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Using Custom Integrated Force Sensing Mechanisms for Interaction Control in Rehabilitation Robots

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ABSTRACT

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This thesis presents the implementation of interaction controllers on two custom integrated force sensing mechanisms and demonstrates their suitability for applications to the field of rehabilitation robotics. One condition in which interaction control is beneficial occurs when the robot’s dynamics are significant and need to be compensated through force-feedback. To address this need, a grip force sensor that measures forces in three planes by using force sensing resistors was developed. The device readily integrates with most rehabilitation robots at the end effector. Additionally, if a robot is non-backdrivable, force measurement is required to render transparent environments during evaluation mode as well as for interaction control. Here, interaction controllers are implemented in a 1DOF MR-compatible actuation module. The MR-compatible device uses a non-backdrivable actuator with series elasticity for force sensing. Experimental implementation of interaction controllers on both devices demonstrates the advantages of closed-loop force control.
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Chapter 1

Introduction

Recovery of upper limb motor function from a stroke or spinal cord injury requires high intensity and high repetition therapy, for which robotic devices are well-suited. Some robotic therapies have focused on improving functional outcomes through increasing muscle strength through constraint-based therapies or re-learning movements through passive robotic assistance [1]. While these training paradigms are important and have shown some success in rehabilitating the upper limb of spinal cord and stroke injury patients [2], research has shown that a patient must be actively engaged during such robotic therapy to maximize the benefits of the therapy [3, 4]. As a result, there is an increasing focus on the ability to control the interactions between the robot and the user in order to maximize engagement and, in turn, to maximize the effectiveness of robotic therapy [4, 5]. One approach to providing more engaging tasks is the implementation of interaction controllers on the robot. In interaction control, the robot’s dynamic properties are varied through control to render a range of impedances to be used for more active engagement during therapy [6, 7]. This thesis describes the implementation of interaction controllers on two 1-degree of freedom (DOF) linear platforms using two different custom force sensing mechanisms and discusses the benefits of these approaches for incorporation in rehabilitation robots.
1.1 Background

Neurological disorders, such as a stroke or spinal cord injury, can have a significant impact on a person’s ability to lead an independent lifestyle. In the United States, 12,000 cases of spinal cord injury (SCI) and 795,000 cases of stroke occur each year [8, 9]. The economic impacts of such disorders are enormous, with the annual costs from stroke, which is the leading cause of long-term disabilities in the United States, being $38.6 billion. By 2030, it is predicted that 4% of adults in the U.S. population will have suffered a stroke, and the cost will rise to $183 billion annually [10]. The total annual costs associated with spinal cord injuries are $21 billion [11].

Incomplete tetraplegia is a common outcome in stroke survivors, and since 2000 it is the most frequent neurological category at discharge after spinal cord injury [12]. Neurologically induced weakness of the upper limbs is observed following tetraplegia and results in partial or complete paralysis of muscles [13]. This weakness can affect one or both of the upper limbs, causing loss of motor control and strength. Although a portion of those affected by tetraplegia in the upper limb recover, a majority do not [14]. The reduced motor control and strength can be detrimental since the upper limb plays a crucial role in the affected person’s daily activities, especially when the wrist is affected.

As a result of reduced upper limb functionality, tasks we often take for granted, such as holding a key to open the door or holding a glass of water, are often difficult or not possible after a stroke or spinal cord injury. These activities of daily living (ADLs) are essential in leading an independent lifestyle and are why more than 70% of tetraplegic persons with SCI reported that upper limb functionality was an important factor in their quality of life [15]. Improvements in functionality have been found to come through intensive rehabilitation in which a therapist helps a patient use the
affected limb to regain strength. These therapies have been shown to be significant, but even cases where improvements are not significant, it is believed that even small improvements in strength in patients with tetraplegia contribute substantially to their ability to use their hands in a functional way [16].

Despite the possible improvements shown from rehabilitation of stroke and spinal cord injury patients, a methodical and repeatable way to administer therapy sessions has not been established [17]. To address this need, robotic devices have been created to help with rehabilitating the upper limb of such patients. These robots have been coined rehabilitation robots and are typically either exoskeleton based [18], or end effector based [3]. In exoskeleton designs, joint axes usually align with those of human axes allowing for directly measuring joint movements and applying torques to specific joints. In end-effector designs, the user interacts with the robot only at one point which allows for better adjustability and easier mechanical design, but does not allow for applying torques directly to specific joints [19].

Rehabilitation robots, such as the RiceWrist [18] and MIT-Manus [3], have shown their ability to effectively rehabilitate the upper limb of spinal cord injury and stroke patients through robotic therapy [2]. Additionally, rehabilitation robots have advantages over traditional therapy in that robots do not tire and are repeatable; that is, they can reproduce the same protocol over time. They also measure movement of the person through the motor encoders. The encoder data can be analyzed to predict outcomes of the patient or used to quantitatively assess improvements in motor coordination. These robots could potentially help drive down the significant costs associated with these neurological disorders [20]. This could be possible through automating therapy, enabling therapy at home or in group therapy sessions in which one therapist supervises multiple patients using rehabilitation robots [3].
The three primary operation modes of robotic rehabilitation devices are patient-in-charge, robot-in-charge, and challenge-based modes, also referred to as active, passive, and active-constrained modes [2, 4, 21, 22]. In the patient-in-charge mode, the user backdrives the robot and so moves their arm through movements without robotic assistance. This condition requires the robot to be transparent, meaning that movements for the user when backdriving the robot should require as small of force as possible to resemble movements in free space. In the robot-in-charge mode, the user is passive and the robot takes the user through pre-planned movement profiles. Finally, in challenge-based modes, the robot in some way constrains or impedes user motion to achieve active training. For example, for the RiceWrist, the active-constrained mode consists of applying a resistive viscous force field to the user such that as velocity is increased, the user must apply more force to move through the field [2, 18]. Subcategories exist within each branch, including haptic simulation strategies in which the user engages with a haptic environment or assist-as-needed protocols in which the robot estimates the user’s contribution to movements and applies more or less assistance as required. For an extensive list of robotic therapy protocols, see [1].

1.2 Motivation

Studies have shown that in persons with spinal cord injury or stroke, pure motion trajectory training might not be the most effective way to provide therapy [3]. As such, research groups have begun exploring assist-as-needed protocols, interaction modalities, or other impedance-based approaches to improve robotic rehabilitation [4, 5, 23]. Typically these control modes are implemented in an open-loop manner, without force sensing. In force-feedback control, the force of interaction between the robot and the environment is measured and fed back to the controller, which
specifies the new desired force commands, driving the actuators. This closed-loop control approach allows for the use of interaction controllers, even when the device’s dynamics are significant or the actuator is non-backdrivable. This thesis explores two such scenarios where force sensing capabilities are required to enable implementation of interaction controllers.

In the first case, a solution is pursued that enables force sensing after the robot has already been manufactured. This solution could be beneficial for robots such as the RiceWrist, and other rehabilitation robots [3], that implement open-loop interaction control because they do not directly measure force. While this open-loop approach can be implemented with success in robots with negligible dynamics, or through model-based dynamic compensation schemes [24], the performance of model-based approaches is often compromised by modeling inaccuracies, such as neglecting higher order and approximating nonlinear dynamical effects (e.g. friction). As such, a means to close the force-feedback loop by measuring forces at the end effector of rehabilitation robots is required to overcome limitations from open-loop control. An additional benefit of measuring force at the end effector of the rehabilitation robot is the ability to monitor grip forces during therapy, which can be of clinical interest.

In the second case, interaction controllers are implemented on a device with an integrated force measuring scheme, necessitated by actuator dynamics. A 1DOF magnetic resonance (MR) compatible series elastic actuated (SEA) device has been developed by our lab which uses an ultrasonic motor for actuation. The MR-compatible ultrasonic motor is non-backdrivable, necessitating the use of force-feedback to implement therapy-like protocols during functional magnetic resonance imaging (fMRI). MR-compatibility is necessary to facilitate the use of the robot during continuous fMRI, so that changes in the brain over the course of robotic therapy can be mea-
sured. Such measurements will assist our research group in determining which rehabilitation protocols should be used to make robotic therapy most effective [1]. The 1DOF MR-compatible actuator will be integrated into a fully MR-compatible version of the RiceWrist [18]. The MR-compatible version of the RiceWrist, which will consist of three of these modules in parallel, needs to be able to perform the general robotic protocols of assisting and resisting the user, as well as being transparent for passive modes. Series elastic actuation was chosen as the means for force-feedback to decouple motor non-linearities and for implementing impedance controllers, that can exhibit these desired behaviors.

