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Series Elastic Actuation: Facilitating Robotic Assessment of Human Neurological and Biomechanical Properties

by

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ABSTRACT

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This thesis proposes integrating series elastic actuators into the design of robots that facilitate assessment of human neurological function and biomechanical properties. Such assessments can provide objective measures of the physiological adaptations which promote recovery after neurological injury, such as stroke or spinal cord injury. These adaptations are currently not well-understood, leading to variable recovery despite intensive rehabilitation. The robot-aided assessments reported in this thesis can be used to identify the neurological catalysts for brain plasticity and the biomechanical factors for everyday function essential for motor recovery. Robots currently used for these assessments do not measure torque and so use backdrivable (i.e., low friction) actuators since nonbackdrivable (i.e., high friction) actuators cannot be used for accurate torque control. However, accurate torque control with nonbackdrivable actuators can be achieved through series elastic actuation, a design in which an elastic element is purposefully incorporated in series between the actuator and user. To overcome limitations found in existing backdrivable robots, this thesis develops two novel series elastic actuated assessment robots with nonbackdrivable actuators for applications in robot-aided assessment. In the first application, brain activity is acquired through functional magnetic resonance imaging during physical human-robot interaction with the Magnetic Resonance Soft Wrist (MR-SoftWrist) robot. The MR-
SoftWrist features series elastic actuation to overcome limitations in using non-ferrous and nonbackdrivable ultrasonic motors. As demonstrated in this thesis, the MR-SoftWrist can interact with the wrist safely and accurately during functional magnetic resonance imaging, while not degrading acquired images of brain activity. In the second application, passive stiffness and active range of motion wrist envelopes are assessed with the novel Series Elastic Assessment Wrist (SE-AssessWrist) exoskeleton. The SE-AssessWrist employs rotary series elastic actuators to provide accurate torque estimation and zero force control, despite using nonbackdrivable actuators consisting of geared motors and a flexible Bowden cable transmission. The SE-AssessWrist can facilitate measurement of wrist biomechanics, in particular determination of the axis of least stiffness and greatest range of motion, which does not coincide with anatomical axes. As an important first step towards improved robotic rehabilitation, the proposed robot-aided assessments in this thesis are validated in able-bodied human experiments.
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<td>ADL</td>
<td>Activities of Daily Living</td>
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<tr>
<td>BL</td>
<td>Baseline</td>
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<td>CNC</td>
<td>Computer Numerical Control</td>
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<td>DAQ</td>
<td>Data Acquisition</td>
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<td>DC</td>
<td>Direct Current</td>
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<td>DOF</td>
<td>Degree of Freedom</td>
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<td>DTM</td>
<td>Dart Thrower’s Motion</td>
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<tr>
<td>EA</td>
<td>Error Augmentation</td>
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<td>ECR</td>
<td>Extensor Carpi Radialis</td>
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<tr>
<td>ECU</td>
<td>Extensor Carpi Ulnaris</td>
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<td>Flexor Carpi Radialis</td>
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<td>FE</td>
<td>Flexion Extension</td>
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<td>FK</td>
<td>Forward Kinematics</td>
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<td>fMRI</td>
<td>functional Magnetic Resonance Imaging</td>
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<td>IK</td>
<td>Inverse Kinematics</td>
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<td>JS</td>
<td>Joint Space</td>
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<td>MR</td>
<td>Magnetic Resonance</td>
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<td>MVC</td>
<td>Maximum Voluntary Contraction</td>
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<td>MVT</td>
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<td>PC</td>
<td>Path Control</td>
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<td>pHRI</td>
<td>physical Human-Robot Interaction</td>
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<td>ROM</td>
<td>Range of Motion</td>
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<td>RUD</td>
<td>Radial/Ulnar Deviation</td>
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<tr>
<td>SEA</td>
<td>Series Elastic Actuator</td>
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<tr>
<td>sEMG</td>
<td>surface Electromyography</td>
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<tr>
<td>TDPA</td>
<td>Time Domain Passivity Approach</td>
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<td>tNSR</td>
<td>temporal Noise-to-Signal Ratio</td>
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<td>TS</td>
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<td>USM</td>
<td>Ultrasonic Motor</td>
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<td>VC</td>
<td>Visual Control</td>
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<td>WEDM</td>
<td>Wire Electrical Discharge Machining</td>
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<tr>
<td>ZF</td>
<td>Zero Force</td>
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To my wife Deanna, my mother Doreen, my father Daniel, 
my sister Danielle, and my brother Matthew
Chapter 1

Introduction

This thesis demonstrates that the adoption of a series elastic actuation scheme enables the application of robotic devices to new domains of neurological and biomechanical assessment. A series elastic actuator (SEA) is a design in which a compliant element is intentionally placed in series between the robot’s actuators and the user [1]. SEAs provide a combination of accurate closed-loop torque control and compliant physical human-robot interaction (pHRI) despite non-backdrivable actuators and transmission schemes. This thesis presents the first instance of using SEAs specifically for robot-aided assessment.

- In the application of neurological assessment, SEAs are employed for accurate force control to facilitate high-fidelity haptic interaction while assessing cortical activation maps during functional magnetic resonance imaging.

- In the application of biomechanical assessment, SEAs are employed for direct torque output and sufficient robot range of motion to assess passive wrist stiffness and active wrist range of motion envelopes with the same device.

The design and characterization of two novel robotic devices are presented to demonstrate the value of the series elastic actuation scheme for robot-aided assessment (see Fig. 1.1). To demonstrate the possibility of using these devices to study human physiological properties, the robot-aided assessments are validated with able-bodied human participants.

In the remainder of the Introduction, Section 1.1 details the motivation for the two assessment robots, Section 1.2 describes the contributions of the thesis, and Section 1.3
outlines the remainder of the thesis.

1.1 Motivation

1.1.1 Robot-Aided Assessment for Improving Robot-Aided Neurorehabilitation

Robotic rehabilitation is a new and burgeoning field with over 75,000 articles published, about 7.5% of the number of articles published in the field of neurological rehabilitation. Neurological rehabilitation seeks to restore lost limb function to those who have been affected by neurological injury, such as stroke or spinal cord injury, so they can independently perform everyday tasks. Rehabilitation robots promise to be one of the most important tools for clinicians in facilitating meaningful restoration of functional capabilities after neurological injury [2–6]. Rehabilitation robots are aptly suited to perform the intensive [7] and interactive [8,9] exercise therapy necessary for recovery after neurological injury since through the use of position encoders and programmable actuators they can perform any
number of rehabilitation training modalities in an accurate and repeatable manner [10–14]. Since the 1990s, researchers have studied the applicability and efficacy of using robotic devices to deliver this rehabilitation. However, results of clinical trials with rehabilitation robots and neurologically impaired patients, often referred to as neurorehabilitation, have been mixed in terms of functional improvement [15–17]. Furthermore, these trials have not demonstrated statistically significant differences in the gains of standard clinical outcome measures for those who perform rehabilitation with robots compared to those who undergo conventional physical therapy [18–22].

While robots have not demonstrated gains over conventional therapy, these early clinical trials validated the capability to perform therapy in a methodical and interactive manner through rehabilitation robots [12, 15]. The trials also showed that positive functional improvements can be obtained during robotic neurorehabilitation [2], although how to maximize these improvements is currently unclear. Most believe that practice is an important component in obtaining functional improvements during neurorehabilitation [23]. Combining repetition with skilled motor learning in a functional task is required for functional reorganization [24], as evidenced in animal studies which demonstrated that if the task was meaningful, greater functional reorganization was achieved [25]. By training patients to learn new motor skills, intact portions of the brain can be recruited to perform the roles previously performed by the motor cortex, a process referred to as brain plasticity [26]; however, the best method for creating this brain plasticity is currently unknown. To improve neurorehabilitation a greater understanding of the recovery processes which best promote patient-specific neuroplasticity, and thereby recovery, is needed [3, 10, 12, 15]. Patient-specific rehabilitation, such as targeted brain plasticity, has thus far not been possible in the neurorehabilitation process [27]. Robot-aided assessments offer the potential to achieve patient-specific assessments which could facilitate patient-specific rehabilitation [28, 29].
The first exploration of robot-aided assessment during robotic neurorehabilitation was the examination of pointing movements [30], a discrete movement between a defined starting and ending point, as has been extensively studied in able-bodied individuals in the neuroscience literature [31–33]. Velocity profiles of pointing movements in able-bodied individuals are very smooth, resembling a bell-shaped curve, with a single peak [34]. Pointing movements made by neurologically impaired patients can be less smooth, with oscillations and an undefined peak [35, 36]. During robotic rehabilitation with the MIT-Manus, Krebs et al. found that over the course of robotic therapy, movements made by the participants became smoother throughout the protocol, possibly indicating the regaining of lost motor control capabilities [13]. In a study by Celik et al., the authors found that in fact smoothness metrics describing pointing movements correlated well with scores on the standard clinical practice Fugl-Meyer Assessment scale [37], indicating that assessments made by robots, which are more rich in information and more objective, could provide indications of recovery throughout the rehabilitation process.

Currently, only human-administered assessments, such as the prominent Fugl-Meyer Assessment, are accepted in clinical practice [38, 39]; however, in addition to being labor intensive, the Fugl-Meyer Assessment scale is subjective, being graded on an ordinal scale [38]. In contrast, robot-aided assessment offers the possibility for objective and automated assessments [13]. Assessment robots might be used to improve our understanding of motor control and brain plasticity in the able-bodied and impaired populations. This information might then be used to incorporate patient-specific data for optimized therapy [28, 29], perhaps by at regular intervals using the assessment information to update the robot interactions. Additionally, providing insights through robotic assessment might lead to the evidence-based recovery necessary for external funding, leading to prolonged and more effective rehabilitation trials. Possible relevant proposed factors for patient-specific
assessments include neurological [40, 41] and biomechanical [42, 43] assessments. The following sub-section highlights the importance of these two assessments and how current technologies are insufficient for performing them.

1.1.2 Limitations in Robot-Aided Neurological and Biomechanical Assessments

Neurological Assessment

It is thought, though not well understood, that the most desirable outcome for rehabilitation is the promotion of brain plasticity for the recovery of motor function. Brain plasticity is a process in which new areas of the brain are employed to cover tasks that are not disconnected as a result of injury [44]. Several approaches to studying brain activity, and thereby brain plasticity, are possible, although perhaps the most prominent in the neuroscience literature has been functional magnetic resonance imaging (fMRI) [45–47]. fMRI uses the blood oxygen level dependent contrast to provide an indirect measure of brain activity [48], and is attractive due to its non-invasiveness and high-spatial resolution. For example, experiments with fMRI have attempted to derive the areas of the brain responsible for vision [49] and motor control [50] as a result of a stimulus. Thus far, kinematic stimuli have been executed through movements which are not measured or controlled. Combining fMRI with interactive robots could lead to identifying specific areas of the brain responsible for carrying out motor control experiments [51], which were previously only possible to perform in laboratory environments [52]. Performing these types of experiments and analysis with the neurologically impaired population, could enable assessment of how rehabilitation changes cortical activation maps, possibly leading to brain plasticity [53].

Designing a device to be used during fMRI requires special consideration due to the large magnetic field necessary for the high temporal-spatial resolution provided by fMRI. Most traditional robots use direct current (DC) motors which contain ferrous materials.
Ferrous materials are dangerous in proximity of the MRI scanner since they are magnetically attracted to each other. If not grounded properly, the robot will be pulled towards the scanner, potentially injuring the person being scanned [51]. In addition, moving parts which are magnetically susceptible, such as any metal, can create eddy currents. These currents if created in proximity to the region of interest where the brain is scanned, lead to artifacts in brain activation maps, potentially leading to false-positives in assessing brain function. Other possibilities for introducing artifacts in the acquired brain activation maps include moving limb mass in proximity of the region of interest, and head movements, which are usually the result of large movements during a task [54, 55].

While many devices have been proposed for use during fMRI [56–58] (see Fig. 1.2 for an example), the devices mainly attempt to leverage existing rigid cable-based robotic design and control strategies used in the laboratory environment. This approach results in the following limitations in an MR-compatible version of these robots:

- Insufficient position measurement accuracy
- Inadequate force control
- Limited force control bandwidth
- Potential danger to the user.

As a result of these limitations, a device which is not significantly limited by the constraints imposed from existing devices is needed. This thesis presents the design, characterization, and validation of an MR-compatible robot for high-fidelity haptic interaction during fMRI, using SEAs to overcome limitations found in previous works.
Biomechanical Assessment

Anatomical joint impedance (i.e., stiffness and damping) is an important characteristic relating to functionality of our joints. Just like mechanical systems, the joints of the human body can be modeled as mass-spring-damper systems [60]. After neurological injury, these impedances often change, resulting in unwanted increased muscle tone [61–63]. This increased stiffness or tone can cause decreased range of motion which is important for performing activities of daily living (ADLs). Unlike mechanical systems, human joint stiffness and damping are non-linear, and active, being modulated through muscular contractions [64–66]. Thus, to have a reference which can be easily compared across populations, joint impedance values are obtained while ensuring the user is passive [42]. In addition to use in rehabilitation, knowledge of these passive joint impedance values enables understanding how our joints produce torque and motion for biomechanical models and can be used for biologically inspired prosthetic designs.

Stiffness of the wrist is especially important for rehabilitation, biomechanical modeling,
and biologically inspired designs, since stiffness dominates wrist impedance [60], and the stiffness and range of motion of the wrist can be used to describe the plane of the dart thrower’s motion [67], an important biomechanical characteristic unique to the wrist [68]. When considering a 2D case, springs form stiffness ellipses, which in linear systems are described by smooth shapes. In the case of human wrist stiffness ellipses, they describe the path of least resistance, which is often referred to as the plane for the dart thrower’s motion [68]. This dart thrower’s motion is important for performing many ADLs, and as a result return of wrist motor control is a high-priority for many neurologically impaired individuals [69]. Wrist rehabilitation is also appealing since studies have observed improvements in proximal joints during distal wrist training [70], although the training of proximal joints has not been observed to improve distal joints [18].

Passive stiffness of the wrist has previously been studied through robotic assessment with able-bodied participants [42, 66] (see Fig. 1.2), while another study with the ankle validated the possibility for these kinds of assessments with neurologically impaired individuals [43]. The devices used for wrist assessment were designed for robotic rehabilitation, and as such were not designed specifically for biomechanical assessment. As a result, they have limited range of motion and torque. Wrist rehabilitation robots [14, 71] also do not measure torque and instead estimate it through motor current commands. Stiffness is estimated through knowledge of interaction torque and displacement. Not measuring torque will lead to inaccuracies in joint stiffness estimation. Additionally, assessment of range of motion with wrist rehabilitation devices is only possible for moderate range of motion [70, 72].

To summarize, the rehabilitation devices currently proposed for stiffness assessment [66] have the following limitations:

- Insufficient torque output
• Insufficient range of motion

• Indirect torque estimation.

These limitations reveal that current rehabilitation robots are inadequate for use in thoroughly and accurately assessing biomechanical properties to understanding recovery from neurorehabilitation. Therefore, this thesis proposes a solution to overcome these limitations which leverages SEAs to create a device with both direct torque measurement and complete range of motion, while also increasing torque output.

1.1.3 Series Elastic Actuators for Robot-Aided Assessment

This thesis proposes using robot-aided assessments with series elastic actuated robots during pHRI. Robots built for pHRI can assess user kinematics and kinetics accurately and repeatably through position and torque sensors [13, 73]. Prominent robots for pHRI, such as robotic haptic interfaces [74], only measure, and therefore enable direct assessment of, user kinematics. Haptic interfaces traditionally use linear torque output actuators, such as DC motors, and backdrivable transmissions, for example pulleys with a rigid cable transmission [75]. A backdrivable transmission is a transmission which can be moved by the user easily, enabling measurement of user movements without powering the device. In contrast, a non-backdrivable transmission requires excessive force to move, usually as a result of high reflected friction and possibly inertia due to motor gearing or the transmission. Backdrivable transmissions lend themselves well to impedance control algorithms [76], in which user motion is input and torque is output. As a result, the control loop is closed around position measurements for accurate position control, while torque commands are performed in an open-loop manner leading to possible inaccuracies as a result of unmodeled device dynamics between the robot and end-effector, such as friction, which are always
present in a haptic interface to some degree. Note that non-backdrivable transmissions require the use of admittance control, the inverse of impedance control, in which a torque sensor measures user effort and then the device is commanded to move the user [77]. This type of control in a rigid system can lead to contact instabilities due to sensor noise and non-collocation of the actuator and measurement [78, 79].

Haptic robot design and control methodologies have found applications in robotic rehabilitation trials [80] and neuroscience experiments [52]. In the robotic assessments thus far [13, 36, 57, 66, 70], existing devices originally designed with the goal of robotic rehabilitation were used (see Fig. 1.2 for representative proposed assessment robots). Rehabilitation robots have been designed with the objective of providing moderate torque and range of motion capabilities [81, 82] to meet the requirement of training for the required torques and range of motion found in most activities of daily living (ADLs) [72]. Rehabilitation robots produce interaction torques through precise actuators, typically DC motors [13, 14, 43, 71, 72, 81–83], with backlash free and backdrivable transmissions, and high resolution position encoders all for the dual purpose of rehabilitation and assessment of patient’s motions [14]. Rehabilitation robots offer an excellent platform for programmable interactive rehabilitation in a clinical setting for automated therapy; however, although using existing rehabilitation robots for assessing movements is justifiable given that robots are designed with direct drive and backdrivable transmissions with zero backlash for transparent behavior when unpowered, the other fields of assessment explored in the literature – neurological [40] and biomechanical [66, 70] – all have limitations as a result of using these existing devices.

In contrast with haptic design methodologies, SEAs provide direct torque estimation, allowing for accurate closed-loop torque control [1]. Torque measurement is provided by measuring deflection of the elastic element, and the added compliance has additional
benefits, such as allowing for the use of high gains for accurate torque trajectory following and the low-pass filtering shock loads. Additionally, having a compliant element located between the actuator and the user enables the use of non-backdrivable elements, such as geared motors with backlash and friction, while achieving this accurate torque control [1, 84]. This thesis provides the first instance of developing series elastic actuated robots for robot-aided assessment, where SEAs provided the actuation architecture necessary for neurological and biomechanical assessments.

1.2 Contributions

This section describes the contributions of this thesis. Portions of this work are represented through the following publications.

1.2.1 Control of Series Elastic Actuators

• Contributed ideas and data analysis in developing a novel time domain control approach for SEAs which enables rendering a stiffness higher than that of the physical spring while maintaining passivity – published in the *IEEE/ASME Transactions on Mechatronics* [85]

1.2.2 Robot-Aided Neurological Assessment: Development and Validation of the MR-SoftWrist for Haptic Interaction during fMRI

• Performed experiments, developed hardware, and ran experiments for a linear MR-compatible SEA – published in the *IEEE/ASME Transactions on Mechatronics* [86]

• Developed and presented preliminary control experiments with a novel three degree of freedom MR-compatible robot, the MR-SoftWrist, which enables assessment of
brain activity during fMRI – published in the proceedings of the *IEEE International Conference on Rehabilitation Robotics* [87]

- Validated the MR-SoftWrist as a tool for facilitating measurement of brain activity via extensive control characterization and a rendition on a classic pHRI experiment performed during fMRI – published in the *IEEE Transactions on Neural Systems and Rehabilitation Engineering* [88]

### 1.2.3 Robot-Aided Biomechanical Assessment: Development and Validation of the SE-AssessWrist for Evaluating Wrist Stiffness and Range of Motion

- Presented the design concept for the SE-AssessWrist, including presentation of a rapid prototyped single-degree-of-freedom module for wrist flexion/extension to validate the concept – poster presentation at the IEEE International Symposium on Wearable & Rehabilitation Robotics [89]

- Presented a Bowden cable-based series elastic one-degree-of-freedom wrist flexion/extension module, highlighting the series elastic element, preliminary control experiments, and a case study – submitted to the ASME Dynamic Systems and Control Conference [90]

- Carried out a study evaluating active range of motion and passive stiffness of the wrist with five able-bodied participants to demonstrate the functionality of the SE-AssessWrist – in preparation for submission to a journal publication

### 1.3 Thesis Organization

This thesis is organized as follows: Chapter 2 presents background literature on the applications, design, and control of SEAs, concluding with one of the contributions of this
thesis: the development of a time-domain approach to control SEAs. Chapter 3 describes the application of neurological assessment through the development of the MR-SoftWrist which uses SEAs to measure and assist wrist movements during fMRI. Chapter 4 details the development of the SE-AssessWrist for biomechanical assessment using SEAs for accurate torque measurement and complete range of motion. Chapters 3 and 4 detail the design methodology, control implementation, performance characterization, and a validation study which illustrate the devices being used for the intended application. Chapter 5 summarizes the benefits of the series elastic actuation design approach for the two robot-aided assessment applications presented in this thesis, and describes the future outlook for the research.
Chapter 2

Haptic Interfaces and Series Elastic Actuators

This thesis demonstrates two applications of robot-aided assessment through the use of two novel robots, the MR-SoftWrist and the SE-AssessWrist. These robots use series elastic actuators (SEAs) to enable accurate torque control and estimation despite the use of non-backdrivable actuators. Details of these robot-aided assessments are discussed in Chapters 3 and 4. This chapter provides a literature review on SEAs, giving the context for why SEAs were developed, and how they expand design possibilities over traditional rigid haptic interfaces.

2.1 Rigid Robotic Haptic Interfaces

The study of haptics in the robotics context began in the early 1990s [74]. Haptics, i.e., the sense of touch, is important to roboticists since robots can be used to display real-world environments virtually through force-reflecting robots. This emulation typically involves a visual display, such as a computer monitor, which displays the user’s virtual position, possibly through a cursor. A virtual environment consisting of objects of various sizes, shapes, and stiffness can be included and as the user’s cursor makes contact with these virtual objects, the robot can display forces to the user to render the material properties in a realistic manner. These types of interfaces have found application in teleoperation, for example during minimally invasive surgery [91], where a leader robot is operated by the user and a remote follower robot is controlled to replicate the leader’s position trajectory.
In this way the operator can control the follower robot over great distances while obtaining force feedback regarding interactions through the leader robot. Haptic interfaces have also been used extensively to further our understanding of the human motor control system through human perception and sensorimotor control experiments [52, 92, 93].

### 2.1.1 Impedance-Controlled Haptic Interfaces

Haptic interfaces are typically controlled through impedance control, an approach founded by Neville Hogan in 1984 [76, 94]. Impedance controllers regulate the flow of energy at the point of interaction between the robot and user. In impedance control, user flow (velocity), typically acquired through the differentiation of high resolution position encoders, is accepted by the robot and in response to this motion the robot outputs a programmable force. A common example of impedance control occurs when the robot is programmed to display a rigid virtual wall. This can be accomplished by making the actuator output a force proportional to user position and velocity measurements to emulate a spring and damper system when making contact with the virtual wall. In the early 1990s, Edward Colgate studied the topic of how to render these virtual walls and the limitations in displaying them. He stated that at low forces, inherent robot dynamics (friction, inertia, damping, and stiffness) dominate the flow of interaction forces between the user and robot, while at high forces the limitations of the actuator and control hardware, which sample and acquire the position and force data, limit performance [95, 96].

Impedance control closes the control loop around position measurement, and force is output in an open-loop manner based on the actuator’s predetermined input-output relationship, typically given by the manufacturer. As such, any unmodeled device dynamics, such as damping, mass, inertia, and friction, will result in a force error, resulting in loss of haptic fidelity, i.e., the rendered object will not feel as realistic. Physical damping is typically
only present in the actuators and is often negligible, and sufficient stiffness can be achieved by using metallic links, although bearings and other joints might introduce some level of play in the system. Additionally, gravitational loading due to device mass of the links can typically be mostly compensated with a model of the gravitational loading, although it is still beneficial for inertial effects to reduce device mass. Therefore, friction and inertia are typically of primary concern during the design process. Although friction and inertia can be compensated through control, this is often not done since friction is a highly non-linear phenomenon [97, 98] and inertia requires knowledge of the measured position’s second derivative [99, 100], which is not easily obtained without introducing significant time delay and amplifying noise. Thus, to maximize performance, haptic devices are designed to minimize perceived device dynamics, especially friction and inertia, by the user.

