Abstract

**Background:** The emergence of commercial haptic devices offers new research opportunities to enhance our understanding of the human sensory-motor system. Yet, commercial device capabilities have limitations which need to be addressed. This paper describes the customization of a commercial force feedback device for displaying forces with a precision that exceeds the human force perception threshold.

**New Method:** The device was outfitted with a multi-axis force sensor and closed-loop controlled to improve its transparency. Additionally, two force sensing resistors were attached to the device to measure grip force. Force errors were modeled in the frequency- and time-domain to identify contributions from the mass, viscous friction, and Coulomb friction during open- and closed-loop control. The effect of user interaction on system stability was assessed in the context of a user study which aimed to measure force perceptual thresholds.

**Results:** Findings based on 15 participants demonstrate that the system maintains stability when rendering forces ranging from 0–0.20 N, with an average maximum absolute force error of 0.041 ± 0.013 N. Modeling the force errors revealed that Coulomb friction and inertia were the main contributors to force distortions during respectively slow and fast motions.

**Comparison with Existing Methods:** Existing commercial force feedback devices cannot render forces with the required precision for certain testing scenarios. Building on existing robotics work, this paper shows how a device can be customized to make it reliable for studying the perception of weak forces.

**Conclusions:** The customized and closed-loop controlled device is suitable for measuring force perceptual thresholds.

**Keywords:** Haptics, Force Feedback Device, Stability, Transparency, Grip Force Measurement, Force Perception

1. Introduction

1.1. Haptic Devices

Haptic devices enable the creation of virtual touch environments, making it possible to touch and grasp and feel the shape and weight of virtual objects, just as head-mounted display devices enable the observation of virtual visual scenes. Benefits of haptic systems include that a single device can generate many different virtual touch environments; environments can be modified on the fly; and data related to one’s interaction can be monitored and stored. Haptic devices include tactile systems, such as Braille displays and cellular phones that vibrate, which interact primarily with the skin, while the focus of this paper is on kinesthetic, or force feedback, devices, which interact primarily with the kinesthetic system (although the contribution of tactile afferents should not be neglected). By allowing real-time control of the physical interaction between the body and virtual touch environment, these devices are offering to a growing number of researchers a promising tool for investigating sensory-motor, cognitive, and neuronal mechanisms involved in areas including touch, proprioception, and object manipulation, e.g., (Shadmehr and Mussa-Ivaldi, 1994; Flanagan and Wing, 1997b; Burdet et al., 2001; Baud-Bovy and Gentaz, 2006; Drewing and Ernst, 2006). Reviews of the history of haptic devices, as well as methods for controlling them, can be found in Adams (1999) and Kern (2009).

Haptic devices are relevant for psychophysical testing, which quantifies the capacity of the human sensory system to sense physical properties, e.g., (Beauregard et al., 1995; Gurari et al., 2013; Dominjon et al., 2005; Kawai et al., 2012). They allow the experimenter to control and adjust the touch sensation using a single apparatus, making it relatively easy to conduct such studies. Further, they enable the experimenter to control the interaction force in a manner which would be impossible to achieve with real objects. For example, geometric and force cues can be dissociated to study haptic shape perception (Robles-De-La-Torre and Hayward, 2001) and visual and haptic cues can be isolated to identify how such information integrates (Ernst and Banks, 2002).

While many of the early studies involving haptic devices have been conducted with custom designed systems, a growing number of commercial apparatuses are coming to the market.
As a result, the use of these systems has expanded from the realm of mechanical engineering and haptic research groups towards use by researchers in other disciplines. As the interest in haptic devices grows, some inherent limitations, or challenges, related to the underlying technology may occur, especially when the systems are used beyond the confines of haptic research labs. In particular, researchers may not always be aware, or might sometimes overlook, the fact that haptic interfaces are real objects exhibiting real world properties, such as mass and friction, which impact the rendering fidelity of virtual touch environments (Salisbury et al., 2009). In robotics, this problem is known as a transparency issue (Hannaford, 1989; Lawrence, 1993; McJunkin et al., 2005). While perfectly transparent haptic devices do not and cannot exist, it is possible to design their hardware and control laws to minimize the undesired artifacts (e.g., Panarese and Edin (2011)). Unfortunately, commercially available haptic devices, as sold off-the-shelf, are not fit for creating precisely controlled stimuli, and thus, are not appropriate for use in many studies. This issue is particularly relevant for psychophysical and neurophysiological studies, where precise control of the stimulus is needed.

1.2. High-Fidelity Weak Force Rendering

The objective of this work was to create a setup which can display weak forces accurately and precisely enough to allow for the identification of force detection thresholds. Although force perception has been studied for many years, starting from the seminal work of Weber (1834), nearly all studies have measured discrimination thresholds and not detection thresholds (see Jones (1986) for a review of this literature). To identify force thresholds, weak forces need to be displayed which are unaffected by possible movements of the object that is in contact with the human. This is not feasible in the real world, especially when the object is in motion, since an object has a mass and, in turn, when moved creates an inertial force. A previous study on weight perception showed that the ability to sense the mass of an object improves when it is moved, and it is hypothesized that this enhancement may be due to the availability of inertial cues (Brodie and Ross, 1985). Thus, the inertial contributions need to be removed in order to compare force detection thresholds at rest or when moving the arm with the same stimuli.

Understanding the mechanism governing the perception and detection of weak forces can assist in better interpreting how humans dexterously manipulate objects. Given that the fingertips are very compliant at low interaction forces (Srinivasan and LaMotte, 1995), the application of weak forces induces a relatively large finger pad deformation which stimulates mechanoreceptors in the skin (Birznieks et al., 2001). When these cues are muffled or absent, such as when the fingertips are anesthetized or are very cold, it can become quite challenging to interact with the world (e.g., fastening a coat button while wearing gloves).

It should be noted that the ability to display weak forces with a high level of fidelity is relevant to areas beyond our aims indicated here. These include the desire to conduct neurophysiological research to investigate the sensitivity of the nervous system to a mechanical stimulation (e.g., stretching of one’s skin (Panarese and Edin, 2011)) and the creation of training simulators to teach one how to perform highly dexterous tasks (e.g. palpating very soft tissue, assembling fragile piece).

In prior work, a custom-designed one-degree-of-freedom haptic device was developed to accurately render weak forces (inertial cues were negligible). One of the main features of this device was the inclusion of a deformable end-effector, which decoupled the actuator inertia from the end-effector inertia. Preliminary research in this area demonstrated that humans can typically sense forces in the range of 0.05-0.10 N, where the force threshold is defined as the minimum amount of force which is needed for the user to correctly identify the direction of the applied force in 75% of the trials (Baud-Bovy and Gatti, 2010). A limitation to these earlier studies is that the user’s grasp was not monitored. Thus, the contribution of the tactile cues is not known.

