CHARACTERIZATION OF THE *FLEXOR DIGITORUM SUPERFICIALIS* AS A PREDICTOR OF GRASPING STRENGTH

by

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ABSTRACT OF THE THESIS

Characterization of the *Flexor Digitorum Superficialis* as a Predictor of Grasping Strength

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The activity of forearm musculature during grasp has been studied by a variety of testing modalities in order to assess muscular conditions, predict grasping strength and control powered prosthetics. Myotonometry has been shown to correlate directly to changes muscle stiffness and tone for the assessment of muscle recovery but has no real time application and is limited to rehabilitative and physical therapy application. Of all testing modes Force Myography (FMG) is the only one that has demonstrated simultaneous multifunctional and multi degree of freedom control of prosthetic hands with sensors placed over the surface of the skin. These studies, however, were not able to identify magnitudes of individual muscle activation for control of volitional movement. In the present study a new testing modality is introduced, targeted Force Myography (tFMG). tFMG detects the change in radial pressure and stiffness of a single muscle during contraction through the voltage response of a single force sensing resistor (FSR) strapped above the muscle body intrinsically combined with the overall change in radial pressure of the harnessed body segment. In this study tFMG of the *flexor digitorum superficialis* is assessed during power grasp. The results correlate well to myotonometry, $r^2 = 0.86$, and show the ability to predict a subject's level of muscle activation, $r^2 = 0.94 \pm .03$. This

was achieved through the development of inexpensive testing fixtures and methods in order to linearize and calibrate the voltage response curves of FSRs, making them an accurate and reliable tool in the assessment of grip force and forearm pressure. It is demonstrated here that tFMG results have the ability record the changing stiffness of the underlying muscle as well as predict grasping strength.

Dedication

This thesis is dedicated to my father. Without his guidance, friendship, and humor I would not be the man I am today. He was more than a father to me, he was a best friend. I thank him and will always love him.

List of Illustrations

Section 1. INTRODUCTION

Isometric grasp is the one of the most common activities of daily living (ADL), and impairment or loss of this ability is devastating. Although there are many forms of grasp, used for various functions, the power grip has been extensively studied. Power grip forces are exerted by the proximal and intermediate phalanges on an object whose axis is perpendicular to the forearm (Freivalds, 1996). In a power grasp, while mechanically controlled by the forearm flexor muscles, the wrist is stabilized by extensor forearm muscles to counteract the wrist flexion torque caused by the finger flexor muscles (Hagg and Milerad, 1997). Thus power grasp, by its nature, sacrifices precision control (Hoozemans and van Dieen, 2005). The behavior of these muscles during isometric contraction has been used for the prediction and monitoring of grip force in both healthy and affected populations, including amputees (Armstrong et al., 1979; Hoozemans and van Dieen, 2005; Hou et al., 2007; Loscher and Gallasch, 1993; Mogk and Keir, 2006; Wininger et al., 2008; Zheng et al., 2006). These, along with other studies, have led to the development of new prostheses and rehabilitation techniques.

In studies of forearm musculature Electromyography (EMG) has been the primary modality, yielding both linear (Armstrong et al., 1979; Duque et al., 1995) and nonlinear (Hou et al., 2007; Woods and Bigland-Ritchie, 1983) relationships with a percent of a subject’s maximum voluntary contraction (% MVC). While EMG has been used extensively for the prediction of muscle activity and control of a prosthetic device (Duque et al., 1995; Farina et al., 2004; Gordon and Ferris, 2004; Hoozemans and van Dieen, 2005; Keir and Mogk, 2005; Mogk and Keir, 2006; Rainoldi et al., 2004), there are several concerns: low frequency noise in the processed signal, changes in skin impendence due to sweat, changes in signal range and frequency characteristics due to movement of electrodes, fatigue and susceptibility to environmental issues (Kenny et al., 2005). Regardless, many studies have demonstrated the potential of EMG models to predict isometric grip force with advanced signal processing, but the issues remain.

Other methods have been developed in order to utilize muscles' natural physiological low-pass filtering of the EMG signal. Dimensional changes detected by hall sensors (Kenny et al., 2005), Mechanomyograms and axial displacement of muscle tissue (Fukunaga et al., 1996; Orizio et al., 1999), changes in interstitial muscle pressure (Mubarak et al., 1976), tendon movement detected with pneumatic sensors on the skin surface (Abboudi et al., 1999; Curcie, 2001), and radial change in limb pressure (FMG) (Wininger et al., 2008) have all been used to predict gripping strength.

Of all the above neuromuscular imaging modalities, force myography (FMG) is the only one that has demonstrated simultaneous multi-digit and multi-degree of freedom control of prosthetic hands with an interface placed over top of the skin. In their studies, Curcie (2001) and Abboudi (1999) were able to identify individual muscle volitions for the control of a prosthetic device by placing force sensing pneumatic sensors in a socket and placing it over the residual limbs of amputees.

Unresolved in the studies on prosthetic control are whether the sensors were able to identify magnitudes of muscle activation. An FMG study of the mechanical force generated by the volumetric change in the underlying muscle complex of the forearm during power grasp has shown a high linear correlation to % MVC (Wininger et al., 2008). This study of unimpaired subjects showed a high correlation between change in overall radial pressure of the entire forearm and grasping strength, it did not address whether an individual muscle could be targeted to estimate grasping strength.

Hypothesis 1

In this study, a new testing modality is created to observe the behavior of an individual muscle during power grasp. Targeted Force Myography (tFMG) is a modification of FMG, whereby a FSR, placed over a specific, targeted muscle will measure the radial pressure and stiffness of the underlying muscle combined with the overall change in radial pressure of the entire limb muscle complex. It is hypothesized that the tFMG recordings of a single FSR strapped to the forearm and overlying the *flexor digitorum superficialis* (FDS) will correlate linearly with the exerted grasping force. The expansion of the muscle belly radius during shortening or contraction alters the pressure applied to the FSR. The piezoresistive sensor reduces resistance under mechanical pressure, the increase in voltage reports change in muscle activity. The FDS was chosen as the muscle of interest due to previous research focusing on the muscle (Abboudi et al., 1999; Duque et al., 1995; Hagg and Milerad, 1997; Heath, 2003; Kenny et al., 2005; Long et al., 1970; Loscher and Gallasch, 1993; Mogk and Keir, 2003). Long et al. (1970) report that the FDS contribution during power grip is direct proportional to the resulting force.

Section 2. MYOTONOMETRY

A. INTRODUCTION

Technology

A myotonometer is a computerized device that non-invasively measures muscle tone based on the force necessary to displace muscle. A myotonometer is typically used to quantify muscle stiffness (tone) (Aarrestad et al., 2004), but has also been shown to correlate to muscle spasticity, compliance (Leonard et al., 2001), as well as strength (Leonard et al., 2004). Muscle stiffness is the relationship between the change in muscle length and active force being applied by the muscle or the amount of force necessary to cause tissue displacement. It has also been show to be proportional to the force exerted perpendicularly to the muscle (Walsh, 1992). Muscle stiffness measurements obtained during muscle contraction provide an indirect but valid measure of muscle strength (Bizzini and Mannion, 2003; Leonard et al., 2004).

A study performed by Aarrestad et al. (2004) in the validation of Neurogenic Technologies, Inc.'s Myotonometer proved that inexperienced users could easily and accurately perform measurements with the device, showing excellent inter- and intra-rater reliabilities when assessing spasticity under isometric contraction, correlating directly to the Ashworth scale.

The operation of the Myotonometer device is very simple and involves running the Myotonometer software and depressing a target area of body tissue with the Myotonometer wand (Figure 1), perpendicular to its surface until a force of 2 kg is reached. The speed of the applied pressure is not controlled. During the operation, software records the displacement of the tissue at .25 kg force increments until the 2 kg force is reached and generates a curve with this data. In a study by Leonard et al (2001), it was shown that the area under the curve (AUC), generated by the force/displacement curve relates directly to the change in activation level of the muscle and correlates well to Electromyography (EMG) readings.

Clinical Utility

After a stroke, brain injury, or conditions such as cerebral palsy or multiple sclerosis, spasticity is not an uncommon occurrence (Pandyan et al., 2005). Measuring the degree of spasticity has proven unreliable by many current methods. The Ashworth scale, which originated in 1964, measures the resistance of muscle to passive motion and is the most common clinical method of assessing spasticity. The two major limitations of using the Ashworth scale is the subjectivity of the examiner and the Ashworth scale is only able to identify cases of severe spasticity, it has been shown that spasticity must reach a certain level of severity before it will impair function (Damiano et al., 2002).

Myotonometry has been used when assessing upper and lower motor neuron injury such as the location of injury as well as recovery of ACL surgery (Chen et al., 2001). Myotonometry can also measure the severity in which the normal dampening of tone is decreased (hypertonicity and spasticity) in stroke and cerebral palsy patients and track the improvement of muscle tone in people recovering from musculoskeletal diseases or sports injuries. In healthy subjects, myotonometry can be used to determine muscle stiffness and tone. Myotonometry is superior to other muscle tone measuring modalities in that it can measure the tone when the muscle is passive as well as contracted, whereas most methods can only measure muscle tone during one or the other.

Hypothesis 2

tFMG, which measures similar muscle behavior, has not been proven to be directly correlated to the assessment of any chronic muscle conditions. A comparative study of tFMG and myotonometry has a variety of potential benefits. The first section of this study will validate myotonometry and tFMG of the FDS as a means for assessing muscle stiffness and tone during power grasp. The hypothesis is that tFMG will correlate with myotonometry.

B. MATERIALS

Since no handheld force gauge was available at the time to assess grasping strength an octagonal custom grip force dynamometer (GFD) (4 cm diameter × 10 cm long) was used (Wininger et al., 2008). The polyvinyl chloride GFD contained of four equally sized FSR strips (model #408, Interlink Electronics Inc., CA) that were affixed to four of the faces of the GFD and ran the length of the GFD. The shape and orientation of the sensors were designed to optimize the contact area between the sensors and the fingers, palm, and thumb. The FSRs were set up in parallel and the output voltage was obtained by a voltage divider circuit with a single 5 kΩ resistor. The dynamic behavior of the FSR was rectified by a third order polynomial in order to linearize the force/voltage response curve.

A single 1.4 cm FSR (model #402, Interlink Electronics; Carpenteria, California) was used to record the mechanical activity of the FDS by means of tFMG. This FSR was incorporated into a voltage divider circuit with a 5 kΩ fixed resistor and placed on the surface of the skin above the FDS muscle body. A cloth strap with a Velcro fastener (McMaster-Carr, Part #: 3955T77) was used to secure the FSR in place.

Sensor signals were acquired with 12-bit accuracy by an external data acquisition board (NI-DAQUSB-6008, National Instruments, Corp; Austin, Texas) and routed to LabVIEW (National Instruments, Corp), which provided the signal processing as well as real-time visual feedback for target tracking. Overall tFMG response was calculated as mean square deviation from the sensor's baseline value and stored using LabVIEW.

The Myotonometer® (Figure 1) and Myotonometer Software were developed by Neurogenic Technologies®, Inc (Missoula, MT). Though there have been other myotonometers developed throughout the literature (Kato et al., 2004; Steinberg, 2005; Veldi et al., 2002), the model used for this testing is the same model of myotonometer as used in the studies being referenced (Aarrestad et al., 2004; Leonard et al., 2004; Leonard et al., 2001).

Figure 1. The Myotonometer developed by Neurogenic Technologies

C. METHODS

Human Subjects

8 healthy subjects were chosen (4 men, 4 women, aged 20-29) with no history of upper arm affliction. All subjects provided informed written consent, and the study was approved by the Rutgers IRB. With the subjects sitting in a comfortable position their right arm was suspended with supports at the elbow and wrist to allow for the most unrestricted behavior of the forearm musculature, the elbow flexion was approximately 20-50 degree flexion and did not change during testing.

Protocol

The thickest area of the FDS muscle was located on each subject based on anatomical landmarks and palpation. The exact site was chosen by the operator by locating the FDS with one or more palpating fingers while the subject is in a relaxed state then asking the subject to create a clenched fist with maximum force and determining where on the muscle body the maximum discernable displacement occurred. This area was then circled and used as the site for placement of both the Myotonometer and tFMG sensor.

Prior to testing, the displayed grip force was calibrated and scaled for baseline and maximum force by following the protocol in Wininger et al (2008). Subjects grasped the GFD by wrapping their fingers and thumb around it with their wrist in neutral position for 5 seconds, merely applying enough force to support the GFD. They were then instructed to maximally grasp the GFD for 5 seconds to establish their maximum voluntary contraction (MVC) force. The average of the recorded data during these two windows was used to scale the visual feedback tank display from 0 to 100 percent MVC. During tFMG testing, the tFMG calibration occurred simultaneously with the gripper calibration; no visual feedback was provided for the tFMG during calibration or testing.

The subjects were given time to practice achieving targeted % MVC grasping levels until a sufficiency was demonstrated in their ability to achieve and maintain target force levels. During testing with The Myotonometer, the subjects were instructed to grasp the GFD levels of 0%, 25%, 50%, 75%, and 100% MVC, and maintain this level for ~20 seconds while myotonometry readings were taken (figure 2). This procedure was repeated 3 times for each subject with a 30 second rest period between force level holds and 2 minutes between runs. At each force level the Myotonometer was pressed into the target tissue area 8 times with an option to redo any trial. An average plot as well as the calculated AUC was then returned and saved by the Myotonometer software.

Figure 2. Myotonometer testing on the Flexor Digitorum Superficialis muscle

During tFMG testing, the cuff was donned on the arm (Figure 3) with a baseline pressure of < 5% full-scale output of the sensors, the baseline was subtracted before each test. Again subjects were instructed to grasp to levels of 0%, 25%, 50%, 75%, and 100% MVC, this time however only ~5 seconds of maintained activity was necessary to get an average reading of tFMG activity. In this case, a small rest period was given between runs. A 5 min resting period was taken between myotonometry and tFMG testing.

