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An EMG-based muscle force monitoring system †

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Abstract

Information about muscle forces helps us to understand human movements more completely. Recently, studies on estimating muscle forces in real-time have been directed forward; however, the previous studies have some limitations in terms of using a three-dimensional (3D) motion capture system to obtain human movements. In the present study, an electromyography (EMG)-based real-time muscle force estimation system, which is available for a variety of potential applications, was introduced with electrogoniometers. A pilot study was conducted by performing 3D motion analysis on ten subjects during sit-to-stand movement. EMG measurements were simultaneously performed on gastrocnemius medialis and tibialis anterior. To evaluate the developed system, the results from the developed system were compared with those from widely used commercially available off-line simulation software including a musculoskeletal model. Results showed that good correlation coefficients between muscle forces from the developed system and the off-line simulation were observed in gastrocnemius medialis ($r = 0.718$, $p < 0.01$) and tibialis anterior ($r = 0.821$, $p < 0.01$). However, muscle lengths and muscle forces were obtained with a maximum delay of about 100 ms. Further studies would be required to solve the delay limitation. The developed system yielded promising results, suggesting that it can be potentially used for the real-time diagnosis of muscle or interpretation of movements.

Keywords: Muscle force; Muscle length; Joint angle; Electromyography (EMG); Electrogoniometer

1. Introduction

Information about muscle forces helps us to understand human movements more completely. Also, since insufficient muscle forces mean that the activities related to roles of the muscles cannot be properly performed, measurements of muscle forces are considered to be very useful clinically needed to enable clinicians to judge a patient's potential for function [1]. However, it is not applicable to non-invasively measure muscle forces in dynamic condition. Therefore, many researches have been directed to develop musculoskeletal models which estimate muscle forces [2, 3].

Since muscle forces cannot be measured non-invasively and can be estimated from off-line processing after experiments, there are few exercise or rehabilitation protocols to control muscle forces. If on-line monitoring of muscle forces is possible, real-time muscle forces estimation allows the design of biofeedback protocols in which subjects can control specific muscle forces [4]. Recently, Bogert et al. developed a real-time system based on the optimization of muscle forces. They showed that calculated muscle forces were similar to EMG data in pattern. Since, however, they used a three-dimensional (3D) motion capture system to obtain human movements, the system developed by them is only used in a very limited space that almost reflective markers can be detected by 3D motion capture system. Force plates are also required to perform inverse dynamics for optimization algorithm [5, 6]. Therefore, there are disadvantages of spatial limits and high cost. In addition, it is limited to only closed chain movement, since inverse dynamics should be performed. Also, an optimization technique for calculating muscle forces cannot consider muscle co-contraction condition.

Inputs in muscle dynamics are muscle length and electromyography (EMG) signals [3]. Muscle length is a function of joint angle [7], and joint angle can be easily measured by using an electrogoniometer. In particular, the output signal of an electrogoniometer is available immediately for recording or converting into the computer. It is possible to obtain information about muscle length by using electrogoniometers instead of the 3D motion capture system.

The purpose of the present study is to realize an EMG-based real-time muscle force estimation system which is available without 3D motion capture system and is applicable to a variety of potential applications such as human machine interface (HMI), visual feedback, and so on. To evaluate the developed system, a pilot study was conducted for two major muscles