Here, we describe the customization and control of a commercially-available force feedback device, which enables the display of weak forces in the range of a few mN with the desired accuracy and precision. To achieve this aim, the device was outfitted with a force sensor, which allows for the control of accurately rendered forces (by implementing a force feedback controller). Additionally, two more sensors were affixed to the haptic device to monitor the user’s impedance by measuring the user’s applied grip force. A large part of this paper is devoted to the analysis of the force feedback controller, and its impact on system transparency and stability.

The manuscript proceeds as follows. In Section 2, a description is provided for how the device was customized with sensors to improve its transparency and monitor the human’s interaction. In Section 3, the haptic device is characterized under both open- and closed-loop control, and the system response is analyzed in the frequency- and time-domain. Additionally, we model the response of the sensors monitoring the grip force and describe the calibration procedure. In Section 4, the system response is characterized based on data collected from 15 participants who interacted with the setup during an experiment. Findings demonstrate that the customized haptic device was reliable for rendering weak forces with various users and interaction methods. In Section 5, the results are summarized and research directions using the experimental apparatus are proposed.

2. Materials and Methods

The experimental setup described in this paper has two subsystems: i) a force feedback device customized to render the force more transparently (while maintaining stability) and ii) a system to measure the user’s applied grip force. The whole system is controlled by a program running on a PC with an AMD Sempron processor and Windows XP operating system. A 16-bit PCI DAQ card (PCI 6034E, National Instruments Corpora-
related artifacts arising from the device (McJunkin et al., 2005; Salisbury et al., 2009; Parietti et al., 2011). This is particularly salient when rendering a free-space environment, because the mass of the device is clearly felt when moving it around.

To address this transparency issue, it is possible to possible to implement feedforward and feedback controllers (Carignan and Cleary, 2000; Lawrence, 1993; Sánchez et al., 2012; Griffiths et al., 2008; Gurari et al., 2013; Colgate and Brown, 1994). Feedforward controllers use a model of the device to compensate for the system’s dynamics. For example, the Omega.3 default controller uses a model of the device structure to compensate for the effect of gravity in a feedforward manner. Feedforward controllers are, however, challenging to implement because they are sensitive to modeling errors. They are limited in their capacity to compensate for inertia by the difficulty to estimate the acceleration accurately, and they also have limitations at low speeds in that they cannot cancel static or Coulomb friction at velocity reversals (Bernstein et al., 2005). For these reasons, we decided to implement a feedback controller to compensate for rendering errors by outputting a correctional force (see Section 2.1.2).

Feedback controllers require the device to be equipped with additional sensors to compute the error between the actual and desired output. To that end, we customized the system with a six-degree-of-freedom force/torque sensor (Nano17, ATI Industrial Automation, Apex, NC, USA), which was affixed via a mounting-side attachment piece and a printed circuit board (PCB) (see Section 2.2 for more discussion about the PCB). We placed the force/torque sensor as close as possible to the user interface, with the aim of identifying the exact forces which were being transmitted to the user. This is because the co-location of the actuator and sensor is important with force feedback controllers to avoid mechanical losses (Colgate and Hogan, 1989).

The device has a right-handed reference frame, with its X- and Y-axes in the horizontal plane (the X-axis points away from the circular Omega.3 basis) and the Z-axis points upward. It connects to the PC through a USB port and is controlled via the Force Dimension DHD C API, which provides an interface for reading the current device position and commanding a desired force output. The entire system and close-up of the user’s interaction and customized end-effector are shown in Figure 1. The reference system of the haptic device is used for describing the forces sensed by the ATI sensor, and the PC communicates with the ATI sensor and Omega.3 via the PCI DAQ card. We created the virtual environment along the Y-axis so that gravitational forces would not be an influencing factor on both perception and rendering performance.

2.1. Force Feedback System

2.1.1. System Overview

The setup that we developed to render weak forces accurately is based on the commercially-available impedance-controlled Omega.3 haptic device (Force Dimension, Nyon, Switzerland). This device, like the Phantom device (Geomagic®, Research Triangle Park, NC, USA), has a lightweight mechanical structure, which is intrinsically more transparent when displaying free-space motions than devices with a heavier structure and gears, such as the HapticMaster (Moog, East Aurora, NY, USA) and Virtuose™ 6D (Haption, Soulgé/Ouette, France). It has a parallel architecture, which allows translational motion along three dimensions (Grange et al., 2001), and a triangular plate connects its three arms.

The Omega.3 device is normally controlled by an impedance control law, where the position of the device is measured and used to command a desired resistive force. The lightness and backdriveability of this device allow the user to move it without motorized assistance. In contrast, heavier and more powerful devices such as the Haptic Master or Virtuose are controlled by an admittance control law and required motorized assistance to be moved (see Ueberle and Buss (2004) for more discussion on impedance and admittance controllers).

As noted in the Section 1.1, the force actually rendered by a haptic device can differ from what is desired because of system-
measured by the ATI sensor, and \( F_{\text{meas}} \) is the filtered interaction force (as discussed below). The closed-loop proportional gain, \( k_p \), was empirically selected as the largest value which could be implemented by the authors while maintaining stability (see Section 4.4 for further discussion on this topic).

While a feedback controller has been demonstrated as an effective method for improving a device’s transparency, it can lead to instabilities, or the creation of undesired oscillations (Lawrence, 1993). In other words, there is a tradeoff between stability and transparency. In our system, the transparency along the Y-axis depends on the proportional feedback gain, \( k_p \), where a larger value gives enhanced performance. Increasing \( k_p \), however, eventually leads to system instabilities, where the region of stability depends on numerous factors including the device’s dynamics, end-effector location in the workspace, sensor resolution, haptic update rate, and user interaction (Abbott and Okamura, 2005; Colgate and Brown, 1994).

Two features were included in our system to improve stability. First, the proportional feedback gain value was reduced progressively when the end-effector approached the limits of the haptic device’s workspace, since the device was more likely to go unstable in this region. More precisely, the gain was progressively tapered off when the end-effector position exceeded 6 cm from the workspace center along the Y-axis. This allowed for a better transparency in the center of the workspace while maintaining stability at its limits.

