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Force sensing glove for quantification of joint torques during stretching after spinal cord injury in the rat model.

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FORCE SENSING GLOVE FOR QUANTIFICATION OF JOINT TORQUES DURING STRETCHING AFTER SPINAL CORD INJURY IN THE RAT MODEL

By

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ABSTRACT

An increasing amount of healthcare resources is used for the treatment and prevention of contractures in patients with spinal cord injury (SCI), with stretch and passive movements remaining the most prominent intervention methods. The results of both clinical trials and animal studies in recent years have shown traditional stretch therapies to be ineffective at preventing contracture and joint immobility, and have encouraged further emphasis on evidence-based practices. However, these studies only analyzed one aspect of stretching, dosage, and failed to look at the characteristic of joint torque. Recent clinical trials have unearthed the fact that the joint torque application of therapeutic stretches in the clinic not only vary by therapist, but also can be well beyond the range of torques tolerated by able-bodied individuals. A glove device utilizing force sensing resistors (FSRs) was developed to gauge joint torques. Coupled with a custom National Instruments' LabVIEW program, the device was able to accurately measure forces, and eventually torques, applied during stretching. This study sought to explain what range of torques were being applied during stretching after SCI in the rat model in the hopes of understanding how to administer safe, effective therapeutic stretches. Six adult female Sprague-Dawley rats were mildly contused at T9 using the NYU impactor device with a 12.5 g-cm weight drop. n=2 rats were stretched 2 days per week and n=2 rats were stretched once per week using an eight minute protocol, for the first 5 weeks post-injury while controls (n=2) received no stretch therapy. Briefly, the tibialis anterior (TA) and triceps surae (TS) muscle groups were stretched by two therapists bilaterally for a minute each, totaling 4 minutes of stretch per rat per day. Kinematic assessments of stretching were accompanied by force measurement data and were used to generate comparisons between therapeutic torque and end range of motion (ROM) of the ankle. The data suggests that both once and twice per week stretching regimens were not enough to inhibit locomotor recovery or elicit a noticeable change in end ROM in such a mild injury model. There were noticeable differences in torques applied during stretching by different therapists, confirming the findings of previous studies. More importantly, the data showed that immediately after injury the normal end ROM can be achieved by applying less torque. The torque necessary to reach the end ROM increases to baseline values by week 5, potentially due to a return of the stretch reflex during spinal shock. This study urges other aspects of stretching therapy to be considered and suggests a tool for therapists to quantitatively apply safe and consistent stretching therapies to patients.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

I. INTRODUCTION

A. Contractures and Passive Movement

An increasing amount of healthcare resources is used for the treatment and prevention of contractures in people with spinal cord injury (SCI), with stretch and passive movements remaining the most prominent intervention methods [1]. Contractures, or reduced joint mobility, are due to loss of extensibility in soft tissues spanning joints and are a common complication of spinal cord injury [2]. In fact, a study performed at Northwestern University in Chicago showed that spinal cord injured patients were afflicted with an average of seven contractures between 6 and 7 weeks after injury [2]. Loss of extensibility of soft tissues spanning a joint – including ligaments, muscles and joint capsules – has been a primary focus of therapists' and researchers' efforts to gain and maintain function in peripheral tissues and for neurological plasticity [3]. Contractures are undesirable for many reasons but primarily because they prevent the performance of motor tasks [4] and create unsightly deformities and are thought to predispose patients to spasticity, pressure areas, sleep disturbances and pain [1].

At the bare minimum, the development of contractures slows the rehabilitation process. At its extreme, contractures can drastically limit a patient's functional potential. Therefore, range of motion exercises are commonly initiated as soon as possible following SCI to prevent the development of contractures. Then during acute care and rehabilitation, the joint range of motion and muscle flexibility should be increased or at least maintained through positioning and daily exercises. There are various ways these range of motion interventions can be administered. For example, passive stretch can be applied using splints, positioning programs or orthoses [5]. When range of motion limitations are seriously impeding functional progress, these splints can be used to help maintain or increase range between therapy sessions [6]. However, when such devices are used, impairments in circulation, skin breakdown, and nerve damage are possible side effects. In this situation, constant evaluation for fit and proper fabrication must be performed [5]. A second approach is the application of passive and active movements using mechanical devices or manually by caregivers and therapists [1]. While stretch and passive movements are most commonly administered manually, this method limits the dosage of stretch and passive movements that can be realistically applied.

As a rule, maintenance of normal joint range of motion and muscle flexibility will enhance function. In the human, the range of motion requirements are well documented for the hips, knees, hamstrings, and ankles and are summarized in TABLE I.

TABLE I

COMMON RANGE OF MOTION REQUIREMENTS FOR HUMANS [5]

Joint	Ideal Range	Functional Significance
Hips	Full extension	Ambulation
	At least neutral extension	Prone lying and prone bed activities
	Normal or near normal flexion	Bed activities, dressing, wheelchair skills
	Normal or near normal external rotation	Dressing
Knees	Normal extension	Ambulation
Hamstrings	110-120° of passive straight leg raise	Long-sitting, mat mobility, floor to wheelchair transfers, coming to stand from the floor
Ankles	At least 10° of dorsiflexion	Ambulation
	At least neutral dorsiflexion	Prevention of pressure ulcers over metatarsal heads and toes

Intuitively, a program limited to strengthening and range of motion exercises will not result in the development of functional skills. The most important component of a functional rehabilitation program is functional training. For example, in human rehabilitation centers, patients are asked to walk on treadmills and stand for minutes on end [7]. In the rat model, simple in cage movement acts as functional training [3]. The purpose of functional training is the acquisition of certain motor skills that can later be utilized in environments other than the therapy gym [7]. By helping patients acquire functional skills that they can perform at home or in the community, therapists can enhance their capacity to return to their accustomed activities and roles. Functional training should be initiated as early as possible in the rehabilitation program and simultaneously with maintenance of strength and range of motion [5].

B. History of Passive Stretching

World War I served as the spark for the organization and management of spinal injuries. Due to the overwhelming numbers of casualties, established spinal units were finally developed. Each had a multidisciplinary team consisting of surgeons, neurologists, and urologists. Though there were constantly high mortality rates, a return to a patient's previous quality of life defined SCI treatments from the First World War through the Second [8]. The polio outbreaks of the 1950s and the associated symptoms of flaccid paralysis and muscle weakness shifted focus toward contracture management in the form of a standardized regimen of passive movements, two to three minutes of stretch every few hours, was recommended for people suffering from paralysis [9]. The rationale for such a recommendation was simple logic: if contractures result from the inability of individuals to move joints, surely therapists moving and stretching these joints will prevent contracture formation [1].

Based on animal studies led primarily by Dr. P.E. Williams in the United Kingdom, an appreciation for prolonged stretching administered through splints, orthoses, and the like developed [10, 11]. These studies demonstrated that in small animals, muscle extensibility could be lastingly improved via prolonged and sustained stretching protocols that stimulate the addition of muscle sarcomeres in series [1]. Whether or not humans responded to this type of stretch as well as the animal models remained shrouded in mystery. Given the lack of evidence to the contrary, it became common practice for therapists to recommended 20 to 30 minutes of sustained stretch per muscle per day for the prevention and treatment of contractures [1].

It is only in the last decade, after a shift toward evidence based practice, that therapists have begun to revisit the issue of contracture management in patients with SCI [12]. At the forefront of this issue are Lisa Harvey and the Rehabilitation Studies Unit at the Sydney School of Medicine. With concerns that the recommended duration of any intervention - five minutes, twenty minutes or even one hour - is not sufficient to attain a therapeutic effect, the dosage of stretching and the relative effectiveness of passive movements and stretches administered in different ways are under scrutiny [13]. Therapists' time and energy may be better spent focusing on the rehabilitation of functional tasks and activities of daily living that naturally place muscles in these lengthened positions [12].

A randomized controlled trial to attempt to demonstrate the effectiveness of passive movements in SCI patients was carried out by Harvey and colleagues in 2009 using the ankle as a model – one ankle was treated and one was untreated for each patient. Twenty people with tetraplegia were exposed to passive movements on their respective trial ankles for 10 minutes, 10 times a week for 6 months while the remaining control ankles were left untreated [1]. The results of this trial indicated that there was a small added benefit of applying passive movements for six months: on average the effect of the 260 10-minute treatments over the 6 month period was 4° [14]. However, what this study failed to bring to light was whether the therapeutic effects of passive movements could accumulate over time. Because 4° every six months is not a striking improvement for the time and energy it takes caregivers to apply passive stretches to patients, an accumulated effect could justify this intensive recovery technique. Moreover, a 10 minute daily stretching regimen, applied to every joint affected by tetraplegia, would be even more onerous for caregivers and therapists to perform as a long-term strategy.

Additionally, the previous study failed to investigate the effect of the dose of passive movement. Though 10 minutes of stretching per limb may seem excessive, would a smaller duration of stretching have similar effects? This daunting yet largely unexplored question could justify the time requirement of passive stretching as a potential therapy for tetraplegia. Two more randomized controlled trials [13, 15], again led by Harvey, essentially demonstrated that passive movement interventions were ineffective when compared to no intervention or usual care. The first study stretched each weekday for 30 minutes for 4 weeks and analyzed ankle mobility in patients with recent spinal cord injuries. However, the stretching was done with a specified torque, 7.5 N-m [13]. The second study followed the exact same timeline but looked at the extensibility of the hamstring muscles in people with recent spinal cord injuries – again with a constant torque of 30 N-m at the hip [15]. The results from both trials showed that the stretching intervention did not significantly change the ankle or hamstring extensibility, respectively, in patients with recent spinal cord injuries [13, 15]. A similar study demonstrated a small effect on ankle mobility after 30 minutes of tilt-table stretching three times per week for 12 weeks [16]. All these studies clearly indicate that minimal stretch intervention – anything applied less than 30 minutes a day over less than 3 months – will not induce clinically meaningful change in joint mobility or muscle extensibility [1, 13, 15, 16].

Optimal dosage has never been clarified, although historically dosage has been based on clinical experience, anecdotal evidence and animal studies [1]. Most human trials investigating stretch in people with SCI have only examined the added benefit of stretch over and above usual care provided to both experimental and control groups. In these studies passive movements may have been administered unintentionally as patients moved during daily activities or rehabilitation programs, such as change in position for skin care and practice of functional activities. Therefore, the findings of these randomized controlled trials in people with SCI must be taken with a grain of salt; though they indicate that stretch as typically applied by therapists does not produce lasting increases in joint mobility, stretch applied as part of usual care may or may not have any effect on joint mobility.

In summary, the results of clinical trials in recent years have shown that traditional stretch therapies are mostly or completely ineffective at preventing contracture and joint immobility, and have encouraged further emphasis on evidence-based practices. In humans, the evidence to date indicates that stretch applied for less than 3 months confers little or no added benefit over and above usual care in people with SCI [16]. There is a small benefit from passive movements when applied for 6 months but it is unclear whether this benefit is clinically worthwhile. For periods lasting longer than 6 months, the effects of stretch remains unknown [12].

In the rat model, it has recently been shown that 8 weeks of a daily 30-minute protocol of bilateral hindlimb passive stretch of the ankle, knee and hip negatively influenced the normal course of recovery for rats mildly contused at the 10th thoracic level (T10) [3]. A second study found the same thing to be true for animals with moderate T10 contusion injuries. As both of these studies modeled their stretching protocols after stretching observed in a clinical setting of SCI management, these findings question whether or not repeated stretching – commonly used to maintain muscle and joint integrity – may conceal or delay functional locomotor recovery in a model that consistently shows substantial recovery by 6 weeks post-injury [3].

