Proceedings of the ASME/ISCIE 2012 International Symposium on Flexible Automation
ISFA2012
June 18-20, 2012, St. Louis, Missouri, USA

ISFA2012-7250

A ROBUST WHEEL INTERFACE WITH A NOVEL ADAPTIVE CONTROLLER FOR COMPUTER/ROBOT-ASSISTED MOTIVATING REHABILITATION

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ABSTRACT

TheraDrive is an effective system for post-stroke upper extremity rehabilitation. This system uses off-the-shelf computer gaming wheels with force feedback to help reduce motor impairment and improve function in the arms of stroke survivors. Preliminary results show that the TheraDrive system lacks a robust mechanical linkage that can withstand the large forces exerted by patients, and it lacks a patient-specific adaptive controller to deliver personalized therapy. It is also not capable of delivering effective therapy to severely low-functioning patients. A new low-cost, high-force haptic robot with a single degree of freedom has been developed to address these concerns. The resulting TheraDrive consists of an actuated hand crank with a compliant transmission. Actuation is provided by a brushed DC motor, geared to output up to 23 kgf at the end effector. To enable a human to interact with this system safely, a special compliant element was developed to double as a failsafe torque limiter. A set of strain gauges in the handle of the crank are used to determine the interaction forces between human and robot for use by the robot's impedance controller. The impedance controller is used to render a one-dimensional force field that attracts or repels the end effector from a moving target point that the human must track during therapy exercises. As exercises are performed, an adaptive controller monitors patient performance and adjusts the force field accordingly. This allows the robot to compensate for gravity, variable mechanical advantage, limited range of motion, and other factors. More importantly, the adaptive controller ensures that exercises are difficult but doable, which is important for maintaining patient motivation. Experiments with a computer model of human and robot show the adaptive controller's ability to maintain difficulty of exercises after a period of initial calibration.

INTRODUCTION
Background on Stroke

With the increasing portion of elderly people in the population, stroke is one of the leading causes of disability and death in the United States [1]. Stroke is an interruption of blood flow to the brain, brought about by an embolism or by a blood vessel rupture, which rapidly causes nerve cell damage or death. The brain damage from a stroke frequently causes cognitive impairments and hemiparesis which often manifest as a loss of motor coordination and impairments affecting the stroke survivor's ability to perform activities of daily living, such as walking, self-feeding, and dressing. Other impairments resulting from stroke can include speech impairment, loss of sensation in affected limbs, and inability to process sensory data. Post-stroke patients must undergo physical therapy to regain lost motor function. Traditionally this therapy has been done manually by a physical therapist, but robotic systems, with their ability to measure and exert forces

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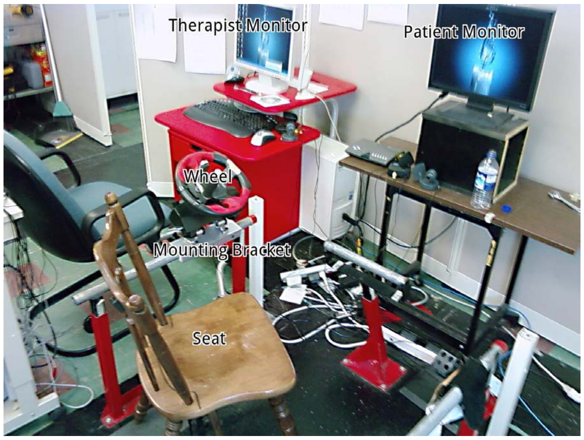

FIGURE 1. THERADRIVE SYSTEM SETUP SHOWING PATIENT AND THERAPIST WORKSTATIONS

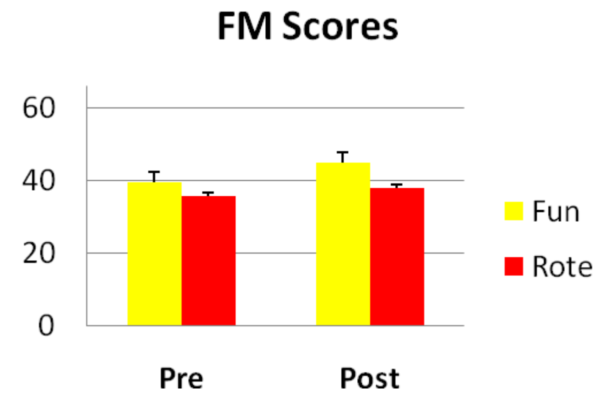

FIGURE 2. PATIENT FUGL-MEYER SCORES PRE AND POST THERAPY

accurately and to control motion, are becoming more common.