1.3 Objective

The objective of this thesis is to characterize and implement interaction controllers on two force sensing mechanisms for intended application to rehabilitation robots. The first mechanism is a grip force sensor which measures interaction forces at the end effector; the second device is a 1DOF MR-compatible SEA. The two approaches to measuring interaction force are differentiated in their location of force measurement on the device. The grip force sensor measures force at the end effector, while the MR-compatible device measures force co-located with the actuator. Experimental implementation serves to demonstrate the feasibility of interaction control and potential for rehabilitation applications for each scenario.

1.4 Contributions

The contributions of this thesis have been disseminated at or submitted to conferences, or submitted to refereed journals. The contributions include the development of a grip force sensor for measuring interaction forces at the end effector, prelimi-
nary experiments evaluating effectiveness of the device for implementing interaction controllers, and implementing interaction controllers on the 1DOF MR-compatible SEA.

A custom sensorized handle was developed, called the RiceWrist-Grip, with the capability of measuring grip or interaction forces during robot-aided rehabilitation therapy. Experiments assessed the suitability of using force sensing resistors (FSRs) to implement force-feedback interaction controllers. In the force-feedback control condition, the applied force for constant speed motion of a linear 1DOF haptic was reduced compared to the uncontrolled condition, demonstrating the possibility of improving transparency through force-feedback via FSRs. The results of this study were published in the proceedings of the ASME Dynamic Systems and Control Conference (DSCC 2013) [25].

A single case study of robotic wrist rehabilitation of incomplete spinal cord injury rehabilitation for an individual with injury at the C3-5 level was conducted. The case study demonstrated the capabilities of the RiceWrist-Grip to measure grip forces and to integrate with the RiceWrist-S exoskeleton rehabilitation robot. When grip strength of the patient was measured pre- and post-therapy with the RiceWrist-Grip, the measurements correlated favorably with the clinically accepted hand dynamometer. Additionally, it was observed that these grip force measurements might offer insight into fatigue of the patient. The fatigue information could be used either in real-time or over the course of therapy to adapt the difficulty of training for a given therapy session. This work is part of a paper conditionally accepted for a special issue in Robotica [26].

The design and preliminary control testing of a novel linear magnetic resonance compatible (MR-C) series elastic actuator (SEA) are presented. The device con-
sists of a non-backdrivable ultrasonic motor connected in series to the load through two extensions springs which act in parallel to each other. System parameters were obtained and preliminary control capabilities were demonstrated. This work was presented at the 6th International IEEE Engineering in Medicine and Biology Conference on Neural Engineering (EMBS 2013) [27] and more recent results of this work have been submitted to the 5th IEEE/RAS-EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob 2014) [28].

1.5 Organization of the Thesis

This thesis is organized as follows: Chapter 2 presents the design and experimental testing of a novel grip force sensor for measuring grip and interaction forces at a rehabilitation robot’s end effector. Chapter 3 characterizes the ultrasonic motor and identifies system parameters and implements interaction controllers on the MR-compatible 1DOF SEA device. Chapter 4 summarizes the research findings in the thesis by comparing the two approaches for measuring force in rehabilitation robots and discusses future work for each project.
Chapter 2

Development of a Novel Grip Force Sensor for Measuring Grip and Interaction Forces

In this chapter, the development of a novel grip force sensor, the RiceWrist-Grip, is presented. RiceWrist-Grip operation, sensor selection, and grip force measurement methods are described. A single subject case study validates the use of the RiceWrist-Grip as a grip force sensor, and experiments show its potential for use as a force measuring device to enable implementation of interaction controllers.

Large portions of this chapter appear in [25, 26].

2.1 Background

While the wrist, forearm, and elbow play important roles in performing everyday activities, grip strength is also an important factor. Currently, however, upper limb rehabilitation devices do not typically measure grip strength, rather they focus on rehabilitating upper limb range of motion [2, 3]. Interestingly, despite not directly rehabilitating grip strength, patients using the RiceWrist have shown improved grip strength over the course of therapy [29]. Grip strength is typically measured through a hand dynamometer [30]. While using a dynamometer is effective, it can only measure grip strength in one direction and is a single function device [31]. Studies have shown that grip strength can vary largely depending on the hands’ orientation, making measurement in only one direction likely non-optimal [32]. Further, the dynamometer
cannot be easily integrated with a rehabilitation robot. Therefore, our research group sought to develop a means to measure grip force at the end effector of the RiceWrist to provide real-time measurement of such forces during therapy.

Although grip strength is not typically the focus of robotic rehabilitation, some groups have developed protocols that focus on grip strength rehabilitation. One such study is seen in the work of Kurillo, Goljar, and Bajd [33] who used a cylindrical handle with a force/torque sensor at the base for measuring grip strength (up to 300 N) of post-stroke patients. Kurillo et al. designed a training plan with the device (shown in Fig. 2.1(a)) that consisted of sessions in which the subject exerted maximum grip strength, followed by sessions involving force tracking of a random signal. The first session could help in increasing maximum grip strength while the second could help in modulating force control. Kurillo et al. found that by including this grip force rehabilitation 15 minutes a day, in addition to the patient’s regular therapy, more than half of ten subjects in a twenty day study experienced significant improvements in grip strength and ability to control grip force. Other examples of grip force sensors have been proposed, such as a cylindrical handle with a pressure mat [32] or a cylindrical handle with six shells (see Fig. 2.1(b)) containing strain gauges as in [34]. These three approaches to grip force measurement have demonstrated accurate grip force measurements, but require the use of bulky and heavy components.

As a result of the bulkiness and weight of the previously mentioned grip force sensors [32, 33, 34], a custom solution is preferred to incorporate a grip force sensor into a rehabilitation robot. All of the previous designs consisted of large metal parts which would either not integrate with the robot or add significant inertia to the device. With the advent of thin film polymer force sensing resistors (FSRs), the ability to incorporate lightweight and small force sensors into a rehabilitation device is
Figure 2.1: (a) The grip force sensor used by Kurillo et al. for grip strength rehabilitation protocols. Users apply grip forces by squeezing the two metal shells. This grip force is measured through a force/torque sensor located at the base of the handle [33]. (b) Grip force sensing device created by Wimer et al. The device consists of six aluminum arms with strain gauges mounted at the base of each arm to measure force applied on the device [34].

possible. FSRs are an appealing choice as the sensor in a cylindrical handle to measure grip forces during robotic therapy since they allow forces in multiple directions to be measured simultaneously. Additionally, FSRs allow for easy integration into the handle and manufacturing of the handle with plastic materials, unlike force/torque sensors or strain gauges [33, 34]. Force sensing resistors have found applications in haptics and physiological studies. For example, by placing FSRs in a glove, the force from each finger and the palm of the hand was recorded [35, 36, 37].

An additional exploratory aim of this project was the use of the integrated grip force sensor as measure of interaction forces applied by the user to the robot. With the device it could be possible to map forces applied to the grip force sensor at the end effector to joint forces/torques to enable interaction controllers with closed-loop force feedback. By closing the force feedback loop, better disturbance rejection of
a device rendering a force field might be achieved. Preliminary testing of the use of FSRs as the force feedback sensor in interaction controllers are presented in this thesis.

This chapter is organized as follows: Section 2.2 details the design and construction of the grip force sensor and describes how it can be used to measure grip and interaction forces. Section 2.3 describes the FSRs used in the device that allow for grip force measurement. The resistance of the sensors at various compression forces is measured to show the potential for obtaining linear force measurements with an inverting operational amplifier. Three different calibrations of the sensors are performed, showing their capability to measure a variety of force ranges depending on the signal conditioning circuit’s components. Section 2.4 validates the novel grip force sensor’s ability to measure grip forces in a single subject case study. Section 2.5 describes the experimental setup used and presents the experimental results for testing the use of FSRs as the force sensor for zero force controllers.