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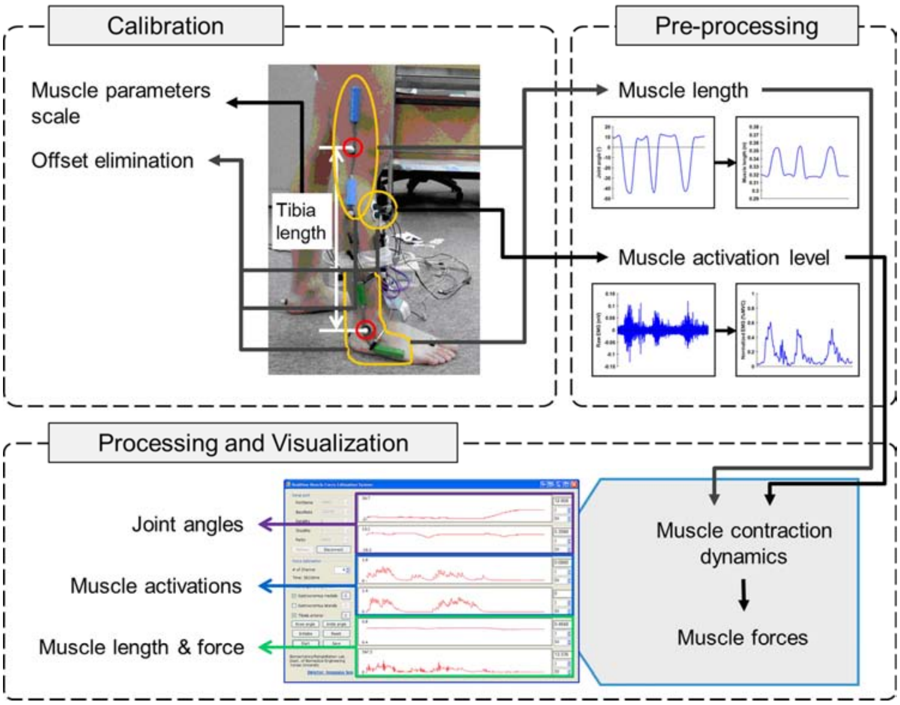

Fig. 1. Schematic diagram to elucidate the concepts of the real-time muscle force estimation system.

about an ankle joint during sit-to-stand, which is one of the most frequent movements in daily life as well as one of the clinical evaluation tools [8, 9]. The results from the developed system were compared with those from the off-line simulation software, SIMM, including a musculoskeletal model.

2. Methods

2.1 Concept of the real-time muscle force estimation system

The suggested real-time muscle force estimation system consists of three parts (Fig. 1): (1) calibration, (2) pre-processing, (3) processing and visualization. In the calibration part, the muscle parameters (tendon slack length, optimal fiber length, and maximum isometric force) are scaled to a subject according to the subject's segment length. In the present study, two major muscles about ankle joint, gastrocnemius medialis and tibialis anterior, were evaluated and tibia length was measured. The scaling algorithm assumes that the ratio among each muscle parameter does not change. After electrogoniometers and surface EMG (sEMG) electrodes are attached on the subject's skin, the offsets of the electrogoniometers' signal and sEMG baseline are eliminated. Electrogoniometer offset was defined as the 5-s mean value when the joint angle is anatomically zero, and the EMG baseline was defined the 5-s mean value when the subject sits a chair.

In the pre-processing part, the joint angles and EMG data are transformed to muscle lengths and muscle activation levels, respectively. In the processing and visualization part, muscle forces are calculated from the muscle lengths and muscle activation levels. To estimate muscle forces in real-time, computational cost should be minimized. The muscle model proposed by Zajac, therefore, was used, since it was designed to be significantly faster than any other model by assuming an infinitely stiff tendon.

2.2 Tibia length and muscle parameters

To scale the un-scaled muscle parameter values in SIMM (Musculographics, Inc., USA) to a subject, the relationship between tibia length and muscle parameters was studied. Tibia length was defined the distance from a point on the skin covering the lateral femoral epicondyle to a point on the skin covering the lateral malleolus [7]. After a subject's motion data captured on the anatomical position from 3D motion capture system (VICON, UK) were imported into the musculoskeletal modeling software SIMM, scaled muscle parameters were then obtained. Linear regression analysis was performed between tibia length and scaled tendon slack length, which is one of the muscle parameters. The ratio of unscaled tendon slack length to estimated tendon slack length was used to scale other muscle parameters: optimal fiber length, and maximum isometric force.