Second, the ATI sensor resolution was improved by continuously acquiring force data from the PCI DAQ card and implementing an on-line filtering scheme. Ellis et al. (1996) cite that an increase in the force sensor’s dynamic range may enhance the system’s transparency when rendering environments that are nearly zero impedance. By obtaining the cleanest and most accurate force signal possible, the signal can be fed back into the control loop and amplified to obtain better transparency. Selecting a single sampled force measurement for use in the controller would give a noisy estimate, and feeding back the average of the force measurements would provide a poor prediction of the subsequent force measurement. Here, the accuracy of the force signal was improved by oversampling the force data at a higher frequency than the control loop and fitting a linear regression to these samples.

The DAQ was configured using the NIDAQmx API to acquire force samples at a rate of 33 kHz and store them in a buffer using Direct Memory Access (33 samples corresponds to data acquired during 1 ms at 33 kHz). The ATI sensor force resolution, as indicated by the manufacturer’s specifications, is 0.006 N, and based on measurements made when the sensor was unloaded, is in the range of 0.006 – 0.007 N along the Y-axis. When a force measurement was requested by the haptic loop (approximately every millisecond), a linear regression was fit to the last 33 force samples in the buffer and filtered on-line. Then the value of the force at the present time, \( \hat{F}_{\text{meas}} \), was computed based on the regression parameters. The ATI sensor force noise was reduced to 0.001 N with this filtering scheme.

A more complicated feedback controller could be implemented. First, one could include an integral and/or derivative term. Empirical testing with integral and derivative terms on our system did not give better outcomes over using a simple proportional controller. The integral and derivative terms rely on the history of noisy signals, the derivative of noisy signals, and possibly filtering; thus, these methods can amplify sensor noise and add a phase lag, resulting in the selection of less aggressive feedback gains to maintain stability.

Second, a model-based feedforward term can also be included in the controller to compensate for system-generated forces, such as frictional and inertial forces. Empirical testing did not show tested feedforward controllers to improve the performance of our device. Information for how to optimize controllers to enhance system performance while maintaining stability can be found in (Adams, 1999; Griffiths, 2008; Kuchenbecker, 2006; Lawrence, 1993; Diaz et al., 2010).

To constrain the user’s motion to the Y-axis, the motion of the device along the X- and Z-axes was restricted via a strong virtual visco-elastic force field:

\[
F_{\text{cmd}} = k_d(x_{\text{meas}} - x_{\text{des}}) + k_d v + F_g,
\]

where the desired device position, \( x_{\text{des}} \), is the orthogonal projection of the measured device position, \( x_{\text{meas}} \), along the Y-axis, \( k_i \) and \( k_d \) are respectively the stiffness and damping of the visco-elastic force field, and \( F_g \) is the gravity compensation force computed by the Force Dimension DHD API.

2.2. End-Effector and Grip Force Measurement System

The grip force measurement system developed is shown in Figure 1(c). The end-effector was designed such that a user grasps it between the thumb and index finger with a pinch, or key grip configuration. That is, the user squeezes the end-effector between the volar portion of the thumb and the distal interphalangeal joint of the index finger (see Figure 1(b) for a representative image). This hand configuration was selected based on empirical testing, since it was a comfortable hand orientation which participants could sustain over possibly numerous hours of testing. The grip force measurement system was built to add as little mass as possible on the user side, with the aim of minimizing its impact on the ATI sensor’s measurements and the overall inertia of the system.

The user’s grip force, as applied by each finger, was measured using Force Sensing Resistors (FSR) \(^{TM}\) 400 Interlink Electronics, Camarillo, CA, USA). These are symmetrically located at vertical and horizontal distances of 14.8 mm and 33.5 mm, respectively, from the mounting side of the Omega3 triangular plate. Each FSR was sandwiched between an end-effector base attachment piece and an user interface piece. These two parts were two-material composites with a stiff Shore D Hardness 86 outer side and a softer Shore A Hardness 61 internal side. The harder regions interfaced with the device and user, while the softer regions interfaced with the FSR so that the user’s applied pressure more evenly distributed to the sensor.

The FSR can sense forces in the range of 0.1-10 N with a repeatability of ±2%, as given by the manufacturer’s specifications. Each FSR was connected in series to a voltage divider...
circuit (Switchable Voltage Divider-1134, Phidgets Inc., Calgary, Alberta, Canada) with a resistance of \( R = 8.45 \, \text{k}\Omega \), and communications were transmitted through thin wires that intersect on the custom PCB. These wires were bent to make them spring-like in behavior. In this way, the tension between the FSR and PCB was reduced, and the undesired forces were not heavily influencing the ATI sensor measurements. The PCB was placed between the Omega.3 triangular plate and ATI sensor so that its weight would not contribute to the ATI sensor measurements. The total mass of the customized end-effector and grip force measurement system is 24 g.

3. System Characterization

An aim of this work was to create a system which could be used to identify force detection thresholds, and additionally, which could control the user’s grasp. Prior to data collection with human subjects, first we needed to verify that the system worked properly. Below we discuss the methods and analyses used to validate that the system would be suitable for human subject testing. Specifically, this section reports on the Omega.3’s transparency in both the frequency- and time-domain for the collected open- and closed-loop control data, as presented in Section 3.2.

3.1. Characterization Methods

Haptic devices can be characterized by their transparency, or ability to virtually render an environment without distortions (Adams, 1999; Griffiths, 2008; Kuchenbecker, 2006). Transparency can be defined as “the ratio between the transmitted and simulated impedance where the ideal ratio is unity” (Lawrence, 1993; McJunkin et al., 2005). Another measure of performance is the Z-width, which identifies the set of proportional and derivative controller gains which can be implemented on a haptic device while maintaining stability (Colgate and Brown, 1994).

While frequency-domain analyses are more commonly used to model systems (Gurari et al., 2013; McJunkin et al., 2005; Griffiths et al., 2008), they capture only part of what is happening, especially when the device is rendering near free-space environments. Coulomb friction becomes a more pertinent contributing factor to the internally generated forces when displaying weak force and free-space environments (Salisbury et al., 2009; Parietti et al., 2011; Abbott and Okamura, 2005; Tashmasebi et al., 2005; Ellis et al., 1996). However, Coulomb friction is nonlinear, so the frequency-domain system characterization methods become much less relevant since they assume system linearity. To-date, the aim of researchers when characterizing their systems have tended to focus on creating high impedance environments (e.g., rigid surfaces or walls) rather than low-impedance environments (Colgate and Brown, 1994; Schauß and Peer, 2012). The latter is the focus of our contribution here.

