# A Predictive Model of Muscle Forces for Children with Cerebral Palsy.

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## Abstract

***This study represents preliminary work in testing our mathematical model's ability to predict isometric forces for children with spastic diplegic cerebral palsy (N=3, mean age 10.4 ± 3.0 years). The model accounted for 87% and 91% of the variance in the shape of experimental forces produced by stimulation trains with mean frequencies ranging from 12.5 to 100 pps from the quadriceps femoris and triceps surae muscles, respectively. The ability of the model to predict forces suggests its use for the identification of optimal activation patterns during functional electrical stimulation and as a tool for investigating physiologic properties of CP muscle.***

## 1. INTRODUCTION

A ubiquitous impairment in cerebral palsy (CP) is muscle weakness for which several mechanisms are postulated. These mechanisms include impaired muscle activation [10], abnormal antagonist co-activation [4, 5], and differences in muscle morphology [6, 8].

Approaches to improve muscle strength and function include strength training using voluntary effort [1] and neuromuscular electrical stimulation (NMES) [11]. Additional approaches include functional electrical stimulation (FES), in which NMES is used to produce and assist muscle contractions during functional activities [7]. Corroborating evidence of different muscle morphology in CP [6,8], recent work suggests that CP muscle may have different physiological properties when compared to typical muscle [10]. Additional work suggests that manipulation of stimulus activation frequency and pattern affects the rate and amount of fatigue and such manipulations can improve FES performance [9]. We have developed a physiologically based mathematical model that can predict muscle force responses to a wide range of stimulation frequencies and patterns for able bodied adults [3] and for children with spinal cord injuries (SCI) [2]. The purpose of the present investigation is to determine the applicability of this model for children with spastic diplegic CP and to gain additional insight on physiologic properties of force responses for this population.

## 2. METHODS

### 2.1. Isometric force model.

Our force model consists of two differential equations. The first equation represents the activation of the muscle by modeling the dynamics of the rate-limiting step leading to the formation of the $Ca^{2+}$-troponin complex. This equation incorporates a factor to account for nonlinear summation of the $Ca^{2+}$ transient when stimulated with two closely spaced pulses [3]. The second equation represents development of mechanical force and was derived from a linear spring, damper and motor in series and is driven by force producing cross-bridges and is mediated by a Michaelis-Menten term [3].

### 2.2. Experimental data.

Three children with spastic diplegic CP (2 male, mean age 10.4 ± 3.0 years) participated. Legal guardians and participants signed informed consent and assent forms, respectively. All testing was performed on a computer-controlled dynamometer (Kin-Com II, Chattecx Corp, Chattanooga, TN) and participants were positioned for isometric testing of their right quadriceps femoris (sitting, knees flexed to 60°) in the morning and their right triceps surae (supine, ankle flexed to neutral) in the afternoon.

Self-adhesive electrodes covered the width of the muscle being tested (Axelgaard, Fallbrook, CA; ConMed Corp., Utica, NY). Two hours before testing, an anesthetic cream (EMLA; 2.5% prilocaine, 2.5 % lidocaine; Astra-Zeneca, Wilmington, DE) was applied to the skin over the electrode area and covered with an occlusive dressing to reduce discomfort during electrical stimulation and to reduce reflex responses. After ~2 hours, the cream was

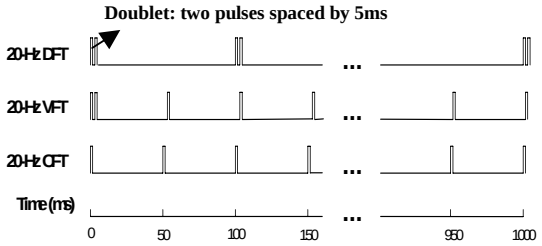


*Fig.1 CFT: constant-frequency train with all pulses within the train separated by constant intervals; VFT: variable-frequency train with a doublet (two pulses separated by 5ms) inserted in the beginning of a CFT; DFT: doublet-frequency trains with doublets separated by longer, constant intervals.*

removed and the skin cleansed with alcohol prior to electrode placement.

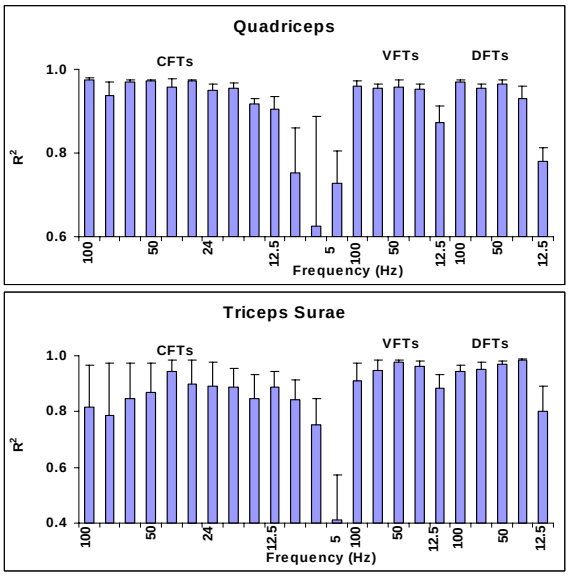


*Fig.2 Correlation coefficients between the predicted and measured forces averaged over three subjects, for each testing train*

Once positioned on the dynamometer, the quadriceps or triceps surae muscles were stimulated with 600 μs square wave pulses using a Grass Instruments S88 stimulator equipped with a SIU8T stimulus isolation unit (West Warwick, RI). First, the maximum electrically elicited force was determined using a 13-pulse, 100 pps electrical train. We termed this stimulus the maximum burst. The stimulus intensity was then lowered and adjusted to potentiate the muscle and to elicit a force equivalent to 40% of the maximum burst force using a 13-pulse, 40 pps train. For subsequent testing, the stimulator was driven by a personal computer using custom-written software (LabView 4.0.1, Austin, TX) to control the timing of each stimulation train. All force data was sampled at 200 Hz and analyzed with custom software.