When stretching began within four days of SCI (acutely) and occurred daily for 8 weeks, locomotor function was severely limited in the first 5 weeks then plateaued to below average levels with continued daily stretching during weeks 5-8. Moreover, significant deficits were evident even weeks after the stretching regimen was terminated. When stretching began in the chronic phase 10 weeks post-injury it caused a significant, but temporary decline in function and did not affect the animals in the following weeks [3].

K. Caudle hypothesized that the reason the stretching protocol brought about a functional plateau earlier than observed in the non-stretched control animals was not because of changes in the muscle physiology, necessarily. The laboratory of JW Grau has demonstrated that randomly applied, noxious inputs can disrupt spinal cord plasticity in the fully transected rat preparation [17, 18]. They also uncovered evidence that only 6 minutes of uncontrollable noxious input, administered via tail shock, can negatively alter locomotor recovery in a contusion model of SCI [19]. Therefore, Caudle proposed that the most practical explanation is that the stretching protocol was activating barrages of afferent activity that disrupted lumbosacral circuitry and plasticity [3].

With human data pointing toward little or no added benefit of prolonged stretching, and animal data indicating that passive movement intervention in fact is detrimental to recovery, current therapeutic practices should rightfully be under scrutiny. However, there still remains another aspect of passive stretching that has yet to be seriously investigated. This second variable in passive movement, other than duration, is intensity, measured by analyzing joint torques.

C. Efforts to Quantify Joint Torques

Efforts to quantify the forces and torques applied during stretching have been few and far between. While it is beneficial, for sake of variable elimination in studies, to apply the same torques to all patients during stretching after SCI, it is another matter entirely to look at what stretching torque is most effective. Beginning in 1998, A. Lamontagne began to examine and quantify the many aspects of stretching after SCI. Spastic hypertonia (SH) is a motor disorder characterized by a velocity-dependent increase in muscle tone with exaggerated tendon jerks, resulting from hyperexcitability of the stretch reflex [20]. Since the response to muscle stretching has been shown to be velocity dependent [21], applying muscle stretches at different velocities allows for the differentiation of muscle resistance that originates from the reflex components from the resistance that originates from the non-reflex components [22].

Spastic hypertonia and the reflex-induced responses to stretch can be assessed using the modified Ashworth scale [23]. However, measurement of the non-reflex component requires a quantitative measurement of the resistive torque during muscle stretches that are performed at a velocity that will not elicit a reflex [22]. The method used by Lamontagne et al. employs a handheld dynamometer, a device capable of measuring torque utilizing force transducers, in the first instance where force transducers are used to assess stretching torques after SCI in relation to ankle position.

In this study, the ankles of nine adults with SCI (average age: 40.6 ± 10.5 years) were stretched into dorsiflexion 5 times at two preset velocities, 5°/s and 180°/s, and ankle position, torque, and EMG were recorded. The results indicated that at both high and low stretching velocities, the handheld force transducers yielded reliable intertrial measurements of resistive torque. Every patient has a distinct velocity that elicits SH, so by adjusting the velocity of the passive movement of the dynamometer, the authors could measure resistive torque that reflected both the reflex components and the non-reflex components of SH, a feat previously impossible with only the Ashworth scale [22]. Without delving too deeply into their analysis of torque and ankle position, the handheld dynamometer recorded a set of 5 passive ankle dorsiflexions at low velocity (FIGURE 1). The arrows indicated where resistive torque and velocity measures were obtained and show that as the ankle approaches full dorsiflexion, the torque rises to about 4 N-m. While the dynamometer needed improvement, this study provided excellent groundwork for further studies on the quantification of stretching.

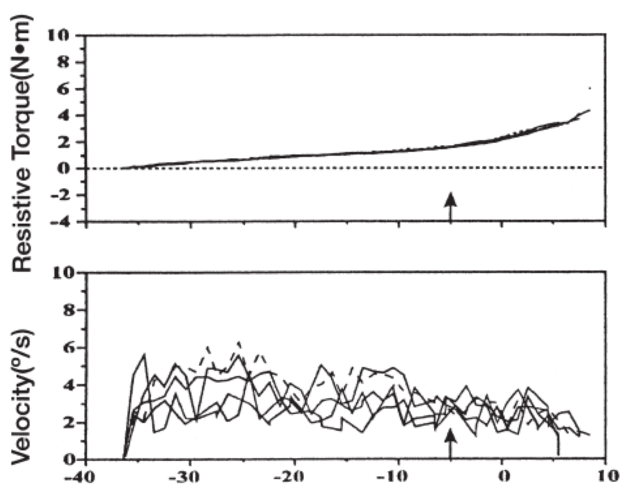

FIGURE 1 - Resistive Torque of the Soleus Muscle and Angular Velocity [22]

About a decade elapsed before another group tackled the issue of comparing torques of passive movement with joint angles. In the meantime, devices were being developed to accurately administer exact, repeatable torques during stretching. Harvey et al., in 2000, refined a previous device to administer a constant stretch each day for 30 minutes by rotating the ankle into dorsiflexion with the knee extended. Patients were either supine or seated in their wheelchairs during stretching. The device used for stretching (FIGURE 2) consisted of a footplate that rotated the ankle in a sagittal plane. A rope attached to the end of the footplate was looped around the rim of a wheel of 15-cm radius and passed through a pulley. A 5 kg weight was suspended from the end of the rope, creating a constant torque of 7.5 N-m (regardless of ankle angle) that rotated the footplate and foot into dorsiflexion. The small torques associated with the weight of the stretching device and foot were ignored [13].

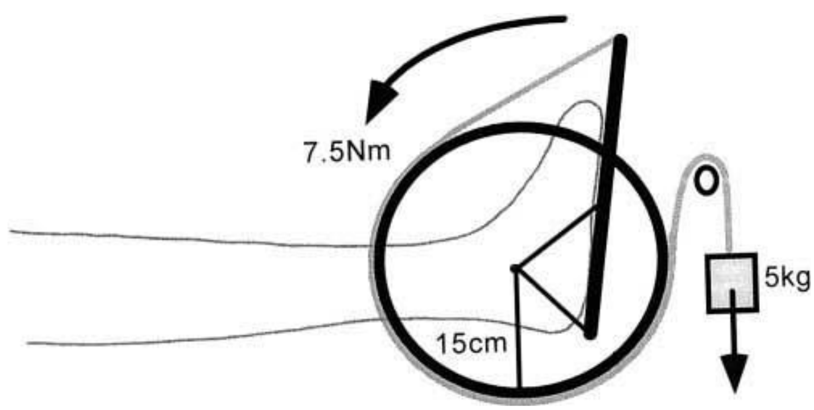

FIGURE 2 – First Constant Torque Stretching Device [13]

Finally, in 2003, Harvey et al. conducted a study that addressed the question “Do therapists administer too much stretch to insensate patients?” by focusing on hamstring extensibility. Therapists commonly apply hamstring stretches by manually holding the knee in extension while moving the hip into flexion. However, the amplitude of the stretch torque is not normally measured because it is difficult to determine how much force the therapist applies, the angle of the hip, and the weight of the leg [24]. When applying stretch to able-bodied individuals, patients can provide verbal feedback regarding degree of muscle stretch and discomfort to ensure that therapists do not utilize excessive stretch torques. But when therapists shift to stretching insensate individuals, they are only guided by their own subjective feel of tension in the patient's hamstring muscles and their own interpretation of what constitutes an "effective" stretch. Excessive stretch, apart from obviously damaging muscle and connective tissues, has been linked to the development of heterotropic ossification, the formation of bone tissue outside the skeleton [25], via skeletal muscle inflammation and articular ossification [26]. Since it is not known how much stretch therapists apply to these insensate patients during a typical stretching session, the aim of the study was to quantify the torques applied by therapists to patients with SCI who are unable to feel muscle stretch and discomfort and to use existing data to determine whether therapists administer stretch torques equivalent to those tolerated by able-bodied individuals [24].

In this study, fifteen patients with either paraplegia (n=7) or tetraplegia (n=8) were given a hamstring stretch by no less than 10 physiotherapists. Hamstring stretches were applied by moving the left hip into flexion while the knee was maintained in extension with a splint [24]. It has been shown that during the first few minutes of stretch, there is a gradual increase in muscle extensibility, or creep [27]. Therefore, to eliminate any possibility of creep in the muscle, a stretch torque was applied to the hamstring muscles for 5 minutes. After this initial stretch, each therapist was instructed to administer a "therapeutic" stretch, without any feedback from the patient.

Torque-angle curves were generated by manually measuring passive hip flexion with the application of a series of standardized torques with a bed-mounted device, as shown in FIGURE 3. Briefly, a leg splint was fixed to a wheel and the wheel mounted on the side of the bed - the purpose of the leg splint was to prevent knee flexion and hip abduction and the wheel to maintain a constant moment arm. Then, increasing torques were applied to the hip flexor by progressively hanging 2.3 kg weights from the rim of the wheel, where each 2.3 kg weight generated an additional 6.35 N-m of hip flexor torque. Hip flexion angle was measured at each increment [24].

FIGURE 3 – Bed-mounted Stretch Device [24]

FIGURE 4 represents the torques applied by the therapists to each subject. The data is presented as ranges, 5^{th} and 95^{th} percentiles, quartiles, and medians. The median stretch torque applied to each subject ranged between 30 and 68 N-m. Compared to the range of torques generally tolerated by sensate individuals, 20 to 60 N-m [28], those values are slightly high. Even more, some therapists applied more than 120 N-m of stretch torque to some subjects [24]. The data is indicative of the fact that therapists apply widely differing torques when stretching the hamstring muscles of patients with SCI, well past the range that would cause pain in able-bodied individuals.

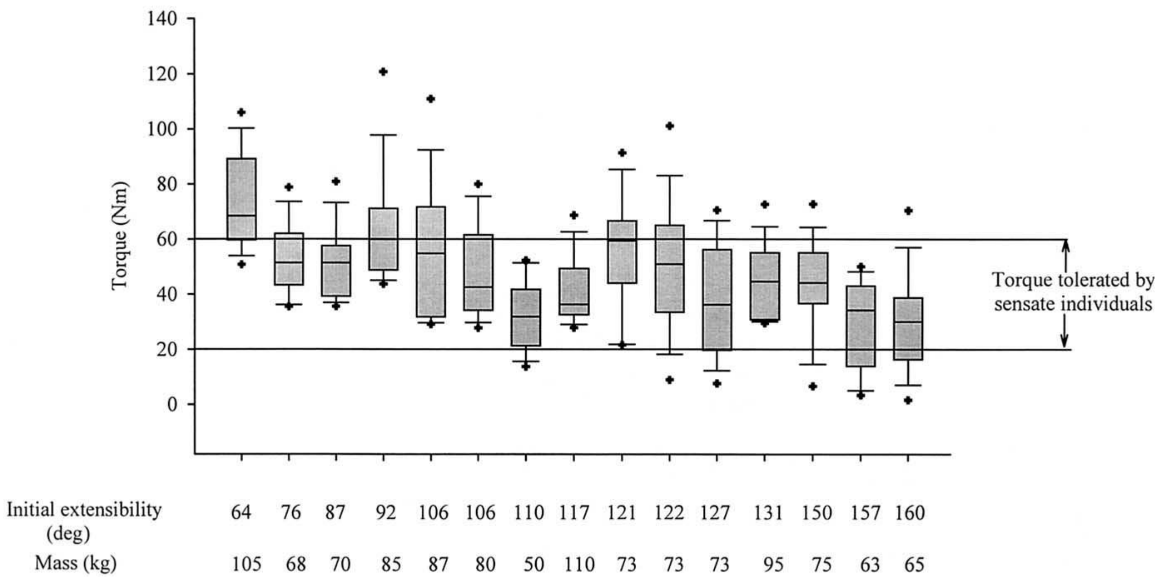

FIGURE 4 – Torques Applied During Stretching in the Hamstring [24]

The results also showed that the torque applied by each therapist was related to both the subjects’ mass and hamstring extensibility. However, the stretch applied by different therapists to any individual subject varied by up to 40-fold. The only explanations these dissimilarities are a lack of consensus about optimal stretch torque and the therapists’ difficulty gauging the magnitude of the stretch torque applied, especially considering that optimal torque varies by individual.

