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System identification of velocity mechanomyogram measured with a capacitor microphone for muscle stiffness estimation ☆

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ABSTRACT

A mechanomyogram (MMG) measured with a displacement sensor (displacement MMG) can provide a better estimation of longitudinal muscle stiffness than that measured with an acceleration sensor (acceleration MMG), but the displacement MMG cannot provide transverse muscle stiffness. We propose a method to estimate both longitudinal and transverse muscle stiffness from a velocity MMG using a system identification technique. The aims of this study are to show the advantages of the proposed method. The velocity MMG was measured using a capacitor microphone and a differential circuit, and the MMG, evoked by electrical stimulation, of the tibialis anterior muscle was measured five times in seven healthy young male volunteers. The evoked MMG system was identified using the singular value decomposition method and was approximated with a fourth-order model, which provides two undamped natural frequencies corresponding to the longitudinal and transverse muscle stiffness. The fluctuation of the undamped natural frequencies estimated from the velocity MMG was significantly smaller than that from the acceleration MMG. There was no significant difference between the fluctuations of the undamped natural frequencies estimated from the velocity MMG and that from the displacement MMG. The proposed method using the velocity MMG is thus more advantageous for muscle stiffness estimation.

1. Introduction

Muscle stiffness plays an important role in human motion, as stiffness can store kinetic energy as elastic energy, and release it to achieve efficient and smooth motion. For example, muscle stiffness is related to the rate of crank torque development (Watsford et al., 2010) and is proportional to power (Uchiyama et al., 2015) during cycling exercise. Musculoarticular stiffness has been investigated using mechanical perturbations such as random disturbance (Ishida et al., 2008), ramp stretch (Gurfinkel et al., 1997), and mechanical impulse (Hunter and Spriggs, 2000; Fukashiro et al., 2001; Murphy et al., 2003). Such techniques assess stiffness around the joint.

Muscle stiffness can be measured using a myotonometer (Bizzini and Mannion, 2003; Ditrolio et al., 2011; Aird et al., 2012), where mechanical impulse-like stimulation is applied to the tissue through a probe, and the damped sinusoidal vibration is measured. This technique assesses transverse muscle stiffness; longitudinal muscle stiffness can be assessed using electrical stimulation and evoked mechanomyogram (MMG). Stimulating the tibialis anterior muscle with an amplitude-modulated electrical pulse train provides the transfer function from the stimulation to the MMG measured with a laser displacement meter. The transfer function has a natural frequency coinciding with that from the stimulation to the torque (Orizio et al., 2008). Recently, an evoked MMG has been regarded as an impulse response, and muscle stiffness has been estimated using a system identification technique. The evoked MMG has been previously measured using both displacement and acceleration sensors (Uchiyama and Shinohara, 2013).

Hereafter, we denote MMGs measured with displacement and acceleration sensors as displacement MMGs and acceleration MMGs, respectively. A displacement MMG system can be approximated with a second-order model, which has an undamped natural frequency corresponding to longitudinal muscle stiffness; however, it cannot provide estimation of transverse muscle stiffness. An acceleration MMG system can be approximated with a sixth-order model, which provides three undamped natural frequencies corresponding to the longitudinal and transverse muscle stiffness, as well as the subcutaneous tissue stiffness. The longitudinal muscle stiffness in this system, however, fluctuates more than that in a

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displacement MMG system. Therefore, we propose a method to estimate longitudinal and transverse stiffness with fewer fluctuations by measuring an MMG as a velocity using a capacitor microphone and a differential circuit. The aim of this study is to demonstrate the advantages of using this method to estimate muscle stiffness via the identification of a velocity MMG system.

2. Methods

Seven young male volunteers aged from 22 to 24 participated in the experiment. None of the participants had histories of neuromuscular disorders. Ethics committee approval (No. 27–34; Faculty of Science and Technology, Keio University) was obtained, and all participants gave their informed consent in accordance with the Declaration of Helsinki.

Electrical stimulation was applied to the common peroneal nerve, and an evoked MMG was measured on the tibialis anterior muscle. Each participant sat on a chair, and had their thigh and foot fixed to the equipment with nylon belts (Fig. 1a). The common peroneal nerve was stimulated with an electrical stimulator (SEN-3301; Nihon Kohden, Tokyo, Japan) and an isolator (S-104 J; Nihon Kohden) using Ag-AgCl electrodes (F-150S; Nihon Kohden). The stimulation was a mono-polar rectangular pulse of 500-μs width, and its strength was supramaximal. The stimulation was repeated 30 times with an inter-pulse-interval of 1 s in a trial. Five trials were conducted.

