

MECHATRONIC AND BIOMECHANICAL CONSIDERATIONS TOWARD THE DESIGN
OF AN ANKLE CLONUS SIMULATOR

BY

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THESIS

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ABSTRACT

Clonus is a rhythmic, involuntary reflex response often seen in individuals with lesions in their descending motor neurons. These lesions can be the result of spinal cord injury or diseases like multiple sclerosis, Parkinson's disease and cerebral palsy. During clinical evaluation of an individual suspected of having lesions in their descending motor neurons, an ankle clonus test is often employed. Clinical evaluation of clonus is currently reported using qualitative rating scales and therefore accurate diagnosis is largely dependent on the clinician's experience. Clinician trainees may not be able to obtain sufficient training time with patients to understand and diagnose clonus effectively. Thus, we, at the Human Dynamics and Controls Lab, designed an electromechanical ankle clonus simulator, based on a series elastic actuator mechanism, that is capable of providing training for administration of an ankle clonus test and therefore substituting the need of human patients. The ankle clonus simulator is intended to accurately mimic the behavior of the ankle joint of a patient demonstrating ankle clonus. However, to obtain confidence in the performance of the ankle clonus simulator, clonus has to be academically explored after which, biomechanical and mechatronic requirements of the simulator must be investigated.

This thesis first presents an overview of ankle clonus and subsequently establishes specific research questions that needed to be answered, based upon which biomechanical targets, anthropomorphic designs, and sensor subassemblies are constructed. Biomechanical models of passive ankle torque to be mimicked by the simulator were developed and preliminary evaluation of sensors designed to be included in the first-generation ankle clonus simulator was conducted. To determine the optimal model for passive joint moment, in both healthy individual case and clonus's case, two different ankle passive stiffness models, derived from literature, were computed and compared. Double exponential functions were consequently selected as the appropriate models for passive stiffness for healthy and clonus cases. Sensor systems were designed to provide appropriate force readings, joint angle measurements and estimates of clinical learner's hand position on the foot structure of the simulator. Off-device testing stations/fixtures for these sensors were built to evaluate the selected sensor's efficacies and also to inform us of the final torque error margins ($<5\%$) and ankle angle margins ($<1\%$) that the clonus simulator would operate in. Overall, enough confidence was established in selected sensors such that an ankle clonus simulator that is manufactured based on the designs presented in this thesis would be able to perform as intended.

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THESIS ORGANIZATION

This thesis presents an overview of clonus and the design of sensor systems incorporated into a first-generation novel training simulator that provides realistic recreation of ankle clonus in patients with upper motor neuron lesions.

Section 1 presents the definition of clonus, current understanding of mechanisms causing clonus and existing assessment methodologies employed in a clinical setting. Additionally, this section highlights the lack of consistent metrics and scales for the accurate quantification of clonus. Furthermore, this section details some of the anatomical features of the ankle joint complex and biomechanical behavior of the ankle joint. A brief overview of other ankle clonus simulators in the biomedical community was included followed by the architecture of our first-generation ankle clonus simulator. A short overview on the benefits of using series elastic actuator mechanisms in biomedical robotics devices is then provided. Lastly, specific research questions pertaining to the development of an ankle clonus simulator were detailed and corresponding requirements that must be met to answer those questions are explained.

Section 2 presents the methods used to determine the biomechanical and anatomical requirements for the simulator. Additionally, the methodology for informing the design of a sensor subsystem (comprising of 3 different sensor systems) is detailed.

Section 3 reports the results of the passive ankle torque estimation techniques, followed by details of the final sensor system design and preliminary evaluation of the sensors used to provide feedback signals for control of the electromechanical systems in the simulator. Off-device tests and were conducted to measure the efficacy of the sensor systems.

Section 4 discusses the key learnings from obtained results. Limitations of the first-generation design are discussed. Furthermore, next steps required to develop a fully incorporated system that simulates both spasticity and clonus are detailed

1 INTRODUCTION

1.1. CLONUS

1.1.1. DEFINITIONS

Clonus is defined as involuntary and rhythmic muscle contractions caused by lesions in the descending motor neurons in the lateral corticospinal tract [1]–[5]. Lesions that cause clonus can exist in the upper motor neurons (UMN) or anywhere along the lateral corticospinal tract, a pathway that extends from the cerebral cortex to the lower end of the spinal cord [6]. These lesions can be the result of spinal cord injury (SCI), multiple sclerosis (MS), cerebral palsy, amyotrophic lateral sclerosis (ALS), Parkinson's disease and other neurological impairments [7]–[10]. Although most commonly observed in the ankle joint, clonus has been reported in muscle groups at other joints, including the jaw, knee and wrists [11]–[13].

Ankle clonus can be elicited in individuals by rapidly dorsiflexing (DF) the ankle and maintaining a stretched state of the ankle plantarflexor muscles [1], [7], [14]. Other peripheral inputs that activate the stretch reflex, such as an Achilles tendon tap [14]–[16], have also been employed to generate clonus. Ankle clonus response is rhythmic with a characteristic frequency anywhere between 5-8 Hz [1], [3], [4], [14], [17], [18]. Each period of oscillation is termed clinically as a beat. However, rhythmic reflex response to rapid dorsiflexion or external stimuli at the Achilles tendon has been observed in healthy individuals [19] and therefore, clonus of more than 4 beats is treated as an indication of irregularities in descending motor pathways [20]. Consequently, clonus of more than 10 beats has been termed as sustained clonus [21], an affirmative indicator of lesions somewhere along the corticospinal tract. Clonus is therefore commonly considered as a symptom that clinicians test for during a neurological exam, making it a marker that is indicative of hyperflexia and consequently lesions in an individual's descending motor pathways [21].

Measurements have shown that for 55% of the clonus period, the ankle is in dorsiflexion while the rest of the period is comprised of plantarflexion (PF) [22]. Usually, the first beat has the highest amplitude and longest period, with the period becoming consistent after the 4th beat [4]. A

division in the research communities exists regarding the variability of ankle clonus frequency. Dimitrijevic et al. reported that the characteristic frequency of sustained ankle clonus has been moderately consistent ($\pm 15\%$ deviation from average) over a period of two years [16] while others [18], [23] have reported a shift in clonus frequency due to changes in applied loads or speeds. The lack of an affirmative understanding of clonus frequency variation is the core challenge in agreement within the community regarding the mechanism of clonus (see section 1.1.3)

Clonus patients usually exhibit additional symptoms of abnormal muscle behavior related to motor neuron impairments like spasticity [1], [5], extensor spasms [24], flexor spasms [25], [26] and weakness in affected muscle groups. It has been observed that there is a close relation between the increase in muscle tone (spasticity) and exaggerated stretch reflexes that cause clonus [16]. However, it is not consistent that all patients experiencing spasticity would also demonstrate clonus [16].

1.1.2. EFFECTS OF CLONUS

The degree of impact of clonus on a patient's life can vary from manageable to highly restrictive to activities of daily living [14], [27]. For some patients, ankle clonus (and the accompanied symptoms of spasticity) can have a debilitating impact on locomotion due to drastic effects on the gait cycle [28], [29]. Sustained clonus can begin during the stance phase of the gait cycle due to the ankle undergoing dorsiflexion, effecting balance and may even completely limit a patient's ability to walk at any speed. Stride length and pressure distribution changes have been observed in patients with clonus that are severe enough to completely disallow walking [28]. Additionally, clonus has been shown to disrupt sleep and cause fatigue of involved muscle groups [30], [31]. Amplitudes and durations of clonus beats can be modulated with cryotherapy (ice water bath) around affected muscle groups [32]. There has been some evidence that clonus frequency and amplitude vary with change in body posture from sitting to standing [33].

1.1.3. ANKLE CLONUS MECHANICS

Clonus is an abnormal muscle stretch reflex response. Broadly, hyperactive stretch reflexes are termed as hyperreflexia and rhythmic stretch reflex responses are termed as clonic reflexes [34].

A consensus on the underlying mechanics of clonus has not been reached. There are two distinct schools of thought that contend to fully explain the mechanics of clonus. In one, the understanding is that clonus has its repetitive response due to self-perpetuating reactivation of peripheral muscle stretch circuits and therefore, each beat produces the subsequent one. In the other, clonus is believed to result from actions of a central oscillator (oscillating response from the descending motor neurons) and that stretch reflex pathways play a very limited role in sustaining clonus. One of main metrics used to provide evidence to support either of the hypotheses is the variance in ankle clonus frequency based on changes in the external stimuli that cause it. Essentially, an unchanging clonus frequency would support the theory that clonus mechanisms are explained by central oscillators, while a varying clonus frequency is indicative of a self re-excitation mechanism. Table 1 summarizes the literature that have published arguments supporting either of the two hypotheses.

Table 1: List of authors and corresponding literature providing merit to defining clonus mechanism caused by (1) a reflex origin or (2) a central origin.

Self re-excitation (reflex origin)			Central Oscillator (central origin)		
Authors	Year	Citation no.	Authors	Year	Citation no.
Szumski et al.	1974	[35]	Walsh	1976	[36]
Hagbarth et al.	1975	[37]	Dimitrijevic et al.	1980	[16]
Iansek	1984	[38]	Walsh and Wright	1987	[12]
Rack et al.	1984	[23]	Beres-Jones et al.	2003	[14]
Latash et al.	1989	[39]			
Rossi et al.	1990	[3]			
Hidler and Rymer	1999,2000	[5], [18]			
de Vlugt et al.	2012	[40]			

1.2. CURRENT CLINICAL ASSESMENT METHODOLOGIES

1.2.1. ANKLE CLONUS ASSESMENT PROCEDURE

Currently, diagnosis is administered by clinicians by counting the number of beats measured in a sustained clonus elicited by rapid dorsiflexion of the ankle or less commonly an Achilles tendon tap. Specifically, the stretch reflex inducing procedure begins by allowing the patient to relax with a passively flexed ankle. Knee flexion angle (0 being fully extended) is usually smaller than 90° for patients being examined while in supine or sitting position. To ensure this initial state, examiners most commonly support the patient's leg with the hand not performing the clonus test. Next, examiners place their hand on the plantar surface of the patient's forefoot (covering the metatarsal head regions) and briskly dorsiflex it, after which, consistent pressure is maintained so as to keep the ankle in a dorsiflexed state and the tactile feedback from any clonus response generated is appreciated (Figure 1). Rhythm of the clonic response and number of beats sustained are evaluated but not explicitly measured [21]. Clinicians may often instruct patients to clench their teeth or perform the Jendrassik maneuver to reinforce the ankle stretch response [41].



Figure 1: Clonus test. Orange arrow indicates direction of clinician applied force. Red arrow and indicates the direction of the rotation of the ankle about the rotation axis (red point).

The clonus clinical assessment is easy to administer since there are no equipment involved and can be performed quickly. Eliciting ankle clonus is most commonly used as a clinical diagnosis tool for detecting hyperreflexia. However, clinicians often evaluate a patient for other commonly associated pathologies like spasticity or weakness before referring them to a neurologist for further diagnosis. Additionally, an ankle clonus test may be employed to evaluate the possibility of serotonin syndrome, which commonly arises from certain drug overdoses [42].

1.2.2. QUANTIFICATION, RELIABILITY AND REPEATABILITY OF ANKLE CLONUS ASSESMENT

Ankle clonus is trivial to recognize in a clinical setting, but quantifying clonus has been difficult. While electromyography (EMG) can be used to measure reflex activity of muscles involved in DF/PF (see section 1.3.2) in a laboratory setting, such tests require a wide range of equipment and prolonged assessment durations. Since spasticity is another clinical condition that is assessed for individuals suspected of UMN lesions, scales used to quantify spasticity are often reported by clinicians. Other than counting the number of beats elicited after swift dorsiflexion, a specific quantitative metric for clonus does not exist in clinical scenarios.

Currently, the assessment tools that quantify spastic behaviors most employed by clinicians are the Ashworth Scale (AS) [43], Modified Ashworth scale (MAS) [44], and Modified Tardieu Scale (MTS) [45], with MAS being the most widely used. However, these scales were originally developed as tools to measure spasticity based on its traditional definition : “a motor disorder characterized by a velocity-dependent increase in tonic stretch reflexes (muscle tone) with exaggerated tendon jerks resulting from hyperexcitability of the stretch reflex, as one component of the upper motor neuron (UMN) syndrome” [46]. None of the scales mentioned above, with the exception of MTS, have evidence of good reliability and repeatability in quantifying clonus. Therefore, their continued employment in quantifying stretch reflex hyperexcitability like clonus and other spasms has been contested.

Numerous arguments have been published listing the limitations of AS and MAS for quantification of spastic motor behavior. Chief among the reasons for moving away from AS is due to its focus on resistance to movement at a single joint that displays signs of hyperexcitability. However, recent evidence suggest the contrary, that reflex hyperexcitability conditions involve

neural impulses originating from multiple joints [26], [47], [48]. Furthermore, AS and MAS do not explicitly quantify the velocity dependence of responses in the muscle tone. Instead, AS and MAS scores are assigned qualitatively, subjecting individual diagnosis to large variations. This is evident in reports of MAS and AS having low interrater reliability [49]–[51], with clinical significance with regard to clonus being moderate at best. Meanwhile, MTS does have an appropriate scale to help quantify clonus (Table 2) but is not as commonly used in reports by clinicians as MAS. MTS has also been reported to have moderate to very good intra/inter rater reliability in assessment of spasticity and abnormal reflex behavior [50], [52].

Table 2 : MTS scale that is appropriate to use for assessment of ankle clonus. Table adapted from [53]

Score	Description
0	No resistance throughout passive movement
1	Slight resistance throughout, with no clear catch at a precise angle
2	Clear catch at a precise angle followed by release
3	Fatigable clonus (<10 secs) occurring at a precise angle
4	Un-fatigable clonus (>10 secs) occurring at a precise angle
5	Joint immobile

Since clonus is measured with a time scale, MTS does not rely on qualitative analysis of muscle abnormalities.

To improve the reliability of clinical quantification of reflex hyperexcitability conditions like clonus, several new scales have been proposed. The Penn Spasm Frequency Scale (PSFS) [54] uses patient self-reported spasm frequencies to judge the severity of an UMN syndrome. Since it relies upon the patients to judge the frequencies, PSFS is prone to poor reliability [55]. Additionally, PSFS does not allow clinicians to delineate between the different types of spasms experienced by patients (namely clonus, extensor spasms and flexor spasms). The Spinal Cord Assessment Tool for Spastic reflexes (SCATS) is perhaps the most suitable, followed by MTS, for quantifying stretch reflex hyperexcitability (Table 3). SCATS was initially validated using correlation analysis of kinematic and EMG signals against SCATS scores, reporting high Spearman rank correlation coefficients ($\rho \geq 0.69$) [56]. A separate validation study suggest that SCATS has good interrater agreement and good repeatability for clonus, extensor spasms and flexor spasms [57], and shown to have moderate to substantial Kappa values using Landis and

Koch test criteria [58]. In the same study, SCATS did not have a good correlation with PSFS, indicated by Spearman rank correlation coefficients less than 0.2. However, one important note that the authors mention was that SCATS is intended to be employed primarily on patients with SCI and may not be entirely applicable for use with other patient populations (e.g. those with MS, cerebral palsy).

Table 3: SCATS scales for quantification of different types of spasms. Table adapted from [57]

Reflex type	Score	Description
Clonus	0	No reaction
	1	Mild: Clonus maintained <3s
	2	Moderate: Clonus persists 3-10s
	3	Severe: Clonus persists >10s
Flexor Spasms	0	No reaction
	1	Mild: <10° flexion at the hip and knee, or extension of the big toe
	2	Moderate; 10° - 30° flexion at the hip and knee
	3	Severe: >30° hip and knee flexion
Extensor Spasms	0	No reaction
	1	Mild: Contraction maintained <3s
	2	Moderate: Contraction persists 3-10s
	3	Severe: Contraction persists >10s

Flexor spasms are measured by applying pin prick stimulus to the medial arch and Extensor spasms are measured by visible contraction of quadriceps muscle after extension of hip and knee about their respective joints.

A recent paper also proposed temporal measurements, devoid of specific scaling criteria like MTS, to quantify ankle clonus. The clonus duration (using a stopwatch) and number of beats (using an optical motion capture system) in a 10 second interval were measured. Clonus duration measurements were reported to have good intra-rater and inter-rater reliability (intraclass correlation coefficient of 1.00 and interclass correlation coefficient of 0.99 respectively). Clonus beats measurements were found to have good correlation with SCATS clonus scores ($r=0.58$) and EMG measurements [7]. However, there was no evidence of interest in the clinical community to

adopt the SCATS nor the methodology for direct temporal measurements as a standard ankle clonus assessment metric.

The lack of a quantitative scale that specifically measures the extent of clonus can result in a clinical misdiagnosis for an inexperienced clinician. The qualitative nature of the assessment for clonus demands an extensive training regimen for trainee clinicians, especially when sometimes healthy patients can exhibit stretch reflex hyperexcitability without the presence of UMN lesions. Additionally, a medical institute may not have access to a sufficiently large patient population for clinicians to get a ‘feel’ of clonus and regularly train on the assessment procedure. Therefore, a clonus simulator designed specifically to recreate physiological performance of an ankle joint of clonus patients can help bridge that gap and better train clinicians to provide accurate diagnosis.

1.3. ANKLE BIOMECHANICS AND ANATOMY

In order to develop a physiologically accurate simulator for ankle clonus, it would be imperative to understand the biomechanics of the ankle joint. The understanding of ankle anatomy is mature and well documented; however accurate biomechanical modeling of the ankle joint complex is still evolving. This is partly due to the complex interactions between a large number of musculoskeletal elements and joints in the foot. The ankle joint and the corresponding muscles bear large amounts of force, ranging from 5 times body weight while walking and up to 13 times body weight during running [59].

1.3.1. DEFINITIONS, BONES, JOINTS AND MUSCLES

Three anatomical planes are defined as shown in Figure 2 and subsequent degrees of freedoms about the anatomical planes are highlighted in Figure 3. DF and PF occur in the sagittal plane, inversion/eversion occur in the frontal plane and abduction/adduction occur in the transverse plane.