2.2 Device Design

Incorporating a grip force sensor as the end effector to an upper limb rehabilitation device necessitates a compact and lightweight design. The design must be compact, so as not to interfere with upper limb range of motion rehabilitation, and lightweight, so as to not increase the inertia at the end effector. This increase in inertia is undesirable due to the increased loading on the actuators and increased effort required from the user to move the exoskeleton. An additional consideration of the device is having a diameter that is comfortable to the user. Studies have shown that people gripping a cylindrical handle have maximum grip strength when the handle diameter is between 33-34 mm [38, 39, 40], although it is possible for maximum grip strength to occur
outside of the 33-34 mm range, for example at 30 mm or 40 mm. The target diameter for the grip force sensor was chosen as 33-34 mm, so as to be in the optimal diameter range. For length, the grip force sensor should be approximately 104 mm, the same size as the current handle on the RiceWrist-S [41], but no more than 150 mm.

To measure grip strength, it is necessary to measure enough components of an applied grip force so that it can be decomposed into equal and opposite components. The developed novel grip force sensor, called the RiceWrist-Grip (see Fig. 2.2), can measure forces in three principal directions using six thin film force sensing resistors (FSRs). The main body of the sensor and the shell slices were created by a rapid prototyping machine with 0.51 mm resolution from Acrylonitrile Butadiene Styrene (ABS) plastic. The RiceWrist-Grip weighs 92 g, is 34 mm in diameter, and 126 mm in length. The RiceWrist-Grip is incorporated into the RiceWrist-S [41] (see Fig. 2.3(a)), a serial version of the RiceWrist [18], to enable real-time monitoring of grip strength for the purposes of assessment and determination of rehabilitation efficacy. A CAD model of the device in use with the RiceWrist-S can be seen in Fig. 2.3(b).

To measure the axis of maximum grip force and directionality of interaction forces between the user and the handle during therapy, measuring force in at least three directions is necessary [32]. For the RiceWrist-Grip, sensors are placed in two sets of three sensors, evenly spaced on the inner circumference of the handle. Three cylindrical shell slices, with two flats facing the sensors, are used so that all load is transferred to the FSRs. To control the load area on the FSRs, 1.5 mm thick and 8.5 mm diameter plastic disks were placed between the sensing area of the FSRs and the flats on the shells, increasing the repeatability of the FSRs response to an applied force. Having two sensors per shell slice allows for the slices to deflect minimally, as opposed to having only one sensor, allowing for a stable grip.
Figure 2.2: The RiceWrist-Grip force sensing handle. The sensor measures grip force through six force sensing resistors (FSRs). Two flats on the shell housing contact plastic disks placed on the FSRs, transferring all the grip force to them for accurate measurement.

Using the measurements from each of the FSRs, the grip force on the sensor can be determined. Due to the flats on the shells and plastic disks placed on the FSRs, all load is transferred directly from the shells to the sensors (forces shown in Fig. 2.4). When grip force is being measured, the forces can be summed from

\[ F_i = F_{i,a} + F_{i,b} \]  

where \( i = 1, 2, 3 \), so that the problem becomes 2D (Fig. 2.5). The resulting forces can be averaged to obtain a measure of grip force [32]. If instead interaction forces are being measured, the difference in applied vertical and horizontal forces could be used to obtain the equivalent 3D force/torque vector applied at the end effector.
2.3 RiceWrist-Grip Sensor Selection

Tekscan’s FlexiForce A301 force sensing resistor was chosen due to its size, 3% linearity, 2.5% repeatability, 4.5% hysteresis, 0-440 N force range, and 5 µs response time. The A301 sensor is 25.4 mm long and 0.203 mm thick with a 9.53 mm diameter sensing area. The FSR is piezoresistive meaning that as force is applied its resistance
changes. This occurs due to the strands on the sensor contacting each other as force is applied, lowering the resistance of the sensor. At no load, few of the strands are contacting each other, resulting in essentially infinite resistance.

When a force is applied to an FSR, its resistance changes as \( \frac{1}{R} \), and so its conductance, \( C = \frac{1}{R} \), is mostly linear with the applied force (Fig. 2.6). To provide a linear voltage reading, an inverting op-amp circuit was used (see Fig. 2.7). With this configuration, the output voltage is given by

\[
V = -V_{in} \frac{R_f}{R_{FSR}}
\]  

(2.2)

where \( V \) is the FSR output voltage, \( V_{in} \) the input voltage to the FSR, \( R_f \) a feedback resistor, and \( R_{FSR} \) is the variable resistance from the FSR. The feedback resistor and input voltage to the FSR need to be chosen appropriately to allow for the output voltage to remain within the op-amp supply voltage range for every admissible value of applied force.

The FlexiForce A301 has a large range of measurable force, from 0-400 N. For low force ranges (< 20 N), the reading from the FSR sensor was calibrated with
Figure 2.6: Experimentally measured resistance vs. force curve for the FlexiForce A301 FSR. The conductance of the sensor is linear to applied force so utilizing an inverting op-amp allows for a linear relation between output voltage and applied force.

Figure 2.7: Inverting op-amp circuit recommended by FlexiForce to create a linear relationship between applied force on the FSR and its output voltage (image taken from www.tekscan.com).

force measured from an ATI Nano17 (SI-12-0.12: 17 N range in z-direction, 3.125 mN resolution) force-torque sensor, by manually applying increasing levels of force in the compression direction. The Transducer Cable and Net Box (components of Net F/T system, ATI Industrial Automation) were used for signal conditioning of the Nano17 force sensor output. The resulting calibration can be seen in Fig. 2.8.
For this calibration, $R_f$ was chosen to be 100 kΩ and $V_{in}$ was chosen to be -6.6 V. Calibration of the FSR was performed by placing a plastic disk between the FSR and Nano17 force sensor to allow for all load to be transferred to the FSRs sensing area. At no load, $R_{FSR}$ was greater than 5MΩ which allowed for zero voltage to be measured at zero force.

The calibration curve in Fig. 2.8 shows a slightly non-linear response of FSR voltage to applied force. The sensor has a higher sensitivity in the low-force region and then settles on a lower (but mostly constant) sensitivity value after this break-in region. A fifth order polynomial fitting was applied to interpolate the data, as already proposed by [42]. In this case, the fifth order polynomial was chosen as

$$F_a = 0.09V^5 - 0.98V^4 + 3.75V^3 - 6.57V^2 + 9.06V - 0.01$$

(2.3)

where $F_a$ is the force applied to the FSR. The polynomial allows for interpolation of force values throughout both the non-linear and linear range of the sensor. With this calibration, the FSR can measure forces between 0-15 N accurately. This accuracy was examined by comparing measured forces from the ATI Nano17 force transducer and the FlexiForce A301 sensor (Fig. 2.9). The comparison shows that the maximum difference between the two measurements is 1.45 N, a difference of about 10%.

While the previous calibration shows the potential for measurement of small forces, a different calibration is required for measuring forces in the range of 0-100 N. For this calibration, $R_f$ was chosen to be 100 kΩ and $V_{in}$ was chosen to be -4.5 V. To calibrate the sensors, an ATI Industrial Automation Nano17 SI-25-0.25 six-axis force/torque transducer with 0.00625 N resolution was used. A single FSR was prepared for calibration by placing a plastic cylinder between the FSR and the Nano17 force sensor and applying manual compression forces to the FSR. The resulting calibration (see
Figure 2.8: Calibration of the FSR voltage with force measured with the ATI Nano17 force transducer. The calibration resulted in a fifth order polynomial fitting of the data to account for the nonlinearity of the sensor at low forces.

Figure 2.9: Comparison between the force measured with the FSR and with ATI Nano17 force/torque sensor, in a dynamic compression test.
Fig. 2.10) shows that at low forces (< 11.5 N), the sensor had a nonlinear output appropriately described with a polynomial fitting (such as that used in [42]), while for larger forces there existed a linear relationship. The coefficient of determination ($R^2$) in the linear portion of Fig. 2.10 was 0.9964, verifying the existence of a linear correlation between the two variables in this region. The ATI Nano17 is limited to 35 N in the z-axis so calibration was only performed up to 35 N. It should be noted that the voltage of the FSR could approach 6 V (limited by the op-amp) and so forces of up to 100 N could be measured. The linearity of the FSRs for voltages above the range shown in Fig. 2.10 is shown in the following calibration, justifying this linear extrapolation.