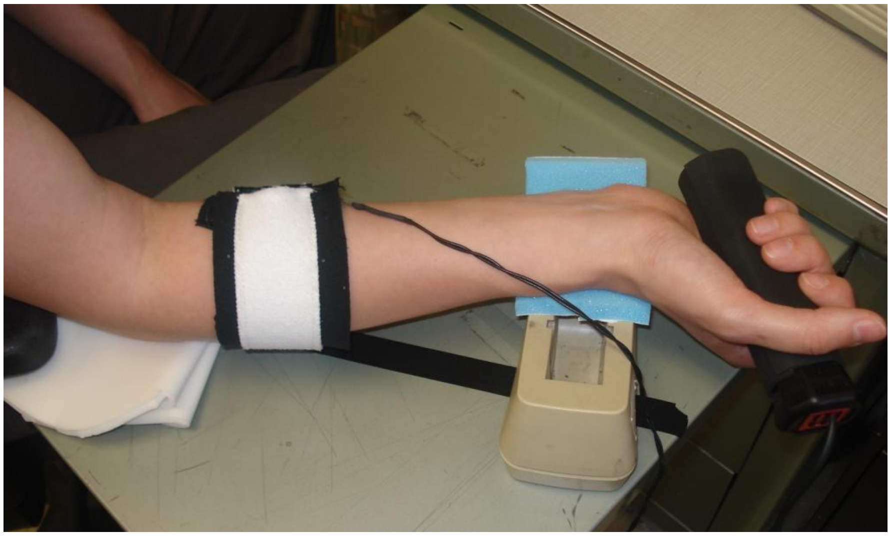


Figure 3. tFMG testing on the Flexor Digitorum Superficialis

D. RESULTS

Myotonometer

The individual plots from the Myotonometer software demonstrate the action of The Myotonometer during depression at various levels of muscle activity (Figure 4). As the muscle contractile force increases, the underlying muscle becomes stiffer and The Myotonometer head cannot depress the tissue as easily. The area under curve (AUC) (to the left as depicted below) is in effect the integrated value of the curve, the force applied by The Myotonometer times the displacement of the tissue or kilograms times millimeters. This effectively represents myotonometer work. The values returned by the Myotonometer software for AUC were averaged over the three runs for each of the % MVC levels. As expected, the averaged AUC values show a good correlation, $r^2 = 0.86 \pm .14$, with the targeted % MVC levels (Figure 5). These average values were then normalized for each subject by baselining the lowest AUC value and dividing by the maximum value. These results showed a good correlation to the targeted % MVC levels (Figure 6). The root mean square values of the subjects' plots had a high of .99 and a low of .64, the overall r^2 value of all the plots was 0.81 with a slope of -0.89. These are good results but cannot be considered a reliable method of assessing relative muscle activity.

Myotonometer Output Graphs

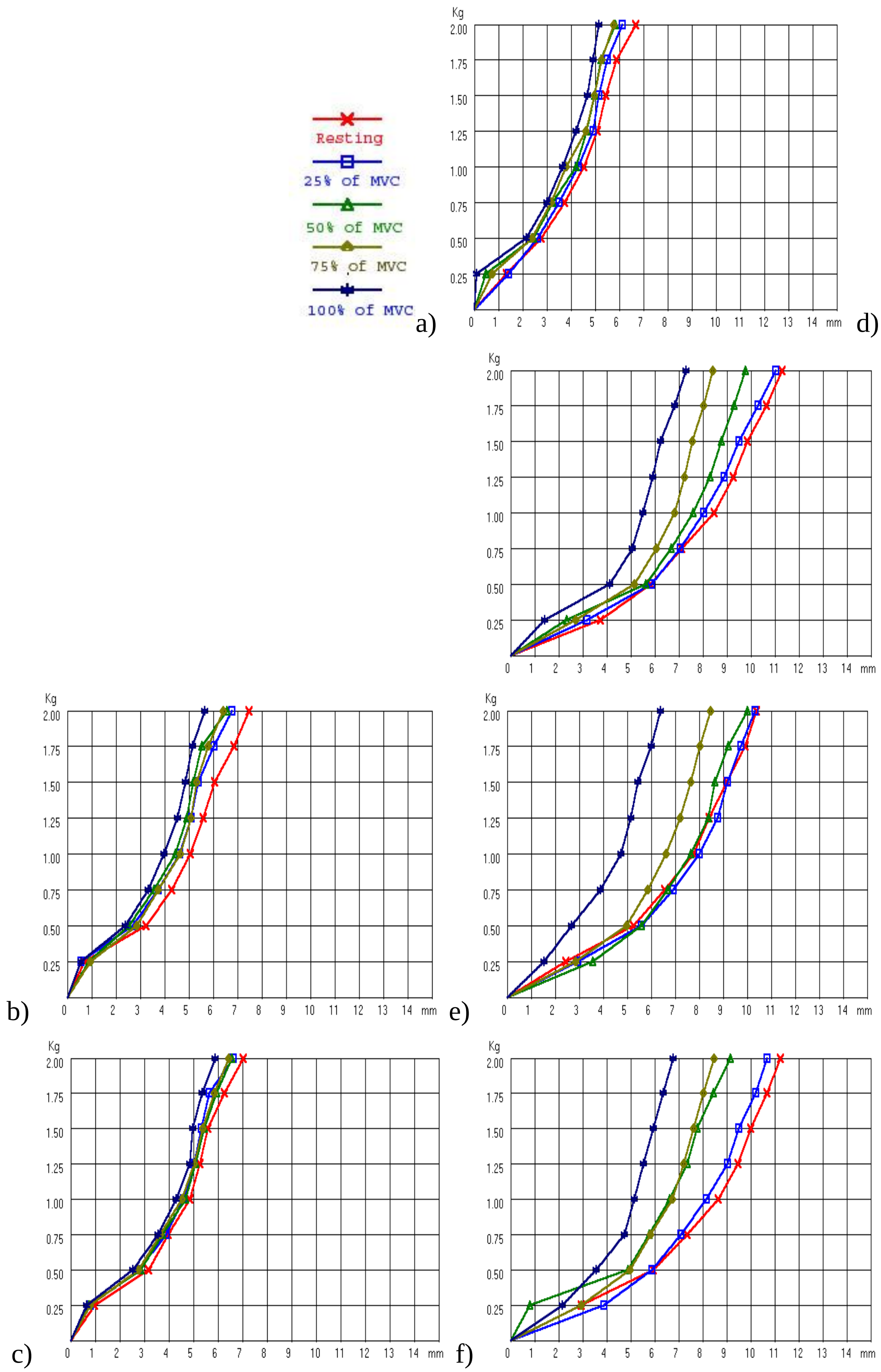


Figure 4a-e. The graphical output by the Myotonometer software of three runs by two subjects: a-c and d-e. The results show the force, in Kg, required to indent the muscle a given depth, in mm. Under a greater contractile force there will be less tissue displaced at a 2kg force. In some instances, d and e for example, the area under (to the left of) the curve even decreased proportionally as the level of muscle activity increased.

Because this is not always the case, as demonstrated, the AUC results for each run is averaged for each subject to reduce the variability.

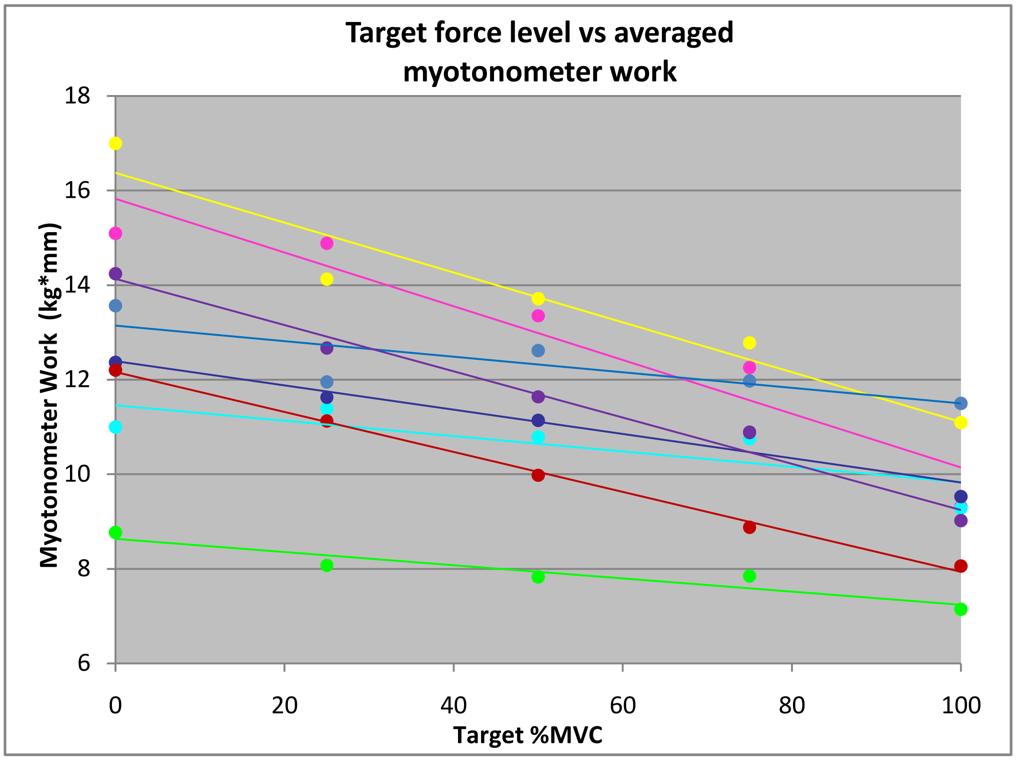


Figure 5. Average AUC for 8 subjects grasping to selected target % MVCs. The change in Myotonometer work vs. grip force activation levels shows a high degree of correlation to subjects' individual fits, but with varying slopes. This confirms that The Myotonometer can be used to assess varying degrees of tone in the muscle of individuals.

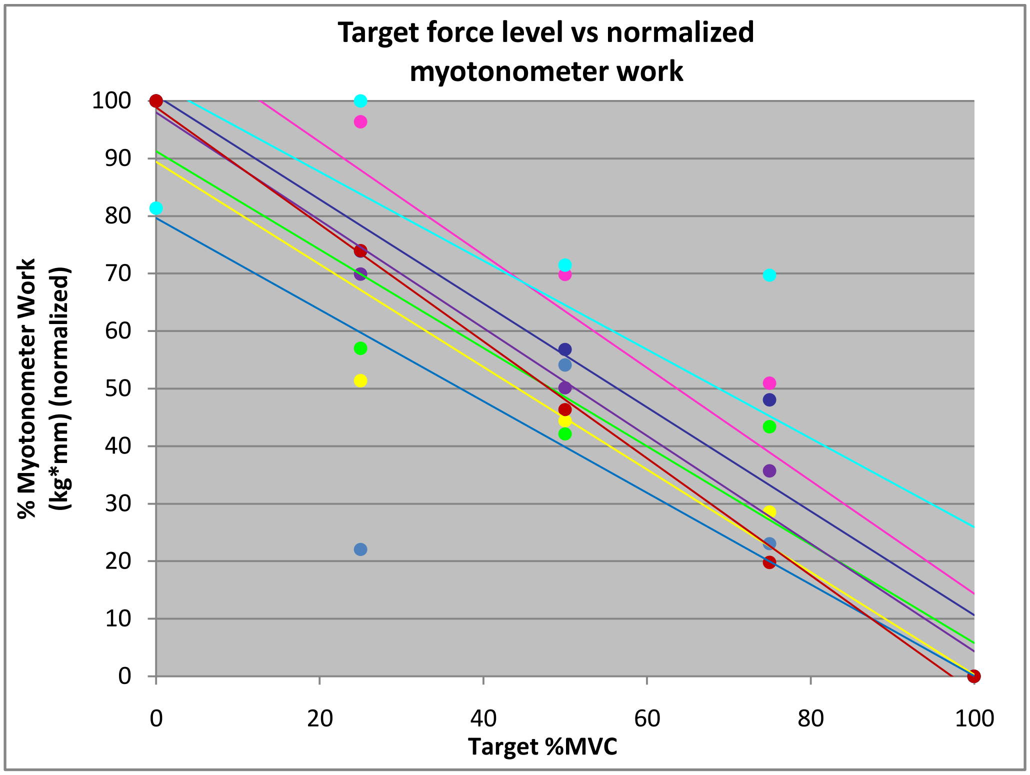


Figure 6. When the AUC$_{\%MVC}$ values are averaged and normalized, the myotonometry results show good correlation to the targeted level of grip force but a high level of variability still exists.

tFMG

The tFMG results of the testing indicated good results, most runs showing high correlation between GFD activity and tFMG response, though a response is not always recognized at low % MVC levels. Figure 7 shows the normalized GFD output, the level of activity that was displayed to the subjects for targeting purposes, and the raw voltage output, used for tFMG. In this case the tFMG response was averaged over the time of gripper activation in order to obtain a single representative data point, these points were then converted to force in grams based on a linear scale (Abboudi et al., 1999) and averaged with the other corresponding tFMG responses for each % MVC level of the subject, again to reduce variability. When the subjects' averaged % tFMG levels are plotted against the targeted % MVC levels, the increase in radial muscle pressure as the grasp level increase is consistent (Figure 8).