Without the possibility of feedback from the patient about feelings of discomfort, therapists must rely on their own discernments of the stretch applied to the muscle. If therapists are going to avoid excessive stretch and possible injury, they need to either routinely measure or be trained to better estimate the stretch torque they apply.

Considering that the therapists involved all had a minimum of 3 years of clinical experience [24], the former option seems most effective. Until an optimal torque is found, it would be safe to assume that therapists should avoid stretching torques that are normally associated with pain in sensate individuals. This request calls for the need to measure stretching torques as they are being applied.

D. Evolution of Glove Design

In 1997, a group in Brazil was attempting to achieve stable grasp force control while performing neuromuscular electrical stimulation (NMES) in patients with quadraplegia after SCI at the C5 and C6 levels [29]. Force control depends on closed-loop control performance, which was previously limited by the practicality of commercially available sensors. The main criterion for this system was that the sensors should not affect the normal behavior of the biological system, or should not interfere with the actual object manipulation. Therefore, light, small, easily mountable, inconspicuous sensors were desired that did not limit the range of movement of the limb. The sensors also needed to be unhindered by frequent calibrations and unaffected by magnetic environments, electrical noise, and shifts in temperature and humidity. The proposed solution mounted force sensing resistors (FSRs) – made by Interlink Electronics - to the distal phalanxes of the thumb, index and middle fingers in a commercial Lycra glove. The sensor readout (40 Hz) was amplified, filtered, and digitized through an analog-to-digital converter. For calibration, a polynomial characteristic curve was generated comparing voltage output and applied force in the range of 0 – 15 N which showed that there was minimal hysteresis in the system and that the unloading cycle values fall within 6% of the loading cycle measurements [29]. Using the idea of a glove design with minimally invasive sensors, further research sought to identify sensors with more promising characteristics for the application of measuring forces in the 0-1 N range.

Similar to the previous study, a group at the Center for Automation and Robotics in Madrid aimed to show the capabilities of contact force sensors in estimating forces exerted by a person in real and virtual manipulation tasks [30]. The authors encountered a similar predicament as the present study as they selected the appropriate sensor for their application. The preferred sensor type would be a force/torque transducer or a load cell [31]. Though these devices are very precise, they have severe inconveniences in their weight, physical size, and high cost. Therefore, the authors chose the FlexiForce model A201 sensor by Tekscan Inc. (Boston, MA) [30], as contact force sensors and piezoresistive sensors have been shown to provide an effective solution for applications involving the measurement of manipulation forces [32-34].

The design utilized by Ferre et al. [30] included four piezoresistive sensors - sensor 1 provides normal forces directly applied by the fingertip, and sensors 2, 3, and 4 estimate the tangential forces (FIGURE 5).

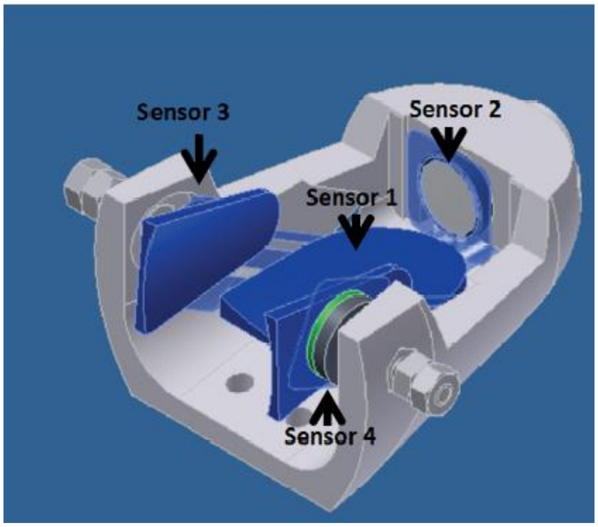

FIGURE 5 – Four Sensor Thimble Interface [30]

Based on manufacturer recommendations and sensor design, to obtain the best sensor repeatability the applied force must be homogeneously exerted over the active sensing area. Therefore, a “sandwich configuration” assembly was used to evenly transmit forces to the sensor’s active area. The sensor was placed in the middle of two cylindrical metal sheets that are located over the active sensor area, as shown in FIGURE 6. This “sandwich configuration” – commonly referred to as a puck or shim – guarantees mechanical isolation between the finger and the thimble, since the user force is thoroughly transmitted to the sensor [30].

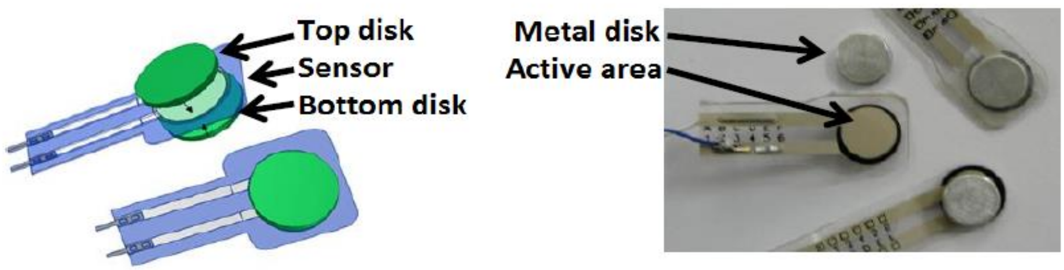

FIGURE 6 – Puck/Shim Sandwich Configuration [30]

To test the effectiveness of the sensors, calibration was performed with a high accuracy six-axis force sensor manufactured by ATI Industrial Automation (Apex, NC), model Nano17. The difference between the data provided by both sensors (thimble and ATI-Nano17) was approximately ±1.43 N with a range of 0 to 20 N [30]. In theory, if this range were scaled down to 0-1 N by altering the drive circuit, the difference would only be about ±0.0715 N, the equivalent of 7 grams of weight.

With this sensor type in mind – an accurate, reliable force sensing resistor (FSR) manufactured by Tekscan, Inc. – research was conducted to build a suitable device that met the design considerations outlined in TABLE II.

TABLE II

DESIGN CONSIDERATIONS

Glove Considerations
1) Fit multiple therapists hands
2) Made of a durable, washable material
3) Include attachment points for 4 removable sensors

Sensor Considerations
1) Low hysteresis
2) Accurate in the range of 0-1 N
3) Commerically available, cost-effective sensors only
4) Non-intrusive size and shape
5) Robust design capable of animal interaction

E. Specific Aims

1. Design a force sensing glove based on the following considerations:
    a. Glove design must be compatible for varying hand sizes
    b. Glove must not affect stretching technique
    c. Force sensing device must be economical enough to allow four sensors to be placed on the fingertips of one hand: thumb, index, middle, and ring fingers
    d. Force sensors must be accurate in the range of 0-1 N
    e. Force sensors must have physical properties that do not interfere with stretching
2. Code a custom LabVIEW program to filter, calibrate, display and record applied force based on the change in resistance of the FSR
    a. Program must be able to record numeric and video data
3. Modify the standard stretching protocol that keeps constant the following:
    a. Hindlimb and body position in all aspects of the stretch

    b. Relative force application
    c. Hand position relative to the two cameras recording 3D kinematics
4. Develop a study that quantifies the torques generated during acute stretching after SCI using the lab's previous stretching protocol
    a. Calculate joint torques
    b. Record range of motion data
    c. Record BBB data
5. Develop a surrogate hindlimb for stretching studies and further characterization of the force sensing glove

II. INSTRUMENTATION AND EQUIPMENT

A. Hardware

Force sensing resistors (FSRs) were chosen for this application for their easy integration, cost effectiveness, and repeatability. As previously discussed, potentially more suitable options include load cells and strain gauges [35], however the physical and cost restrictions these types of sensors would place on a glove design dramatically outweigh the increase in measurement accuracy.

The principle behind FSRs is quite old, dating back to use in the first microphone. Using the example of conductive foam sandwiched between two contacts, when there is no force applied to the foam, there are voids in the foam, leading to a poor connection between the two contacts and thus a lower conductivity. An increase in force condenses the foam, resulting in a larger contact area and more efficient routes for electricity to flow, and the conductivity increases; as it turns out, over a wide range of forces, conductivity is approximately a linear function of applied force. Modern, commercial FSRs use a resistive polymer to obtain this effect and, because of the excellent compression properties of the polymer, they achieve a much more uniform change in conductance with force [36].

FSRs are constructed of two layers of flexible substrate film, and in the case of Tekscan's sensors that substrate is polyester. On each flexible layer, a conductive silver layer is added, followed by a layer of pressure-sensitive ink. Then the two layers of substrate film are laminated together using adhesive to form the final force sensor, as shown in (FIGURE 7) [37].

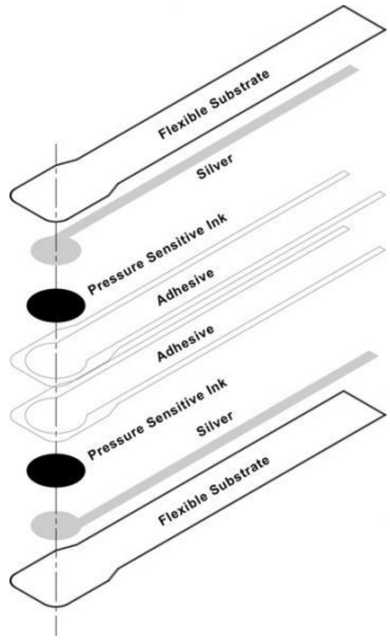

FIGURE 7 – FlexiForce Sensor Construction [37]

Three different sensors were evaluated based on the following criteria: accuracy of the pressure sensor at low pressure (0-0.3 N), total pressure range, linearity, hysteresis, diameter of the pressure sensing area, cost, length of the sensor, and durability. These characteristics were evaluated using a decision matrix (TABLE III, below). For each criterion, the sensors were ranked 1, 2, and 3, with 1 representing the sensor with the best characteristic and 3 the worst. The sums were taken for each sensor and the sensor with the lowest value, the FlexiForce A301 sensor by Tekscan Inc., was selected for use in the final glove design. Range, hysteresis, sensing area, cost, and length were all evaluated based on the manufacturers' specifications, a summary of which can also be found in TABLE III.

TABLE III

DECISION MATRIX - FSR SELECTION

	Range	Linearity	Hysteresis	Sensing Area	Cost	Length	Durability	Sum
FSR400	3	3	3	1	1	2	3	**16**
A201	2	2	2	2	2	3	2	**15**
A301	1	1	1	2	2	1	1	**9**

	Range (N)	Hysteresis	Sensing Diameter (mm)	Cost	Length (in)
FSR400	0.1-10	10%	7.62	$5.95	15
A201	0-4.4 N	<4.5%	9.53	$19.95	6
A301	0-4.4 N	<4.5%	9.53	$19.95	2

A separate test was performed to confirm the linearity of each sensor in the range of 0-300 grams (~0.3 N) of force, as this is the range in which the sensors will most often function. Briefly, each sensor was connected to the same two terminals of our NI-DAQ system and resistance, and thus conductance, was measured using the LabVIEW program. A uniform puck covered each sensing area to ensure that each sensor captured the entirety of the applied load, since the contacting surface was larger than the sensor diameter. Forces were applied using 0, 100, 200, and 300 grams of weight, and 600 data points were taken during each of three trials. These 1800 values were averaged to determine the conductance of each sensor at each of the four loads, conductance versus applied force was plotted (FIGURES 8-10) and the coefficient of determination (R^2) was calculated for each linear best-fit curve. The four R^2 values of each sensor contributed to the ranking of linearity, with R^2 values closest to one being the most preferred.