Background on TheraDrive

TheraDrive is a system developed to examine issues of robot/computer motivating rehabilitation [2, 3]. It uses off-the-shelf force-feedback steering wheels (Logitech Wingman) as therapy robots with a single degree of freedom. This steering wheel is mounted to an adjustable frame, allowing the plane of motion to be moved (Figure 1). In the center of the frame is a movable chair on rails to ease the process of seating an impaired patient, and in front of the frame is a task display for visual feedback. The entire system can be built from readily available parts, and it can be disassembled and folded for added portability. Ultimately, the goal of this project is to create a low-cost portable physical therapy system for stroke survivors that can be used at home or in a clinic. Most existing rehabilitation robots are not able to fill this niche because they are too bulky or too expensive.

Therapy with the system involves positioning and tracking exercises (rote therapy) as well as driving games (fun therapy). For rote therapy, the patient is presented with a point-to-point positioning task or a trajectory-following task by the UniTherapy software program [2]. The wheel is used to move a cursor to a specified point or to guide the cursor along a moving path. The therapist can select from a large array of point layouts and trajectory shapes to adjust the task difficulty to the patient's ability level. Assistive or resistive forces can be generated by the wheel's force feedback motor, and the magnitude of this force is set by the therapist before each exercise. This assistance/resistance usually takes the form of a spring force that attracts/repels the patient from the target position, but it can be changed to emulate a mass, a damper, or random perturbations. Game therapy simply involves having the patient play a computer game using the wheel as a controller. UniTherapy logs controller data in the background during gaming sessions to analyze smoothness of motion. Typically a driving game such as Need for Speed or TrackMania is played, as these are well-suited to the wheel interface and designed to support force-feedback controllers.

A pilot study to assess the TheraDrive system tested ten subjects, each of whom underwent 24 one-hour training sessions over the course of six to eight weeks [3]. Subjects were divided into a group that did only rote therapy and a group that did only game therapy. Every four sessions, the subjects were evaluated using metrics including the Fugl-Meyer scale and the Ashworth test for spasticity. Both groups showed improvements in motor function (Figure 2) and decreases in spasticity, with the fun therapy group showing better improvement trends. Additional subjects were needed to show statistical significance of these improvements. Additionally, the game-based therapies enhanced the motivation and engagement of patients and caused a slight increase in functionality gain over simple tracking and positioning exercises. However, one low-functioning patient showed little to no gain from therapy because the wheel could not exert sufficient assistive force to allow him to perform the tasks. This highlights a major weakness of the TheraDrive system—the wheel cannot exert enough assistive force to provide effective therapy to very low-functioning patients.

Problem Definition

The TheraDrive system has shown success in preliminary human subject trials, but these trials have also revealed severe limitations of the system. Stroke patients using TheraDrive showed improvements in motor function after a typical course of therapy and found the training regimen to be interesting and engaging [3].

There is a need to improve force-feedback magnitudes and profiles for low functioning subjects. One stroke survivor with inability to use his hand for grasping and very little arm movement used the system. An external forearm and upper arm support sling was employed so that he was able to get some training. He needed maximum assistance to complete tasks, but the system could only provide limited forces to move his impaired arm. This revealed a need to improve the wheel interface to better support low-functioning stroke survivors who have difficulty with hand opening/closing and difficulty supporting the forearm against gravity.