3.2. Data Collection and Preparation

To identify the transparency of the system, data were collected when the experimenter translated the end-effector along the Y-axis during both open-loop control \((k_p = 0)\) and closed-loop control \((k_p = 10)\). A null force was rendered \((F_{des} = 0 \, \text{N})\), and the experimenter moved the device across the workspace for a total of 59 s – 30 s of slow motion, 20 s of normal motion, and 9 s of fast motion. The position and force data were saved at 1 kHz for off-line analysis. The manner with which the system was interacted during both open- and closed-loop control is shown in Figure 2. These data were analyzed to make indications as to whether the customized and controlled device would allow for force detection threshold testing (has a rendering fidelity which is better than 0.05 N (Parietti et al., 2011;
Table 1: RMS of velocity, acceleration, and force error during slow, normal, and fast interaction speeds when the system was controlled in open- and closed-loop.

<table>
<thead>
<tr>
<th>Speed, m/s</th>
<th>Velocity, ( \dot{x} )</th>
<th>Acceleration, ( \ddot{x} )</th>
<th>Force Error, N</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow</td>
<td>0.034</td>
<td>0.166</td>
<td>0.327</td>
</tr>
<tr>
<td>Normal</td>
<td>0.095</td>
<td>0.800</td>
<td>0.394</td>
</tr>
<tr>
<td>Fast</td>
<td>0.124</td>
<td>1.489</td>
<td>0.582</td>
</tr>
</tbody>
</table>

Data were filtered to remove noise from the signals. To that end, we used a smoothing spline method, which pieces together cubic polynomials at points called knots so that the function, along with its derivatives, is continuous (Wood, 1982). The smoothing spline gives an analytical function which can be used to compute derivatives. The amount of smoothing is controlled by a parameter which gives the relative weight of the squared second derivative of the spline in the fit. A thorough and nice review of smoothing splines as well as other methods to smooth and differentiate data is provided in Wood (1982).

We used the smooth.spline function in R with all data points as knots and a high level of smoothing \((\text{spar} = 0.9)\) to filter the position data, \(x\). Velocity, \(\dot{x}\), and acceleration, \(\ddot{x}\), were computed by taking the respective first and second derivatives of the position’s cubic smooth splined results.

Table 1 reports the main characteristics of the excitation movements. The root-mean-square (RMS) of the velocity and acceleration for the end-effector’s movements show that movements were comparable for the two data sets, giving a valid basis from which to compare the transparency in the open- and closed-loop conditions. For a sinusoidally-shaped waveform, the RMS corresponds to 71% of the sine amplitude. Moreover, Table 1 show the RMS force error was much smaller in the closed-loop condition than in the open-loop one, but increased with speed in both conditions.

### 3.2.1. Frequency-Domain Response

The MATLAB function \texttt{tfestimate} was applied to obtain a linear, time-invariant transfer function, which is quantified as the quotient of the cross power spectral density of the input and output functions and the power spectral density of the input function. Here, the input was position and the output was force, and analyses were conducted using 20 periodic hanning windows and 50% section overlap. Responses are given in Figure 3 and resemble a damped second-order system.

The ideal magnitude response is 0 N/m across all frequencies, which corresponds to a system with no effective mass or damping. Results demonstrate that this ideal scenario is not achieved. However, the closed-loop controller has a significant impact on the system’s performance in terms of masking the device’s friction and inertia. The mean magnitude response for data spanning 0.9 to 2.5 Hz during open- and closed-loop control is respectively 44.73 ± 16.80 N/m and 4.47 ± 1.96 N/m. This demonstrates that the force controller improves the system’s performance by a factor of 10, corresponding to the proportional feedback gain, \(k_p\). Additionally, the plot demonstrates that as the interaction frequency increases, the system’s performance worsens.

The ideal phase response is 0°, which corresponds to the sensed force being perfectly in-phase with the measured position. The 90° out-of-phase response at low frequencies demonstrates the viscous damping in the system, with position lagging force. The 180° out-of-phase response at higher frequencies demonstrates the mass in the system, again with position lagging force. Thus, the lower frequencies are dominated by the system’s viscous damping, and the higher frequencies are dominated by the system’s mass.

Note that the to-and-fro movement during the last 5 seconds of acquisition reached a maximum frequency of approximately 2.25 Hz during both open- and closed-loop control. The maximum interaction frequency was due to safety checks in the software, which turned off the controller at high-interaction forces (which corresponds to fast movements).

### 3.2.2. Time-Domain Response

As noted in Section 3.1, frequency-domain methods are not well suited to model non-linear systems. The objective of the time-domain analysis is to estimate the values of the parameters which characterize the system’s dynamics and affect the measured force, such as the mass, viscous friction, and non-linear Coulomb friction. To that end, we modeled the force as:
The model parameters were estimated using a least-squares regression and are reported in Table 2. All estimated parameters were statistically significant (p < 0.001). The model explains 91.1% of the variance in the force signal during open-loop control and 94.5% of the variance during closed-loop control, as indicated by the adjusted $R^2$ value (all interaction speeds inclusive). Comparing the parameters in the open- and closed-loop conditions demonstrates that the closed-loop controller improves the device’s transparency by masking the system’s natural behavior by an amount that is proportional to the $k_p$ value; all parameters decrease by approximately a factor of 10.

It should be noted that if the model is applied to only the linear components of mass and viscous friction during closed-loop control, then the model explains 75.9% of the variance in the force signal. The parameter estimates become $m = 0.027$ kg and $b = 0.292 \frac{N}{s}$, where the viscous friction estimate increases by nearly a factor of 10.

Figure 4 gives the breakdown for how the RMS force for each component varies as a function of the user interaction speed during closed-loop control. The contribution of the Coulomb friction, viscous friction, and mass to the force error is computed as the RMS of $c \frac{\ddot{x}}{|\dot{x}|}$, $b \dot{x}$, and $m \ddot{x}$, respectively. At slow interaction speeds the Coulomb friction is the main contributing force error source, while at fast speeds mass is the main source. Viscous friction does not have a large impact and is less of a concern across all speeds when compared to the device’s Coulomb friction and mass. This analysis demonstrates the increase of the force errors with movement speed in Table 1.

3.3. Grip Force Measurement System Modeling and Calibration

The second subsystem of the experimental setup is the grip force measurement system, which is based on data obtained from the uncalibrated FSR sensors. In this section, we describe the modeling of the sensors, present the calibration procedures followed, and report on the precision of the grip force measurements.

Table 2: Estimated model parameters during open- and closed-loop control, along with the corresponding goodness-of-fit.