![Figure 2.10](image)

Figure 2.10: Calibration of a single FSR with using the ATI Nano17 force sensor. For forces under < 11.5 N, a polynomial fitting was used, while a linear fit was used for after this non-linear break in region. The $R^2$ value shows the correlation coefficient in the linear region.

For larger force range calibrations (> 100 N), an Instron Model 4500 testing machine was used to apply compression forces on an FSR. In this calibration, a 22 kΩ feedback resistor and ~4.19 V input voltage were used. The displacement of the
Instron was set to 1 mm/min for quasi-static loading. Voltage data was collected using a National Instruments myDAQ. Both measurements were sampled at 25 Hz and the data sets were synchronized through a trigger. The force data from the Instron and voltage data were post-processed with a moving average filter with a 10 sample window (400 ms) to filter load cell noise. The resulting calibration curve (Fig. 2.11) shows linearity between the applied force and output voltage ($R^2 = 0.999$) with a slope of $0.03 \frac{V}{N}$.

![Figure 2.11: Calibration of a FlexiForce A301 sensor using an Instron testing machine. The sensor was placed between materials made of ABS plastic and had a small plastic disk on the sensing area to ensure load was transferred to it. This reproduced the same loading conditions as in use with the RiceWrist-Grip, for an accurate calibration.](image)

The three described calibration methods demonstrate the ability of FSRs to measure forces accurately over a wide range of forces. By changing the values of the feedback resistor ($R_f$) and the input voltage ($V_{in}$) in the inverting op-amp circuit (Fig. 2.7), the range of measurable forces can be adjusted. Depending on the appli-
cation, one of the three calibration methods should be used for calibrating the FSRs. If the application dictates measuring small forces (< 15 N), a fifth order polynomial can be used for converting output voltage to applied force. For forces up to 100 N, while still having good sensitivity to low forces (< 11 N), a combination of a polynomial fitting and linear fitting can be used. Finally, for large forces ranges (> 100 N), a linear relationship is obtained between the two variables. A summary of the measurable force ranges, circuit component values, and fitting type can be seen in Table 2.1.

Table 2.1: FSR Calibration Parameters

<table>
<thead>
<tr>
<th>Force range (N)</th>
<th>Feedback Resistor (kΩ)</th>
<th>Supply voltage (V)</th>
<th>Fitting</th>
</tr>
</thead>
<tbody>
<tr>
<td>0-20 N</td>
<td>100</td>
<td>-6.6</td>
<td>polynomial</td>
</tr>
<tr>
<td>0-100 N</td>
<td>100</td>
<td>-4.5</td>
<td>polynomial &amp; linear</td>
</tr>
<tr>
<td>0-175 N</td>
<td>22</td>
<td>-4.19</td>
<td>linear</td>
</tr>
</tbody>
</table>

2.4 Validation of the RiceWrist-Grip

The RiceWrist-Grip was used to measure grip forces of a single subject pre- and post-wrist rehabilitation therapy with the RiceWrist-S. The subject (a 45-year-old male, right handed, incomplete SCI at the C3-5 level, American Spinal Injury Impairment Scale (AIS) C, 83 months post-injury) participated in ten sessions of robotic-assisted arm training over the course of twenty days of training. One objective of the study was to evaluate whether the RiceWrist-Grip could be used accurately as a grip force measuring device during robotic therapy. Measurement accuracy would be deter-
mined by examining the correlation between post-therapy grip force measurements from the RiceWrist-Grip with those of a clinically accepted hand dynamometer.

Grip strength data were collected before and after each training session, first with the RiceWrist-Grip and then the dynamometer. Manual grip strength was measured with a hand-held standard adjustable dynamometer (Lafayette Instrument, Model 78010) with the subject in a seated position with shoulder adducted, elbow flexed, and forearm in mid-position. The subject was asked to perform three maximal voluntary contractions, and the maximum value of each attempt was recorded. For all cases, the subject was instructed to exert maximum grip effort for approximately two seconds with 10-20 seconds of rest in between maximum grip efforts [43]. The participant had a few minutes to rest between using each device.

Hand dynamometer grip force was taken as the average of the three trials and was visually recorded (device resolution of 2 N [31]). For the RiceWrist-Grip, grip force was calculated by first converting the voltage readings from the FSRs to force through the calibration in Fig. 2.10. This calibration was chosen due to its range of measurable forces (0-100 N) while still having sensitivity to low forces. To calculate total grip force, the measured forces were considered to be as pressures on the circumference of the handle, similar to [32], and so total grip force was calculated as

\[ F_{\text{grip}} = \frac{1}{n} \sum_{i=1}^{n} F_i \]  

(2.4)

where \( F_i \) are the equivalent forces on the circumference of the handle, \( F_{\text{grip}} \) is the grip force, and \( n \) is the number of forces on the handle, which for the RiceWrist-Grip \( n = 3 \). Each component of the grip force is computed by summing the values of the forces on the corresponding vertical pairs of FSRs (as described by equation 2.1 and illustrated in Fig. 2.5). Grip force was averaged across the extent of the trial (\( \sim 2 \))
s), and then the three average values for each trial were averaged to obtain the mean grip strength.

The mean post grip force measurements are reported for Sessions 2 through 10 for the participant’s left hand (Session 1 was excluded due to incomplete data collection), and compared with the estimate of grip force obtained through the commercial dynamometer. Results for the mean grip force measurements from the RiceWrist-Grip and hand dynamometer are shown in Fig. 2.12. The resulting correlation coefficient ($R^2 = 0.8335$) shows that there is good linear correlation between the two methods of grip force assessment, thereby providing preliminary validation of the use of the RiceWrist-Grip for assessment of an individual’s grip force capabilities.

Figure 2.12: Comparison of post-treatment grip forces across Sessions 2-10, measured by the RiceWrist-Grip and Lafayette hand dynamometer.
2.5 Force Feedback Control using FSRs on a 1DOF Linear Platform

2.5.1 Experimental Setup

A 1DOF testbed was used in order to evaluate the use of FSRs as a force measurement device for implementation of interaction controllers. An ABS housing was mounted to an aluminum platform which was translated on linear bearings. The platform was connected to a brushed DC motor (RE40 model 148877, Maxon Motors Corp.) that controlled the motion of the platform through a cable transmission (see Fig. 2.13). In this setup, direct force-feedback controllers can be applied by measuring the force at the port of interaction between the actuated device and the subject. A proportional-integral (PI) force controller (as shown in Fig. 2.14) acted on the force estimated from the FSR(s). By setting a value of desired force as $F_{\text{des}} = 0$, the robot attempted to render a transparent haptic interface to the user, such that a very small force was required to move the device.

During the experiments, the motor was either powered off, or current-controlled as in the scheme shown in Fig. 2.14, using the VoltPaq Q8 from Quanser Inc and a Matlab-Simulink model sampled at 1 kHz, translated in Real-Time code with a Quanser USB Q8 board for data acquisition. The two experimental conditions allow comparison of the effort required to complete the task. For these conditions, the main source of non-linearity in the transfer of force between the actuator and the subject is provided by the static friction in the linear bearings.
Figure 2.13: The FSR is attached to an ABS housing fixed to an aluminum slider, which slides on a linear bearing, driven by a DC-motor actuated cable transmission. The ABS housing also serves as an alignment of the center of the ATI Nano17 and FSR allowing for forces to be transmitted from the user to the load cell, to the FSR, and finally to the platform.

Figure 2.14: Block diagram representation of the zero-force control experiment. In the figure, $F_{des}$ is the desired force (0 for this experiment), $F_e$ is the external force applied by the user, $k_p$ is the proportional gain and $k_i$ is the integral gain applied on the error signal to the motor, and $F_m$ is the motor force. The external and motor force are both applied to the system and if larger than static friction result in motion of the system. When the user force is non-zero, an error signal is sent to the motor in order to achieve transparent behavior.