In addition it was clear that, for low functioning subjects, the maximum force-feedback moment produced by the wheel is not sufficiently large that they cannot overpower it. This means that the wheel cannot be used to build strength or to emulate a sufficiently rigid constraint when working with these types of patients. The motor driving the wheel is a small brushed DC motor, which is underpowered for this task. Another major concern about the wheel is its ability to withstand off-axis forces and moments. This is an important issue because patients using the wheel have varying levels of motor coordination, so they often exert lifting or bending forces on it. Since the wheel uses plastic-on-plastic bushings instead of roller bearings, its already-short wear life is shortened even further by increased loading on these surfaces. These sliding surfaces produce a significant amount of wear debris, which falls onto the gears and embeds itself between the gear teeth, causing damage to the gears. Backlash is also a concern with the wheel, since it uses molded plastic gears to couple the force-feedback motor. These gears have considerable backlash and compliance, both of which increase as the gear wears. The backlash of the gearing creates a dead zone where the force feedback does not influence the wheel's motion. Essentially, the wheel is a toy, and it lacks many features that would make it ideal for use as a therapeutic tool For low-functioning subjects who need more assistance during training. Several Logitech wheels wore out over the course of the pilot study as a result of the large forces in the normal and tangential directions that users placed on the wheel. There is a need to improve the wheel interface and improve its ability to be effective for patients with all levels of ability throughout the therapy.

Patient-specific adaptive control is another important feature that the TheraDrive system currently lacks. Patients experience differing levels of impairment at different points in their range of motion, and a controller should be developed to account for this fact in order to deliver personalized therapy. The current controller for the wheel is only able to simulate a linear spring with constant stiffness. This means that during an exercise, patients can have great difficulty moving the wheel at some points along its travel but move the wheel with relative ease at other points. Thus, a controller must be designed to adapt to each patient's individual form of impairment, based upon range of motion, torque, and speed.

This paper will discuss the development of a new low-cost, high-force haptic robot with a single degree of freedom that has been developed to address these concerns. The mechanical design for the new robot crank arm will be described along with its integration into the TheraDrive set-up. Simulation models for the system along with the new patient-specific adaptive controller will be discussed. Results from experiments with a computer model of human and robot show the adaptive controller's ability to adapt difficulty of exercises to a patient's ability after a period of initial calibration.

FIGURE 3. NEW THERADRIVE SYSTEM SETUP AND HAPTIC ROBOT

DESIGN OF NEW THERADRIVE SYSTEM

Design Goals and Constraints

The problem definition above give rise to the our key design criteria:

1. The new robot crank system must be integrated into the existing Theradrive system and be able to mount to the adjustable frame in front and on the sides, allowing exercises to be performed in different planes.
2. The robot must be low-cost to maintain the affordable theme (less than $3000 USD).
3. The robot must support torques on the crank larger than 25 N-m.
4. The controller must be patient-specific and adapt forces at the crank to accommodate a variety of patients with strokes, especially low-functioning patients with motor weakness and poor coordination.
5. The robot must be backdrivable and safe (torque limited).
6. The forces on the crank arm must be measured and recorded.

Description of Design

Servomechanism Figure 3 and Figure show the new TheraDrive system and the new haptic robot. A key feature of the robot is its motor. Due to budget and weight constraints, the motor selected is an aftermarket treadmill motor from Turdan Industries. It is a 2-pole brushed DC motor rated at 2000 W at 70% duty with a maximum speed of 8000 rpm and with windings rated for 130 V. Although the motor is only rated for rotation in one direction, its windings and commutator are symmetrical, so it achieves the same performance turning backwards as it does turning forwards. The motor is overpowered for this application, but through the servo amplifier and a mechanical torque limiter, the maximum mechanical power delivered to the patient is 950 W (45 N-m at 200 rpm). A maximum output torque of 45 N-m translates to a linear force at the end effector of 23 kgf, a value on the same order of magnitude as the linear output force of the human arm at the hand. To couple the motor to the end effector, a 40:1 planetary gearbox and custom-made torque limiter are used. A planetary gearbox was selected over a harmonic drive because the planetary gearbox is more easily backdriven, allowing for smoother haptic interaction. Driving the motor is a plug-in analog pulse-width modulated (PWM) servo amplifier from Advanced Motion Control (model 30A20AC) interfaced with a DAQ system. This amplifier does not need an external power supply and instead runs on rectified mains voltage (40-190 VDC), saving space and money. The servomechanism is controlled in impedance mode, using a load cell in the end effector to provide feedback to the impedance loop. A 1000-line encoder is attached to the torque limiter output shaft to measure the end effector position, which is used by the impedance controller to render force fields.