<table>
<thead>
<tr>
<th>Model Parameter</th>
<th>OL Control</th>
<th>CL Control</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coulomb Friction, $c$, N</td>
<td>0.244</td>
<td>0.025</td>
</tr>
<tr>
<td>Viscous Friction, $b/\dot{x}$</td>
<td>1.036</td>
<td>0.041</td>
</tr>
<tr>
<td>Mass, $m$, kg</td>
<td>0.290</td>
<td>0.027</td>
</tr>
<tr>
<td>$\sigma$, N</td>
<td>0.119</td>
<td>0.009</td>
</tr>
<tr>
<td>$R^2$, %</td>
<td>91.9</td>
<td>94.4</td>
</tr>
</tbody>
</table>

$$F_{meas} = \beta_0 + c \frac{\ddot{x}}{|\dot{x}|} + b \dot{x} + m \ddot{x} + \varepsilon. \tag{3}$$

The coefficients $c$ and $b$ correspond to Coulomb and viscous friction, $m$ corresponds to the apparent mass, $\beta_0$ is an offset which can take into account possible biases in the sensed force, and $\varepsilon$ is unexplained system noise.

The resistance of the FSR sensor, $R_{FSR}$, varies as a function of the inverse of the user applied force, $F$:

$$R_{FSR} = a + \frac{b}{F}, \tag{4}$$

where $a$ and $b$ are unknown parameters. Given that the FSR sensor was mounted to a voltage divider circuit, the relationship between the input voltage to the circuit, $V_{in} = 5$ V, and the output voltage of the FSR/Voltage Divider circuit, $V_{out}$, is:

$$V_{out} = V_{in} \left( \frac{R}{R + R_{FSR}} \right), \tag{5}$$

where $R = 8.45$ k$\Omega$ is the fixed value of the resistor in series with the FSR sensor in the voltage divider circuit. Plugging Equation 4 into 5 and rearranging gives the following relationship between the output voltage and the user applied force:

$$F = \frac{bV_{out}}{RV_{in} - (a + R)V_{out}}. \tag{6}$$

Values of the unknown parameters $a$ and $b$ can be obtained by calibrating the sensors using a known force.

The top FSR sensor was calibrated by commanding the haptic device to apply a sinusoidally-shaped time-varying force ranging from 0 to 12 N upward while the experimenter held the position of the device constant with the thumb placed on the top FSR in the upper “half” of the key grip. During the calibration of the bottom sensor, the device applied a downwards force while the experimenter held the position of the device with the index finger on the bottom FSR in the bottom “half” of the key grip configuration. Both the output voltage of the voltage divider circuit and the force applied on the FSR sensor (as measured by the ATI sensor along the vertical direction, or Z-axis) were saved at 250 Hz and used for the calibration of the
The values of the free parameters $a$ and $b$ were obtained by fitting the force predicted by the force-voltage model, as given by Equation 6, using a gradient descent algorithm, which minimized the sum-of-squares error between the force measured by the ATI sensor and the force predicted by the model.

Figure 5 displays the calibration data for the top FSR, giving the actual force measurements (as obtained by the ATI sensor) and the modeled force (as predicted by Equation 6 with the parameters $a$ and $b$ obtained from the calibration procedures) as a function of the output voltage. This figure shows that the model of the force-voltage relationship is appropriate, yet not very precise. The mean absolute error between the actual and modeled parameters is 0.850 N. Reasons for the lack of precision include that the FSR sensor is quite sensitive to the manner by which pressure is distributed, and the sensor exhibits hysteresis which is not accounted for in the model.

Taking a closer look at Equation 6 shows that the model has an asymptote and predicts an infinite force when the output voltage is $V_{\text{asymptote}} = V_{\text{in}} \left( \frac{R}{R + a} \right)$. This asymptote should not be reached, in practice, as long as the force applied to the sensor does not exceed the maximum force used during the calibration procedures, $F_{\text{maxCal}}$. To allow for relatively accurate estimates of applied forces which are greater than $F_{\text{maxCal}}$, we modified Equation 6 by linearizing the output voltages which exceeded the maximum voltage used during the calibration procedures, $V_{\text{maxCal}}$:

$$F = \begin{cases} \frac{bV_{\text{in}}}{dV_{\text{out}}} (V_{\text{out}} - F_{\text{maxCal}}) + F_{\text{maxCal}}, & \text{if } V_{\text{out}} < V_{\text{maxCal}}; \\ \left( \frac{dF}{dV_{\text{out}}} (V_{\text{out}} - V_{\text{maxCal}}) + F_{\text{maxCal}}, & \text{otherwise.} \end{cases}$$

$\frac{dF}{dV_{\text{out}}}$ is the derivative of Equation 6 with respect to $V_{\text{out}}$ estimated at $V_{\text{maxCal}}$.

$$\frac{dF}{dV_{\text{out}}} = \frac{bV_{\text{in}}R}{[R(V_{\text{in}} - V_{\text{maxCal}}) - dV_{\text{maxCal}}]^2}$$

Figure 5 shows that Equation 6 predicts an infinite force when the measured voltage, $V_{\text{out}}$, reaches the asymptote, $V_{\text{asymptote}} = 4.545$ V, and it depicts the linear extrapolation scheme implemented to deal with such a possibility. Even though the estimates with the modified model somewhat underestimate the true force when $V_{\text{out}} > V_{\text{maxCal}}$, the force estimated in this manner never returns meaningless infinite (or even negative) values as it would with the nonlinearized equation.

Repeated measurements spaced over several months demonstrate that the FSR sensor calibration remains valid as long as the user applies the pressure in a similar manner.

4. User Interaction

In this Section, we analyze the quality of the haptic interaction that took place during a psychophysical study, which aimed at measuring the minimum force one can perceive. In particular, the goal of the study was to inform on the effect of grip force on one’s ability to sense a weakly applied force when the end-effector is held fixed in space or is moved back and forth. The general objective of this analysis is to verify that the device can render forces with an accuracy which is less than 0.05 N; we validate that the system is suitable for the purpose of the study when slow to normal interaction speeds were employed. For this manuscript, the participants can be viewed as an “excitation source”, since the movements they produced allow for us to characterize the system’s performance under various user interactions. The behavioral results are not analyzed here, but, rather, they will be presented in a separate article.

Specifically, the first objective of this section is to show that the system rendered a constant force with a precision which allowed for one’s ability to detect weak forces to be evaluated. The second objective of this analysis is two-fold: to identify whether the system was stable during the interaction, and to determine whether the system had a comparable level of transparency across the tested range of load forces.