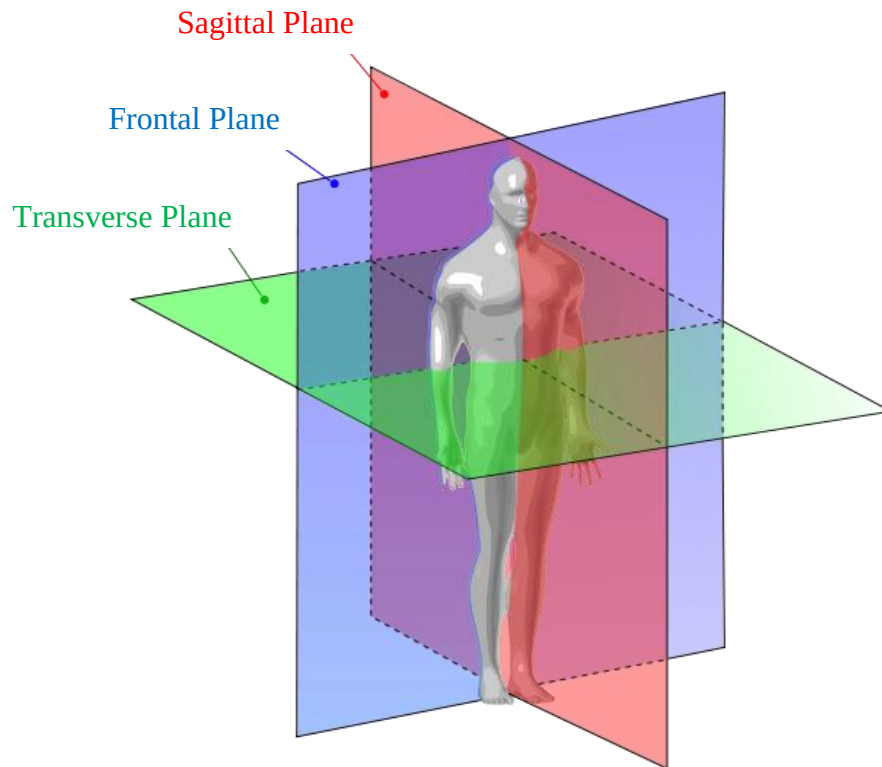


Figure 2 : Anatomical planes in humans. Adapted from [60]

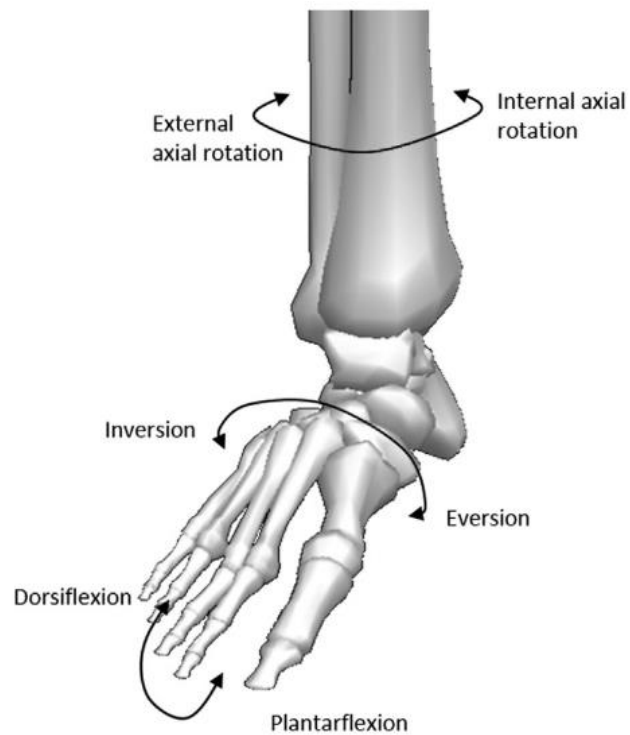


Figure 3: Definitions of ankle degrees of freedom. Adapted from [59]

A total of 26 bones make up the foot, some of which are depicted and labeled in the Appendix A. Consequently, the connections of the foot with the long bones of the lower limb make up a total of 33 joints in the ankle foot complex [61]. A joint is defined as a position about which two different bones articulate. Three main types of joints exist in the human body:

- 1) Synarthroses (immovable joints due to direct surface contact)
- 2) Amphiarthroses, (slightly movable due to connections via ligaments of cartilage)
- 3) Synovial/Diarthrosis (mobile joints where bone surfaces are separated by a synovial fluid that is contained in a fibrous joint capsule)

Of specific interest to biomechanics researchers are the synovial joints in the foot. There are two main joints that contribute to the articulation of the ankle in the different anatomical planes (Figure 2) namely, the Tibiotalar joint and the Subtalar joint. The three modes of movement in the ankle joint complex are PF/DF, rotation in the sagittal plane; abduction/adduction in the transverse plane and inversion/eversion in the frontal plane (Figure 3). The combinations of these three movements culminate in a three dimensional motion of the foot during gait more commonly called as supination and pronation [62]. During pronation, the foot undergoes DF, eversion and abduction while PF, inversion and adduction occur during supination.

The Tibiotalar joint (Talocrural joint) is a diarthrosis hinge largely responsible for ankle mobility in the sagittal plane allowing for PF and DF of the ankle about its oblique axis[59]. The joint, more commonly referred to as the ankle joint, forms a junction between the tibia, fibula of the lower leg and the talus bone of the ankle. Specifically, a tenon like feature in the talus bone (trochlea) fits into a mortise structure formed by the distal ends of the tibia and fibula, whose malleoli inhibit degrees of freedom in the frontal and transverse planes. Recently, however, it has been shown that the Subtalar joint is also slightly articulated during DF and PF [63]. Classically, a line drawn from the medial to the lateral malleoli forms the oblique axis about which the joint articulates. However, studies have indicated that the rotation axis of the Tibiotalar joint varies across its range of motion, resulting in proposals of multiple independent axes of rotation [64]. Three groups of ligaments provide stability to the joint: tibiofibular syndesmosis, deltoid ligaments and the lateral collateral ligaments.

The Subtalar joint (Talocalcaneal joint) is a diarthrosis contributing primarily to ankle mobility in the frontal and transverse planes primarily allowing for inversion, eversion of the ankle

[59]. The joint operates in a gliding motion rather than a hinge, making it also responsible for adduction and abduction of the foot. The transverse tarsal joint (Chopart's Joint) is a subset of the subtalar joint that is also articulated along with the subtalar joint, consequently contributing to the inversion and eversion of the ankle. The subtalar joint is mainly a junction between the calcaneus (heel bone) and the talus, with the talus resting on the anterior surface of the calcaneus. The main ligaments associated with the connection between the talus and the calcaneus are the interosseous talocalcaneal ligament, lateral talocalcaneal ligament and the anterior talocalcaneal ligament. Figure 4A depicts the rotation axis for DF/PF and Figure 4B depicts the rotation axis for inversion/eversion.

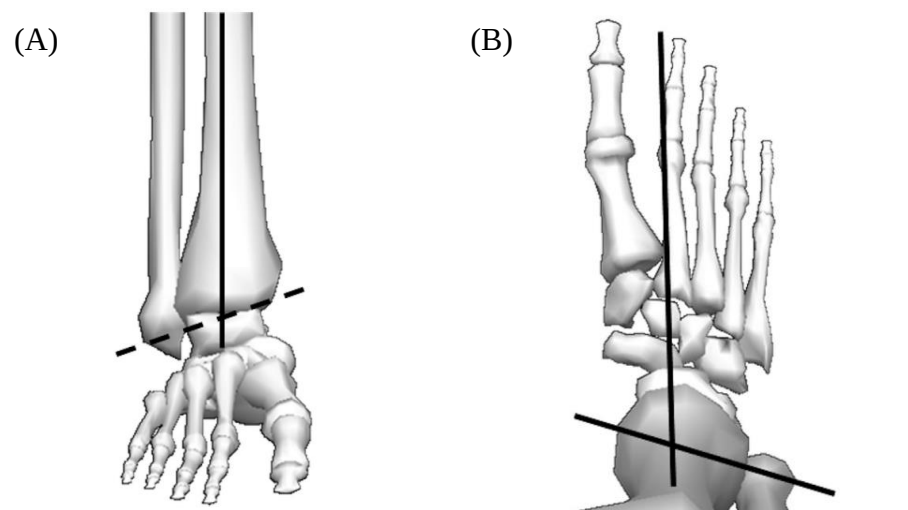


Figure 4 : (A) Depicts the ankle rotation axis for DF/PF (dashed line) and (B) depicts the ankle rotation axis for inversion/eversion. The intersection of the dashed line and solid line in (A) forms the origin for inversion/eversion axis. Figures adapted from [59].

Four muscle compartments comprising of twelve different extrinsic muscles provide forces necessary to actuate the foot about the Subtalar and Tibiotalar joints. The anterior compartment, which consists of the tibialis anterior, the extensor digitorum longus, the extensor hallucis longus, and the peroneus tertius muscles primarily produce DF movements about the Tibiotalar joint, along with some contributions to eversion and inversion about the subtalar joint. The posterior compartment consists of three muscles: the soleus, the gastrocnemius, and the plantaris muscles, all of which contribute primarily to PF movements about the Tibiotalar joint. EMG measurements in a laboratory setting are measured from these muscles, more commonly from the soleus muscle

to understand active contributions to stretch reflex during dorsiflexion. The lateral compartment comprises of two muscles: the peroneus longus and the peroneus brevis muscles, which contribute to the PF motion about the Tibiotalar joint along with some contributions to eversion about the subtalar joint. Lastly, the deep posterior compartment comprises of three muscles: the tibialis posterior, the flexor digitorum longus, and the flexor hallucis longus which produce PF and inversion about the Tibiotalar joint and the Subtalar joint respectively [59], [61], [65].

1.3.2. ANKLE RANGE OF MOTION

The range of motion (ROM) of the ankle is dependent on the geometry of the articulating surfaces in the joints, musculotendinous structure differences in the two joints, ligamentous attachments, joint capsules and other soft tissue components [66]. This dependency on the many anatomical features causes the range of motion between humans to vary. Age and gender related differences along with difference in physical activity levels attribute to the variance in anatomical differences in the ankle joint. The following Tables 4 and 5 depict the ankle ROM (DF/PF, inversion/eversion) data as a function of age and gender obtained from expansive studies conducted by Grimston et al. [66] and Nigg et al. [67]. Presence of ROM inhibiting pathologies like SCI were treated as exclusion criterion for subjects in both the studies.

Table 4 : Average ankle ROM values in the saggittal plane as a function of age and gender.
Adapted from [67]

Gender	Age Ranges	Dorsiflexion(DF)			Plantarflexion (PF)		
		Avg	Range (min,max)	Std dev.	Avg	Range (min,max)	Std dev.
Female	21-39 (N=15)	26	18.0 ; 43.4	1.7	48.2	32.9 ; 57.0	1.5
	40-59 (N=15)	23.7	13.6 ; 33.0	1.5	40.6	19.5 ; 52.2	2.4
	60-69 (N=15)	20.6	11.5 ; 33.0	1.2	41.6	29.4 ; 54.9	1.8
	70-79 (N=15)	18.5	9.9 ; 26.1	1.3	40.5	31.4; 52.9	1.5
Male	21-39 (N=14)	25.2	15.0 ; 42.0	1.9	41	18.0 ; 54.0	2.7
	40-59 (N=15)	27.1	18.0 ; 38.8	1.5	40	30.0 ; 55.3	1.6
	60-69 (N=15)	25.4	16.7 ; 31.9	1.2	36.2	26.3 ; 47.0	1.4
	70-79 (N=15)	26.4	17.4 ; 36.7	1.2	33.7	19.1 ; 47.8	1.7

Table 5: Average ankle ROM values in the frontal plane as a function of age and gender.
Adapted from [67]

Gender	Age Ranges	Inversion			Eversion		
		Avg	Range (min,max)	Std dev.	Avg	Range (min,max)	Std dev.
Female	21-39 (N=15)	21.4	10.1 ; 35.8	1.9	17.2	8.2 ; 27.0	1.3
	40-59 (N=15)	19.6	9.8 ; 34.0	2	14.2	7.7 ; 19.0	0.8
	60-69 (N=15)	15.3	8.4 ; 30.2	1.4	11.6	4.0 ; 15.5	0.8
	70-79 (N=15)	15.3	5.2 ; 28.5	1.4	11.4	6.1 ; 17.9	0.9
Male	21-39 (N=14)	19.6	11.2 ; 32.0	1.7	13	6.0 ; 20.0	1
	40-59 (N=15)	21.8	11.0 ; 41.6	2.1	13.3	2.8 ; 17.7	0.9
	60-69 (N=15)	19	8.0 ; 25.5	1.4	13	6.0 ; 18.8	0.9
	70-79 (N=15)	18.6	8.6 ; 28.8	1.4	11.4	5.1 ; 15.4	0.8

From the data above, it is evident that DF/PF, inversion/eversion and abduction/adduction ROM decreases as a function of age. Grimston et al. propose that this decrease in ROM is possibly due to reduction of physical activity levels. Since SCI and other UMN lesion syndromes can drastically reduce activity levels of the foot, it can be assumed that ankle ROM for SCI patients would be less than healthy individuals. Additionally, ROM for individuals with UMN lesions is lesser than healthy patients [68].

1.3.3. ANKLE JOINT STIFFNESS

In classical biomechanics, a joint's stiffness is influenced by the viscoelastic forces produced by muscles, tendons and ligaments that connect them to other bones [69]. Joint stiffness in a biomechanical context can be defined as the joint position-dependent component that stores or releases energy in a joint [70]. Care should also be taken in distinguishing stiffness from quasi-stiffness which is the derivative of the torque-position relationship at a certain joint [71]. In this work, ankle joint stiffness shall always refer to the torque-angle relationship for DF/PF motion about the Tibiotalar joint. The angle formed by the plantar surface of the foot with respect to the frontal plane during DF/PF shall be termed as ankle angle. Positive ankle angles will denote DF

while negative ankle angles will denote PF, and zero ankle angle is defined as the position where plantar surface of the foot is parallel to the transverse plane.

Ankle joint stiffness is comprised of two components: passive stiffness and active stiffness. Both components contribute to the stability of the ankle joint during activities of daily living. In this work, passive stiffness at the ankle joint is the resistance to DF/PF motion without any muscle activation that provide torque to the joint for pronation/supination. Passive stiffness is believed to originate from the visco-elastic elements of muscle (tendons, structural proteins within the myofibril, and connective tissue around the muscle fibers) [72], [73]. Active stiffness at the ankle joint is the resistance to DF/PF caused by stretch reflex mechanisms that activate muscles in order to resist contraction. With the classical definition of spasticity as provide by Lance et al. [46], increase in the velocity dependent stretch reflex i.e. increase in active stiffness at the ankle contribute to the overall increase in ankle joint stiffness. However, recent studies have shown that both active and passive components increase in patients diagnosed with ankle spasticity [74]. As the ankle joint stiffness (both passive and active) can have variances based on patients age, anatomical differences, gender and general physical activity, no universal consensus on joint stiffness values exists.

In literature, many researchers have calculated the passive joint stiffness as a linear relationship between torque and ankle angle [73], [75], [76] while some have developed a nonlinear model (double exponential function) [71], [77], [78] that matches closely with experimental measurements of torque-ankle angle relationship at various hip and knee joint positions. Others have reported modelling passive ankle joint stiffness as a 2nd order dynamic system, having angle dependent inertial, vicious and elastic parameters [79]. Roy et al. [73] and Singer et al. [80] considered a simplified piecewise passive ankle stiffness over ankle ROM using the equation:

$$T_{ankle}(\theta) = \begin{cases} K_{DF} \times \theta & \theta \in [0,20] \\ K_{PF} \times \theta & \theta \in [-45,0] \end{cases} \quad (1)$$

Where K_{DF} is the stiffness in the DF range of the ankle and K_{PF} is the stiffness in the PF range of the ankle. Chung et al. [76] reported quasi - stiffness of the ankle joint for both spastic and healthy patients using a similar formulation as equation (1). The data gathered from the researchers have

wide variances, with standard deviations almost equal to the nominal values themselves. The reasons behind these can be due to the physiological differences in ankle muscle structure between subjects (data was not normalized by anatomical differences between subjects), different testing methodology and spurious activations of ankle muscles that were not picked up by EMG sensors. The linear mappings and quasi-static relationships of passive ankle torque to ankle angle are tabulated below:

Table 6: Passive stiffness of the ankle joint values from literature.

Authors	Ref	Tested ankle angles	Knee flexion angle (position - deg)	Test speed (deg/s)	Subject Pathology group	Subject Ages	K_{PF} (Nm/deg)	K_{DF} (Nm/deg)
Chung et al.*	[76]	at 10°DF at 30°PF	Seated - 60°	NA	Healthy (N=32)	42.1±20.5	0.11±0.09	0.35±0.2
					Stroke (N=24)	55.3±10.1	0.2±0.14	0.54±0.19
Singer et al.*	[80]	10-15°PF 10-15°DF	Supine - 90°	5	Healthy (N=18)	17-48	0.03±0.01	0.44±0.21
					Brain Injury (N=13)	22-50	0.07±0.05	0.53±0.36
Roy et al. ⁺	[73]	full ROM	Seated - 45°	5	Healthy 1 (N=10)	32±5	0.31±0.12	0.52±0.31
					Healthy 2 (N=10)	59±6	0.22±0.01	0.65±0.26
					Stroke (N=10)	60±8	0.26±0.06	0.83±0.20

Symbols * indicate quasistiffness calculations and ⁺ indicates linearly fitted stiffness results

Another reason for the discrepancies between the stiffness values seen in Table 6 stiffness parameters may be dependency on speed of ankle joint articulation and initial knee flexion angle. Therefore, merely using linear variables like K_{DF} and K_{PF} may not be the most accurate to model stiffness. This indicates that stiffness parameters might have dependency on ankle angle. Therefore, nonlinear models may be more appropriate, especially when a robotic system would potentially have signal readings of angle joint angle, speed and torque in real time. This warrants a modelling-based approach, supported by literature to accurately determine the ankle angle torque

relationship. A deeper study of ankle joint stiffness is detailed in section 2.1.1 (methods) and results of modelling the ankle torque relationship is provided in section 3.1.1 (results).

