Introducing intraoperative direct measurement of muscle force and myofascial force transmission in tendon transfer for cerebral palsy
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CHAPTER 3

Intraoperative Measurement of Length-force Relationship of Human Forearm Muscle

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Abstract

The specific relationship between force and length is one of the most important characteristics of vertebrate muscle. The only accurate method to measure the length-force characteristics is to generate a set of isometric force-time plots at different muscle lengths. In humans, such length-force characteristics mostly are based on indirect measurements that have their limitations. A method of direct, in-vivo measurement of length-force characteristics of the human flexor carpi ulnaris muscle using relatively simple equipment during transposition surgery is presented. The method is proven reproducible, with an overall estimated error of 2.8 %.

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Introduction

The relationship between force and length is one of the most characteristic features of vertebrate muscle function. It reflects the potential exertion of force at every muscle length and determines the available active range of motion (ROM) of the joints passed by the muscle. The only accurate way to establish this length-force relationship is to generate a set of isometric force-time plots at different muscle lengths. Ideally, this is accomplished by in-vivo measurement of the force of the maximally activated muscle at a series of discrete, isometric muscle lengths. The length-force relationship of several muscles in numerous animal muscles has been determined in this way. The length-force relationship of human muscle has seldom been directly measured in-vivo because direct access to active human muscle is limited. To avoid the problem of limited access several indirect methods have been proposed to estimate the length-force relationship in human muscle of the upper extremity, all of which have their limitations. As such, human cadaver studies, cannot provide measurements on
active muscle function. Second, extrapolation of animal data to the human is not accurate because animal muscles may behave different from human muscles. Finally, the application of extracorporeal in-vivo measurements on human muscle is limited because only the net moment, rather than the actual force that a muscle exerts at a joint, can be assessed. This moment not only depends on the force production, but also on the moment arm. Because the moment arm varies among subjects and is difficult to assess accurately, estimating actual muscle force has limited accuracy. In addition, the moment exerted at a joint represents the combined moment of many muscles and passive structures.

Given the limitations of these indirect methods, there is need for direct measurement of force exerted in the human muscle. Even though direct measurements have been done for a few human muscles, the reproducibility of these measurements has never been reported. Although one group of researchers controlled for corporeal movement during contraction, the applied methods did not control for movement artifacts of the arm. Such movement artifacts during measurement influence the length of the muscle, and, therefore, interfere with an accurate estimation of muscle length during contraction. The aims of the current study are to introduce an accurate method for direct measurement of length-force characteristics of the in situ flexor carpi ulnaris muscle (FCU) using relatively simple equipment during transposition surgery and to quantify the reproducibility of these measurements.

Materials and Methods

Patients

All patients having a transposition of the tendon of the FCU to the extensor carpi radialis brevis muscle were eligible to participate in this study (table 1). During surgery, force measurements were done at the distal tendon of the FCU. The study was approved by the authors' institution's ethical committee and the patients were included in the study after giving informed consent. Excluded were patients with a peroperatively estimated maximal muscle force of greater than 200 N because the maximum level of calibration of the measuring device was reached and because sutures could tear out of the tendon if exposed to such forces. Two male and three female patients with cerebral palsy (mean age, 14 years; range, 10 - 18 years) were included in this study.
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\(l_{\text{forearm}}\) = Forearm length  
\(F_{A_{\text{max}}}\) = Maximal active force  
\(F_{P_{\text{max}}}\) = Maximal passive force  
\(l_{\text{opt}}\) = Optimum length  
\(F_{P_{\text{opt}}}\) = Passive force at optimum length

**Experimental setup**

Surgery was done with the patients under general anesthesia. No muscle relaxants were administered. Measurements were done without a tourniquet. Two gel-filled skin electrodes (Red Dot 2560, 3M Inc, Minneapolis, Minnesota) were placed on the skin over the cubital tunnel of the elbow and connected to a custom-built, constant current peripheral nerve stimulator. For safety, the stimulator was isolated from the electric mains using an isolation transformer. The electrodes were covered with plastic foil to allow for a sterile surgical field. A 15-cm long incision beginning at the pisiform bone was made in proximal direction. A Kirschner wire (K-wire) was drilled into the medial epicondyle of the humerus. The insertion of the most distal muscle fiber on the FCU tendon was marked with a thin suture. Tenotomy was done of the tendon of the FCU at its insertion on the pisiform bone. Subsequently, the FCU was dissected from the surrounding connective tissue in the proximal direction, to halfway up the muscle belly. A 2-mm metal ring was sutured to the cut FCU tendon of the muscle. A custom-built, sterile, stainless steel bar designed to carry a linear strain gauge force transducer was placed perpendicular to the K-wire (figure 1). This fixation of the measurement system onto the patient’s body is a novel feature in human length-force measurement. The strain gauge had a maximal output error of 0.1% and a compliance of 0.0048 mm/N (AMC-Medical Technical Development, Amsterdam, the Netherlands). The stainless steel bar with the strain gauge connected was held
in place by the assistant investigator. The distal end of the bar and the strain gauge were covered with sterile plastic cover. Preoperatively, the strain gauge was connected to a data acquisition system that was isolated from the electric mains by the isolation transformer (2000 gain bridge amplifier, 100 Hz AD converter, Pentium III computer (Intel Corp., Santa Clara, CA), resolution of force 0.05 N; AMC-Medical Technical Development, Amsterdam, the Netherlands) and calibrated with a 200 N weight. The strain gauge was linked to the metal ring on the tendon using a stiff (less than 0.5 mm strain under 200 N stress), sterile, stainless steel chain which passed through the plastic cover. The weight of the chain previously was shown to be of negligible contribution to the force [Smeulders MJC 2001, unpublished].