To measure interaction forces between human and robot, a custom load cell was built into the crank handle. These interaction forces must be known in order to close the impedance control loop and for clinical data logging. The load cell is composed of a cantilever beam with eight strain gauges mounted, fixed to the crank arm. The handle grip is fitted around the load cell and rotates freely on bearings. Fixing the load cell to the arm rather than the grip allows forces to be measured in the radial-tangential coordinate system of the crank arm and simplifies the conversion of interaction forces to joint-space torque. Strain gauges are mounted around the circumference of the beam at 90-degree intervals at two points along its length. Diametrically opposed gauges are wired in half-bridge configuration and measure bending strains at two points due to moments about the radial and tangential axes, for a total of four measured strains. The half-bridge configuration also provides temperature compensation because opposing gauges will experience the same amount of thermal strain, creating no change in the difference between the two gauges' resistances. Temperature compensation is important because it cannot be assumed that the strain gauges will remain at a constant temperature due to their proximity to the patient's hand, a source of heat.

The strain gauge amplifier is designed around the Burr-Brown INA125 instrumentation amplifier with precision voltage reference, a chip designed to power strain gauges and amplify strain gauge signals. These four strain signals are used to calculate bending stresses, then bending moments, and finally shear and moment reactions at the fixed end of the beam. The interaction force is converted to joint-space torque by multiplying the tangential shear reaction force by the crank arm radius, and this is used as the feedback to the impedance controller. RF noise created by the PWM servo amplifier is a significant source of error in the load cell signal. To combat this, the strain gauges are connected to the strain gauge amplifier through shielded twisted-pair leads, and the output of each half-bridge is filtered with a RF choke and shunt capacitor. Additionally, a RF choke was added to the motor leads to reduce the emitted RF noise.

Compliant Torque Limiter In order for the robot to apply large forces to the patient safely, a torque limiter was added to the robot's transmission. Additionally, the transmission must be compliant to protect the patient from impact loads during a sudden reversal of the motor or during an accidental collision between patient and robot [4]. Both of these requirements are met with a single transmission element in the robot. This compliant torque limiter is essentially a cross between a drill clutch and the VS Joint in [5]. It consists of a crown cam, splined to the input shaft, and a cam follower keyed to the output shaft. The cam is held against the follower by a spring, but it is free to slide along the spline of the input shaft. Deflecting the output shaft causes the cam to compress the spring, generating a restoring torque. This creates a torsional spring from a linear spring, and the stiffness profile of the torsional spring can be altered by changing the shape of the cam surface. For an angular deflection of θ, the restoring torque $T(\theta)$ is defined by the relationship

$$T(\theta) = \left(\frac{dz(\theta)}{d\theta}\right)(r_c)\left(F_{ko} + kz(\theta)\right), \tag{1}$$

where $z(\theta)$ defines the shape of the cam surface, r_c is the cam radius, k is the spring stiffness, and F_{ko} is the preload in the spring. Preload in the spring is adjusted manually using a telescoping shaft collar to compress the spring. The cam shape chosen for this robot is parabolic. With a parabolic cam path, increasing the preload in the spring increases the torsional stiffness together with the torque limit. When the spring preload is low, this provides patients who cannot handle large interaction forces a softer robot to reduce discomfort. When the preload is high, the increased stiffness reduces control loop delay, enhancing the stability of the impedance controller when rendering large forces. The increased stiffness also presents patients a robot that feels stiffer and stronger, hinting that they will be experiencing larger interaction forces.

Software and Control The robot is controlled using a dedicated computer running Mathworks Simulink xPC Target OS for real-time control and data acquisition. This PC is fitted with two DAQ boards to read sensors and communicate with the servo amp: an encoder board from Measurement Computing (PCI-QUAD04) and a multipurpose board from National Instruments (NI PCI-6251). A host computer interfaces with the target

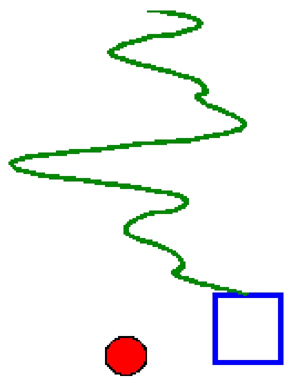

FIGURE 4. TASK INTERFACE FOR PATIENTS

computer to upload executable code and display the patient interface. The interface presents the patient with a cursor that follows the position of the end effector. The cursor is situated at the bottom of the screen, and the desired trajectory scrolls down the screen towards the cursor, presenting a box at the current target location, as shown in Figure 4. The cursor changes from red to green while it is within the target box.

