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Development of a Haptic Interface for Natural Orifice Translumenal Endoscopic Surgery Simulation

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Abstract

Natural orifice translumenal endoscopic surgery (NOTES) is a minimally invasive procedure, which utilizes the body's natural orifices to gain access to the peritoneal cavity. The NOTES procedure is designed to minimize external scarring and patient trauma, however flexible endoscopy based pure NOTES procedures require critical scope handling skills. The delicate nature of the NOTES procedure requires extensive training, thus to improve access to training while reducing risk to patients we have designed and developed the VTEST[®], a virtual reality NOTES simulator. As part of the simulator, a novel decoupled 2-DOF haptic device was developed to provide realistic force feedback to the user in training. A series of experiments were performed to determine the behavioral characteristics of the device. The device was found capable of rendering up to 5.62N and 0.190Nm of continuous force and torque in the translational and rotational DOF, respectively. The device possesses 18.1Hz and 5.7Hz of force bandwidth in the translational and rotational DOF, respectively. A feedforward friction compensator was also successfully implemented to minimize the negative impact of friction during the interaction with the device. In this work we have presented the detailed development and evaluation of the haptic device for the VTEST[®].

Index Terms

Force feedback; haptic device; surgical simulation

1 Introduction

The advancements in the field of surgery have focused on the development of a variety of minimally invasive surgical access techniques such as laparoscopy, Single Incision Laparoscopy (SILS) and Natural Orifice Transluminal Endoscopic Surgery (NOTES). The primary focus of minimally invasive surgery (MIS) techniques is to reduce post operative patient trauma, which are frequently attributed to obtaining surgical access rather than the actual surgical task [1]. The NOTES procedure is designed to minimize external scarring and patient trauma [2], by gaining access to the target organ through the lumen of the colon, cervix or stomach via the oral, vaginal or anal cavities [3], [4]. The potential benefits of NOTES are driving the surgical community to actively explore the technique, and multi-center clinical trials have been set in motion with some positive results already published [5], [6].

Given the enthusiasm around the development of NOTES procedures [6]–[9], a cautioned approach is still necessary to appropriately address patient safety concerns that were also seen to hinder the further development of laparoscopic cholecystectomy in the early 1990's. As of now the limited clinical exposure, minimal post-procedure follow up data and overall complexity inherent in the NOTES technique warrants a cautioned yet progressive approach to the development of NOTES. With these concerns in mind, a working group called the Natural Orifice Surgery Consortium for Assessment and Research (NOSCAR) was formed. A whitepaper written by the group in 2005 detailed eight major issues that the surgical community must address to facilitate safe introduction of NOTES to the clinic [10]. The whitepaper listed the development of effective training paradigms and development of a multitasking platform to accomplish procedures as two of the goals. These two goals emphasize the need for training and further procedural development, which can be facilitated by simulation interfaces.

NOTES procedures can be characterized as a pure or hybrid NOTES. A pure NOTES procedure consists of using a flexible endoscope to gain access to the target organ via a natural orifice, for example the gall bladder for a pure NOTES cholecystectomy. A hybrid NOTES technique uses a rigid endoscope along with the additional assistance of a laparoscopic tool through a different access port, such as the umbilical [8]. It is the control and maneuvering of a flexible endoscope in pure NOTES procedure that requires substantial training. The complexity in a NOTES procedure arises from the difficulty in safely maneuvering a flexible endoscope to approach the target organ while breaching the respective lumen [11] and then achieving proper luminal closure after the procedure is completed [12].

The inherent difficulty of flexible endoscopy techniques are widely understood and have warranted the development of a variety of simulators, specifically for flexible endoscopy based procedures. Some simulators were built for training in a specific procedure while others allow general endoscopy based training. A 2 DOF haptic device for colonoscopy training was developed by Samur et al. [13]. The device was designed to provide active and passive rotational and translational force-feedback to the user. The portable Virtual Endoscope system (Portable VES) was designed by Ikuta et al. for endoscopic insertion