Normalized GFD output and raw tFMG response

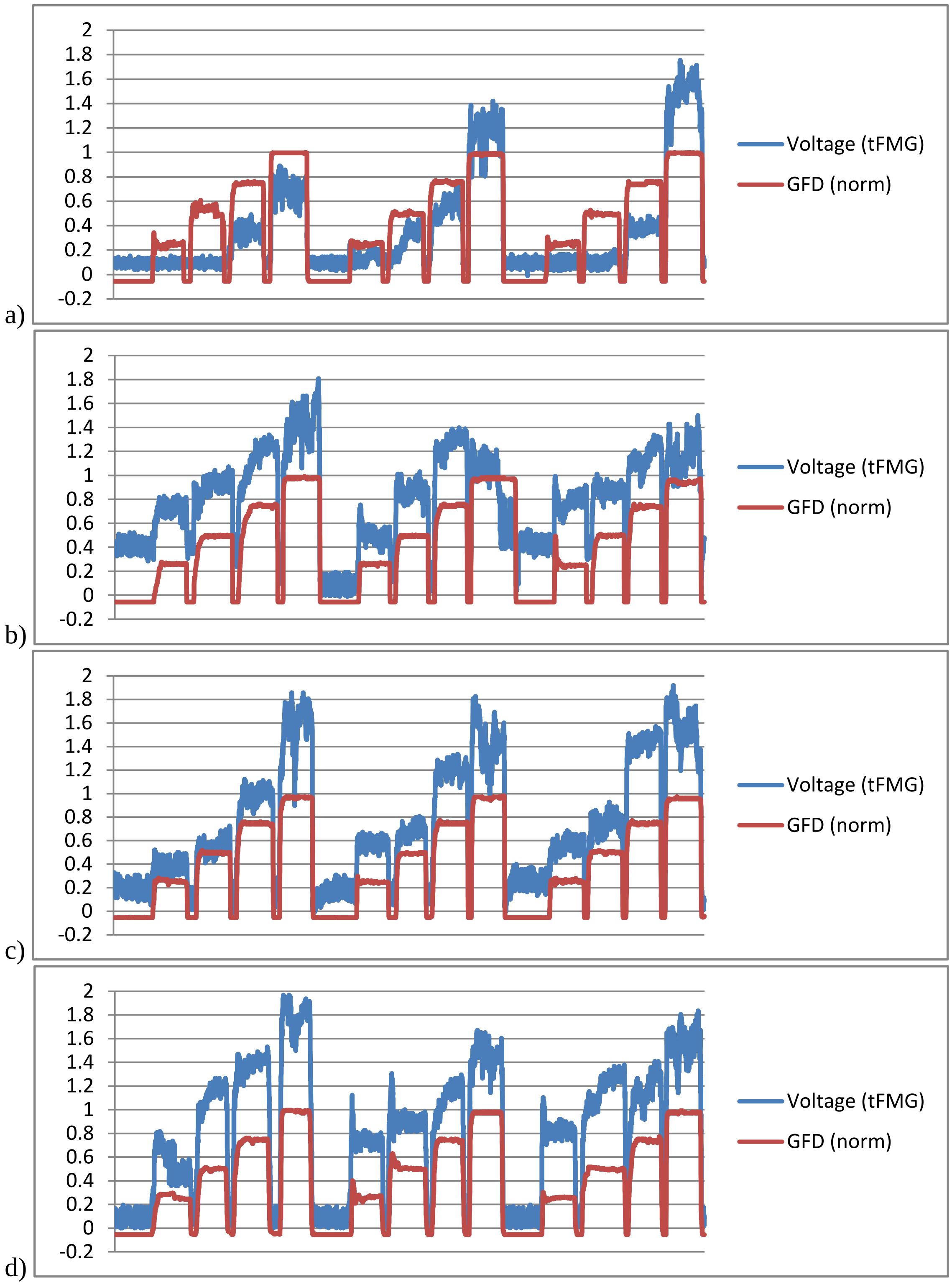


Figure 7 a-d. The normalized GFD activity and tFMG voltage output for four subjects shows either relatively good correlation or little activity. This is a good indication for the ability of tFMG of the FDS to correlate to hand grip force but that the methods must be adjusted.

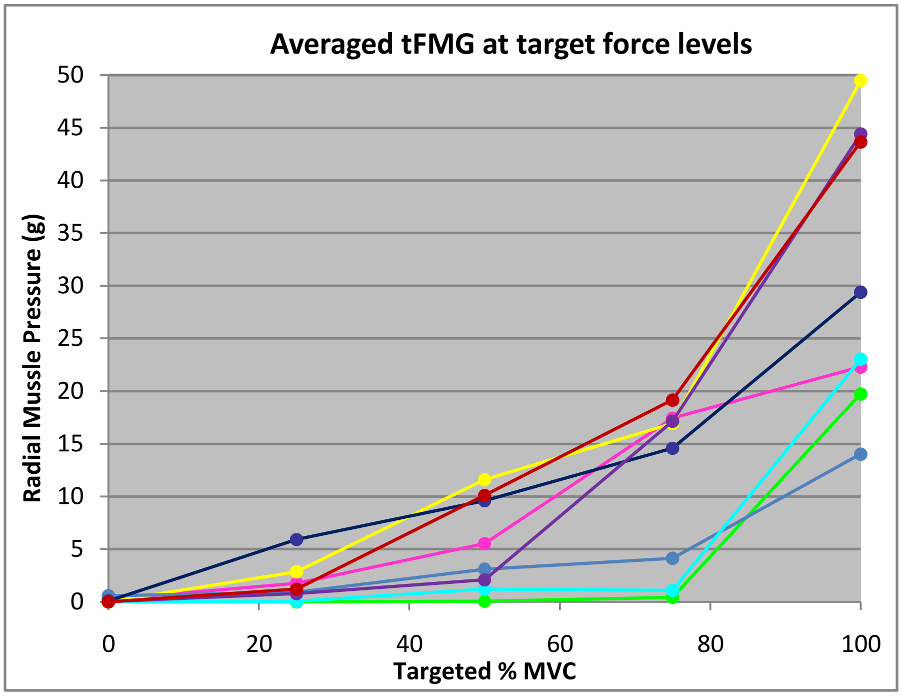
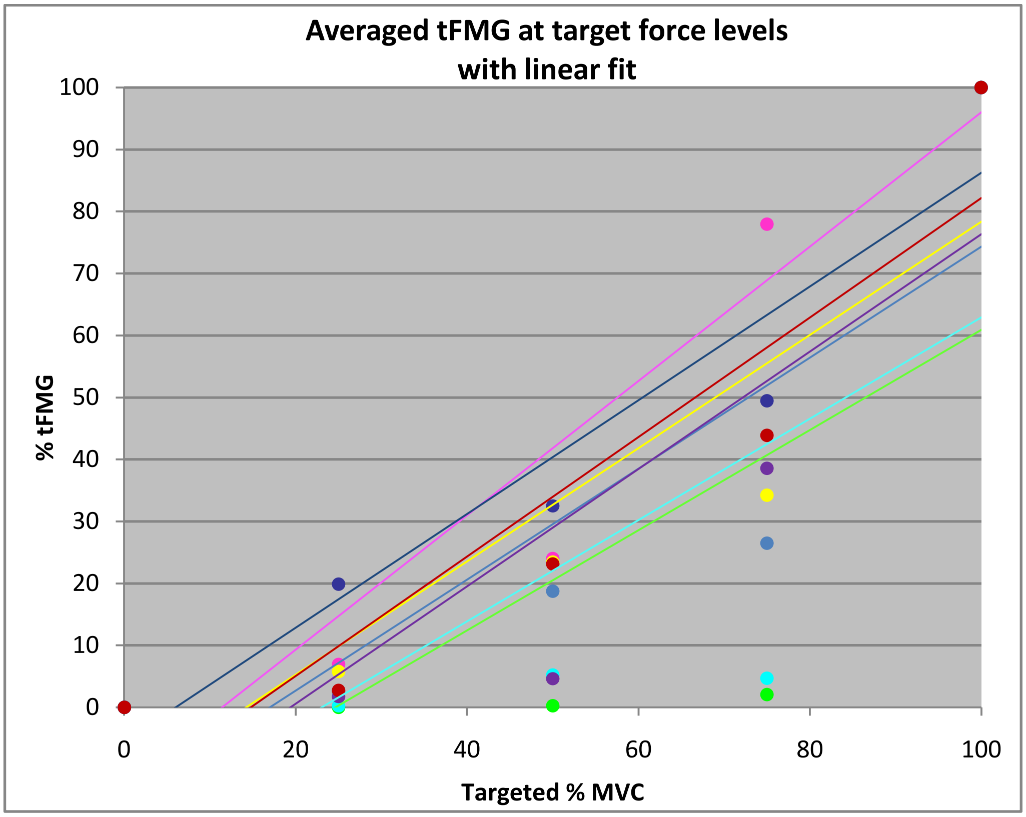
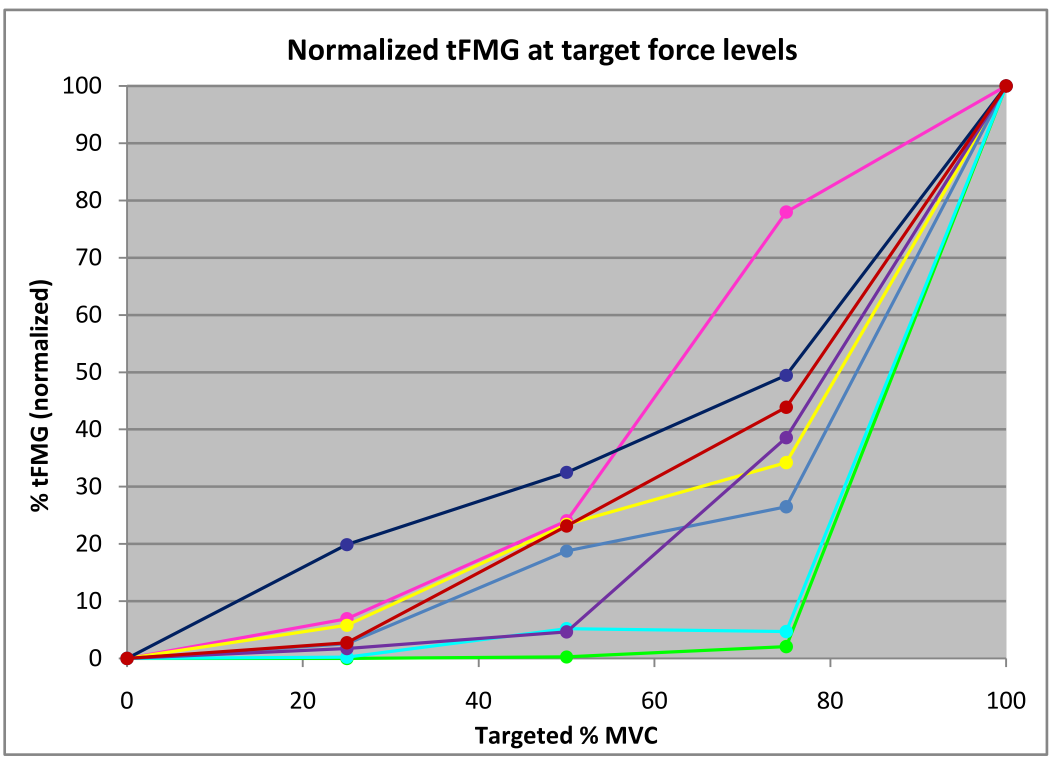


Figure 8. The pressure exerted on the FSR is constantly increasing when the grasping force on the GFD is increased. The exponential jump observed in each subject is most likely

The averaged tFMG values were also normalized for each subject's targeted % MVC levels by subtracting the averaged resting tFMG level and dividing by the maximum value. These results did not show a good correlation to the targeted % MVC levels (Figure 9a). The root mean square values of all the plots averaged to 0.76 with a σ of 0.16, a high of .92 and a low of .51. A high correlation was not expected though due to the non-linear nature of the results in figure 8, but a continuous, increasing trend in the relationship becomes more apparent (Figure 9b).



a)



b)

Figure 9. a) The normalized tFMG results show very poor linearization. b) The non-loinear nature of the curve with connected lines seems to be the best fit to represent the trend of the data.

E. DISCUSSION

Myotonometer

As was expected, the results from The Myotonometer show a high correlation to the relative level of muscle activation, the change in muscle stiffness is known to be a good predictor of muscle activity. The relationship of these values to actual muscle tone can only be inferred based on previous research since no trained physical therapist was in attendance and the operator was not trained in the interpretation of Myotonometer results for the prediction of muscle tone. Stiffness was inferred based on the change in AUC from rest to full contraction. All subjects were of similar athletic backgrounds, no (semi)professional athletes or body builder, as could be expected, all subjects exhibited similar slopes and are assumed to have similar tone.

Normalizing the myotonometry data does not create overlapping plots as would be desired in a repeatable, marketable product. With an overall slope of -.89 the correlation of the data, while relatively reliable for individual subjects, could not be inferred simply based on a maximum level and a baseline, the targeted force values are necessary.

tFMG

The tFMG exhibits a consistently increasing output corresponding to increased grasping levels for all discernable responses, there are, however, issues that must be addressed. The first issue is when no/minimally discernable activity is observed (figure 7a and 8); this is most likely a product of the cuff not being properly attached or sensor slippage. The second issue is the non-linear behavior of the results; based on the literature, although there is no study relating the change in radial pressure of a single forearm muscle to grasping force, it is assumed that the relationship should be linear (Abboudi et al., 1999; Kenny et al., 2005; Wininger et al., 2008; Zheng et al., 2006). This non-linear relationship can be attributed to two factors. One, the stiffness of the round muscle body increases the pressure causes the sensors to bend as well as compress, this bending may cause different response than compression alone. Two, as will be profiled in a different section, the force voltage curve of FSR sensors is an exponential or multi-ordered polynomial curve. The increased response on the sensor would exponentially increase the outputted voltage; this indicates that the linear conversion from voltage to force for the 1.4 cm FSR was invalid and merely an arbitrary unit change.

tFMG vs Myotonometry

The Myotonometer results are very good for determining differences in muscle stiffness, showing a distinct change in stiffness over the entire range of muscle activation. Though, since no Ashworth Scale tests were performed, it is impossible to determine if these reading are related to the subject's tones. What can be observed is the correlation of The Myotonometer to the tFMG results, indicating a potential for this relationship in subjects afflicted with muscular disorders. When this comparison is done the resting tFMG value is ignored, this value is an arbitrary initial pressure that is determined by how tight the operator attaches the cuff. When AUC results are plotted against the tFMG results for the varying muscle activation levels while ignoring the resting level (Figure 10) the slopes of are very similar. Indeed, when the difference in responses between $AUC_{25\%MVC}$ and AUC_{MVC} is plotted verse the difference between $tFMG_{25\%MVC}$ and $tFMG_{MVC}$, a highly correlated relationship is observed, ignoring the results of the outlying subject, the r^2 value is .86 (Figure 11). The subject with the most noticeably different slope and outlying relationship value was the only subject tested with noticeable obesity, this may account for deviations in the results of either myotonometry or FMG since no study has observed the effects of subcutaneous fat on either modality.