Regarding active stiffness, it is known that there exists a dependency on joint angle [81] and joint velocity [82]. Active stiffness of the ankle is often reported by measuring the change in ankle torques with different amplitudes of EMG signals obtained from gastrocnemius and soleus muscles during PF and Tibialis Anterior, Triceps Surae muscles during DF. During the administration of clonus, active muscle properties will be relevant after a certain reflex threshold velocity has been reached, where reflex resistance to the DF stretch is exhibited by contraction of the Gastrocnemius and Soleus muscles. At lower velocities, the component of active stiffness contributing to ankle torque will be minimal.

Stretch reflex behavior is known to be elicited at certain stretch velocities. For ankle clonus, however, few studies exist that have determined the stretch reflex threshold velocity. The rotation speed of the foot about the Tibiotalar joint at which stretch response is activated was investigated by Lorentzen et al.[74]. Seventeen different stretch velocities ($8^{\circ}/s$ to $200^{\circ}/s$) were tested among individuals with UMN lesions ($N=56$) and healthy individuals ($N=31$) and EMG signals were obtained from the soleus muscle. It was determined that the average ankle joint velocity at which a reflex response was evoked in spastic subjects was $77.1 \pm 9^{\circ}/s$ (1.35 ± 0.15 rad/sec) while average threshold velocities for non spastic subjects and healthy controls was between $70.3 \pm 9^{\circ}/s$ and $74.3 \pm 7^{\circ}/s$ (1.23 ± 0.15 and 1.29 ± 0.12 rad/sec) respectively. However, the authors caution that the overall group stretch velocities between spastic and healthy patients was not significantly different. This can be due to several physiological differences among participants. Another study by Muramatsu et al. [83] reported stretch reflex activity to begin at $80^{\circ}/s$, which is consistent with the reflex speed determined by Lorentzen et al.

1.4. SIMULATOR ARCHITECTURES

1.4.1. EXISTING CLONUS SIMULATORS

There exist very few devices that physiologically attempt to recreate ankle clonus for clinicians to train on. Additionally, no commercially available clonus simulator exists. An exoskeleton device that provided clonus like behavior was prototyped by Okumura et al. [84] (Figure 5A). The exoskeleton device relied on the intrinsic muscle dynamic stiffness of the wearer and therefore only provided an additional torque that simulated the resistance felt by a clinician during a clonic response. However, several limitations were evident in their design. First being the lack of support of plantar flexion torque by the device. In a real patient, the gastrocnemius and soleus muscle resist fast dorsiflexion and provide a reflex resistance torque about the ankle joint axis during clonus. The lack of this torque modeled into their device might create a different haptic feedback than that experienced during clonus in human patients. The second limitation that stood out was the lack of clear justifications for choosing the resistive torque values that were applied by the device. In real patients, the reflex resisting torque about the ankle joint axis can be a complex function of the ankle joint angle and rate of ankle joint rotation during dorsiflexion/plantarflexion. In the exoskeleton device, the ankle joint stiffness values during and preceding clonus were modelled as constant resistances and did not employ a viscoelastic model that more accurately describes ankle joint dynamics [79], [85].

A complete robotic clonus simulator was built by Kikuchi et al [86] (Figure 5B). The electromechanical leg-shaped device made use of a magnetorheological fluid clutch to vary the ankle torque experienced by the clinician during different stages of the algorithm that simulated clonus. The control sequence used to actuate the DC motor and fluid clutch followed a similar protocol as the one seen in [84]. Critical parameters like threshold stretch speed to initiate clonus and clonus torque were set by experience physicians. However, there are few drawbacks for the robotic ankle simulator as well. Firstly, the device lacks an ergonomic foot shape and has a large deviation from average human foot inertias. Especially, since this device was ‘calibrated’ by clinicians, the parameters selected may not match real human patients due to the variances in the way clinicians may perceive a different inertia than foots they are used to assessing on. Secondly,

there exists no experimental validation that compares the simulator to some data gathered from a clonus afflicted individual's ankle.

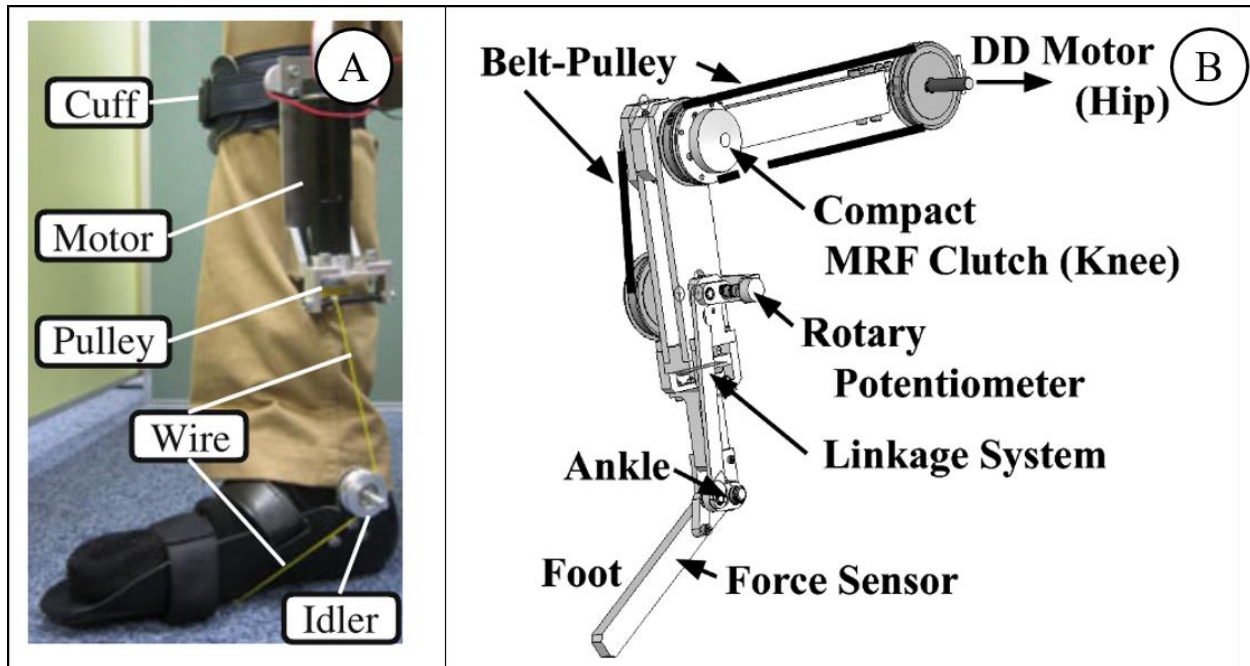


Figure 5: Existing clonus simulators. (A) A wearable simulator developed by Okumura et al. [84]. (B) The structure of a novel simulator built by Kikuchi et al [86], employing a magnetorheological fluid clutch

The limitations listed in the two existing clonus simulators can be treated as areas that can be improved upon in our own simulator design. The selection of appropriate control mechanisms for a robotic device directly dictates its quality of performance. Robot control mechanisms suitable for large industrial robots may not always be applicable to high accuracy requirements of medical applications. Therefore, for the ankle clonus simulator, where haptic feedback of the clonus simulator is of prime importance, actuation using a Series Elastic Actuator (SEA) were chosen. The following section details our design and explains how SEA mechanism can better suit biomechanical robotics applications.

1.4.2. OVERVIEW OF GEN 1 ANKLE CLONUS SIMULATOR

The ankle clonus simulator being developed in the Human Dynamics and Controls Lab (HDCL) in the Department of Mechanical Science and Engineering at the University of Illinois will use a different approach lying between purely mechanical (springs and dampers) and electrical (motors) designs to generate resistance with active force control by using a series elastic actuator (SEA), which has the advantages of both.

The other actuation methodology that was initially considered was a Quasi-direct drive (QDD) motor supplemented with a transmission system. However, it was soon realized that a significant challenge would be met to assemble the mechanism within the size constraints of a human male foot and body due to requirements of a large electric motor that can operate at torque and speeds required.

Pioneered by Pratt et al. [87], SEAs have seen increased research interest recently, primarily due to the advantage of providing highly accurate force control for dynamic robotic applications. The need of utilizing SEAs for robotics applications arose due increasing need for higher stiffness and lower impedance of loads connected to motors for more accurate and high bandwidth force control. However, for smaller scale applications, smaller motors that have low output torque capacity would need to operate at higher speeds (hence lower torques) and are often supplemented with high reduction gear transmissions systems [88]. SEAs can help bridge the gap for high fidelity force control needs and small form factor designs by intentionally adding an elastic member, whose forces can be monitored in real time, in series between the actuator and the load [89]. This increased compliance increases force control fidelity, allows for near zero impedance (or inertia), and provides a limit to the maximum impedance of the actuator at high frequencies.

However, the caveat from using SEAs is that the bandwidth of operation will be reduced, and hence only suited for operation in lower frequencies. Low bandwidth limitations are non-consequential, however, for robot devices in the biological space (simulators, exoskeletons, rehabilitation robotics, etc.), where actuation speeds are rarely too high to be controlled. The performance of a SEA mechanism is directly dictated by accurate measurement of the forces in the elastic elements and appropriate friction compensation. Most SEA designs have measured the displacements in the elastic members using high resolution sensors like strain gages [87], [90], [91] or linear displacement potentiometers [92] attached parallel to the elastic element.

Consequently, elastic members that are trivial to model like torsional springs and compression springs are most common. For our ankle clonus simulator, two compression springs, modelled using Hooke's law and having high spring stiffnesses have been employed. The higher stiffness in our linear springs provide a larger control bandwidth, such that clonus frequencies falling between 5-8 Hz can be easily reproduced. Additionally, the springs are assembled in a 'spring cage', where a slider block is situated between the two compression springs and coupled to the internal structures using low friction linear bearings.

The primary actuator in the ankle clonus simulator was chosen to be a brushless direct current motor (BLDC), in order to utilize a small form factor motor and to provide sufficient torque. It was also determined that a gear transmission system would need to be supplemented along with the BLDC motor. Loosely inspired by anatomical findings of the ankle joint complex, the simulator would constitute of an anatomically accurate outer shell that encloses rigid metallic structures. The down selection from viable mechanisms and design of the mechanical portions of the simulator was the work of Yinan Pei (ypei2@illinois.edu). An initial design (version 3) of the ankle clonus simulator is shown in Figure 6 below, highlighting mechanical design choices made. In later design revisions, the coupler was modified to be parallel to the spring deflection axis so that the control scheme would not require nonlinear modelling.

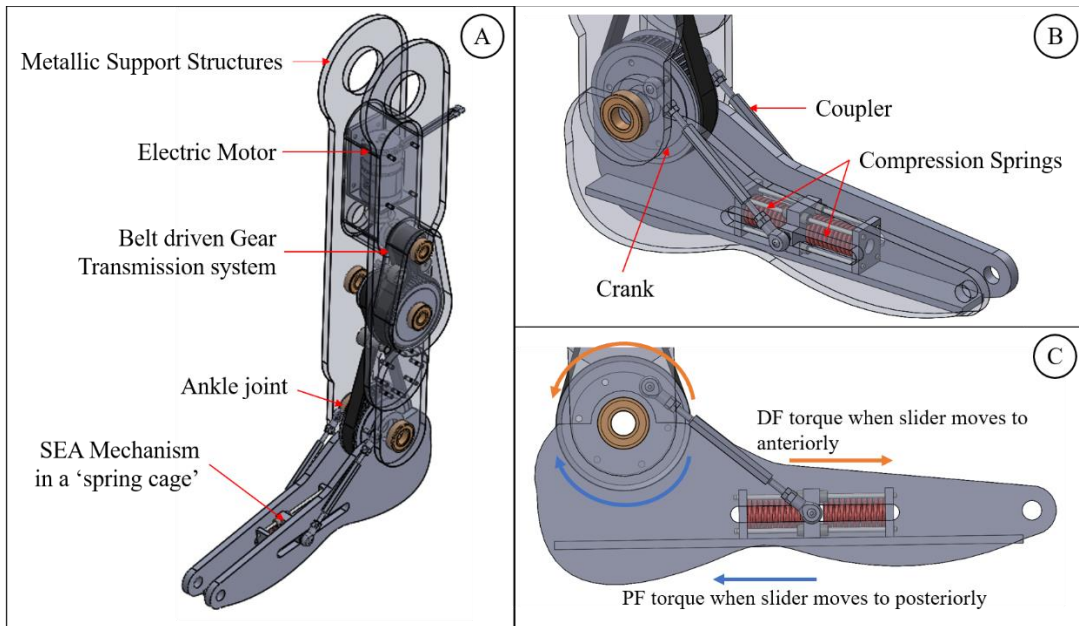


Figure 6: Internal design of ankle clonus simulator. (A) Main components of the simulator. (B) Crank-slider type SEA mechanism. (C) SEA torque control functionality.

Based on control bandwidth needing to be at least 10 Hz, the resultant spring stiffness was chosen to be around 170 Nm/mm. Through the SEA mechanism, the ankle simulator would have the capacity to track 15.1 Nm of ankle torque but can be increased up to 25 Nm by simply changing the gear ratios (by selecting different belt drive pulley diameters). Based on the crank slider analysis for this design, where the lever arm of the crank is designed to be 45mm, supporting 15 Nm of torque would require springs to have deflections of +1.96 mm. Manufacturing of the simulator would require machining the structural elements out of aluminum sheets. The shrouds were 3D printed in segments and attached using fasteners to the support structures and relevant mechanisms. The design shown in Figure 6 was then iteratively modified to converge at a complete simulator design (version 26) depicted in Figure 7.

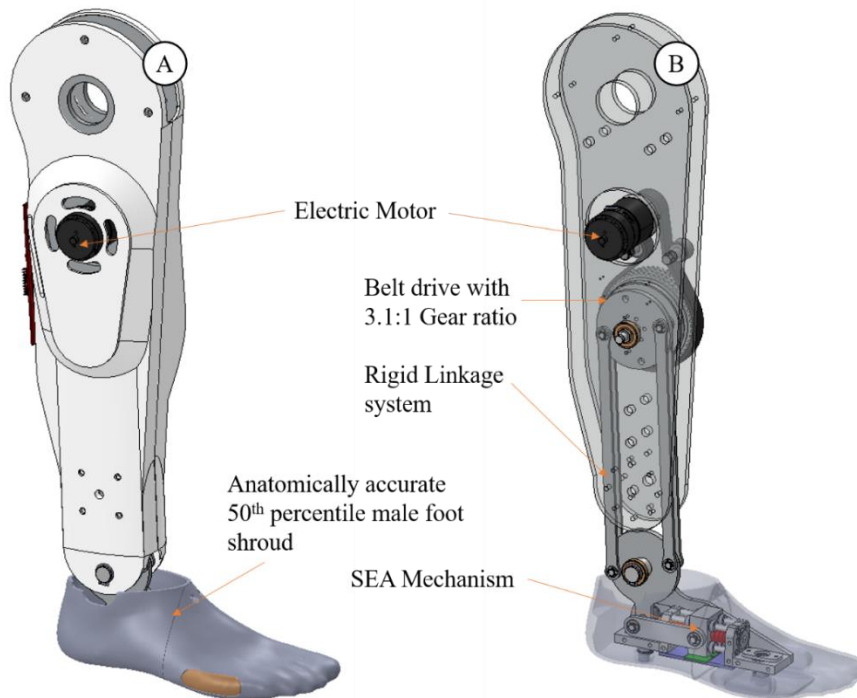


Figure 7: Final design of the first-generation ankle clonus simulator. (A) The complete assembly. (B) Internal mechanisms.

Having decided on the general actuation strategies of our simulator, the next step was to consider mechatronic requirements to ensure appropriate functionality of the ankle simulator. The following section therefore considers the specific research questions and corresponding electromechanical requirements that need to be considered to develop a more physiologically accurate ankle clonus simulator based on our architecture.

1.5. RESEARCH QUESTIONS

Broadly, the challenge is to recreate the behavior of an ankle joint as closely as possible to a 50th percentile American male demonstrating clonus. Based on what we know about ankle clonus, clinical methods, ankle anatomy and simulator architecture in the preceding sections, the following research questions arose:

- 1) What considerations need to be made with regard to the biomechanics and anatomy of the ankle and foot to facilitate the development of an ankle clonus simulator?
- 2) What are the required electromechanical signals, and the corresponding optimal sensor systems in order to:
 - a. measure and control the force applied on the ankle by the learner so that the simulated clonus response correctly mimics real patient behavior?
 - b. measure the speed of the ankle joint of the simulator, upon actuation by the learner during clonus simulation procedure?
 - c. estimate the learner's hand position on the forefoot of the simulator external structure (shroud) during clonus assessment training so as to the better advise the learner on proper clonus test administration techniques?

These research questions constitute the backbone of this thesis research and will inform the engineering decisions during the development process of a SEA-based ankle clonus simulator. The specific measurements that were deemed necessary to answer the research questions posed above are elaborated in the following section.

1.5.1. SPECIFIC REQUIREMENTS BASED ON RESEARCH QUESTIONS

In the biomechanical literature, it has been established that passive and active torques about the ankle joint are dependent on the amount of stretch of the muscles involved and hence the ankle joint angle. Therefore, of primary concern is an accurate measurement of the rotation angle about the ankle joint and consequently the accurate torque about the ankle joint as a function of the measured angle. The torque at each angular position should track the required torque-angle relationship at a bandwidth at least twice as fast as the fasted clonus beat frequency. Based on choice of using a SEA mechanism for force control, the critical real time signal required is the spring deflection. Additionally, the structures used to provide feedback of the ankle angle cannot

be too large, since all the subassemblies are required to be able to fit comfortably inside the silhouette of a 50th percentile human American male foot.

In order to be anatomically accurate, the simulator will need to recreate the same ankle ROM as a human individual. Consequently, since the simulator must support both healthy and clonus patient mechanics, the DF/PF range of motion for the simulator was set to match that of the average ROM for healthy males around the ages of 40-70. Based on ROM values in Table 4, this would equate to roughly 26° of DF about the neutral position (plantar face of ankle parallel to the horizontal) and 37° of PF about the neutral position.

Additionally, it was understood early in the development process that the clonus simulator should allow for a secondary degree of freedom of rotation in the frontal plane, presenting inversion/eversion movements of the ankle. The extra degree for freedom is critical since clinicians should be aware of the importance of keeping the patient's foot as close as possible to zero inversion/eversion angle. Therefore, it would be critical to monitor the inversion/eversion angle of the foot structure on the simulator. Similar to DF/PF range constraints, we targeted the inversion/eversion angle of the simulator to meet the average ROM reported for males aged 40-70 years (Table 5). Therefore, the engineering goal would be to achieve around 19.8° and 12.6° ROM for inversion/ eversion respectively.