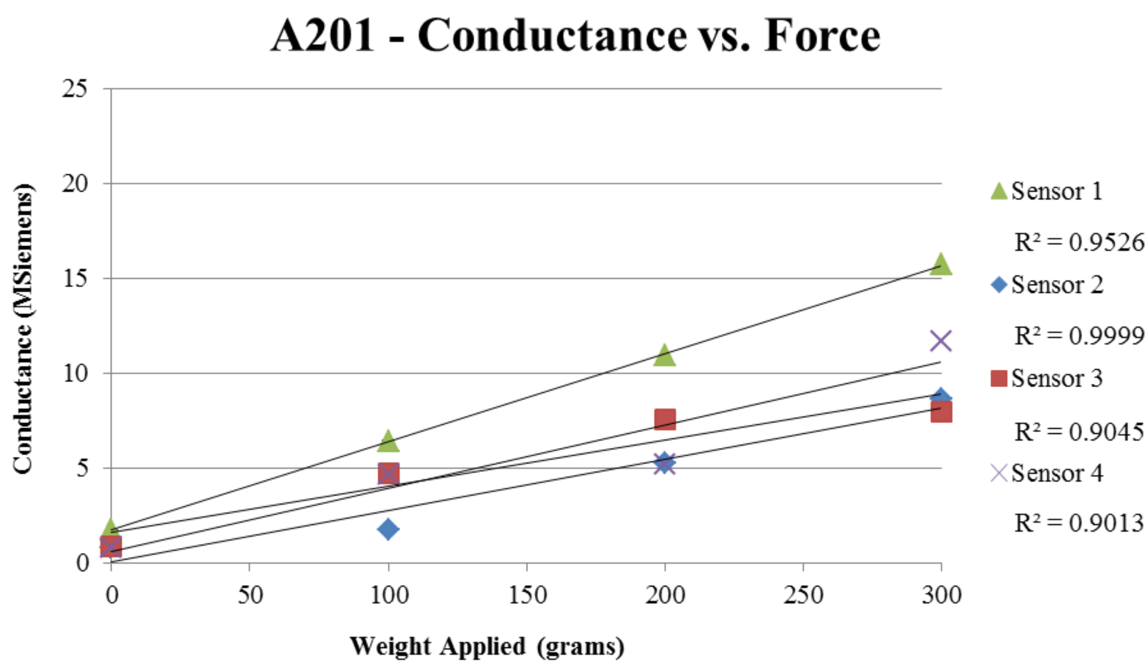

FIGURE 8 - A201 Sensor Linearity Plot

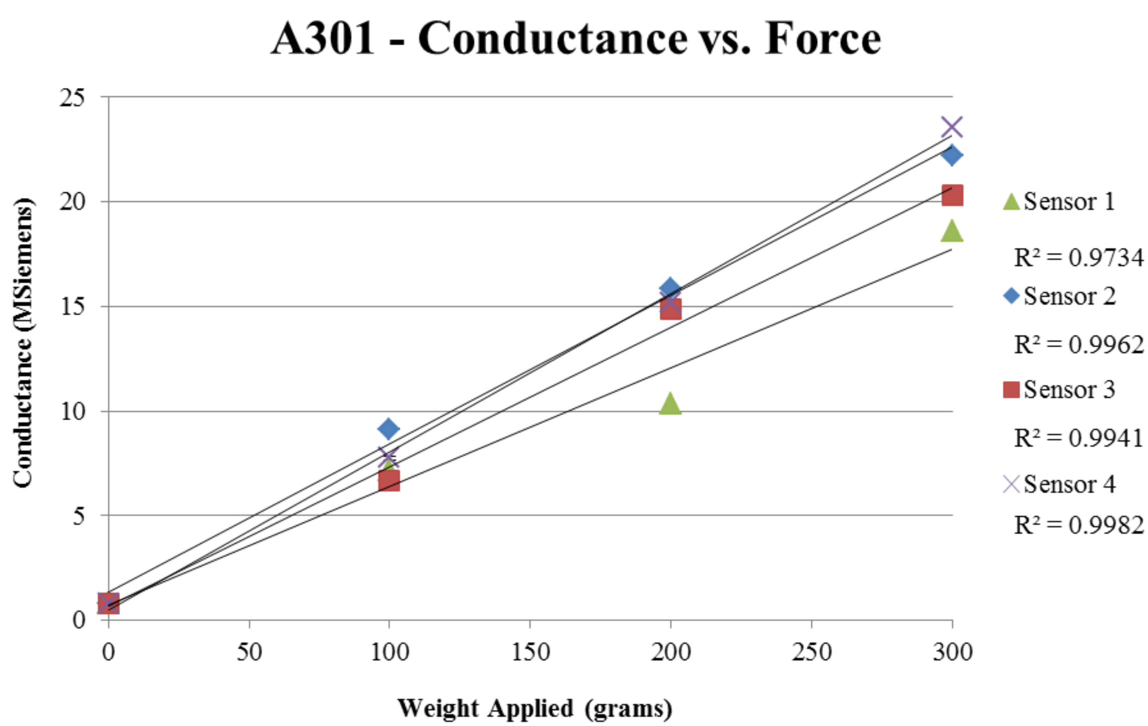

FIGURE 9 - A301 Sensor Linearity Plot

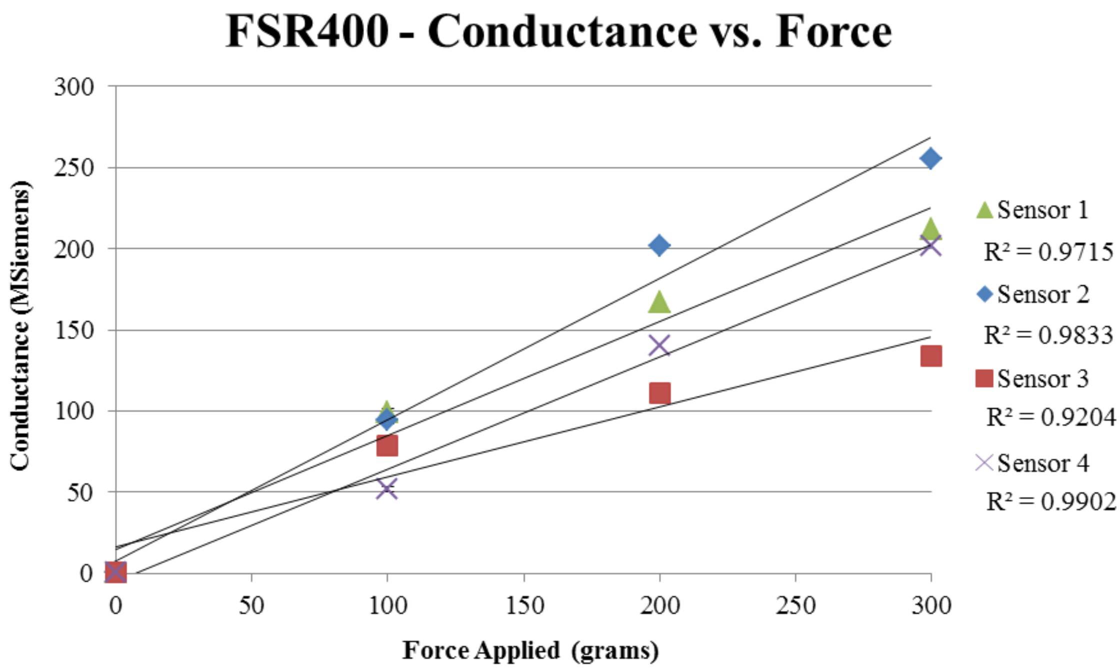

FIGURE 10 - FSR400 Sensor Linearity Plot

Fabrizio Vecchi and others at the Advanced Robotics Technology and Systems Laboratory in Pisa, Italy echoed these findings when they compared FSR sensors from Interlink Electronics (Camarillo, CA) and the FlexiForce sensors by Tekscan Inc. while evaluating force sensors for biomechanics and motor control applications [38]. Interlink FSRs come in a number of shapes and sizes, are inexpensive and robust, but they tend to experience noticeable drift over time and have a larger response time. For applications where the measurement of exact forces is not necessary, these FSRs would be well suited [39]. FlexiForce sensors, on the other hand, are more accurate and repeatable – they rely on the enhanced piezoresistive effect of the pressure-sensitive ink in the sensor – though they are a bit more expensive and delicate [38].

Because these FSRs change conductance linearly with force application, by measuring the exact conductance at the time of force application we can accurately quantify the force applied. To measure the resistance and conductance, the sensor was integrated as the first of two resistors in a voltage divider circuit, as seen in FIGURE 11.

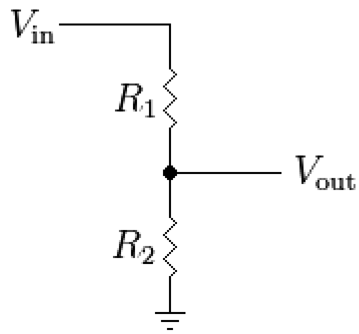

FIGURE 11 – Common Voltage Divider Circuit

By measuring the voltage going into the sensor and the voltage coming out of the sensor, the exact resistance (and thus conductance) of the FSR can be calculated based on the fact that R_2 = 100 kΩ in the formula:

$$R_1 = \frac{R_2 * V_{IN}}{V_{OUT}} - R_2 \quad (1)$$

The circuit that incorporates all four sensors can be found in FIGURE 12. Tekscan, Inc. has published a recommended drive circuit that utilizes a constant supply voltage and the operational amplifier (OpAmp) MCP6004. While the circuit Tekscan, Inc. recommends would be extremely successful in powering one sensor, it did not work while powering multiple sensors.