Control of the robot is achieved through three loops, all running at a sample rate of 1000 Hz to allow for smooth controller response. The innermost control loop is the impedance proportional-integral-derivative (PID) controller. This control loop is used to regulate forces at the interface between patient and robot, taking an input at the load cell and sending a torque command to the motor. Next is the assistive/resistive controller, which renders a force field in the robot's workspace. The controller divides the workspace into 17 regions, spaced $1/16$ of a revolution apart. Each region has a value assigned to it to define the assistive/resistive stiffness of the robot at that location, with negative stiffnesses being assistive and positive stiffnesses being resistive. The stiffnesses between regions are interpolated to produce smooth transitions between regions of different stiffness. Assistive stiffness is rendered as a linear spring pulling the patient towards the target end effector position with a damper added to reduce overshoot. Resistive stiffness is rendered as a linear spring that repels the patient from the target position, creating an unstable system that the patient must stabilize. It is necessary to implement position-dependent stiffness because stroke patients usually have inconsistent abilities in the range of motion of their impaired arm.

The outermost control loop is the adaptive controller. This controller evaluates patient performance in real-time and continuously adapts the stiffness of the robot to the patient's ability. Patient performance is quantified as the root-mean-square (RMS) trajectory tracking error over the past three seconds. This error is compared to a desired error of 0.25 radians (50 mm of arc length), a value corresponding to the width of the target that the patient tracks. A nonzero tracking error is desired because this maintains difficulty of exercises. If the desired tracking error were zero, the robot would always provide maximum assistance to every patient, and if the desired tracking error were too large, the robot would always provide maximum resistance to patients. Stiffness values in a quadrant centered around the end effector are adjusted proportionally to the difference between actual and desired tracking error, essentially establishing a proportional controller around the gains of the assistive/resistive controller. Over time, the adaptive controller shapes the stiffness profile to suit the patient's ability level, ensuring that exercises will be difficult but doable. "Difficult but doable" means that patients should be presented with a challenge sufficient to maintain motivation and a moderate degree of exertion, but the challenge should not be so great as to cause patients to become frustrated or to fail to complete exercises.

METHODS AND EXPERIMENTS

Simulation Setup

A model of the robot and human patient was written in Mathworks Matlab and SimMechanics to aid in controller design and testing. The crank and human arm are modeled as a four-bar linkage, with the motor coupled to the crank through a spring. Parameters for the model robot, such as link inertia and motor torque constant, were found through characterization of the motor and through analysis of 3D part drawings in SolidWorks. Equations of motion were created and solved numerically within SimMechanics based upon the mechanical model. Using SimMechanics, all nonlinearities in the system, such as encoder quantization and Coulomb friction, could be modeled easily; and a more accurate model of the system than a linearized analytic model was produced. The human arm model was ported from another computational model used by Formica et al. for the MIT-MANUS robot [6, 7]. This model simulates the dynamics and control of the human arm with or without stroke impairment. It is a planar model consisting of a forearm/wrist segment, an elbow joint, an upper arm segment, and a fixed shoulder joint. The forearm/wrist segment is 35 cm long with a mass of 1.54 kg, and the upper arm is 25 cm long with a mass of 1.96 kg, corresponding to metrics of a human of average height and weight. A trajectory planner and joint-space proportional-derivative (PD) controller are used to model human control of the arm. The PD controller is of the form

$$\begin{pmatrix} T_e \\ T_s \end{pmatrix} = \begin{bmatrix} P_{ee} & P_{se} \\ P_{es} & P_{ss} \end{bmatrix} \begin{pmatrix} u_e \\ u_s \end{pmatrix} + \begin{bmatrix} D_{ee} & D_{se} \\ D_{es} & D_{ss} \end{bmatrix} \begin{pmatrix} \dot{u}_e \\ \dot{u}_s \end{pmatrix} \tag{2}$$

with cross-coupling terms in each gain matrix being nonzero, where u is the position error vector in radians in joint space and