2.3 Muscle activation level

To estimate muscle activation level, raw sEMG signals are first high-pass filtered at 20 Hz; second, full-wave rectified; third, low-pass filtered at 5 Hz, and last, normalized by MVC value. This processed EMG was used as the muscle activation level to reduce computational cost, since it is the simplest muscle activation [10]. It was also roughly assumed that EMG-to-force relationship is linear for low cost calculation. All mentioned filters are Butterworth 1st order.

2.4 Joint angles and muscle length

To estimate muscle length, joint angles were used instead of the 3D motion capture system. To estimate muscle length from joint angles, cubic regression analysis was performed between joint angles and muscle length. After joint angles were inputted to SIMM as 0 ~ 120˚ and -50 ~ 30˚ for knee and ankle joints, respectively, muscle lengths were then obtained. In gastrocnemius medialis, it was assumed that changes in muscle length are independent for each joint.

2.5 A pilot experiment

Ten male subjects (height: 174.6 ± 8.5 cm, weight: 70.0 ± 7.1 kg, age: 26.1 ± 1.9 years old) participated in this study, who self-reported no history of a neural and musculoskeletal disease. Before the beginning of experiment, subjects read and signed an informed consent. Sixteen reflective markers were attached on subjects' anatomical landmark, and sEMG electrodes were placed on the gastrocnemius medialis and tibialis anterior muscles by instructions for fine-wire placement [11]. For the measurement of joint angles, two electrogoniometers (SG110 and SG150, Biometrics Ltd., UK) were placed on the lateral side of the knee and ankle joints. MVC tests were conducted during ten seconds according to the instructions [12], and MVC values were defined maximum from 1-s sliding mean. Subjects were asked to stand on a chair. During the experiment, trajectories of the reflective markers were stored at 120 Hz.

The muscle lengths and muscle activation levels were wirelessly transferred at 100 Hz to the computer using Bluetooth communication. The processing and visualization of the signals were performed in an Intel® Pentium4 - 3.2 GHz personal computer with 2 GB memory.

2.6 Comparison methods

To evaluate the developed system, the correlation coefficient (r) and significance (p) were calculated between muscle forces from the developed system and from SIMM using the statistical software SPSS (SPSS Inc., USA). RMS difference (d) was also compared.

3. Results

Tendon slack length of gastrocnemius medialis and tibialis anterior muscles significantly increased in accordance with increasing tibia length (Fig. 2). There was a good linear correlation between tibia length and tendon slack length of gastrocnemius medialis ($r = 0.998$). Another good linear correlation was also observed between tibia length and tendon slack length of tibialis anterior ($r = 0.999$). Correlations were significant in their two relationships ($p < 0.01$). Relative errors between tendon slack length from the developed system and from SIMM were less than ±1% (Table 1). Since tendon slack length was used to scale optimal fiber length and maximum isometric force, relative errors of optimal fiber length and maximum isometric force were the same as those of tendon slack length.

Table 1. Comparison of average muscle parameters from the developed system and from SIMM (n = 10).

	GCM		TA	
	Developed	SIMM	Developed	SIMM
Tendon slack length (m)	0.453794 (0.031124)	0.453688 (0.031162)	0.247350 (0.016215)	0.247276 (0.016254)
Optimal fiber length (m)	0.050051 (0.003433)	0.050039 (0.003437)	0.108388 (0.006898)	0.108677 (0.007132)
Maximum isometric force (N)	1237.9 (85.0)	1237.6 (85.0)	668.9 (44.0)	668.8 (44.0)
Error rate (%)	0.02321		0.03025	