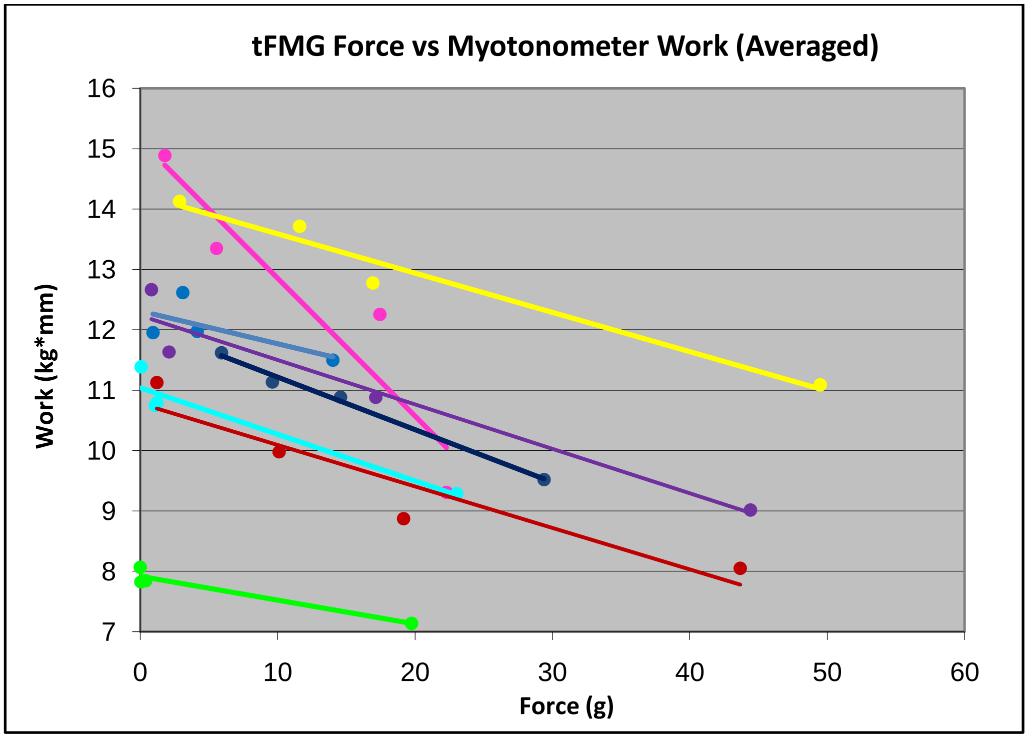


Figure 10. The plot of Myotonometer work vs. tFMG pressure shows a similar slope for all but one subject who was noticeably obese.

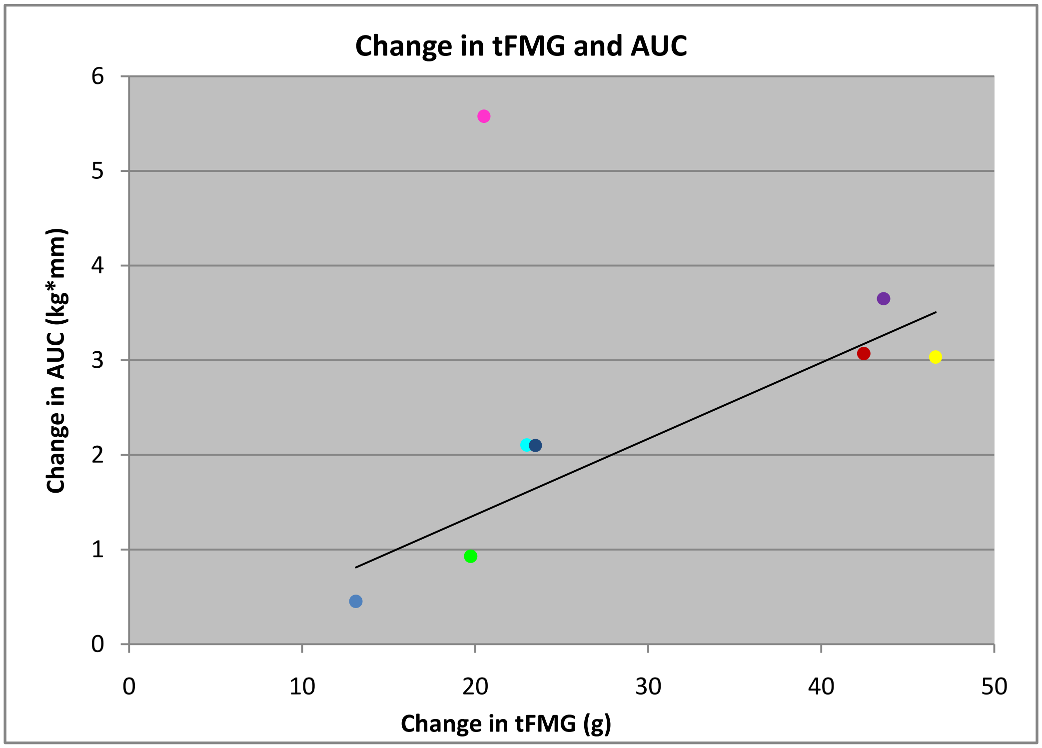


Figure 11. The change in AUC and tFMG between 25% MVC and MVC show a good correlation. This indicated that tFMG has the potential for assessing muscle tone based on the subject's distance away from the normal line if a reliable method was developed for assessing tFMG readings of a relaxed muscle.

Spasticity is determined by a myotonometer by comparison of difference of the work done by the myotonometer when the muscle is at rest and when the muscle is in MVC. Figure 11 reveals that the relationship of change in myotonometer work has a continuous, positive correlation with the change in tFMG. It is then conceivable that tFMG could be used to assess spasticity if a device and method were developed to accurately take tFMG readings when a muscle is in a relaxed state. This device would have to monitor both the length and the tension at which the FSR sensor is strapped to the forearm. However, without comparisons to spastic individuals this relationship can no go no further than hypothesis.

F. CONCLUSION

Both tFMG and myotonometry show correlation to the level of muscle activation. This, however, is only achieved with knowledge of the targeted level of activation and averaging of the subject's trials, and still yields high variability. With the current setup and method both modalities are insufficient to truly predict grasping strength at a functional level.

Section 3. A Novel Grip Dynamometer

A. INTRODUCTION

Most research studies have used custom-built dynamometers that measure grip or pinch strength on particular objects or manipulandums, that give digital outputs (Blennerhassett et al., 2007; Dun et al., 2007; McDonnell et al., 2006; Nowak et al., 2007; Nowak and Hermsdorfer, 2006; Raghavan et al., 2006; Schenker et al., 2006). Commercially available grip dynamometers are based on hydraulic, pneumatic, mechanical, or strain gauge methods. Unfortunately, current popular dynamometers are designed more for use at the high-end of the force scale, and thus their accuracy and range for disabled users is relatively limited (Bohannon, 1999). There are inexpensive wired dynamometers available, and there are also high-end grip sensors, that sell for many thousands of dollars (PPS, 2008). Thus, there are commercial electronic hand dynamometers, spanning 2 orders of magnitude in price range; however, they generally lack crucial features and the programmability required for generating scientific rehabilitation protocols and biofeedback. This Section deals with the subject of attempting to fabricate a cheap, reliable, and accurate hand help force gauge that can fill this niche.

As discussed, and utilized, in the previous Section, force sensing resistors (FSRs) can assess relative grip force of the hand, while affixed to a cylindrical gripping device. FSRs are a polymer thick film overlaid with electrode teeth; with a fixed voltage passing though, the polymer exhibits a decrease in resistance with an increase in the pressure applied to it and an infinite resistance with no applied pressure. Its output is related to the amount of sensor surface addressed, the type of actuator, and the ratio of the area of the applied force to the electrode pitch, in this way the FSR is neither a pure force nor a pure pressure sensor.

FSRs have many advantages; small size, low weight, inexpensive, and are easy and versatile to utilize. The disadvantage of using FSRs comes in their low inter-sensor and intra-sensor reliability as well as their uncharacterizable response to non-uniform pressures and mechanical moments. However, the set-up of the sensors and the methods of calibration can be refined to maximize reliability and accuracy. One such adjustment, in order to improve intra-sensor repeatability, is to attach a solid structure to the pressure sensing area (Pylatiuk et al., 2006), holding the sensor contact area constant and preventing bending, subsequently converting the FSR into a force sensor. A solid structure will uniformly distribute the pressure exerted on the sensor, independent of the actuator contact area. Holding the sensor contact area constant, intra-sensor reliability, as force sensor accuracy, is reported as ranging from ±5% to ±25% (FSR), but as high as ±2% with proper mechanical arrangements (Yaniger, 1991). Additionally, a rubbery or soft overlying layer above the sensor has been shown to distribute the applied load over a larger pitch, effectively increasing the slope of the force/voltage curve at low forces and a decreasing the slope at high forces (Yaniger, 1991). In this way, it will be attempted to use FSR sensors to both accurately measure the grip force of individuals and the change in radial force exerted by their muscles.

Inter-sensor repeatability is reported as ranging from 15% to 25% (FSR). This high variability requires that the force/voltage curve must be individually calibrated for each sensor or sensor configuration under the same mechanical arrangements. This can be done easily and effectively with either a force gauge or an Instron mechanical testing system.

FSR force vs. voltage curves have been expressed as a simple logarithmic regression (Yun et al., 1997), a 3rd order polynomial (Pylatiuk et al., 2006; Wininger et al., 2008), and a 7th order polynomial (Castro and Cliquet, 1997). This fit to the force/voltage curve must be generated and implemented under the testing conditions discussed above. Each of the fits mentioned above are valid since the behavior of each of these curves is dependent on the surface size of the sensor, the relative area on the sensor that is being utilized, and the fixed voltage value that is placed in the voltage divider circuit.

Hypothesis 3

A grip dynamometer using FSRs can yield a linear dynamic response over the useful range of power grip.

Development of Dynamometer

In the previous section an octagonal GFD device with thin foam wrapped around it was used to assess relative levels of grip force. No actual grip force units were ever inferred from the output of the GFD, rather a MVC was calibrated and all gripping forces were relative to that. Without an accurate and reliable knowledge of gripping force it is impossible to truly quantify tFMG behavior relative to %MVC.

The first step to the calibration of the GFD was done by a comparison of 10 subjects' MVC on the GFD to that on a quantifiable grip force gauge. Currently, the most effective and widely used method for the clinical assessment of hand grip force is the JAMAR, a uni-axial, pneumatic force gauge. This device is the gold standard in this field for its high levels of reliability and repeatability. In this study a Jamar-type, uni-axial hydraulic hand dynamometer (Fabrication Enterprises Inc., NY) was used.

B. VERSION 1.0 GFD

With a 5k Ω fixed resistor value in the voltage divider circuit, 7 subjects grasped the GFD with a MVC and a moment later grasped the Jamar with a maximal effort, the task was then repeated with their opposite hand. Peaks were read from the recorded data and a maximum force was read off the Jamar indicator. The results show that voltage from the GFD was able to accurately and reliably predict an accurate representation of true grip strength to a best fit line with moderate to low accuracy (Figure 12). Through testing, a new resistor value of 220 Ω was chosen to maximum the linear range of the force/voltage curve and the range of voltage response, meaning lowest voltage change per force increment. The lower fixed resistor value will help linearize the curve, making lower and higher loads more distinguishable. This two peak grasps on each instrument was performed by the 13 subjects (Figure 12), several weaker subjects were purposefully obtained. Data collection was performed with a Nian-Crae board and Myotonus software (Nian-Crae, Somerset, NJ) and analysis was done in Microsoft Excel.

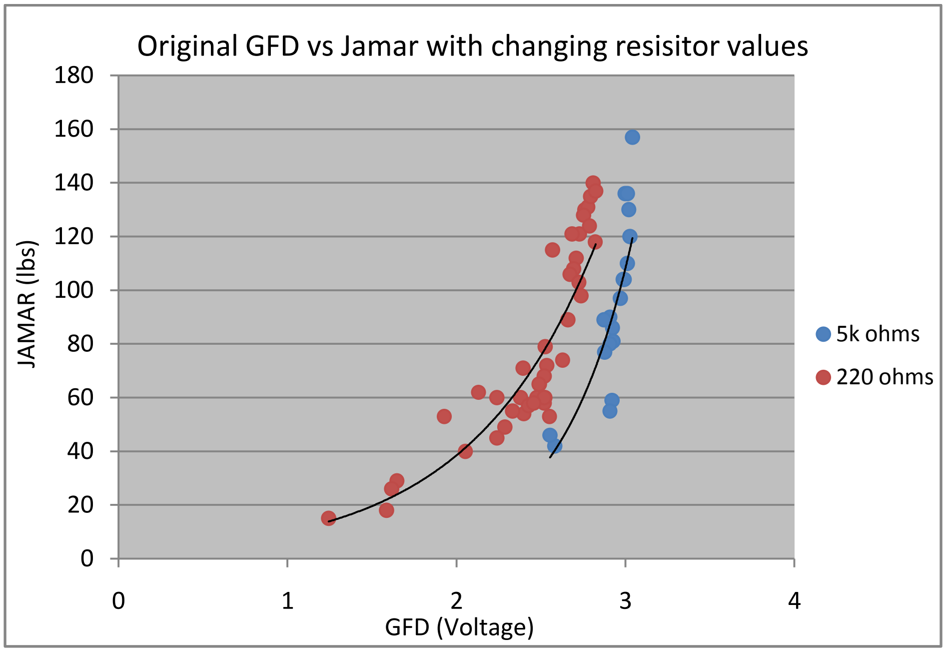


Figure 12. A comparison of a subject's MVC on both the GFD and Jamar show moderate to low accuracy in comparison when a subject's MVCs are compared.

Modifications to the Handle

In order to validate the use of FSRs as force gauge a direct comparison must be made to the Jamar hand grip force gauge at all force levels. Two FSR strips (Model #408, Interlink Electronics Inc., CA) were affixed to two concave adaptors with flat exteriors which were then placed and secured over the rounded fore and aft handles of the dynamometer. The fixed resistor value was changed to 200 Ω and two FSRs and 1/8" foam was cut to the size of the fixtures and placed in line with the axis of gripping force on the Jamar. The dynamometer was secured onto a desktop surface to maintain neutral wrist position of the subject forearm. In this way the grip force was simultaneously measured by the Gripper and the Jamar. From starting position with zero force, subject was asked to reach the target force level which varied from 5 to 60 lb with increments of 5 lb force per each run with 1 minute resting period between each trial to avoid fatigue.

The force range covered in this study (0~60 lb) is sufficient to deal with most of the typical forces generated by the hemiparetic population. Force values from the dynamometer were acquired by the maximum force indicator. The corresponding FSR output value in voltage was acquired, and the peak was selected. (Figure 13).