Of tertiary concern is the ankle speed at which stretch response is elicited in individuals with UMN lesions. A stretch reflex threshold speed is seen in the elbow joint for spasticity and has been studied widely. Considering existing knowledge of clonus causing mechanisms (listed in section 1.1.3), clonus is most likely the cause of self-reinforcing reflex response to a fast stretch of the gastrocnemius and soleus muscles. For the purposes of the ankle clonus simulator however, the data from [74] are adequately informative and the stretch reflex threshold velocity (ω_{thres}) for this work is selected as 77.1°/s. The chosen stretch reflex speed for our simulator is lower than the 4 rad/sec (229°/s) threshold speed chosen by other ankle clonus simulators [84], [86], where the speeds were set qualitatively by physiologists and not supported by scientifically accurate measurements by researchers.

Lastly, it is crucial to estimate the pressure applied by the clinician on the plantar surface of the patient's foot (ball of the foot) during the clonus assessment procedure. In order to sustain clonus, clinicians have to rest their palms/fingers and apply sufficient pressure on the metatarsal

head regions of the patients after a brisk dorsiflexion to feel the haptic feedback from any beats generated by ankle during clonus. The focus of estimating the examiner's hand placement was proposed by clinical practitioners involved in the training of medical students. Additionally, neurologists and experienced clinicians were consulted to validate the engineering decision to include the additional sensors. It was also deemed necessary to estimate the learner's hand placement along the anterior regions of the foot (specifically the locations of the fingers and thumb) along medial and lateral surfaces of the forefoot.

2 METHODS

2.1. BIOMECHANICS AND FOOT ANTHROPOMETRY

From information gathered from literature (section 1.3.3), it was learned that the passive stiffness of an ankle can be better understood using a nonlinear dependency of ankle torque on ankle stiffness. To provide an adequate answer to the first research question, choosing the correct passive muscle biomechanical model is critical. Additionally, precise anthropometric data have to be utilized to create an anatomically accurate foot design for the clonus simulator. Therefore, the following sections detail (1) the nonlinear ankle stiffness models considered and (2) the methodology employed to produce an anatomically accurate foot shroud design.

2.1.1. ANKLE ANGLE-TORQUE RELATIONSHIP

To select the best model for our simulator, we tested two different methods. The first, was a viscoelastic stiffness model established by Weiss et al. [79] where passive ankle torque ($T_{passive}$) was modelled as a 2nd order dynamic system, having angle dependent inertial $I(\theta)$, viscous $B(\theta)$ and elastic $K(\theta)$ parameters, with θ denoting the ankle joint angle:

$$T_{passive}(\theta) = I(\theta) \cdot \ddot{\theta} + B(\theta) \cdot \dot{\theta} + K(\theta) \cdot \theta \quad (2)$$

Due to the ability to provide torque-ankle angle relationships based on ankle actuation speed, passive ankle torque ($T_{passive}$) for the healthy individuals in the ankle clonus simulator was initially modelled by Equation 2. To derive the quantities expressed in Equation 2, exponential functions were used to fit data reported in [79] by obtaining expressions for angle dependent passive stiffness parameters $B(\theta)$ and $K(\theta)$. Modelling was performed in MATLAB (R2016b, MathWorks Inc., Natick, MA) with the use of the curve fitting toolbox.

However, the data from Weiss et al. are older (1984) and therefore, newer models may be better suited for the clonus simulator. To verify the viscoelastic model, a second passive stiffness model was derived from Reiner et al. [77], where double exponential functions were used to determine the ankle torque at different knee flexion angles (Equation 3). The study conducted by

Reiner et al. does not clearly mention what actuation speeds were involved but were reported to be lower than $10^\circ/\text{s}$. The equation developed by Reiner et al. is provided in Equation 3.

$$T_{passive,healthy}(\theta, \varphi_{knee}) = e^{2.35 - 0.085 \theta - 0.021 \varphi_{knee}} - e^{-7.8 + 0.19 \theta} - 1.6 \quad (3)$$

The Reiner et al. model may be more accurate due to good agreement with collected experimental torque values (0.24 – 0.29 mean square error). However, this model does not differentiate between inertial, viscous and elastic components and hence would not support passive muscle properties with ankle speed dependencies. The difference between the two models were compared to inform the decision of choosing the correct passive stiffness model for healthy patients.

For the clonus case, due to lack of viscoelastic data available from literature, double exponential models for torque angle relationships were used to determine the torque angle relationship. The models would not have any ankle speed dependence. The parameters for this model were based on a least squares fit to experimental data conducted by Edrich et al. [69] on patients with spinal cord injury. The model also included the effect of knee flexion angle on the passive torque response (Equation 4).

$$T_{passive,clonus}(\theta, \varphi_{knee}) = e^{2.01 - 0.110 \theta - 0.0300 \varphi_{knee}} - e^{-8.070 + 0.189 \theta} - 0.950 \quad (4)$$

To understand the difference in passive torque for a spastic ankle and a healthy individual, we used a modified version of Equation 4 based on data published by Edrich et al. (Equation 5).

$$T_{healthy\ controls}(\theta, \varphi_{knee}) = e^{2.1 - 0.0830 \theta - 0.0176 \varphi_{knee}} - e^{-7.96 + 0.195 \theta} - 1.792 \quad (5)$$

It is known that the passive stiffness among spastic patients increases with stretch speed. For the clonus case, it was initially decided that a spastic ankle torque response with movement speed dependency would be employed. An initial plan was that this model could be derived from the work of Edrich et al. [69], which gives a double exponential function dependent on ankle angle and knee flexion angle. However, this model does not have any speed dependent terms either and hence cannot model the passive ankle properties in the clonus case at different speeds (below the stretch reflex threshold). In most active stiffness studies, reflex spasms like clonus have been

treated as an exclusion criterion. Few establishments of torque-angle relationship during ankle clonus have been made [18],[40]. From both studies ([18],[40]), it was found that patients demonstrating clonus had increases in passive muscle properties and motor neuron activation delay as compared to healthy patients. An increase in reflex muscle forces among spastic ankle subjects was evident, but it was theorized that this increase in force was due to the changes in neural activation rather than intrinsic active muscle behavior (force-length and force-velocity parameters) [40]. Additionally, change in passive stiffness of muscle properties is far larger than change in active muscle behavior. Due to time constraints, for the clonus case, it was decided to investigate reflex torque based models developed by Vlught et al. [93] and Hidler et al [5] at a later time in next steps, with passive stiffness models being employed on the clonus simulator as a start.

2.1.2. HUMAN FOOT MODELLING

The design of the outer shell had to match the dimensions of a 50th percentile Caucasian male foot to have an anatomically accurate representation for clinicians to train on. With that goal in mind, it was decided to utilize some anatomically accurate models, scan them digitally and design our internal mechanisms around those constraints. First, a literature search was conducted to obtain dimensions of a 50th percentile Caucasian male foot. The largest anthropometric database for foot measurement found was the expansive measurements of US army recruits taken from 1976-1981 by White [94].

Table 7: 50th percentile American army male foot measurements based on [94] along with the measurements from the current foot shroud model.

Dim no.	Measurement Name	50th percentile male - mm	Standard Deviation (mm)	Current Foot shroud (mm)
1	Foot Length	267	13	266
2	Instep Length	196	10.4	193
3	Foot Breadth	98	5.5	93
4	Heel Breadth	68	4.7	64
5	Bimalleolar Breadth	74	3.9	71
6	Ball of foot Circumference	251	13.1	247
7	Instep circumference	265	16.5	274
9	Lateral Malleolus Height	71	5.4	80

The current foot model is well within the standard deviations for the important dimensions considered.

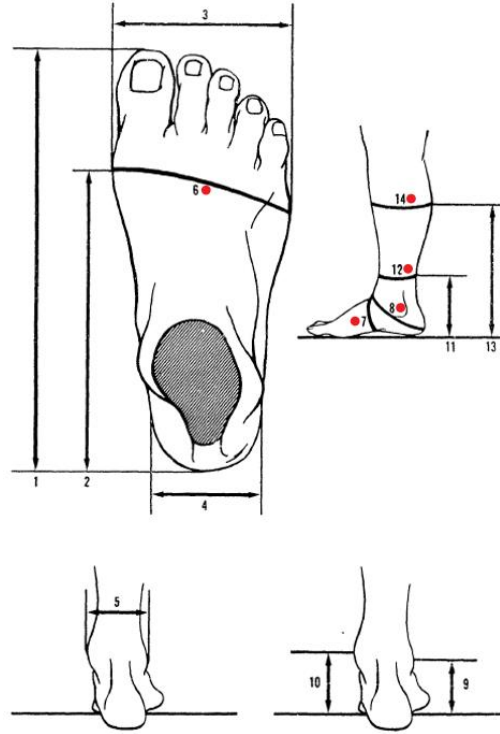


Figure 8: Anthropometric male foot measurements taken by White [94]. Dimensions with a red dot are circumferential measurements

Although this database could contain measurements from a mixture of racial categories, due to the population of the US, it is assumed that the large majority of the recruits were Caucasian, and therefore this database was appropriate for this simulator project. Table 7 and Figure 8 show the measurements taken into consideration to design the foot model. To accurately model the foot shroud for the simulator, the team acquired a prosthetic foot (male size 10 shoe) (Omni Prosthetics, Urbana, IL, USA) that matched as closely as possible to the 50th percentile male dimensions seen in Table 7. We then proceeded to perform a high-resolution 3D scan of the prosthetic foot and imported it into a computer aided design software (Solidworks 2019, Dassault Systèmes, Vélizy-Villacoublay, France) to modify the CAD such that its dimensions closely matched that of a 50th percentile male foot. Additionally, stiffening structures were added and cutouts were made so that we could allow as much inversion/eversion of the foot shroud as possible. Since the learning opportunity for clinician learners is basically focused on maintaining a near zero inversion/eversion state of the foot during the assessment procedure, it was decided that the engineering limitations for inversion/eversion ROM due to the design of the simulators mechanical components and structural frame would be acceptable.

2.2. SENSOR SELECTION AND DESIGN PROCESS

Based on Research Question 2, optimal sensors systems for measuring applied force, movement speed, and rotation angles (DF/PF and inversion/eversion) had to be selected and designed into the simulator. Additionally, sensors capable of differentiating different hand positions on the forefoot shroud of the simulator had to be investigated.

Regarding Research Question 2a, the SEA mechanism is the main force control system in the ankle simulator, and therefore the accurate real-time measurement of linear spring deflection is critical. Sagittal plane (DF/PF) angular position and speed, and Frontal plane (inversion/eversion) angular position are to be measured as per Research Question 2b.

Sagittal plane ankle angle will be estimated by using the signal from the motor encoder and assuming that there is no slipping along the motor, belt transmission and DF/PF axis of rotation. This ankle angle will be digitally differentiated to determine the ankle joint velocity ω_{ankle} and ankle joint acceleration α_{ankle} . If in the future it is found that the fidelity of the encoder signal may not be good enough to estimate DF/PF angle, an additional rotary magnetic encoder around the DF/PF rotation axis will be considered, which can be easily disassembled from the simulator if deemed unnecessary.

Measurement of ankle angle in the frontal plane (inversion/eversion) would require a separate high-resolution angle sensor.

Lastly, as per Question 2c, the most viable option was to measure the touch force across different areas of the clonus simulator foot shroud.

The specific processes used to define the designs of the different sensors are detailed in the following sections. The individual sensor systems were designed to be processed using a digital signal processing microcontroller (MCU) specifically manufactured for control system applications. The MCU chosen was LAUNCHXL-F28379D (Texas Instruments, Dallas, TX, USA). However, for simplicity and ease of computation, preliminary sensor validation tests were conducted on smaller MCUs. The overall electrical schematics for all of the sensors involved are shown in Appendix B. The C++ code for all experiments can be obtained from the project code repository [95].

2.2.1. LINEAR SPRING DISPLACEMENT MEASUREMENT

The important feedback measurements needed from the SEA mechanism are the compression in each of the springs, which is used to estimate the force being applied by the learner. A few sensor solutions were considered, and their operating conditions are listed in Table 8. Space requirements were determined based on volume of the sensor compared to volume of the SEA subassembly, with high being closer to 50%, and low being around 10%. Assembly complexity considered the details of sensor operation and failure mechanisms during operation.

Table 8: Sensor types considered for measurement of spring deflection in the SEA mechanism.

Sensor Type	Typical electrical output	Relative space requirements	Typical operating ranges	Relative assembly complexity	Accuracy (% of range)
Linear potentiometer	DC Voltage	High	10 mm to 150mm	Low	10-15%
Optical distance sensors	Digital signal	High	5 mm to 80 mm	High	5-10%
LVDT (linear variable differential transformer)	AC voltage	Medium-Low	5 mm to 50 mm	Low	1-2%
Magnetic linear sensor	Digital signal	Low	>1 mm	Low	0.1-2%

Through the dynamic analysis of the SEA mechanism, the total displacement that the spring assembly would undertake during its operation was determined to be ± 1.85 mm. Therefore, if we require a force error less than 2%, the sensor selected would have to have an accuracy target of roughly ± 0.038 μ m. This meant that only LVDT and magnetic sensors were viable for our purposes. Consequently, it was decided that a magnetic sensor would be the best solution for the intent of measuring small linear displacement. The specific sensor selected was the AMS AS 5311 high resolution magnetic linear sensor (Austria MicroSystems AG, Premstätten, Austria), which is located on an integrated circuit board. The sensor operates using internal Hall effect elements whose voltages vary based on the change in magnetic field direction across their surface. Therefore, a linear, multiple pole magnetic strip or tape was secured with epoxy glue on the spring

block, and the magnetic sensor integrated circuit board was secured via tape to a 3D printed bracket spanning the spring cage at a fixed distance of 0.3 mm away from the magnetic tape surface. As the spring block undergoes small displacements, the magnetic tape will apply a varying magnetic field on the Hall effect elements in the sensor, therefore changing the final output reading. The resultant design of the linear sensor to be included in the ankle clonus simulator is depicted in Figure 9. The distance between the linear magnetic encoder sensor IC face and the magnet tape surface can be carefully controlled using plastic shims such that it does not exceed 0.5 mm.

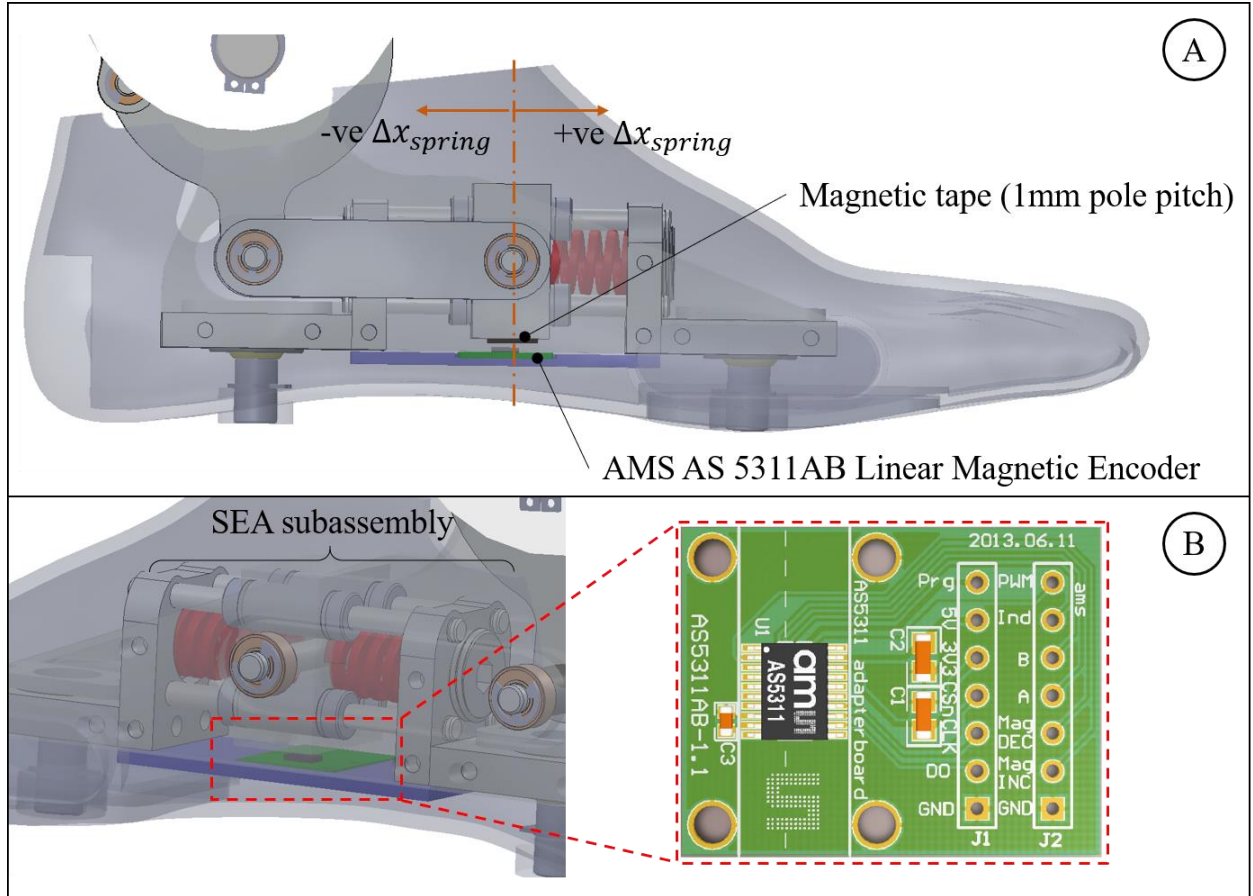


Figure 9: Linear sensor assembly. (A) depicts the side view and (B) depicts an isotropic view along with a real image of the linear magnetic encoder sensor IC (AS5311 AB). The magnetic tape is placed on the slider block at a nominal gap of 0.3 mm away from the sensor IC face, which was attached to a bracket spanning the spring cage. Other sensors and fasteners are not shown.

It was decided to read the incremental quadrature outputs from the sensor integrated circuit and then convert those readings in MCU code to determine the total spring displacement.

Therefore, during operation, once the displacement of the spring block is determined, spring force direction and magnitude can be computed.