As a result of low inherent dynamics and fast dynamic response, high-fidelity haptic impedance-controlled interfaces can display environments from free space to rigid walls. Note that these goals are competing in that displaying a rigid wall requires a powerful actuator which in turn increases resistance during free space motion. The perfect haptic interface would be able to render completely free environment (no impedance) when unpowered while providing infinite force bandwidth, the ability to track a force command with a given accuracy at maximum frequency. While possible in theory, these goals are ultimately not realizable due to imperfections such as device dynamics, sensor noise, and time delays in the control hardware [74].

In addition to high-fidelity haptic interfaces needing low inherent dynamics, these devices also use rigid linear force amplifiers and smooth (i.e., low friction with no backlash) transmissions for accurate open-loop force control. This approach was first carried out in the foundational three degree of freedom (DOF) Phantom haptic device created by Thomas Massie and Kenneth Salisbury in the 1990s (see Fig. 2.1) [73]. The Phantom uses direct
current (DC) motors in conjunction with pulley transmissions which transmit force through rigid cables. Since DC motors are often low-torque and high speed, torque amplification is required and is accomplished through pulleys. These pulleys were reduced to capstan arcs to save weight and space. Although devices such as the Phantom are designed with minimal friction, any brushed DC motor will have some friction due to brushes, and a cable transmission creates additional friction, as described by the capstan equation, introducing force command error. This error has been quantified to be up to 40%, and often above 10% in standard dynamic interactions with a Phantom 1.5 device, as estimated from Fig. 7 of [101]. While not a concern for many haptics studies since position measurements are mostly used for analysis [52, 102], studies which use this force estimate should consider the limitations of such hardware.
Rehabilitation Robots

Although several variations of the Phantom device now exist, such as the Phantom Omni [75] or the Novint Falcon [104], perhaps the greatest extension of these devices has been to the field of rehabilitation robotics (see Fig. 2.1) as started in the early 1990s [5]. Rehabilitation robots were created to aid physical therapists in performing the high-intensity and interactive therapy necessary for recovery [5, 10, 12, 13, 105]. In the early 2000s, the number of rehabilitation devices exponentially grew, and is still growing to this day [6, 71, 83, 105, 106]. Almost all preliminary and subsequent studies use impedance-controlled devices that are made to be backdrivable with low effort. Rehabilitation robots are designed with the paradigm that they must be able to interact with the user in three ways: human-in-charge, robot-in-charge, and programmable human-robot challenged-based modes [107]. With respect to operating the robot, these three modalities translate to a transparent mode, position control, and interaction control. Impedance-based rehabilitation robots can accomplish these goals proficiently as demonstrated in many clinical trials [19, 20, 108]. In addition to rehabilitation, these robots were initially envisioned as dual rehabilitation and assessment devices [13, 109]. Robots certainly offer accurate and repeatable assessment during training with the device; however, new assessments such as facilitating neurological assessment during functional magnetic resonance imaging and biomechanical assessment of stiffness have been proposed [42, 57] which suffer performance limitations as a result of these design methodologies.

2.1.2 Admittance-Controlled Devices

While impedance control has been the most popular approach for interaction control in robotic haptic interfaces, other control approaches exist, such as admittance control. To implement interaction control through admittance control, the corollary to impedance con-
control, the controller accepts a force as input and outputs a position through its actuators [77]. Any number of actuators and transmissions can now be used in the design process since the device does not need to be backdrivable when unpowered. In admittance control a force/torque sensor measures the user’s intent and can be accomplished by incorporating a six axis force/torque sensor at the end-effector [79, 110] or single axis force sensors at each joint of the robot [111–113]. Admittance control is useful for high force and large range of motion applications as demonstrated in the Haptic Master [114], an admittance-controlled device with significantly more force output and workspace volume, i.e., the achievable positions of the end-effector of the robot based on position limitations of the joints, than the Phantom has [115]. This device uses a strain gauge sensor at the end-effector to compensate for the device’s friction and inertia, that are caused by leadscrew spindle transmissions and heavy links.

Although often used in industrial applications [111, 112], the use of leadscrews and gears as transmissions have largely been avoided in haptic interfaces. Geared motors are beneficial in industrial applications, as they eliminate the need for cable tensioning and can have a high torque density as a result of significant gear reductions in a small volume compared with capstan torque amplification [115]. Additionally, many industrial applications only require rigid position control with force sensing to guide the robot along some path, for example as in arc welding [116]; however, using gears results in variable friction, backlash, and noise being introduced into the transmission. Humans can perceive backlash readily, which is in large part why gear trains have been avoided in haptic interfaces [115]. Additionally, if the gearbox has a large enough ratio it becomes non-backdrivable due to perceived static friction and inertia being amplified from the motor to gearbox output, eliminating the possibility for traditional impedance control. Due to rigidity in the force sensor and flexibility between the sensor and actuator, contact instabilities can arise from measure-
ment of impact forces and non-collocation of the sensor and actuators leading to unstable force control [78, 94, 117]. Therefore, gains must be carefully selected with admittance controllers when attempting to render low impedance environments, which is the opposite dilemma as in impedance control.

While traditional impedance-controlled and admittance-controlled devices have been the primary mechanisms for interaction control in haptic interfaces, each suffer limitations at the extreme. In impedance-controlled devices, rendering low impedance environments is simple as the device can be backdriven when unpowered. While the interface can also provide high bandwidth interactions, these occur at low forces, with limited range of motion, and possibly loss of force accuracy. On the other hand, admittance-controlled device are difficult to render low impedance environments, but they can perform rigid position control with great accuracy. Additionally, using force sensors for force control can reduce the perceived dynamics significantly, although force control still suffers from non-colocation of the actuator and rigid force sensor. Offering a solution between the two approaches, series elastic actuators have been created for accurate interaction control despite non-backdrivable actuators and transmissions.

2.2 Series Elastic Actuators

Although the industrial robots of the past could use simple position controllers [118], because robots now interact with people more closely, more sophisticated and safe controllers are necessary [119]. Additionally, impedance-controlled devices are too restrictive for design selection, and implementing interaction control with non-backdrivable admittance-controlled devices can be challenging. As a result, SEAs have arisen, a design in which an elastic element is intentionally incorporated between the user and robot (see Fig. 2.2). This design was formally introduced by Gill Pratt and Matthew Williamson in 1995 for the
Figure 2.2: Linear plant model of the SEA: torque ($\tau$), position ($\theta$), inertia ($J$), damping constant ($B$), and spring rate ($K$). Note that in some models the damping term is neglected as in some cases it can be considered negligible compared with the inertia of the motor. Subscript “M” refers to motor while “L” refers to load.

benefits created by intrinsic compliance: added shock tolerance, possible energy storage, and improved force control accuracy and stability [1]. By introducing elasticity between the actuator and load, a low-pass filter is made such that when the robot is unpowered, the user will feel the stiffness of the spring instead of the actuator. In addition to offering human-friendly interaction, introducing an elastic element can decouple actuator dynamics at high frequencies, so that instead the user feels the spring, and at low frequencies direct, compliant, and accurate force control can be achieved. As a result, actuators and transmissions traditionally not used for rigid haptic interfaces can now be used for accurate interaction control.

There is extensive literature on the control of flexible joint manipulators where the flexibility is not intentionally incorporated into the robot and is a result of elasticity in the transmissions, possibly due to gears, cables, bearings, or robot links. Although SEAs were introduced in 1995, the related field of flexible joint control has been around since the 1980s. Flexible joint manipulator control is interested in improving force control performance and stability by modeling the undesirable joint elasticity. Flexibility in the joints can cause unwanted position error due to elastic deformation and chatter through resonance with the controller [120–122]. In contrast, SEAs intentionally incorporate elasticity, ideally
only in the direction or axis of actuated motion, for improved force control performance. Since the elasticity is known and the position of the element is measured on both sides, the degrading effects from unwanted joint elasticity literature are eliminated. Interestingly some of the main benefits of SEAs were discovered in 1985 by Roberts et al., where the authors determined that the lower the stiffness of the force sensor, in this case a strain gauge on a cantilevered beam, the higher the proportional gain could be used to increase the accuracy of force control while correspondingly lowering position control accuracy [79]. Interestingly, the possible application of this work to interactive robotic devices was not discovered until 10 years later.

2.2.1 Applications of SEAs

Although introduced in 1995, only recently have the merits of SEAs been explored for haptic interfaces. The first prominent application of SEAs was the development of the 8DOF LOPES exoskeleton for gait rehabilitation [107, 123, 124]. The LOPES exoskeleton leveraged SEAs to locate bulky geared DC motors off the main exoskeleton frame through a Bowden cable transmission. A Bowden cable transmission, which is commonly found in bicycle brake and derailleur systems, consists of a cable routed through a flexible conduit. In this way the cable can be flexibly routed from one end to the other, in the case of the LOPES between the off-board motors and the on-board series elastic joints. Although Bowden cables introduce significant force transmission problems, such as friction and hysteresis, the series elastic actuated joints of the LOPES enabled overcoming these limitations. As a result, the LOPES could lower its intrinsic inertia compared with the Lokomat, a rigid exoskeleton for gait rehabilitation, while still achieving accurate interaction control [19].

Since the development of LOPES, several other applications of SEAs have been pre-
sent. Series elastic joints have been incorporated into two of NASA’s humanoid robots: Robonaut [125] and Valkyrie [126], as well as into the Baxter robot [127] developed by Rethink Robotics for industrial applications. All of these robots can interact with user in close proximity in a safe manner as a result of SEAs. Another exciting application of SEAs is a modular snake robot which takes advantage of rubber as the elastic element for implementing torque control despite friction and for added compliance in the presence of backlash [128, 129]. Recently, SEAs have gained traction in the field of robotic rehabilitation with exoskeletons for the knee [130], hand [131], and shoulder and elbow [132], as well as for robotic assistance during locomotion in the elderly population [133] and a shoulder and elbow orthosis [134]. Although many exciting applications for SEAs have been put forth, using SEAs for neurological assessment was not considered until 2016 [135, 136], well after the work related to the MR-SoftWrist was published [86–88, 137], and has still not been validated in human trials. Additionally, leveraging SEAs for biomechanical assessment has still not been considered, and no wrist exoskeleton has been developed with series elastic actuated joints.

2.2.2 Spring Selection

A critical design decision for SEAs is selection of the elastic element. The choice of elastic element depends on the desired stiffness and space constraints of the application. Fortunately, the elastic element can be designed with a variety of form factors and customized stiffness, as demonstrated in the SEA literature (see Fig. 2.3). A simple implementation is the use of conventional linear or torsion elastic springs [88, 107, 131, 138, 139], although these springs are typically largest in the length dimension. Torsional springs can be created from linear springs in series [140, 141], but these result in errors due to the linear displacement approximation and can be difficult to assemble. Newly designed springs with
Figure 2.3: Various spring implementations and design for incorporation as elastic elements in SEAs. (left) Linear spring composed of a helical tension and compression spring [139], (center) torsion spring composed of six linear compression springs [141], and (right) custom torsion spring [142]. Images were obtained from cited sources and cropped for depiction in this thesis.

Archimedes spirals, possibly with variable width to optimize stress for maximum torque output, have been created through wire electrical discharge machining, although designing and manufacturing these springs is both more costly and time intensive than using conventional off-the-shelf springs. However, as opposed to traditional torsional springs which are longest in the length dimension, these springs are largest in the radial dimension, which can be beneficial for exoskeleton design as evidenced in their many applications [132, 142–145].

Spring rate is typically selected by performing a compromise of force control accuracy and resolution, as well as position control bandwidth. While increasing the stiffness of the spring improves position control accuracy, this comes at the cost of lowering the possible control loop gains during force control while maintaining stability. As a result, softer springs result in improved force control, but a decrease in position control accuracy. Note also that to achieve direct force control, measurement of the deflection of the spring is required. Further, the spring rate and resolution of the encoders determine the force resolution of the SEA, which directly impacts bandwidth and benefits for safety. Thus,
with sufficient sensor resolution, a relatively stiff spring might still be used for soft force control, although generally inexpensive encoders will result in the need for softer springs and therefore reduced position control accuracy. Thus, when creating a SEA, a trade-off between force resolution and position control bandwidth must be made.

2.2.3 Series Damping and Variable Impedance Actuators

With the invention of SEAs, more variations on incorporating passive, variable, and active elements into the design of haptic interfaces have arisen. Series damping, a newly presented alternative to SEAs, incorporates a damper in series between the actuator and user [146]. The damper can be made through a viscous or friction coupling, altering the unpowered perceived dynamics of the device [147]. This design concept has also been extended to variable dampers, in which the damping coefficient can be modulated in real time [148, 149]. Unlike elastic elements which are passive, damping elements are dissipative and can improve position control performance compared with SEAs. SEAs can introduce position oscillations due to the passive nature of the elastic elements and they can add energy to the system, but in many cases can also increase energy requirements as a result of the need to modulate the springs’ displacement to control force. On the other hand, series dampers add design complexity, change the unpowered perceived dynamics, and reduce safety during interaction control.

Several variations and combinations of adding springs and dampers to designs have been presented, from connecting the components in parallel and series, to making each element adjustable. An overview of approaches such as these are presented in a review by Vanderborgh et al. [147]. Making the elements variable has appeal in that the stiffness and damping can be modulated for a given situation to optimize device transparency and dynamic force control; however, the ability to vary these component’s properties comes at
the cost of added time-delays, design complexity, and bulk to the system since adjustability typically corresponds to the need for an additional actuator. As such, the application and the benefits of adding dampers or the capability to vary elements must be carefully considered for each device. Due to the limitations currently found in variable stiffness actuators, in this thesis non-variable SEAs are employed for their more favorable form factor, as well as the many benefits they provide, e.g., direct torque estimation, accurate force control with nonlinear force transducers, and compliant behavior when unpowered.

2.2.4 Control of SEAs

The main objective in an interaction control loop with an SEA is to regulate the deflection of the spring to control the interaction torque ($\tau_L$) between the user and the robot since $\tau_L = k(\theta_M - \theta_L)$. Since the robot has control over the actuator’s position and the user has control over the other end of the spring, the torque control problem becomes a position control problem in which the position of the spring on the motor’s side ($\theta_M$) is controlled in response to user motion ($\theta_L$). In this way direct torque control is achieved by regulating the deflection of the spring through $\theta_M$. Many control approaches have been suggested to regulate $\theta_m$ in response to a user’s motion. First approaches attempted to control $\tau_L$ through specifying a desired $\tau_{M,d}$ with open-loop current control [1]. This approach results in limitations since SEAs often use geared motors and other transmission elements which introduce static friction, creating error in regulating $\tau_M$. Other methods of achieving closed-loop torque control exist such as using an inner position or velocity loop to implicitly modulate torque [150]. To improve on these force control approaches, for a specific case, the now ubiquitously utilized [84, 86, 144] cascaded torque control for SEAs was proposed [84, 140, 151] (see Fig. 2.4),

In cascaded torque control, the outermost control loop specifies $\theta_L$, while an inner loop
controls $\tau_L$, and a further nested inner loop regulates $\dot{\theta}_M$. Controlling motor velocity is easily implementable and robust to low frequency force disturbances, such as static friction, and high frequency disturbances are filtered out by the spring. Linear theory analysis is readily applied to this control architecture, a topic thoroughly explored by Vallery et al. for the case of rendering a pure spring [84]. Vallery found that to passively render a pure spring, the outer impedance control loop cannot render a spring stiffer than the physical spring. The analysis by Vallery et al. was extended by Tagliamonte et al. to include virtual damping, which can extend the range of allowable stiffnesses [152]. Passivity was used as the control criteria in these studies, as in most robotic controllers intended for human-robot interaction, since while a stable system can become unstable when interacting with a passive environment, two passive systems interacting will remain stable. Since humans are generally modeled as passive systems, using a passivity-based approach to control provides more robust stability criteria than traditional isolated stability analysis, although this robustness comes at the cost of decreased control performance.

While cascaded torque control is sufficient for many applications, various control approaches have been presented to improve interaction control accuracy and bandwidth. In work by Kong et al., a disturbance observer (DOB) based approach was recommended.

Figure 2.4: Block diagram of the cascaded torque control scheme presented in [84]. In this case an outer impedance loop regulates a virtual spring by measuring and specifying load position. The load torque is also measured, and an inner force control loop modulates the desired interaction torque, while at the motor level a nested inner velocity loop controls the motor as a velocity source for robustness and to overcome limitations from friction. The block diagram was adapted from [84] and [86].
for SEAs in which the DOB views $\theta_L$ as a time-varying disturbance. In this approach, various feed-forward terms were presented to compensate for actuator and transmission nonlinearities. This approach led to significant improvements in low-impedance control and in tracking force profiles [153]. Further control approaches include a more simplified model-based DOB has been presented in [154], a sliding mode controller [155], and an $\mathcal{H}_\infty$ controller in [156]. While the aforementioned control approaches exceed cascaded torque control in performance, this increased performance comes at the cost of knowing system parameters accurately and increased control complexity, leading to possible instabilities as a result of errors in model estimates. This also means that each system must be extensively characterized to obtain an accurate system model. A more in-depth survey of controllers for compliant manipulators can be found in a review article by Calanca et al. [157].

It is interesting to observe that interaction control of a SEA can be viewed as a hybrid impedance-admittance controller. At the highest level of a SEA interaction controller, the controller accepts a position input ($\theta_L$) and outputs a torque ($\tau_L$). However, unlike the case of open-loop torque control with current controlled DC motors, $\tau_L$ is regulated through $\theta_M$. Thus, SEAs can be viewed as having an inner admittance control loop wrapped by an impedance controller to perform accurate, programmable interaction control. In addition to force control, SEAs might also need to be used for pure position control, which can be achieved through closed loop position control of $\theta_L$ without considering the effects of $\tau_L$. Introducing an elastic element, while beneficial for interaction control, degrades position control since now position control attempts to regulate $X_L$ through $x_M$, but the two are connected through an oscillatory spring which introduces unwanted oscillations. This is why position controlled devices attempt to make the robot as stiff as possible, but since position control performance is not as essential in SEA applications, as it is on a closed-environment industrial floor, this loss in performance is acceptable [1, 158]. Note that both
position and force control can be significantly improved when a user is coupled to the robot, since the most destabilizing case occurs when the load is a pure inertia, and a user typically introduces damping [86, 158–160].

**A Time-Domain Passivity Approach to Control**

In seminal work by Vallery et al., analysis in the frequency domain states that for the actuator to remain passive, the rendered stiffness of a pure virtual spring cannot exceed the physical spring stiffness [84]. In contrast, one of the contributions of this thesis to the field of SEAs is in the development of a time domain passivity (TDPA)-based approach for SEAs [85]. Time-domain approaches to control were originally discussed in the application of teleoperation by Hannaford and Ryu for rigid haptic interfaces [161]. This same TDPA was extended to applications with SEAs by Losey, Erwin, and colleagues [85]. By monitoring and ensuring that the virtual energy does not exceed the physical energy, i.e., $E(t) \geq 0 \forall t$ where

$$E(t) = E_p(t) - E_v(t),$$  \hspace{1cm} (2.1)

passivity is guaranteed. Note that the physical energy is given by

$$E_p(t) = \int_0^t \tau'_L(\lambda)(-\dot{\theta}_L(\lambda))d\lambda$$  \hspace{1cm} (2.2)

and the virtual energy is given by

$$E_v(t) = \int_0^t \tau'_{L,d}(\lambda)(-\dot{\theta}_L(\lambda))d\lambda$$  \hspace{1cm} (2.3)

By monitoring the physical and virtual energy of each system, when $E_v$ is lower than $E_p$ by a given threshold, the physical stiffness can be exceeded virtually, but below a given threshold, the controller must dissipate energy. Unlike passivity-based approaches in the frequency domain, the TDPA can passively render virtual stiffness higher than that of the
physical spring, increasing the potential applications of the actuator. Being able to passively render a stiffness higher than that of the physical stiffness, allows for using lower stiffness springs for improved force control, while still providing more rigid haptic interactions. Validation of this control approach is demonstrated in Fig. 2.5.

In addition to the TDPA to control, the paper presents a model reference adaptive controller. The model requires no knowledge of system parameters and drives the system to behave according to desired closed-loop torque control characteristics. The model reference adaptive controller was shown to converge quickly to the system parameters and to perform more robustly than the disturbance observer approach presented in [153]. This approach to control overcomes previous limitations in model-based SEA controllers since it does not require knowledge of system parameters and even permits time varying parame-
ters [85]. Although beyond the scope of this thesis, these control approaches could be used for the series elastic actuated device presented in Chapter 3 to improve its dynamic range and to perform accurate control without accurate knowledge of system parameters.

2.3 Summary

In this chapter the trade-offs in interaction control performance for rigid impedance and admittance-controlled devices were discussed. The concept of SEAs was presented including their viability as a compromise between traditional impedance- or admittance-based haptic interfaces. The elastic element in a SEA functions as a low-pass filter between the actuator and user, masking nonlinear actuator and transmission phenomena, as well as providing shock tolerance to the gearbox and enabling direct torque control. In this way SEAs have the benefits of admittance-controlled devices in that geared motors and other nonlinear force transmission schemes can be used, but SEAs have much more accurate interaction control as a result of the compliance that increases force control performance. The ubiquitous frequency-based approach to SEA control was discussed, and an alternative time-based approach was presented. Unlike the frequency domain approach which limits rendering stiffness up to that of the physical spring, the time domain approach enables rendering higher stiffnesses for short periods of time. With this time domain approach to control, force resolution can be increased by using a softer spring while still being able to render stiff environments. In the subsequent chapters, the benefits of SEAs in robot-aided assessment will be presented in the development of the MR-SoftWrist and the SE-AssessWrist for neurological and biomechanical assessments.
Chapter 3

Robot-Aided Neurological Assessment: Development and Validation of the MR-SoftWrist for Haptic Interaction during fMRI

The MR-SoftWrist is a novel three-degree-of-freedom series elastic actuated magnetic resonance compatible robot. The MR-SoftWrist was created with the objective of providing accurate multi-degree-of-freedom kinesthetic feedback during functional magnetic resonance imaging. To achieve this high-fidelity haptic interaction, and while using non-ferrous materials, the MR-SoftWrist uses commercially available non-magnetic ultrasonic motors. However, these motors are non-backdrivable and have a low-velocity nonlinearity. Thus, to achieve accurate interaction control, series elastic actuation are incorporated into the device’s architecture.

In this chapter, design decisions regarding structural parameters of the MR-SoftWrist are detailed. Characterization and control experiments are also presented, illustrating the device exceeds capabilities of previous MR-compatible devices as a result of the combination of the series elastic actuated ultrasonic motors. Finally, a single participant case study in which the MR-SoftWrist provided kinesthetic feedback during functional magnetic resonance imaging, is presented. As hypothesized, the study revealed distinguishable brain activity in response to three interaction tasks regulated by the MR-SoftWrist.

Portions of this chapter were published in [87, 88], the preliminary work for this chapter were published in [86, 137], and further demonstration of the MR-compatibility of the device was published in [162]. I gratefully acknowledge the contributions of my collaborators.
in these publications.