Figure 1
A schematic drawing of the experimental setup is shown. The sterilized, stainless steel bar holding the strain gauge force transducer is fixed on the patient’s arm at a K-wire that is drilled in the medial epicondyle of the humerus. A chain connects the tendon with the force transducer. The bar and the chain are pierced through a sterile plastic cover. Inside this cover, the measuring system is held by an assistant so that the chain is aligned. A computer is used to control parameters of electrical stimulation of the ulnar nerve and data collection of the force signal.
Measurements

The force transducer was moved up the metal bar and fixed when the chain was in line with the muscle. The initial length (li) of the muscle belly was measured as the distance in centimeters between the K-wire and the marker-suture on the distal tendon. Subsequently, the force transducer was positioned 2 cm farther up the bar, stretching the muscle to a length of li +2 cm to pre-stretch the viscoelastic components of the muscle. One electrical pulse was delivered to the ulnar nerve 300 ms before every train of pulses to straighten the chain and allow the muscle to shorten to the desired length at low muscle lengths. Subsequently, trains of biphasic, intermittent pulses of 125 mA amplitude at a stimulus frequency of 50 Hz and a pulse duration of 0.1 ms were delivered for 1000 ms. A pilot study showed that this evoked full recruitment yielding an isometric, tetanic contraction of the FCU muscle, the medial part of the deep flexor muscles of the fingers and the intrinsic muscles of the hand innervated by the ulnar nerve [Smeulders MJC 2001, unpublished]. In all patients, such stimulation could be done without the risk of burn marks or other discomfort. After the tetanic contraction, the muscle was allowed to recover for 2 minutes at low length. This procedure was repeated three times at the particular length of li +2 cm to tighten the sutures of the metal ring and to precondition the muscle's internal connective tissue. Subsequently, the force transducer was moved 3.5 cm in a proximal direction along the bar, releasing the chain and the next contraction allowed the muscle to shorten to a new length of li -1.5 cm. Force measurements were repeated at this new length. After that, the muscle was stretched by 0.5-cm increments by moving the force transducer on the metal bar. At each new length the procedure of stimulation and rest was repeated until the active force decreased rather than increased, followed by two more length increments. In this way a series of force measurements were collected at lengths of li -1.5 cm, li -1 cm, li -0.5 cm, li, li +0.5 cm, li +1 cm, li +1.5 cm, through li +n cm.

To test for reproducibility, the series of force measurements at lengths of li -1.5 cm through li +n cm was repeated twice after 2 minutes. Peroperative setup of the experimental devices and the repeated acquisition of length-force data took approximately 50 minutes.

Data Analysis

Two representative data points from each force-time profile were identified for use in construction of the length-force relationship: one before stimulation representing the passive (elastic) muscle force, and one at the tetanic plateau representing the total muscle force.

Additional treatment and averaging was done as previously described. In short, the force-passive length curve was constructed by plotting the force data obtained
before stimulation against muscle length. The total length-force curve was constructed by plotting the maximum of the force during stimulation against muscle length. Passive length-force data were least square fitted using an exponential function and active muscle force was calculated by subtracting the measured passive force from total force during muscle activity. Active length-force data were assigned a polynomial function. The degree of the polynomial function that most adequately described a particular set of length-force data was selected using an analysis of variance (ANOVA). Optimal FCU force was defined as the maximum of the selected polynomial, and the corresponding FCU length as optimum length ($l_{opt}$). From each fitted polynomial function, six data points were selected at length intervals of 5 mm for averaging. Because the absolute length at which the measurements had started and the absolute measured range of lengths differed in every patient, the length relative to $l_{opt}$ was calculated to allow for comparison among patients. Averages and standard errors of the length-force data of all patients were calculated.

To test for reproducibility, a two-way ANOVA for repeated measures was done on the normalized force-relative length data. The factors were the repetition of measurements and muscle length. Reproducibility of the testing procedure was obtained by calculating the expected standard deviation from the residual variance. From this, a 95% confidence interval of the expected error of reproduced measurement was calculated.