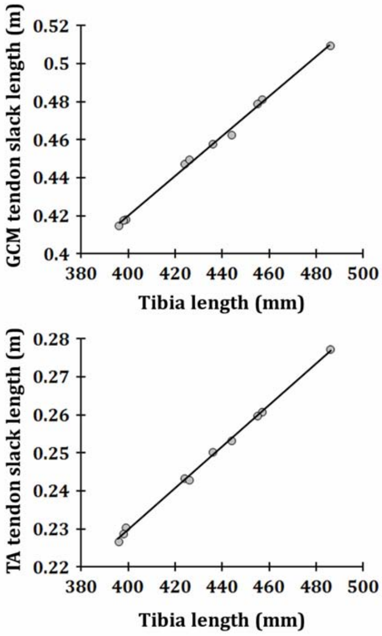

Fig. 2. Relationship between tendon slack length and tibia length. Excellent linear correlation was found in both cases. Abbreviations: GCM = gastrocnemius medialis, TA = tibialis anterior.

Joint angles from the developed system were similar to those from motion analysis (Fig. 3). Excellent correlations were observed in knee joint angle ($r = 0.999$, $p < 0.01$) and ankle joint angle ($r = 0.950$, $p < 0.01$). RMS differences were about 4˚ for knee joint (range: 4.268 ~ 87.021˚) and about 0.2˚ for ankle joint (range: 5.569 ~ 16.195˚) (Table 2). Quantization errors of approximately 1.5˚ were observed due to low resolution of the electrogoniometers that we used.

Muscle lengths from the developed system were similar to those from SIMM (Fig. 4). Excellent correlations were found in gastrocnemius medialis ($r = 0.997$, $p < 0.01$) and tibialis anterior ($r = 0.975$, $p < 0.01$). RMS differences were less than

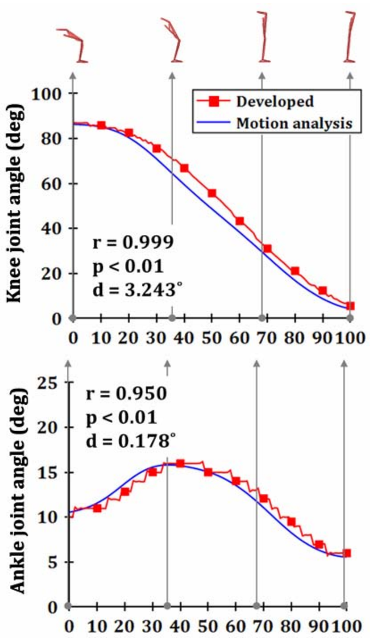

Fig. 3. Joint angles in sit-to-stand movement. Excellent correlations were observed in knee and ankle joints. 0~100 indicate the normalized time phase from sit-to-stand.

approximately 0.4 mm in gastrocnemius medialis (range: 439 ~ 457 mm) and approximately 0.2 mm in tibialis anterior (range: 290 ~ 297 mm) (Table 2). Quantization errors were also found and these originated from joint angles.

Muscle forces from the developed system were similar to those from SIMM, but with relatively low correlations in comparison to muscle length (Fig. 4). Good correlation coefficients were observed in gastrocnemius medialis ($r = 0.718$, $p < 0.01$) and tibialis anterior ($r = 0.821$, $p < 0.01$). RMS differences were less than approximately 28 N in gastrocnemius medialis and approximately 12 N in tibialis anterior (Table 2).

4. Discussion and conclusion

The objective of this study was to implement a real-time muscle force estimation system without a 3D motion capture system. Up to now, there has been no gold standard method for calculating muscle forces, but at least, the developed system is meaningful in terms of implementing real-time muscle force estimation with similar results of widely used off-line software, SIMM.

The developed system was performed on an Intel® Pentium4 – 3.2 GHz personal computer environment with 2 GB memory. The operating system was Microsoft Windows XP® and the main program for processing and visualization was only executed to reduce the system load. The developed system delayed approximately 100 ms, and it may be taken in transferring, processing and visualizing the signals. This amount of delay may not be serious for clinical application, but cannot be allowed for the real-time HMI system or visual feedback applications. Most delay occurred during the wireless communication. The signal transmission along wire is generally faster than the wireless communication. The computer specification is also important to reduce the delay time. Therefore, a wired version with the recent personal computer specification may operate in real-time for the HMI system or visual feedback applications.