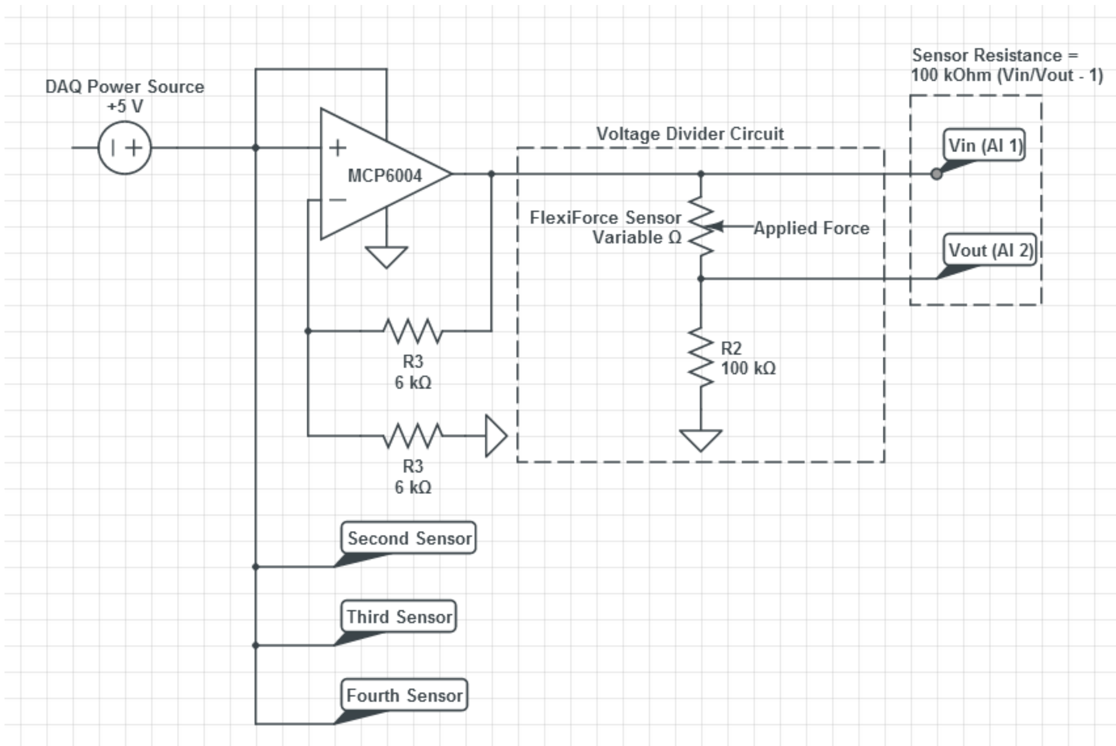

FIGURE 12 – Drive Circuit for All Four Sensors

Because each FSR changes resistance, the amount of current each FlexiForce sensor requires changes as well. When four sensors are being powered by the same +5V power source, a change in one sensor ripples through all three sensors because the amount of current going to the other three sensors is fluctuating. Therefore, a means of isolating the power sources for each sensor was required. To accomplish this task, the MCP6004 OpAmps were placed before each FSR in the circuit - rather than after, as recommended by Tekscan's drive circuit - and four OpAmps were used to effectively isolate the supply voltage of each sensor. Based on the MCP6004 manufacturer's instructions, the OpAmps (FIGURE 13) were provided with a gain of 2 using Formula 2 and resistance values $R_1 = R_2 = 6\ \mathrm{k\Omega}$:

$$\text{Gain} = 1 + \frac{R_1}{R_2} \tag{2}$$

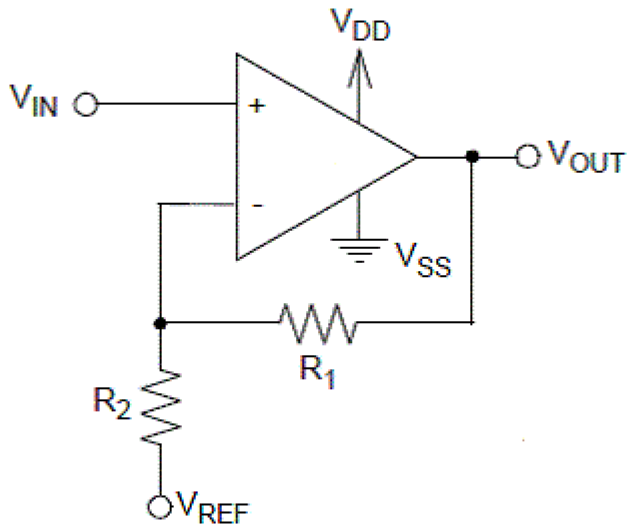

FIGURE 13 – MCP6004 Schematic

As stated earlier, it was necessary to mechanically isolate the interaction between the stretcher's fingertip and the sensor. To get any semblance of repeatability from this sensor type, a puck was developed and refined through multiple iterations. Based on work done by F. Vidal-Verdu in Spain, the first puck consisted of a polyurethane cone, as shown in FIGURE 14 [40]. This cone had a base diameter equal to that of the sensing area (9.53 mm) and a top diameter roughly half that size (4.76 mm). It was noted that when stretched was applied to a rat the point of force application was not wide enough to cover the entire sensing area. Because homogenously applied force was the goal, it was necessary to distribute the small contact area to the larger sensing area; therefore, a slightly deformable cone provided a firm contact point and a means of accurate pressure distribution. Though the concept behind this design was well grounded, there was difficulty affixing the cone to the sensing area, so an alternative design was proposed.

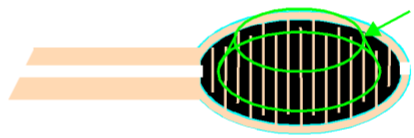

FIGURE 14 – Initial Puck Design

The alternative design utilized the same polyurethane as the cone. However, this design consisted of a flat polyurethane back and front attached by a hinge that rested at the top of the sensor. This puck joined together two flat surfaces, ensuring homogenous force application. The only downfall of this design was that the hinge placed a very small amount of resting force (<10 grams) on the sensor at all times, offsetting the force measurements. One final iteration was taken to solve this issue, adapted from the thimble design discussed earlier [30, 34]. Simply, two metal plates were cut to exactly cover the sensing area on the front and back of each sensor, as shown in FIGURE 6 [30]. This sandwich configuration was held together using Gorilla glue and was the most accurate and least intrusive design.

A primary design consideration for the glove apparatus was that the glove be unobtrusive to its operator. That is, the glove and the sensors should minimally interfere with the manipulation required for stretching. Therefore, the base of the glove was created out of a regular white cotton and spandex fabric. The flexible fabric allowed multiple handlers to wear the glove and the white color provided excellent contrast for computer digitization purposes, to be later discussed. The sensors needed to be both removable and interchangeable, so Velcro strips were fixed to the backs of each sensor's puck configuration and larger Velcro strips were sewn where the distal phalanxes of the thumb, index and middle, and ring fingers would be positioned. Thus, the sensors could be positioned anywhere on the fingertip and readily removed for calibration, glove washing, or sensor exchange.

B. Software

Off the shelf systems for reading and logging force measurements are available, however no company offers a truly customizable system suitable for this particular application; the closest available solution is the ELF (economical load and force measurement) system by Tekscan Inc., though the lack of modification options and $950 price tag make this system undesirable. Therefore, a hardware and software interface was developed to allow for complete customization of all components of the system, including sensor selection, sensor range, data collection frequency, coupled video and force recordings, and filtering. Partnering National Instruments' (Austin, TX) USB-6210 data acquisition system with NI LabVIEW provided a simple, powerful, mobile solution for the hardware and software interface. A custom LabVIEW program was developed and refined to collect voltage data from both the input and output of each sensor, as well as collect two video sources for use in 3D kinematics. Based on these voltages and the fact that each sensor acts as a variable resistor in a voltage divider circuit, the program calculates the real-time resistance and conductance of each sensor.

The operation of the program is simple to understand. Once the program is executed, the DAQ device is immediately reset, clearing all previously logged data. The program requests that each sensor be independently calibrated with four user-defined weight values - the default being 0, 100, 200, and 300 grams - and generates a linear calibration curve that correlates each sensor's conductance with applied force, or weight. Once calibrated, the program reads and graphically and numerically displays the weight applied in one of five units: weight (g), force (N), resistance (kΩ), conductance (μS), and output voltage (V) (FIGURE 15). When the record button is selected, the program begins to simultaneously save a spreadsheet containing the force on all four sensors and two separate AVI video files.

FIGURE 15 - Custom LabVIEW Graphical User Interface

The “Settings” tab of the program allows the user to customize numerous aspects of the sensor function (FIGURE 16). The user can select the number of points to be averaged for the calibration curve, the standard weights used for the calibration curve, as well as the data collection frequency (Hz). Most importantly, though, the user can save calibration data to use from one session to the next with the “Keep Values” and “Manual

Override" buttons, allowing the same calibration to be used for each animal during a given stretching session.

FIGURE 16 – Settings of the Glove Program

III. PROCEDURE

A. GL-AN Study

The GL-AN study, short for Glove-Anesthetic, was the pilot study conducted to test the efficacy of the glove interface during stretching. Before this study, the operation of the glove and sensors were tested and characterized (as described in the Instrumentation section), however this was the first study involving live animals.

In this study, two female, uninjured, adult Sprague-Dawley rats (225-300 g) were used. They were gentled for a week (Week 0) before stretching would commence. Gentling simply acquaints rats to being handled by humans, and was performed every day for five days. Each rat was held, swaddled in a towel to simulate the stretching environment, and gently handled for 30 minutes twice a day, totaling an hour of gentling per rat per day.

Following gentling, two weeks of stretching were performed under varying anesthetics. Three anesthetics were used: a ketamine-xylazine combination in a dosage of 75-100 mg/kg and 5-10 mg/kg for ketamine and xylazine, respectively, injected intraperitoneally (IP); isoflurane gas administered as an inhalant; and Nembutal (sodium pentobarbitol) administered IP at a dosage of 40-50 mg/kg. Rats were given dosages based on their body weight to come as close to a surgical plane of anesthetic as possible.

Two stretches were performed that isolated the tibialis anterior muscle and triceps surae muscles, respectively. The ankle flexor, or tibialis anterior (TA), stretch was performed by placing the index finger under the heel and the thumb on the dorsal surface of the paw and extending the ankle joint to its terminal position. When stretching the right hindlimb, the handler used his or her left hand, and vice versa. The ankle extensors, or triceps surae (TS), stretch involved both hands and was accomplished by stabilizing (extending) the knee using the index finger and the thumb of one hand, while flexing the ankle with the other hand. Both TA and TS muscle groups were stretched by a single therapist bilaterally for a minute each, totaling 4 minutes of stretch per rat per day. Stretching was performed 6 times total during a two week period, twice with each anesthetic in the order isoflurane, ketamine-xylazine, Nembutal each week.

During stretching, force and kinematic data were collected using the custom LabVIEW program. Using the motion analysis tool MaxTRAQ (Innovision Systems, Inc., Columbiaville, MI), the video recordings were analyzed, as shown in FIGURE 17.

(a) (b)

FIGURE 17 – (a) TA Stretch and (b) TS Stretch of an Anesthetized Animal

As seen in the figure, four points were digitized using this software. Point 1 was located on the toe, point 2 on the ankle, and point 3 on the knee. Point 4 was placed along the foot at the location of sensor contact. Points 1-3 were digitized and provided the angle of the ankle, and point 4 provided the means to calculate distance from the ankle to the point of force application. The distance between point 2 and point 4 (“d”) was expressed in centimeters and was used to calculate “Torque” (N-cm) using the following equation and the force value (“F”, expressed in grams) from the LabVIEW program:

$$\text{Torque} = \frac{d * F * 9.81\frac{m}{s^2}}{1000\frac{g}{kg}} \tag{3}$$

The torque and angle relationships were examined using Microsoft Excel (Microsoft, Redmond, WA).

B. GLOVE Study

The main study of this thesis was labeled GLOVE. Adult, female Sprague-Dawley rats (250-300 g) (N=6) received moderate contusion injuries at the T10 level using an NYU impactor device. Developed in 1992 by John A. Gruner, the NYU impactor device consists of a 10 g impactor that can be raised 6.25, 12.5, 25, 50, or 75 mm above the surface of the exposed spinal cord, producing a range of chronic injuries from very mild to complete paraplegia [41]. For a mild injury, the rats were given a 12.5 g/cm contusion injury and allowed to recover for three days.

Before rats were injured, a week of gentling was performed following the same protocol as seen in the GL-AN study. Following gentling, two therapists each applied stretches bilaterally to the TA and TS muscles for a minute each, totaling 8 minutes of stretch per rat per session under isoflurane. This was done twice weekly for two weeks, followed by a week of rest before injury.