It was quickly realized that although lowering the fixed resistor values allows the force range to be more distinguishable, there is still a high amount of variation that will be inherent in inter-subject grasping; contact area on the sensors will be dependent on the size of a subject’s hand. The contact area must be made constant in order to create an accurate and reliable force gauge. The next method to smoothing the curve was to uniformly distribute the load over the sensors entire active area. A 1/8 inch rigid metal strip was cut to the size of the sensors and affixed above the 1/8 inch thick piece of non-hysteresis foam cut to the same size (Figure 14). These rigid plates distribute the load evenly over the entire sensing area, regardless of contact area on the plate, and the foam ensures an even interaction with the polymer in the sensors. This distribution of force causes a further linearization of the force/voltage curve of the FSR (Figure 13). For this and future testing, data analysis was performed in though a program developed in Matlab. The program used a windowing method to find the peak force values of every power grasp. The program ensured that each peak was separated from other peaks by a reliable temporal distance.

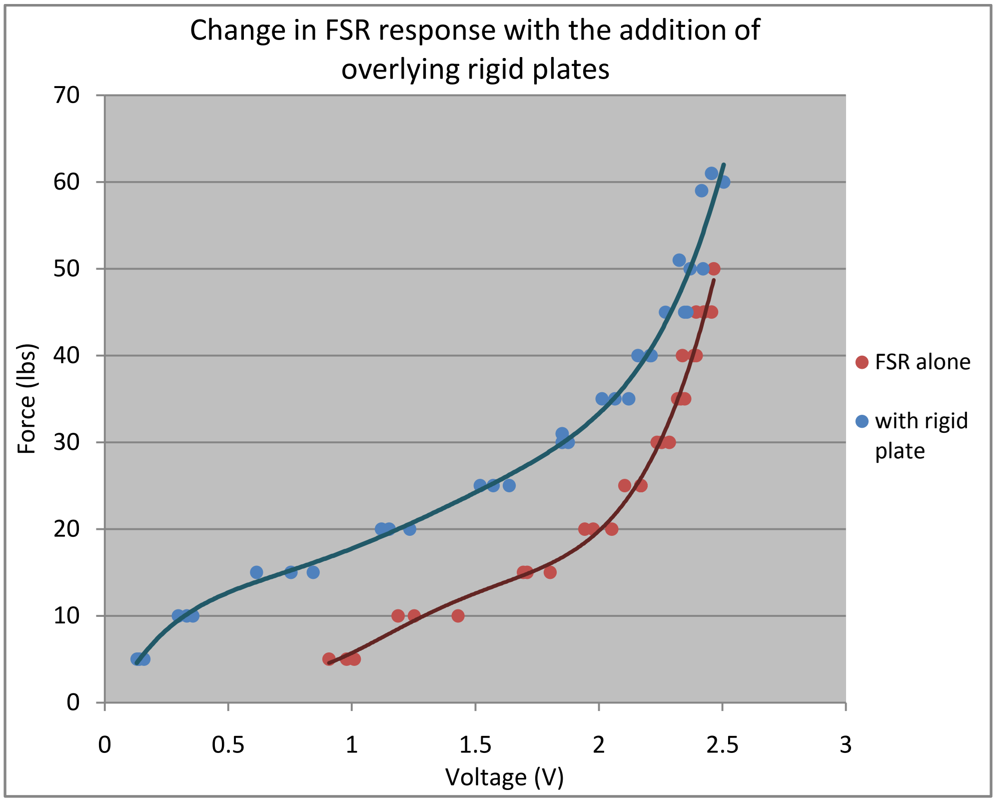


Figure 13. With the FSRs aligned with the gripping axis of the Jamar it can be seen that the addition of rigid plates above the sensors increases the linearity of the curve.

Figure 14. An FSR (Model #408, Interlink Electronics) with 1/8" thick non-hysteresis foam and 1/8" metal strip are shown in their overlying pattern. An ideal configuration for creating uniform pressure distribution with resting load applied to the sensor.

With rigid metal strips and foam fixed to the FSRs with 3M double sided scotch tape 9 subjects were asked to apply force to the Jamar set up with bursts of grasping strength from their MVC to 5 lbs in 5 lbs increments, again the Jamar reading were done by eye on the Jamar force gauge scale (Figure 15). This testing was done over a period of 3 weeks and in order to ensure that the FSR force/voltage response is the same regardless of the subject. The results of these tests showed a relatively consistent curve at low force levels but the consistency in the curve begins to falter at higher forces.

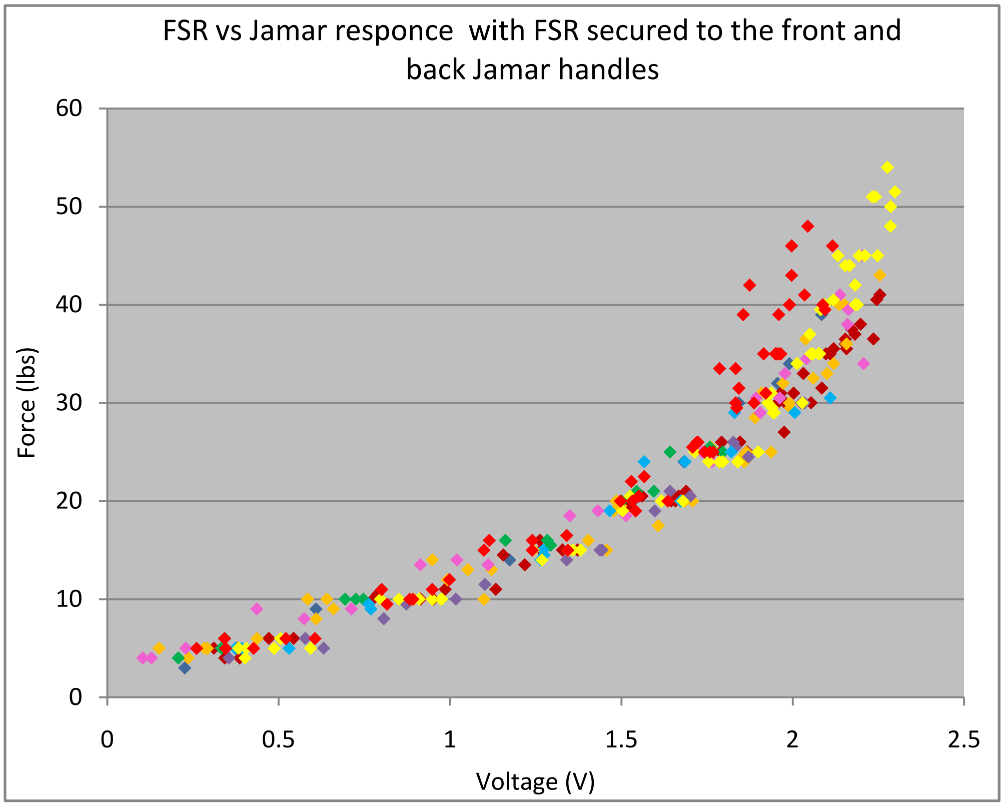


Figure 15. The comparing the Voltage peaks from the FSRs mounted to the Jamar with the 9 subjects grasping in bursts the peak strength for each burst read off the Jamar shows a consistent trend at low force levels but scatter increases at higher force levels.

Instron Testing

The increased scatter at higher force levels, seen in Figure 15, was assumed to be the natural variability of FSRs until the two FSR strips were removed from the Jamar and a calibration curve was established by compression testing in an Instron mechanical testing system. For this method the FSR, foam, and plate sensor units were placed on a flat base plate with a 1/16" aluminum plate lying across both sensors. The Instron compressed the sensors in pulses to the same target force levels as the Jamar testing. This method revealed a flaw with the results from gripping on the Jamar. The results of grasping the FSRs on the Jamar show a lower voltage response than during Instron testing (Figure 16). This is presumed to be due to a bending moment on the fixture attached to the curved Jamar handle. This bending has an unexplained effect on the FSR response; it is presumed that a lower voltage response when the FSRs are fixed to the Jamar is due to the change in behavior of the piezoresistive film when the FSR bends. This testing modality is unacceptable, different subjects will cause different bending moments on the fixture and FSR. It was determined that the rear handle fixture alone, was experiencing bending and was removed in order to allow the FSRs record a purely compressive force for every subject grasping the Jamar.

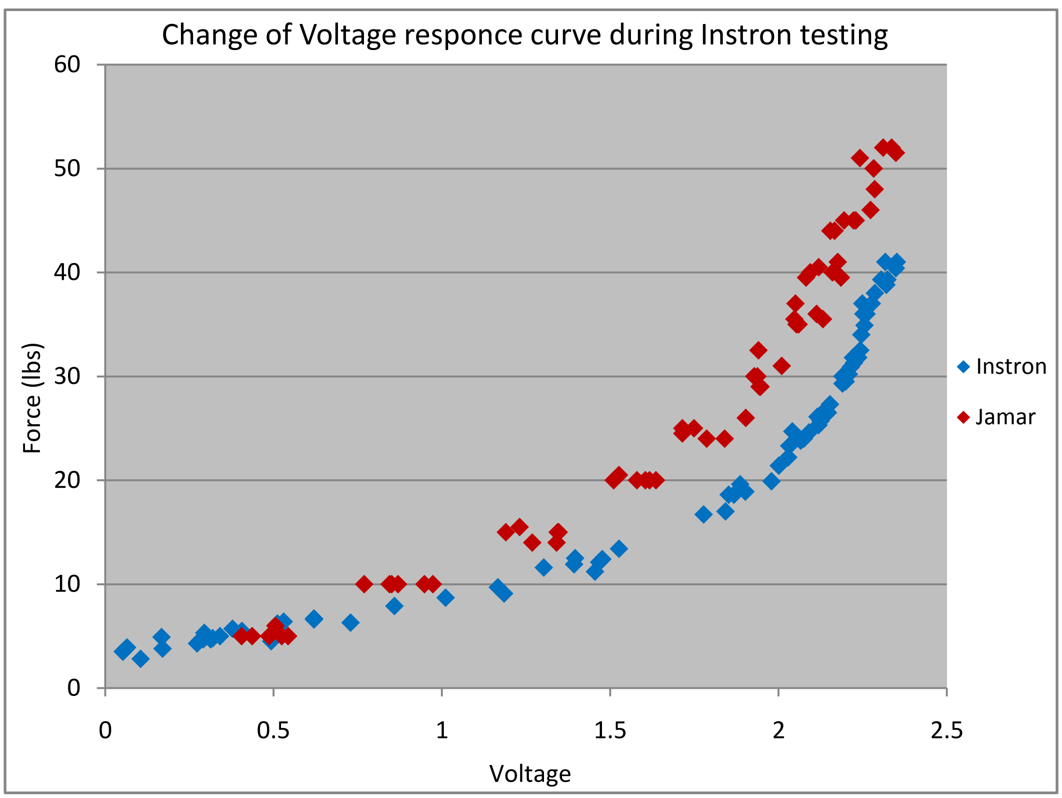


Figure 16. The voltage response of a single subject grasping the Jamar and the FSRs placed under uniform compression in an Instron testing system. The FSRs' response changes when they are in pure compression in

the Instron. To make a reliable testing modality the sensors must be able to be calibrated repeatedly and reliably by an Instron.

A single FSR sensor, sized to fit on the front handle of the Jamar, was then placed into an Instron by itself and a calibration curve was created (Figure 17). After reattaching the FSR fixture to the Jamar (Figure 18), 6 subjects were tested with the same test method and found to follow the curve calibrated by the Instron very well (Figure 17). This indicated that a purely compressive calibration test performed on an Instron mechanical testing system is transferable to a handgrip force gauge. The curve loses linearity due to the smaller sensor area reacting in parallel; the resistor value was not changed as to not reduce the active voltage range.

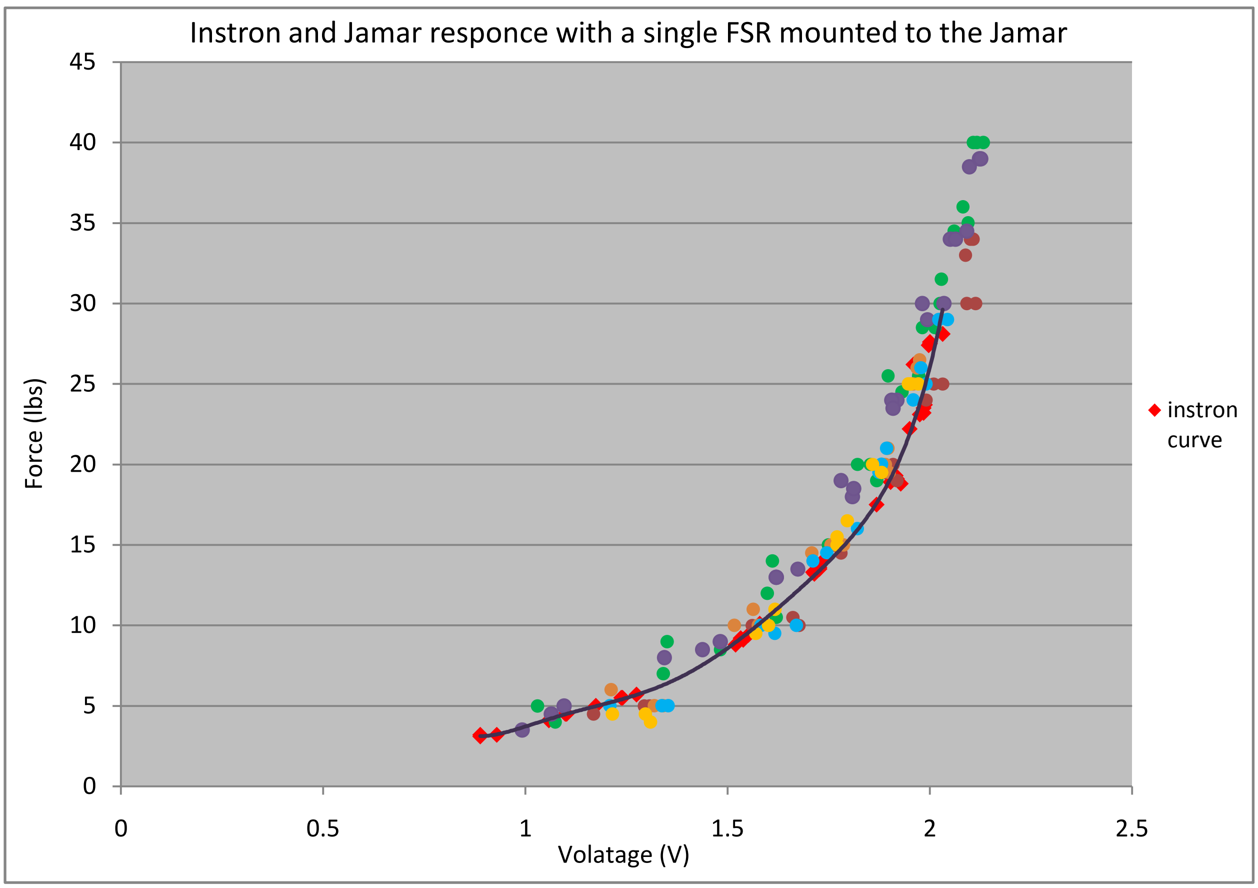


Figure 17. With only a front FSR fixture on the Jamar the force curve shifts to be less linear but allows for more accurate and repeatable results when interpreting force.