2.5.2 Pushing (Uni-directional) Interaction Forces

The first interaction scenario was that of a user applying a pushing force on a single FSR (Fig. 2.13). Due to the low forces required to move the slider, the calibration described by Equation 2.3 was used. In this condition, a subject applied the pushing
interaction force to the mounted FSR to regulate the motion of the slider, imposing a movement of 25 mm in approximately 1 sec at a relatively constant speed.

Friction compensation was successfully obtained with the simple force-feedback scheme shown in Fig. 2.14, validating the use of the FSR as a sensor suitable for implementation of interaction controllers in human-interacting robots. To quantify the improvements in interface transparency, the mean force required by the user was calculated for representative tasks of the two conditions, as shown in Fig 2.15. Using the forces measured from the ATI Nano17, the mean force applied for the no control case was 1.71 N and 0.28 N for the controlled case. Thus, with the interaction controller, 6.1 times less work was required to move the system, compared to having no controller. Although FSRs may not measure force as accurately as commercial force sensors, they have been demonstrated as useful for measuring interaction forces and for serving as a sensor for force-feedback controllers.

Figure 2.15 : (top) Applied position during the interaction test, during the considered experimental conditions. (bottom) Interaction force measured in the two experimental conditions. The average force required for the same motion in the controlled condition is approximately 6.1 times lower.
Pinch (Bi-Directional) Interaction Forces

In addition to testing uni-directional forces, the ability to implement zero force control using FSRs when interacting with bi-directional pinch forces was also investigated. The setup was similar to that used in the uni-directional force experiments, except that an additional FSR was mounted on the opposite side of the FSR housing to allow for measurement of force in both directions (see Fig. 2.16(a)). With this setup, a user could interact with the setup by pinching the FSRs. By measuring the force on each FSR and taking the difference of the measurements, the interaction force on the system was estimated (see Fig. 2.16(b)). In the following experiments the FSRs were calibrated similar to Fig. 2.10, so that a non-linear polynomial was used for forces < 5 N, and a linear correlation for larger forces. Each FSR needed to be calibrated individually to allow for accurate pinch force measurement.

Figure 2.16 : (a) Experimental setup for pinch interaction forces. An additional FSR was added to the FSR housing to allow for measuring forces in both directions. (b) User applying a light pinching interaction force to the FSRs.

An experiment was performed where a user applied a light pinching force to the setup while moving the platform dynamically. This experiment was performed under
two conditions. The first was with applying the zero force controller (in this case only with a proportional gain on the error) and a no control case with the motor powered on. A sample segment of one of these tests can be seen in Fig. 2.17 with two cases of proportional gains and a no control case. A regression was performed for the three cases, which included the effects of velocity and acceleration as well as a constant force. The regressions estimated the mass of the platform to be approximately 150 g. The damping coefficients were estimated to be 1.8, 0.7, and 7.6 $\frac{Ns}{m}$ for the cases of $k_p = 2$, $k_p = 6$, and no control respectively. The validity of this fit can be seen in Fig. 2.18 for the case of $k_p = 6$.

![Figure 2.17](image)

Figure 2.17: (top) Velocity profile applied by the user to the platform and (bottom) measured force (difference of the two FSR forces) applied on the system to move the system for the given velocity profile.

As can be seen from the regressions, the essential effect of the zero force controller is to reduce the damping felt by the user. By increasing the proportional gain on the force error, the perceived damping is decreased. This trend can be visually observed in Fig. 2.19 which plots force against velocity for the three experiments. Although
increasing the proportional gain decreases the perceived damping, having too low of damping will result in instability of the system. However, these experiments show that by using FSRs in pinch experiments, the perceived damping of the device can be reduced.

2.6 Summary

In this chapter, the development and characterization of a novel grip force sensor was presented. The grip force sensor, called the RiceWrist-Grip, easily integrates with most rehabilitation robots by using it as the end effector of the device. Integration of the device with the RiceWrist-S demonstrated its potential benefits to robotic therapy by providing a means to measure grip force with the rehabilitation robot. Additionally, with access to interaction forces on the handle, it might be possible
Figure 2.19: Estimated pinch interaction force from the FSRs plotted against rail velocity. The solid lines show the velocity dependent force found for each case from the regression. The data sets are down sampled to 10 Hz for visualization.

to use the device as a force sensor for interaction controllers. Preliminary control testing in this thesis indicated that on a 1DOF platform, zero force control could be implemented in a stable manner. By using FSRs for zero force control, static friction and damping of the device could be compensated for, demonstrating the need for future experiments to further evaluate the efficacy of the device for dynamic compensation.
Chapter 3

Interaction Control of a 1DOF Magnetic Resonance Compatible Device

Robotic devices that are MR-compatible enable researchers to study the neural correlates of movement while manipulating a person’s limbs through prescribed motion trajectories and force fields. MR-compatibility requirements pose stringent constraints on material and component selection. Our group has developed a 1DOF series elastic actuated (SEA) prototype that demonstrates compatibility in an MRI scanner [27, 28, 44]. The 1DOF module will be replicated in the same kinematic structure as the RiceWrist to enable therapy like wrist motor protocols during continuous fMRI [18].

The 1DOF device employs a series elastic architecture (Fig. 3.1) in order to allow for implementing force-feedback control. Being able to measure force was necessary since the robot will need to be able to implement transparent environments to the user. The ultrasonic motor used as the actuator of the device is non-backdrivable and so without a measure of force, rendering a transparent environment with the device would be impossible. Our group’s design incorporates a series elastic element, as introduced in [45], for force sensing that uses readily available off-the-shelf components (i.e. extension springs). The added benefit of compliance is that the user is decoupled from the motor’s nonlinearities and its non-backdrivability [45, 46, 47, 48]. With compliance, if the motor is blocked, the user feels the stiffness of the springs instead of the motor’s non-backdrivability. Additionally, interaction forces are estimated by
measuring the springs’ deflection and knowledge of the springs stiffness allowing for force-feedback control (see Fig. 3.1). This control is necessary for the type of interactive and transparent protocols that will be implemented in future wrist motor protocols during fMRI. An alternative approach to the use of SEAs for force sensing would have been to use an MR-compatible force/torque sensor. These sensors use an optical sensor and fiber to measure a differential in light intensity [49, 50]. Although these sensors have been shown to be MR-compatible, they are often difficult to manufacture.

Figure 3.1 : Schematic of a series elastic actuator (SEA). An SEA consists of an actuator connected in series to a load through a compliant (i.e. elastic) element (from [51]).

In this chapter, the control of this 1DOF linear magnetic resonance compatible series elastic actuator is presented. The characteristics of the ultrasonic motor (USM) used for actuation of the system are determined through static calibration and examining the built-in velocity controller’s load disturbance rejection capabilities. Experimental results with the 1DOF device indicate that it has suitable performance for implementing high fidelity virtual stiffnesses along with high transparency during zero force control.

This chapter is organized as follows: Section 3.1 details the components and operation principles of the linear 1DOF MR-compatible SEA system. Section 3.2 provides experimental results to characterize the USM’s built-in velocity controller. Section 3.3
uses system identification techniques to find the parameters of the 1DOF system, including spring rate, natural frequency, mass, and damping. Finally, Section 3.4 presents the implementation of virtual stiffness and zero force controllers on the system, along with a frequency domain analysis to determine the range of application of these controllers.

Portions of this chapter are published, submitted, or in preparation in [27, 28, 44], and I greatly acknowledge my fellow researchers’ contributions to these publications.

3.1 System Components

The 1DOF module consists of an ultrasonic motor (Shinsei USR60-E3N), a threaded spool (12.5 mm pitch diameter, 2 mm pitch) attached to the motor shaft through a brass set screw, a cable transmission (0.4 mm diameter SpiderWire Stealth Teflon-coated polyethylene cable), two custom-designed pretensioned phosphor bronze extension springs (stiffness: 1.9 N/mm, max force: 36 N, pre-tension: 12 N), a custom 3D printed ABS plastic slider, and a non-magnetic linear ceramic ball bearing, produced by DelTron Precision Inc. The rotational motion of the USM is converted to linear motion through a cable transmission which connects the motor’s rotations to two springs in parallel to the load (or end effector) which are connected to a slider through eye bolts. The total travel of the linear actuator is 35 mm, accounting also for the reduction in workspace required by maximum spring deflection under loading. The system can be seen in Fig. 3.2.