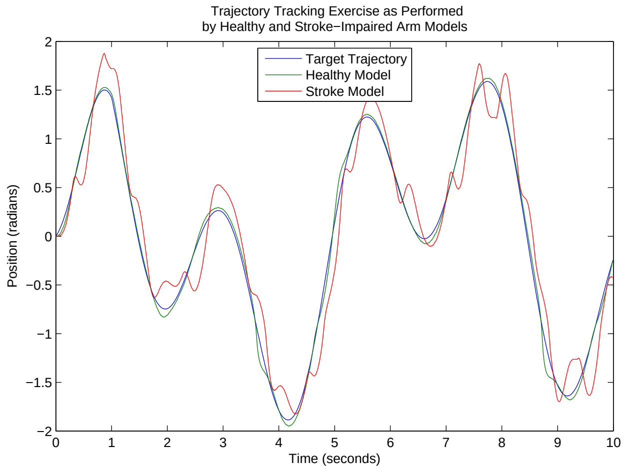

FIGURE 5. SIMULATED PATIENT UNASSISTED TRAJECTORY TRACKING WITH ROBOT

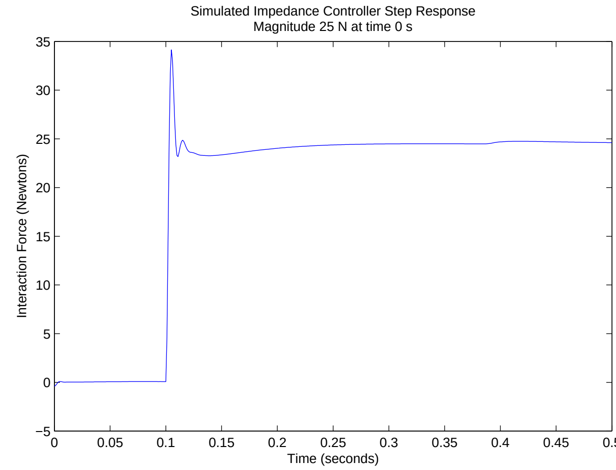

FIGURE 6. IMPEDANCE CONTROLLER STEP RESPONSE (SIMULATED)

T is the joint torque output vector in N-m. The PD controller gains are based on the stiffness and viscoelasticity of the average human arm and are defined as

$$P = \begin{bmatrix} 8.67 & 2.83 \\ 2.51 & 10.8 \end{bmatrix}; \quad D = \begin{bmatrix} 0.76 & 0.18 \\ 0.18 & 0.63 \end{bmatrix}. \tag{3}$$

Stroke impairment is simulated by perturbing the planned trajectory with a sawtooth function, creating a piecewise linear trajectory that deviates from the ideal trajectory to simulate lack of coordination. For example, if the ideal trajectory were defined as $\theta(t)$, the perturbed trajectory would be defined as

$$\theta_p(t) = \theta(t) + a\left(\theta(t) \bmod b - b/2\right), \tag{4}$$

where a and b define the amplitude and period of deviation, setting the severity of the simulated impairment. The difference in tracking ability between healthy and impaired simulated arms with no assistance/resistance can be seen in Figure 5. Trajectories used in exercises are a sum of two sine waves of different frequency and phase that create a pseudorandom smooth path.

Impedance Control Performance

The impedance PID controller was designed and tuned in simulation using Matlab's Control System Toolbox. Controller gains found were then used as a starting point for manual tuning of the real impedance controller. Figure 6 shows the step response of the simulated impedance controller. The controller was made to track a step function of magnitude 25 N at the end effector while the end effector was held by the simulated arm. The 90% settling time of this controller is 12 ms. The initial overshoot is caused by the integrator winding up to break the cogging in the motor (modeled as static friction), which is initially stationary. During exercises, peaks like this only occur when the direction of interaction force reverses or when the motor stops or reverses direction, as seen in Figure 7. The response of the controller is underdamped, but the transient vibration is small enough that it is imperceptible to the patient. The initial peak in force output can be felt by patients, but the energy imparted by this peak is not enough to influence motion of the arm; it is just a slight bump. A fast controller response was desired over steady-state accuracy because this improves the transparency and smoothness of motion of the human-robot interface. High steady-state accuracy is not crucial because this robot does not work to build fine motor skills, and small errors in interaction force have little effect on gross motor skills such as reaching.