3.1 Introduction

Rehabilitation robots have demonstrated their effectiveness in improving functional outcomes of patients with neurological disorders; however, the underlying mechanisms that promote recovery from a neurological injury are still not well understood [12, 20, 163]. As a result, optimal therapeutic regimens remain to be found in neurorehabilitation [164]. One approach to understanding the recovery process at the neurological level is the use of functional magnetic resonance imaging (fMRI). Since the advent of functional magnetic resonance imaging (fMRI) in the early 1990s, neuroscientists have been able to non-invasively study neural activity of the entire brain with high spatial resolution [48], starting with tasks such as recognition and memory [49], and in surgical robotics applications [165]. Shortly after, fMRI was also applied to the study of neural activity during human motor control. In one such study, researchers examined the difference in neural activity during rhythmic and discrete movements through one degree of freedom (DOF) wrist flexion/extension (FE) movements [166], while another study used finger tapping or wrist flexion to examine differences in brain activity of patients with chronic stroke pre- and post-therapy [41]. More recently, researchers performed the first multi-DOF visually guided task with the wrist during fMRI to study motor acuity during a semicircular arc pointing task [167]. Among the potential applications for fMRI in the study of motor control, using fMRI to study neuroplastic changes induced by motor neurorehabilitation may be one of the areas to benefit the most [168]. To date, studies examining cortical reorganization over the course of therapy have only been able to use passive devices to measure fMRI pre- and post-therapy [169, 170].

While previous fMRI motor control studies yielded results that expanded our under-
standing of the neural correlates of motor skill learning, they still lacked a critical component – the controlled combination of kinematic measurement and kinesthetic feedback. Since it is known that variation in amplitude and velocity of movements can modulate brain activation patterns [47], it is important to have knowledge of limb kinematics to account for confounding factors when analyzing brain activity. Additionally, kinesthetic feedback is vital for exploring the world around us, and is especially important in robot-aided neurorehabilitation, a promising approach to motor rehabilitation [20]. Investigation of sensorimotor learning under kinesthetic feedback via fMRI requires an MR-compatible haptic device to apply force fields, while accurately measuring movement kinematics. Researchers have already demonstrated viability in using haptic robots to study motor control outside of an MRI scanner [171], paving the way for the field of haptic fMRI, which studies neural activity during physical human-robot interactions. With the aid of a robot that can display arbitrary dynamics to the user, rehabilitation-like protocols can be performed during fMRI to determine which best promote recovery [45]. Using a haptic interface is necessary in this endeavor for its force control, backdrivability, repeatability, accuracy, and precision [172].

3.1.1 MR-Compatible Robots

Combining haptic robots with fMRI requires special considerations. The principle of fMRI is based on measuring the blood oxygen level dependent signal [49], which is accomplished by measuring inhomogeneities of the MR machine’s large static magnetic field (up to 9.4 T for human studies), and requires switching magnetic field gradients and radio frequency (RF) pulses. Any robot used during fMRI must be MR-safe (i.e., it does not pose any threat to the participant) and MR-compatible (i.e., it will not interfere with the quality of scanned images or have its own performance affected by the fMRI process) [172]. To be MR-safe, no ferrous components should be used and conductive loops that couple to the switching
gradients or to the RF pulses should be avoided. Such loops can cause heating and burn the skin of the participant. In contrast, being MR-compatible is more complicated. Some of the factors that play a role in MR-compatibility are the amount of metallic and electrically active components, especially those near the region of interest, i.e., the brain, or those that move through magnetic field lines. Moving electrically active or metallic components can produce electromagnetic interference [51]. The robot is not the only factor potentially involved in image degradation. The task itself should be carefully selected, since the quality of fMRI images is substantially degraded by head movements and displacement of body mass in proximity of the region of interest [54].

Distal arm movements are more suited to fMRI studies, since it has been shown that shoulder and elbow movements result in much larger head motions than those involving the wrist [55]. Despite this fact, studies have mainly focused on sweeping arm movements [51, 173] which may be a result of the challenges to create a multi-DOF haptic device targeting the wrist. Also complicating matters, participants typically lie supine on the scanner bed inside a 60-70 cm scanner bore, making it challenging to fit both the robot and person inside the same small area. Thus far, researchers have developed two strategies to address the constraints imposed by the MR environment [137]. The first is a remote actuation scheme where non-compatible actuators and materials are located at a great distance (>1 m) from the user and are connected to the end effector through hoses when using fluidic actuation or long rod or cable transmissions when using electromechanical actuation [51, 174, 175]. The alternative solution is to co-locate the actuators with the end effector, requiring all materials and actuators to be MR-compatible [135, 136, 176].

Although single-DOF haptic interfaces [135, 176, 177] are attractive because of their simplicity and capability to study isolated joint movements, multi-DOF haptic interfaces offer a broader range of possibilities for neuroscience purposes, studying, for example,
adaptation to lateral force fields [171], or redundant tasks [178]. However, the design of multi-DOF MR-compatible robots with accurate kinesthetic feedback capabilities has so far proved challenging. Previous approaches to multi-DOF MR-compatible haptic devices have included a 2DOF hydraulic system [56], a 2DOF pneumatic system [173], a 3DOF system using long rods connected to a Phantom Premium [174], a 3DOF system using shielded electromagnetic motors [57], and a proposed design for a 6DOF device using ultrasonic motors (USMs) [179]. All of these devices are limited in terms of position measuring accuracy since they do not measure joint movement directly. Additionally, the use of hydraulic actuation [56] or long carbon fiber rods to connect the device with the user [57] limits force bandwidth. Finally, the latter two functional devices are limited by the fact that they are intended for use with unconstrained arm movements, an intrinsically non MR-compatible task.

To address these constraints, this thesis develops an MR-compatible wrist robot, the MR-SoftWrist [87], equipped with MR-compatible compliant USM actuation for accurate force feedback [86]. The MR-SoftWrist is the first MR-compatible multi-DOF haptic wrist robot, and is capable of measuring and supporting wrist motions during fMRI. The MR-SoftWrist is placed at the edge of the MR machine’s scanner bore and extends its links to a wrist ring end effector which connects the device to the user’s wrist for kinesthetic feedback.

This chapter presents the MR-SoftWrist [87, 88], a parallel 3DOF co-located haptic MR-compatible robot for wrist motor protocols during fMRI (Fig. 3.1 [88]). The device employs series elastic actuators (SEAs) to achieve interaction control despite the use of non-magnetic, non-backdrivable, and non-linear velocity sourced ultrasonic motors [86,137]. To demonstrate the importance of the SEAs, dynamic interaction control capabilities of the series elastic actuated MR-SoftWrist are thoroughly characterized. Finally,
the device is validated through a single participant case study in the fMRI environment [88].

3.2 MR-SoftWrist

In the following section, an overview of the robot, the MR-SoftWrist, used in this chapter is provided, describing its kinematic structure, components, and properties. The MR-SoftWrist interacts with a user’s wrist instead of more proximal limbs as done in [56, 57, 173], which should reduce head movements and moving body mass during scanning, thus improving fMRI image quality [54]. General design requirements for the manipulator are capability of position, low-impedance, and force-source control modes [81, 107]. In previous work [86], an actuation approach was proposed which uses USMs in a parallel kinematic architecture in conjunction with series elastic elements for accurate force control. It was determined that the achievable circular workspace for wrist FE and radial/ulnar deviation (RUD) is 20 deg, and that 1.5 N·m of continuous torque output can be provided [87,88].
3.2.1 Mechanical Design

A parallel design was pursued to achieve high structural rigidity, velocity, and torque output with low inertia as compared to an equivalent serial manipulator. Additionally, the parallel design places the actuators on a stationary base frame, reducing potential imaging artifacts from active electrical components moving in the scanner’s magnetic field. A three revolute-prismatic-spherical kinematic structure was chosen (see Fig. 3.2), which consists of a base ring and three legs, each of which include a revolute, prismatic (actuated), and spherical joint [180]. Each spherical joint is mounted equidistantly on a wrist ring, which serves as the robot’s end-effector. The mechanism provides 3DOFs corresponding to \( z_c \) (the platform height), \( \theta_{FE} \) (wrist FE, a rotation about \( \hat{x} \)), and \( \theta_{RUD} \) (wrist RUD, a rotation about \( \hat{y} \)) (see Fig. 3.1).

Although \( z_c \) is a DOF of the device, it is fixed through control to provide alignment to the participant’s wrist anatomical axes. As a consequence, when coupled to a participant, the robot task space effectively reduces to the 2D space defined by \( \theta_{FE} \) and \( \theta_{RUD} \). With a suitable value for \( z_c \), this solution places the actuators at the edge of the scanner bore, which is beneficial for the device’s performance and for avoiding interference with scanned images. The device is therefore in its nominal configuration when \( \theta_{FE} = \theta_{RUD} = 0 \) and \( z_c = z_{c,\text{nom}} \), which corresponds to the extensible links \( l_i \) being equal to \( l_{i,\text{nom}} \) [87, 88].

A design was pursued such that the user’s hand moves in the space between the base and wrist rings (Fig. 3.1(right)). This solution places the actuators far from the scanner bore, which is beneficial for the device’s performance and for avoiding interference with scanned images. Additionally, fixing the forearm during operation allows for measurement and support of wrist FE and RUD. A drawback of this solution is that it requires large values for the wrist ring radius \( (r) \) and base ring radius \( (R) \) to fit the user’s hand inside the device, since if \( r \) is not large enough, the extensible links \( (l_i) \) will make contact with the
Figure 3.2: Kinematic schematic of the MR-SoftWrist with relevant parameters labeled [87, 88]. A stationary base with radius $R$ is connected to a moving platform with radius $r$ through extensible links with length $l_i$, corresponding to a platform height $z_c$. This figure is adapted from [81, 180].

A relevant measure of the robot workspace size is the radius of the circle inscribed within admissible solutions in the 2D task space (for $\theta_{FE}$ and $\theta_{RUD}$) defined by the selection of a specific $z_{c, \text{nom}}$ value. Such a measure is largely determined by $r$, since the workspace radius increases as $\frac{1}{r}$. To maximize the device’s ROM then, $r$ should be minimized. The value of $R$ has a smaller impact on workspace, but must be chosen to ensure the user does not contact the links while not being larger than the scanner bore. The distance between the two rings ($z_{c, \text{nom}}$) must be large enough for the hand to fit between them. As the distance is increased, the actuators can be placed further from the scanner isocenter; however, this increase comes at the cost of reduced workspace size and structural stiffness. Thus, parameters $r$, $R$, and $z_{c, \text{nom}}$ were carefully selected to guarantee a large workspace size while avoiding interference between the participant’s hand and the moving links, as
Table 3.1: MR-SoftWrist Structural Parameters [87, 88]

<table>
<thead>
<tr>
<th>$R$ [mm]</th>
<th>$r$ [mm]</th>
<th>$z_{c,\text{nom}}$ [mm]</th>
<th>$l_{i,\text{nom}}$ [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>168</td>
<td>67</td>
<td>255</td>
<td>274</td>
</tr>
</tbody>
</table>

well as between the device and the scanner bore [87, 88].

The effect of $r$, $R$ and $z_{c,\text{nom}}$ on the workspace radius can be seen in Fig. 3.3. By carefully selecting the base platform ($R$) and end effector ($r$) radii, a 20 deg workspace radius was achieved such that the user’s hand would not make contact with the device within this workspace and that required each link to travel about 40 mm. To accommodate users of varying size, linear bearings with 75 mm of travel were selected to allow for the desired 20 deg workspace radius to be achieved for $z_{c,\text{nom}}$ +15 mm/-20 mm. The nominal link lengths ($l_i$) were selected such that the workspace requirements were met and the platform height could be adjusted as much as possible to accommodate users with different arm lengths. The chosen structural values were 168 mm for $R$, 67 mm for $r$, 255 mm for $z_{c,\text{nom}}$, and 274 mm for $l_{i,\text{nom}}$ (see Table 3.1 for a summary of the structural parameters [87,88]). Note that the resulting $l_{i,\text{nom}}$ values were achieved by shifting the encoder readings 11.43 mm (in the $\hat{z}$ direction) to account for the fact that due to the geometric shape of the device, equal displacements in opposite directions for the link lengths do not result in equal changes in linear platform height or wrist angles. Additionally, adjustability was built into the device through mechanical vertical (up to 63.5 mm) and angular (up to 30°) adjustments of the device for aligning the user’s wrist with the axes of the wrist ring [87, 88].

3.2.2 Theoretical Performance Characteristics

Theoretical performance characteristics with respect to the device’s operational 2DOF ($\theta_{FE}$ and $\theta_{RUD}$ since $z_c$ is fixed during operation) are obtained using the device’s nominal struc-
Figure 3.3: Variation in workspace radius (of a circle for $\theta_{FE}$ and $\theta_{RUD}$) dependent on $r$, $R$, and $z_{c,nom}$. To maximize this radius, $r$ was minimized while ensuring that the user’s hand would not contact the links while inside the workspace, $R$ was maximized before the device would contact the scanner, and $z_{c,nom}$ was chosen to place the device just at the edge of the scanner [87].

Structural parameters ($R = 168$ mm and $z_{c,nom} = 255$ mm). Although the results will change slightly depending on these structural values, the results below give a representative sample of how the device’s performance changes as the link lengths are varied throughout the device’s workspace, meaning $l_i$ are within the bearing limits. To obtain performance characteristics as the device configuration changes in the allowable workspace, the device Jacobian is required and is defined as

$$\dot{\mathbf{q}} = \mathbf{J}\dot{x}$$  \hspace{1cm} (3.1)

where $\dot{\mathbf{q}} = [\dot{l}_1 \dot{l}_2 \dot{l}_3]^T$ is a vector of joint space velocities, $\dot{x} = [\dot{z}_c \dot{\theta}_{FE} \dot{\theta}_{RUD}]^T$ is a vector of task space velocities, and $\mathbf{J}$ is the device Jacobian which is a 3x3 matrix [87].
Velocity and Torque Limits

To determine the maximum possible velocities $\dot{\theta}_{FE}$ and $\dot{\theta}_{RUD}$ for every configuration in the workspace, the inverse Jacobian was used to relate the maximum joint space velocities to the task space ones as

$$\dot{x} = |J^{-1}k\dot{q}_{max}| \forall q \subset W$$

(3.2)

where $W$ is the workspace of the device, $k = [k_1 \ k_2 \ k_3]^T$ is a vector of scalar values between $\pm 1$, and $\dot{q}_{max}$ is the maximum joint space velocity of 86 mm/s. The maximum task space velocities at each configuration are found through careful selection of the components of $k$. Using (3.2) leads to a system of three equations which can be solved by satisfying constraints that 1DOF be maximized while the other DOF and $z_c$ have zero velocity. For example, for the configuration $\theta_{FE} = \theta_{RUD} = 0$, $\dot{\theta}_{FE,max}$ is obtained with $k = [0 \ 1 \ -1]^T$ and $\dot{\theta}_{RUD,max}$ with $k = [1 \ -0.5 \ -0.5]^T$. Using this method in (3.2) for all possible configurations, the resulting maximum angular velocities are $\dot{\theta}_{FE} = 1.43$-2.05 rad/s and $\dot{\theta}_{RUD} = 1.24$-2.04 rad/s [87].

The maximum torque for every possible configuration was found in a similar way, except that the equation relating joint space forces to task space torques is given by

$$\mathbf{f}_E = |J^Tk_f_{L,max}| \forall q \subset W$$

(3.3)

where $f_{L,max}$ is the maximum linear task space force and $\mathbf{f}_E = [F_z \ \tau_{FE} \ \tau_{RUD}]^T$ are the task space forces and torques. The maximum load force is limited by the springs since their yield deflection occurs at 25 N. Note that this force on the USM only corresponds to 0.138 N·m, which is almost 4 times less than its maximum continuous torque of 0.5 N·m. The resulting maximum end-effector torques for $\theta_{FE}$ and $\theta_{RUD}$ are 1.94-2.83 N·m and 1.76-2.46 N·m respectively [87].
Spatial and Torque Resolution

Spatial resolution of the device was computed as

\[
\delta_x = \max \{|J^{-1}k\delta_q| \forall q \subset W\} \tag{3.4}
\]

where \(\delta_q\) is the joint space position resolution and \(\delta_x\) are the task space spatial resolutions. The joint space resolution is obtained from the resolution of the linear optical encoders, which have a resolution of 0.0127 mm of link travel using a quadrature reading. Applying this in (3.4) results in a maximum quantization of \(3 \times 10^{-4}\) rad for \(\theta_{FE}\) and \(\theta_{RUD}\) [87].

The torque resolution of the device is computed as

\[
\delta_{fe} = \max \{|J^T k\delta_{fe}| \forall q \subset W\} \tag{3.5}
\]

where \(\delta_{fe}\) and \(\delta_{fe}\) are the joint and task space force resolutions respectively. The USM has a 1000 cnt/rev encoder corresponding to 0.0086 mm of link travel, and with a spring constant of 3.8 N/mm, summing the resolutions of the USM and load encoders leads to a worst case resolution of 0.08 N. Applying this in (3.5) leads to a maximum quantization of 9 N·mm for \(\tau_{FE}\) and \(\tau_{RUD}\) which is 0.5% of the 1.75 N·m maximum interaction torque [87].

3.2.3 Mechanical Properties

For the torque output of the MR-SoftWrist, the limiting factor is the maximum force on the springs (25 N), resulting in a torque of 138 N·mm on the USM, almost four times less than its maximum continuous torque of 0.5 N·m. Depending on the configuration of the device, its maximum velocity and torque output varies. Through analysis described in Section 3.2.2, the maximum torque and velocity limits of the device were computed for \(z_{c,nom}\) while varying the task space coordinates within the device’s 20° circular workspace. The MR-SoftWrist’s specifications are presented in Table 3.2 [88], which includes the device’s
Table 3.2: MR-SoftWrist Specifications [88]

<table>
<thead>
<tr>
<th>Joint</th>
<th>MR-SoftWrist</th>
<th>ADL</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$k$</td>
<td>$\theta_{res}$</td>
</tr>
<tr>
<td>FE</td>
<td>370 [N·mm/deg]</td>
<td>0.02 [deg]</td>
</tr>
<tr>
<td>RUD</td>
<td>370 [N·mm/deg]</td>
<td>0.02 [deg]</td>
</tr>
</tbody>
</table>

stiffness coefficients and workspace radius, as well as its velocity and torque limits and respective resolutions. As a comparison, wrist workspace and torque values for activities of daily living (ADL) are given using values obtained from [72]. The task space stiffness of the MR-SoftWrist, $k$, is about 7 times higher than the highest stiffness of the human wrist for combined RUD and FE rotations, which is estimated in several studies such as in [66]. The task space stiffness was determined through the relationship

$$K_X = J^T K_L J$$

(3.6)

where $K_L$ is a diagonal matrix of linear spring stiffness values (3.8 N/mm) at the load side (joint space), $J$ is the device Jacobian, and $K_X$ is a diagonal matrix of the task space stiffness values. The linear spring elements have been selected by trading off position control bandwidth during interaction with a passive participant (which increases for increasing stiffness values) with force measurement resolution (which instead decreases for increasing stiffness values) [87, 88].
3.2.4 Components

The MR-SoftWrist is a co-located robot, requiring all components to be MR-compatible. Due to its high rigidity and light-weight properties, Delrin was selected for all structural components. Brass screws were selected for securing all components due to their low magnetic susceptibility [181] and high strength. The actuated prismatic joints are linear series elastic actuators [86] composed of a rotary USM (Shinsei Corp. USR60-E3NT with the D6060E/24V motor driver) with an 11.5 mm diameter pulley, secured to the motor shaft through a brass set screw, connected to custom brass extension springs (Spring Engineers of Houston LTD., 1.9 N/mm) through a nylon cable transmission (SpiderWire Stealth 0.4 mm diameter braided fishing line, 80 lb test). The two springs are connected at each end of a slider through nylon eyebolts which are also used for pre-tensioning the springs. The springs are placed in parallel to each other but in series between the USM and load. Load deflection is measured through a linear optical encoder (US Digital EM1-0-500-I, with 101.6 mm long encoder strips). The slider is supported by brass housed linear bearings with ceramic balls and titanium shafts (Del-Tron Precision Inc. S2-3-NMS-Brass). The actuated prismatic joints are supported by two ceramic radial ball bearings (Impact Bearing) which are mounted on a brass shoulder screw secured to the base. Since no commercial MR-compatible spherical joint exists, the spherical joints were approximated through a revolute-universal-revolute kinematic chain. Ceramic ball bearings were used for the revolute joints (Boca Bearings) while the universal joints (Ondrives size 6 plastic universal joint) are made of brass and Delrin. Using two radial bearings introduces redundancy, which does not affect kinematics [81], in the kinematic chain allowing for greater ROM since the universal joints are less restricted [87, 88].

In addition to its mechanical components, an important feature of the MR-SoftWrist is the shielding and filtering of its motor and encoder lines. The MR-SoftWrist is comprised
of several electrically active components – six optical encoders and three USMs (voltages on the order of 200 V at 40-45 kHz). Operation of such elements during imaging can introduce electromagnetic interference, if the signal is not filtered properly to a ground reference, such as the one provided by the scanner penetration panel. In an effort to reduce electromagnetic interference introduced by these active components, tripolar twisted-pair shielded cable with an additional outer shield was used for encoder lines, and the load optical encoder cases were wrapped with aluminum foil. The USMs were wrapped in aluminum foil along with their respective encoders. The shield of the load optical encoders was connected to their respective foil on one side and to the penetration panel on the other to ensure a low-impedance path to ground, attenuating the noise generated by active components through a Faraday cage. To ensure decoupling of the signal references and to avoid introducing noise in the scanner room coming from the unshielded control room, the motor signal and encoder lines were low-pass filtered using 1300 pF and 5600 pF capacitive filters respectively. The filter frames were grounded by the penetration panel [87, 88].

3.2.5 MR-Compatibility Experiment

An MR-compatibility experiment [87] was performed to determine if the SEA MR-SoftWrist can function inside the MRI room as well as not distort images obtained during fMRI. The first requirement of MR-compatibility, namely safety due to ferrous components, was not of concern since the device is made entirely of low susceptibility components. To determine if operating of the MR-SoftWrist interfered with acquired brain images during fMRI, an experiment was performed with the MR-SoftWrist in three different conditions:

- Baseline (BL): no robot
- In (IN): robot in the scan room, but not operating
• Movement (MVT): robot moving while scanning.

A measure of interference, the temporal noise-to-signal ratio (tNSR), was calculated in each condition and examined. Additionally, whether the scanner had any impact on the MR-SoftWrist was evaluated by examining error in position measurements from the MR-SoftWrist while operating inside the scanner room and within a lab environment. In each case, the MR-SoftWrist was found to pass the MR-compatibility tests, verifying its use during haptic fMRI. Details of the methodology and results for this section can be found in Appendix A [87].

3.3 Device Dynamic Characterization

A variety of experiments are presented which thoroughly characterize the dynamic performance of the MR-SoftWrist. The device’s control scheme is presented, along with experiments evaluating its position control and force measurement accuracy. These experiments warrant use of the device as a kinesthetic feedback machine, capable of rendering virtual environments such as a transparent mode, with null displayed force, to different values of virtual stiffness, to unstable diverging environments [88].

3.3.1 Kinesthetic Feedback Capabilities

The MR-SoftWrist can be controlled to render a variety of virtual environments through impedance control as shown in Fig. 3.4 [88], using a low-level cascaded force-velocity control scheme typical of series elastic actuators (SEAs) [84]. The implementation of the low level force controller, labeled as “JS SEA Force control” in Fig. 3.4, on the USMs is described in detail in [86, 88].

Control of the MR-SoftWrist is performed through real-time software in a Matlab-
Figure 3.4: Block diagram of task space impedance control, in this case the impedance is a virtual spring, for the MR-SoftWrist. Desired task space position \( (x_{\text{des}}) \), measured task space position \( (x_{\text{meas}}) \), task space position error \( (e) \), task space virtual spring stiffness \( (k_v) \), force error without considering gravitational loading \( (f_{\text{E,k}}) \), gravity compensation model shown in Fig. 3.7, task space forces and torques \( (f_E) \), Jacobian \( (J) \), joint space load forces \( (f_L) \), force control performed at the joint space level as in [86], measured joint space position \( (q_{\text{meas}}) \), and forward kinematics \( (FK(\cdot)) \) [88].