2.2.2. INVERSION/EVERSION MEASUREMENT

Of specific need was to select and design a rotational sensor assembly that provided an accurate reading of the rotation of the foot shroud about the inversion/eversion axis. Several sensor options were considered, and their advantages and disadvantages are listed in Table 9. Space requirements were based on the ability of the sensor assembly to fit in the remaining space of the foot shroud after subtracting the space for the SEA mechanism. Assembly complexity also considered the required mounting constraints within the foot shroud.

Table 9 : Sensor types considered for the measurement of the inversion/eversion angle.

Sensor Type	Typical electrical output	Size smaller than foot shroud?	Relative assembly complexity	Accuracy ($\pm^\circ$)
Rotary Potentiometer	Voltage	Some sensors	Low	0.05-0.1
Capacitive Rotary Encoder	Voltage/Digital output	No	Medium	0.02
Magnetic Rotary Encoder	Digital	Yes	Medium	0.02

Since all of the sensor types listed in Table 9 had acceptable accuracy ratings, it was decided that a contactless magnetic rotary encoder sensor was the optimal solution since it required minimal assembly tolerance control and provided no resistance to inversion/eversion during operation. The specific sensor selected was the AMS AS5048 magnetic rotary encoder (Austria MicroSystem AG, Premstätten, Austria). This sensor also operates using Hall effect elements. Therefore, a rotation of the magnetic field vectors on the sensor face would provide a different sensor output. The sensor would require an appropriately sized diametrically magnetized cylindrical magnet so as the Hall effect elements can receive enough magnetic field to detect the change and also not become oversaturated due to the field being too high. Based on the operation of the sensor Hall effect elements, the diametric magnet would have to apply a magnetic field strength of 30 to 70 milli Tesla (mT) on the sensor face. An online simulation tool (NVE magnetic

field calculator, NVE corporation, Eden Prairie, MN, USA) was used to calculate the magnetic fields and therefore determine the range of distances allowable for several selected magnets. Table 10 below depicts these values.

Table 10: Candidate magnets selected and their corresponding sensor distance constraints. D82DIA was determined to have the best balance between size and the required field strength.

Magnet name	Company	Diameter (mm)	Thickness (mm)	Grade	Surface field BrMax (mT)	Min distance to sensor face (mm)	Max distance to sensor face (mm)
D82DIA	K&J magnetics	12.7	3.175	N42	1320	4.17	7.86
D42DIA	K&J magnetics	6.35	3.175	N42	1320	2.94	4.95
D1008D	Supermagnetman	3	1	N45	1370	1.17	2.07

The proposed magnetic inversion/eversion sensor assembly design is shown in Figure 10. The diametric magnet is bonded to the foot shroud at a distance of 5.2mm away from the sensor face.

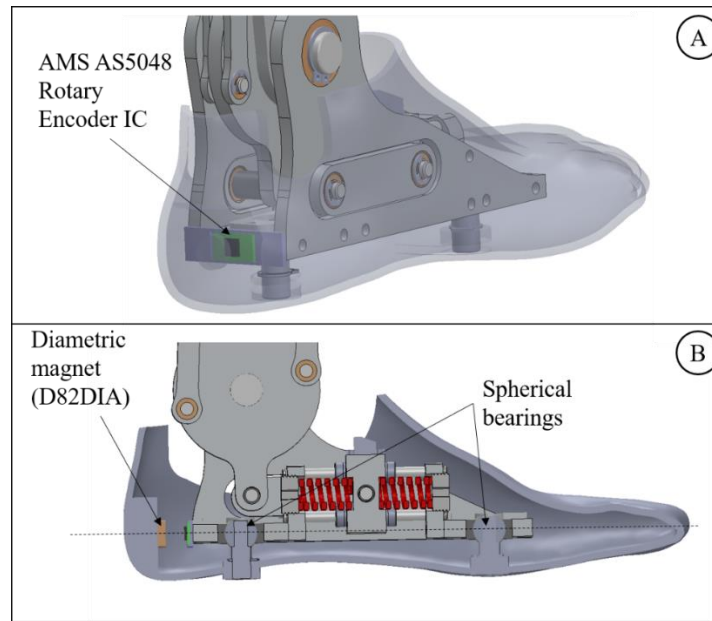


Figure 10: (A) Inversion/Eversion sensor IC assembly. (B) Diametric magnet at a calibrated distance and aligned to rotation axis. Wiring, fasteners and other sensors not depicted.

SPI communication protocol will be employed between the sensor and the MCU to determine the absolute inversion/eversion angle. The angle reading will be input into the main clonus algorithm as a case statement to determine if the inversion/eversion angle used by the learner is inside or outside a certain prescribed range (e.g., 6°), upon which clonus response would not be initiated.

2.2.3. TOUCH SENSORS

To estimate the learner's hand position on the forefoot of the simulator, it was decided that it was necessary to detect the applied pressure on the plantar metatarsal head region of the foot shroud and the lateral and medial surfaces of the forefoot. The requirements for sensor accuracy were not as restrictive as the other two sensor systems because the intent, based on Research Question 2c, was to measure hand positions and not forces. (If necessary in the future, a more accurate measurement of applied force could be derived by digital means, where the force differences between motor output and SEA spring force could be used to estimate the force on the plantar surface of the foot during actuation.) Therefore, it was decided to use a low-profile touch sensor that would fit into the space of a human foot rather than bulky load cells. Force sensitive resistors (FSRs) are a good choice for low profile and easy to assemble sensors to detect contact loading. FSRs operate by increasing their internal electrical resistance as the load applied on the active surface area increases. Therefore, a simple voltage divider circuit can be employed to relate relative force applied to change in electrical resistance. To obtain antagonistic signal readings for FSR pairs, a Wheatstone bridge circuit design can be employed. Antagonistic signals are beneficial in medial and lateral surfaces since the difference between touch magnitude within a FSR pair can be computed without the need for corrections of drift and offset errors. Additionally, in a Wheatstone bridge, the common mode noise arising from each sensor can be cancelled out and some of the force measurement inaccuracies can be resolved.

Figure 11 depicts the arrangement of the sensors. The first group of four smaller FSRs record the hand posture on the lateral and medial surfaces of the foot and are labeled as L1, L2, M1, M2, respectively. These FSR pairs (L1,M1 and L2,M2) are arranged in a Wheatstone bridge (Figure 24, Appendix B) format to provide antagonistic signal readings (one pulls voltage high while activated and other pulls voltage low while activated), therefore the distinction from each

sensor pair can be made more clearly. The second group recorded the average pressure applied on the plantar surface of the foot and are labeled P1-P4. These plantar surface FSRs were connected as a simple voltage divider circuit (Figure 24, Appendix B) so as that a very rough estimate of the center of pressure on the plantar surface can be determined.

Circular FSRs (Interlink Electronics, Camarillo, CA, USA) were chosen with diameters of 12.7 mm (for P1-P4) and 5.08 mm (for L1, L2, M1, M2). The fastening of the FSRs to the foot shroud was done using adhesive tape (VHB 4646, 3M Corporation, MN, USA).

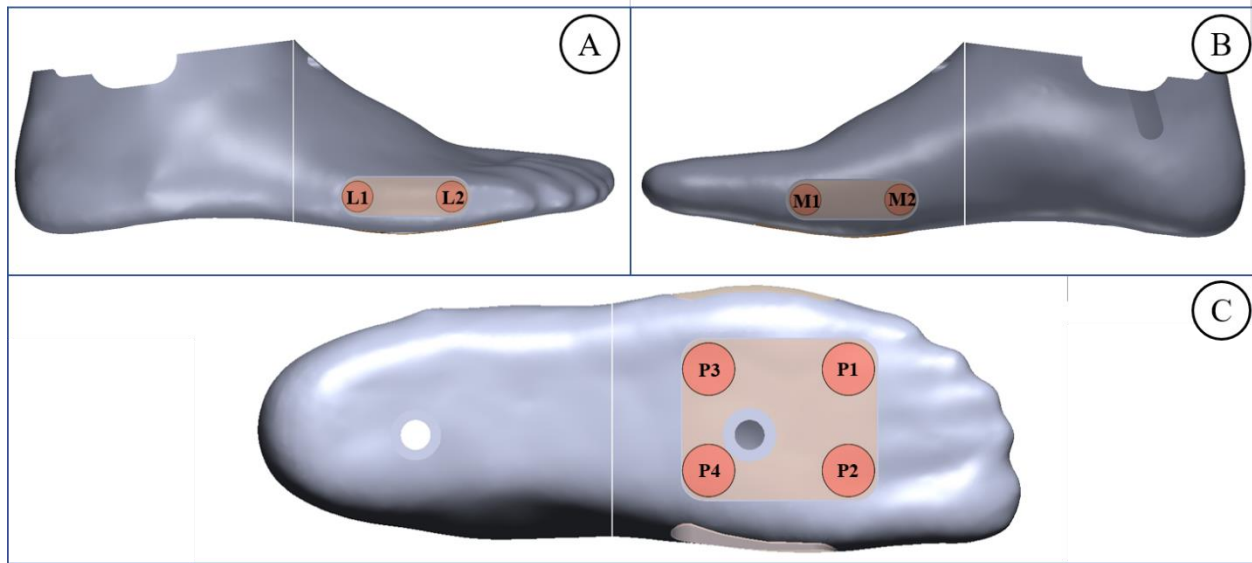


Figure 11: FSRs sensors placement in the foot shroud. (A) L1 and L2 are the FSRs on the lateral surface of the foot shroud. (B) M1 and M2 are the FSRs on the medial surface of the foot. (C) D1-D4 are the four FSRs that detect hand pressure on the plantar surface.

2.3. OFF-DEVICE SENSOR VALIDATION

Off-device testing incorporated sensor validation testing outside of the assembly of the simulator so that the efficacy of each sensor system could be evaluated before installing them on the ankle simulator. To determine the limitations of each sensor system, errors in force control and frontal plane foot position were quantified, such that any errors could be accounted for digitally when the sensor systems are assembled into the full simulator.

2.3.1. SPRING DEFLECTION SENSOR

It is imperative to quantify the error in readings due to magnetic field interference by nearby electronic and metallic components. A magnetic sensor is highly sensitive to changes in magnetic field and therefore magnetic fields created by nearby electronic wires running variable current (e.g., the analog sensing wires of the FSRs) may contribute to some inaccuracies. Therefore, to evaluate the accuracy of the linear magnetic sensor, an Instron vertical test machine (Instron 3367; High Wycombe, United Kingdom) was utilized. The crosshead displacement accuracy of ± 0.01 mm (± 10 μ m) of the Instron machine was regarded as a standard against which readings from the linear magnetic sensor could be compared. Signals were obtained using a smaller MCU (Arduino Uno, Arduino LLC, Italy) and a 4-point moving average filter was applied on the sensor data. The test setup is shown in Figure 12.

Testing procedure started by fixturing the 3D printed magnet fixture to the LVDT sensor mounting locations on the Instron compression platen. Then, the sensor fixture was placed on the Instron. The fixture is adjusted so that sensor face and markings on the fixtures (depicting center of the magnet and center of the sensor) were aligned visually. A plastic shim of 0.25mm was used to determine the gap between the magnet tape face and sensor IC. A cyclic loading test was programmed into software on the Instron machine computer with displacement amplitudes of 3.3 mm. Both the Instron machine and the sensor readings were started simultaneously. After one full cycle of displacement was completed, data were obtained from both sensors (Instron displacement and magnetic sensor displacement) and saved as CSV files for post-processing. Detailed electrical schematic for this sensor is provided in Figure 22, Appendix B

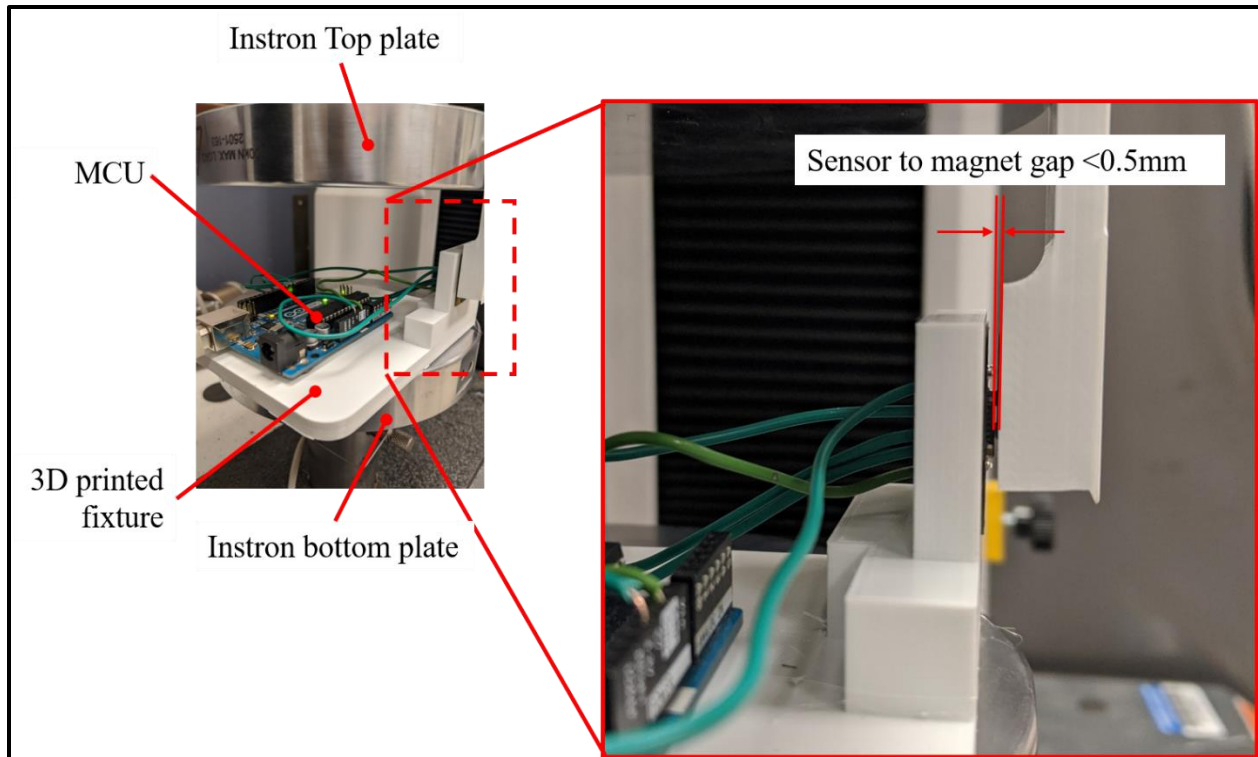


Figure 12: Images from the test setup of the linear spring deflection sensor off device evaluation.

2.3.2. INVERSION/EVERSION SENSOR

To understand the limitations of the magnetic rotary sensor, a customized test setup had to be developed. The testing fixture was built upon the design by Seung Yun (Leo) Song and is depicted in Figure 13. The test setup included a stepper motor controller by a high-resolution optical encoder (HEDS-9040 B00, Broadcom, San Jose, CA, USA) that was actuated and processed using a separate microcontroller (Teensy 3.6 USB, Development Board, PJRC; Portland, OR, USA). It was also critical to understand if the magnetic rotary sensor was sensitive to the placement of the diametric magnet being off axis. A well calibrated but weaker magnet was used that had a surface magnetic field intensity of 120 mT (Cyl0401D, NVE Corp, MN, USA) so as to understand the minimum field strength that would cause the sensor to start deviating from the desired measurement. Detailed electrical schematic of the sensor is provided in Figure 23, Appendix B

Excursion angles of 0° - 180° were programmed on the MCU, upon which encoder readings from both the motor and the magnetic sensor were collected simultaneously, sampled at

25Hz. Five trials at good alignment were conducted. Next, the magnet was shifted in the vertical direction by around 1.2 mm and then misaligned magnet readings were recorded. Lastly, a test at higher excursions speeds (1 trial each) were conducted to determine if there was any change in sensor accuracy. Tested speeds were 50°/s , while higher speed tests were 100°/s and 200°/s.

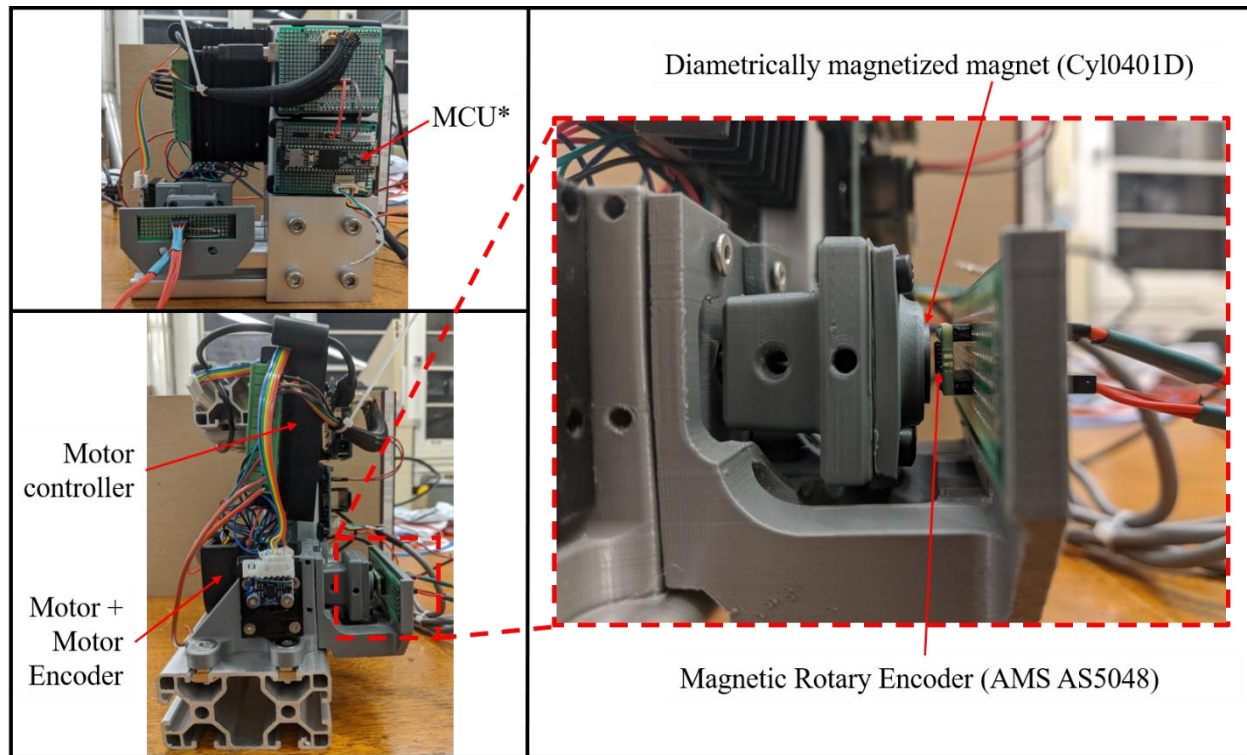


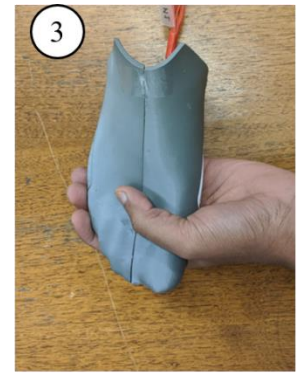
Figure 13: Test station for inversion/eversion sensor. Front view (top left), Side view (bottom left) and zoomed view of the calibrated test magnet and sensor IC (right).