training [14]. The 2 DOF colonoscopy simulator developed by Yi et al. incorporated a unique sensor to measure the jiggling forces applied during a procedure, a maneuvering technique unique to colonoscopy [15]. The haptic interface was further developed and came to be known as the KAIST-Ewha Colonoscopy Simulator II. It allowed for free motion of the tip of the endoscope while employing deflection sensors to accurately render the dial manipulation in the virtual scene [15][16]. Korner and Manner [17] developed a 1 DOF device to provide translational force feedback driven by an actuated toothed drive belt, however back drivability is a concern with such mechanisms. A variety of commercial devices designed for colonoscopy simulation are available on the market. CAE healthcare has commercialized the EndoVR system designed to provide training for gastrointestinal and bronchial assessment [18]. The system provides active force feedback while inserting and removing the endoscope from the mannequin. The GI mentor developed by Symbionix USA Corp. has also commercialized and validated an endoscopy simulator specifically for gastrointestinal procedures [19], [20]. The EndoSim simulator by Surgical science particularly offers multiple high fidelity models to learn fundamental endoscopy skills as well as therapeutic skills [21]. A unique feature of the EndoSim is that the system has a detachable desktop haptic unit, allowing easy transport for remote training.

All research and commercial devices present different actuation mechanisms but it is their final force feedback output capability which dictates whether the device is well suited for simulating a certain procedure or not. In this work we demonstrate that the actuation requirements of a NOTES procedure are substantially smaller than traditional colonoscopy procedures. Traditional endoscopy procedures such as colonoscopy have shown peak pushing force = 43N, pulling force = -17.6N, anti-clockwise torque = 1.0 Nm, clockwise torque = 0.8Nm [22]. Other non-navigation based endoscopy procedures such as knotting and suturing actions have shown forces of 0.4N to 10.5N at the end-effector, with friction forces of 0.5 to 1.5N [23]. Our work will show that NOTES procedures demonstrate peak translational forces of 25N with nominal forces of 5.5N and a peak rotational torque of 0.2Nm with a nominal torque of 0.026Nm. Some of these devices that can provide sufficient translational force feedback do not possess rotational force feedback capabilities. One device can render translational forces up to 20N [17] with two other devices rendering up to 8N [14] and 15N [21]. Some devices such as the device by Korner and Manner [17] do not possess rotational force feedback, disqualifying the device since rotational force feedback is important in flexible endoscopy procedures. Devices such as the VES-IV by Ikuta et al. [14] possess the rotational force feedback however use stiction prone actuation mechanisms such as a decoupled ball drive mechanism. The existing commercial devices, such as the EndoSim may be able to render greater forces but using them for NOTES procedures would mean a gross underutilization of the force feedback capabilities, potentially restricting the systems dynamics to only a small portion of its dynamic range. In particular, it has been shown that higher levels of force-feedback and system bandwidth may improve the realism of the virtual object, but does not convey more information on the size of the virtual object [24]. Also, devices with large force outputs have been shown to have negative impacts on performance due to high friction and inertia [25]. This highlights the fact that devices with force output ranges greater than the application requirement do not necessarily provide a better haptic device and in some cases may prove detrimental to the performance due to high

inertia and friction. The discussion section provides a technical overview of haptic performance specifications in comparison to a NOTES procedure's requirements illustrating the need to build a NOTES specific haptic device.

At the Center of Modeling, Simulation and Imaging in Medicine (CeMSIM) at RPI, we have designed and developed a dedicated VR-NOTES training platform known as the VTEST© (Virtual Transluminal Endoscopic Skills Trainer). The VTEST© consists of a dedicated haptic device for pure NOTES using a flexible endoscope. In this work we present the design, development and system identification of the 2 DOF haptic device.

2 System Components and Design

2.1 Interface Design

Achieving appropriate face and content validity for a surgical simulator is critically important to aid in learning, one way to achieve that is by hardware realism. The VTEST© hardware interface has been given the form factor of a patient lying on an operating table in the supine position while covered by surgical drapes to mimic an OR like environment. In this design of the VTEST© hardware interface, we have incorporated a transvaginal entry port to simulate a trans-vaginal pure notes procedure. The insufflated abdomen is constructed with tissue mimicking silicone foam while the transvaginal entry port is a silicone model of the vagina from Simulab Corp (Part# PRT-20). Figure 1 illustrates the design of the interface by means of a CAD model along with the internal haptic hardware detailed in the following sections. The simulation interface operates with a simulation PC rendering the virtual simulation environment and is displayed to the user using a standard computer monitor as shown in Fig. 2.

