);
result=zeros(numCols,2);
fi_result=zeros(1,8);
fs = 1000;

for z = 1: numRows
    %MVC torque 측정
    [time torque velocity angle emg_VM emg_RF emg_VL markers activity]
    =
```

```
textread([fnindex(z,:),'_TC_MVC_01.txt'],'%f%f%f%f%f%f%f%f%f%d','header
lines',5);
```

```
[lowf_b,lowf_a]=butter(1,15/500);% 15Hz
```

```
force_ex = filter(lowf_b,lowf_a,torque);
```

```
RAW_ex = [emg_VM emg_RF emg_VL];
```

```
RAW_ex_notch = RAW_ex;
```

```
[force_ex_max,MVC_ex_time] = max(force_ex);
```

```
for i = 1: length(force_ex)
```

```
    if (force_ex(i) > force_ex_max * 0.85)
```

```
        MVC_ex_start2 = i;
```

```
        break;
```

```
    end
```

```
end
```

```
force_ex_MVC = mean(force_ex(MVC_ex_start2:
MVC_ex_start2+1000));
```

```
% 피로운동 중 peak power 계산 및 근전도 선처리
```

```
EMG = [0 0 0];
```

```
Force = 0;
```

```
Velocity = 0;
```

```
for l=1:5
```

```
    [time torque velocity angle emg_VM emg_RF emg_VL markers
activity] = textread([fnindex(z,:),'_TC_fatigue_0',num2str(l),'.txt'],
'%f%f%f%f%f%f%f%f%f%d','headerlines',5);
```

```
[lowf_b,lowf_a]=butter(1,15/500);% 15Hz
```

```

force = filter(lowf_b,lowf_a,torque);

Power = force.*velocity;

RAW = [emg_VM emg_RF emg_VL];

RAW_notch = RAW;

EMG = [EMG',RAW_notch]';

Force = [Force',force]';

Velocity = [Velocity',velocity]';

end

%FIhr1 계산 및 peak power 와의 correlation coefficient 계산함수를
통한 최적 주파수 도출

[x, fval] = fmincon(@(x)FIhr1(x, EMG(:,3), Force, Velocity,
force_ex_MVC),350,[],[],[],[],1,479,[],options);

%최적 주파수에서의 peak power 와 FIhr1 값 추출

[cc, pp_tmp] = Fdataanl(x, EMG(:,3), Force, Velocity, force_ex_MVC);

pp_result = fi_result;

[numRows numCols] = size(fi_result);

fi_result = zeros(numRows+50,numCols);

for i= 1:7

    fi_result(:,i) = [pp_result(:,i)', pp_tmp(:,i+2)'./((pp_tmp(1,i+2)+
pp_tmp(2,i+2))/2)*100]';

end

result(z,1) = x;

result(z,2) = cc(3,8);

result(z,3) = cc(3,4);

result(z,4) = cc(3,5);

```

```
result(z,5) = cc(3,6);  
result(z,6) = cc(3,7);  
result(z,7) = cc(3,9);  
  
end  
  
fi_result = fi_result(2:end,:);  
fi_result(:,4) = log10(fi_result(:,4));  
fi_result(:,5) = log10(fi_result(:,5));
```

Abstract (in Korean)

새로운 필터 기반 피로 지수의 개발 및 검증

연세대학교 의공학과 대학원

김정윤

본 연구에서는 근피로 추정을 위하여 근전도 신호를 이용한 새로운 필터 기반 근피로 지수를 개발하였다.

근피로는 힘든 일로 인하여 일시적으로 힘과 에너지를 잃는 것으로, 운동의 방해요소로 작용하며 이로 인해 근피로는 골격근 부상에 주요한 원인이 되므로 근피로를 측정하는 것이 중요하다.

필터 기반 근피로 지수는 근전도의 고주파수 대역 성분과 저주파수 대역 성분의 비율을 이용하여 근피로를 추정하도록 개발되었다. 개발된 근피로 지수의 주파수 대역을 결정하기 위해, 고역 통과 필터의 차단주파수가 최대 관절과워와 개발된 근피로 지수의 상관관계를 최대화 하도록 최적화 하였다. 개발된 근피로 지수를 평가하기 위하여 다른 근피로 지수들과 피로 추정 정확성을

비교하였고, 다양한 근육, 다양한 운동, 다양한 전극간 간격에 적용하여 최적화된 주파수 대역을 검증하였다.

제안된 필터 기반 근피로 지수는 다양한 근육, 다양한 운동 방법, 다양한 전극간 간격에서 기존의 근피로 지수들 보다 높은 정확도를 보였다. 이를 통하여 필터 기반 근피로 지수가 기존의 근피로 지수들 보다 정확하게 피로를 추정 할 수 있음을 확인하였다. 또한, 필터 기반 근피로 지수의 고역통과 필터의 최적화된 차단주파수가 모든 근육, 모든 운동방법, 모든 전극간 간격에서 350Hz 근처로 통계적으로 유의한 차이를 보이지 않았다. 이 결과는 350Hz 고역 통과 필터가 필터 기반 근피로 지수의 일반적인 해법이 될 수 있음을 보여주며, 필터 기반 근피로 지수가 운동 단위의 점화율을 대변함을 확인할 수 있었다. 이 결과는 향후 근육의 피로를 추정하는데 많은 도움이 될 것으로 판단된다.

본 연구에서 개발한 필터 기반 근피로 지수는 운동단위의 점화율을 대변 할 수 있는 반면 실제로 운동단위의 점화율을 피로 운동 도중 측정하지는 못하는 한계를 가지고 있다. 향후 운동단위 활동전위 분해 연구나 삽입 전극 근전도 측정을 통하여 운동단위의 점화율을 직접 측정한다면 필터 기반 근피로 지수의 특징을 검증할 수 있을 것이다.