2.3.3. TOUCH SENSORS

To verify the performance of the touch sensors, a standalone forefoot shroud with the FSRs attached was fabricated (Figure 11). The detailed electrical schematics are shown in the Figure 24 Appendix B. Three different hand positions were chosen as acceptable clinical positions and two were chosen as unacceptable positions (Figure 14). These positions were selected based on the placement of the appropriate hand on the foot and position of the learner during the administration of a clonus test. For example, an unacceptable hand position would be where a right hand is used for administration of clonus and the left hand is used to support the calf with the learner located on the medial side of the foot (position 5), which could cause the right hand to be impeded by the support hand. Qualitative evaluation of the sensor performance was conducted by several members

of the team where voltage change data were collected for each of the five hand positions. The readings from each of the FSRs were recorded using a smaller MCU (Arduino Uno, Arduino LLC., Italy). These recordings will be used as the first step toward determining the algorithms that estimate hand position of the user during the training process.

Acceptable hand positions



Unacceptable hand positions

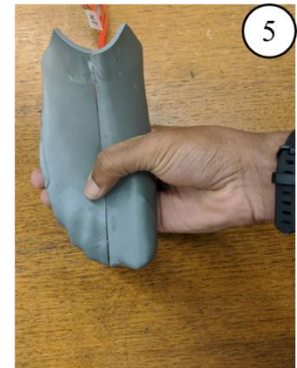
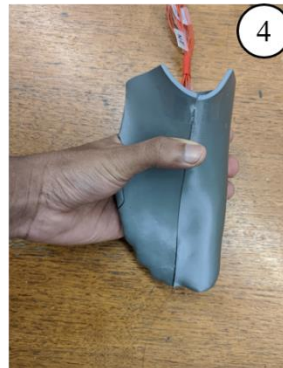


Figure 14 : Test hand positions (right hand dominant). Positions 4 and 5 would cause the clinician to have the test hand be impeded by the other hand supporting the calf/knee during test.

3 RESULTS

3.1. BIOMECHANICS AND FOOT ANTHROPOMETRY

3.1.1. ANKLE ANGLE TORQUE RELATIONSHIP

The resultant expression for the values of $I(\theta)$, $B(\theta)$ and $K(\theta)$ are tabulated in Table 11. All expressions were converted to degrees and were assumed valid within the ROMs considered in Table 4. The fit of the viscoelastic model with the data was moderate. Figure 25 in Appendix C illustrates the goodness of fit plots for the passive elastic and viscous components at the ankle joint with respect to ankle angle. In all equations and plots depicted henceforth, positive angles are in DF and negative angles are in PF.

Table 11: Derived expressions by fitting data from [79] (0° knee flexion angle) for passive stiffness parameters as a function of ankle angle.

Parameter	Units	Fitting function	Expression	(r^2)
$I(\theta)$	kgm^2	constant	0.0073	-
$B(\theta)$	$\frac{Nms}{^\circ}$	exponential	$2.883 \times 10^{-6} e^{-707.4(\theta)} + 0.2335 e^{41.87(\theta)}$	0.7328
$K(\theta)$	$\frac{Nm}{^\circ}$	exponential	$0.2186 e^{-345.0(\theta)} + 20.13 e^{71.55(\theta)}$	0.7835

From the expression derived above, it is evident that the viscous and inertial components play limited roles in passive ankle torque especially at smaller angles, as indicated by low coefficient values in Table 11. To validate the use of the viscoelastic model (Equation 2), we compared the torque calculated based on subject information in [79] with averaged torque ankle angle relationship established in another study with a double exponential function developed by Reiner et al. [77] (Equation 3). A comparison between the two models was plotted in Figure 15.

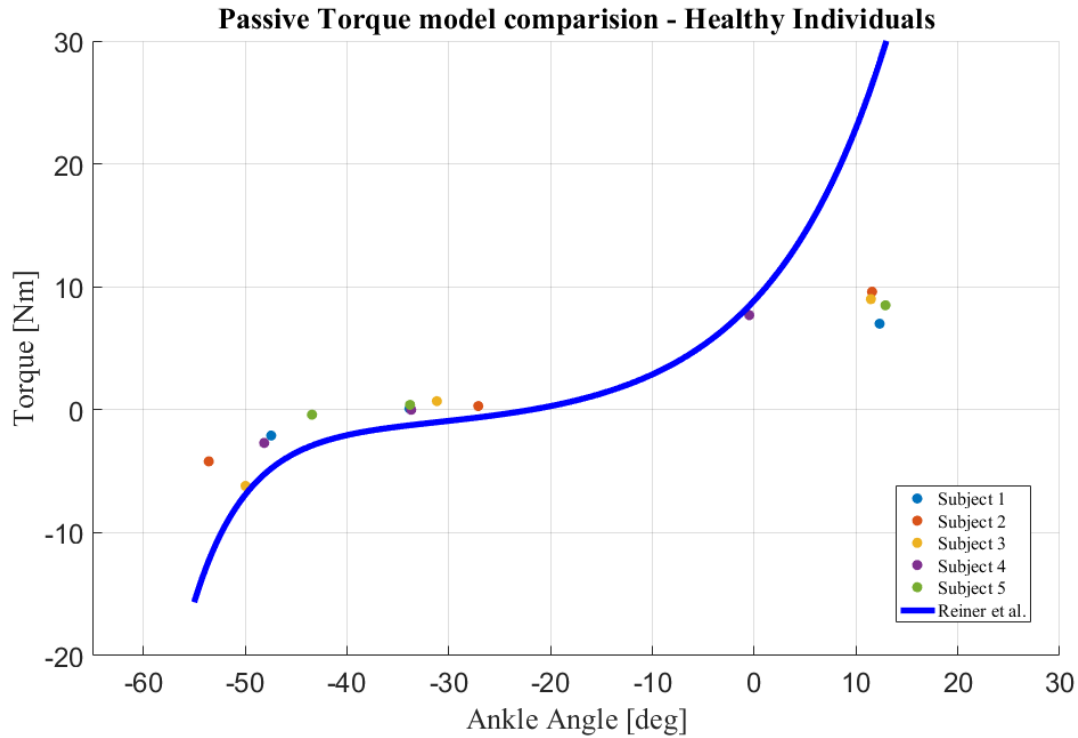


Figure 15: Models for passive ankle torque in healthy test subjects (with 0° knee flexion). The blue line is from the model in Equation 3 using the double exponential function for healthy subject from Reiner et al. [77]. Dots depict results from the viscoelastic model from Equation 2 based on discrete ankle angle values for specific subjects.

There was moderate agreement between the models in the PF region while larger variances were observed in the DF region. The differences between the two models can be accounted for by physiological variances between patients in the different studies. Also, normalization of the initial data gathered based on anatomical variables could not be made due to the lack of available data in the source. It was decided that for the clonus simulator, torque based on the viscoelastic model in Equation 2 would not be employed, instead, the newer double exponential function developed (Equation 3) would be written into the ankle clonus simulator algorithms.

Ankle stiffness in the clonus case was modeled using a double exponential expression that is a function of ankle angle and knee flexion angle. The expression was adapted from the study by Edrich et al. [69], in which, experimental ankle torque data collected from SCI patients who exhibited heightened passive stiffness (sign of spasticity). A modified expression for healthy controls (Equation 5) was also used to compare the average difference in stiffness between healthy individuals and SCI patients. Since SCI patients are most likely to exhibit clonus, Equation 4 was

deemed as a valid mathematical description of passive ankle stiffness in clonus patients. Figure 16 depicts the result of the ankle stiffness comparisons over the entire ankle ROM. It was found that changes in knee flexion angle impact the passive ankle torque properties substantially, with higher knee flexion angles decreasing the DF torque of the ankle. It can be hypothesized that this relationship is due to a larger dependency of passive muscle properties on lengthening of the gastrocnemius and soleus muscles, which have different initial lengths at different knee angles.

Passive ankle torque behaviour at different knee flexion angles - Healthy vs Clonus case

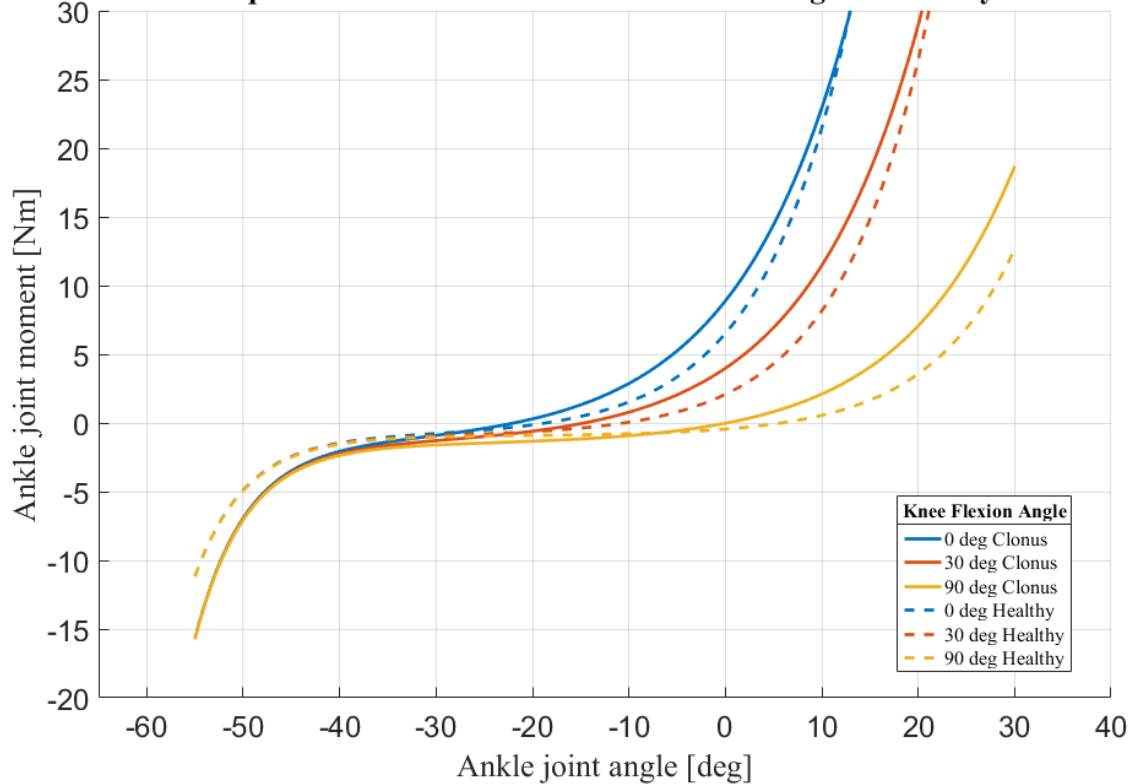


Figure 16: Ankle joint angle – passive joint moment relationship based on knee flexion angle for SCI (clonus) vs. healthy case. Models are based on expressions by Edrich et al. [69]

Based on our testing, it was decided that using double exponential models (Equation 4) for the clonus case as well as for the healthy individual case (Equation 3) would be ideal and that the viscoelastic model (Equation 2) would not be as accurate as initially hypothesized. To increase the accuracy of our clonus model, further research into active muscle properties will be required and algorithms will need to be updated accordingly.

3.1.2. ANTHROPOMETRICALLY ACCURATE SHROUD DESIGN

The PF and DF ranges of motion achievable were 37° and 28° , respectively (Figure 17 A), which are greater than the target ROM values in the sagittal plane as indicated in section 1.5.2. However, the maximum eversion and inversion ranges that the simulator was capable of achieving were limited to 12° and 10.2° , respectively (Figure 17 B, C). The reasons for us not meeting our frontal plane targets described in section 1.5.1 are due to engineering limitations based on the SEA mechanism chosen. Possible paths to increase eversion/inversion ROM were considered but due to potential ramifications to the structural integrity of the simulator internal structures, it was decided that frontal plane ROM achieved in this design would be acceptable.

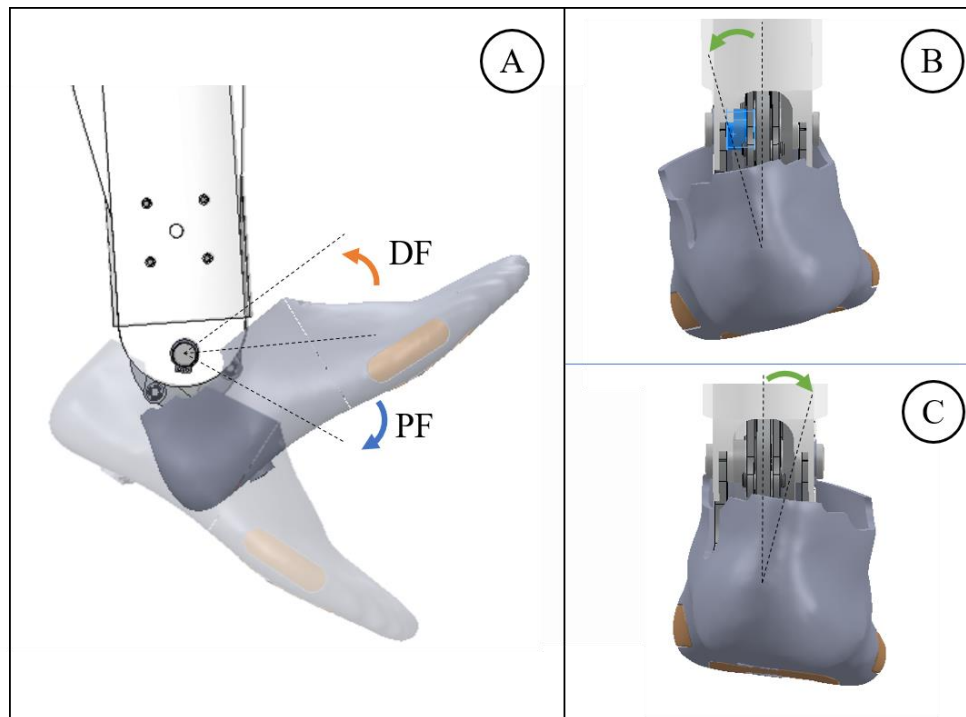


Figure 17: Final ROM capacities the ankle clonus simulator. (A) DF/PF ROM. (B) Eversion ROM. (C) Inversion ROM

3.2. OFF-DEVICE TEST RESULTS

3.2.1. LINEAR SPRING DISPLACEMENT SENSOR

Results from the off-device tests of the linear magnetic sensor system were informative. It was discovered that a slight misalignment of the magnet travel axis from the center of the IC (tested at 1.0 mm) could cause a reading error of up to 0.54 mm. Additionally, there were significant errors when the sensor was too far apart from the magnetic tape, where not enough magnetic field strength was imparted on the Hall effect elements in the sensor IC, causing a deviation in displacement readings of up to 0.75 mm. Figure 18 depicts the output signals of the sensor when the subassembly was properly aligned and misaligned (i.e., the magnetic tape at a distance greater than 0.7mm away from the IC) as compared to the calibrated Instron machine encoder.

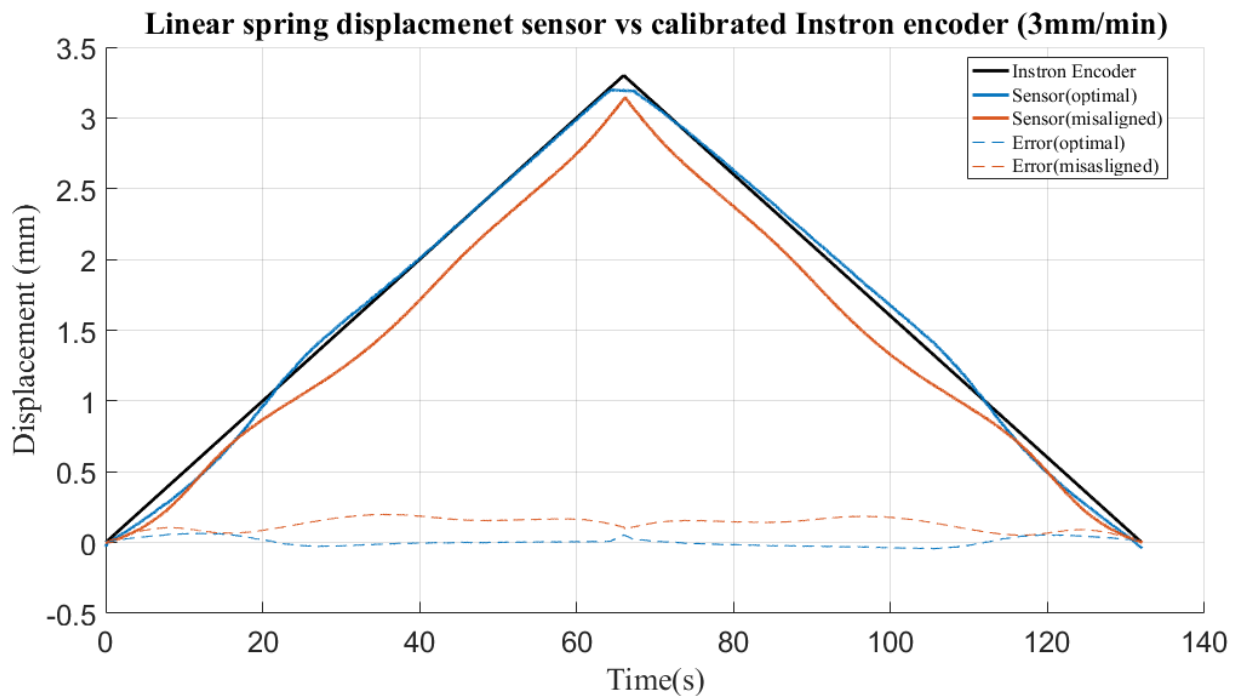


Figure 18: Sample trial results of linear displacement sensor output plotted against calibrated Instron machine encoder measurements. Optimal sensor result curve is clipped at 3.2 mm since the magnetic tape field exceeded the limits of measurement range for the IC in this trial. Error was computed in the range of 0 to 1.8 mm, reflective of spring displacements in the ankle simulator.