Adaptive Controller Testing

The adaptive controller was tested by running exercises for extended periods of time in simulation. Long simulations were necessary because the adaptation rate of the controller was made to be slow to reduce the influence of exercise trajectory shapes. Figure 8 shows the adaptive controller responding to the performance of a simulated healthy patient. A sine tracking exercise was run for ten minutes of simulation time, and the resistive stiffness, averaged over the workspace, was plotted versus time. The controller response is approximately first-order with a time constant of 90 s, entering a limit cycle at steady state. Steady-state oscillations are caused by the shape of the tracked trajectory, and both have the same period of oscillation. This is because there are slight variations of difficulty in the motions that compose the

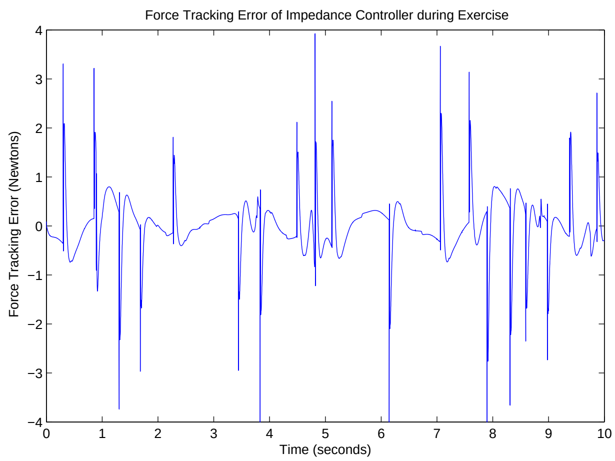

FIGURE 7. IMPEDANCE CONTROLLER FORCE TRACKING ERROR DURING EXERCISE (SIMULATED)

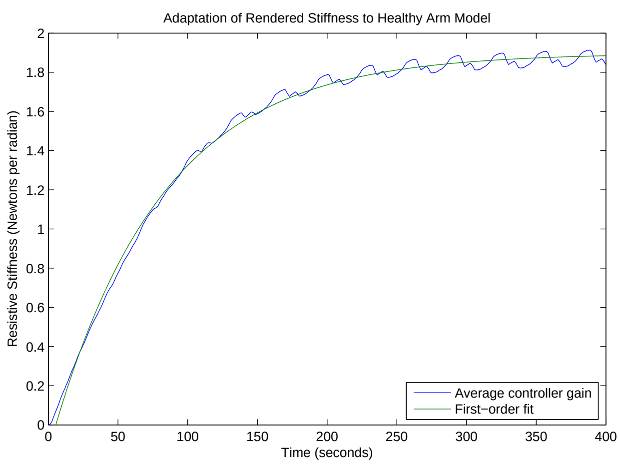

FIGURE 8. RENDERED STIFFNESS BEING ADAPTED TO HEALTHY PATIENT PERFORMANCE (SIMULATED)

trajectory. For example, tracking fast movements is more difficult than tracking slow movements, so changes in the trajectory velocity will perturb the gains of the adaptive controller. However, the perturbation is so small that it is imperceptible to patients, as the exercises themselves present variations of force of much greater magnitude and speed than those caused by small variations of stiffness.

Performance of a simulated healthy patient before and after controller adaptation are shown in Figure 9. The baseline performance of the patient with zero resistance has RMS tracking error well below the desired level, indicating that the patient tracks the trajectory with ease. However, the desired RMS tracking error of the patient is 0.25 rad, so the adaptive controller adds resistance to the exercise until this is achieved. With the proper amount of resistive gain, the patient is still able to track the reference trajectory, but with larger errors, indicating that the exercise has become difficult but doable.