Figure 18. The single FSR fixture mounted to the front handle of the Jamar.

It was also found that the FSRs have a highly sensitive response to low forces that could not be resolved by the Jamar, being poorly responsive and inaccurate below 10 lbs. This is an important advantage of this testing mode, since some user populations, such as hemiparetic, stroke, and cerebral palsy patients, often cannot produce much force. In later testing, a group of 5 subjects grasped the Jamar setup in with a real time, linearized voltage curve, giving visual feedback to grasp strength. The peak forces recorded through the linearized curve showed an excellent correlation to the force observed by an operator on the Jamar (Figure 19).

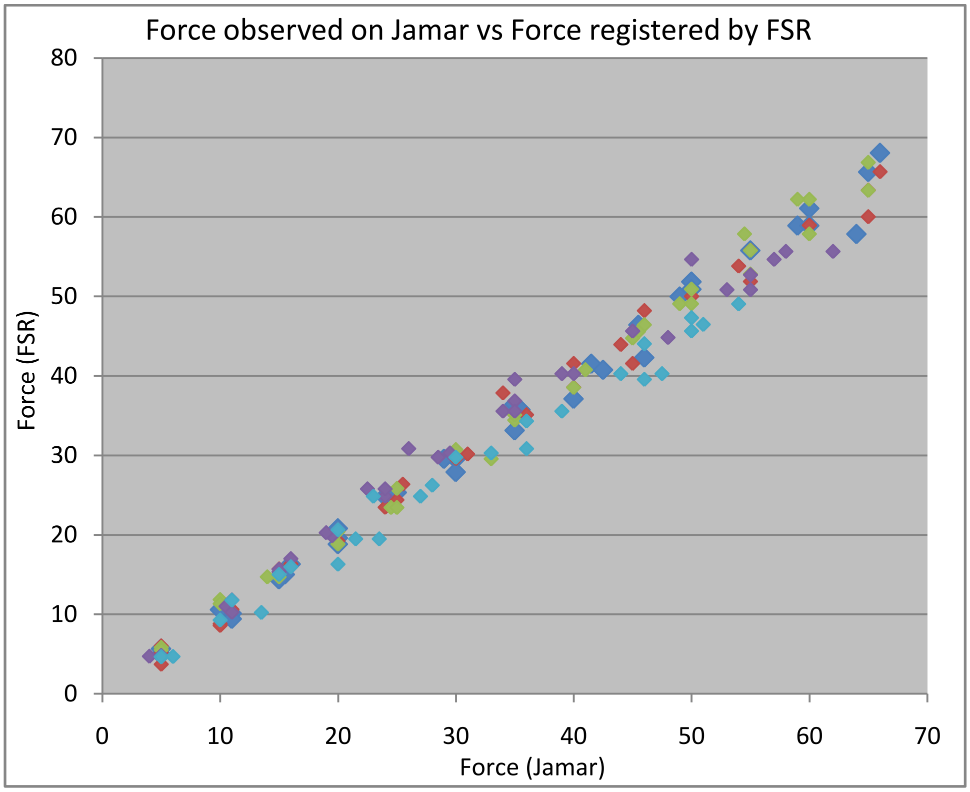


Figure 19. The linear fitted data from the FSR is highly correlated to the force recorded by the Jamar style hand held dynamometer. This shows that the force being displayed to subjects and recorded by LabVIEW is accrate to the actual grasping force exerted by the subject.

Forearm Sensor

Another concern addressed from the previous section was the behavior of the 1.4cm FSR that lays prone to the skin over the muscle body. In order to accurately test the tFMG of the FDS a reliable and accurate force gauge must be utilized that will lay flush to the surface of the skin while causing minimal interference to muscle behavior. Previous FMG studies have merely held the FSR, alone, on the skin surface of the skin (Curcie, 2001; Wininger et al., 2008). This will cause non-uniform pressure distributions and bending of the sensors. As stated in the previously, in order to convert an FSR into a reliable and accurate force gauge these two issues must be resolved by creating a fixture to house the 1.4 cm FSR. A fixture housing the FSR will also allow for linearization of the voltage response curve.

With this in mind, a sensor housing was developed of a minimal thickness, .26 in, which would apply all force due to radial changes in forearm pressure normal to the FSR as well as prevent any sensor bending. 1/16" PVC was cut into two circles approximately the size of the 1.4 cm sensor. The FSR and a 1/16" piece of foam, cut to the size of the active area of the sensor, were secured between the two pieces of PVC by double sided tape (3M) (Figure. 20). A piece of Velcro was also cut to the size of the sensor in order attach the sensor to the forearm strap used in testing, this will allow accurate targeting of the FDS and prevent sliding of the sensor.

The FSR was again put into a voltage divider circuit with a 200 Ω fixed resistor value and Inston compression testing was performed in order to linearize the curve. Data collection was performed in LabVIEW. The force voltage curve shows a good linear range at the expected force values, < 10 lbs (Figure 21).

Figure 20. The FSR fixture for testing tFMG. The FSR and foam are sandwiched between two pieces of PVC; the overlying Velcro is for affixing the sensor to a forearm strap, allowing the sensor to accurately target a specific muscle.

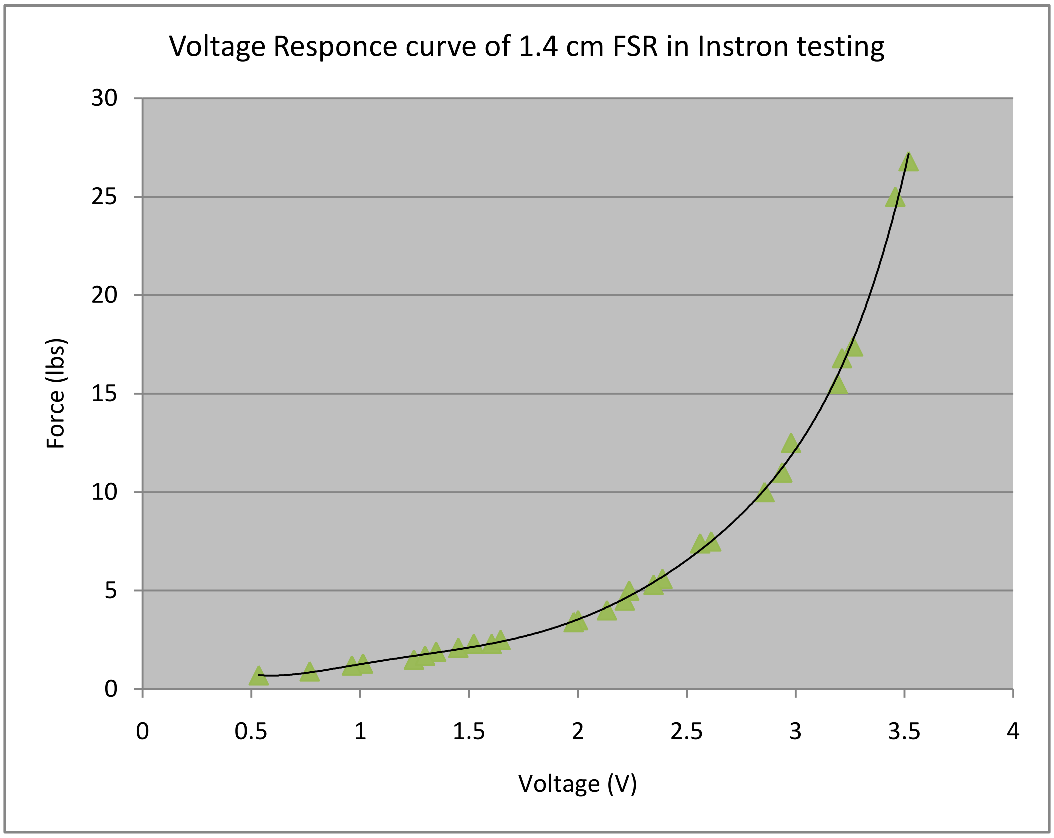


Figure 21. The force voltage curve of a 1.4 cm FSR in a custom fabricated housing during Instron testing. The curve calculated here is used to linearize voltage response of the FSR to mechanical pressure exerted radially by the forearm. The linear range of the curve is very good for the expected force values.

The final concern about the use of FSRs is their propensity to drift. Drift can occur due to temperature and humidity or merely over time. Drift was notice on the single FSR fixed to the front handle of the Jamar over a period of months when temperatures were constant (Figure 22). The force/voltage curve does not seem to fluctuate much on a daily scale but a clear drift can be seen over a period of months. This can be easily rectified by calibrating FSR sensors at least once a week. A single calibration run is all that is required for a new linearization equation due to the high reliability of the FSRs in the present setup.

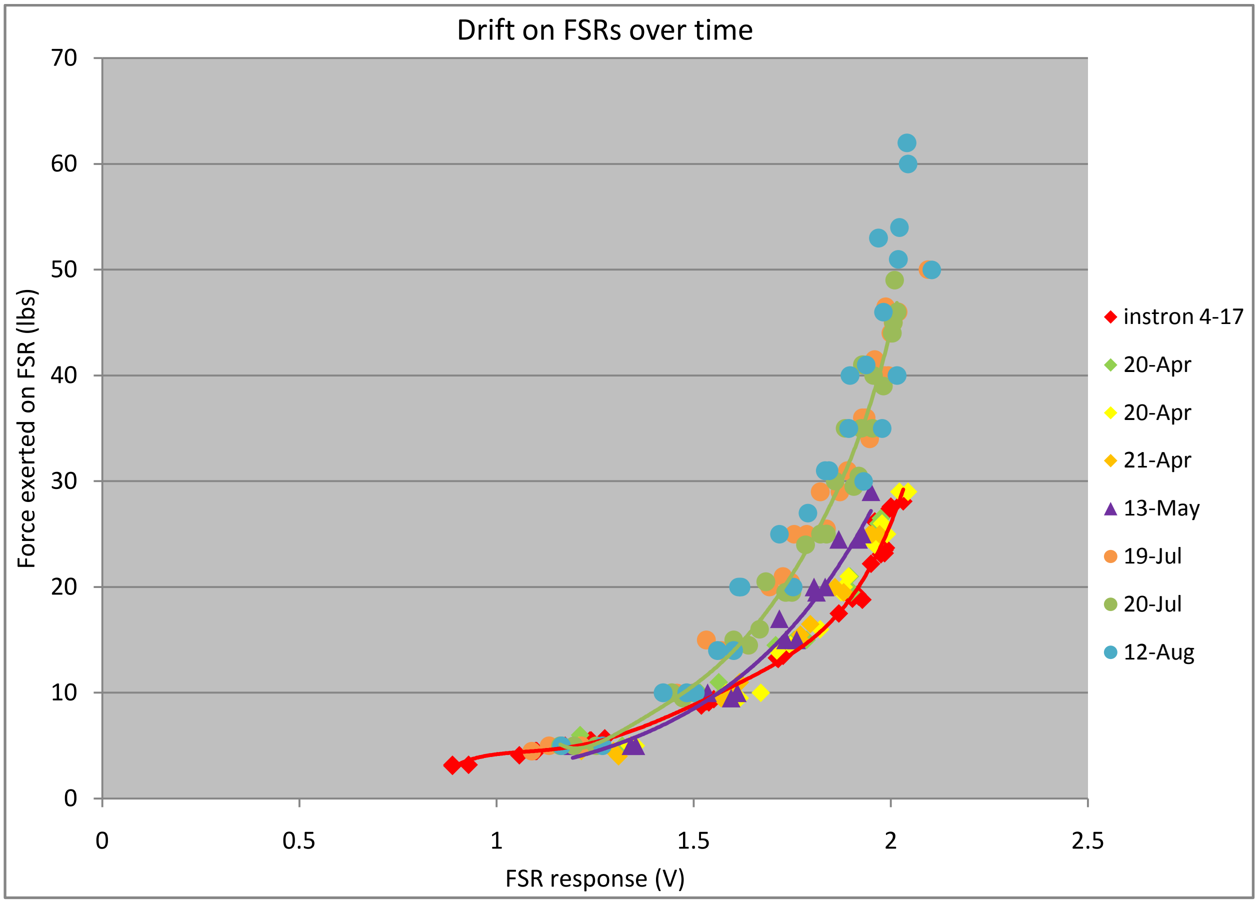


Figure 22. Drift is a concern in the use of FSRs. Drift is generally negligible over a period of days but over a period of week and months the drift is much more noticeable. This can be easily accounted for by calibrating FSR sensors at least once a week to ensure proper force measurements are being taken.

C. VERSION 2.0 GFD

Persons with upper-limb paralysis with loss of sensation can benefit from proprioceptive exercises combined with visual-tactile biofeedback to help regain their grip strength control. But there is a lack of available modality among currently accessible rehabilitation training facilities which offers a way of monitoring patients' grip force with visual feedback which would be suitable for rehabilitation, but difficult to achieve. Current one-dimensional grip devices only allows partial contact between the hand and the device, which is more prone to interfacial and person-to-person errors; people with different hand anatomy will have different leverage effects depending on the exact contact points of their phalanges and metacarpal bones with the same device (Kaneko and Tanie, 1994; Montana, 1988). This makes a unidirectional grip sensor unable to ascertain the true grip force, especially when variation of hand sizes and contact points of the hand exist between subjects. Using a hexagonal grip device can compensate these contact and size errors up to a certain level by ensuring a more comprehensive palmar contact which enables better grip force detection for wider population with different hand anatomy.