The 1DOF MR-compatible actuation module uses a 60 W Shinsei USR60-E3N ultrasonic piezoelectric motor. The MR-compatibility of Shinsei’s USM has been tested up to a 7T environment [52]. The motor comes with a factory built velocity controller, allowing for precise control of its velocity, and a 1000 CPR (counts-per-revolution)
optical encoder. Drawbacks of the USM are its nonlinear velocity characteristics and 8 ms time delay when switching directions. In fact, using the built-in velocity controller, the minimum velocity that can be regulated is 14 rpm, and the Shinsei built-in velocity controller does not provide a means to regulate motor velocity at lower amplitudes.

The ultrasonic motor (USM) has an operating curve as shown in Fig. 3.3. Based on maximum torques required for use in the 2DOF wrist device, the maximum interaction force is estimated to be 20 N [44]. With the effective spool radius being 5.5 mm, the resulting torque on the USM is 0.11 Nm. From the motor performance curve, this torque value is well within the limits of the continuous operation region.

In our MR-compatible SEA, the compliance is realized with two extensions springs, one on either side of the USM, having the same displacement such that when one spring compresses, the other extends. Both springs are pre-loaded to approximately the same load with the range of measurable forces equal to twice the pre-load. Since the USM is non-backdrivable, the springs do not have to be pre-tensioned to precisely
Figure 3.3: Torque vs. velocity curve for the ultrasonic motor included in the co-located force feedback actuation scheme (from [51]). Data points were taken from the manufacturer’s catalog.

the same load since any discrepancy will simply result in a small loading on the USM shaft. A schematic of the USM SEA can be seen in Fig. 3.4. A cascaded control approach with an inner velocity loop (closed by the USM’s velocity controller) and an outer force-feedback loop is depicted. Measuring spring deflection is accomplished through measurement of both motor rotation (via a 1000 CPR incremental encoder) and of the load displacement (via a 500 LPI (lines-per-inch) incremental linear optical encoder (US Digital)). Taking into account quantization errors (approximately 0.01 mm for both encoders), the resulting quantization error for measuring interaction force is 0.04 N.
3.2 Motor Characterization

Knowledge of the the motor’s characteristics is required in order to implement closed-loop control. Although the motor is packaged with a built-in velocity controller, limited information on its capabilities is available from the manufacturer or in the literature. The controller converts commanded voltages into ultrasonic waves and uses the motor encoder for closed-loop feedback on the motor velocity. Specifically required was the relationship between a voltage command to the controller velocity, and how externally applied loads to the USM would affect this relationship.

3.2.1 Steady State Motor Velocity Calibration

The first characterization was the relationship between commanded voltage to the USM velocity controller and output motor velocity. A sequence of step voltages was applied to the built-in velocity controller, and held for a set amount of time (3-5 s) to allow for the motor shaft displacement to settle. The voltage sequence and resulting motor velocity profile can be seen in Fig. 3.5. The steady state values of velocity to an applied voltage were obtained through averaging over the steady state region of the velocity. Using this data, the relationship between commanded voltage and output motor velocity is shown in Fig. 3.6. The calibration reveals that there
exists a linear region between voltage and velocity (49 RPM/V), as well as saturation at large velocities (170 RPM), and a discontinuity near zero desired velocity from the minimum regulated velocity (14 RPM). These characteristics are defined in the following conditions:

\[
f(v) = \begin{cases} 
  v_{min} \cdot \text{sign}(v) & \text{if } |v| < v_{min} \\
  v & \text{if } v_{min} < |v| \leq v_{max} \\
  v_{max} \cdot \text{sign}(v) & \text{if } |v| > v_{max}
\end{cases}
\] (3.1)

where \( v \) is desired motor velocity, \( v_{min} \) is the lowest possible motor velocity (14 RPM), and \( v_{max} \) is the motor velocity’s saturation limit (170 RPM).

Figure 3.5: Desired (or commanded) voltage and measured velocities of the motor. Although some overshoot exists at higher velocities, in general the controller tracks the signal quite accurately and quickly (< 10 ms).
Figure 3.6: Calibration curve of commanded voltage to the built-in velocity controller and USM velocity. As can be seen, the motor has a linear region, saturation at large velocities, discontinuity around zero desired velocity, and a minimum regulated velocity.

3.2.2 Velocity Response to an Externally Applied Torque

With the relationship between commanded voltage and motor velocity known, it was desired to determine how this relationship held for external disturbances. In this section, the velocity responses when connected to a viscous damper and with an active torque applied through a DC motor are evaluated.

Constant Torque

The first load disturbance test was to evaluate the motor controller’s ability to reject a torque applied from a viscous damper. A viscous damper (Rotary Viscous Damper RVD-010 \( b = 0.01 \frac{Nm}{RPM} \), ITT Enidine Innovations) was connected to the output shaft of the USM through gearing. The gear ratio was 4 with the larger gear connected to the damper. By commanding a desired value and then determining the steady state velocity when connected to the damper, the controller’s ability to reject a variety
of loads can be evaluated. Repeating the same voltage command sequence as in Fig. 3.5, the calibration curve shown in Fig. 3.7 was obtained. As can be seen, the resulting calibration is essentially the same, and thus constant loads have no effect on the steady state value of desired velocity. There is a difference in slope for both the undamped and damped cases in Fig. 3.7 compared to Fig. 3.5, but is approximately 2.5% and so within typical calibration accuracy. Note that the load disturbances were rejected for almost twice the expected maximum torque (0.11 Nm) on the USM.

![Graph of command voltage vs. motor velocity](image)

Figure 3.7: Effect of a velocity dependent damping torque on the motor for a desired constant velocity. The torque has little effect on the motor’s velocity profile.

### Varying Torque

With the motor’s characterization to constant loads known, it was also of interest to determine the response to large step torques and the admittance transfer function (ratio of USM velocity change to a unitary external torque). Experiments were conducted connecting the USM to a Maxon RE40 DC motor through gearing \( G = \frac{\omega_{\text{RE40}}}{\omega_{\text{USM}}} = 4 \), as shown in Fig. 3.8. The gear ratio was chosen to amplify the torque from the DC motor so as to stay within its torque performance limits. Current control was used
to regulate the torque of the DC motor with the torque constant being $60.3 \, \text{mNm/} \text{A}$. Current control was implemented on the DC motor using a 12A8 PWM servo drive (Advanced Motion Controls) to regulate the perturbation torque on the USM. The servo gain was set to 4.9 A/V.

![Figure 3.8](image)

A sequence of step torques with amplitude of 0.05 Nm was applied to the USM to investigate its velocity response to small changes in torque while the torque increased. The resulting velocity response and torque profile can be seen in Fig. 3.9. The USM’s velocity shows insignificant changes to the step torques, even when the torque becomes large. However, in this case the magnitude of the steps were small. The USM was then subjected to step torques at half the maximum continuous torque (0.25 Nm).
The step torques were applied in both directions so as to examine the effects when going in the same direction as the motor and opposing it. The velocity responses can be seen in Fig. 3.10. Here there is some overshoot and settling time (< 10 ms), but the motor controller again rejects the step disturbances.

![Figure 3.9: Velocity response of the USM when given a desired constant velocity to step torques of varying sizes. As in the previous experiments of constant loading, the built-in velocity controller was able to reject these disturbances.](image)

With the motor’s response to constant loads fully characterized, it was also of interest to determine the admittance transfer function. To apply a perturbation torque of varying frequency, a Schroeder multisine profile was generated with a variable peak amplitude (max of 0.092 Nm) and flat frequency spectrum in the range of 0.5 - 10 Hz (see Fig. 3.11). Load velocity was obtained using a Savitzky-Golay filter on the USM position. Measuring the change in velocity resulting from the application of torque, the admittance transfer function was found at four different velocities. These transfer functions were obtained by using the Matlab tfestimate command, which provides a non-parametric estimate of a transfer function. Additionally, the coherence of the estimated transfer functions were obtained with the mscohere Matlab command.
Figure 3.10: Step torque of half of the maximum rating of the USM’s continuous torque limit. As can be seen, the torque has some effect on the USM’s velocity, but it is able to quickly converge on the desired velocity.