2.2 Instrumented Flexible Endoscope

The hardware interface uses an actual flexible endoscope instrumented with high precision rotary encoders (CE300-40, MicroE Systems, Bedford, MA) to simulate the flexible endoscopy procedure [26]. The rotary encoders are ergonomically installed to prevent any interference to the user while measuring the manipulation to the dials as the user navigates through the procedure. The exact placement of the encoders and optical discs can be seen in Fig. 3. During manipulation of the distal end of the endoscope through the dials, resistance encountered, if any, has to be reflected to the user for realism. However, in this work the dials are disconnected from the end effector and no force feedback is provided, which will be part of our future work.

2.3 Device Actuation Requirements

Prior to the development of the hardware interface for the VTEST© simulator, a study was conducted to identify the desired actuation specifications of the haptic device [27], [28]. The study illustrated the force and torque-rendering requirements in a flexible endoscopy based NOTES procedure for the two primary DOF, translation and rotation. Figure 4 shows the range of pull (+ve) and push (−ve) forces applied by the users during the procedure. Figure 5 shows the range of clockwise (+ve) and counter-clockwise (−ve) torque applied by the users. The data collected was for a NOTES appendectomy procedure consisting of all the NOTES

specific maneuvering and soft tissue interaction steps as per the procedural tasks analysis. The mean pull and push resistance in the translation DOF was found to be 1.95 N ($\sigma=1.97\text{N}$) and 4.06 N ($\sigma=3.25\text{N}$) respectively. The peak observed force was 25.2 N and the nominal force was 5.5N. The nominal force was determined as 75th percentile of the measured forces across all the subjects. The mean CW and CCW torque resistance in rotation was 0.026Nm ($\sigma=0.028\text{Nm}$) and 0.023.2Nm ($\sigma=0.030\text{ Nm}$) respectively. The peak torque was 0.214Nm with a nominal torque of 0.026Nm.

With the actuation specifications as a design requirement, the actuators were specifically chosen to match the peak and nominal torque outputs for this NOTES procedure.

2.4 Haptic Interface

The actuators responsible for rendering force feedback can be found inside the simulated insufflated abdomen, as shown in Fig. 6.

The haptic hardware is a decoupled 2 DOF system designed to render translational and rotational force feedback through the flexible endoscope and eventually to the user. The translational DOF provides workspace as long as the flexible endoscope while the rotational DOF has an unlimited rotational workspace. The flexible endoscope remains in a pre-inserted configuration inside the VTEST© interface where the tip of the endoscope interacts with the two decoupled DOF. Ideally endoscopic insertion should also be a part of the training procedure, however a NOTES procedure is typically performed by a multidisciplinary team where insertions are usually performed by a gynecologist in the case of a transvaginal approach and a gastroenterologist in the case of anal or oral cavity approach. With our specific focus on target-organ manipulation and intra-luminal navigation training, endoscopic insertion is not part of this NOTES trainer. Also some design constraints made insertion and removal mechanically unfeasible but these concerns can be addressed in future device iterations. The primary reason for decoupling the two active DOF is to allow minimal interference of the active/passive characteristics of one particular DOF onto the other's active/passive characteristics. For example, the rotational DOF can be actuated with minimal interference from an inactive translational DOF.

The translational force feedback unit consists of two dual-hub omni-directional frictional rollers to grip the flexible endoscope. The staggered nature of the 18 individual rubber rollers on each friction wheel allows for 6 distinct points of contact with the endoscope at all times. It is these individual rubber rollers that decouple the translational force feedback from the rotational force feedback. Figure 7 illustrates the method by which the friction wheels grasp the endoscope and apply the desired force for user to feel via the endoscope. The two friction wheels are rigidly coupled to two RE40 DC motors from Maxon Motor inc., Switzerland (0.04 m diameter, 48V operating voltage, 0.187Nm nominal torque, 0.796Nm stall torque), which operate synchronously, to apply the desired force without any slippage. To further bolster the grip of the friction wheels onto the endoscope, the motors are placed within spring-loaded housings from McMaster Carr, Elmhurst, Illinois (steel compression springs, 1.75" long, 0.187" OD, .012" wire diameter, $K=50\text{N/m}$). The springs apply a constant force onto the motor and the rigidly coupled friction wheel, to essentially squeeze the endoscope for an improved grip and prevent any slip during actuation.