When the magnetic tape was properly aligned and at distance less than 0.4 mm from the sensor IC, results were borderline acceptable. A worst-case reading error of 0.082 mm was

recorded in this case, being 4.2% of the displacement range of the spring and corresponding to an error of ± 14 N (8.2% of maximum spring force). From the dynamic analysis of the crank slider system, the force error would translate to an error of 0.63 Nm (4.2 % of max) of torque about the ankle joint. Initially, we targeted to achieve less than 2% torque error in our design. However, due to limitations of the sensor fidelity and high springs stiffnesses used, the actual torque error margin for the simulator was determined to be less than 5%.

3.2.2. INVERSION/EVERSION SENSOR

Off-device validation results of the inversion/eversion sensor were favorable. Testing was performed with an excursion range of 0° - 180° to understand how the readings varied with a full rotation of the magnet. The maximum and average error were computed from 0° - 26° (representing maximum inversion ROM). The inversion/eversion sensor had an average error of about 0.095° (0.8% and 0.094% of eversion/inversion ROM, respectively). Good agreement between the high accuracy optical encoder and the inversion/eversion sensor was obtained when the diametric axis of the magnet was properly aligned to the sensor and roughly 50mT of field (simulated) approached the sensor face. It was observed that up to 1.27° (10.5% and 12.6% of eversion/inversion ROM, respectively) of reading error could occur if the magnet was mounted slightly off axis or if the magnetic field strength on the sensor face was toward the lower operational limit. This case was tested with the magnet axis offset at around 1.2 mm (10% of magnet diameter) away from the axis center of the sensor, which is reflective of worst-case assembly tolerances. To indicate performance with assembly errors, the readings for the worst-case scenario are plotted in Figure 19 with the error multiplied by 10 for better visibility.

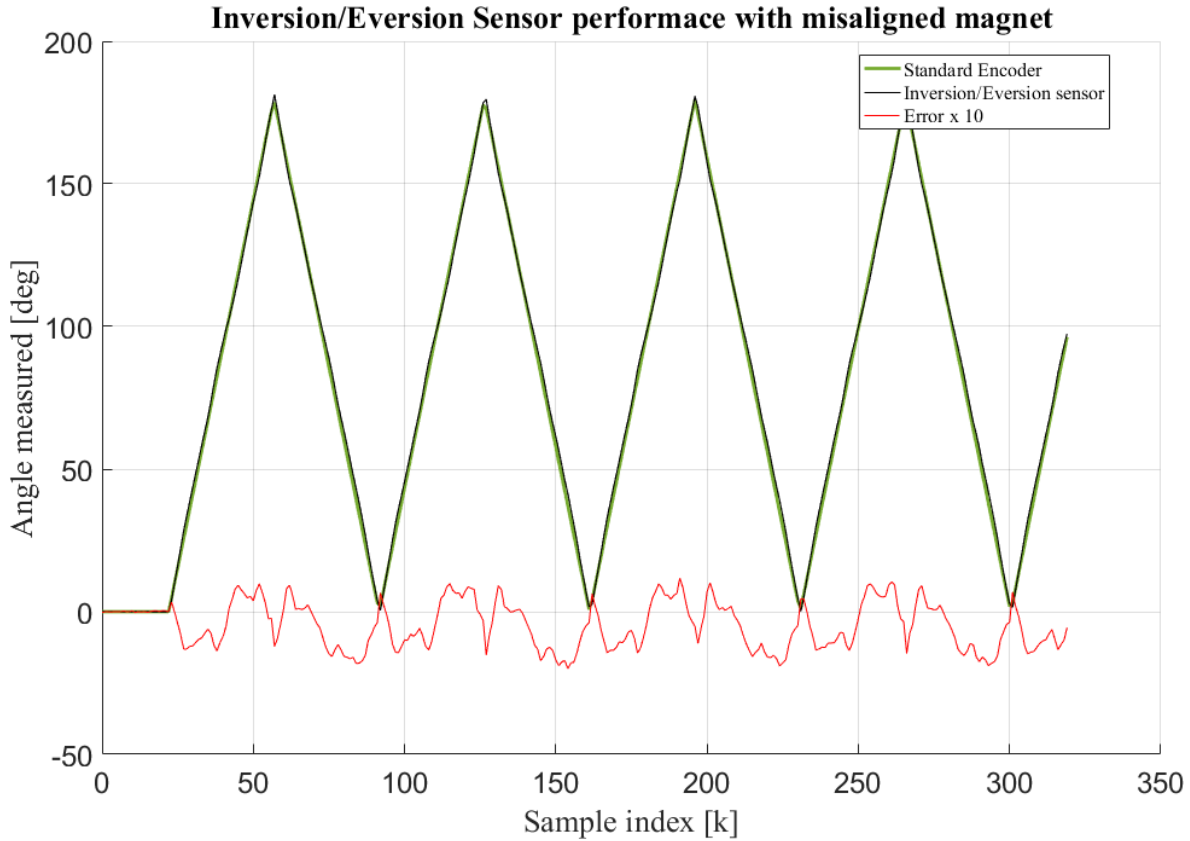


Figure 19 : Sample data for a slightly misaligned magnet. Sensor output in inversion/eversion ROM are marginally acceptable. (Error was multiplied by 10 for better visibility)

Testing the inversion/eversion sensor at multiple actuation speeds also yielded no difference since the sampling rate from the sensor was very high (25 Hz). In all, the inversion/eversion sensor errors and risks were deemed acceptable for our application.

3.2.3. HAND POSTURE ESTIMATION

Results from the hand placement on the prototype shroud tests were critical in understanding the performance of the touch sensors. By iteratively testing the required touch force to activate each sensor, it was decided a threshold voltage change of ± 2.45 V was appropriate to delineate between a “touch” or “no touch” threshold. Therefore, any voltage changes due to the Wheatstone bridge going below zero are termed as ‘low’ (L), and changes exceeding the threshold were termed as ‘high’ (H). From the data gathered, a case statement algorithm will be developed in the future. For each of the hand postures tested, results from one of the evaluators (right-handed)

are shown in Table 12. Additionally, it was observed that some sensors might exhibit drift in values over time. This drift may cause false positive activations in the future especially when using tape adhesive to hold the parts of sensor actuation surfaces. To address this issue, digital calibration steps when the IC is first booted up can be implemented, where average initial readings can be subtracted from following readings to eliminate any offsets. Additionally, drift in the sensor signal may not be detrimental (>10% error) until at least 11-12 months have passed, after which the sensors may need to be replaced [96].

Table 12: Sample touch sensor outputs for different hand positions. Figures corresponding to each hand position is shown Figure 14.

Hand Position	Clinically acceptable?	L1+M1	L2+M2	P1	P2	P3	P4
1	Y	L	0	H	H	H	0
2	Y	0	0	H	H	H	H
3	Y	H	0	H	H	0	H
4	N	0	L	H	H	0	0
5	N	0	H	H	H	0	H

‘0’ indicates minimal or no change in voltage (i.e., below threshold or “no touch”). ‘L’ indicates voltage decrease and ‘H’ indicates voltage increase. L1/M1 form an antagonistic pair in of sensors in a Wheatstone bridge configuration and L2/M2 form the other antagonistic pair. (Appendix B)

Based on the data obtained, an algorithm to differentiate the different hand postures will be trivial to implement. P1-P4 voltage readings can be recorded into a single byte as binary variables (1,0), while L1/L2 and M1/M2 voltages can be recorded into a signed byte (high, low, 0). These bytes can then be compared to stored bytes and user hand position cases can be established. While the FSRs do not provide a high fidelity reading of the forces, the estimated hand positions can be successfully recovered along with presence of continuous force on the plantar surface of the foot (P1-P4 readings, at least two of them should be ‘H’).

4 DISCUSSION

4.1. BIOMECHANICS AND ANTHROPOMETRY

To select an appropriate expression that mathematically models the passive biomechanics at the ankle two different approaches were studied: the traditional viscoelastic 2nd order function and a newer double exponential function fitted to experimental values. It was found that the double exponential models were better suited for the ankle clonus simulator and since viscous components of the viscoelastic model were small enough to neglect the speed dependency on ankle stiffness. The resultant ankle torque expression derived for the ankle clonus case and healthy patient cases were applicable to be programmed into the MCU for the simulator (Equations 3 and 4). The learnings from literature study detailed in section 1.3.3 and results obtained will be critical in developing the control algorithms for the ankle simulator. For instance, the reflex threshold velocity of 77°/s can be actively derived by the motor encoder to turn on a clonus behavior sequence. Additionally, it was discovered that a significant change in passive torque mechanics occurs when the knee flexion angle is changed (Figure 16). For our simulator, we will employ a fixed knee flexion angle of 45° and advise learners accordingly. However, it might be beneficial for us to include a setting in the code where the learner can change the knee flexion angle, which would update the force control algorithms and hence produce more accurate clonus behavior.

Anthropometric accuracy of the foot shroud was achieved by careful modelling of the shroud structure and heeding anatomical learnings of ankle ROM among human males. The design allows for 28° of DF and 37° of PF. However, there was a significant challenge in meeting the target ROM requirements of inversion/eversion of the ankle due to the design of the spring cage and foot structure of the simulator. Upon consulting with the neurologist in collaboration on the project (Dr. Chris Zallek), these limitations in inversion and eversion angles (10.2° and 12°, respectively) were deemed acceptable for the first-generation simulator since the key learning requirements for the clinician trainees would be to try and maintain a near zero inversion/eversion angle during the ankle clonus test. Overall, the basic biomechanical and anatomical considerations explored in this work were able to provide preliminary information for the proper functioning of the ankle simulator for basic clonus activity. However, more research into active muscle properties would be required obtain higher fidelity of the force-angle relationship similar to that of an actual

patient. The addition of a sinusoidal reflex response that follows the active muscles properties during clonus can then be implemented to complete the functionality of the ankle clonus simulator.

4.2. SENSORS

The linear spring deflection sensor selected would be appropriately accurate for the purpose of measuring the SEA mechanism force. Coupled with the sensor information of spring displacement and motor torque, the estimated torque error at the angle is acceptable (within 5 %). From the off-device testing of the sensor, it was found that the sensor is highly sensitive to assembly tolerances; care must be taken to ensure that the magnetic tape is properly aligned with the sensor IC. A gap of more than 0.5 mm between the magnetic tape and the sensor IC surface would deem the linear displacement inaccurate, and by consequence the torque control at the ankle joint. Noise in the spring deflection readings may also occur due to fluctuations in ambient temperature. The polymer based magnetic tape used has a heat expansion coefficient of 17-20 ppm/°K, which would equate to roughly 0.02 mm change in reading per °K change in ambient temperature. It must be noted however, that sub-millimeter accuracy in reading is rarely achieved by miniature linear potentiometers and optical sensors. The only other alternatives to this design would be to design a custom strain gage sensor. To conclude, the force and speed measurements derived from the linear deflection sensor appear to be marginally acceptable for force control in the ankle simulator (5% error or ~0.75 Nm), and within the bandwidth limits of the mechanical design. To help reduce error in force estimation, a sensor fusion approach could be undertaken with the use of a high resolution rotary encoder (AMT223B, CUI Devices, Tualatin, OR, USA), attached to the second stage of the gear transmission system to get a better estimate of the spring deflection. A sensor fusion design could employ a Kalman filter approach to reduce spring deflection measurement error. Specifically, the spring deflection measurement signal and the ankle angle measurement signal can both complement each other in the filter and thus provide a better true estimate of the ankle position and spring positions.

The inversion/eversion sensor selected sufficiently met our design intent. Care has to be taken during the final assembly of the sensor system in the simulator to make sure that the magnet and the sensor IC are concentric about the inversion/eversion axis. While some misalignment of the magnet (<10% of magnet diameter) can be tolerated, a reading error greater than 10% may provide false negatives for learners that do not perform the clonus test at an appropriate

inversion/eversion angle. For the purpose of the development of an ankle clonus algorithm, the exclusion criterion for the ankle simulator to begin clonus behavior should be set at $\leq 6^\circ$ eversion and $\leq 5^\circ$ inversion as a start. Upon further consultation from collaborating clinicians and neurologists, these values can be modified accordingly.

The touch sensors were deemed viable for the intent of estimating the learner's hand position during clinical assessment. The exclusion criterion for the simulator to actuate a clonus behavior during a test could be if the FSR readings match one of the incorrect hand position cases. A graphical user interface feedback of incorrect hand position would be beneficial for the clinician learner. Like the inversion/eversion angle exclusion criterion, hand position cases can be modified, and thresholds can be adjusted upon consultations with neurologists/clinical training instructors.

4.3. LIMITATIONS

Biomechanically, the ankle simulator is intended to successfully match the passive ankle torque for both healthy individuals and clonus patients. While the clonus passive properties discussed in this work can be reflective of average passive properties of the spastic ankle, the delineation between different levels of spasticity (MAS 1-4) has not been explored. Additionally, there are certain spasticity cases, such as MAS levels of 3 and 4 [93], where the patient torque may exceed the torque capability of the design. While the simulator is built specifically for clonus, further details regarding active muscle properties and different passive properties for different spasticity levels will need to be considered.

The anatomical accuracy of the foot is a key novelty in our simulator. By undertaking a significant engineering challenge of constraining all of the SEA mechanisms within a 50th percentile male foot, we can deliver a more realistic experience of ankle clonus than previously developed simulators [84], [86]. Subsequently, by learning correct hand positions required and associating some quantifiable metrics (e.g., minimum stretch speed required) in the training, clinician learners will be able to administer ankle clonus tests more effectively. A feature that is not present in this design, however, is the compliant nature of skin. A lack of the haptic feel of human skin may cause learners to perceive the robot device as less similar to that of a real human foot. Additionally, the foot shroud in our design does not support range of motion in the transverse plane. The inclusion of a 3rd axis of rotation in the transverse axis about the ankle joint was a

complexity that would interfere with the correct functionality of the SEA mechanisms, and adding the corresponding sensors to detect the added movement would require more space in an already overcrowded foot shroud. Therefore, true pronation/supination of the ankle cannot be completely simulated.

Sensors to be assembled into the simulator were tested independently and verified for their functionality; however, their combined behavior after assembly is not known. The linear sensor chosen was not providing a reading fidelity as originally envisioned (4.4% actual vs. 2% targeted). While the resulting torque error (± 0.63 Nm) may not be easily discernable by learners, future revisions of the design may include rethinking the spring deflection sensor such that better force control accuracy may be achieved. Regarding the touch sensors, the limitations of FSRs include drift due to creep of laminating polymers over time (5% per $\log_{10}(\text{min})$) and electrical noise ($\pm 10\%$ of initial resistance) [96]. This drift can cause a total variance of up to 15% of the force measured during the lifetime of operation of the FSRs. These limitations might mean that the FSRs might need to be replaced after prolonged use (> 11 months).

4.4. FUTURE WORK

For the modelling of the ankle joint stiffness, two main cases (clonus and healthy) were considered in this work. However, due to spasticity and clonus often both being demonstrated by individuals with UMN lesions, it would be imperative that spasticity behavior be included in the cases of the ankle simulator. Spasticity by its traditional measurement (MAS scale) would have five distinct scales and more research needs to be done to establish a correlation between ankle stiffness and MAS scales. To support higher levels of spasticity, the DC motor selection should be reevaluated to provide greater maximum torque of up to 30 Nm [93]. Coincidentally, control algorithms will need to be revisited to support different scales of spasticity.

Currently the foot shroud is rigid, and a fabric sock will be used to cover the foot and ankle area of the simulator. However, this assembly can be changed to contain a layer of compliant silicon material that is over molded onto the stiff shroud. Foot shroud parts anterior to the metatarsal head regions could be made fully compliant to mimic the anthropomorphic behavior of the toes, providing more realistic representation of a human foot. Our first-generation simulator does not support foot excursion in the transverse plane so any future devices (second generation)

might want to address this limitation by changing the architecture of the primary simulator mechanisms.

As for the sensors, it would be worthwhile to investigate the use of metallic foil strain gages as the primary linear deflection sensor instead of a magnetic linear encoder. Sensor code on smaller MCUs would need to be rewritten into a more advanced signal processing controller (LAUNCHXL-F28379D). This new controller would create a unified central processing location for all sensors systems and algorithms to be housed, minimizing electrical board space and complexity of the clonus simulator assembly. Additionally, an ankle clonus algorithm would need to be devised and implemented in this MCU. This algorithm would combine the learnings from this work to actuate the clonus feedback mechanism when stretch reflex speed has been reached during administration of the clonus test on the simulator. Additionally, the algorithm would need to negate clonus behavior of the simulator if the exclusion criterion based on clinician hand position (touch sensors) and inversion/eversion angle (rotary sensor) have been met. Lastly, clinical validation tests need to be completed once the simulator has been assembled. Collaborators from the medical communities (clinicians, neurologists) will be consulted and appropriate protocols must be established for such tests.