After injury, the six rats were divided into three groups: a group stretched once a week (n=2), a group stretched twice a week (n=2), and injured control animals (n=2). The stretching rats were given three days to recover and stretched initially four days post injury (dpi), to capture the effects of spinal shock on stretching. Then for four weeks after recovery (Weeks 2-5), the rats were stretched either once or twice a week as described previously. Kinematic and force data were collected and analyzed during each stretching session, and the Basso, Beattie, Bresnahan (BBB) Locomotor Rating Scale was used to measure functional recovery both before and after every stretching session [42]. A detailed timeline can be seen in FIGURE 18.

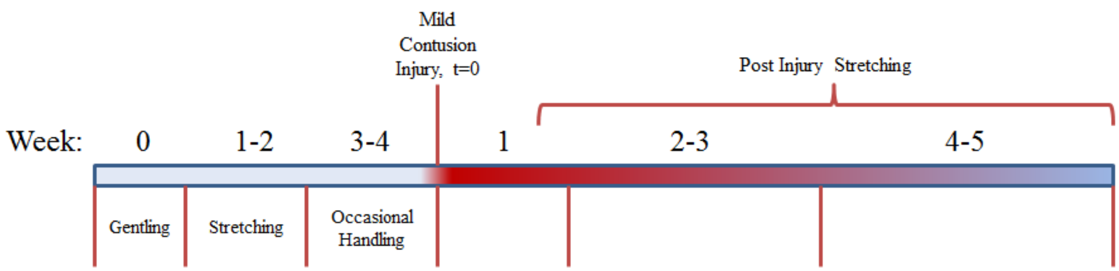

FIGURE 18 – Detailed Timeline of the GLOVE Study

IV. RESULTS AND DISCUSSION

The GL-AN study was merely a pilot study to prove the effectiveness of the glove device. At the same time, it allowed the therapists to gain valuable practice manipulating the device while effectively administering stretch therapy to the TA and TS muscle groups. Because of this and the fact that the duration of the study was so short and without an injury model, data from the GL-AN study will not be investigated.

As previously mentioned, the animals in the GLOVE study were numbered and randomly divided into three groups, summarized in TABLE IV.

TABLE IV

GROUP CLASSIFICATIONS BY RAT ID NUMBER

Rat ID	Group
1	Twice-per-Week
2	Once-per-Week
3	Twice-per-Week
4	Once-per-Week
5	Control
6	Control

To compare the effect of stretching dosages it is necessary to look at the stretching characteristics as well as recovery. Analysis was performed as shown in FIGURES 19-20 to determine whether stretching once a week or twice a week influenced the two aspects of stretching this thesis focuses on, torque and end ROM. Though there appeared to be variation between the once-per-week and twice-per-week groups as recovery progressed, no consistent patterns emerged. There appeared to be an acute loss in extensibility in both groups between 3 and 4 weeks post injury, although a steady increase in torques seems to characterize both stretching groups. However, a general linear model analysis comparing the once and twice-per-week groups found no significant differences in both stretching torques and end ROM between the two groups and over time.

The parallels between the groups suggest that both once-a-week and twice-a-week stretching produces effectively similar physiological changes in the function of the TA and TS groups. As mentioned earlier, it has been shown that stretch applied daily for less than 3 months confers little or no added benefit over and above usual care in people with SCI [16]. A previous study by Caudle et al. using the same rat injury model as the present study justified this conclusion when it showed that a regimen of 30 minutes of daily stretch over 8 weeks only caused a slight change in ROM [3].

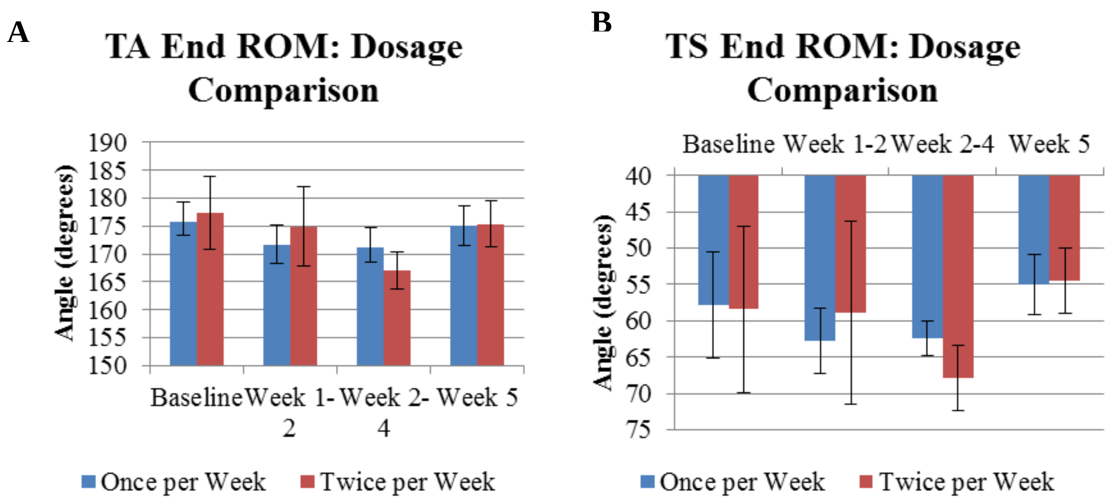

FIGURE 19 – End ROM Comparisons Based On

Dosage in (a) TA and (b) TS Muscle Groups

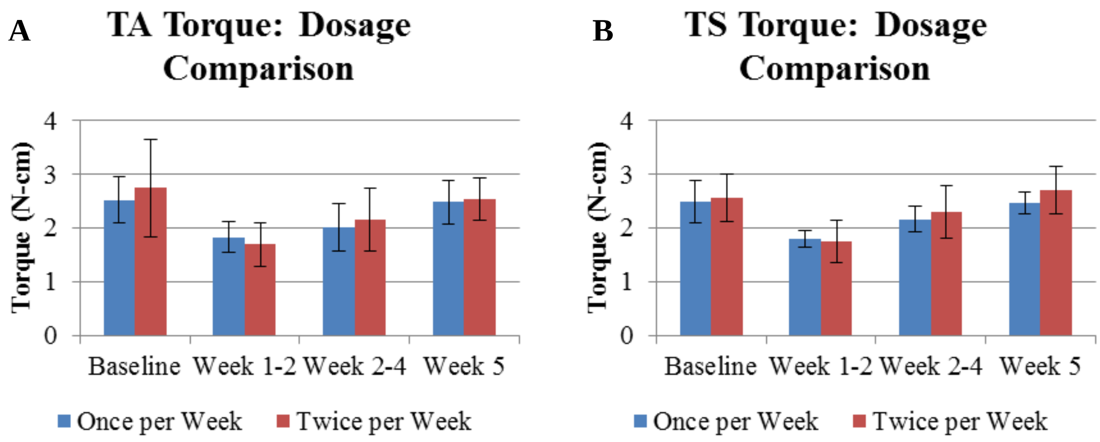

FIGURE 20 – Torque Comparisons Based On

Dosage in (a) TA and (b) TS Muscle Groups

However, this same study by K. Caudle showed drastically reduced recovery in the daily stretched group (FIGURE 21) evident at only 2 weeks post injury [3]. Compared to the recovery of the animals in this study, there is a striking distinction. The recovery pattern in the present study showed that the once-per-week stretching group recovered as quickly and to the same extent as the control group, as seen in FIGURE 22. However, a non-parametric ranks test showed that the twice-per-week group was significantly different from the control group ($p<.0001$) and once-per-week group ($p<.001$). Therefore, the group receiving the largest stretching dosage had statistically lower BBB scores without a significant change in end ROM. This finding supports the hypothesis by K. Caudle that the stretching protocol is activating barrages of afferent activity that disrupted lumbosacral circuitry and plasticity.

This contrast between the dosages in this study and prior dosages points to a couple things. It is well documented that paralyzed tissue below the level of a clinically "complete" SCI has enormous capacity to adapt [43], but is hindered by degradation of the musculoskeletal system of the paralyzed limbs. Without normal use, paralyzed muscle rapidly atrophies, creating a catabolic state, and increased risk for secondary complications [44]. However, in our contusion model, the animals seemed to recover to their maximal ability by 30 days post injury. This leads to the conclusion that in such a mild injury muscle atrophy and extended paralysis are not a problem and that chronic physiological changes in muscle spindles did not affect stretching after injury. Moreover, it goes on to support the claim made by K. Caudle that such extensive stretching regimens, 30 minutes daily for 4 months, can actually be detrimental to recovery [3], whereas more relaxed stretching regimens may maintain joint ROM without physiological or neurophysiological damage.

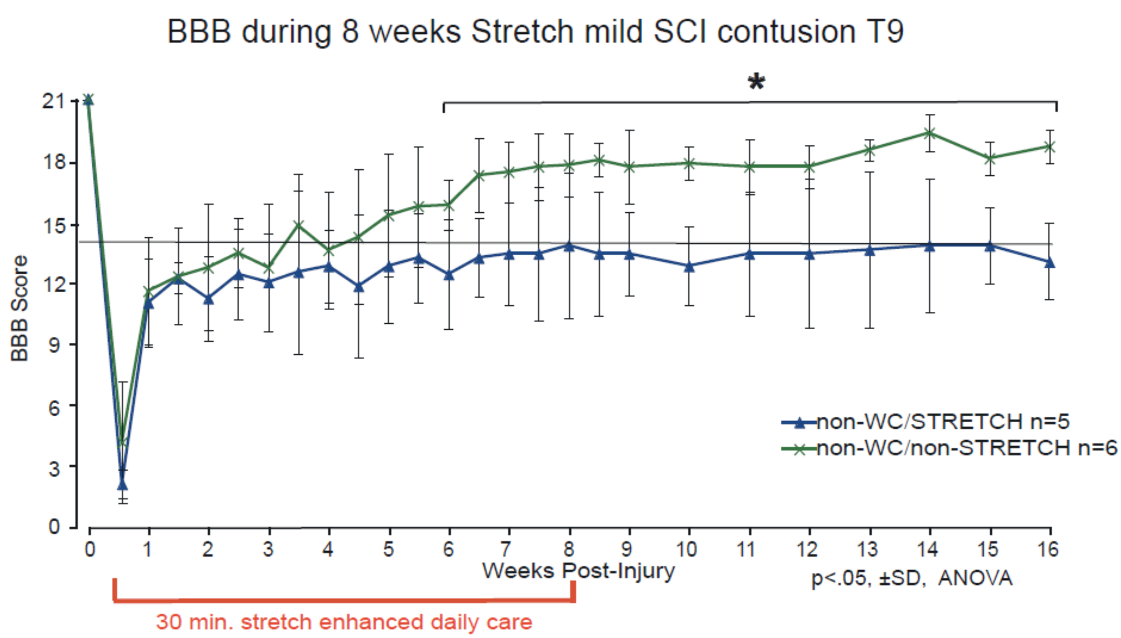

FIGURE 21 – Deterioration in Recovery Due to Stretching in Mildly Contused Rats [3]