The device provides force feedback in the rotational DOF by transmitting torque using a 20-inch long 316-stainless steel flexible drive shaft (max torque 2.2Nm). A RE40 DC motor (Maxon Motor inc., Switzerland) is attached to the drive shaft which then interfaces with the tip of the endoscope using a rigid coupling machined from aluminum. The drive shaft is guided and kept restricted within a channel to prevent kinking during high torque transmission thus minimizing undesired loss in transmission. A slit the length of the guidance channel runs on the underside of the guidance channel to allow the flexible drive shaft to exit and remain connected to the DC motor. The mechanism is detailed in Fig. 8 highlighting the torque transmission, eventually felt by the user.

2.5 Device Control Electronics

The RE40 DC motors are attached to high-resolution optical rotary encoders (HEDL 5540, 500 counts per rotation) that are used to measure the endoscopes linear and angular orientation. These measurements then allow the simulation engine to calculate the appropriate location and orientation of the endoscope. In conjunction with the information from the optical encoders on the endoscope handle, the virtual scene renders the correct endoscopic view for the user to view. The DC motors are controlled by three separate servo controllers (ESCON 50/5, Maxon Motor Inc., Switzerland), which output the desired torque by commanding a certain current to the motors. The simulation PC then controls the servo controllers through a high speed DAQ (National Instruments NI-USB-6341).

3 System Characterization & Response

Once the haptic hardware has been designed and implemented, it is important to elucidate the characteristics of the system in order to be able to control it. Each haptic device possesses its own impedance (Z_{device}), which interferes with the transmission of the desired virtual environment impedance (Z_{VE}). Thus the goal of identifying the characteristics of the system is to actively compensate for the devices own impedance (Z_{device}) and achieve a better system transparency such that the user experiences the desired virtual environment impedance (Z_{VE}).

Haptic device performance parameters have been well documented and been used to describe the performance and capabilities of haptic devices. Most notably the works of Ellis et al. [29] from Queen's University at Kingston in Ontario, Canada and Hayward et al. [30] at McGill University, Montreal, Canada have described in great detail haptic device performance parameters. Here in this work we have evaluated the static, impulse and frequency response of the VTEST© hardware and eventually implemented a feedforward controller to compensate for device impedance (Z_{device}).

In order to measure the performance characteristics of the 2 DOF haptic device we imposed three types of boundary conditions, open-ended, close-ended and a hand grip as shown in Fig. 9.

In the open-ended configuration, the end effector (location of user's grip on endoscope) was instrumented with a high precision 6 DoF (1.4 mm RMS, 0.5 degrees RMS) magnetic position-tracking sensor (TrakStar, Ascension Technology Corporation, Shelburne, VT) to

measure the position in response to actuation. The close-ended configuration was a locked condition where the end effector was arrested using a vice grip with a 6 axis load cell (ATI Industrial Automation, Apex, NC) interfaced between the two to measure all force and torque (max force=50N, max torque=0.5Nm, Force resolution=0.0125N, torque resolution=6.25×10e-5Nm). These two conditions were used to extract device characteristics for both the translational DOF and rotational DOF.

To further elucidate the performance of the device with a human in the haptic loop, a hand grip boundary condition was also imposed while the device was actuated. A custom handle instrumented with the 6 axis load cell was used to grip the endoscope allowing the real-time recording of forces felt at the user's hand while the device was in use. While grips can be subjectively differentiated as loose or firm grips, we performed the test with a regular grip, sufficient to maneuver the endoscope, ensuring no slippage or restriction to the endoscopic maneuvering.