Figure 3.11: Multisine wave used for commanding the current of the RE40 DC motor to provide a multisine torque disturbance on the USM, allowing for excitation frequencies between 0.1 - 10 Hz.

The resulting admittance transfer function can be seen in Fig. 3.12 and the coherence of the transfer functions can be seen in Fig. 3.13. From the admittance transfer functions, at 3 Hz the magnitude is 20 dB. With the largest interaction torque
on the USM of 0.11 Nm, the difference in velocities \((\omega - \omega_{\text{des}})\) is equal to 1.1 RPM. When \(\omega_{\text{des}} = 50\) RPM, the resulting change in velocity is 2%. Thus, even for the case of the maximum expected interaction force at 3 Hz, which is faster than typical human movements [53], is negligible. The coherence plot reveals that, in general, the model achieved good coherence (> 0.8); however, for the 90 RPM case, the coherence was poor. This was a result of the experimental setup since the two spur gears were not perfectly aligned. At faster speeds, the misalignment increased, resulting in more variable torque transfer from the RE40 to the USM.

Figure 3.12: Bode plot of the USM’s velocity for a desired constant velocity in response to an externally applied torque (about one-fifth the maximum of the USM). The experiment was performed motor speeds in the typical expected operating region of the device.

3.3 System Identification

3.3.1 System Parameters

To be able to use the system as described in Section 3.1 for force-feedback control schemes, the spring stiffness constant, system natural frequency, and static friction
Figure 3.13: Coherence of the model for the Bode plot. The coherence is generally quite good (> 0.9) over the frequency range of interest (0.5 - 10 Hz), indicating good model accuracy. Lower coherence for the 90 RPM case is likely attributable to experimental conditions.

needed to characterized. The spring stiffness was manually characterized by statically pushing on an ATI Industrial Automation Nano17 (SI-25-0.25: 35 N range in z-direction, 6.25 mN resolution) force/torque transducer connected to the end of the slider with the force being aligned with the spring deflection. Given the non-backdrivability of the motor, the resulting force vs. position curve corresponds to the intrinsic characteristic of the series elastic element (Fig. 3.15). The resulting spring constant was found to be 3.8 N/mm, with a maximum nonlinearity error below 0.2 N, 1% of the total actuator force (20 N).

The natural frequency of the system was found in a blocked motor condition (i.e. motor powered off) by manually displacing the system from equilibrium and releasing it (see Fig. 3.16). Since the system is underdamped, the system went into a mode of free vibration until brought to rest by the small amount of dissipation in the system. Using this method, the natural frequency was found to be 207.35 rad/s (33 Hz). With
Figure 3.14: MR-compatible system showing the ATI Nano17 attached for measuring force applied by the user for system characterization.

Figure 3.15: Force vs. load position to determine the spring rate of the co-located MR-compatible device \(k_s = \frac{F_{\text{max}}}{x_L} = 3.8 \text{ N/mm}\)

the spring constant and natural frequency known, the mass of the system could also be estimated from

\[
m_L = \frac{k_s}{\omega_n^2}
\]  

(3.2)

and was found to be 90 grams. Using the logarithmic decrement method, the damping
coefficient \((\zeta)\) was found to be 0.1. As demonstrated in [54], it is possible to also use the free response data to measure static friction; however, the resulting static friction found using this method (0.6 mN) was smaller than the force resolution of 0.04 N of the SEA as limited by the load’s optical encoder. The low damping and static friction present in the system allows for accurate force estimation from measuring load deflection. A summary of the system parameters is seen in Table 3.1.

![Figure 3.16: Free response of the system from an initial load displacement.](image)

<table>
<thead>
<tr>
<th>Table 3.1: 1DOF System Parameters</th>
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<tbody>
<tr>
<td>Mass</td>
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<td>90 g</td>
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3.3.2 Position Control

Reference trajectory tracking performance was analyzed to assess system behavior and validity of motor characterization. Feedback of the load displacement was used when tracking a reference sine wave, since this would emulate a robotic therapy scenario in which the user is carried through a sinusoidal trajectory. In this mode, the position control error was fed back as a desired signal to the inner velocity control loop of the motor, using a simple nested proportional control scheme. Position control was evaluated under the condition of no user interacting with the device (i.e. free motion).

The position control scheme was to apply a proportional gain from the error in load position. Before applying the gain, the error signal was passed through a 0.05 mm deadzone in order to prevent a switching mode of the motor. This mode can result from noise in position measurement that fluctuate around zero. Since the USM cannot regulate low velocities (as seen in Fig. 3.6), the result is a constant switching between the clockwise and counter clockwise smallest velocities. This switching can result in large vibrations in the system from the sudden changes in velocity (28 rpm peak-to-peak). After the deadzone, the proportional gain converted the position error into a desired velocity of the USM. Finally, this signal was converted into a voltage command to the USM’s velocity control box. A block diagram of the position controller is shown in Fig. 3.17.

The resulting position profile in response to a 10 mm amplitude and 0.5 Hz sine wave can be seen in Fig. 3.18. The device is able to track the input with a maximum error less than 0.5 mm, but is unable to perfectly track the desired position profile, due to the fact that the system is underdamped with very low friction in the ceramic ball bearings. If the proportional gain is increased (which would decrease position error), then the large velocity errors when changing direction (amplified by the low-
Figure 3.17: Block diagram of closed-loop position control on the load. The control is closed-loop since the load position is fed into the controller through measurement with the load’s optical encoder. The measured error is passed through a deadzone to avoid large changes in motor velocity. A proportional constant converts this error into a desired motor velocity which is converted to a voltage applied to the USM’s velocity controller.

Figure 3.18: Closed-loop control of the load’s position. The desired position profile is a sinusoid, with 10 mm amplitude, and 0.5 Hz frequency.

velocity nonlinearity of the motor) would cause the system to vibrate at the system natural frequency. Thus, the ability to track certain reference trajectories is limited by non-linearities of the USM.
3.4 Interaction Controllers

3.4.1 Rendering Virtual Springs

The capability of the device to render springs of different stiffnesses was evaluated. All experiments were performed under a constant gain scenario and in the condition of no user interacting with the load. The gain was tuned so that the system would be stable under free motion and for the stiffest desired virtual spring. Additionally, gains were chosen such that if the system was released from equilibrium, it would not become unstable. This is the most conservative method of choosing the proportional gain since, in general, the device will be connected to the user who will have a stabilizing effect on the system [55, 56]. These studies have shown that the human hand can be modeled as a mass-spring-damper, and thus is passive. The damping from the user acts to damp out the oscillations from the underdamped 1DOF system. Also, the gains can be changed depending on the desired virtual spring (generally higher allowable gain for lower spring constant), but gain variations are not explored in the scope of this thesis.

Fig. 3.20 shows the resulting loading curves for a variety of virtual springs under static loading with a proportional gain ($k_p = 75$ RPM/N). The system is able to render stiffnesses much lower and even stiffnesses slightly larger than that of the physical stiffness. The capability to render these virtual springs was also determined under dynamic conditions. Dynamic perturbations were applied manually by the experimenter who applied forces at the location of the ATI force sensor. The resulting Bode plot can be seen in Fig. 3.21. The SEA is able to accurately render all haptic environments up to approximately 3 Hz. For the higher stiffness values, the system could keep realistic haptic rendering until even greater frequencies.
Figure 3.19: A desired zero position for the spring is given ($X_{L,d}$) which results in a position when the user interacts with the device. The desired output force is found by multiplying this force by the physical spring stiffness. This force is compared to the estimate load force and is acted upon by a proportional controller after being passed through a deadzone. Finally, this signal is sent as a velocity command to the USM's controller. This block diagram representation of rendering a virtual spring through series elasticity has been adapted from [46].

Figure 3.20: Force vs. load position of the co-located device for rendering various virtual springs under static loading conditions. A constant proportional gain of 75 RPM/N was used in all experiments. The gain was chosen such that stability was achieved even when rendering the highest virtual stiffness. Note that the system is able to stably render stiffness greater than that of the physical system.