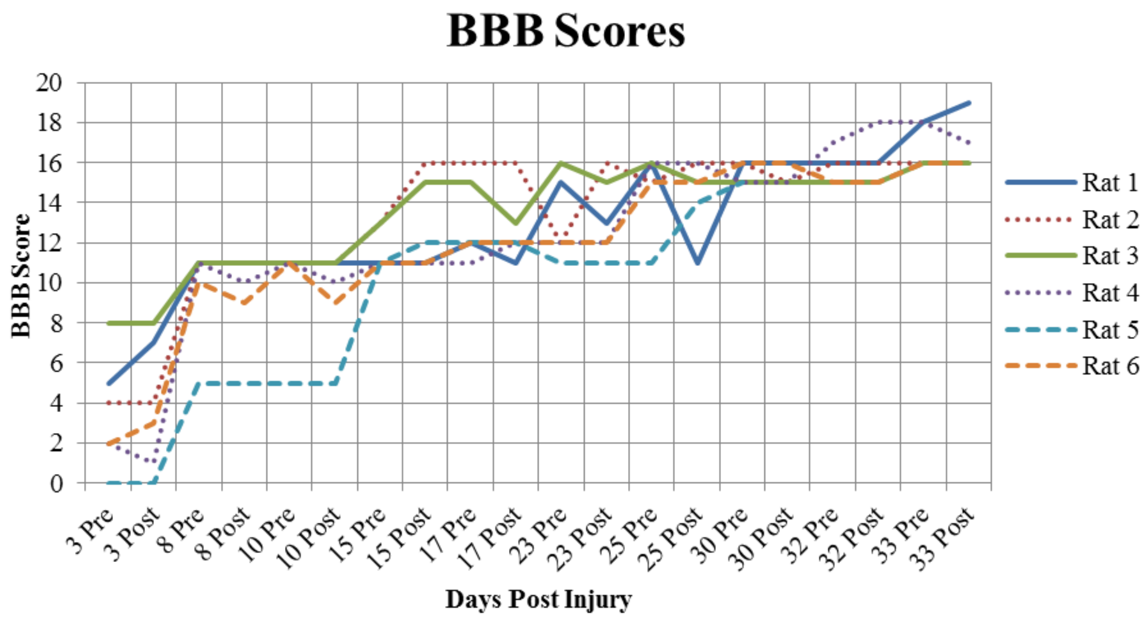

FIGURE 22 - BBB Locomotor Rating Scale Scores

Examining the plots of end of range torques for both the TA (FIGURE 23) and TS (FIGURE 24) muscle groups, one trend is prominent. There is an immediate and drastic decrease in end of range joint torque following SCI, with a gradual increase toward baseline levels from week 3-5 after injury. Keeping in mind that all baseline stretches were performed under light anesthesia and represent the muscle at its most flaccid level before injury, this raises striking questions.

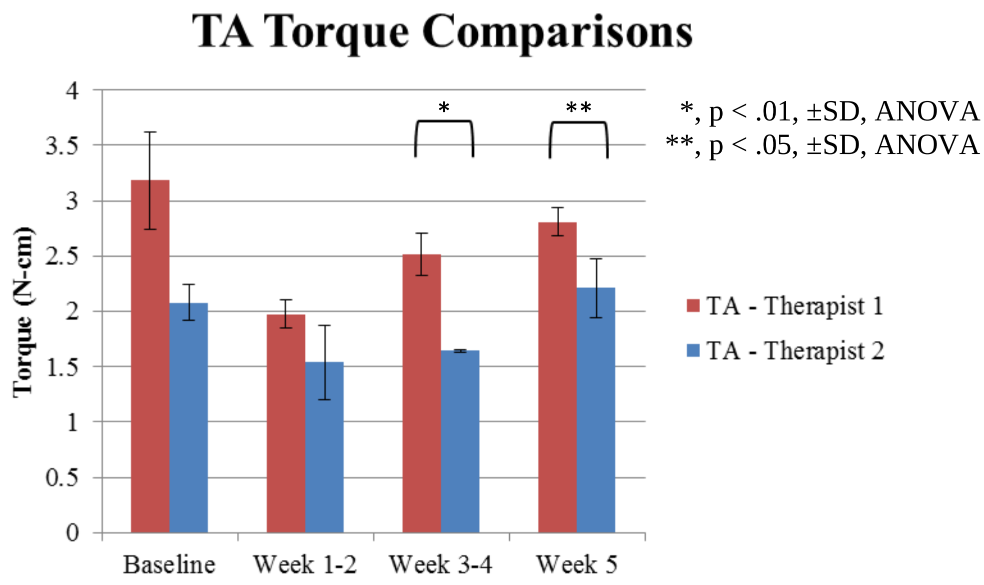

FIGURE 23 – End ROM TA Torque Comparisons between Therapists over Time

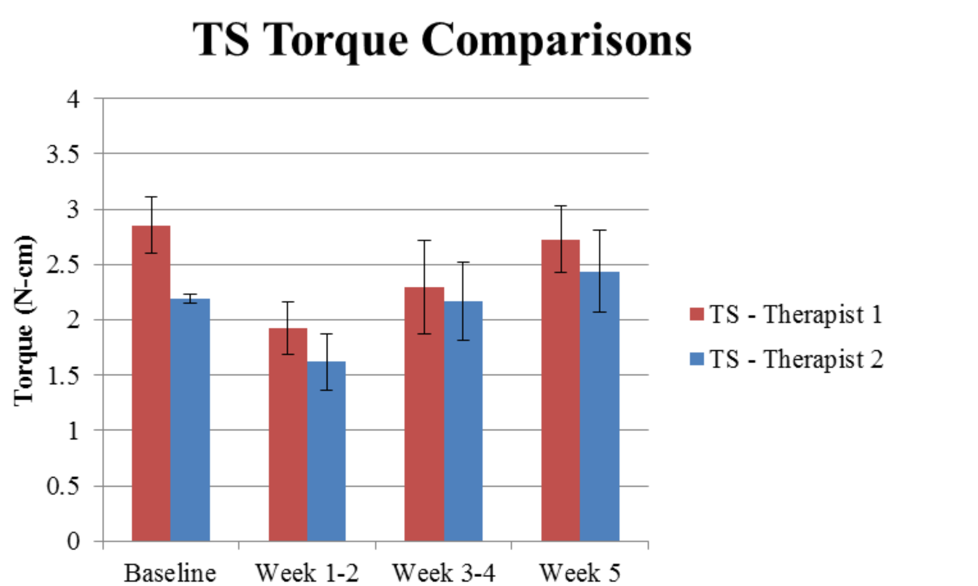

FIGURE 24 – End ROM TS Torque Comparisons between Therapists over Time

Severe SCI results in an immediate and prolonged depression of stretch reflex excitability in spinal segments lying caudal to the injury [45]. The term "spinal shock," classically defined as a "state of total abolition of all tendon reflexes and profound depression of other reflex activity below the level of cord transection which immediately follows such a lesion," [46, 47] describes this condition. When examining acute spinal cord injury, dramatic differences in lower limb tendon reflexes were seen between subjects with respect to injury severity [45]. At a more chronic stage, patients with incomplete SCI tend to have enhanced spinal cord excitability, characterized by frequent spasms and enlarged tendon response amplitudes, compared to subjects with complete injury [48]. One study noted that spinal shock varied with injury severity, and might last only minutes or hours in mild to moderate injuries, but this finding was not elaborated [49]. Very few investigations have examined reflex properties after acute SCI in persons with incomplete injury. However, one study found that deep tendon reflexes were present at only 2–3 dpi in ASIA 'D' subjects and 75% of the ASIA 'B' and 'C' subjects [50], a period in which spinal shock consistently appears in more severe injuries.

Throughout all the stretching sessions, the rats were either within the period of spinal shock or were in the recovery phase. When stretching began they were most likely in a degree of spinal shock during which deep tendon reflexes (DTRs) are still absent with a possible return of the tibial H-reflex and potential muscular flaccidity and paralysis [51]. The rats most likely began and finished stretching as their DTRs began to gradually reappear [50] and became evident in almost all subjects within 30 days, though the timing of this reflex return is highly variable [52]. Reflex return timing could be dependent on the pre-injury gentling each rat experienced [52] or by each rat's independent in-cage activity. Further characteristics of this later stage of recovery are increased excitability that is clinically associated with the development of brisk deep tendon reflexes, increased resistance to muscle stretch, muscle spasms and clonus [53]. Neurophysiological correlates include prominent accentuation of H-reflexes [54].

Examination of the H-reflex after SCI leads to a possible explanation of the initial decrease and subsequent rise in joint torques following the mild SCI. Most likely the H-reflex has not returned to the animals in the first few weeks. The H-reflex is analogous to the spinal stretch reflex; the pathway for both is the same: after the appropriate stimulus (electrical or physical), action potentials travel along Iα afferents to α-motor neurons and ultimately result in a twitch response of the muscle. The only difference is that the spinal stretch reflex is induced after a muscle stretch, whereas the H-reflex is the result of electric stimulation [55]. Without the H-reflex, the pathway to oppose physical stretch is absent and the same end ROM can be achieved with less torque. As the reflex gradually returns to the animals (weeks 3-4), twitches are produced in the muscle in response to stretch and more torque is required to achieve the same ROM. By week 5, the H-reflex should be accentuated to the point that the torque applied should meet or exceed the baseline torques, as brisk deep tendon reflexes and increased resistance to muscle stretch are seen, as noted earlier. General linear model tests found significance only in the plot of TA Torque, where both therapist ($p < .0001$) and week after injury ($p < .001$) were significantly different from one another. Post hoc t-test calculations went on to show that during weeks 3-4 ($p < .01$) and week 5 ($p < .05$) the TA torques between therapists were significantly different.

It should be noted that baseline recordings were performed under isoflurane, as sensate animals cannot be safely stretched. Therefore, the baseline recordings were not included in the data when statistical tests were performed. The effects of isoflurane on the H-reflex have been extensively studied [56-58] and studies have shown that isoflurane causes H-reflex suppression of about 50% [56, 57] and therefore decreases motoneuronal excitability in the spinal cord [56]. In this case, it would not be implausible that stretch reflexes 5 weeks post-SCI and those seen in anesthetized animals during baseline recordings are very similar. The return of stretching torques to baseline values, when looked at under this light, makes sense.

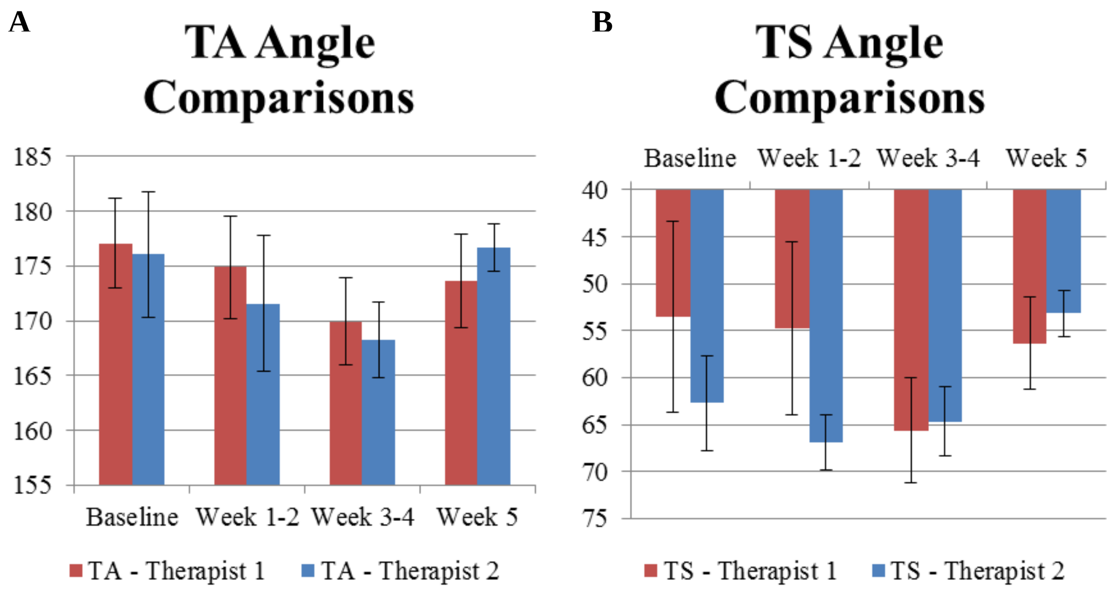

FIGURE 25 – End ROM Angle Comparisons for (a) TA and (b) TS Muscle Groups Between Therapists over Time

Another observation that can be drawn from FIGURES 23-24 is that of therapist variability. The torques applied by Therapist 1 are significantly higher than that of Therapist 2 for over 90% of the baseline recordings and also during TA stretches 5 weeks after injury. Though not statistically significant, the contrast between torques applied by the two therapists is apparent. The method by which stretching was performed each day involved Therapist 1 stretching each rat immediately before Therapist 2 applied stretch. Therefore, the variability in stretching torques could stem from muscular "creep." It has been shown that during the first 5 minutes of stretch, there is a gradual increase in muscle extensibility [27]. Since Therapist 1 applied a minimum of 4 minutes of passive stretch to each rat before Therapist 2, the likelihood that the muscle groups were at their highest extensibility for only Therapist 2 is high. This would imply that Therapist 2 reaches a greater range of motion while applying torque less than or equal to that of Therapist 1. However, a comparison of the end ROM achieved by each therapist showed no perceptible difference, as can be seen in FIGURE 25. Therefore a more plausible explanation for the contradictory joint torques would be an inconstant perception of "suitable" torque. Because therapist variability is a well-documented issue in stretching after SCI [24], this remains the most likely explanation.