3.1 System Static Response

The device's static response highlights the maximum continuous force and torque output capabilities. These characteristics are important since the device will mostly be rendering continuous force feedback, and these parameters can directly be compared to our previous actuation requirements study performed on an ex vivo porcine simulator. The device was placed in the close-ended condition, allowing the measurement of force for the translational DOF and torque for the rotational DOF. Once in the locked condition, both DOFs were independently commanded a steady ramp up and ramp down output. Figure 10 and Figure 11 show the results of the close-ended experiments for both DOFs. The results illustrated a maximum continuous translational force and rotational torque of 5.62N and 0.190Nm respectively. When commanding an output beyond these ranges, we observed excessive saturation of the DC motors and thus limited the actuation range, as shown in Figs. 10 and 11. Here the input force refers to the commanded force and output refers to the measured force.

The results from the static experiments for both DOFs indicate a clear non-linearity in the input-output relationship. As observed from the results, a given input force/torque produces a different output force/torque depending on the direction of actuation. The hysteresis in the input-output relationship can be attributed to the presence of friction in the system. Since the device must overcome friction prior to transmitting the commanded force, we observe a transmission loss. The friction in the system is a part of the devices impedance (Z_D), which in an ideal system is zero thus making the system transparent. Specifically for the rotational DOF, we can observe that the clockwise (+ve) direction has the best reproduction of the desired torque, while the counter clockwise (-ve) direction indicates a significant loss in transmission. This direction-based dominance of the rotational DOF's static response can be attributed to the unidirectional winding on the metal threads that constitute the flexible drive shaft. However this direction-based dominance was constant throughout an insertion length of 1 m and did not vary depending on the linear position of the tip of the endoscope, but differences could be expected on the extreme ends of the length of the scope. A unique feature of the translational DOF highlighted by sudden changes in the slope can be seen in

Fig. 10. This repeatable change in slope of the input-output relationship can be attributed to the inner metal windings of the endoscopic metal sheath found under the outer rubber layer. This sudden change in slope occurs when the windings are compressed to the maximum, in effect changing the stiffness of the scope. The stiffness of the scope in translational motion has been accounted for in the model and is compensated for.

3.2 System Impulse Response

Studying the system's impulse response provides the device's capabilities to render forces in terms of velocity and acceleration of the end effector. This response is important to study since in a NOTES procedure there may be instances of sudden contact with an abdominal wall, requiring the haptic device to render a 'virtual wall'. The velocity and acceleration characteristics dictate the device's ability to provide a realistic experience in a 'virtual wall' environment.

In order to observe the device's impulse response, an approximate impulse input was provided to both DOF while the end effector was in the open-ended configuration. A square wave of amplitude 5.62N and 20ms duration provided a maximum velocity of 0.52m/s and acceleration of 7m/s² in the translational DOF as shown in Fig. 12 and Fig. 13. The rotational DOF provided a peak velocity of 2.3rad/s and a peak acceleration of 29rad/s². The peak velocity and acceleration data points provide an understanding of the device's speed while responding to hard stops such as virtual walls.

3.3 System Frequency Response

In order to determine the haptic device's useful bandwidth, a frequency response analysis was performed. Similar frequency response analysis on a 2 DOF haptic device has been performed before by Samur et al. [13]. The translational and rotational DOF were both independently subjected to a 1–30 Hz chirp input signal excitation sequence with translational and rotational amplitudes of 2N and 0.1Nm respectively, while placed in the open ended and close ended configurations, separately.

From the open ended experiment we measured the position of the end effector and calculated the velocity, which yielded the relationship between input force (F_{des}) and output velocity (V_{ee}) known as $V(s)$. The velocity was calculated by taking the derivative of the position data with respect to the sampling rate. In order to filter out the noise the derivative introduced, we implemented a 5th order 30 Hz lowpass butterworth filter.

$$V(s) = \frac{V_{ee}}{F_{des}} \quad (1)$$

From the close-ended configuration we determined the relationship between the input force (F_{des}) and output force (F_{ee}) known as $F(s)$.

$$F(s) = \frac{F_{ee}}{F_{des}} \quad (2)$$

In Fig. 14 we can see the bode plot of the experimental data collected, illustrating the frequency response of $F(s)$, the input force and output force relationship. It can be observed from the experimental data that the force output begins to sharply drop off after $\sim 15\text{Hz}$. In Fig. 15 we can see the frequency response of $F(s)$ for the rotational DOF, indicating a rather quicker drop off in output with respect to the input. Figure 14 and 15 also shows the fit of the model, which was designed to model the behavior of the haptic device. The details of the modeling are in the following section. From the estimated model we determined that the translational DOF force bandwidth of the device was 18.1Hz and 5.7Hz for the rotational DOF.