Simulink model communicating with Quanser’s Q8 USB board at a 1 kHz loop rate. The device Jacobian \( J \) is obtained through the formulation found in [182]. In [182] the Jacobian is given as \( \rho(q') \), a matrix relating task space to joint space velocities, where \( q' \) is a vector including all of the device’s generalized coordinates (12 for the MR-SoftWrist). A closed form of the inverse kinematics was obtained through a method similar to that presented in [180]. The forward kinematics (FK) of the device are acquired through the method discussed in [183]. This solution seeks to find a coordinate transformation \( q' = \sigma(q) \) between the independent and generalized coordinates. This is accomplished through an iterative solution which guarantees the existence of \( \sigma(q) \) and can be computed in real-time. The iterative solution is formulated by deriving the nine kinematic constraint equations of the MR-SoftWrist, with a vector equation (three scalar equations) derived for each of the robot legs, as is standard in parallel manipulators [88].
Figure 3.5: Block diagram of the task space position controller for the MR-SoftWrist. Desired task space position ($x_{\text{des}}$), measured task space position ($x_{\text{meas}}$), error ($e$), proportional gain ($k_p$), desired task space velocity ($\dot{x}_d$), Jacobian ($J$), desired joint space velocity ($\dot{q}_d$), USM velocity controller, measured joint space position ($q_{\text{meas}}$), and forward kinematics ($FK(\cdot)$) [87].

### 3.3.2 Position Control Accuracy

Although not a priority for control, position control of the MR-SoftWrist was evaluated to demonstrate that the device can accurately follow a position trajectory at a pace representative of movements during wrist tasks. Position control might be used for playing back user motion to account for any brain activity due to the movement themselves [88].

To illustrate the ability of the device to track reference trajectories, the device was commanded to track a 20° circle in 2D task space coordinates ($\theta_{FE}$ and $\theta_{RUD}$) through a sinusoidal reference oscillating at 0.4 Hz. Control of the device was accomplished through a simple task space position control scheme as shown in Fig. 3.5 [87]. The task space controller converts an error into a desired USM velocity, which is regulated with the USM’s factory tuned velocity controller [86]. The result of applying this control scheme to the circular task space reference can be seen in Fig. 3.6. This experiment shows that the device meets the desired workspace while achieving good position tracking capabilities, resulting in a root mean square error of 2.25 deg and a maximum error of 3.45 deg in tracking either DOF [88].
3.3.3 Gravity Compensation

The MR-SoftWrist is force controlled by regulating the deflection of its springs. Since incremental encoders are used, a reading of zero force is determined by the amount of deflection present on the springs upon system initialization. However, when the manipulator moves, gravitational loading from the links and wrist ring assembly deflect the device’s springs, resulting in force readings despite no user interaction forces. This gravitational loading effect must be estimated and accounted for to accurately measure interaction forces and display virtual impedances in the task space [88].

An experiment was performed by position controlling the device to follow a $15^\circ$ circular trajectory in 2D task space coordinates with platform heights of $z_{c,nom}$ and $z_{c,nom} \pm 5$ mm. The device started with $\theta_{FE} = 15$ degrees and $\theta_{RUD} = 0$. Once the device achieved steady state, the forces measured by the springs were recorded. Using a Fourier series expansion

\begin{align*}
\theta_{FE} [\text{deg}] & \quad [\text{Top}] \\
\theta_{RUD} [\text{deg}] & \quad [\text{Bottom}]
\end{align*}

Figure 3.6: Position control experiment validating the workspace of the MR-SoftWrist. (top) Validation that the MR-SoftWrist can perform a $20^\circ$ circle in 2D task space coordinates. (bottom) Accurate tracking by the MR-SoftWrist of a 0.4 Hz sine wave [88].
Figure 3.7: Task space gravitational loading force readings obtained from steady state measurements while the MR-SoftWrist’s end effector was at various locations on a 15° radius in the 2D task space coordinates, for three $z_c$ values. The solid lines represent the model used to fit the experimental data [88].

to the second term, including a linear contribution term dependent on $z_c$, models were obtained for all three DOFs. The resulting task space force values, with respect to the variable $\phi$, which is defined as $\phi = \text{atan2}(\theta_{RUD}, \theta_{FE})$, can be seen in Fig. 3.7 along with the fit obtained from the model obtained through a multiple linear regression. Similar experiments were performed for $\pm 15^\circ$ of FE and RUD separately, along with varying the platform height, which was used for implementing gravity compensation schemes for those DOF for the impedance control experiments in Sections 3.3.5 and 3.3.6 [88].

3.3.4 Force Measuring Accuracy

As the MR-SoftWrist acquires force measurements in joint space coordinates, the accuracy of task space interaction force/torque measurements can be compromised by non-ideal
transmission elements. To estimate the accuracy of such task space force measurements, a six axis force/torque transducer (ATI Nano17 SI-25-0.25) was used as a ground truth reading (see Fig. 3.8). The ATI Nano17 was mounted on a handle attached to the wrist ring. In this way, the loading path of forces and torques applied by a user would propagate through the ATI Nano17 before being transferred through the wrist ring, spherical joints, and then the springs. Measurements of the ATI Nano17 were transformed to task space coordinates through a force-torque transform based off of distance and angular offsets obtained from a CAD rendering of the device. The transform matrix, given by the relationship

$$\begin{bmatrix} \mathbf{x}_{\text{T}} \\ \mathbf{s}_{\text{R}} \end{bmatrix} = \begin{bmatrix} \mathbf{x}_{\text{R}} \\ 0 \\ \mathbf{x}_{\text{P}} \times \mathbf{s}_{\text{R}} \end{bmatrix}$$

(3.7)

where $X$ denotes the end effector frame and $S$ denotes the sensor frame. Torques in end effector frame were computed through the equation

$$\mathbf{x}_{\text{T}} \mathbf{F} = \mathbf{s}_{\text{T}} \mathbf{F}_{\text{S}}$$

(3.8)

where $\mathbf{F}$ is a 6x1 vector of forces/torques. These measurements were compared with task space force measurements of the MR-SoftWrist by multiplying joint space force readings (from the linear springs) by the transpose of the device Jacobian [88].

To estimate the force measuring accuracy of the device in its neutral configuration, a human experimenter applied quasi-static loads (see Fig. 3.9) on the handle by grasping it and pushing/pulling in various directions with roughly sinusoidal profiles. In this experiment, the robot was blocked, so the springs are expected to measure the force applied to the MR-SoftWrist handle. Forces up to 8 N were applied in the $\hat{z}$ DOF ($F_z$) and torques of up to 460 N·mm and 630 N·mm were applied to the FE ($\tau_{FE}$) and RUD ($\tau_{RUD}$) DOFs respectively. The comparison of force measurements over a 10 s time period can be seen in Fig.
Figure 3.8 : CAD Rendering of the MR-SoftWrist with the ATI Nano17 force sensor. 1) Handle attachment, 2) ATI Nano17, 3) end-effector attachment.

3.10 which resulted in mean errors of 0.3 N, 17 N·mm, and 34 N·mm for $F_z$, $\tau_{FE}$, and $\tau_{RUD}$ respectively. Given the accuracy in force measurements, reported torque values in future experiments will be those derived from spring estimates. A possible explanation for the small error in force measurement is the compliance of the plastic structural components, as well as the play and friction in the universal joints [88].

### 3.3.5 Zero Force Control

Interaction control of the device was realized with the task space impedance control scheme shown in Fig. 3.4. To implement zero force control, the virtual spring constant ($k_v$) was set to the null vector, resulting in the desired task space torques being equal to zero. The low-level force controller presented in [86] was implemented on each actuation module. Such a cascaded force-velocity series elastic actuator controller contains a proportional gain for the outer force loop that determines the control effort based on force measurement
error. When a user attempts to make transparent movements during zero force control, they feel a damping force which is inversely proportional to this inner loop proportional gain. This gain can be experimentally tuned so that it is high enough to provide a transparent environment without the risk of creating an unstable system. There is also a deadband of 0.2 N on joint space force measurements since the USM cannot regulate velocities under 14 rpm. If there was no deadband present, the device would enter a limit cycle due to its low-velocity nonlinearity [88].

Three separate zero force control experiments were performed for three different types of unidimensional trajectories $\theta_L(t)$, i.e., linear trajectories along the $\theta_{FE}$ and $\theta_{RUD}$ axes, and circular movements in the robot workspace, defined by angle $\phi = \text{atan2}(\theta_{RUD}, \theta_{FE})$, using zero force control with gravity compensation. A user performed sinusoidal movements of varying velocity and frequency with velocities at the MR-SoftWrist's limits (~100
Figure 3.10: Resulting task space force and torques from an experimenter applying quasi-static loads on the device’s handle. Force measurements from the springs were transformed with the Jacobian to task space variables while measurements from the ATI Nano17’s sensor frame were transformed to task space variables via a force-torque transform [88].

deg/s). The results of these experiments can be seen in Fig. 3.11. A fit was performed separately for each experiment’s coordinates with the model

\[ \tau_L = I \ddot{\theta}_L + b \dot{\theta}_L + \tau_c \text{sign}(\dot{\theta}_L) \]  

(3.9)

where \( \tau_L \) is the applied load torque as measured by the springs, \( I \) the device inertia, \( b \) the perceived damping, and \( \tau_c \) the perceived Coulomb friction. A multiple linear regression was performed which gave estimates of \( I, b, \) and \( \tau_c \) for each coordinate. The values are presented in Table 3.3 [88]. Plots of velocity versus torque for the three experiments can be seen in Fig. 3.11, demonstrating that at low velocities the user perceives a Coulomb fric-
Figure 3.11: Experimental validation of zero force control for FE, RUD, and circular movements. The damping force is inversely proportional to the proportional gain at the joint space force control level. The data has been down sampled to 50 Hz for visualization. The maximum interaction torques are approximately 150 N·mm for wrist FE and RUD and 100 N·mm for circular movements [88].

Table 3.3: Zero Force Control Parameters [88]

<table>
<thead>
<tr>
<th></th>
<th>$I$ [kg·m$^2$]</th>
<th>$b$ [N·mm·s/deg]</th>
<th>$\tau_c$ [N·mm]</th>
<th>$R^2$</th>
</tr>
</thead>
<tbody>
<tr>
<td>FE</td>
<td>0.0016</td>
<td>1.25</td>
<td>15</td>
<td>0.84</td>
</tr>
<tr>
<td>RUD</td>
<td>0.0018</td>
<td>1.14</td>
<td>15</td>
<td>0.87</td>
</tr>
<tr>
<td>$\phi$</td>
<td>0.0005</td>
<td>0.58</td>
<td>5.4</td>
<td>0.93</td>
</tr>
</tbody>
</table>

The friction force, while at higher velocities the damping force dominates. There is an additional inertial effect from the mass of the legs and end effector that can be noticed at intermediate velocities from the deviation of torques from the straight line portion of the data. It should be noted that the main cause of the perceived Coulomb friction is the use of the deadband on joint space force measurements, required to compensate the USM low-velocity nonlinearity [86, 88].
3.3.6 Rendering Virtual Stiffnesses

To render virtual stiffnesses, the MR-SoftWrist uses the control structure from Fig. 3.4 [88] with the choice of $k_v$ as a symmetric, positive semidefinite 3x3 matrix. Two sets of experiments were performed to evaluate rendering virtual stiffnesses with the MR-SoftWrist. Both tests were performed for either wrist FE or RUD where one DOF (FE or RUD, respectively) was commanded to display a virtual stiffness, while the other two (including $z_c$) were commanded to display zero force. In the first experiment, a static interaction test, an experimenter applied quasi-static loads through the device’s handle for various virtual stiffnesses in the direction of the desired DOF. This experiment evaluated how well the device can track desired force trajectories for varying impedances [88].

In the second experiment, to characterize the device’s dynamic range, an experimenter applied sinusoidal motions of varying frequency (see Fig. 3.13), with frequency content up to 5-10 Hz, for the same varying impedance values. Estimates of the virtual stiffness transfer function $K_v(f) = \frac{T_L(f)}{\Theta_L(f)}$ relating the Fourier transform of output torque ($T_L(f)$) to the Fourier transform of input position ($\Theta_L(f)$) were obtained using the Matlab function `tfestimate` for each experiment. The results of both experiments are presented in Fig. 3.12 [88]. Defining bandwidth as the frequency range for which the ratio of virtual to desired spring stiffness is within ±3 dB, the device has a minimum bandwidth of 5.5 Hz which occurs for the lowest virtual stiffness for RUD [88].

3.4 System Validation

A case study with one able-bodied participant involving a semicircular arc pointing task under controlled kinesthetic feedback during fMRI was conducted (see Fig. 3.1). The specific goal of the case study was to validate the device as an MR-compatible haptic
Figure 3.12: Characterization of the task space stiffness control of the MR-SoftWrist, as described by the torque vs. angular displacement plots (top) for FE (left) and RUD (right) under the application of slowly increased loading; and by Bode plots of the virtual stiffness transfer function $K_v(s) = \frac{K_v(s)}{s}$ under dynamic manual perturbations (bottom). Experiments at different values of commanded virtual stiffness $k_v$ are conducted, and values are reported as a function of the natural manipulator stiffness in task space coordinates $k$. The Bode plots only show estimated transfer function values with coherence values greater than 0.8 [88].

Figure 3.13: Representative manual perturbations applied through the dynamic characterization of rendering various virtual stiffness values. Displayed are the trials for $k = 0.5k_v$. (top) Sinusoidal velocity profiles applied by the experimenter, and (bottom) corresponding Fourier Transforms.
interface for wrist sensorimotor protocols with kinesthetic feedback, by demonstrating the possibility of computer-controlled force display to elicit distinguishable kinetics and neural response without image artifacts due to head movements and/or noise [88].

### 3.4.1 Methods

An able-boded individual (25 yr, M) participated in an fMRI experiment consisting of a 2DOF wrist circular arc pointing task, modeled after the task presented in [167], requiring wrist FE (angle mapped on the horizontal axis) and RUD (angle mapped on the vertical axis), along a circular trajectory with radius \( \rho = 15 \) deg. The task involved pointing from an east target (\( \theta_{FE} = \rho, \theta_{RUD} = 0 \)) to a west target (\( \theta_{FE} = -\rho, \theta_{RUD} = 0 \) deg) (see Fig. 3.14 (left)). During the pointing task, the participant received online visual feedback; movement initiation and direction were triggered by visual display of a ghost cursor, which moved along the circle in either a clockwise or counterclockwise direction, reaching the target \( T_d = 3 \) s after departure. After the participant reached the target and maintained his position over the target for 0.5 s, the ghost cursor started movements toward the next target, along a circular trajectory that was calculated from polar coordinates with constant distance \( \rho \) and angle \( \phi(t) \) determined from a minimum-jerk trajectory with extent 180 deg. Guide lines were displayed around the desired trajectory, and the participant was instructed to try to remain inside them during the circular pointing task and to reach the target before the ghost cursor, moving along the same direction (either clockwise or counterclockwise) [88].

During the experiment, the MR-SoftWrist was controlled with the task space impedance controller shown in Fig. 3.4 [88], with \( \mathbf{x}_{des} \) defined at each iteration as the point minimizing the Euclidean distance in the task space between the measured position \( \mathbf{x} \) and the circle with radius \( \rho \), obtained through iterative numerical optimization. After calculation of \( \mathbf{x}_{des} \), the task space desired force vector \( \mathbf{f}_{E,k} = k_v(\mathbf{x}_{des} - \mathbf{x}) \) (with \( k_v \), scalar) was projected along
the normal direction to the circle and applied via the task space force controller shown in Fig. 3.4. Three modes were implemented and tested in the pilot study: i) zero force (ZF) mode, with $k_v = 0$; ii) path control (PC) mode, with $k_v > 0$; iii) error augmentation (EA) mode, with $k_v < 0$. Constants of $k_v = -30 \text{ N-mm/deg}$ and $k_v = 230 \text{ N-mm/deg}$ were chosen for the EA and PC modes respectively. The gain constants were manually tuned to result in noticeable and measurable task space assistance/perturbation forces within conservative safety and stability margins. To prevent possible hardware failures deriving from task instabilities, force output generated in the EA mode was also limited to 100 N-mm [88].

After a training session to familiarize themself with the robot and its control modes (roughly 30 min in a side lab outside the scanner), the participant executed the wrist pointing task during fMRI in repeated block design experiments (one per each control mode). Each experiment consisted of the alternation of visual control (VC) blocks and active blocks. In the VC blocks, the ghost cursor was moving and the participant was informed by a “STOP” message appearing at the beginning of the block to follow only visually the movements of the ghost cursor. In the active block, the participant was instead asked to perform circular wrist pointing movements. Each block consisted of six repetitions, each with random direction, with a minimum block duration of 21 s (actual block durations could vary depending on the time required for the participant to complete the movement). Each experiment switched between seven VC blocks and six active blocks, for a total experiment duration of $305 \pm 5$ s. For details on acquisition of brain images, see Appendix B [88].

### 3.4.2 Data Analysis

Kinematic and force data were collected continuously during the experiments and logged at 50 Hz. A scalar error value $e$ was defined at each time sample as the distance between the current position and the desired trajectory $e(i) = |x(i) - x_{\text{des}}(i)|$; for each movement $k$
(composed of $M$ samples) an average error $e_k = \sum_{i=1}^{M} \frac{1}{M} e(i)$ was defined and used for statistical analysis. Interaction force values were similarly measured from spring deflections, converted to task space torques, and averaged for each movement for comparison between different experimental conditions. The estimated task space interaction forces were projected along the tangent to the circle measured by the current point $x_{\text{des}}(i)$, which enabled distinguishing between lateral perturbation forces $F_{\text{lat}}$, normal to the desired trajectory, and resistance forces $F_t$, along the desired trajectory [88].

3.4.3 Results

During the semicircular pointing task conducted in the scanner, the MR-SoftWrist was capable of implementing kinesthetic feedback to manipulate error depending on the selected control mode ($e_{EA} = 1.7 \pm 0.6$ deg, $e_{ZF} = 0.9 \pm 0.3$ deg, $e_{PC} = 0.3 \pm 0.1$ deg, mean ± standard deviation) (see Fig. 3.14 (right)), with all paired comparisons (one-tailed $t$ test, dof = 70) rejecting the null hypothesis with high statistical significance ($p < 0.001$). Error manipulation was achieved through regulation of task space interaction force in the perpendicular direction to the task ($F_{\text{lat,EA}} = 40 \pm 12$ N·mm, $F_{\text{lat,ZF}} = 10 \pm 2$ N·mm, $F_{\text{lat,PC}} = 22 \pm 8$ N·mm), with all paired comparisons for the $F_{\text{lat}}$ values significant at $p < 0.001$. In contrast, differences in resistance force displayed along the tangential task direction were minimal ($F_{t,EA} = 41 \pm 2$ N·mm, $F_{t,ZF} = 43 \pm 2$ N·mm, $F_{t,PC} = 44 \pm 3$ N·mm), and mostly explained by variability in tangential velocity among different control modes that results in slightly different interaction force values (refer to the zero force behavior shown in Fig. 3.11 right). Images of the resulting brain activation can be seen in Fig. B.1 [88].
Figure 3.14: Pilot study results. (left) GUI of the force feedback schematic for the circular wrist pointing task. The participant position ($x$) is displayed with the cyan cursor, and movement direction is visually cued with the red ghost cursor towards the correct target. The force feedback controller, shown in Fig. 3.4, is implemented, with $x_{des}$ being the nearest point along the circular trajectory. With positive $k_v$, the robot implements path control (PC), while with negative $k_v$ the robot implements error augmentation (EA). When $k_v = 0$ the robot implements the zero force (ZF) mode (colors in this image differ from the experiment for improved quality in grayscale print). (right) Kinematics measured in the three different control modes, during the pilot study experiment. EA is only shown in the top portion, and ZF and PC are shown only in the lower portion [88].

3.5 Conclusion

In this chapter, the design and fabrication of a parallel 3DOF co-located haptic MR-compatible robot was presented. The MR-SoftWrist is a novel MR-compatible wrist robot capable of accurate kinesthetic feedback during 2DOF wrist pointing movements executed during fMRI. The development of an accurate and high-bandwidth kinesthetic feedback robot suitable for operation during fMRI required investigation of non-standard mechatronic design practices. The MR-SoftWrist uses a parallel architecture, placing its three actuators on a stationary base frame, as well as providing increased structural rigidity and increased torque and speed output compared to a serial design along with benefits for scanned image quality since the actuators are housed on a stationary base and are located far from the isocenter. To guarantee MR-compatibility, the MR-SoftWrist is actuated via non-
backdrivable piezoelectric ultrasonic motors, and interaction control is achieved via force control based on the measurement of the linear deflection of three compliant elements. To the best of the author's knowledge, the MR-SoftWrist is the first parallel manipulator employing non-backdrivable actuators and a compliant transmission to render task space virtual impedances. With its workspace including wrist FE and RUD rotations in a circle with a 20 deg radius, high spatial (0.02 deg) and torque (9 N-mm) resolution, and capability for producing 1.24 rad/s of angular velocity and 1.5 N-m of torque around both wrist axes, the device is suitable for investigating wrist sensorimotor protocols under force feedback during fMRI [88].

Position control was demonstrated for a sinusoidal trajectory covering the entire 2D workspace, oscillating at a frequency of 0.4 Hz. The root mean square error was 2.25 deg while the maximum error was 3.45 deg in tracking either DOF. Although this error is large compared to the performance of non MR-compatible rehabilitation robots [14], it is a result of the use of series elastic actuation, which facilitates interaction control but does not allow high position control bandwidth and accuracy, and the nonlinear velocity sourced ultrasonic motors. Since position control would mainly be used for playing back user motions in visual- and proprioception-controlled experiments, the limitation in position control accuracy is justified by the more pertinent requirement of achieving accurate task space impedance control for interactive tasks. Torque measurement accuracy of the MR-SoftWrist was evaluated in a static test, yielding mean force estimate errors of 17 N-mm and 34 N-mm for $\tau_{FE}$ and $\tau_{RUD}$ respectively. This test shows that the MR-SoftWrist is capable of measuring user interaction torques accurately within the series elastic architecture, validating the use of the device as an accurate force measuring device for haptic fMRI experiments [88].

The ability of the MR-SoftWrist to display a transparent mode was evaluated in three
experiments: one for each DOF and one for a representative task (in this case wrist pointing along a circular trajectory in FE and RUD coordinates). The maximum backdrivability torque for wrist FE and RUD was found to be approximately 150 N·mm at 100 deg/s in each case. Estimated Coulomb friction from each DOF were the same (15 N·mm) while other parameters were similar, i.e., inertia estimates of 0.0016 and 0.0018 kg·m² and damping estimates of 1.25 and 1.14 N·mm·s/deg for FE and RUD respectively. The MR-SoftWrist’s maximum backdrivability torque and inertia estimates are almost identical to those of the wrist module of the MIT-MANUS [71]. Additionally, the maximum interaction force of 150 N·mm felt by the user is only 10% of the maximum output torque [88].

Impedance control of the device was evaluated by validating its ability to display a range of virtual stiffnesses for both static and dynamic cases. Static experiments showed accurate tracking, while dynamic experiments revealed that the device is capable of achieving at least 5.5 Hz of impedance control bandwidth for a 14 dB range of virtual stiffness values, corresponding to 25-125% of the device’s physical reflected stiffness in the nominal configuration. This is more than sufficient for the range of 2-5 Hz for human movements [184] and far exceeds the approximately 1 Hz position control bandwidth of previous MR-compatible force feedback haptic interfaces [177]. Additionally, the device can stably render impedances higher than those of the device’s physical impedance, although these impedances are not passive [84, 185]. Finally, unlike previous MR-compatible devices which require custom force sensor, which are difficult to manufacture [186], the MR-SoftWrist achieved accurate force control through easily manufactured linear elastic springs [88].