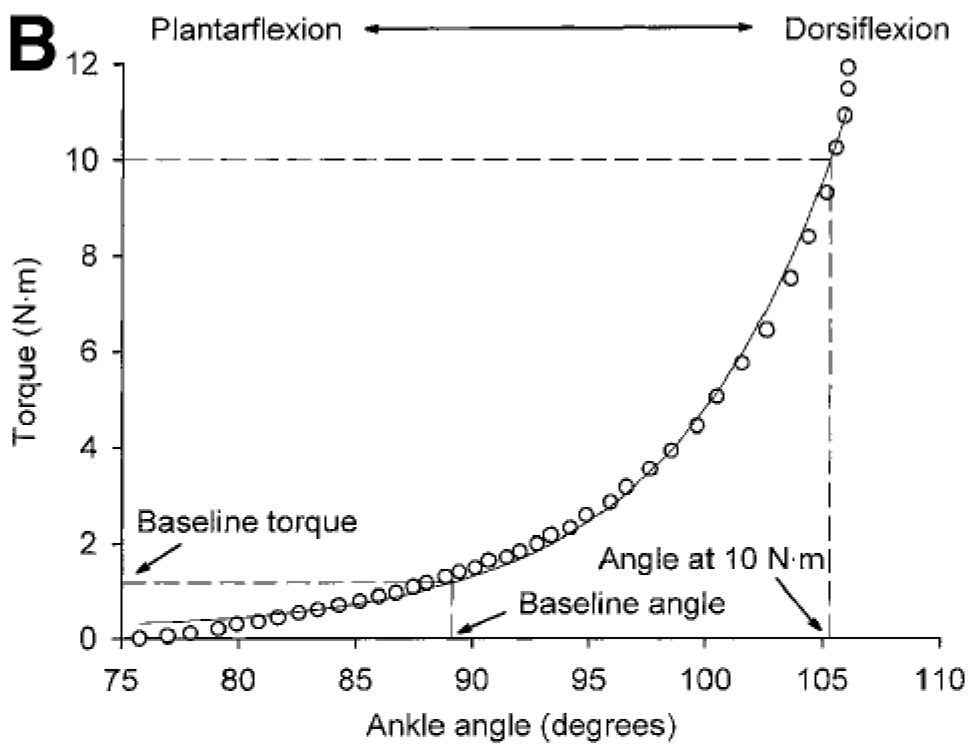

FIGURE 26 – 14 Patients (66 ± 17 kg) Received Passive Stretch into Dorsiflexion after SCI [13]

For purposes of translation, it is necessary to attempt to compare the published data on stretching in humans to the present data in the rat model. Due to physical dissimilarities in the ankle joints between the two species, it would be foolish to attempt to compare ranges of motion. However, a comparison can be drawn between torques applied during stretch. In 2000, L. Harvey published data comparing the ankle angle and torque applied during dorsiflexion in a human patient after SCI [13]. In FIGURE 26 the range of torque applied during stretching begins at the baseline torque, the torque caused by the weight of the foot alone, and ends at the location where the end ROM is first evidenced. Dorsiflexion in the human and the triceps surae stretch in the rat both focus on the medial gastrocnemius. Making the assumption that the ratio between the average body weight of the human patients in Harvey's study and the rats in the present study is similar to the ratio of torque applied during dorsiflexion in the human and rat, one can make mathematical inquiries into the TS torque data in the present study. The ratio of the human patients (66 ± 17 kg) to the rats (275 ± 10 g) is 240:1. Applying this same ratio to the range of torques elicited in Harvey's study (2-10 N-m), a theoretical range for the rat model would be 0.8-4.2 N-cm. As a testament to this comparison and to the efficacy of our therapists' stretching techniques, all of the torques elicited during stretching fell well within this range. This range can be put in perspective by presenting torque as total body weight applied a certain distance from the joint in question. Common end ROM torques seen in the rat model, 2.5 N-cm for example, can be viewed as the body weight of the rat applied to a point 0.9 cm from the ankle, about three quarters of the length of the rat paw.

However, it is very possible that this previous comparison between human and animal torques is inadequate. Measuring torque merely allows scientists to have a surrogate measure of stretching. In reality, the best aspect of stretching that we could measure would be muscle twitch force, but that is impossible to measure *in vivo*. There are many differences between the animal and human models that come into play when we compare two surrogate measures. For instance, the moment arms of each muscle, designated by their respective bone attachment points, are surely different for a quadruped than a bipedal animal. The cross sectional areas and fiber types for each muscle group are sure to be different between species as well. Therefore, the comparison proposed in this Discussion merely acts as a spark for further investigation into the translation of this study’s findings.

V. CONCLUSION

The findings presented in this study offer promising strategies to problems found in SCI rehabilitation clinics and physical therapy clinics alike. In rehabilitation, it has been demonstrated clearly that therapists administer differing torques to patients in an attempt to maintain muscle extensibility and improve functional training. After multiple attempts, no study has clearly defined an optimal dosage of stretching after SCI. Therefore a shift in the current analysis of stretching is not only required, but potentially more rewarding.

This thesis provides the means by which intensity of stretching can be gauged and compared. From the data presented, it seems that passive stretching is not as “passive” as the name suggests. Stretching a muscle to an end ROM appears to be an active process, requiring therapists to temporally tailor the torque applied to each joint. Due to the viscoelastic nature of muscles, the effect of body weight on necessary torque for maximal range of motion, and the apparent change in joint ROM after acute SCI, there is an obvious need to measure and modify stretching delivery.

Searching for an optimal stretching dosage without keeping the actual stretching constant has led nowhere. However, a study holding stretching dosage constant and only comparing differing stretching torques could unearth breakthroughs previous studies could not. The presented force measurement device allows therapists to measure torques applied during stretching in animals in real-time and adjust the intensity accordingly. While it is prudent to begin such studies in a rat model, the concept of live force measurement is not restricted to use in animals.

VI. RECOMMENDATIONS

It is recommended that studies in animals and humans be undertaken simultaneously, as translating findings in animal models to patients in the clinic may not be straightforward. The characteristics of balance, trunk stability, and posture and the consequences of errors in stepping vary considerably and dramatically influence the capabilities of animals during the critical acute and sub-acute post-injury time period [3]. Rodents are very active within days of injury whereas patients are largely immobile for several weeks or months before they enter an activity-based rehabilitation setting [59].

A future study will need to examine acute stretching torques in the rat model, but with a different sample size. An extremely low number of animals were used in this study (N=6) due to time and resource constraints. This small sample size places serious limitations on the significance on the data presented in this study. While it is doubtful that a similar study utilizing more animals will uncover different discoveries, measures need to be taken to ensure that enough animals are used that the data carries enough power to justify transitional and translational research. To this end, a power analysis can be performed to determine an appropriate sample size.

To ameliorate variability in stretching protocols in rehabilitation clinics, this same type of device can be applied to human patients. Obviously, therapists apply much greater torques to humans than animals during stretching. The type of sensor used in the glove device can only accurately sense loads up to about 100 pounds (445 N), so it would be necessary to choose a different type of sensor. Though load cells are costly and physically larger than FSRs, as previously discussed, their interface with patients would not be as problematic as it is in animals. Therapists stretch patients using their entire hand, and commonly their entire body weight. A palm-sized, cushioned load cell should not hinder limb manipulation or force application and would offer extremely accurate force and torque measurements during stretching. Even if it is too premature to begin searching for effective stretching torques in humans, measures can be taken to ensure stretching in patients is performed safely and consistently.

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APPENDIX I: STATISTICAL TESTS

Statistics were performed using IBM (Armonk, New York) SPSS Statistics software package. For sake of brevity, the only tests included are those that found significant differences in the data. The general linear model analysis performed for "FIGURE 23 - End ROM TA Torque Comparisons between Therapists over Time" can be seen below:

General Linear Model

Descriptive Statistics

	Therapist	Mean	Std. Deviation	N
TATORQUE12	1.0	1.973523	.1265600	4
	2.0	1.539098	.3367327	4
	Total	1.756311	.3307285	8
TATORQUE34	1.0	2.514164	.1956699	4
	2.0	1.643477	.0143332	4
	Total	2.078821	.4827999	8
TATORQUE5	1.0	2.807297	.1248771	4
	2.0	2.209012	.2712148	4
	Total	2.508155	.3748042	8

Mauchly's Test of Sphericityb

Measure:MEASURE_1

Within Subjects Effect	Mauchly's W	Approx. Chi-Square	df	Sig.	Epsilona		
					Greenhouse-Geisser	Huynh-Feldt	Lower-bound
Weeks	.995	.024	2	.988	.995	1.000	.500

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.b. Design: Intercept + Therapist Within Subj

Tests of Within-Subjects Effects

Measure:MEASURE_1

Source		Type III Sum of Squares	df	Mean Square	F	Sig.
Weeks	Sphericity Assumed	2.276	2	1.138	27.997	.000
	Greenhouse-Geisser	2.276	1.991	1.144	27.997	.000
	Huynh-Feldt	2.276	2.000	1.138	27.997	.000
	Lower-bound	2.276	1.000	2.276	27.997	.002
Weeks * Therapist	Sphericity Assumed	.194	2	.097	2.389	.134
	Greenhouse-Geisser	.194	1.991	.098	2.389	.134
	Huynh-Feldt	.194	2.000	.097	2.389	.134
	Lower-bound	.194	1.000	.194	2.389	.173
Error(Weeks)	Sphericity Assumed	.488	12	.041		
	Greenhouse-Geisser	.488	11.943	.041		
	Huynh-Feldt	.488	12.000	.041		
	Lower-bound	.488	6.000	.081		

Tests of Between-Subjects Effects

Measure:MEASURE_1Transformed Variable:Average

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Intercept	107.299	1	107.299	2272.315	.000
Therapist	2.415	1	2.415	51.149	.000
Error	.283	6	.047		

VITA

Erik Seibt is originally from Covington, KY and received his Bachelors of Science from the University of Louisville in 2011. He was introduced to the Kentucky Spinal Cord Injury Research Center working as an undergraduate co-operative education student in the lab of Dr. David Magnuson. During this time, he was involved with multiple studies focusing on activity based rehabilitation. Prior studies investigating stretching and functional recovery after SCI in this lab suggested the idea for the present thesis. Erik Seibt will receive his Masters of Engineering degree in 2013 and will go on to the University of Louisville’s School of Dentistry as part of the Class of 2017.