With this in mind a new GFD was designed in *Pro-Engineer* software and rapidly prototyped at Rutgers University. The morphology of the GFD is crucial to accurately register grip force, which should enable maximal contact of the hand. The gripping action by human can be best matched with an oblong cylindrical. The diameter of the Gripper (long axis distance) was selected as 4cm to ensure the most stress-free cylindrical grip actions from general population. Previous work shows optimal handle diameter to be 19.7% of the user's hand length (Kong and Lowe, 2005) but also that handle diameter is does not affect muscle activity in power grasp (Hoozemans and van Dieen, 2005; Kong and Lowe, 2005). For the new GFD base material, PVC mold was chosen for its structural rigidity to ensure the detection of isometric force.

A drawing of the new and current design is shown in Figure 23. It is designed to use 2 identical pieces of PVC which sandwich the FSRs together with 1/16" compressible foam. The external profile is oval for ergonomic gripping and to ensure proper hand position to direct prehensile forces normally to the sensors. Pins spanning the two halves will prevent off-axis sliding. Force registration occurs by compression of the sandwiched FSRs. Preliminary tests with this design have indicated that precision requirements for orientation of the hand on the unit are not high and thus are not critical for accurate output. Optionally, the unit can be covered with a foam grip, which may be silk-screened with fingers and thumb to assist consistent hand placement. A nylon sock may also cover the foam for hygiene. A Velcro wrist safety strap will also be attached to unit. The design is robust and rugged, with all components well protected.

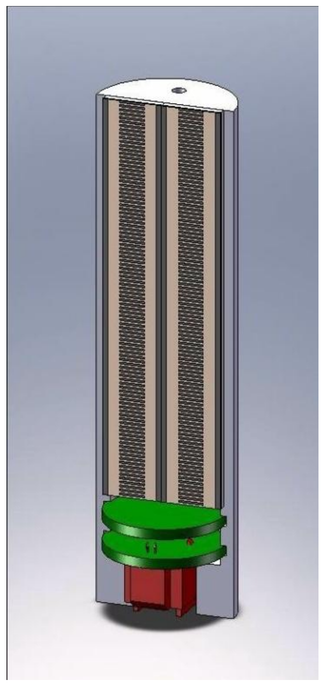


Figure 23. The new design of the GFD incorporates two FSRs with 1/16" foam cut to the size of the reactive areas, sandwiched between two half cylinders. This design, while uniaxial, utilizes the maximum contact area with the user's hand. The two FSRs are placed in parallel and connected to a voltage divider circuit.

This new low cost hand-grip force gauge was calibrated with a custom fixture holding the sandwiched faces of the two halves tangent to the ground. This custom fixture was placed in the testing area of an Instron and a force was applied normal to the FSR sensors. National Instrument's ADC device (NI-DAQ-6008) was used with PC for data acquisition. FSR's raw voltage output from voltage divider circuitry was fed into ADC channel. A 6^{th} order curve was fitted to the resulting force/voltage curve which showed very good accuracy and distinguishability up to 80 lbs (Figure 24).

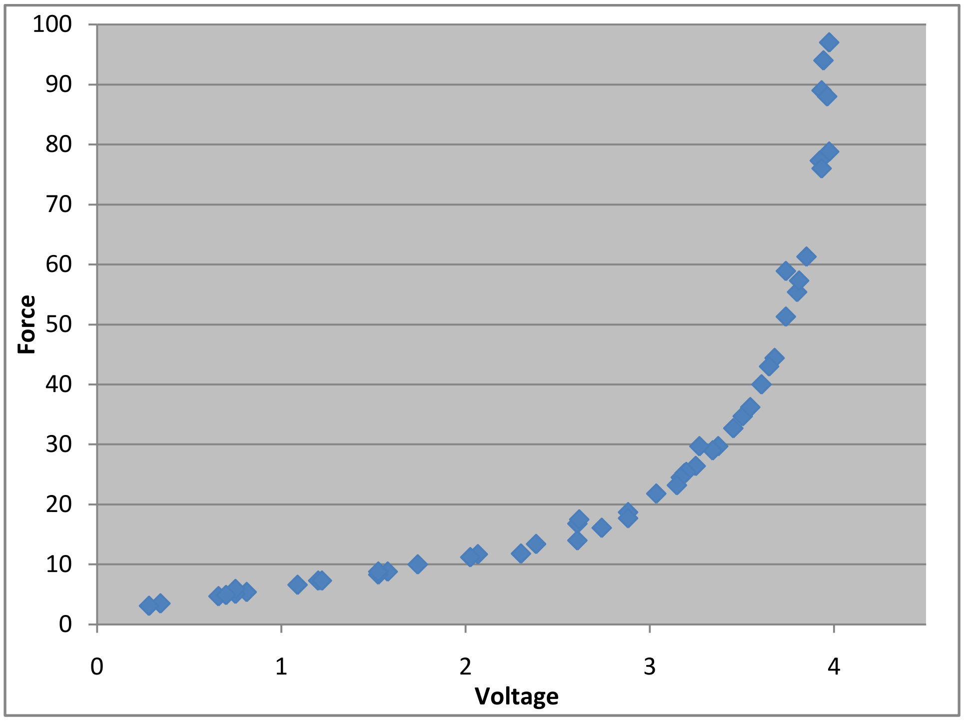


Figure 24. The force/voltage curve on the version 2.0 GFD shows good linearity and if very distinguishable up to 80 lbs.

Section 4. tFMG and GRASP STRENGTH

A. INTRODUCTION

Here, Hypothesis 1 is tested, specifically, that the *flexor digitorum superficialis* (FDS) can predict force of power grasp. A sensor on the surface of the skin measures the change in radial pressure and stiffness of the underlying muscle combined with the overall change in radial pressure of the entire forearm (tFMG). The FDS was chosen as the muscle of interest due to previous research focusing on the muscle (Abboudi et al., 1999; Duque et al., 1995; Hagg and Milerad, 1997; Heath, 2003; Kenny et al., 2005; Long et al., 1970; Loscher and Gallasch, 1993; Mogk and Keir, 2003) and Long et al.'s (1970) report that the FDS contribution during power grip is direct proportional to the resulting force.

B. METHODS

The grip strength testing performed in this Section utilized testing methods from previous sections. A single FSR with foam and a rigid metal plate was placed on a fixture and affixed to the front handle of a Jamar-style hand dynamometer and the Jamar was anchored to a table. Calibration was performed weekly to ensure drift was not a factor in the calibration curve.

An arm restraint was fabricated to keep the subjects' arm parallel to the ground and wrist at a neutral angle when grasping the Jamar. The FDS was found by palpating the subjects forearm during rest and maximum contraction and identifying the largest area of the muscle body (Abboudi et al., 1999; Curcie, 2001; Kenny et al., 2005). Once the largest area of the muscle body was identified the sensor, affixed to the forearm strap, was placed over the target area of the resting muscle and the strap was tightened to ~1 lb of force (Figure 25). This force was determined by visual feedback of force on the sensor displayed by LabVIEW. This section assumes that all force applied to the sensor is applied normal to the sensor face.

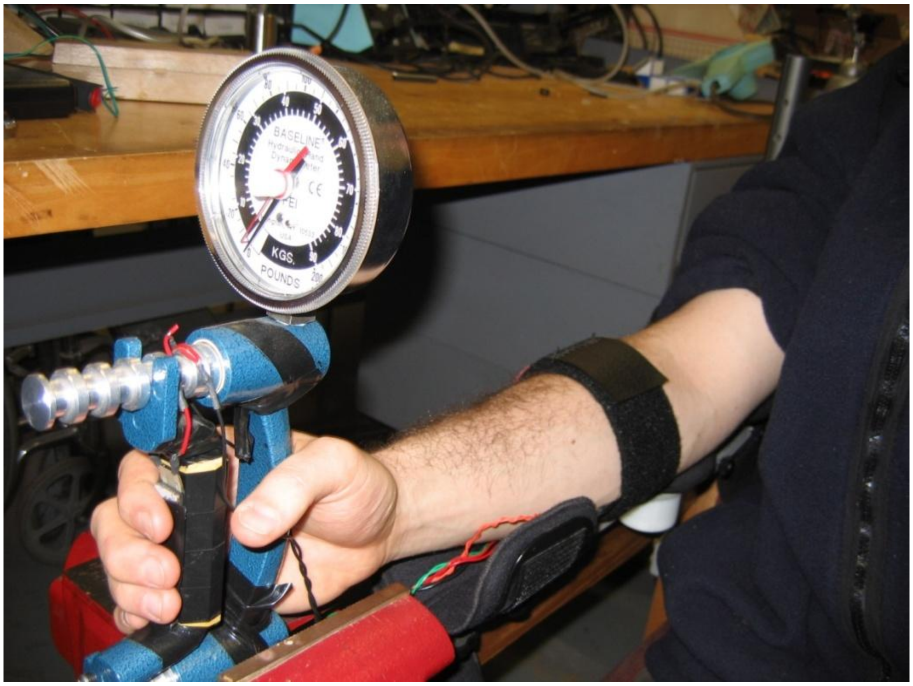


Figure 25. Testing setup shows the subjects arm movements restrained and a neutral wrist angle at rest when grasping the Jamar. The support is set up in such a manner as to not prevent any support from coming in contact with the forearm strap, preventing any interference in reading. The FSR located on the front of the Jamar handle allows the subject to monitor their grasping strength through LabVIEW.

This study recruited 11 subjects (7 males, 4 females, age 20-30), all subject reported no history of upper extremity or other musculoskeletal complaints, all testing was done on the right hand. Prior to testing the subjects were then given time to practice gripping the Jamar before data was recorded. Subjects were asked to grasp to their MVC with a burst of grasp force, then bust grasp to decreasing 5 lb force increments from their MVC to 5 lbs. Accurate achievement of the targeted force increment was not required since the study is concerned with how well FDS activity can predict grasping strength. This was repeated 3 times with a short rest period between in which subjects were asked to assess their own fatigue and comfort. Peak detection software was run on Matlab (see Section 2) with the addition of a threshold, incorporated into the program which ensured that any peaks detected were above the resting level of FSR response. This automated peak detection process was incorporated into the LabVIEW data collection program for visual feedback purposes but not utilized due to an intrinsic data averaging function in the LabVIEW peak detection subroutine.

C. RESULTS

Although the grip force plotted vs. tFMG of the FDS showed a linear correlation for subjects in the low-to-moderate force range (Figure 26), this plot cannot be used for inter-subject comparison. The variable resting levels and slopes of the data trend of the plots are influenced by the baseline pressure on the FSR, the location of the sensor on the subject's forearm (Orizio et al., 1999; Roman Liu and Tokarski, 2002), and the change in overall radial pressure exerted on the cuff. For comparable observation the data can be normalized.

The subject's observed peak values for each grasp were normalized by the equation:

$$\%MVC = \frac{GF - GF_{min}}{MVC - GF_{min}} \times 100$$

$$\%tFMG = \frac{tFMG - tFMG_{min}}{tFMG_{max} - tFMG_{min}} \times 100$$

Where GF is the grip force peak value recorded from the FSR mounted to the Jamar during each grasp burst, GF_{min} is the lowest non-zero grasping force peak, and MVC is the maximum voluntary contraction force. tFMG is the peak radial force value applied by the FDS to the FSR strapped on the forearm during each grasp burst, $tFMG_{min}$ is the lowest recorded peak radial muscle force above their activation threshold, and $tFMG_{max}$ is the largest peak radial muscle force. The values were not normalized to zero GF and resting tFMG because the resting tFMG value is an arbitrary value that is based on how tightly the operator applies the arm band; instead GF_{min} and $tFMG_{min}$ were used at the baseline.

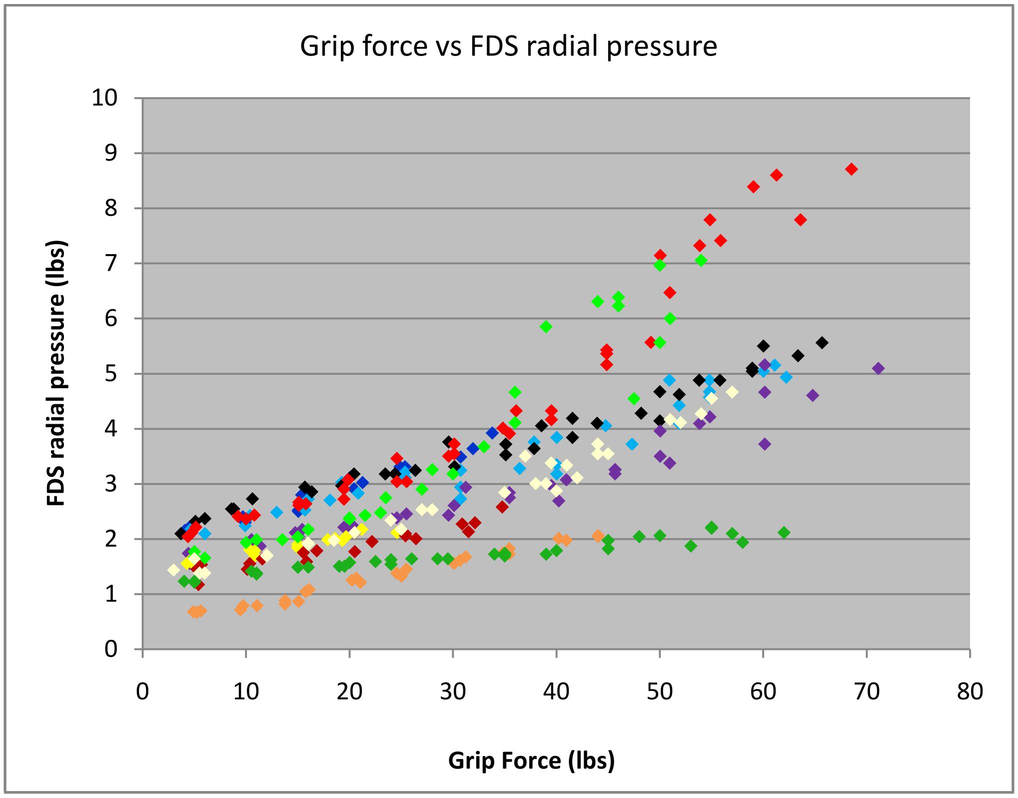


Figure 26. When the grip force is plotted vs. tFMG force from a single muscle a good linear relationship is observable on all subjects. Each color represents a different subject, colors are held constant for each subject throughout the section.