4.1. Data collection

15 participants partook in the study. All provided their informed consent, were naive to the task, and were healthy with no neurological illnesses, neuromuscular disorders, or medical concerns. Participants self-declared experience with haptic environments ranged from ‘None’ to ‘Having participated in numerous demos’. Three participants were IIT employees who volunteered their time, and the remaining were non-IIT employees who were paid 30 Euro.
The experimental setup described in the previous sections was used to display the virtual force and to measure the grip force. At the beginning of each trial, the haptic device displayed a load force along the Y-axis, which ramped up from 0 N to one of nine possible target levels ranging from –0.20 to 0.20 N in increments of 0.05 N.

Four user interaction methods were employed, which corresponded to the combination of two levels of grip force (Low and High) and two modes of interaction (Static and Dynamic). The grip force, defined as the amount of squeezing applied between the thumb and index finger, was controlled throughout the trial by visually relaying to the participant the force sensed by each FSR in the grip force system. The desired grip force levels corresponded to 0.75 N and 6.25 N in the Low and High conditions, respectively. The interaction mode was set by instructing the participant to hold the hand still in the Static condition, and by allowing the subject to move the hand back and forth in the Dynamic condition.

During each trial, data recorded included the force displayed by the device (as measured by the ATI force sensor), force applied to the top and bottom FSR sensors, and motion of the device. Data were saved at 250 Hz for off-line analysis. The experiment was comprised of a total of 360 trials per participant, in which one of 36 possible combinations (corresponding to the nine load forces, two exploration modes, and two grip forces) were repeated 10 times. Altogether, data were collected for 5,400 trials.

4.2. Data Pre-Processing

The original 5,400 trials were pre-processed to remove those which did not provide meaningful information for characterizing the system. The first 6 s of an example trial is given in Figure 6. For every trial, the force linearly ramped up to the desired load force across the span of 3 s; thus, trials which did not span this length of time were discarded from future analyses. Additionally, the closed-loop controller was set to be fully active between ±6 cm with relation to the workspace center. Therefore, any trial in which the user’s movement exceeded this distance at some point was thrown out. In total, 161 trials were removed, or 3% of the original data set.

Every trial was classified as a No Motion or Motion trial, depending on whether or not the average trial velocity exceeded a threshold value of 0.013 m/s. The threshold was defined by identifying the average velocity across all trials, and visual inspection concurred that this metric was reliable for identifying system motion. The classification between a No Motion and Motion trial does not exactly correspond to the classification between a Static and Dynamic trial, since some participants chose not to move the device or made a very small motion during the Dynamic trials. Thus, the classification of No Motion/Motion is used in place of the Static/Dynamic classification to inform on the effect of the interaction mode on system performance for the following analyses.

4.3. User Grip Force Interaction

The average grip force values applied to both the top and bottom FSR ± standard deviation (SD) were 0.97 ± 0.43 N and 6.19 ± 0.67 N in the Low and High conditions, respectively. This demonstrates that all participants adjusted the grip force according the level specified on the visual display.

4.4. System Stability

The likelihood of observing an instability increases as the controller is made more aggressive (Lawrence, 1993). Figure 6 gives an example force response when an instability occurs. Participants commented on feeling this buzzing sensation at various points throughout the data collection process. In this section, every trial was examined to determine when the system went unstable. This analysis is a novel method presented for identifying instabilities based on data in the time-domain.

The occurrence of an instability was identified as when the residual error between the measured and modeled force was greater than one standard deviation from the mean estimate of all residual errors. A large error between the measured and modeled force implies that there is some unexplained "noise" present, which may take the form of this instability, and visual inspection concurs that there indeed was an instability/vibration/buzzing in the system.

The device went unstable in 99, or 1.9%, of the analyzed trials. A logistic regression was used to test the effects of grip force and interaction mode on the stability of the system. This analysis showed that the probability of observing an unstable trial increased in the High grip force ($\chi^2_{1, N=5239} = 6.214, p < 0.005$) and Motion ($\chi^2_{1, N=5239} = 1.873, p < 0.034$) conditions. Figure 7 shows the probability of observing an unstable trial for each combination of grip force and interaction mode. 
4.5. System Transparency

To analyze the system transparency, the 99 trials with bouts of instability were excluded, leaving 5,140 trials for the following analysis. For all trials, we computed the RMS force error, or the RMS of the difference between the measured and desired force along the Y-axis. The mean value ± SD of the RMS force error was 0.019 ± 0.003 N for the No Motion trials and 0.017 ± 0.004 N for the Motion trials. Overall, the mean value ± SD of the RMS force error across all trials was 0.018 ± 0.004 N.

We also computed the maximum absolute value of the force error for each trial to measure to what degree the rendered force could differ from the desired force. The mean value ± SD of the maximum value of the absolute force error was 0.041 ± 0.013 N, and the distribution is shown in Figure 8. 0.05 N was exceeded in 10.4% of the trials, with 8.5% of all trials lying between 0.050 N and 0.075 N. Less than 0.6% of trials exceeded 0.1 N. Visual inspection of the trials with a large maximum force error typically exhibited the presence of a single very brief and isolated peak in the force error signal. The occurrence did not seem to be closely related to the movement of the user and may correspond to a very brief bout of instability caused, for example, by an occasional delay in the haptic feedback loop (due to the operating system).

A three-way analysis of variance (ANOVA) was run to determine the impact of absolute load force, interaction mode, and grip force on the RMS force error. Rendering transparency was significantly affected by the interaction mode ($F_{(1,5120)}=244.222$, $p<0.005$, $\eta^2=0.045$), which coincides with the results in Section 3, where the RMS force was observed to increase as a function of user interaction speed. The absolute load force ($F_{(4,5120)}=6.044$, $p<0.005$, $\eta^2=0.004$), grip force ($F_{(1,5120)}=7.094$, $p=0.008$, $\eta^2 = 0.001$), and interaction between interaction mode and grip force ($F_{(1,5120)}=4.195$, $p=0.04, \eta^2=0.001$) also all exhibited significant effects. Even so, the effect size is quite small for all parameters, indicating that these factors have very little influence on the system’s transparency (less than 0.5%). The significance was most likely obtained due to the large number of samples used in the analysis (5,140 trials).