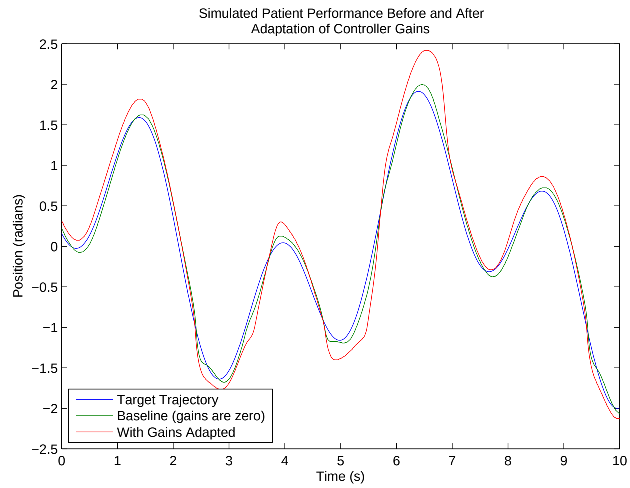

FIGURE 9. SIMULATED HEALTHY PATIENT PERFORMANCE BEFORE AND AFTER CONTROLLER GAINS ARE ADAPTED

Position-dependent assistive gain is important for therapy exercises with stroke patients, as they have varying levels of ability throughout the robot's workspace. To verify that the adaptive controller is able to provide position-dependent gains, a simulated stroke patient was created with large tracking errors at ± 1 rad. A sine tracking exercise was run until the adaptive controller reached steady-state, and the resulting gains were plotted in Figure 10. This graph shows two peaks where the assistive gain is larger ; one large peak at 1 rad and a smaller, wider peak at -1.3 rad. The peaks are of different magnitudes due to the differing mechanical advantage of the arm at these points. At -1.3 rad, the arm is extended and has little mechanical advantage over the robot, and less assistance is required to correct tracking errors. However, at 1 rad, the arm is close to the body, giving the patient a large mechanical advantage, so more assistance is required to correct the trajectory. Mechanical advantage is also the reason the peak at -1.3 rad is not at -1 rad. The mechanical advantage of the patient is lower at the angle of -1 rad than at -1.3 rad because around -1 rad, there is a point where the tangent to the path of the elbow intersects with the axis of the crank, creating a singularity in the kinematics. At this point, the shoulder joint has zero mechanical advantage over the crank because it cannot exert any moment on it, putting the arm's overall mechanical advantage at a minimum. The arm has more mechanical advantage at -1.3 rad, so more assistance is required at this point to correct

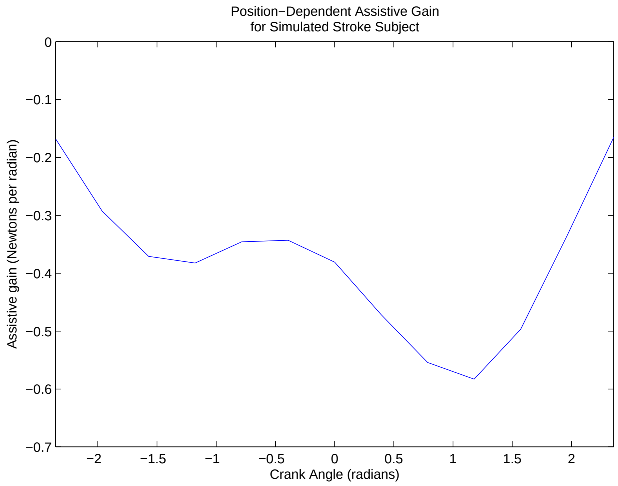

FIGURE 10. POSITION-DEPENDENT GAINS ADAPTED TO SIMULATED STROKE PATIENT PERFORMANCE

the arm's trajectory than is required at -1 rad.

CONCLUSION AND FUTURE DIRECTION

The design and simulation of a low-cost, high-force haptic robot for use with the TheraDrive system was discussed. This robot is capable of providing assistive forces up to 23 kgf to very low-functioning patients, expanding the capabilities of TheraDrive. Simulations show that the robot controllers are safe and robust, and the controllers offer adequate performance. The impedance controller has a settling time of 12 ms for smooth motion, and the adaptive controller successfully adapts gains to fit each simulated patient's unique abilities.

Preliminary human testing has begun at Marquette University. Previous TheraDrive stroke subjects and normal healthy subjects (a total of nine) will use the old and new Theradrive configurations. Subjects will complete surveys to determine their perception of both systems, and these survey results will provide an assessment of the robot's utility as a post-stroke therapy device.

An area of future investigation is the adaptive controller. Very little investigation has been done regarding the use of adaptive algorithms to modify therapy exercises to fit patient performance. It would be beneficial to investigate automated means of measuring patient performance as well as intelligent adaptation strategies. This controller can also be modified to provide real-time adaptive control of other systems with variable dynamics and/or unknown states, such as pick-and-place robots that move objects of unknown and non-trivial mass or temperature controllers for vessels containing a variable mix of fluids.

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