The MR-SoftWrist was validated in a case study with a single able-bodied participant, demonstrating the capability of the device to apply force feedback to elicit distinguishable kinetic and neural response. The pilot validation study demonstrated the capability of si-
multaneous acquisition of functional images during interaction with the MR-SoftWrist as demonstrated by the limited head movements. Activation maps were determined for different control modalities, including path control (PC), zero force (ZF) and error augmentation (EA). Comparison of the activation maps showed that activation in the premotor cortex was bilateral in the EA mode, but only contralateral in the ZF and PC modes. The contrast between activation in the EA and ZF conditions revealed areas with higher activation in the EA mode relative to the ZF mode, which included bilateral portions of the premotor cortex and somatosensory cortex, and bilateral BA44 (part of Broca’s area). Significant activation was observed in the EA mode also in parts of the contralateral cerebellum. These functional neuroimaging results indicate the capability to generate experimentally-controllable activation maps in response to haptic environments, but the results of this study are not at all meant to provide any findings on the neural correlates of robot-assisted motor control, given the inclusion of only one participant in the analysis [88].

The MR-SoftWrist is a new tool that can be useful to study motor learning under haptic guidance. The device can control multi-DOF movements directly unlike most other MR-compatible robots, which control unconstrained movements through an end effector design [57, 173, 174]. Additionally, the MR-SoftWrist interacts with the wrist and thus does not induce significant head motion, which has shown to be a significant issue when whole arm movements with large feedback forces are employed [56]. Previous studies addressing wrist movements during fMRI involved 1DOF tasks which lacked control or repeatability of wrist movements [41, 166], measured only isometric torques [187], involved only neonates [188], or have not been validated during fMRI with a user [136]. Recently, a multi-DOF wrist pointing study was conducted using motion tracking, but did not have the capability of force feedback [167]. To the best of the author’s knowledge, the MR-SoftWrist is the first MR-compatible haptic robot that can measure and support 2DOF wrist
movements with complete control and MRI-compatibility validation, including validation during an actual fMRI experiment [88].

The MR-SoftWrist is a promising tool for neuroscientists to investigate human sensorimotor control under haptic guidance. In the future, this device will be used in experiments with impaired participants to examine the effects of different force fields or robot-aided therapeutic protocols on brain reorganization promoting motor learning and neurorecovery. With a better understanding of brain plasticity of patients with neurological disorders, personalized treatments might be conducted, possibly increasing the specificity of movement-based robotic therapy after neurological injury [88].

The capabilities of the MR-SoftWrist to perform human motor control experiments during fMRI are due to its use of series elastic actuation. The series elastic actuators enabled a co-located design with non-magnetic, non-backdrivable, and nonlinear velocity sourced ultrasonic motors. In this thesis, the force accuracy and dynamic interaction capabilities of this MR-compatible series elastic actuation scheme have been thoroughly characterized, thus demonstrating the feasibility of using the MR-SoftWrist in the mentioned future applications.
Chapter 4

Robot-Aided Biomechanical Assessment: Development and Validation of the SE-AssessWrist for Evaluating Wrist Stiffness and Range of Motion

The SE-AssessWrist is a novel two-degree-of-freedom series elastic actuated wrist exoskeleton. The SE-AssessWrist was created to assess wrist stiffness and range of motion. To measure wrist stiffness, the device must be able to regulate slow ramp position trajectories and measure interaction torque, while to measure wrist range of motion, the device must have sufficient range of motion to match that of the human wrist. To arrive at a design which meets these requirements, the SE-AssessWrist uses series elastic actuation and a Bowden cable transmission. The series elastic actuators provide torque measurement for assessing wrist stiffness, while the Bowden cables decouple device torque and range of motion capabilities. Additionally, the series elastic actuators provide the means to regulate a transparent environment, despite friction from the Bowden cable transmission, while assessing range of motion.

In this chapter, the design of the device, including the description of components and specifications of the series elastic actuation and Bowden cable transmission, are detailed. The control performance and range of motion of the device are characterized with respect to the intended assessment application. Finally, a cast study is presented which provides the first instance of evaluating both wrist stiffness and range of motion with the same robotic device.
4.1 Introduction

Rehabilitation robots, which are used to promote motor recovery after neurological injury [164], have become prominent in the clinical research setting [189]. Although clinical improvements using these devices have been observed [20, 70], the best way to perform robotic rehabilitation is still unclear [12, 15]. In an attempt to increase our understanding of recovery throughout neurorehabilitation, accurate, descriptive, and repeatable assessments are required to aid in determining patient-specific rehabilitation protocols [28, 29]. The only current assessments which are regularly incorporated into neurorehabilitation are clinical measures [20], which are typically subjective [190], labor intensive [39], and graded on an ordinal scale [38]. In contrast, robotic measures offer the possibility for accurate, objective, repeatable, and easily implemented assessments throughout neurorehabilitation [191].

One of the most commonly proposed robotic assessments is examining movement smoothness [192], in which studies have observed increase in smoothness over the course of therapy [71]. This assessment has been shown to correlate well with existing clinical measures [37], offering the potential for more automated evaluations of recovery. Another important suggested assessment in the literature is using robots to study the underlying neural processes for brain plasticity by using functional magnetic resonance imaging in conjunction with haptic robots [45, 170]. In Chapter 3 of this thesis, the development and validation of such a device – the MR-SoftWrist – was described. A less developed robot-aided assessment comes from biomechanical motivations to assess joint impedance [60].

Muscular impedance is an important characteristic relating to functionality of our muscles. Similar to mechanical systems, the joints of the human body can be modeled as mass-spring-damper systems [60], as shown in see Fig. 4.1. However, in the human body, these properties are not necessarily linear, for example, exhibiting stretch-reflex at the onset of motion, variation in stiffness at the limits of the workspace, and stretch re-
Figure 4.1: A linear mass spring damper system with force generation potential. Within the limits of the wrist’s ROM, sufficiently far from the maximum ROM, the wrist can be modeled as a linear mass spring damper system for inbound and outbound movements. The wrist is an active system ($F(t)$), therefore to measure passive properties the system's force production must be monitored. This ensures the measured properties are those of the passive system and not of the controlled system.

laxation [42, 63, 66, 193]. Unlike most mechanical systems, the stiffness and damping of human impedances are variable as they can be modulated through muscular contractions. Therefore, joint stiffness, which is due to stretching muscle and tendon tissue, is typically studied while the user is passive. Passivity of the user can be monitored through measurement of muscular activity using surface electromyography (sEMG) electrodes. By normalizing muscle activity obtained during stiffness measurement to maximum activity, the passivity of the user can be evaluated. Using robots and sEMG measurements allows for precise estimation of passive joint stiffness due to the possibility of high resolution position and torque sensors. Studies estimate user stiffness by commanding the robot to smoothly move the user at constant velocity over some range of motion (ROM) while measuring user position and torque. So long as the velocity of the robot is sufficiently low to not elicit significant damping or inertial affects during acceleration, stiffness estimates are invariant to strain rate due to the variance of connective tissue to strain rate as demonstrated in [42, 63, 64, 66, 194, 195].
4.1.1 Wrist Stiffness and Range of Motion

Studies evaluating passive joint stiffness have been reported since the 1970s [63, 64, 194, 195], although only recently has multi-DOF wrist stiffness been evaluated [42, 66, 68]. This is in part because assessing multi-DOF wrist stiffness is particularly challenging from a robotic design standpoint due to the wrist’s two perpendicular and intersecting degrees of freedom (DOF), i.e., wrist flexion/extension (FE) and radial/ulnar deviation (RUD). Wrist biomechanical assessment is of particular interest due to the wrist’s importance for performing activities of daily living (ADLs) and rehabilitation. In a survey of 500 spinal cord injury survivors carried out by Snoek et al., one of the respondents’ top priorities was to recover hand function, likely due to the wrist’s importance in performing ADLs [69]. As for rehabilitation, studies of wrist rehabilitation have reported improvements in proximal joints despite not being directly trained [70], although the corollary has not been observed [18]. In addition, studying wrist stiffness [42, 66] and its related ROM [68, 70] could offer insight into recovery from wrist disorders, such as unwanted increased tone, leading to increased wrist stiffness [61–63], and recovery during rehabilitation. From a biomechanical perspective, wrist stiffness, as opposed to or in addition to wrist viscosity and inertia, is important to study as wrist stiffness dominates wrist dynamics as evidenced in the work of Charles et al. [60, 196, 197]. Furthermore, a study of wrist stiffness and ROM with cadavers found that the orientation of wrist stiffness and ROM are oblique, oriented in the direction of the dart thrower’s motion [67], which is unique to humans and to each individual [66]. This dart thrower’s plane, as illustrated in Fig. 4.2, is oriented in the direction of radial-extension to ulnar-flexion with an angle of about 20-30° from pure flexion towards ulnar deviation [66, 68, 198]. This plane has been called the plane of greatest functionality, important from an evolutionary perspective for throwing spears and using a hammer [67].

The only in vivo 2DOF wrist stiffness studies [42, 66] have been carried out with the
InMotion Wrist, a wrist rehabilitation device which for the wrist consists of two direct current (DC) motors driving a differential gear transmission. In the preliminary work [42], wrist stiffness was studied for a circle in the FE and RUD axes with about a 20° radius. In the work by Pando et al., this ROM was expanded, but still only covered 70% of the target wrist ADL ROM. This is in part because wrist exoskeletons have been designed for ADL torque and ROM [72], and generally fall short of the wrist’s complete ROM capability since ADL ROM is significantly smaller than maximum ROM (see Table 4.1).

While the studies in [42, 66] provided in-depth analysis and computation of wrist stiffness, the studies had limitations due to the hardware. Interaction torque was estimated through motor current, leading to possible inaccuracies in stiffness estimation as a result of the device dynamics, such as static friction. The device was also limited in its continuous torque output (1.95 N·m). Additionally, the device did not measure joint angles directly, due to offsets between the device and anatomical axes. Other robots designed for wrist

Figure 4.2: Illustration of the dart thrower’s motion with reference to representative wrist stiffness and range of motion envelopes. Referring to the dart thrower’s axis as the path of least stiffness or greatest range of motion, reveals an axis rotated 20-30° from pure flexion towards ulnar deviation. Representative envelopes were referenced from [66, 68].
Table 4.1: ROM and Torque of the Wrist and Prominent Wrist Robots

<table>
<thead>
<tr>
<th>Wrist FE</th>
<th>Wrist RUD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max [68, 200, 201]</td>
<td>185 12.3</td>
</tr>
<tr>
<td>ADL [72]</td>
<td>115 0.35</td>
</tr>
<tr>
<td>SE-AssessWrist</td>
<td>185 4.9</td>
</tr>
<tr>
<td>ARMin-II [202]</td>
<td>85 N.A.</td>
</tr>
<tr>
<td>CADEN-7 [72]</td>
<td>120 N.A.</td>
</tr>
<tr>
<td>IIT Wrist [203]</td>
<td>144 1.53</td>
</tr>
<tr>
<td>InMotion Wrist [71]</td>
<td>120 1.43</td>
</tr>
<tr>
<td>MAHI Exo-II [204]</td>
<td>72 1.67</td>
</tr>
<tr>
<td>OpenWrist [83]</td>
<td>135 3.6</td>
</tr>
<tr>
<td>RiceWrist-S [14]</td>
<td>130 3.37</td>
</tr>
<tr>
<td>WRES [205]</td>
<td>75 1.62</td>
</tr>
<tr>
<td>Wrist Gimbal [206]</td>
<td>180 1.77</td>
</tr>
</tbody>
</table>

rehabilitation would have a subset of the same limitations in torque output, ROM, torque estimation, or joint offset. Furthermore, most wrist rehabilitation devices are designed for a handle which must be gripped [82, 93, 103], which is suboptimal for estimating passive stiffness as the hand must be completely wrapped in order for the participant to remain passive during the movements. Additionally, a grip posture puts wrist muscles in a more active state, and ROM estimates from a motion capture study demonstrated that wrist ROM was larger when the fingers were left natural and not having to grip an object [199, 200].

Although passive stiffness of the wrist has been studied to a limited degree for small
ROM, complete active wrist ROM with a robot has not, since existing wrist exoskeletons do not have sufficient ROM to achieve this. Using a robot to assess ROM offers advantages over previous studies using electrical goniometers, which are challenging to use – especially in measuring multi-DOF movements [207, 208]. While it might be possible to study ROM with a passive device or through motion capture, neither solution would offer the possibility for assessing both stiffness and ROM through the same mechanism, which would be beneficial for repeatability. By assessing both properties with the same device, wrist alignment can be maintained allowing for direct comparison between the two. The only study examining maximum stiffness and ROM with a robotic device was a study performed with cadavers [68]. The robot used in the study, only applicable for cadavers, was commanded to move the wrist in a given direction until 2 N·m of torque was applied, and this was considered the maximum ROM [68]. Note that in this case, passive ROM was being studied, which while important, is different than active ROM in which the user determines the maximum ROM [191].

While assessing wrist stiffness and ROM is important for understanding recovery from neurorehabilitation, the devices used thus far are insufficient as they have a limited combination of torque output and ROM. In addition, rehabilitation devices do not measure torque, which is required to estimate joint impedance. Instead these devices estimate torque from motor current leading to inaccuracies [42, 66, 70, 205]. In this chapter, the development of a device, the SE-AssessWrist, which overcomes these limitations, is described. The SE-AssessWrist features SEAs in conjunction with a Bowden cable transmission scheme for increased torque output and ROM, as well as direct torque measurement. The following subsection give more details on Bowden cable actuation.
Figure 4.3: (a) Traditional co-located haptic actuation architecture in which the motor is grounded on the exoskeleton and motor position is directly transmitted to the output through a fixed pulley. (b) Bowden cable actuation architecture with the motor grounded to the earth, thereby not imposing its mass on the device. This comes at the cost of increased and variable friction between the motor and environment. (c) Illustration of the capstan model which relates input and output cable tension for situations in which the cable is wrapped around a pulley. Bowden cable actuation can be compared to the case of a cable sliding over a fixed cylinder, resulting in a much larger loss of tension than the case of a cable running over a pulley allowed to rotate freely on a bearing.

4.1.2 Bowden Cable Transmission

Rigid haptic displays, such as the devices in Table 4.1, are limited by design in that the motors almost always must be housed directly on the device as a result of co-located actuation from cable-pulley mechanisms [73]. An alternative to this actuation scheme is the use of the so-called Bowden cable transmission scheme (see Fig. 4.3). The use of Bowden cables for force transmission was first realized for bicycles. In bicycles Bowden cables enable flexible re-routing of force transmission between the brake levers and calipers as well as shifters and derailers. Bowden cable transmissions have since been adopted for body-powered prosthetic devices [209, 210] and rehabilitation robots [131]. For the case of robotic manipulators, motors are used to precisely control either Bowden cable position or tension. Using Bowden cables in robotic devices allows the actuators to be housed remotely, decreasing device mass and increasing device compactness [107, 131, 211].

The main operating principle of the Bowden cable transmission scheme is a cable (also
referred to as a tendon, usually steel or nylon) being routed through an outer sheath (also referred to as the conduit). The cable and sheath are both stiff in the longitudinal direction, but flexible laterally. The sheath is clamped at both ends, while the cable is free to travel inside the sheath. The sheath is usually made of a flexible material (such as nylon), and might also include helically coiled cable between the tendon and sheath to provide some stability of structural shape. If the cable and sheath are perfectly aligned with no bend, there is no contact between the two and the case of a traditional haptic display is realized; however, this is often not the case as the main advantage of Bowden cables is to re-locate the motors off-board, often requiring routing of the cables with a bend. This results in variable friction between the cable and sheath depending on the bend radius of the conduit and the tension on the cable. When the cable is pressed against the sheath, a normal force is created which leads to friction proportional to this force. Friction which results from this interaction can be modeled with the capstan equation,

\[ T_o = T_i e^{\mu \alpha} \] (4.1)

where \( T_o \) is the output tension, \( T_i \) is the input tension, \( \mu \) is the coefficient of friction, and \( \alpha \) is the wrap angle, since the relative motion of the cable on the sheath can be thought of as a cable moving on a fixed cylinder. As can be seen from equation 4.1, as the wrap angle is increased, so is the amount of tension loss in the transmission as a result of friction. Since the normal force between the cable and sheath need not be constant, a Bowden cable transmission can be modeled as a cable-pulley system with highly variable friction.

Bowden cable transmissions result in highly nonlinear force transfer as a result of Coulomb friction, stiction, viscous friction, and flexibility which lead to backlash, hysteresis, slip, and loss of tension from the motor to the load. These losses in tension have been both theoretically [212–214] and experimentally [209, 215, 216] modeled. Theoret-
cal models are often complex and idealized for limited sheath motion which is not always realistic [213, 217]. Experimental characterizations have been extensively performed for analyzing different parameters and how they affect force transmission efficiency, defined as the ratio of output to input tension [209, 215, 216]. These studies have observed that using nylon cables is more force efficient than steel. Additionally, adding a liner, such as Teflon, also increases force transmission efficiency, although these coatings can wear off leading to increased friction. Note that for small bend angles in the cable-sheath system, the differences in materials are small, but as bend angle increases they become increasingly important. In addition, a Bowden cable scheme requires pre-tensioning the inner cable, which results in some friction always being reflected to the user. Although in [218] a mechanism was developed for low force applications to essentially eliminate the need for pre-tension, this mechanism added much additional complexity to the system. Since the inner cable cannot be compressed, these schemes are limited to applications in which the cable remains under tension, which can be accomplished with one motor actuating a continuous cable or two motors pulling two separate cables for a given joint. This redundancy is more efficient as with the case of a single motor, the motor most overcome the tension of the cable that is not being pulled [212].

Bowden cable transmissions were used in early haptic devices, such as the Utah/MIT dextrous hand [219], due to difficulties of co-locating motors with the device. The first rehabilitation robot that benefited from Bowden cable actuation was the 8DOF LOPES exoskeleton for lower-limb rehabilitation [107]. Unlike the Lokomat [6, 220], which was mainly operated in position control mode, the LOPES could be operated in impedance control modes, allowing for low-impedance backdrive modes [221]. This was possible through the use of series elastic actuators (SEA), with elastic elements placed at the actuated joints to directly measure joint torque. Using SEAs greatly simplifies the problem of force con-
trol, but comes at the disadvantage of reduced position control bandwidth [138]. Reduced position control bandwidth is typically not a concern for a series elastic actuated rehabilitation robot, as force control performance is prioritized due to its importance in implementing the interactive controllers necessary for engaging rehabilitation [84, 88, 107, 134, 145, 222]. An alternative solution would be to use rigid force sensors; however, combining these rigid sensors with non-collocated actuators for direct force control can lead to contact instabilities [117]. As a result of the success of the LOPES exoskeleton, Bowden cable transmission with SEAs has also been used for a knee [130], elbow [223], and a finger device [131]. All of these devices have exploited the actuation scheme for different benefits, whether it be decreased inertia [130] or the possibility for a more natural design [131]; however, Bowden cable actuation has yet to be employed in a 2DOF wrist exoskeleton. Using Bowden cables in a wrist exoskeleton could enable a design with larger ROM and increased torque output, due to the ability to located the actuators off-board.

Unlike previous wrist exoskeletons which were designed with traditional haptic design methodologies (i.e., backdrivable actuators and open-loop torque control through direct drive transmissions), this thesis proposes a design using SEAs in conjunction with a laterally flexible Bowden cable transmission and geared DC motors. These SEAs are created by placing an elastic element between the user and Bowden cable transmission at each joint, enabling accurate torque estimation while achieving compliant physical human-robot interaction (see Fig. 4.4). The SEAs provide a direct measure of torque for both assessment of passive wrist stiffness, in addition to a measure of user interaction torque required for transparent zero force control used during wrist ROM assessment. To achieve increased ROM compared with previous wrist exoskeletons, the motors are located off-board, which enables a compact end-point solution, greatly increasing the ROM of the device. While in this way range of motion is increased, torque capabilities are not sacrificed as the torque
amplification occurs off-board through the gear train.

### 4.2 SE-AssessWrist

The following section provides an overview of the robot, the SE-AssessWrist. The SE-AssessWrist, unlike other existing exoskeletons designed for robot-aided rehabilitation, was specifically designed for robot-aided assessment. Additionally, users interface with the device through an open hand configuration, providing a more natural and relaxed hand configuration which is essential for stiffness and ROM estimation. Through the use of SEAs, the device also directly measures torque, which also serve to provide a transparent environment through zero force control during ROM assessment. ROM assessment is facilitated through a Bowden cable transmission that allows a design with higher ROM and torque than previous exoskeletons (see Table 4.1).
4.2.1 Mechanical Design

A serial design was pursued for its increased ROM capabilities and more straight-forward mechanical structure compared with a parallel design. Since the standard in wrist modeling and therefore exoskeleton design is to place wrist FE as the first axis and RUD second, the design of the SE-AssessWrist was restricted to such a serial kinematic architecture. Additionally, to provide accurate wrist position estimation, the design was also restricted to an exoskeleton structure, a structure in which joint axes align with those of the wrist. To create such a design, a serial RR mechanism with a passive prismatic joint at the handle was pursued. The passive prismatic joint allows the wrist axes to remain aligned despite potential wrist axis translation during movement, increasing joint alignment.

Additional considerations for the device were to have sufficient ROM to enable wrist ROM assessment for most individuals, and to have sufficient torque to estimate wrist stiffness. Previous wrist stiffness studies have cited limitations in ROM and torque, limited to 1.95 N·m \([42,66]\), and as such torque of at least 3 N·m was desired for the SE-AssessWrist. Since stiffness measurement necessitates an estimate of torque, the device must measure torque, unlike previous devices. Correspondingly, the device needs sufficient torque resolution to adequately estimate wrist stiffness. Based on preliminary estimates, at least 10 N·mm of torque is desired. The following subsections detail the components that satisfy these requirements.

Bowden Cable Transmission

To achieve the desired torque and ROM, a Bowden cable transmission was selected. A Bowden cable transmission is advantageous for these constraints since a large actuator can be used off-board without affecting compactness, which affects achievable device ROM while maintaining structural integrity and control performance, of the exoskeleton portion
of the robot. Additionally, capstan arcs and differential mechanisms make it difficult to maintain compactness with a desired torque and ROM, providing further support for the Bowden cable approach.

A design tradeoff of this approach is that a Bowden cable transmission adds friction to the system. This friction can be alleviated through special consideration of materials. For the SE-AssessWrist, Bowden cables (Jagwire Road Shop Kit) with L3 slick lubed inner tubing and a 4mm outer diameter were chosen. This conduit is designed for 1.2 mm bike cables, and so 1.168 mm cable (McMaster-CARR 34235T28) were chosen. This cable has a 7x19 18-8 stainless steel braided construction, a nylon coating, and is rated to 133 N with a 5:1 safety factor.

A Bowden cable transmission can be actuated through either one or two motors. Using a single motor requires pre-tensioning the cable to half of the maximum force, and since friction in the cable-conduit system is proportional to this force, the control performance suffers. Additionally, such a constraint is more beneficial for wearable systems, since weight is a larger factor. Given the testbed nature of the SE-AssessWrist, the design here was chosen with two motors per DOF (see Fig. 4.5). In this way, the cables do not need to be excessively pre-tensioned, important for control performance.