The relationship between joint angles and muscle lengths was considered for only flexion and extension. Since the range of motion (ROM) on the sagittal plane is the largest in sit-to-stand movement, reasonable results could be obtained from the developed system. If ROM on the other planes is relatively large, good results may not be expected. Also, two muscles, gastrocnemius medialis and tibialis anterior, were only evaluated in the present study, but there are many important lower extremity muscles such as the hamstring muscle group. Therefore, the relationship between joint angles and muscle lengths on the other planes and muscles needs to be studied.

The results in muscle forces showed relatively less correlation than those of joint angles or muscle lengths. Muscle forces from the developed system included much noise. This result can be explained as follows. The resolution of electrogoniometers was low, and muscle lengths also included the same amount of error with electrogoniometers. Fiber contraction velocity is one of factors that affect to generate active force, and is calculated by differentiating fiber length which is related to muscle length. Quantization errors that were amplified through the differentiation process affected muscle forces. If these muscle forces are used to determine joint moment or joint power without eliminating much noise, the resultants will not be reasonable. This problem should be solved and it could be possible by reducing quantization errors, i.e., replacing electrogoniometers with the higher resolution, or low-pass filtering with adequate cut-off frequency.

In the present study, the muscle model designed by Zajac was used because it is significantly faster than any other model. However, for some simulations the assumption of an infinitely stiff tendon is not appropriate, such as those involving muscles with very long tendons or high forces. Therefore, other muscle models, Schutte's model [13] or GTO model [14], need to be considered for higher accuracy. However, if the cost of model complication were increased, the delay time would be increased. Therefore, further studies would be required to solve this trade-off problem.

Even though there are problems that need to be resolved, the present study has some advantages compared to the previous study in which a 3D motion capture system was used with optimization technique. First, almost reflective markers should be sensed by infra-red cameras in using a 3D motion capture system to detect motions. Moreover, the ground reaction force should be obtained by force plates due to use of optimization technique. Therefore, it can be possible within the space where markers can be detected by infra-red cameras and in which ground reaction force can be obtained by force plates. On the other hand, the developed system is available anywhere in the communication distance of Bluetooth (maximum

Table 2. Comparison of joint angles, muscle lengths, and muscle forces from the developed system and from SIMM.

Subject	Joint	Joint angles(˚)	Muscle	Muscle lengths (m)	Muscle forces (N)
		d (r*)		d (r*)	d (r*)
SRH	Knee	2.146 (0.991)	GCM	0.002 (0.912)	3 (0.967)
	Ankle	3.596 (0.928)	TA	0.005 (0.844)	9 (0.863)
YJY	Knee	1.964 (0.996)	GCM	0.001 (0.994)	11 (0.936)
	Ankle	2.092 (0.840)	TA	0.006 (0.679)	8 (0.738)
LGD	Knee	1.194 (0.997)	GCM	0.001 (0.985)	55 (0.848)
	Ankle	3.335 (0.966)	TA	0.001 (0.969)	9(0.875)
LJH	Knee	0.155 (0.999)	GCM	0.001 (0.984)	47 (0.774)
	Ankle	0.206 (0.971)	TA	0.002 (0.967)	20 (0.775)
JS	Knee	1.818 (0.964)	GCM	0.004 (0.982)	29 (0.804)
	Ankle	5.104 (0.867)	TA	0.001 (0.825)	0 (0.939)
CYK	Knee	3.243 (0.999)	GCM	0.000 (0.997)	28 (0.718)
	Ankle	0.178 (0.950)	TA	0.000 (0.975)	12 (0.821)
CDS	Knee	1.998 (0.974)	GCM	0.001 (0.742)	30 (0.938)
	Ankle	1.027 (0.900)	TA	0.001 (0.903)	1 (0.877)
LJS	Knee	1.937 (0.988)	GCM	0.001 (0.952)	2 (0.725)
	Ankle	1.029 (0.923)	TA	0.001 (0.941)	29 (0.560)
PSW	Knee	0.001 (0.959)	GCM	0.002 (0.935)	14 (0.806)
	Ankle	2.844 (0.884)	TA	0.001 (0.860)	1 (0.691)
KHD	Knee	6.132 (0.997)	GCM	0.001 (0.988)	9 (0.915)
	Ankle	0.203 (0.966)	TA	0.004 (0.425)	1 (0.872)