Figure 9 shows the RMS force error for all trials as a function of the average velocity and acceleration in each trial. Black and gray dots indicate trials in the No Motion and Motion conditions, respectively. For the No Motion trials, the force error varied greatly from trial to trial, but in a range that is approximately limited by the estimated value of the Coulomb friction, which was 0.025 N (see Table 2). This estimate is indicated in the figure by the dashed horizontal line, and is the primary
force derived from the device for low interaction speeds, as was discussed in Section 3. Hence, during the No Motion condition participants were essentially interacting with the Coulomb frictional force. For the Motion trials, the force error was mostly dependent on the inertia, as expected from the characterization of the system in Section 3. Thus, the RMS force error increased as the acceleration increased due to the dependency of the inertial force on the user’s motion.

5. Discussion

An aim of this work is to encourage the use of haptic devices in sensorimotor control studies in a manner which is informed so that meaningful results can be obtained from the data which is collected. A lack of stability and/or transparency may be perceptible to the human interacting with the haptic environment, and it may influence his/her ability to identify a particular haptic sensation and/or to interact with a virtual object. Recent work is underscoring the benefits of using haptic devices for such testing scenarios (Kawai et al., 2012); however, there exist possible pitfalls to using these devices if one is not aware of their limitations.

5.1. System Control, Stability, and Transparency

Ideally, a haptic device renders a perfectly stable and transparent environment. Realistically, this is not feasible due to hardware and software constraints (e.g., friction and inertia in the system, time lags, controller update rate, sensor resolution, maximum force output capabilities (Hannaford, 1989; Lawrence, 1993; Salisbury et al., 2009; Tatti et al., 2014)). When working with such devices, one should first ensure that the system is stable, and then second attempt to create a transparent virtual environment (McJunkin et al., 2005). Moreover, it is advisable to identify the desired level of transparency, feasible level of transparency, and controller which would achieve the desired rendering results (Lawrence, 1993). Additionally, the system’s transparency should be quantified across the range of environments which will be rendered (Schauß and Peer, 2012).

The aim here was to render force errors which were less than that of human force detection – 0.05-0.10 N (Baud-Bovy and Gatti, 2010). To achieve this, first the experimenter characterized the response of the system based on her controlled interaction with the system. She translated the Omega.3 device for nearly a minute when it was open-loop controlled, and the system’s response demonstrated that the internally generated forces due to the device’s mechanical attributes would be perceptible to the user, with RMS force error estimates spanning 0.327 N to 0.582 N (nearly a factor of 10 greater than perception). To address this issue, the haptic device was customized with a force sensor and a closed-loop controller was implemented with proportional feedback on force. The response of the system under this scenario was identified to be good enough to allow for perceptual testing of force thresholds (users would not detect the internally generated forces when movements were in the Slow to Normal interaction speed range, as forces were < 0.05 N).

Next, data were collected with 15 users when manipulating the device under various controlled conditions. Findings demonstrate that rendering performance was accurate, with RMS force errors of 0.019 ± 0.003 N when the device was held still and 0.017 ± 0.004 N when it was in motion. These values are less than half the force threshold of humans. The mean value ± SD of the maximum absolute force error observed during each trial was 0.041 ± 0.013 N, which is near the human threshold. We hypothesize that a very short peak force which is near or slightly above the sensory threshold would not be detected, since detection of a stimulus may depend both on its intensity and duration (temporal summation). In the tactile modality, for example, the detection threshold for vibrations delivered over a large surface area decreases as the stimulus duration increases (Gescheider et al., 2004). These findings demonstrate that the implemented closed-loop controller with the selected proportional gain of $k_p = 10$ was appropriate for obtaining improved rendering fidelity while avoiding the creation of undesired force artifacts across multiple users.

Another challenge faced when using force feedback devices and closed-loop controllers is ensuring that the system remains stable. Transparency is a function of the system, while stability is dependent on both the system and user. The likelihood of observing an instability increases as the controller aims to enhance the rendering fidelity (Lawrence, 1993). Additionally, variations between users, as well as variations within a single user’s arm impedance (Hogan, 1989) impact the system’s stability. A number of researchers accounted for user variations in their setups by methods including customizing the controller for each user (Gurari et al., 2013) and controlling the user’s applied grip force when characterizing the system (Kuchenbecker, 2006).

To verify the stability of the control loop, we examined the stability of the system based on the data collected with the 15 users. A small percentage of the trials gave instabilities (1.9% out of 5,239 trials), with a majority of these trials occurring in the High grip force condition when the hand and arm of the user are presumably more rigid (due to a co-contraction of the muscles when squeezing the object). This observation is well in agreement with the conclusions of Hannaford and Anderson that the system’s stability is dependent on the human operator’s impedance (Hannaford and Anderson, 1988).

5.2. System Modeling

We aimed to identify from where the system generated forces were derived. Therefore, we modeled the force errors arising in the system using the data collected from the experimenter who translated the device across the mechanical workspace in a controlled manner for 59 seconds at various speeds. We should note that the system characterization was done in a particular Cartesian reference frame and is valid only for the trajectory employed here. Varying system responses and characterizations will be obtained, depending on the region of the system’s mechanical workspace (Tatti et al., 2014) and how it is explored during the data collection process. Even so, these data allowed for a more complete description of the system, since they were rich in terms of movement speeds and device locations.
Typically, a frequency-domain analysis is conducted in which an assumption is made that the system is linear; thus, the contributing factors to the internally generated forces are the system’s mass and viscous friction (and possibly an elastic spring) (Gurari et al., 2013; Griffiths et al., 2008; Carignan and Cleary, 2000). We modeled the system in this manner, as it can provide an intuitive description of the system’s performance across a range of frequencies.

However, a number of works demonstrated that Coulomb friction is also a contributing factor to the internally generated forces, albeit a nonlinear parameter (Salisbury et al., 2009; Parietti et al., 2011; Abbott and Okamura, 2005; Tahmasebi et al., 2005; Ellis et al., 1996). Coulomb friction becomes much more pertinent when rendering weak forces, which was the aim of this work. Hence, it was crucial to characterize the nonlinear Coulomb friction, since it would be a main contributing factor when rendering forces with a magnitude ranging from 0.00 N to 0.20 N.

Therefore, we modeled the system as having the linear components of viscous friction and mass, as well as the nonlinear component of Coulomb friction. A linear regression estimated these parameters for the range of tested speeds. This time-domain manner by which we characterized the system is not well adopted in the haptics literature, where only a small number of other researchers have chosen to perform similar analyses (Flanagan and Wing, 1997b; Parietti et al., 2011; Abbott and Okamura, 2005; Tahmasebi et al., 2005). These other works also identified the importance of the Coulomb friction, and, hence, Incorporated it into their system models.