**Motor and Gearbox Selection**

To achieve the desired torque requirements, a Maxon Motor RE40-148877 was chosen (see Table 4.2). Since this motor can only output 0.187 N·m of continuous torque, torque amplification was required. To accomplish this, a planetary gearhead with a 43:1 gear ratio was selected (Maxon Motor Planetary Gearhead, ceramic version, GP 42C 203120). This gearbox can withstand up to a continuous output of 15 N·m, and a peak torque of 22.5 N·m. It has an average backlash of 1°and a maximum efficiency of 72 % [224].
Figure 4.5: Illustration of the selected Bowden cable actuation architecture. Two motors actuate a single joint, but since cables can only pull, essentially each motor controls a single direction of movement. Due to friction from the cable contacting the conduit’s inner nylon sheath, friction results in loss of tension between the motor pulleys and the input pulley. Variables from the figure are motor \( M \), radius \( r \), torque \( \tau \), position \( \theta \), angular velocity \( \omega \), and force \( F \) while subscripts refer to input \( i \) and output \( o \) pulleys.

**Torque Sensing**

Torque measurement was accomplished through the adoption of a series elastic actuation architecture (see Fig. 4.6). Series elastic actuation provides the benefit of added compliance to the system for improved torque control over traditional rigid torque sensors. Additionally, through using a sufficiently stiff spring, position control performance can still be maintained for this application.

The elastic element chosen for the design was a double Archimedes spiral. This design

<table>
<thead>
<tr>
<th>( \tau_{cont} ) [N·m]</th>
<th>( i_{cont} ) [A]</th>
<th>( i_{max} ) [A]</th>
<th>( k_t ) [N·mm/A]</th>
<th>( P ) [W]</th>
<th>( V_{nom} ) [V]</th>
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<tbody>
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<td>42.4</td>
<td>60.3</td>
<td>150</td>
<td>48</td>
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</table>
Figure 4.6: High-level block diagram of the series elastic Bowden cable actuation scheme chosen for the SE-AssessWrist. By using an SEA, direct torque control is achieved, and enables force control despite nonlinear friction as a result of the gearbox and Bowden cable transmission.

is advantageous for this application since it is customizable, and largest in the radial direction where space constraints are less problematic than longitudinal ones. To measure the deflection of this spring, US Digital’s EM2 transmissive optical encoder module (EM2-2-10000-I) with a 50.8 mm diameter transmissive rotary hubdisk (HUBDISK-2-10000-375-IE) was selected. The encoder has 10,000 cnt/rev, which, with quadrature encoding, leads to a position resolution of $1.57 \cdot 10^{-4}$ rad. To achieve the desired torque resolution of 10 N-mm, a spring rate of approximately 64 N-m/rad was required.

The spring was designed through an iterative process, considering the relationship between various parameters, such as the spring’s radius, width, thickness, and other variables. Additionally, from a preliminary spring design, an offset of approximately 30% was de-

| Table 4.3: Spring Specifications |
|-------------------------------|----------------|-------------|---------------|---------------|
| Radius [mm] | Spiral Width [mm] | Thickness [mm] | Fillet Radius [mm] | No. Spirals |
| 30.48       | 2.654            | 4.826        | 1.143          | 1             |


Figure 4.7: Custom-designed double Archimedes spiral spring. (left) CAD rendering including key physical dimensions: spiral radius \( r \), spiral width \( w \), spiral thickness \( t \), inner and outer fillet radius \( f \), and number of spirals \( s \). (center) CAD rendering including the connection to the output pulley. The two components are connected through 4x dowel pins (1) and 6x 6-32 screws (2). In addition, the spring includes a thru-hole for the set screw which secures the end of the cable (3). The cable runs in a race (4) in the pulley which includes an end cable hole (5). (right) Physical spring with an integrated hub (6) and 2x (one is not visible) 90° offset threaded set screws holes (7) for mating with the output shaft.

Terminated between finite element analysis and actual spring constant. As such, and to not underdesign the spring rate, physical parameters were chosen which resulted in a predicted spring rate of 73 N·m/rad. The chosen parameters are presented in Table 4.3 and an image of the corresponding spring is shown in Fig. 4.7.

Note that unlike most other double spiral springs, the rotational spring in this work was made from aluminum (7075-T651) for a softer design, compared to steel, while still maintaining sufficient maximum torque. Additionally, the spring was created through computer numerical control machining, as opposed to the ubiquitous wire electrical discharge machining. Using computer numerical control machining enabled a design with the spring and spring hub, which mates with the output shaft, to be made as one piece, thus reducing mechanical play due to the small screws that would have been necessary to connect the two parts had wire electrical discharge machining been used.
Figure 4.8: A user interacting with the SE-AssessWrist during the demonstration experiment. The image highlights the arm cuffs, attachment between device and user, and the off-board actuators. 1) Forearm support, 2) visual display, 3) distal wrist cuff, 4) exoskeleton interface, 5) SE-AssessWrist, 6) FE motors, and 7) RUD motors.

4.2.2 SE-AssessWrist

An image of a user connected to the SE-AssessWrist is shown in Fig. 4.8. To see more images of the device, please see Appendix C. The complete mechanical specifications for the SE-AssessWrist are outlined in Table 4.4. The continuous torque was estimated by considering the motor torque constant and gear ratio, as well as inefficiencies in the gearbox (72%) and Bowden cable transmission (85% as estimated from [209]), resulting in a continuous torque of 4.9 N·m. The maximum torque is limited by the spring, as finite element analysis predicted local yielding at 5.1 N·m. Note that this is a conservative estimate, as gross yielding was not predicted to occur until 7.5 N·m, providing some safety factor. As can be seen, the device meets the desired torque requirements (Table 4.4) and ROM requirements (Table 4.1) for the biomechanical assessment application of measuring wrist passive stiffness and active ROM.
4.3 Device Characterization

Experimental validation of the SE-AssessWrist is performed to characterize its capabilities to estimate passive wrist stiffness and active ROM. These assessments necessitate implementation of low-velocity position and zero force control modes. The position and zero force control experiments presented in this section demonstrate the performance necessary for the intended assessment of wrist passive stiffness and active range of motion.

Based on previous passive wrist assessment [42], it was determined that the device must be able to position control its joints to follow low-velocity ramp position trajectories. For assessing active range of motion, the user backdrives the device. To achieve this, the device must be controlled through zero force control. In this assessment, the device should not adversely affect measured active range of motion due to excessive interaction torque between the device and user. Since no active wrist device has been used to measure active wrist range of motion, no benchmark for this interaction torque exists, but in this thesis the resulting interaction torque are compared to the torque required to backdrive other wrist exoskeletons. Additionally, as measurement of torque is an essential feature of the device, the stiffness of the spring is characterized through a custom testbed. Since movements made during zero force are low-acceleration, the relevant dynamics of the robot are its friction, which can effectively be reduced through zero force control, as demonstrated in this section.

| Table 4.4 : SE-AssessWrist Specifications |
|---------------------------------|---------------------------------|----------------|-----------------|-----------------|
| $\tau_{cont}$ [N·m] | $\tau_{max}$ [N·m] | $\tau_{res}$ [N·mm] | $[k_{FE} \ k_{RUD}]$ [N·m/rad] | $\theta_{res}$ [rad] |
| 4.9 | 5.1 | 12 | [75.96 77.23] | $1.57 \cdot 10^{-4}$ |
4.3.1 Control Hardware

Control of the SE-AssessWrist is performed through real-time software in a Matlab-Simulink Real-Time model communicating with Quanser’s Q8 USB data acquisition (DAQ) board, which is run at a sampling rate of 1000 Hz. Velocity estimates of encoder positions were obtained through the Q8’s built-in instantaneous velocity estimator, which runs at 100 MHz. Analog voltage commands from the Q8-USB were sent to servo amplifiers (Advanced Motion Controls AMC 12A8) which converted the voltage commands to current control the brushed DC motors.

4.3.2 Spring Stiffness

The stiffness of the two springs of the SE-AssessWrist were each characterized separately using the same experimental protocol. A custom testbed was designed for the purpose of measuring spring stiffness (see Fig. 4.9). The testbed consisted of a torque sensor, which was mechanically grounded on one side, while the other was rigidly connected to the outer race of the spring through a spring plate with the same bolt patterns of the load side pulley used in the final assembly. The inner race of the spring was connected to a shaft which was able to rotate freely through the use of a radial ball bearing. A handle was connected to the shaft to enable an experimenter to deflect the spring readily. An encoder mounted on the same part which housed the ball bearing, measured this input position. Since the other side of the spring was mechanically fixed, the corresponding position and torque measurements provided the data to determine the spring rate.

The torque sensor used in the testbed was Transducer Techniques TRT-50-in-lb. The sensor can measure torque up to 5.649 N·m, with a safe overload of 150%, a rated output of 2 mV/V, nonlinearity and hysteresis of 0.1% of the rated output, and nonrepeatability of 0.05% of the rated output, providing an accurate ground truth estimate of torque. The
torque sensor’s output voltage needed further amplification before being sent to the DAQ, and so was amplified through an inverting amplification circuit using an AD620 operational amplifier. The gain of the amplification was set to 100.45. The encoder used in the testbed was US Digital’s EM2 transmissive optical encoder module, which is the same encoder used in the final assembly. The encoder has 10,000 cnt/rev, which, with quadrature encoding, leads to a position resolution of $1.57 \cdot 10^{-4}$ rad.

An experimenter applied quasistatic loading, mean velocity of approximately 0.006 rad/s and maximum velocity of 0.04 rad/s in each experiment, on the spring to evaluate its spring stiffness. The experimenter applied up to 3 N·m of torque in each direction. The results of this experiment are presented in Fig. 4.10, which plots the torque vs. displacement for each spring separately, and on the same plot for visual inspection. As can be seen, the spring provides an accurate and linear estimate of torque. The resulting spring rates are very similar: $k_s = [k_{FE} \ k_{RUD}] = [75.96\ 77.23]$ N·m/rad for the FE and RUD springs, and matched the adjusted finite element analysis estimate (73 N·m/rad) closely. The exper-
Figure 4.10: Characterization of the custom-designed springs which are used in the (left) FE and (right) RUD joints. Both springs were found to have very similar spring rates: $k_s = [75.96 \ 77.23]$ N·m/rad with high coefficients of determination $> 0.9996$. Note that the choice of which spring was selected for each joint was arbitrary. Measured data were downsampled to 5 Hz for visualization.

iment for the FE spring demonstrated a coefficient of determination ($R^2$) of 0.9998 with an average error of 0.012 N·m and a maximum full scale output error of 2.06%, while for the RUD spring the coefficient of determination was 0.9996 with an average error of 0.03 N·m and a maximum full scale output error of 2.52%.

4.3.3 Position Control

Position control of the SE-AssessWrist is an important feature in that the device must be able to provide relatively constant velocity over a wide range of positions. Position control of the SE-AssessWrist is achieved through PD control with feed-forward torque compensation of static friction,

$$\tau_M = k_p(\theta_d - \theta) - k_d \dot{\theta} + \tau_{ff}$$

(4.2)

where $\tau_M$ is the motor torque, $\theta_d$ desired position, $\theta$ measured position, $\dot{\theta}$ measured velocity, $k_p$ is the proportional gain, $k_d$ is the derivative gain, and $\tau_{ff}$ the feed-forward torque.
The feed-forward torque was experimentally found by slowly increasing the torque command to the motor and recording the torque required to pull the cable at the output pulley through the Bowden cable transmission. Feed-forward compensation of friction was found to improve the initial portion of the position control trajectory, as oscillations introduced from backlash and friction were reduced.

Due to the use of two motors for a single-DOF, the module is over-actuated; however, since the cables only produce rotational motion of the output pulley when under tension, only one motor can move the output joint in a given direction. As such, for the constant velocity needed in the stiffness assessment experiments, each joint consisted of a leader and follower motor. The lead motor was commanded to follow the desired position control trajectory through PD control and feed-forward friction compensation (see Equation 4.2). On the other hand, the follower motor was sent a constant negative torque command, as found through experimentation, which provided sufficient current to maintain some slack in the cable. This resulted in the lead motor not having to overcome friction present in both Bowden cable transmissions, in addition to backdriving the follower motor.

The described position control approach is demonstrated through two experiments. In both experiments, \( \tau_{ff} = 0.3 \text{ N}\cdot\text{m} \), while the follower motor’s feed-forward torque is -0.1 N·m. The first experiment uses low PD gains, as in other passive wrist assessment studies [42, 66]. Compared with larger gains, using lower gains can reduce oscillations. A second set of experiments with higher gains is provided to illustrate the trade-off between position error and resulting velocity. Given that position control is used to assess passive wrist stiffness, which only requires segmenting the position and torque data over a movement, thus not being affected by errors in following the desired trajectory, low gain controllers are preferred.

In the first experiment, the device made movements to 12 targets spaced evenly in the
Figure 4.11: Position control performance of the FE joint to various ramp position commands with low control gains. The low gains result in steady state error, which is acceptable for the given application since the stiffness profiles in the validation profiles are simply segmented over the movement range. The ramp trajectories were chosen to replicate the methodology for assessing passive wrist stiffness in [42].

FE and RUD space, starting with wrist extension. Gains of $k_p = [10 \ 30]$ N-m/rad and $k_d = [0.1 \ 0.1]$ N-ms/rad were chosen specifically to be low as in [66] since they will be used in the demonstration study for generating less oscillatory velocity inputs (see Figs. 4.11 and 4.12). As can be seen, the device is able to track the desired position trajectory well, but with some steady state error. The average error over the experiment was $e_{\text{avg}} = [0.037 \ 0.021]$ rad with a maximum error of $e_{\text{max}} = [0.092 \ 0.048]$ rad, and as can be seen the velocity of the joints is relatively smooth.

To demonstrate that the device can achieve accurate position control, a second experiment with higher proportional gains, $k_p = [25 \ 40]$ N-m/rad, was performed (see Figs. 4.13 and 4.14). Note that $k_d$ was not changed since it is limited by the noise in the velocity signal. In this case, the joints are able to more accurately track the desired trajectory, but the velocity profiles are slightly more oscillatory. The average error over the experiment was $e_{\text{avg}} = [0.014 \ 0.014]$ rad with a maximum error of $e_{\text{max}} = [0.040 \ 0.036]$ rad.
Figure 4.12: Position control performance of the RUD joint to various ramp position commands with low control gains. The RUD joint requires a higher gain than the FE joint due to the presence of greater friction from the Bowden cable conduit system.

Figure 4.13: Position control performance of the FE joint to various ramp position commands with higher proportional gains. The higher proportional gain greatly reduces error, but also introduces much more oscillations into the joint velocity.
Figure 4.14: Position control performance of the RUD joint to various ramp position commands with higher proportional gains. The device is able to track the desired RUD position signal more closely, but introduces more oscillations.

4.3.4 Zero Force Control

In addition to position control, the SE-AssessWrist must also be able to perform zero force control, a specific subset of force control in which the desired spring force is always zero. Through zero force control, a non-backdrivable actuator, such as the SE-AssessWrist, can render a transparent environment so that the user can backdrive the device while the device measures user position. This is important for the case of active ROM, since the user must be able to backdrive the device, with ideally minimal effort. Although, zero force control capabilities for such an experiment are not as stringent as those which, for example, evaluate the velocity of point-to-point movements.

Due to the high static friction as a result of the Bowden cable transmission, to achieve zero force control, a controller which leverages the capabilities of the device to perform actuator position control was chosen [150] (see Fig. 4.15). In the force control approach presented in [150], to regulate torque, the motor attempts to control deflection of the spring through position control of the spring’s input position. As in all series elastic devices, the module cannot regulate arbitrarily low torque since it has a practical lower bound based on
the torque resolution of the spring. Additionally, to overcome backlash, the device’s default state in this control mode is to provide tension on both sides of the spring such that the user can create a torque to inform the controller to perform active zero force control. Once a deadzone limit is exceeded, the zero force controller, shown in Fig. 4.15, is used.

To illustrate the effectiveness of this control approach, an experimenter moved the device around at a pace similar to that expected during the ROM portion of the validation study while going to the limits of their ROM to 12 targets as in the position control experiment. The zero force controller consisted of PD gains of $k_p = [175 \ 225]$ N·m/rad and $k_d = [0.05 \ 0.05]$ N·ms/rad, as well as a deadzone of $\tau = [0.15 \ 0.15]$ N·m. The results of this experiment are shown in Fig. 4.17. Due to the zero force control, the user was able to actively backdrive the device to find their ROM limits, verifying the device’s capability for wrist ROM assessment. Additionally, the spring torque during the experiment was low with $\tau_{\text{avg}} = [0.082 \ 0.067]$ N·m and a maximum absolute torque of $\tau_{\text{max}} = [0.38 \ 0.29]$ N·m. Considering that just considering the static friction of most wrist exoskeletons is on the order of 0.1 N·m [14, 71, 83], the interaction torques here are acceptable for the intended application of backdriving the robot at a modest pace to find active ROM.

### 4.3.5 Range of Motion

To characterize the SE-AssessWrist’s ROM, an experimenter manipulated the device while it was commanded with the the zero force controller presented in Section . The user moved the device to its flexion limit, and then moved the handle to its limit ulnar deviation limit. In this configuration, the device was moved to its extension limit, where the RUD joint was then moved to its radial deviation limit, before moving to its radial deviation limit. The results of this experiment are shown in Fig. 4.16. With reference to the ROM used in previous wrist assessments, the SE-AssessWrist provides much greater ROM capabilities,
Figure 4.15: Force control scheme selected for the SE-AssessWrist during ROM assessment. A desired torque $\tau_{s,d}$ is selected, and is converted into a desired spring deflection through the spring rate ($k_s$) and adding the position of the output side of the spring ($\theta_{s,o}$). This desired position is compared to the measured input side spring position ($\theta_{s,i}$) which is directly controlled by the motor through the position controller. During ROM assessment, the desired torque is zero, which results in the corresponding motor regulating $\theta_{s,i}$ to match $\theta_{s,o}$. This block diagram was adapted from [150].

while also increasing torque capabilities. The ROM limits of the device were originally chosen based on those found in [68, 200, 201], and as can be seen the device meets the objective.

4.4 Demonstration Study

A demonstration study was conducted to evaluate using the SE-AssessWrist to measure active wrist ROM and passive directional stiffness in five able-bodied individuals for wrist directions of FE, RUD, and several combinations of the two.

4.4.1 Methods

The same experimental control hardware presented in Section 4.3 was used in this study. Additionally, the document used to administer the protocol is provided in Appendix D. Note that where applicable, portions of the protocol described in this section are based off of previous studies evaluating passive wrist stiffness [42, 66, 68].
Figure 4.16: Comparison of the SE-AssessWrist range of motion to that found in [68], used during [66], and of an experimenter.

Figure 4.17: Zero force control characterization experiment. (left) A user operated the device while connected to determine their wrist ROM. During the experiment, the device was able to modulate the transparency torque well for the (center) FE and (right) RUD joints. Note that the main effect of the controller is to simulate static friction similar to levels observed in other wrist exoskeletons, despite the highly nonlinear force transmission scheme used here.
Participants

Five participants (1 female, 4 male) with an age range of 22-32 years old ($\mu = 26.4$, $\sigma = 4.16$) participated in the experiment. All participants were right hand dominant with no current injury or known history of neuromuscular injury in their wrist. Approval for the experiment was obtained through the Rice University Institutional Review Board (IRB-FY2018-331). The consent form is provided in Appendix D.

Measuring Wrist Muscle Activity

As in previous works [42, 66], recordings of muscle activity through sEMG were used as a means to determine if the user was being non-passive. By having a measure of sEMG, how passive the participant is being while the robot moves the wrist can be determined. Since if the user is not careful about regulating wrist activity, stiffness much greater than the passive stiffness can be observed. As such, sEMG can be used as a check in such situations. Previous studies found that sEMG recordings were generally very low (< 5% of maximum voluntary contraction) during assessment, which is a desirable result. Note, like in related studies [42, 66], the sEMG recordings were not collected for any additional purposes other than to provide a means of identifying any potential non-passive movements, although a rigorous definition of what such sEMG levels correspond to a non-passive movement has not been presented, and is left for future work.

Muscle activity relating to wrist FE and RUD were recorded through four sEMG electrodes. As in [66, 226], the four muscles targeted for the wrist were: flexor carpi radialis (FCR, related to flexion), extensor carpi ulnaris (ECU, related to extension), flexor carpi ulnaris (FCU, related to ulnar deviation), and extensor carpi radialis longus (ECR, related to radial deviation). Representative locations for placement of the electrodes on a participant’s forearm are shown in Fig. 4.18. Although these muscles are not purely used for
the movement directions listed, since the wrist is an overactuated joint, these muscles are located near the surface of the forearm and so provide a good measure of activity for wrist movements.

Measurements from sEMG are typically very low voltage, and need to be amplified before sending to the DAQ. The sEMG electrodes chosen for this study are the Motion Lab Systems MA-411 with a x20 pre-amplification. The 20x pre-amplification occurs at the electrode, providing a better signal to noise compared to not pre-amplifying since noise in the low voltage line can accrue between the electrode and DAQ. In addition, the sEMG signals were processed through analog circuitry including a bandpass filter and further amplification before being sent to the DAQ. The signals from the sEMG electrodes were further processed digitally through a first order high-pass Butterworth filter with a 20 Hz cutoff frequency, a rectifier, and a first-order low-pass Butterworth filter with a 2 Hz cutoff frequency, similar to [66, 87, 227]. For more information regarding the sEMG electrodes, and the corresponding analog signal processing, see Appendix E.

The first step in the experiment was to affix the sEMG electrodes to the participant (see Fig. 4.18). Before placement of the electrodes, the skin and electrodes were cleaned.
to ensure integrity in signal acquisition. Since sEMG is a relative measure, maximum voluntary contraction (MVC) was performed to provide a reference for sEMG activity obtained during the stiffness experiment discussed in Section 4.4.1. As in [66], participants performed 3 measures of MVC for each movement direction, and the maximum activity for each channel during this experiment was recorded. In the wrist stiffness experiment, the corresponding sEMG activity is normalized by the maximum signal from the MVCs performed during this experiment. This portion of the experiment took 10-15 minutes.

**Wrist Range of Motion Assessment**

The second part of the experiment measured maximum active wrist ROM. This portion of the experiment was performed for the dual purpose of evaluating wrist ROM, and using these ROM measurements to select an appropriate region over which to estimate wrist stiffness in the wrist stiffness portion of the experiment. The robot was operated under the zero force control scheme demonstrated in Section 4.3.5 with gains of $k_p = [175 \ 225]$ N·m/rad and $k_d = [0.05 \ 0.05]$ N·ms/rad, as well as a deadzone of $\tau = 0.15$ N·m.

Participants’ active ROM, as opposed to passive ROM in which a therapist or the robot moves the limb, was evaluated for 24 movement directions. During the experiment, participants sat with a posture consisting of moderate shoulder flexion, shoulder abduction, elbow flexion, and a neutral forearm orientation. The forearm was secured through a cuff that could be comfortably compressed around the forearm. The neutral orientation of the forearm was defined visually with the top of radial styloid in line with the device’s FE axis of rotation. The neutral orientation of the wrist was defined with respect to having an open grasp. As in [68], neutral wrist orientation was defined visually by aligning the dorsal surfaces of the forearm and hand until flush (FE neutral), and then aligning the the third metacarpal’s long axis was parallel to the forearm’s long axis (FE neutral) similar to [228].
Images of a participant near the neutral position are shown in Appendix E.

To find wrist ROM, each participant was asked to actively move the farthest they could in a set of 24 directions. Angles chosen aligned with the traditional anatomical FE/RUD axes, and directions spaced 15° apart. Participants repeated the 24 directions 3 times with the directions being presented in three blocks. Within each block, the movement directions were presented in random order. This approach was used as opposed to a set order, or true random order, since it would mitigate any potential learning of use of the apparatus or stretching to not bias any particular directional ROM.