**Results**

*Length-force Relationship*

Full recruitment of the FCU at different lengths induced tetanic contractions, each with a specific plateau force (figure 2). The fluctuations in the curves were small. Maximum values of the plateau of the force-time profiles plotted against the muscle length yielded the typical parabolic shape of the force-active length curve (figure 3). Expressing force as a percentage of optimal force and length as a difference from $l_{opt}$, there still was considerable variability among subjects, as could be deduced from the standard deviation (figure 4). Considerable variation was also found for the passive force at the highest length studied and at optimal length (table 1).
Figure 2
Typical example FCU force-time profiles at different lengths are shown for Patient 2. The pre-tetanic twitch and the tetanic contraction can be seen. The maximum force and the shape of the profile is highly muscle-length dependent.

Figure 3
Three repeated measurements of active and passive force-length curves of the flexor carpi ulnaris muscle are shown for patient 5. The active profile is parabolically shaped and the passive profile is exponentially shaped. Repeated measurements did not significantly differ.
Reproducibility

Repeated measurement of length-force curves within one patient was comparable (figure 3). The ANOVA showed the length-force curves of repeated measurement not to be significantly different ($p = 0.117$). The residual variance was 10.82, which implies the expected standard deviation within a subject for a repetitive measurement of force at any specific length to be 3.2% 1. The 95% confidence interval for the expected error in reproduced measurements was calculated to be 2.8%. Because of the high level of reproducibility of the authors’ method the high variance in length-force curves (figure 4) cannot be caused by measurement artifacts, but have to be ascribed to individual variation among patients.

Discussion

Ralston et al. 96 were the first to report actual measurement of length-force profiles of human muscle in amputees from the second World War and they showed that human muscle had similar characteristics to frog muscle. In their study, they controlled for large corporal movement during measurement by placing an extra force-transducer between the arm and the measuring system. However, small arm movements and rotational movement of the arm could not be controlled for with their system 96. Freehafer et al. 24 and Lacey et al. 52 were the only ones, so far, to intraoperatively measure length-force profiles of human lower arm muscles.
The experimental setup that was used for their studies was limited in that movement artifacts of the arm were not controlled for by attaching the measurement system to the patient's arm. In isometric testing, any movement of the arm or body during measurements may profoundly affect the measured forces because movement of the arm will lead to a change of muscle length and allow force to be exerted onto the force transducer from sources other than the target muscle. That the length-force profiles observed by Freehafer et al.\textsuperscript{24}, Lacey et al.\textsuperscript{62}, and Ralston et al.\textsuperscript{96} were not as smooth as those found in the current study may have been caused by movement artifacts. Because of the construction of our measuring device and the attachment of the metal bar directly on the patients arm in the line of pull of the FCU, unwanted corporal and arm movements did not interfere as the measuring device moved along with the arm. Although some swaying of the force transducer was inevitable, this was kept within 1 cm. Movement of the force transducer of 1 cm was calculated to maximally result in a 0.02 cm length change in the muscle, so the movement artifacts in this study were accepted to be negligible. Although fixing the measurement system on the upper arm did not prevent possible elbow rotation, no elbow rotation was observed during contraction, as the hand was held in a fixed position by the surgeon.

Another explanation for the difference in smoothness of the tetanic contraction between the current study and those reported earlier may be the difference in stimulation methods. Although simultaneous maximal recruitment of all muscle fibers can be achieved in voluntary contraction in healthy subjects\textsuperscript{86}, such contraction depends on factors such as motivation and training status\textsuperscript{55}. For reliable muscle testing in patients with neuromuscular disease such as cerebral palsy, the lack of motor control demands external electrical stimulation\textsuperscript{14}. The current authors therefore stimulated the innervating nerve, whereas Ralston et al.\textsuperscript{96} used voluntary contraction. Freehafer et al.\textsuperscript{24} and Lacey et al.\textsuperscript{62} used needle electrodes that were placed in the muscle. This latter method may be more susceptible to irregularities of recruitment because of shifting of the electrodes during contraction.

During transposition surgery in patients with paralysis and cerebral palsy, knowledge on length-force characteristics is particularly important because the general goal of transposition surgery is to maximize strength over the largest operating range that is required of the muscle in its new role\textsuperscript{12, 24, 26, 75}. The patients in the current study suffered a spastic disorder of different severity. The individual differences in the active and passive force curves may have been caused by the difference of spasticity, since spastic muscle may differ from normally innervated muscle. Given the large standard deviation around the mean active force profiles,
even after normalization for maximum force (figure 4), an individual approach for each patient is indicated. The surgeon’s subjective estimate of this passive tension is the only clinical guide to quantify muscle length-force characteristics. On the basis of the current study, however, we challenge the accuracy of such estimates. The wide variation of passive length-force characteristics at lopt in the current five patients shows the limitation of estimation of active length-force characteristics based exclusively on passive tension during stretch. Although the current setup involves the use of a data-acquisition system and may thus as yet be not very practical in a clinical setting, the active length-force curve informs the surgeon about the most important properties of the donor muscle and is therefore a better guide for quantifying its function. With the method presented, it is possible to make reproducible intraoperative length-force measurements that may be used to optimize tendon transfer surgery.