Results from our characterization demonstrate that our model was accurate, since it explained 94.5% of the variance in the force errors. A model based on solely the linear viscous friction and mass components explains only 75.9% of the variance in the force errors, and the viscous friction estimate increases by nearly 10-fold to account for the unexplained Coulomb friction component. These findings correlate well with those obtained in the frequency-domain, yet they put less emphasis on the viscous friction component, as Coulomb friction is modeled.

Another contribution of this work was explicitly demonstrating the effect of the system’s physical properties at various interaction speeds based on a time-domain analysis. In general, this point is more readily conveyed using a frequency-domain analysis (Gurari et al., 2013; Carignan and Cleary, 2000). For example, a friction feedback controller is useful for masking force errors in the quasistatic range, but it quickly looses its effectiveness at approximately 0.1 Hz, when the forces generated by the device’s mass take over (Carignan and Cleary, 2000). Here, we demonstrated that Coulomb friction is the main source of force for slow movements (RMS velocity: 0.031 \( \frac{m}{s} \)), contributes about the same force as the mass for normal speeds (RMS velocity: 0.097 \( \frac{m}{s} \)), and contributes less force than the mass for fast movements (RMS velocity: 0.117 \( \frac{m}{s} \)). In addition, our findings concurred with those of Tahmasebi et al. (2005) in that the forces generated from the Coulomb friction were “unanimously higher” than the forces generated from the viscous friction.

The modeling efforts described in this paper can be applied to the characterization of other systems. Depending on the desired rendered force level and user interaction speed, one may choose to consider the Coulomb friction and, in turn, rely on such an analysis which incorporates this nonlinear component. The characterization also is valuable in that it allows for an automatic and accurate identification of instabilities.

5.3. Grip Force Monitoring

Mechanoreceptors in the skin play an important role for controlling one’s grasp, specifically detecting contact with an object and possibly conveying finger slip (Johansson and Flanagan, 2009). In force perception studies where forces are transmitted externally by a hand-held object, monitoring grip force is important since mechanoreceptors in the skin provide information about the contact force in both the normal and tangential directions (Birznieks et al., 2001). The response and degree of excitability of these sensors is known to vary as a function of one’s grip force. As described in Westling and Johansson (1987), some mechanoreceptors adapt quickly during an increase of the grip force and stop firing or fire irregularly when the load is constant. Other mechanoreceptors have a sustained response, but saturate at a relatively low grip force level (1-2 N).

Given these points, we outfitted the aforementioned closed-loop controlled haptic device with a grip force measurement system to allow for monitoring and control of grip force. The task of affixing the grip force measurement system is somewhat contradictory to the aims of improving the device’s transparency (as was presented in Section 5.1). The force feedback control scheme executes best when the force sensor measuring the system’s applied force is located near the location of the user’s grasp. However, in order to study grip force, force sensors needed to be placed in contact with each finger to quantify the user’s grasp. One possible solution was to use the same force sensors for measuring the user’s applied grip force and the system’s interaction force. For example, two multi-axis force/torque sensors could have been mounted to the end-effector of the haptic device to measure grip force and net load force simultaneously.

Here, we chose a different solution which was based on affixing light-weight and low-cost force sensing resistors. This approach was selected to add as little mass as possible in order to keep its impact on the system’s transparency low (see Section 2.2). While this solution gave less precise grip force measurements, it had the advantage of requiring only one relatively heavy and expensive multi-axis force/torque sensor.

Testing with the grip force measurement system demonstrated that, once properly calibrated, two distinct grip force levels (0.75 N and 6.25 N) could be identified across several months of usage with the same calibration settings (see Section 3.3). Given that grip force can influence the ability to perceive an external force (Flanagan and Wing, 1997a) and can inform on the mechanism governing perception, we recommend others to incorporate such a system and calibration procedure into their own devices for controlling and monitoring the user’s interaction.
6. Conclusions

The experimental setup presented here is meant to inform on the mechanism governing weak force perception. While the motivation for developing the system was for studying force detection, the experimental setup could be used in any application requiring high-fidelity force rendering. Moreover, the solutions adopted to model the system, improve the system’s transparency while maintaining stability, and monitor the user’s applied grip force are not limited to the particular hardware used in this study, but could be applied in other contexts.

We made the following adjustments to a commercially-available haptic device to improve its fidelity for human subject testing:

- sensorized and controlled the system using a closed-loop feedback loop to enhance its transparency,
- modeled the system to characterize the effect of various parameters on its stability and transparency, and
- customized the system with a grip force measurement system to monitor and inform about the user’s grasping effort.

Altogether, the device and closed-loop controller performed according to our aim. As indicated by Carignan and Cleary, “it seems likely that force feedback controllers will play a vital role in the successful implementation of future haptic interfaces” (Carignan and Cleary, 2000). The approach described here for improving a device’s transparency while maintaining stability can be applied to different hardware (e.g., haptic devices), and we would recommend investing in a good quality force sensor for implementing such controllers. Panarese and Edin (2011) used a similar approach, attaching an expensive six-axis ATI force/torque transducer (approximately 6,000 USD) to a low-cost Falcon haptic date (<200 USD). By implementing a proportional-integral-derivative (PID) controller on force, they obtained internally generated force errors averaging 0.013 N when the system was stationary and 0.122 N when the system was in motion. Reasons for why force sensors may not have been incorporated in devices to-date are the added mass and cost (Carignan and Cleary, 2000). However, the positive impact on the system’s rendering fidelity, as evidenced here, can make the investment worthwhile. Additional considerations for how to improve the system and controller are discussed in Colgate and Hogan (1989).

To further enhance the device’s transparency, other control algorithms and considerations can be tested. For example, perhaps an admittance control scheme could be designed to give a better system response. Carignan and Cleary (2000) compared admittance control to impedance control, and the former was shown to be less susceptible to instabilities for rendering low impedances. Furthermore, the controller could be adjusted to the user’s grip force, or impedance, in order to achieve improved transparency for a range of interactions while maintaining stability (Hannaford and Anderson, 1988). Here, we demonstrated that our setup “as is”, with a force feedback controller, was suitable for the human subject weak force perception study, so we did not pursue further enhancement of the device’s transparency.

The experimental setup could be also improved by combining the measurement of the system’s interaction force and the user’s grip force by mounting two multi-axis force/torque sensors at the extremity of the haptic device, as described in the previous section.

The customized system can be extended to research areas other than what we identified. It can be used for advising the design of novel haptic devices, identifying sensory deficiencies in patient populations, training individuals in highly dexterous tasks, and informing on the development of sensory perception in young children. A future paper will report on the human perceptual results obtained based on the human subject study that was briefly presented here.

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