To assist with finding active ROM, during the experiment, participants viewed a virtual display consisting of a cursor identifying the user’s 2D position \( \boldsymbol{\theta} = [\theta_{FE}, \theta_{RUD}] \), and a line with angle \( \phi \) in the direction of the desired movement (see Fig. 4.19). The line was used so as not to bias participants’ ROM attempts by not suggesting a ROM to the user. With respect to the visual display, the angle \( \phi \) was defined as \( \text{atan2}\{\theta_{RUD}, \theta_{FE}\} \), where positive was defined as being in the wrist extension and radial deviation directions for ease of visual interpretation. Participants were asked to move in the direction of the line as far as they comfortably could. Prior to data collection, if needed, participants were allowed to practice a few trials to become familiar with the setup. The entire active ROM experiment typically took 10 minutes.
Wrist Stiffness Assessment

The last portion of the experiment was performed to evaluate passive wrist stiffness. To accomplish this, the robot moved the participant’s wrist while they attempted to remain passive, i.e., attempt to not interact or invoke muscle activity. Many of the procedures in this experiment are similar to the seminal work in wrist stiffness presented in [42, 66, 68].

The wrist was moved in 24 equally spaced wrist directions starting from wrist extension and moving counterclockwise as is standard in the literature [42, 66, 68]. Each movement consisted of an outbound (neutral to outer target) and inbound (outer target to neutral) component, and each movement was repeated three times prior to moving to the next target. Unlike previous experiments, which either used torque control due to the use of cadavers [68], or limited ROM, as a result of limited device ROM [42, 66], the targets in this study custom-tailored to each participant, using the ROM found in the previous portion of the experiment (see Section 4.4.1). Due to the soft gains employed, which resulted in less oscillatory motion than higher gains used during position control experiments in Section 4.3.3, the final target positions were generally undershot by a few degrees, as in related work [42, 66]. Participants were instructed to relax their wrist during the task. Since it can be counterintuitive to remain passive while a limb is being moved, participants were given a familiarization trial (approximately 2-3 minutes) prior to data collection. In addition, a follow-up experiment with sEMG provided as bio-feedback was also carried out. The passive wrist stiffness portion of the experiment took 35-40 minutes.

The robot was commanded to follow a ramp position trajectory (constant velocity) with a velocity magnitude of 0.2 rad/s so as to assist with avoiding muscle activity due to stretch reflex. Proportional derivative gains of $k_p = [15 \ 30]$ N·m/rad and $k_d = [0.1 \ 0.1]$ N·ms/rad were used with the same control scheme as discussed in Section 4.3.3. However, the experiments in Section 4.3.3 were performed in an un-loaded condition. During the experiment,
when the robot moves inbound, the user’s wrist will provide a restoring torque in the direction of movement which can result in unwanted oscillations. As a result, the feed-forward torque ($\tau_f$) was reduced to 0.1 N·m, which was experimentally found to reduce these oscillations. Note that while collected in the experiment, the analysis of these inbound movements is left for future work as they were not evaluated in [68], and were found to be similar to outbound movements in [42, 66].

After the experiment, each movement was segmented, removing movements until the position magnitude was greater than 3° to eliminate any short-range stiffness effects, or oscillations introduced at startup. Additionally, the movement segments were stopped when the reference position reached its final position to avoid collecting any data after the robot stopped moving. Finally, any position dependent torques, due to device gravity, encoder offsets, or conduit flex, were subtracted from the measured joint torques, similar to what was done in [42, 66]. From the segmented data, the linear stiffness of the wrist in that direction was calculated using multiple-linear regression for outbound movements. No down sampling or filtering of the signals was performed. Finally, during the experiment, software limits were set to ensure that the robot would not apply excessive joint torque, displacement, or velocity. Additional safety measures were taken through setting limits on motor position, motor velocity, command current, and the inclusion of a stop button.

4.4.2 Results

Maximum Voluntary Contraction

Participant’s EMG signals acquired in the MVC task are presented in Fig. 4.20. The mean, standard deviation, and range of maximum muscle activity found during MVC are presented in Table 4.5. Note that these measurements are unitless and given the identification of the arbitrary units of “muscle activity,” as they only provide a relative signal. As in
previous work [42, 66], these measurements were obtained to provide a reference accounting for the variability in sEMG recordings between participants, as a result of participant to participant variability. By normalizing sEMG recordings in the stiffness portion of the experiment by the MVC, an estimate of user passivity is obtained.

Table 4.5 : MVC Measurements

<table>
<thead>
<tr>
<th>Muscle</th>
<th>$\mu$</th>
<th>$\sigma$</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>FCR</td>
<td>3.29</td>
<td>2.29</td>
<td>1.63-7.32</td>
</tr>
<tr>
<td>ECU</td>
<td>3.21</td>
<td>1.74</td>
<td>1.45-5.38</td>
</tr>
<tr>
<td>FCU</td>
<td>1.81</td>
<td>0.67</td>
<td>1.27-2.97</td>
</tr>
<tr>
<td>ECR</td>
<td>2.87</td>
<td>0.85</td>
<td>1.47-3.57</td>
</tr>
</tbody>
</table>
Range of Motion

From the active range of motion measurements, participants’ maximum values were calculated for each direction of movement. This set of data were analyzed together to calculate group mean and standard deviation for every movement direction. With reference to previous work analyzing passive wrist ROM in cadavers [68], the directions of most interest were those of the anatomical axes and axes of greatest ROM. These values are shown in Table 4.6, and in a 2-DOF wrist ROM envelope plot in Fig. 4.21.

With reference to the passive ROM measurements in pure flexion of 65.2°, extension of 64.2°, ulnar deviation of 42.3°, and radial deviation of 20.5° [68], the active ROM measurements in this thesis are similar with flexion of 69.8°, extension of 53.6°, ulnar deviation of 48.0°, and radial deviation of 30.1°. Additionally, in [68], the authors calculate the direction of greatest ROM with a flexion component (26.6°) and with an extension component (15°). The direction found in this study with a flexion component was similar (28.9°); however, the average across participants for the direction with an extension component was different (-10°). Overall, considering the sample size in this study ($n = 5$) and in [68] ($n = 6$ cadavers), the general agreement is encouraging. Lastly, the average completion time for the active ROM experiment was 5 minutes and 56 s, which considering the comprehensive

<table>
<thead>
<tr>
<th>Direction</th>
<th>$\mu$ [deg]</th>
<th>$\sigma$ [deg]</th>
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</thead>
<tbody>
<tr>
<td>Flexion</td>
<td>69.8</td>
<td>17.3</td>
</tr>
<tr>
<td>Extension</td>
<td>53.6</td>
<td>8.4</td>
</tr>
<tr>
<td>Ulnar Deviation</td>
<td>48.0</td>
<td>5.4</td>
</tr>
<tr>
<td>Radial Deviation</td>
<td>30.1</td>
<td>6.5</td>
</tr>
</tbody>
</table>
Figure 4.21: Results from the ROM experiment. The dots are the maximum values for each direction over all participants. The solid line is the mean maximum values across participants for a given direction. As found in a study with cadavers [68], the ROM plot is ellipsoidal in shape and oriented obliquely to the anatomical axes.

ROM results obtained, is a rapid assessment of wrist ROM.

**Stiffness**

The main objective of the stiffness experiment was to determine user’s 2-DOF wrist envelopes, anatomical wrist stiffness, and identify the directions of least stiffness as in [66, 68]. In the stiffness experiment, participants underwent 2 conditions while the robot slowly moved the wrist to 24 evenly spaced targets, 3 times to each target, in the user’s 2-DOF wrist ROM. From the stiffness measurements, participants’ stiffness for each of the 24 directions was calculated as the average of the 3 trials within a given condition. Stiffness was calculated using only the torque contribution in line with the movement [66], thus representing the restoring stiffness for each of the 24 directions. Compared with these previous studies, this thesis presents the first instance of performing both in vivo passive stiffness and active range of motion assessment with the same device.
An important component of the wrist passive stiffness experiment is for the participant to remain passive. Previous work [66] has used sEMG as a measure of passivity, but has not provided any rigorous definition as to how sEMG might affect stiffness measurements. From pilot work, it was identified that viewing sEMG activity might assist with maintaining low levels of muscle activity. Thus, to assist with maintaining passivity, this thesis employed a novel protocol in which user’s underwent two wrist stiffness conditions. As in related work [42, 66], in the first condition, the user was instructed to relax and attempt to be passive as the robot moved them. In the second condition, the user was given the same instructions but were also shown throughout the condition their sEMG activity through a visual display. Participants were instructed to attempt to regulate these measurements of muscle activity to be as low as possible. So as to standardize the feedback, the vertical axis was set to a constant 0.5 [unitless muscle activity].

As a measure of the difference in performance in regulating sEMG activity between the two conditions, the mean muscle activity for a given participant during analyzed movements was computed (see Table 4.7). Additionally, the fidelity of stiffness estimation from the two conditions was analyzed as the mean coefficient of determination ($R^2$) in estimating user stiffness for these movements (see Table 4.8). As can be seen from these tables, comparing the first condition to the second, user’s were in general able to down-regulate their sEMG activity. Correspondingly, the second condition resulted in higher-fidelity measurements of wrist stiffness. These results provide some evidence that providing sEMG as bio-feedback could be useful for assisting individuals with regulating low muscle activity during passive stiffness experiments, and that such bio-feedback might also lead to higher-fidelity passive stiffness estimations. Given this result, the remaining presented stiffness measurements will be from data acquired in the second condition.

The main result of the study, the 2-DOF stiffness envelope, is shown in Fig. 4.22. As
can be seen, the orientation of the stiffness envelopes is oblique to the anatomical axes, an observation originally found in a cadaver study [68]. The direction of least stiffness found in this experiment for a movement with a flexion component was $30^\circ$ from pure flexion in the direction of ulnar deviation. Additionally, the direction of least stiffness with an extension and radial deviation component was $17.7^\circ$ from pure extension in the direction of radial deviation. These measurements correspond closely with the $21.7^\circ$ axis of least stiffness found in [68].
Figure 4.22: Main result from the passive wrist stiffness experiment. In the plot, the dots correspond to the average directional stiffness over all participants. The solid line is the mean of these mean stiffness values across participants for a given direction. As found in a study with cadavers [68], the stiffness is ellipsoidal in shape and oriented obliquely to the anatomical axes.

The corresponding mean and standard deviation of the anatomical stiffness values obtained from the experiment are presented in Table 4.9. As stiffness varies with arm shape, muscle strength, and other properties, stiffness values will vary between participants. Additionally, considering the different mechanisms used to measure stiffness, for example, other studies might have used a grip handle, used a 1-DOF device, or have slight offsets in neutral configuration, making comparison not a demonstration of validity. However, it is encouraging that the stiffness estimates obtained in this work are in the same range as other studies, which have for example estimated wrist FE stiffness to be between 0.15-3 N·m/rad and for RUD stiffness to be between 1.45-1.63 N·m/rad as estimated in 1-DOF wrist stiffness studies [42, 195, 229, 230]. Comparison to the only other study [66] using a robot [71] for in vivo 2-DOF passive wrist stiffness measurements, the estimates of user wrist stiffness found in [66] are higher (> 0.8 N·m/rad and > 2 N·m/rad for RUD) than the estimates in this thesis. However, unlike the other study, using the SE-AssessWrist enables
measurement of not only passive wrist stiffness, but also active ROM with the same device, which was previously only demonstrated in a study with cadavers [68].

4.5 Conclusion

In this chapter, the design and fabrication of a serial 2DOF series elastic actuated Bowden cable-based device, the SE-AssessWrist, was presented. The device was designed to interact and measure wrist flexion/extension (FE) and radial/ulnar deviation (RUD) for the biomechanical assessments of measuring passive wrist stiffness and active range of motion (ROM). To perform these assessments, it was determined that the device should 1) measure torque directly, 2) output sufficient torque, and 3) allow complete wrist ROM. These objectives were accomplished through implementation of a Bowden cable-based series elastic actuation architecture. The incorporation of series elastic actuators (SEAs) accomplished the objective of direct torque measurement, while the combination of SEAs and a Bowden cable transmission provided the appropriate torque and ROM capabilities. These objects were accomplished while achieving accurate position control for the stiffness assessment and the capability to provide accurate zero torque for the range of motion assessment. In this chapter, the capabilities of the device were evaluated, including description of device capabilities, characterization of the spring, control experiments, and a validation study.

As presented in Table 4.1, the device has enough ROM capabilities to measure complete ROM, and has greater continuous torque capabilities than previous wrist exoskeletons. In addition, the spring designed for the SE-AssessWrist was characterized, exhibiting highly linear behavior ($R^2 >= 0.9996$). Demonstration of the performance capabilities of the device, as necessary for the assessment application, were presented in two characterization experiments. The first was demonstration of position control performance of the SE-AssessWrist with both low and high gains, demonstrating its capability to accurately
tracking a position trajectory despite the Bowden cable transmission. Additionally, since
the device is non-backdrivable, it needs an active transparent mode to assess wrist range
of motion. This was accomplished through the adoption of a zero force controller with an
inner position control loop [150]. In this way, the device was able to accurately regulate
the spring to within a small deadzone, 0.15 N·m, similar to the friction found in other wrist
exoskeletons [14, 71, 83].

To validate the robotic hardware and experimental hardware, five right-hand dominant,
able-bodied individuals were recruited for the study. As such, this validation demonstrates
the capabilities of the SE-AssessWrist, but the results are not at all meant to provide any
biomechanical findings due to the limited number of participants. The objective of the
first portion of the experiment was to measure active wrist ROM. The participants actively
found the limits of their ROM, while the robot was backdriven using the zero force control
scheme. ROM envelopes were plotted, and were observed to be oblique, i.e., the direc-
tions of maximum ROM were not aligned with the anatomical axes of the wrist, which is
consistent with an observation made in a cadaver study [68].

The 2nd objective was to determine passive wrist stiffness. This was accomplished by
commanding the robot to follow a ramp position trajectory while instructing the participant
to remain passive. A constant velocity trajectory was used since it has the benefit of re-
moving any inertial wrist torques, and constant viscous torques do not affect linear stiffness
measurements since stiffness is a differential measurement. To monitor user passivity, sur-
face electromyography (sEMG) was employed to measure muscle activity during the pas-
sive wrist stiffness measurement. Wrist stiffness was also observed to be oriented obliquely
to the wrist’s anatomical axes, and relatively perpendicular to the ROM envelopes.

To the best of the author’s knowledge, the SE-AssessWrist provides the first instance of
an active wrist exoskeleton being used to measure wrist ROM, something that was previ-
ously only possible in cadavers [68], although in that case only passive ROM was measured. With this capability, the SE-AssessWrist not only measures an important parameter related to activities of daily living, but can use the ROM test, as done in this thesis, to develop a user-specific stiffness experiment based upon their own ROM. In this way, more samples can be obtained to determine user stiffness, potentially providing a more reliable estimate. Additionally, although not done in this proof-of-concept work, the SE-AssessWrist can be used to examine end-point stiffness, i.e., stiffness at the workspace limits, in addition to the linear directional stiffness estimates found in this study.

The SE-AssessWrist provides a non-traditional compliant approach to the design of a wrist exoskeleton. This was in part because the SE-AssessWrist was designed specifically for wrist biomechanical assessment, compared with other devices which are designed for rehabilitation. In addition to the performance benefits previously mentioned, adoption of the Bowden cable-based series elastic actuation include the need for only a half rotation of cable wrapping, which is much easier to assemble than the complex multiple-wraps and pre-tensioned cable wrapping needed in traditional capstan arcs. Using computer numerical control (CNC) machining to create the spring was found to be beneficial, as compared with the standard approach of wire electrical discharge machining (WEDM). By using a CNC, the spring hub could be directly integrated into the spring’s design, and currently using a CNC is much more cost effective than WEDM. Additionally, since most similar springs were designed for gait training exoskeletons, the SE-AssessWrist spring uses aluminum for its softer modulus. Due to the specific use of aluminum 7075-T651, a spring with a high yield torque was still achievable.

In regards to the Bowden cable actuation, it enabled larger torque output without the need for locating heavy and bulky direct current (DC) motors on the device. Additionally, the device achieved higher ROM without sacrificing compactness due to the Bowden cable
transmission. Increasing torque and ROM would be the opposite for a traditional haptic device, which would need to include bulky motors on board as well as large, possibly infeasible, capstans to achieve the same ROM achieved here. A larger torque requirement was set since previous studies reported that 1.95 Nm of torque was insufficient in some instances for exploring a small ROM [42, 66]. Additionally, ROM was targeted to achieve a more complete picture of wrist biomechanical properties: both with estimation of wrist ROM and stiffness. Measuring ROM enables identification of the wrist’s primary axis: the so-called dart thrower’s motion which is a motion between FE and RUD in the direction of radial-extension to ulnar-flexion. The general shape and orientation of the stiffness and ROM envelopes were not aligned with the anatomical axes, and were similar in shape to that presented in a similar study which used cadavers [68].

The SE-AssessWrist is a promising tool for biomechanics efforts to better understand the passive and active properties of the wrist. In the future, this device will be used in experiments with able-bodied individuals to develop a database of ROM and stiffness. This database will serve as a reference for further experiments with neurologically impaired individuals. Using the device to study wrist biomechanical impairment after neurological injury, such as stroke, could reveal insights into the recovery process of ROM and stiffness, as well as serve as a tool for evaluating spasticity. Through such assessment, the device can be used to help close the loop on robotic rehabilitation to enable patient-specific robotic therapy after neurological injury.

The capabilities of the SE-AssessWrist to measured both wrist passive stiffness and active range of motion are due to its use of series elastic actuation and a Bowden cable transmission. The series elastic actuators and Bowden cable transmission compliment each other, with leveraging Bowden cables creating a design with more range of motion than previous wrist exoskeletons, and the series elastic actuators providing measurement of torque
and enabling zero force control. In this thesis, a demonstration of the developed apparatus was presented, providing an example of how series elastic actuation can benefit robot-aided assessments.
Chapter 5

Conclusions

This thesis adopted the series elastic actuation scheme, traditionally used for lower-limb robots due to the benefits of shock tolerance, human-friendly actuation, and possibility for inertia reduction, for the design of two novel assessment robots. In both assessment robots designed and evaluated in this thesis, the incorporation of series elastic actuators (SEAs) provided meaningful performance gains with respect to previous devices which are used, but use the traditional rigid rehabilitation robot design paradigm, for these assessments. Specifically, the incorporation of SEAs led to the creation of a high-fidelity haptic device during functional magnetic resonance imaging (fMRI), which has not been previously demonstrated during fMRI for a multi-degree of freedom (DOF) device. Additionally, the incorporation of SEAs led to the creation of a wrist exoskeleton for robot-aided assessment with direct torque measurement in addition to more range of motion and torque output than any previous wrist exoskeleton. These performance gains were made possible through SEAs, since the devices could use nonbackdrivable (i.e., high friction) actuators and transmissions, not allowable in traditional open-loop torque controlled devices.

The two series elastic actuated robots design and evaluated in this thesis are demonstrated in applications of neurological and biomechanical robot-aided assessment. These demonstrations provide the first instance of using series elastic actuators to specifically design a robot for these assessments. The current robots designed for such assessments are the same robots, or use the same design principles, as those that currently provide rehabilitation, resulting in performance limitations for robot-aided assessment. The design
limitations, i.e., the need to use rigid, backdrivable actuators for open-loop torque control, imposed on these rehabilitation robots are addressed in the two series elastic actuated assessment robots presented for in this thesis.

Most rehabilitation robots use rigid (i.e., no springs), backdrivable (i.e., low friction) actuators and transmissions, which is generally accomplished through the use of direct current (DC) motors with torque amplification through backlash-free pulley transmissions. Due to the backdrivability of these devices, they lend themselves well to impedance control, where the interaction between the robot and user is modulated by measuring user position and outputting torque in an open-loop manner, which results in inaccuracies due to unmodeled device dynamics. In contrast to rigid haptic interfaces, a series elastic actuated device uses a design in which an elastic element is intentionally placed in series between the actuator and user. Direct and accurate torque control is achieved through measurement of the elastic element’s deflection.

A more thorough discussion of the design benefits and trade-offs of using SEAs in relation to rehabilitation robot design paradigm is discussed in Chapter 2 of this thesis. One such further benefit of SEAs is that, compared with a rigid device, they increase user comfort since the actuator’s dynamics are decoupled through the elastic element. Additionally, in applications where humans interact closely with robots in the industrial floor, SEAs provide a safety measure as a result of this decoupling. Although SEAs provide many benefits, a common limitation is that their upper bound on passively (i.e., not injecting energy into the system) displaying a rigid environment to the user is the spring constant of the elastic element. However, through the use of the TDPA, by systematically monitoring the flow of interaction energy between the robot and user, energy can be timely injected, while still maintaining global-time passivity. This is accomplished since energy is not injected unless previous energy has been dissipated. This approach provides the ability to display
much more rigid environments, while maintaining passivity, which is important for safety considerations, with the same SEA. Using softer springs maximizes the benefits of SEAs: safe, compliant interaction with the user through the decoupling of device dynamics, and a softer spring results in greater force control accuracy due to higher torque measurement resolution and the ability to use higher control gains.

In Chapter 3 of this thesis, the importance of SEAs in the application of robot-aided neurological assessment is demonstrated. The ultimate objective of neurological assessment is to discover the underlying neurological processes, in response to a meaningful stimulus such as robotic rehabilitation, which promote brain plasticity to improve motor control in neurologically impaired individuals. It is widely believed that brain plasticity is essential to recovery, but the answers to questions such as to what extent and how to best promote it through robotic rehabilitation are not known. To evaluate brain plasticity through functional magnetic resonance imaging (fMRI) during robotic interactions, the MR-SoftWrist was developed for this thesis. The MR-SoftWrist is a 3DOF device which can interact with the user’s wrist in flexion/extension (FE) and radial/ulnar deviation (RUD). The third linear DOF of the device is used for alignment of the user’s wrist axes with those of the device. Since the MRI scanner uses a large magnet, traditional DC motors are not suitable as a result of their ferrous magnetic properties, which, if not properly grounded, could result in the DC motors being attracted into the scanner, posing a danger to the user due to them motors being pulled into the scanner at high velocity. As a result, the MR-SoftWrist uses non-ferrous ultrasonic motors; however, these motors are nonbackdrivable and have a low-velocity deadband. To overcome the limitations introduced by the ultrasonic motors, SEAs were incorporated into the design. In this way direct torque control was achieved through measurement of spring deflection, enabling rich haptic interactions with the user despite the use of nonbackdrivable motors.
Chapter 3 provides analysis of the design decisions regarding structural properties of the MR-SoftWrist to achieve the necessary range of motion and torque output. Additionally, several experiments demonstrate the accuracy and efficacy of the SEA torque control. These experiments included validating torque measurement accuracy with a ground truth torque sensor and validating the capability to provide accurate impedance control through a virtual stiffness control experiment. MR-compatibility of the device, i.e., demonstrating that the device’s electromagnetic properties do not interfere with scanned images, has been demonstrated in related works and is discussed in the Appendices. To validate the application of the MR-SoftWrist for robot-aided neurological assessment to assess brain activity, and possibly brain plasticity, in response to rehabilitation, an interactive wrist movement task was performed with a single able-bodied individual during fMRI. The user interacted with the robot in three control modes while making circular arc pointing movements. The three control modes required usual (zero torque control), minimal (path control), and maximum (error augmentation) effort to stay on the circular path. It was found that the device could elicit both distinguishable kinematics and brain activations from the user depending on the three control modes employed, validating the MR-SoftWrist as a viable tool for systematically studying brain activity in response to a haptic interaction during fMRI.

In Chapter 4 of this thesis, the importance of SEAs in a case study of robot-aided biomechanical assessment is demonstrated. Such a device was created since biomechanical factors are important to study during recovery from neurological injury since neurological injury can lead to reduced joint range of motion and increased stiffness as a result of muscle tone or spasticity. Having a device which can accurately estimate these factors for the wrist could assist in tracking the return of function and how these two related factors interact. Specifically in this thesis, the wrist was chosen for biomechanical assessment due to its importance in performing activities of daily living. To more thoroughly study both
properties with the same device, sufficient torque output and range of motion of the device are needed and current devices do not meet these requirements. Additionally, previous devices use only an estimate of torque to estimate user stiffness, a limitation which will lead to errors in stiffness estimation. This limitation is overcome through the incorporation of SEAs in the design of the SE-AssessWrist, which can measure torque directly, while also having greater range of motion and torque capabilities than previous exoskeletons. The SE-AssessWrist is a 2DOF wrist exoskeleton for assessing wrist range of motion and stiffness in flexion/extension (FE) and radial/ulnar deviation (RUD).