The use of linear techniques to characterize the system’s ability to display a virtual spring is validated by the coherence of the model. The Bode plot obtained in Fig. 3.21 has good coherence (> 0.9) in the range of 0.5-10 Hz (see Fig. 3.22). One possible explanation for regions of lower coherence is that the same gain that guaranteed
Figure 3.21: Bode plot of force/position for the system rendering various virtual springs. As can be seen up until 3 Hz for low virtual stiffnesses and 8 Hz for larger ones, the system is able to accurately render the desired impedance. After this region, the system no longer has the bandwidth to render these springs and so instead the user feels the physical stiffness of the device. Note how the phase goes negative for stiffness larger than that of the physical system. This is a result of the system losing passivity for these gains, but the system was stable for the described conditions.

stability for rendering stiff springs was used for lower stiffness virtual springs. Another explanation might be that the user applying the dynamic loads may have not excited all frequencies. By applying the Laplace integral operator \( \frac{1}{s} \) on the Bode plot in Fig. 3.21, the impedance of each case can also be observed.

### 3.4.2 Zero Force Control

The final interaction control scenario presented in this thesis is a zero force controller. This employs the same interaction control scheme as in the previous section (Section 3.4.1), where the value of the desired virtual stiffness is zero (Fig. 3.24). In this way, the motor tries to get out of the user’s way when the system measures a force. The amount the motor moves is a function of the proportional gain applied to
Figure 3.22: Coherence of the model used to obtain the Bode plot of Force over position. In general quite good coherence was observed ($> 0.9$) except in rendering the spring constant of $0.25k_s$. This could a result of many factors, but one likely is the result of using a constant gain for all experiments. Using a larger gain for this stiffness could improve rendering a more accurate virtual spring under dynamic loading conditions.

Figure 3.23: Impedance of the device when rendering various virtual springs. The plot shows how much force is applied to a given velocity. Note that after about 4 Hz, all impedances approach the same value.
Figure 3.24: Control block diagram of the zero force controller using series elastic actuation with the USM. The desired interaction force is zero, and so any interaction force is a force error to the system. Interaction force is estimated as the difference of motor and load position multiplied by the spring constant. After being passed through a deadzone and multiplied by a proportion gain, it is sent as a velocity command to the USM.

The force error. While in general increasing the gain increases transparency, increasing the gain too much will cause the system to become unstable.

Using the same gain as in the virtual spring experiments \((k_p = 75 \text{ RPM/N})\), the response of the system to a dynamic interaction can be seen in Fig. 3.25. These plots show the position, velocity, and interaction force in the time domain. The impedance rendered by the device is seen by plotting the interaction force vs. load velocity. For very small constant velocities \((< 10 \text{ mm/s})\), the device acts as a zero impedance device, but after this region, the device behaves like a damper with a damping constant of \(30 \frac{Nm}{rad/s}\). With the maximum velocity of 100 mm/s, a force of 3 N is required to back drive the robot. This force is somewhat significant (15% of the maximum interaction force), but still shows that the device is backdrivable with a zero force controller.
Figure 3.25: Zero force control of the MR-compatible SEA. The plot shows in the time domain the resulting force needed to move the platform for a given position profile.

Figure 3.26: Estimated force plotted against load velocity during zero force control. As can be seen, there exists a region at low-velocities where the device acts as a zero impedance device (no force to an applied velocity). Outside of these regions the device acts as a damper with a value of $30 \frac{Nmrad}{s}$.

3.5 Summary

In this chapter, it was shown that the 1DOF MR-compatible SEA could implement the controllers necessary to implement general robotic protocols. First, a closed-loop
position controller was presented that could be used during the robot-in-charge mode. Rendering of virtual springs was shown to be stable for a wide range of stiffness, indicating the potential to render different impedances to the user. This interaction control combined with position control could be used for the challenge-based protocols where virtual springs help the user achieve a desired position trajectory. Finally, zero force control was implemented and showed that low force was required to backdrive the robot, which would be used during the patient-in-charge mode. Note that the interaction control experiments had a bandwidth of about 3 Hz, which is sufficient for the proposed use of the robot. If more bandwidth were desired, one solution could be to increase the physical stiffness of the springs [48]; however, this would result in less compliance and masking of non-linear motor effects.

Additionally, it should be noted that with a passive user attached to the load, better system performance can in general be achieved through increasing the proportional gain for the position control and interaction control experiments. All gains were tuned for the case of interacting with a massless end effector (position control) or being released when given a large displacement (virtual spring). These cases are known to be the most destabilizing interaction modes for series elastic actuators. When a user interacts with a device, the effect is usual to add stable damping to the system. Through this damping, the small vibrations in the springs in the massless cases, which can result in instability, can be damped by the user. This allows for increasing the proportional gains to increase, thereby increasing performance of the device.
Chapter 4

Conclusions

This thesis describes the implementation of interaction controllers on two custom integrated force measuring mechanisms and discusses the applicability of these controllers for the field of rehabilitation robotics. The first device presented was the RiceWrist-Grip which uses six force sensing resistors to measure forces on the handle of a wrist rehabilitation robot. Preliminary testing showed that the force sensing resistors could regulate interactions successfully using force-feedback controllers. By specifying a simple proportional-integral force feedback controller with a null desired force, an experiment using FSRs as the force sensor found that user effort was decreased 6.1 times when moving a 1DOF linear platform. Additionally, in the 0-15 N range, error in force measurement of the FSR can be kept to 1.5 N, thereby allowing for accurate force measurement. A second experimental condition where two FSRs were used on the same 1DOF platform demonstrated the efficacy of implementing zero force control during pinch interactions. Accuracy of grip force measurements with the RiceWrist-Grip was evaluated in a case pilot study of incomplete spinal cord injury. Post-grip strength measurement using the RiceWrist-Grip showed good correlation ($R^2 = 0.84$) with a commercial hand dynamometer. Measuring grip force with the RiceWrist-Grip in place of the hand dynamometer offers the advantage of an integrative solution as well as the ability to measure interaction forces in real-time during therapy.

The characterization and interaction control design for a previously developed
linear 1DOF MR-compatible series elastic actuator was also presented. A series of experiments characterized the relationship between commanded voltage and motor speed for the non-magnetic ultrasonic motor. The results showed that the motor’s velocity had a linear region as well as a low deadband and high saturation region to commanded voltage. Experiments connecting the USM to a viscous damper and to a Maxon RE40 DC motor evaluated the effect of load disturbances on the USM’s built-in velocity controller. The velocity controller was able to reject all load disturbances from the viscous damper. Even at large step torques, the system would re-achieve the desired velocity in less than 0.3 seconds. Some loading effect was found from a sinusoidal load, but the loading was insignificant. For a 3 Hz maximum interaction force (20 N), the change in set-point velocity was less than 2%, indicating that interaction forces expected during protocols would have a negligible effect on the motor’s velocity controller.

In addition to characterizing the USM’s performance, the linear platform’s system parameters were identified. System identification revealed low damping and negligible friction as well as high force measuring accuracy through the use of the extension springs in a series elastic configuration. Two interaction controllers implemented on the device tested its ability to render different impedances to the user. The first was the rendering of various virtual stiffnesses. It was shown that the device can render virtual stiffness both lower and higher than the physical stiffness of the system within a bandwidth suitable for fMRI protocols. It should be noted that at stiffnesses higher than the physical stiffness, the system was no longer passive; however, the system was still able to render these stiffnesses in a stable manner. A zero force controller was also implemented, which is equivalent to the virtual spring case with the desired virtual spring equal to zero, which tested the ability to render a transparent environment. In
the zero force mode, the user was required to exert less than 3 N for the maximum expected velocity of 100 mm/s.

In conclusion, this thesis examined two force measuring schemes for implementing interaction controllers. Both were found to be suitable for implementing zero force controllers, which could be used to improve the patient-in-charge mode of therapy when the rehabilitation robot is either non-backdrivable or has significant dynamics. Future work should examine the effectiveness of the grip force sensor in being used as a force measuring device to implement zero force control with the RiceWrist or RiceWrist-S. Using this force measurement could improve transparency and enable a closed-loop constrained mode for improved disturbance rejection over open-loop strategies. Further work with the 1DOF MR-compatible system includes fabrication of an MR-compatible version of the RiceWrist. The control testing with the 1DOF platform will be used to implement therapy-like protocols with the MR-compatible RiceWrist, thereby enabling examination of brain adaption to different interaction control protocols during continuous fMRI.
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