While an understanding of the device performance in an ideal condition is useful, it is also important to be conscious of the device performance with a human in the haptic loop. Since the primary mode of endoscopic manipulation is a human hand, it can be expected that the dynamic characteristics of the human will have an impact on the haptic performance. Thus using the hand grip boundary condition, a human was introduced into the haptic loop while separately exciting the 2 DOF using the 1–30 Hz chirp signal as done previously. Figure 16 shows the frequency response of $F(s)$ highlighting the input-output relationship of the translational DOF with the human in the loop. We see that the force bandwidth with the hand grip is $\sim 11\text{Hz}$. In Fig. 17 we can see the frequency response of the force output for the rotational DOF, which illustrates a force bandwidth of $\sim 4.4\text{Hz}$. The frequency response data provides an understanding of how the haptic device performs with a human in the loop and will be utilized in further studies to develop control algorithms incorporating human hand dynamics.

4 System Modeling

Using the experimental data, the relationships in equation (1) and (2) were used to estimate the transfer function of the haptic device by fitting the collected experimental data to a second order mass-spring-damper model, as shown in Fig. 18. The transfer functions were estimated using the greybox modeling method built into the system identification toolbox within Matlab.

The transfer functions $F(s)$ and $V(s)$ that were estimated using the greybox modeling method were fit to the above mentioned dynamic model, with the following expressions describing the transfer functions

$$\begin{aligned} V(s) &= 1/(Ms+B) \\ F(s) &= k/(Ms^2+Bs+k) \end{aligned} \quad (3)$$

4.1 System Output Impedance

A haptic device's impedance is defined as the relationship between a given velocity input and its corresponding force output. As mentioned earlier, friction in the system is a factor contributing to the uncontrolled impedance of the device. Other physical characteristics such as inertia and damping in the system contribute to the impedance of the device. In order to identify the impedance of the VTEST© haptic device, we used the relationships between the input force to output force, $F(s)$ and input force to output velocity, $V(s)$. By implementing the electrical analogy of Thevenin's theorem to a 2-port mechatronic haptic device, which compares open-ended and close-ended responses of the system, the device's uncontrolled impedance, Z_{device} was determined using the following relationship [31]

$$F(s) = Z_{\text{device}} \cdot V(s) \quad (4)$$

Figure 19 demonstrates the impedance response of the rotational DOF. From the bode plot we can see that the experimentally determined impedance output drops off distinctly after ~6Hz. The rotational DOF indicated a drop in impedance output after ~12Hz. This range of impedance output for the translational DOF coincides well with the frequency range of human hand motion. It has been shown that human hand motions do not surpass 10Hz, which indicates that our device is sufficiently equipped to perform within the range of human motion [32], [33].

The estimated transfer function $F(s)$ and $V(s)$ provided us with the transfer function for the device's impedance. From the Z_{device} transfer function we determined the overall system characteristics, shown in Table 1.

The parameters shown in Table 1 are the model-fitted parameters, mass, spring and dampers of the dynamic model as shown in Fig. 18. These dynamic device parameters are critical to the design of the closed loop controller, as they will be used to compensate for the hardware's inherent friction.

4.2 Friction Identification & Compensation

In order to compensate for the device dynamics, particularly the friction in the system, a model consisting of stiction and viscous damping was implemented. Figure 20 below shows the implemented friction model with the stiction force (F_{stiction}) and the velocity dependent viscous friction (F_{dynamic}).