Chapter 4 provides analysis on the design decisions relating to the actuators and transmission of the SE-AssessWrist, which were selected to achieve the desired range of motion and torque output. Additionally, the application of the SE-AssessWrist for robot-aided biomechanical assessment was validated in a study with three able-bodied individuals. Individuals first donned four surface electromyography electrodes to measure important muscles of wrist activity and performed maximum voluntary contracture to serve as a reference for baseline activity. The participants then donned the SE-AssessWrist, through an open hand interface, to find their maximum range of motion by moving the wrist in 12 different directions while the SE-AssessWrist was operated under the zero torque control scheme. In the last portion of the experiment, participants attempted to remain passive while the SE-AssessWrist was operated in position control at low-velocity to measure passive wrist stiffness. The range of motion and stiffness envelopes were observed to be oriented not in the direction of the anatomical axes, an axis which might have important implications for rehabilitation efforts. This experiment demonstrated the potential of the SE-AssessWrist to assess wrist biomechanics with a range of motion only previously possible in work with cadavers. In conjunction with rehabilitation efforts, the SE-AssessWrist can next be used in a study with a larger number of able-bodied individuals to develop a database of normative
range of motion and stiffness properties.

In addition to applications in rehabilitation, the contributions of this thesis have implications in many other areas. For the case of the time domain passivity based approach to control (TDPA) of series elastic actuators (SEAs), designers of SEAs can now increase the perceived rigidity of the actuators, while still maintaining the advantages of having a compliant actuator. Advantages of a compliant actuator include human-friendly interaction, whereby if the human and robot impact, the springs provide a buffer between the rigid actuators. Additionally, being able to use a softer spring, while still being able to display rigid environments, leads to increased torque resolution, since softer springs lead to higher resolution in torque measurement, and improved force control accuracy. Advancements to control and design of SEAs is vital to the field of interactive robotics, since humans and robots are beginning to interact more closely than ever with robots: both on the industrial floor and in the home. As this trend continues, more of these robots might incorporate SEAs due to their human-friendly compliant approach to actuation.

In the future, robot-aided assessments, such as those presented in this thesis, will be essential to enhance rehabilitation outcomes after neurological injury. Currently, gains observed during robotic rehabilitation after neurological injury are mixed as a result of the lack of understanding of how recovery occurs. Devices such as the MR-SoftWrist and the SE-AssessWrist can be used to provide robot-aided assessment during this robotic neurorehabilitation. By including descriptive and objective assessments, how an individual responds to therapy can be carefully measured. After obtaining a large enough database in both the able-bodied and impaired population, patient-specific therapies might be created to increase the efficacy of rehabilitation. Additionally, these objective assessments can be used to provide further evidence of recovery which other standard clinical assessment scales might not show. This evidence-based recovery could be used to increase efficacy
in current rehabilitation trials by providing the high-resolution evidence-based assessments necessary to guide rehabilitation trials. By creating patient-specific rehabilitation therapies, further recovery in the same rehabilitation window might be observed.

In summary, this thesis presents two design implementations of series elastic actuators for assessment robotic devices and a novel time-domain approach to control series elastic actuators. Regarding robot-aided neurological assessment, sensorimotor control experiments with a haptic interface were previously only possible in the laboratory environment. However, through the application and control validation of high-fidelity haptic interaction with the MR-compatible MR-SoftWrist, such studies can now be conducted during fMRI. In this way the MR-SoftWrist can be used as a tool for neuroscientists to further our understanding of how specific components of the brain are responsible for various aspects of sensorimotor control. As for robot-aided biomechanical assessment, combined wrist stiffness and range of motion envelopes have only previously been studied in a study with cadavers. Now these biomechanical properties can be determined with the SE-AssessWrist in vivo. Findings from such studies might be used to advance biomechanical models which could be used for bioinspired robotic and prosthetic design.
Appendix A

MR-Compatibility of the MR-SoftWrist

This appendix contains details relating to the MR-compatibility of the MR-SoftWrist. While not a focus of this thesis, these details are included here for completeness. The content in this appendix appeared in [87]. See [87, 88] for more details on these experiments and [162] for a more in-depth analysis of the MR-compatibility of the MR-SoftWrist.

A.1 Methods

MR-compatibility scans were conducted at the Baylor College of Medicine Center for Advanced MR Imaging, using a Siemens 3T MAGNETOM Trio with a 12 channel head coil and 60 cm bore. We assessed MR-compatibility of the MR-SoftWrist by assessing the introduction of higher temporal fluctuation in the fMRI signal, through measurement of the temporal noise-to-signal ratio (tNSR) during fMRI sequences. Each excitation protocol was conducted in three different experimental conditions, i.e., baseline (BL), device in (IN), device moving (MVT). In the BL condition, only the phantom was in the scanner. In the IN condition, the device was located at a position representative of its normal operation (see Fig. 3.1). In the MVT condition, the device was commanded to track 6 mm peak-to-peak sinusoidal joint space movements [87].

tNSR, a measure proposed in [231] and introduced in the MR-compatible robotics literature in [232], involves scanning a phantom with electrical properties matching the ones of a human head. For this purpose, a spherical phantom filled with doped agar gel
was used. A standard T2-weighted sequence employed in fMRI studies was used (voxel size=2.5x2.5x2.5 mm, image size=80x80 px, flip angle=78, TE=35 ms, TR=2000 ms, no. slices=38). Processing of tNSR involved definition of a region of interest, a rectangular set of 16x16x21 adjacent voxels in the center of the phantom, resulting in a total of 5376 voxels. Signal intensity measured from each voxel \( i \) was concatenated in a timeseries \( s_i \), with mean \( \bar{s}_i \), that was detrended using a linear regression. Detrending increases specificity of the measurement, compensating for the slower thermal scanner dynamics in the computation of temporal fluctuations of the signal. From the detrended timeseries \( \ddot{s}_i \), noise \( n_i \) was calculated as

\[
\dddot{n}_i = \text{median}(|\dddot{s}_i|), \tag{A.1}
\]

and tNSR was calculated as percentage of the mean timeseries signal, as

\[
tNSR_i = \frac{n_i}{\bar{s}_i} \times 100. \tag{A.2}
\]

MR-compatibility was assessed through a non-parametric Kruskal-Wallis test, comparing the distributions of tNSR from all 5376 voxels, in the three experimental conditions BL, IN, and MVT [87].

### A.2 Results

No significant difference was determined for the effect of “experimental condition” on tNSR, as demonstrated by the Kruskal-Wallis test \( (p = 0.40) \), (see Fig. A.1(a) [87]). This is also confirmed by estimate of the cumulative distributions of tNSR obtained through bootstrapping \( (N = 10000, \text{Fig. A.1(b)}) \), which estimates the tNSR confidence intervals to be \( 0.457 \pm 0.003 \) for the IN condition, \( 0.456 \pm 0.003 \) for the MVT condition, and \( 0.457 \pm 0.003 \) for the BL condition [87].
In addition to showing MR-compatibility with respect to not disturbing images, the effect of the scanner on the MR-SoftWrist was examined. Movement data of the device from the MVT condition was compared to the same protocol executed inside the control room. Comparing the data, the maximum position error of any link was 0.38 mm, a maximum full-scale-output error of 6%. This is a reasonable result since when comparing the two MVT conditions data, a similar maximum error of 0.34 mm and maximum full-scale-output error of 4% were found, indicating that the device did not lose any functional abilities due to the MR scanner [87].

Additional MR-compatibility experiments further validating the MR-compatibility of the MR-SoftWrist have been described in [162]. These experiments include those involving human participants during fMRI, establishing the possibility of using the MR-SoftWrist to replicate motor control experiments (such as in [52, 233]) by interacting with a user’s
wrist with accurate kinesthetic feedback while not interfering with fMRI images. The experiments presented in detail in [162] confirm that any increase in signal fluctuations introduced by the MR-SoftWrist are not sufficient to significantly degrade the quality of functional images measured during operation of the robot.
Appendix B

MR-SoftWrist System Validation: Measurement of Brain Activation

This appendix contains details regarding a fMRI experiment with the MR-SoftWrist. These details, while not a focus of this thesis, are included here for completeness. The content in this appendix appeared in [88]. See [88] for further information regarding this experiment.

B.1 Methods

During the functional experiment, a standard echo-planar imaging sequence was used (voxel size: 2.5 mm isotropic - no gaps, image size: 80x80 px, no. slices: 42, scanned volume: box with edges 200x200x105 mm, flip angle=78 deg, TE=35 ms, TR=2000 ms, pixel bandwidth=1453 Hz/pixel) covering the entire cerebrum and the superior part of the cerebellum. After the functional experiment, a high-resolution structural scan (magnetization-prepared rapid acquisition with gradient echo, voxel size: 1 mm isotropic - no gaps, scanned volume: box with edges 256x256x176 mm, flip angle=8 deg, TE=3.03 ms, pixel bandwidth=130 Hz/pixel) was conducted to allow registration of the functional images [88].

B.2 Data Analysis

The images acquired during the three block design experiments were analyzed with a standard fMRI processing batch, including realignment (to the first image, using SPM8 – Wellcome Department of Imaging Neuroscience, London, UK – realignment function, with
options quality = 95%, separation 2.5 mm), coregistration to the structural magnetization-prepared rapid acquisition with gradient echo (using the SPM8 coregistration function with the normalize mutual information option, with progressively decreasing separation of 4 mm, 2 mm and 1 mm), spatial normalization (using SPM8 preset values), smoothing with an isotropic Gaussian filter (full width half maximum 8 mm), and high-pass filtering (time constant=128 s). A general linear model was constructed for each experimental condition, using the block variable (stimulus on/off), convolved with the SPM8 canonical hemodynamic response function, as the regressor of interest, and adding head motion parameters estimated through realignment as nuisance regressors. To determine differential activation in response to different haptic environments, the EA and ZF runs were concatenated in time and underwent the same pre-processing steps, with the exception of high-pass filtering, which was disabled for the analysis including concatenated data. A second general linear model was constructed, using the block variables (stimulus on/off) convolved with the canonical hemodynamic response function as regressors of interest, and using constant and linearly increasing regressors to account for the non-continuous acquired data. Model estimation yielded $t$-maps for the first three general linear models, one for each experimental condition relative to baseline (Active $> VC$), and estimation of the second general linear model yielded parametric maps for the difference between activation in the two experimental conditions (EA-ZF $> 0$). Using SPM8 correction for multiple comparisons (family wise error correction at $p < 0.05$), the voxel-level false discovery rate thresholded $t$ score for the whole brain analysis was $4.8 \pm 0.1$ [88].

### B.3 Results

The total head displacement during the experiments, as estimated from image realignment parameters, was within 1 mm and 1 deg in all conditions. No large artifacts (i.e., volume
Figure B.1: Task related activation for the Active > Visual Control conditions (EA, PC and ZF), and for the contrast between EA and ZF conditions (EA-ZF). Statistical parametric maps are overlaid on the standard Montreal Neurological Institute 152 template, as axial multislices, cut at z values labeled below the images. Note that t values higher than the colormap maximum are saturated to the highest intensity values [88].

distortions or RF noise lines) could be detected from visual inspection of the scanned volumes. Task-related activation maps revealed activation in the contralateral primary motor cortex, and bilaterally in the somatosensory cortex. Activation in the premotor cortex was bilateral in the EA mode, but only contralateral in the ZF and PC modes (see Fig. B.1 [88]). The contrast between activation in the EA and ZF conditions revealed areas with higher activation in the EA mode relative to the ZF mode, which included bilateral portions of the premotor cortex and somatosensory cortex, and bilateral BA44 [88].
Appendix C

SE-AssessWrist Images

This appendix contains additional images of various components and features of the SE-AssessWrist, both through CAD renderings or pictures, presented in Chapter 4.

C.1 Spring Assembly

![CAD renderings of the spring designed for the SE-AssessWrist. The spring was designed to be CNC’ed so that the hub, where the shaft would connect to the spring, would be one part with the spring. (a) Isometric and (b) top view of the spring. Note that in the top view, the two holes in the mid plane of the spring are thru holes for set screws to secure the cables on the pulley. The other holes are for threading or dowel pins.](image-url)
Figure C.2: The components considered a part of the essential spring assembly. (a) Isometric view highlighting the dowel pins and screws used to rigidly connect the pulley and spring. There is also a designed ridge on the pulley such that the spring does not rest on the pulley. (b) A side view highlighting the retaining ring and grooves for the cables.

C.2 FE and RUD Modules

Figure C.3: Assembled modules for the (a) FE and (b) RUD joints. The RUD module connects to the FE module through the shaft collar, which enables ease of assembly and disassembly. Similarly, the handle (not shown) connects to the RUD output through the shaft collar.
Figure C.4: Images of the motor assemblies used to actuate the (a) FE and (b) RUD joints through a Bowden cable transmission.

C.3 Complete Assemblies

Figure C.5: CAD renderings of the SE-AssessWrist, providing unique perspectives for visualizing (a) the entire assembly and (b) the RUD motors.
Figure C.6: (a) Completed SE-AssessWrist 2DOF assembly and (b) 1DOF FE assembly used for testing. Due to the use of shaft collars to attach the RUD unit to the FE joint, the SE-AssessWrist can easily be adapted to a single degree of freedom testbed. This testbed was used for preliminary prototyping, and might be useful for future experiments looking to characterize single degree of freedom wrist biomechanics or as an experimental platform for control of a series elastic actuated joint through a Bowden cable transmission.

Figure C.7: CAD renderings of the 1DOF module of the SE-AssessWrist, providing unique perspectives for visualizing (a) the entire assembly and (b) the FE motors.
Appendix D

SE-AssessWrist Validation Study Methods Document

This appendix provides the study methodology document created for assistance with running the validation study presented in Chapter 4. Additionally, the IRB used for the study is also included in this section. Some additional images of a participant during the experiment are also provided in this section.
Participants

- $n = 5$
- Right hand dominant, healthy, with no known physical, cognitive, or other disabilities that would impede their wrist, and normal or corrected to normal vision.

Estimated Procedure Time [70 min overall]

1. Sign IRB [3 min]
2. Attach sEMG electrodes [15 min]
3. Measure maximum voluntary contraction [5 min]
4. Don SE-AssessWrist [1 min]
5. Measure wrist ROM [10 min]
6. Measure wrist stiffness [35 min]
7. Doff equipment [1 min]

2. Attach sEMG electrodes

- Participants will be equipped with 4 sEMG electrodes on the forearm to measure muscle activity related to FE and RUD, and a ground electrode on the bony part of the elbow.

3. Measure maximum voluntary contraction (MVC)

- Participants will be asked to perform MVC (against manual resistance) in flexion, extension, ulnar deviation, and radial deviation, three times for each direction.
- Participants will be given rest between efforts.

4. Don SE-AssessWrist

- Adjust alignment variables to align wrist with SE-AssessWrist axes.
- Attach user's hand to open handle setup using Velcro and foam padding for comfort.
- Representative neutral configuration shown in pictures below.

Neutral configuration. Note: distal wrist cuff not shown in the left and center images, but is used during the study (see right image).
5. Measure wrist ROM

- Participants will make 72 movements in 24 different wrist directions. The 72 movements will be separated into 3 blocks. Each block will contain a movement in all of the 24 directions, with the movements within a block presented in random order.
- For assisting with this task, the user will be presented with a visual display consisting of a center target (neutral), a cursor (user position), and a line in the direction of the desired movement. For ease of visualization, movements on the screen will be presented with extension+ (x axis) and radial+ (y axis).
- The participant will be instructed to move in the direction of the line as far as they comfortably can.
- The lines will all have the same length so as not to provide a reference for ROM.
- The target angle will automatically update once the user returns approximately to center.
- Prior to data collection, participants can perform a few trials for familiarization, if needed.

![Sample GUI during ROM task. Participants will move in the direction of the green line (flexion and ulnar deviation in this case).](image)

6. Measure wrist stiffness

- The robot will move the participant's wrist slowly in sequential order (same as in 5. ROM) between 24 targets (spaced 15 deg apart in the FE/RUD space).
- The distance from neutral to the outer limit will be determined from the ROM found 5. ROM.
- The robot will move to the targets by being commanded to follow a ramp position trajectory (constant velocity) with a velocity magnitude of 0.2 rad/s (about 10 deg/s).
- Each movement repetition, which consists of moving from the neutral configuration to the outer limit and back, will be repeated 3 times.
- For familiarization, the participant will undergo a full circuit with only 12 targets (about 2-3 min).
Consent Form for Participation in Research

Study Title: Evaluating Wrist Biomechanics through Human-Robot Interaction

Principal Investigator:
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Other Investigators:
Andrew Erwin
Graduate Student
Nick Moser
Undergraduate Student

Participant’s Name: ___________________________ Participants ID Number: ____________
__________________________________________

You may be eligible to take part in a research study. This form gives you important information about the study. It describes the purpose of the research, the risks and possible benefits of participating in the study.

Purpose of this Study
The study is designed to advance the understanding of wrist biomechanics, which is important to fields such as rehabilitation and sports science. For example, by investigating wrist biomechanics with healthy participants, conclusions can be drawn about the potential for using the assessment robot for clinical use. This project represents the first step towards clinical implementation.

Procedures / What will happen to me in this study?
This experiment will focus on interacting with a wrist robot. There will be two types of interactions. In one interaction, the user will move their wrist while the robot measures the user’s movements. In the second type of interaction, the robot will move the user while the user attempts to relax their wrist. The relaxation level of the wrist will be measured through up to 4 surface electromyography (EMG) electrodes placed on the forearm, near the elbow. The experiment will take place on the Rice University Campus. All sessions will last no more than 2 hours, with sufficient breaks to prevent fatigue and maintain focus.

Participant Requirements
Participants must meet the following criteria: healthy, with no known physical, cognitive, or other disabilities that would impede their wrist, and normal or corrected to normal vision.

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Rice University

Consent Form for Participation in Research

Risks
The risks and discomfort associated with participation in this study are no greater than those ordinarily encountered in daily life.

Benefits
There may be no personal benefit from your participation in the study, but the knowledge received may be of value to humanity.

Compensation & Costs
There will be no cost to you if you participate in this study. Similarly, you will not be financially compensated for your participation in this study. If you are enrolled in a qualifying class, you may be awarded extra credit for your participation in the study.

Ending Your Participation
Your participation in this study is entirely voluntary. You are free to refuse to be in the study and your refusal will not influence current or future relationships with Rice University and participating sites.

Confidentiality
By participating in the study, you understand and agree that Rice University may be required to disclose your consent form, data, and other personally identifiable information as required by law, regulation, subpoena, or court order. Otherwise, your confidentiality will be maintained in the following manner:

Your data and consent form will be kept separate. Your consent form will be stored in a locked location on Rice University property and will not be disclosed to third parties. By participating, you understand and agree that the data and information gathered during this study may be used by Rice University and published and/or disclosed by Rice University to others outside of Rice University. However, your name, address, contact information, and other direct personal identifiers in your consent form will not be mentioned by Rice University in any such publication or dissemination of the research data and/or results. Each participant in this study will be assigned a number in order to facilitate ease of data processing and confidentiality. The researchers will record any data collected during the study by number, and not by name. The document detailing the relationship of any participant to their respective number will be stored separately, and securely. Any original recordings or data files will be stored in a secured location accessed only by authorized researchers.

Optional Permission
I understand that the researchers may want to use a short portion of any video or audio recording for illustrative reasons in presentations of this work for scientific or educational purposes. I give my permission to do so provided that my name and face will not appear.

☐ YES  ☐ NO  (Please initial here ________ )

Version: February 2018
Consent Form for Participation in Research

Rights
Your participation is voluntary. You are free to stop your participation at any point. Refusal to participate or withdrawal of your consent or discontinued participation in the study will not result in any penalty or loss of benefits or rights to which you might otherwise be entitled. The Principal Investigator may at his/her discretion remove you from the study for any of a number of reasons. In such an event, you will not suffer any penalty or loss of benefits or rights which you might otherwise be entitled.

Right to Ask Questions & Contact Information
If you have any questions about this study, you should feel free to ask them now.

If you have questions later regarding the study or a research-related injury, or if you have complaints, concerns, suggestions about the research, desire additional information, or wish to withdraw your participation please contact the Principal Investigator by mail, phone or e-mail in accordance with the contact information listed on the first page of this consent.

For questions about your rights as a research participant, or to discuss problems, concerns or suggestions related to the research, or to obtain information or offer input about the research, you should contact Stephanie Thomas, Compliance Administrator, at Rice University. Email: irb@rice.edu or Telephone: 713-348-3586

Conflict of Interest
Graduate student members of this project team may have the potential for personal benefit in the form of potential use of the results of this study as a part of a thesis or dissertation.

Voluntary Consent
By signing below, you agree that the above information has been explained to you and all your current questions have been answered. You understand that you may ask questions about any aspect of this research study during the course of the study and in the future. By signing this form, you agree to participate in this research study.

______________________________
PARTICIPANT SIGNATURE

DATE

I certify that I have explained the nature and purpose of this research study to the above individual and I have discussed the potential benefits and possible risks of participation in the study. Any questions the individual has about this study have been answered and any future questions will be answered as they arise.

______________________________
SIGNATURE OF PERSON OBTAINING CONSENT

DATE
Appendix E

Surface Electromyography Specifications and Circuitry

This appendix provides additional information regarding the surface electromyography (sEMG) electrodes used during the validation study presented in Chapter 4.

E.1 Performance Specifications

The four sEMG electrodes used during the experiment are the Motion Lab Systems MA-411. The following information is from the corresponding product datasheet [234]. These electrodes are pre-amplified with a gain of x20, which will produce 0.02 V at the output for a 1 mV input. The electrodes require a ground reference on the participant for common mode signal rejection, which in the study was accomplished through placement of an electrode on the olecranon. The input impedance of the electrodes is 100,000,000 Ω and has overload protection. The sensor contacts are 12 mm medical grade stainless steel disks spaced 18 mm between the centers. The size of the sensor is 38 x 19 x 8.5 mm and weighs 20 g (including the cable) [234].

E.2 Additional Amplification Circuitry

Since the pre-amplified gain of x20 is too low to acquire an accurate signal by the data acquisition board (DAQ), an additional amplification circuit was built. As recommended by Motion Lab Systems [234], the EMG signals were first passed through an analog bandpass filter. The amplification circuit used an AD620 amplifier with a gain of 742. See Fig. E.1
for a circuit schematic of the signal processing circuit and Fig. E.2 for a Bode plot of the bandpass filter. Additionally, see Fig. E.3 for a circuit schematic of the power signals sent to the sEMG electrodes, and Fig. E.4 to see the physical circuit.

Figure E.1: Electronic schematic of the signal processing circuit of sEMG signals. The circuit includes a bandpass filter and an amplification circuit. The schematic was created through Autodesk EAGLE and modified in Inkscape.

Figure E.2: Bode plot of the bandpass filter used in the signal amplification circuit. The low cutoff frequency is 2.57 Hz and the high cutoff frequency is 5134 Hz, while the center frequency is 114.8 Hz.
Figure E.3: Electronic schematic of the power sources for the sEMG amplifiers. The circuit includes an op-amp to lower the power signals, while also providing a buffer, and a diode to provide a smooth supply to the sEMG electrodes. The schematic was created through Autodesk EAGLE and modified in Inkscape.

Figure E.4: Physical circuit board for the sEMG signal processing and power supply.
Bibliography