Within 10 s after potentiation, the testing sequence commenced using 23 testing trains each delivered once every 10 s to prevent fatigue. With the exception of the twitch (1 pps), trains consisted of either 50 pulses or had train-durations of ~1-sec, whichever produced the shorter train duration. The trains covered a wide range of frequencies (1 to 100 pps) and 3 pulse patterns (Fig. 1).

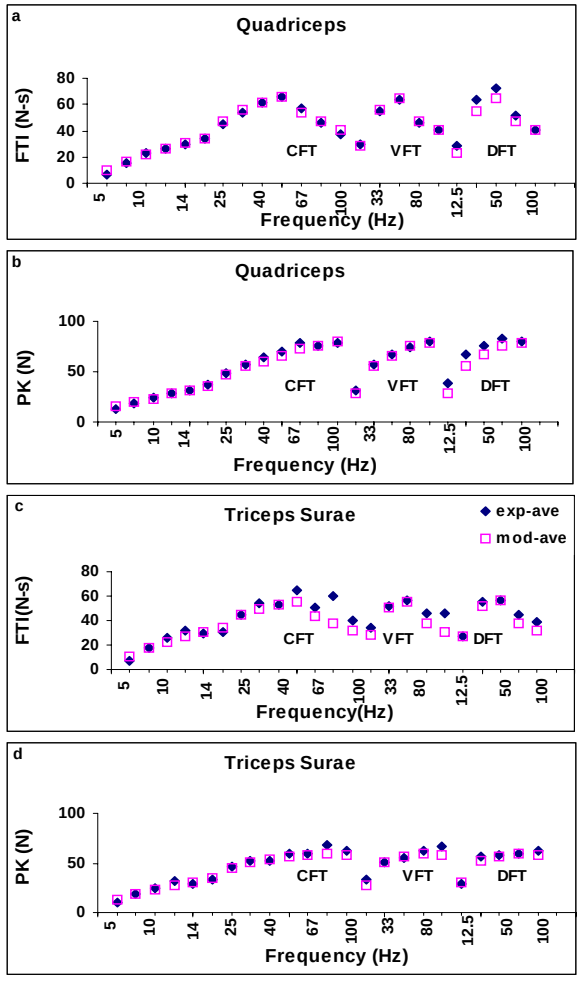


*Fig.3.Group (n=3) force-time integrals (a, c) and peak forces (b, d) calculated for experimental and predicted forces for the quadriceps and triceps surae muscles, respectively.*

## 2.3. Simulation

For each subject, force responses to the 50Hz-CFT and the 12.5Hz-DFT were used to calculate the model parameter values. Then the parameterized model was used to predict force responses to each of the other testing trains.

## 3. RESULTS

The ability of model to emulate the shape of the experimental force record in response to each stimulus train was assessed by comparing the predicted to the experimental force for each 5 ms interval of the force record. Correlation coefficients were calculated between the predicted and experimental forces for each testing train for all three subjects (Fig. 2). For both quadriceps and triceps surae muscles, the $R^2$ values are above 0.9 for most testing trains except the low-frequency CFTs suggesting that the model predicted well the shape of the force trace for most trains tested. The poor predictive ability for low-frequency CFTs is probably due to the very low forces that these trains produced. Force-time integral (FTI) and peak force (PK) were also calculated for each testing train and the experimental and predicted data compared by calculating the percent difference between them. On average, the model predicted well the FTI and PK for both muscle groups in response to all three stimulation patterns. Histograms of the percentage error in FTI reveal that 61% and 65% of the predicted responses were within ± 10% of the experimental responses for the quadriceps and triceps surae, respectively. For PK, 65% and 81% of the predicted responses were within ± 10%, respectively. As with the group data, the model was able to predict the stimulus frequency for each pattern that produced the greatest FTI and PK (Fig. 3).

## 4. DISCUSSION AND CONCLUSIONS

The present model was developed from quadriceps muscles of healthy subjects. These results demonstrate that the model can be used for children with CP. By fitting force responses to just two stimulation trains, the model was able to predict forces to different test trains for both quadriceps and triceps surae muscles.

For healthy adults [3] and children with SCI [2], greater than 80% of predicted FTI and PK values for different testing trains differed by ± 10% of experimental values. Participants with CP had greater percentage differences in part due to involuntary responses and difficulties to relax throughout data collection.

Parameter values for $Ca^{2+}$ sensitivity and time constants for $Ca^{2+}$ handling were lower and greater, respectfully, for the triceps surae than the quadriceps muscles. This is consistent with slower contraction times observed for the triceps surae suggesting that it is slower than the quadriceps [10]. This model has potential for optimizing stimulation patterns for FES and investigating physiologic properties of muscle in CP.

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