The normalized values of all subjects were then combined and plotted; a clear linear relationship can be observed (Figure 27). The overall best fine like has a slope of 0.95 and a y intercept of -2.2, with a correlation of determination of 0.92. The average error to this fit line for all data is 7.1 % MVC with a STD of 5.3. Figure 28 shows the overall normalized data displayed as discrete bins at 5% MVC increment with standard deviations taken over a 10% MVC range, the best-fit line, and a line with a slope of one. The STD envelope stays relatively constant across the entire range, 7.5 ± 1.8.

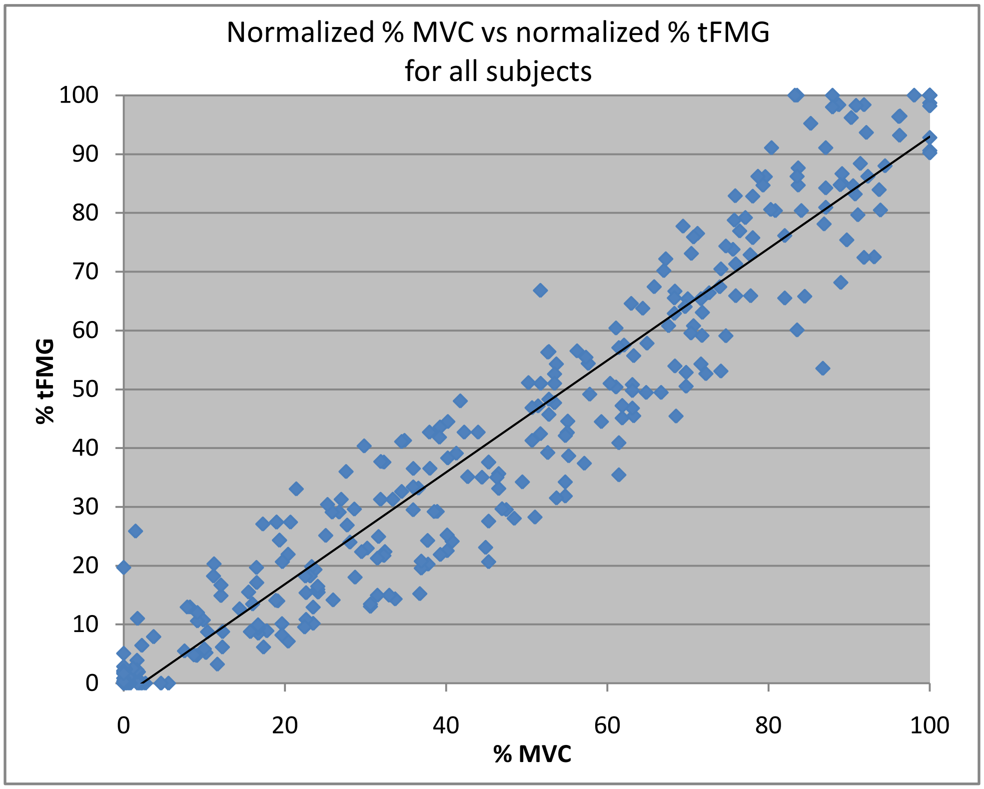


Figure 27. Normalized data for all subjects shows a strong correlation to a single linear fit line.

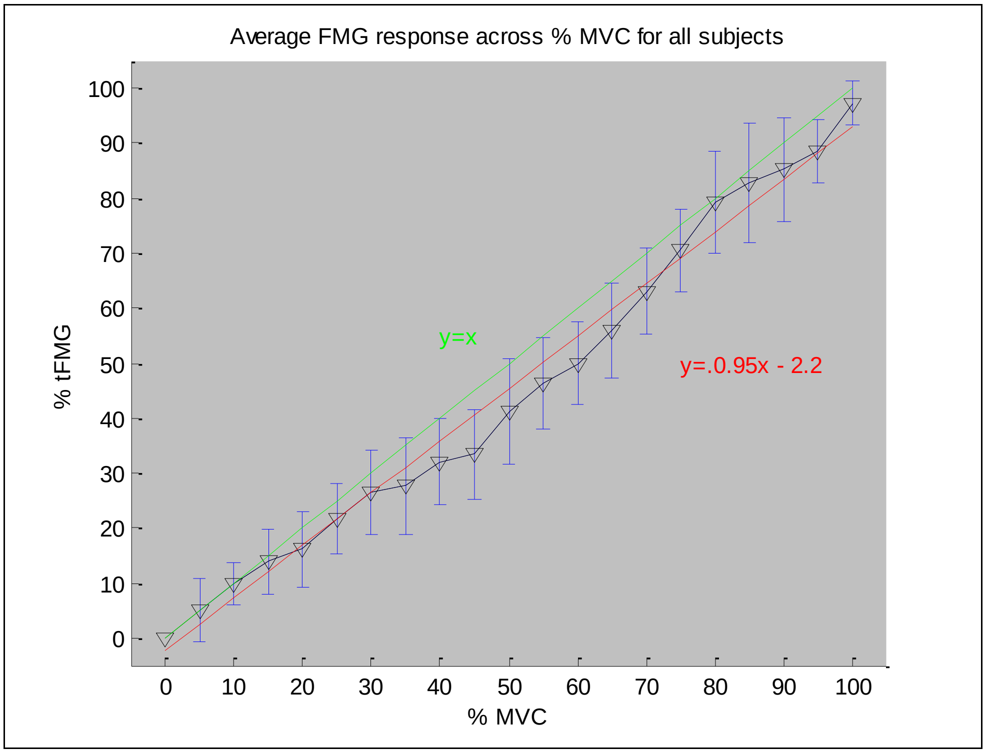


Figure 28. The moving average of all subjects' data reveals the behavior of the overall normalized data. The moving average follows the best fit line very well, staying within one σ at all times but the average of all the subject falls consistently below a pure linear relationship between % MVC and % tFMG.

In order to show the ability of tFMG of the FDS to control grasp strength the axis were reversed and linear best fit lines were then plotted for each subject's peak values (Figure 29). Flipping the axis shows the ability of a single muscle to predict % MVC, where in the case of prosthetic control, % tFMG is the independent variable. The individual slopes ranged from 1.21 to .87 with an average of .99 but the correlation of determination ranged from 0.90 to 0.98 with an average fit of 0.94 and a STD of 0.03. This indicates that although there may be a higher variation to the overall fit line of the FDS's behavior during isometric contraction, intra-subject repeatability is higher. This result is more important than overall fit; it indicates that this method could be used to create a fit line from a subject's perceived MVC and a minimal effort for real-time control of a prosthetic device.

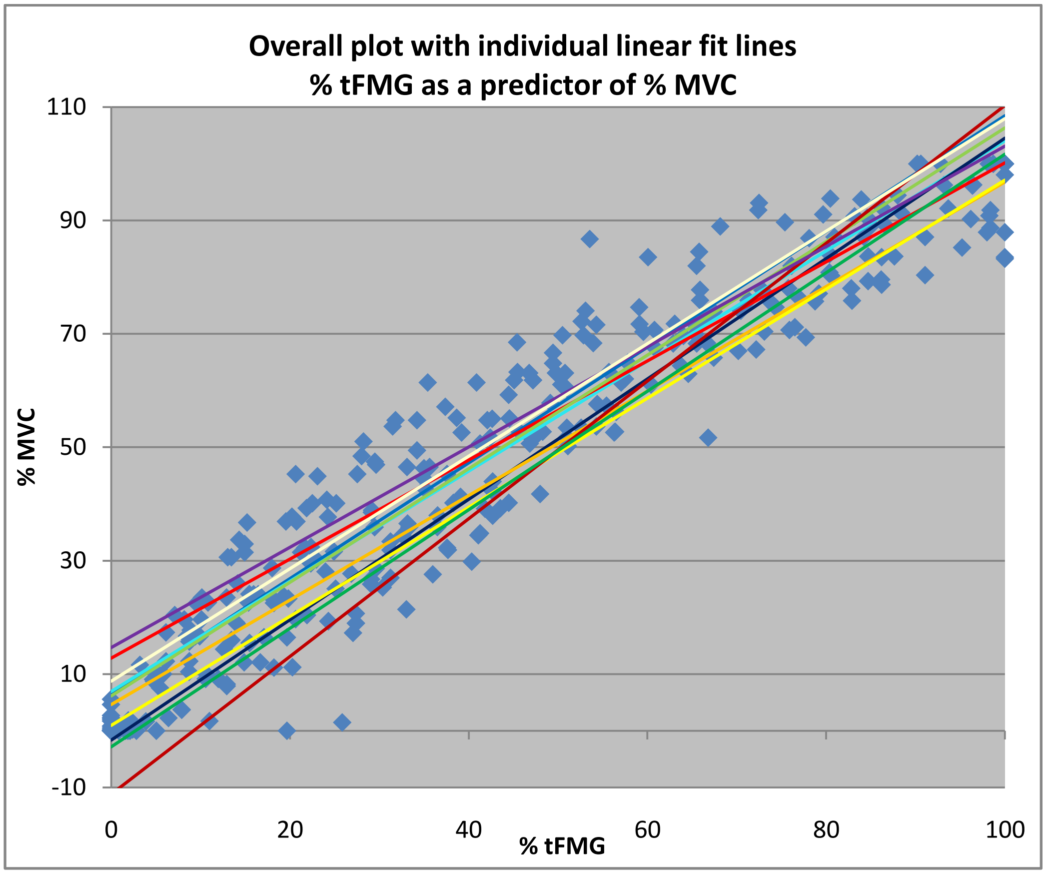


Figure 29. Best fit lines of individual subjects corresponding to figure 5. Though there is some variation in the slopes, the correlation of each subject to their own linear fit is high, 0.94 ± .03.

D. DISCUSSION

According to Redfern (1992), significant antagonist and synergistic activities occur during isometric force production and monitoring of these muscles activity is necessary for EMG estimation. The novel method of this study mechanically combines the change in radial pressure of the entire forearm, thus incorporating antagonists and synergists with the increased stiffness of the underlying muscle and soft tissue by the strap wrapped around the subject's forearm. If the sensor were placed on a different muscle or an area with underlying fatty tissue the sensor response would be different and not entirely related to the grasping strength.

Though no testing was done, the linear relationship between the normalized values of % tFMG and % MVC reported here is predicted to have a strong enough linear correlation to allow for the accurate control of the magnitude of volitional movement in a prosthetic device, extending this testing modality to a more refined multifunctional and multidimensional control. This can be stated despite the fact that muscle axial deformation in amputated limbs reveals a less smooth curve than healthy subjects (Zheng et al., 2006). Duque et al. (1995) reported a high correlation (.895) between % EMG activity and subject's expected muscle force, indicating relative levels of muscle activity remain consistent when a subject is predicting their level of force. This phenomenon indicates that a subject with connected muscles in an amputated limb would still retain the ability to maintain levels of volitional control based on relative levels of muscle activation.

Comparing the results reported here to EMG results of similar methods, it can be seen in Figures 4, 5, and 6 that the relationship is not purely linear and varies between subjects. Duque et al. (1995), reported a linear relationship between EMG activity of the FDS and % MVC. A definitive explanation is beyond the scope of this study, besides for errors in the testing methodology, such as misplacement, variation of placement, and sliding of the sensor or drift of the FSR force/voltage curve, there are some hypotheses based on previous research.

Variations of muscle composition is one hypothesis, Woods and Bigland-Ritchie concluded that the basis of the relationship between EMG and % MVC was muscle fiber composition, muscles with uniform muscle composition led to a linear relationship while mixed fiber compositions yielded non linear curves. Johnson et al. (1973) report a wide variation in the fiber type ratio across subjects for a given muscle, with no way of predicting how and when these muscle fibers will be recruited during activation. Related to that, Zheng et al. (2006) reported that muscles with smaller thicknesses deformed less and had a smaller change in muscle circumference to change in length during contraction. These variables may affect the results presented here but this study is not equipped to observe muscle thicknesses or muscle fiber composition *in vivo.*

A more likely explanation is that the variation in the profile of the curve is due to the behavior of the synergist and antagonist muscles. During activation the synergist and antagonist muscles are activated and contribute to the change in overall radial pressure of the forearm. Subjects, regardless of their overall strength, will have variation in the size and activation profile of these muscles. Mogk and Keir (2003) reported that below in low level grip force, 0-5% MVC, extensor activation was always higher than flexors; around 50% MVC the levels of activation became similar, and by 70% MVC and above flexor muscle activation always exceeded extensor activation. As was observed here, differences in activation patterns varied across subjects but were consistent and repeatable within the same subject (Buchanan and Lloyd, 1995). This selective recruitment patterning is most likely the reason for the variation across subjects.

The results reported here are not consistent with findings reported by Orizio et al. (1999) who studied the lateral displacement of muscle size *ex vivo*. They reported that the relationship of force output and lateral displacement can be split into two components; first a large geometrical change of the muscle at low forces followed by less lateral displacement and large force increments. These results are however taken from the lateral displacement of a single muscle *ex* vivo; the sensors in this study are recording pressure from the muscle combined with the total radial pressure of the entire forearm muscle complex.

The spacing on the Jamar hand grip was not varied between subjects, meaning that the subjects' relative grip size to hand length ratio and muscle length varied between users. Previously, handgrip forces were found to be dependent of grip width, where grip widths of 19.7% of the users hand length were accompanied with the maximum observed grip forces (Kong and Lowe, 2005). However, the grip diameter does not affect the results of EMG correlation to percent MVC (Hoozemans and van Dieen, 2005; Kong and Lowe, 2005) showing repeatable results between 25 and 75 mm grip diameters. This behavior is assumed to carry over to the results of tFMG.

Fatigue is not a concern in this study due to the high reliability and correlations of subjects' individual fit lines. Localized muscle fatigue is typically observable by loss of force production capabilities, localized discomfort, or pain (Chaffin, 1973). Between runs of grasping bursts subjects when subjects were asked to assess their own level of discomfort or fatigue and no subject reported any level of discomfort.

E. SUMMARY

In this Section it is demonstrated that the tFMG of the FDS recorded by a single FSR is adequate for determining the level of force activation. Although the variation may seem unacceptable when inter-subject data is compared, the more important behavior of intra-subject correlation is high. These results show that the improved testing methods are accurate and reliable but more importantly can be interpreted without knowledge of the actual grasping force, all which is required is an MVC and a very minimal contraction. The work presented here can easily be incorporated into future development of a multi-functional, multi-degree of freedom prosthetic limb with proportional control.

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