Velocity and displacement MMGs were measured using a capacitor microphone (MX-E4758; Primo, Tokyo, Japan). The microphone was fixed to an air chamber made with ABS resin (MX-5072D; Primo) as shown in Fig. 1b. The total weight of the microphone and the air chamber was 0.78 g. The sensitivity of the microphone was −43 ± 3 dB at 1 kHz, where 0 dB was 1 V/Pa. The microphone fixed to the air chamber had flat gain characteristics from 0.5 to 4000 Hz, and was connected to the driver circuit involving an RC high-pass filter that separated the signal from the bias voltage of the microphone (Fig. 2). The cutoff frequency of the high-pass filter was 106 Hz for the measurement of the velocity MMG, and 0.16 Hz for the measurement of the displacement MMG. The frequency range of a displacement MMG is less than 100 Hz, so the RC high-pass filter with a cutoff frequency of 106 Hz could well differentiate the displacement MMG because the slope of the gain characteristics was 20 dB/decade. The velocity and displacement MMGs were amplified (×67.7 for velocity, ×3.1 for displacement) and then filtered with a Sallen–Key low-pass filter (cutoff frequency: 100 Hz). The filtered signal was sampled at 2000 Hz with a 16-bit analog-to-digital converter (cRIO-9215, National Instruments, Austin, TX, USA), then stored on a computer. The acceleration MMG was measured with an acceleration sensor (MP-110-10-101, Medisens, Saitama, Japan) attached at the measurement point with adhesive tape. The acceleration was amplified and filtered (1–250 Hz) with an MMG amplifier (MPS110, Medisens; Saitama), and the acceleration MMG was also sampled at 2000 Hz and stored on a personal computer.

The capacitor microphone with the air chamber was fixed over the tibialis anterior muscle using ring-shaped double-sided adhesive tape. The fixed position was $L/3$ from the caput fibulae, where L is the distance from the caput fibulae to the lateral malleolus. The acceleration MMG was also measured at the same position.

The evoked MMGs were synchronously averaged and then used for system identification. The transfer function from the stimulation pulse to the evoked MMG was identified by the singular value decomposition method, as demonstrated in previous studies (Uchiyama and Shinohara, 2013; Uchiyama and Sakai, 2013; Uchiyama et al., 2015; Fukawa and Uchiyama, 2016). Here, we provide a brief description of this method. The system was assumed to be linear time invariant for a short time after the electrical stimulation. The system can be described in a state space as follows:

$$\begin{cases} \boldsymbol{x}(k+1) = \boldsymbol{A}\boldsymbol{x}(k) + \boldsymbol{b}u(k) \\ y(k) = \boldsymbol{c}^T\boldsymbol{x}(k) + du(k) \end{cases}, \quad (1)$$

where $u(k)$ is an input (electrical stimulation), $y(k)$ is an output (evoked MMG), and $\boldsymbol{x}(k)$ is a state vector. The input and output were regarded as ideal impulse and impulse response, respectively. The coefficient d is equal to $y(0)$. The matrix and vectors, $\boldsymbol{A}$, $\boldsymbol{b}$, and $\boldsymbol{c}$ were

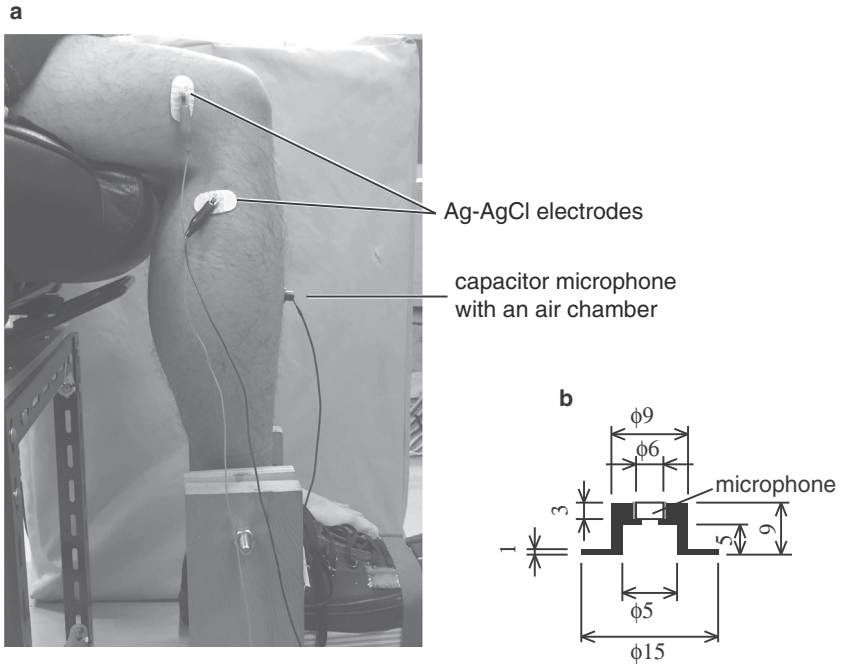

Fig. 1. Experimental setup. (a) Position of the capacitance microphone (diameter = 5.8 mm, height = 2.1 mm) with the air chamber unit and stimulation electrodes. (b) The capacitance microphone was fixed to the top of the air chamber unit. The total weight of the microphone and the air chamber unit was 0.78 g.