d: RMS difference
r: correlation coefficient
GCM: gastrocnemius medialis
TA: tibialis anterior
* Above all correlation coefficients were significant ($p < 0.01$).

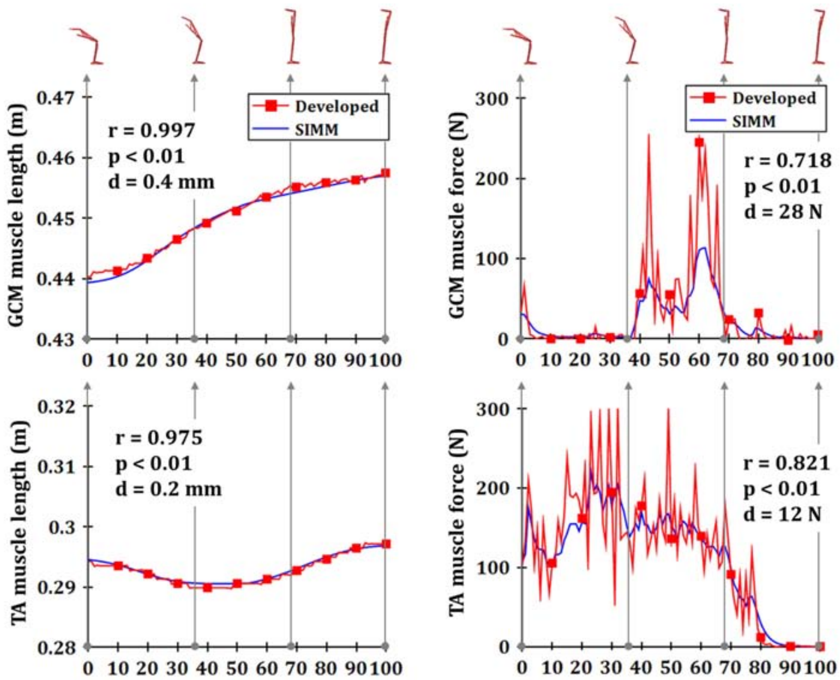

Fig. 4. Muscle lengths (left column) and muscle forces (right column). Excellent correlations were found in GCM and TA muscle length. As results of muscle forces good correlations were observed with similar patterns, but relatively low in comparison to muscle lengths. Abbreviations: GCM = gastrocnemius medialis, TA = tibialis anterior.

100 m in case of this study). This allows applying to an analysis of movements which require a wide space, i.e., sprint. Second, many reflective markers are required to detect human motion with a 3D motion capture system. In this study, however, muscle lengths can be obtained by a small number of electrogoniometers. Lastly, the developed system is economic in terms of prices compared to a 3D motion capture system, which makes clinically wide use possible.

The developed system yielded promising results, suggesting that the developed system can be potentially used for the real-time diagnosis of muscle or interpretation of movements. Another possible application of the developed system is muscle force-based human machine interface (HMI) system. There are many studies related to EMG-based HMI [15-17]. On the other hand, muscle force-based HMI research rarely follows the studies that have been developed for EMG-based HMI. Moreover, the developed system can be used for on-line monitoring of muscle forces. Therefore, the developed system makes it possible to confirm that subjects are performing well according to the experimenter's instructions and to modify experimental protocols without stopping the on-going measurement [18].

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