4.5. CONCLUSIONS

This thesis presents the evaluation of biomechanical and electromechanical requirements for the development of an ankle clonus simulator. Novel mechanical actuation using series elastic actuation provides accurate force control to match ankle torque behavior at the ankle joint of a healthy individual and an individual demonstrating clonus. Double exponential functions were best suited to model the ankle-angle and torque relationship for both healthy individual case (Equation 4) and clonus patient case (Equation 5). More research into reflex muscle activations could provide a better match with ankle torques measured among patients demonstrating spasticity. Sensors systems designed for the proficient functionality of the ankle simulator met the design intent. Total torque error at the ankle joint was found to be less than 5% (0.63 Nm for a 15 Nm reference torque); therefore, the current linear magnetic sensor system is appropriate as long as an error of ~5% in estimated ankle torque is acceptable for the torque feedback loop and the estimated applied load by the learner. The inversion/eversion angle measurement errors in the frontal plane were below 1% (where inversion is 10.1° and eversion is 12.0°); thus, this sensor system is acceptable

although care must be taking when assembling into the final simulator. The locations for touch sensors detecting user hand positions on the lateral, medial, and plantar surfaces of the forefoot shroud were defined. Electrical circuits using thin FSR sensors were proposed and evaluated for some simple test cases. Additional work is needed to create algorithms for detecting acceptable/unacceptable hand positions. Embedded code for processing signals and data from these sensor systems need to be ported to the final simulator controller system. Overall, adequate exploration into biomechanics and sensors was performed to be able to answer specific research questions that arose during development of the ankle clonus simulator.

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APPENDIX A: ANKLE BONES

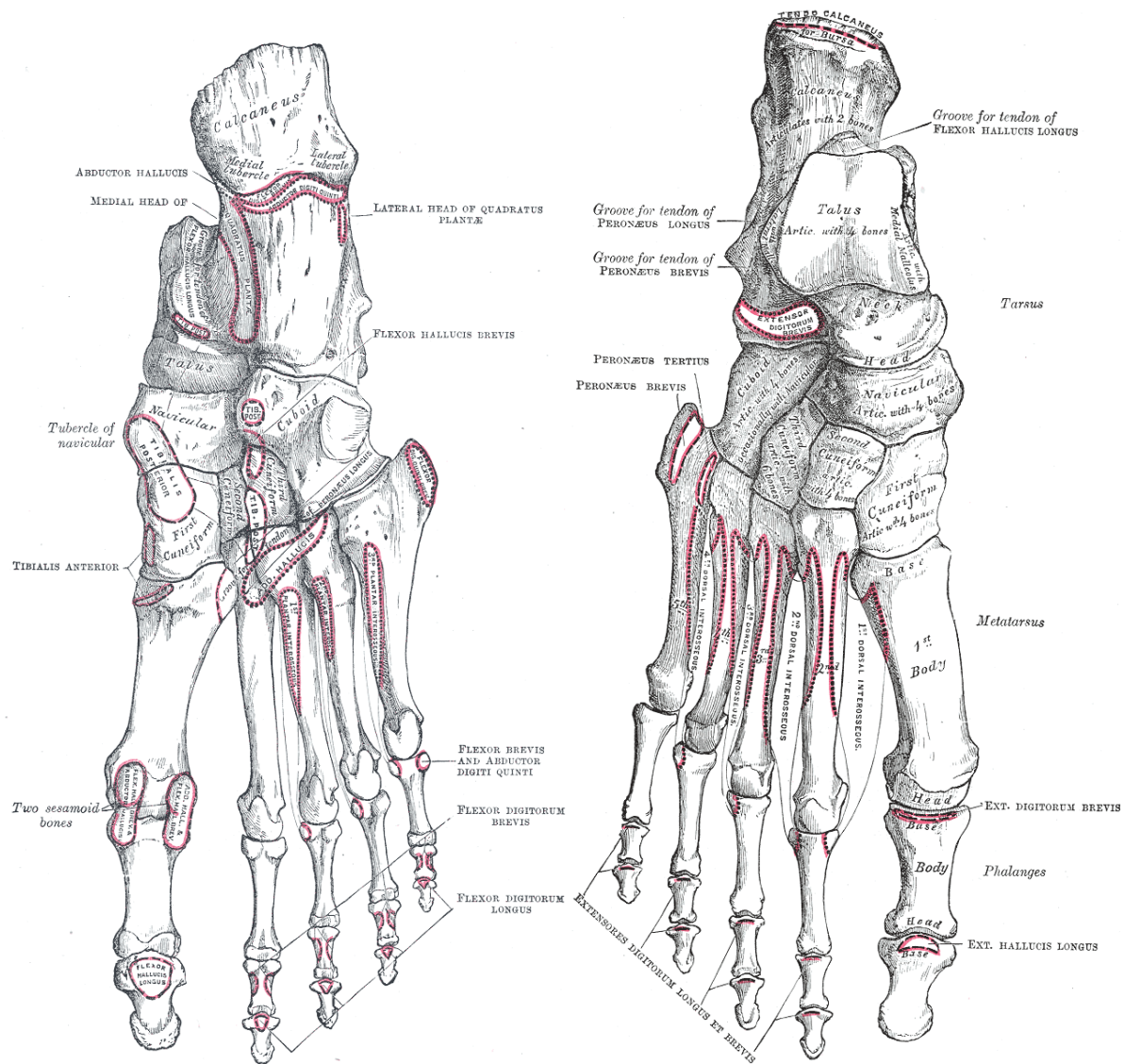


Figure 20 : Ankle and Foot bones. For the clonus simulator, it is important to consider the location of the axes of rotation in DF/PF and inversion/eversion directions. Additionally, the metatarsal head regions were selected as locations for touch sensors to be placed.

Images reproduced from [97] (Open source).

APPENDIX B: ELECTRICAL SCHEMATICS

Complete System:

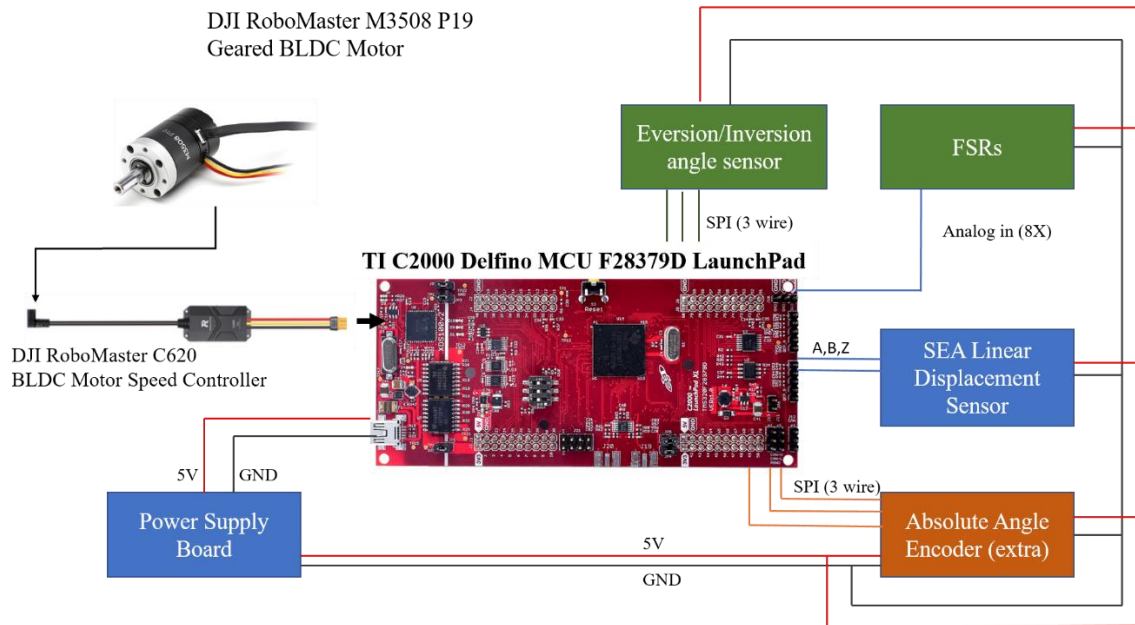


Figure 21 : Overall electrical architecture of the system.

Linear Displacement sensor:

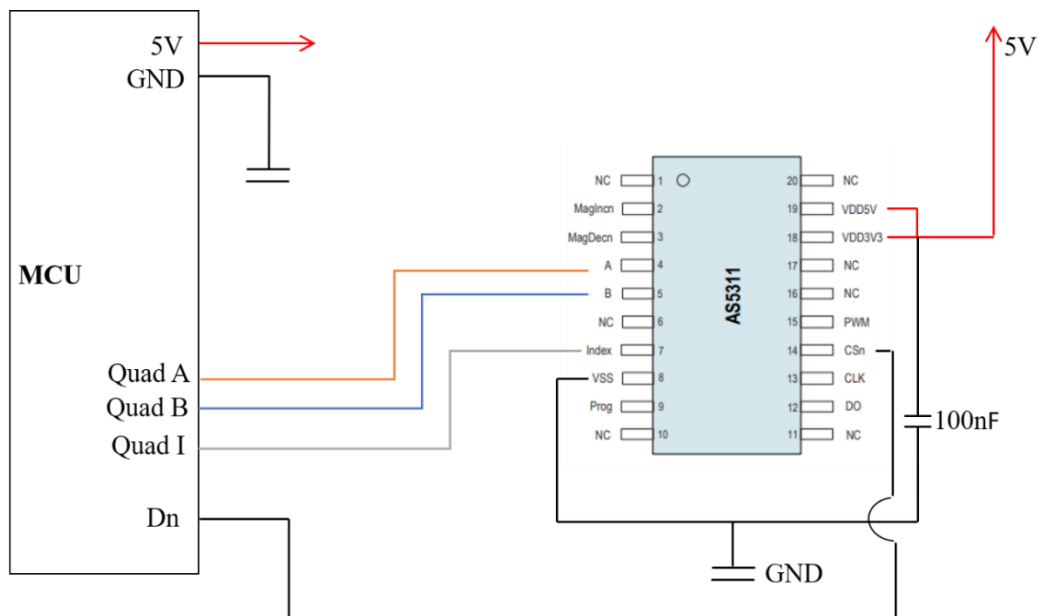


Figure 22: Linear sensor electrical schematic. Same connections were used for off-device tests.

Inversion/Eversion Sensor:

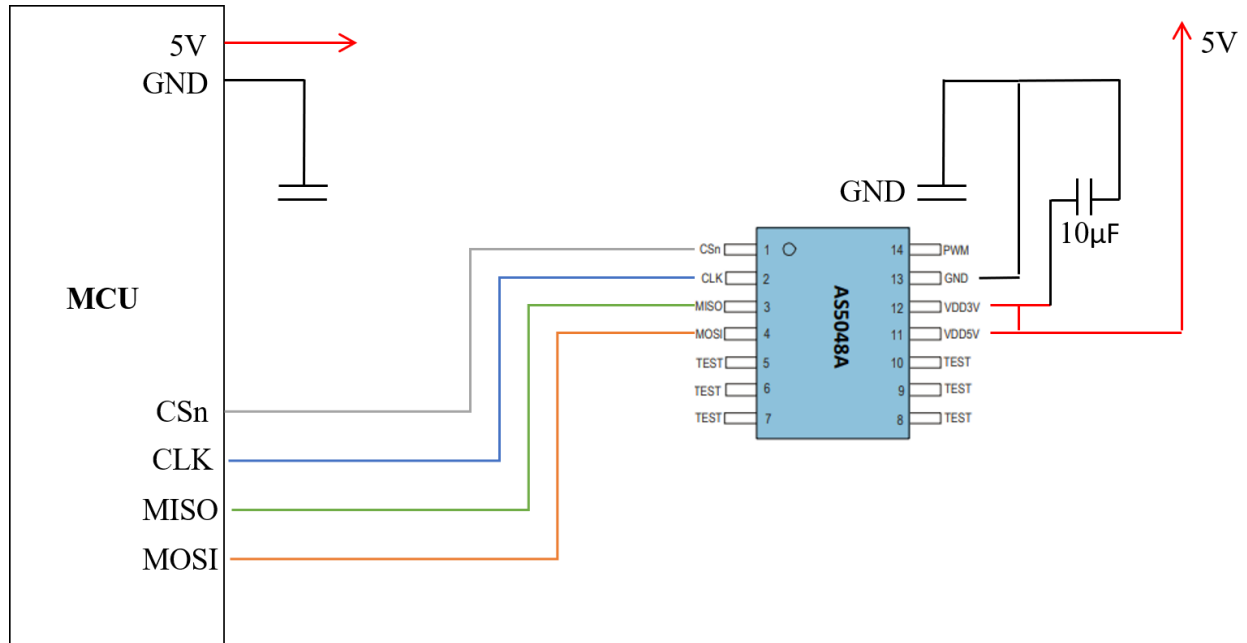


Figure 23: Inversion/Eversion sensor. Same connections were used for off-device tests.

Touch Sensors:

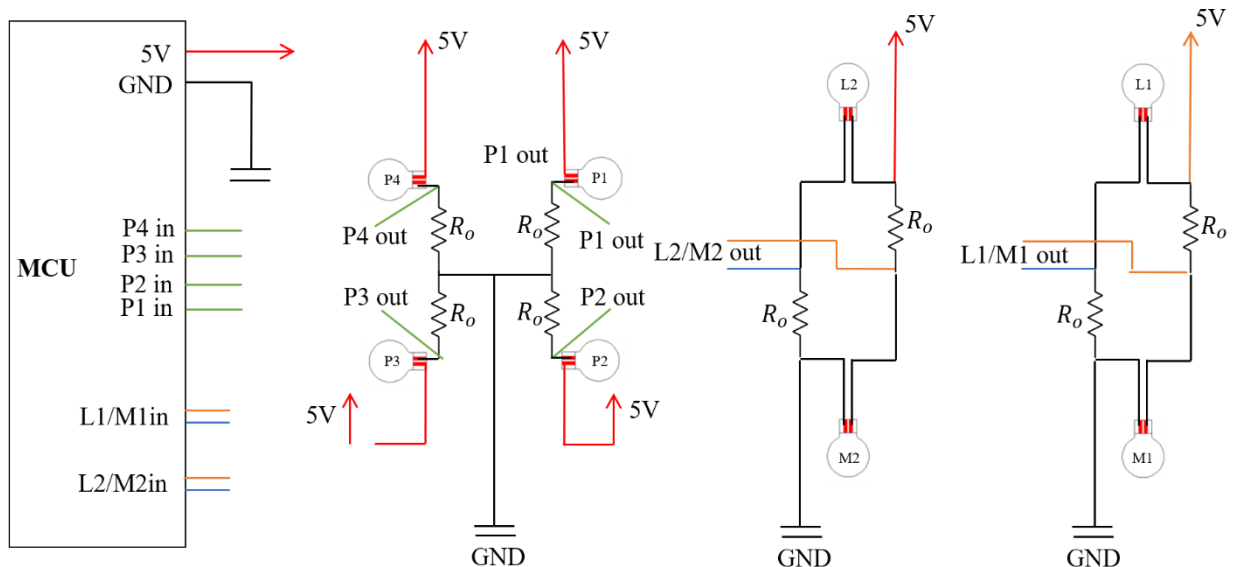


Figure 24: Schematic for touch sensors on foot shroud. Same connections were used for prototype shroud. Green lines indicate standard analog readings while orange and blue indicate differential inputs. R_o was chosen as 10000 Ω resistors.

APPENDIX C: VISCOELASTIC MODEL FITTING

Figures below present the fitting curves to determine the expressions for ankle angle-torque relationship in healthy individuals using MATLAB Curve Fitting Toolbox (R2016b, MathWorks Inc., Natick, MA) Angles are depicted in radians to help compare with original plots in Weiss et al. [79]

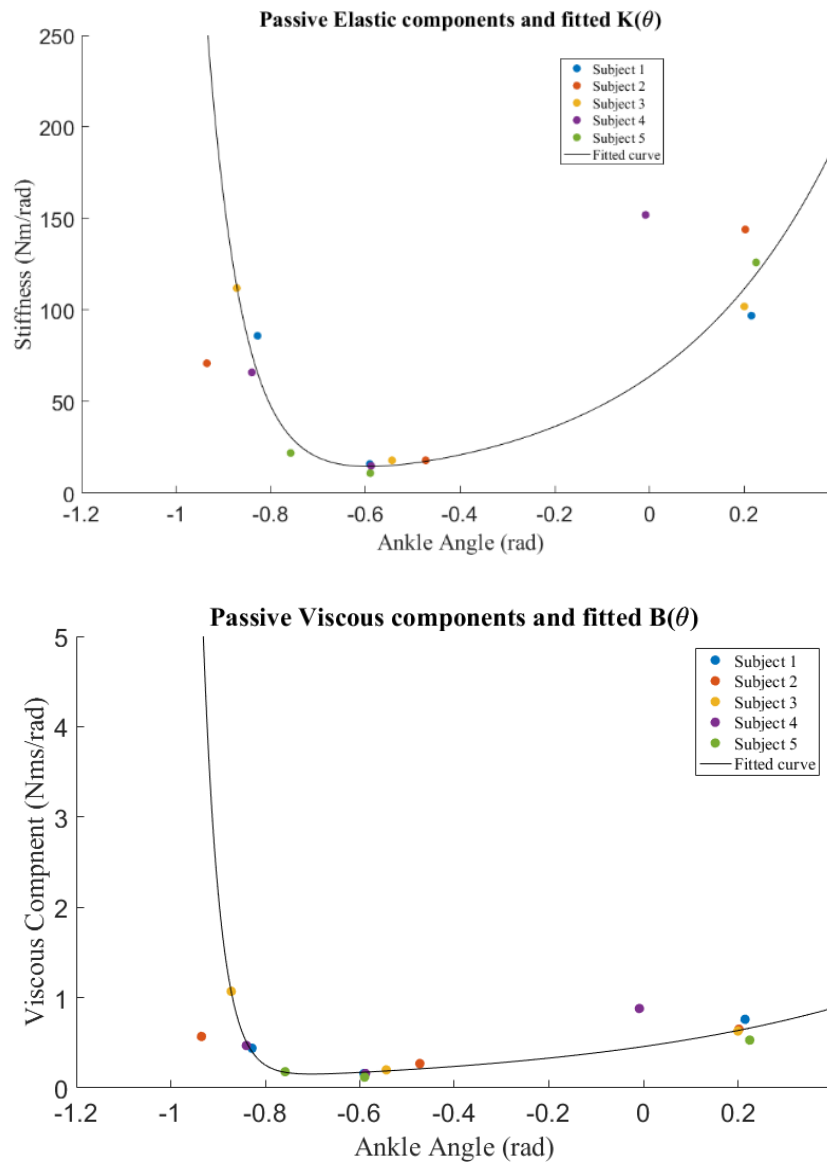


Figure 25: Closeness of fit of viscoelastic stiffness model (Equation 3) to values published in Weiss et al. Elastic component (top) and viscous components (bottom) are plotted as a function of ankle angle.