The relationship in equation (5) identifies the friction compensation implemented as a function of velocity. The non-zero velocity condition includes the damping parameter B that was experimentally identified using the 2nd order device model from the device dynamics identification in the previous section. During the zero velocity condition, the compensator overcomes the stiction force to break the system free. F_{stiction} and F_{dynamic} were experimentally obtained by having a user move the end effector repeatedly in a periodic fashion. Figure 21 shows the implementation of friction compensation in an impedance type haptic device.

$$F_{comp}^n = \begin{cases} F_{dynamic} \text{sgn}(v^n) + Bv^n & \text{if } v^n \neq 0 \\ F_{stiction} \text{sgn}(v^{n-1}) & \text{if } v^n = 0 \end{cases} \quad (5)$$

The results of the friction compensation can be seen in Fig. 22 and Fig. 23, which compares the response of the device while in user-induced periodic motion with and without friction compensation. The compensator is able to reduce the stiction in the system at the points of near zero velocity. The viscous damping, however minimal, in the system was also well compensated for as illustrated by Fig. 23.

4.3 Performance with Simulated Virtual Wall

Simulating a virtual wall using the VTEST© haptic device provides a realistic example of how the device may be required to behave during a simulation. For example, while simulating gastrotomy, which requires a hole be made through the gastric wall, large abrupt force feedback would be required. The use of a virtual wall simulation illustrates that this device is capable of rendering stiff virtual tissue interactions. In order to quantify the performance of the device, a virtual wall with stiffness, $K=0.34$ N/mm was implemented for the translational DOF. The translational DOF was driven to the maximum stable output of 5.6 N with $K=0.34$ N/mm which was the maximum stable stiffness as shown in Fig. 24. No slipping or instability occurred and $K=0.34$ N/mm and was determined to be the maximum stable virtual stiffness, which can also be claimed as the highest tissue stiffness which can possibly be rendered by the device.

5 Discussion

A new haptic device was built for interfacing with the VTEST© simulator for the NOTES procedure. A number of experiments have been performed on the device to determine a variety of system describing parameters in order to eventually quantify the performance of the haptic device. The primary actuation specifications of the device came from the porcine ex-vivo simulator based NOTES procedure conducted by a diverse group of expert and novice surgeons [27]. These specifications most directly correlate to the haptic devices range of continuous forces, which was quantified by the static response of the system. The VTEST© haptic device is capable of rendering a maximum continuous force of 5.62N and a maximum continuous torque of 0.190Nm. In comparison with the actuation requirements, the device can sufficiently render the full range of pull forces, but falls short for the push force. However Fig. 4 highlights that forces above 5.62N only account for 23.5% of the total range of forces. The device can render forces up to 0.5 standard deviations above the design requirement of mean push forces. With regards to the rotational DOF, the device can provide the full range of torque about the insertion/retraction axis and meets specification. This does highlight the shortcoming in the devices push force capability and leaves an opportunity to improve upon the device by employing gears to improve the actuators output ratio, however employing gears will impact the devices back drivability which could negatively impact the user's experience during free motion.

A detailed study of the frequency response of the device showed the available range of frequencies at which a force can be transmitted to the user. It was shown by Katsura et al. that a narrow force bandwidth negatively impacts the system performance by slowing down the system convergence in a scenario requiring a step input, such as a sudden stop or collision [34]. The VTEST© haptic device demonstrated 18.1 Hz and 5.7 Hz of force bandwidth in the translational and rotational DOF respectively. The high force bandwidth in the translational DOF is especially useful when rendering a sudden collision encounter (hitting abdominal wall), requiring a greater frequency of actuation. The haptic device with 18.1 Hz of bandwidth in the translational DOF ensures a quick settling time of 0.12 seconds and overshoot-minimization. It was also noted that a settling time of 0.12 seconds was fairly short for a haptic device with significant damping. The high damping of the system is a desirable feature for impedance type haptic devices in order to increase its dynamic range [35]. When the same frequency response study was performed on the system with the human in the loop, force bandwidths of 11 Hz and 4.4 Hz were observed for the translational and rotational DOF, respectively. This is an interesting observation as we noticed a drop in force bandwidth of the system for both DOF. This is attributed to the increased damping introduced into the system to the human grip. While there is a slight drop in force bandwidths, depending on the nature of the human grip we can expect greater stability and stiffness of the system due to increased damping, as it is a desirable characteristic for impedance type haptic devices. It is to be noted that different arm postures and gripping forces will yield different dynamics of the system as has been shown in past work [36]. Gastrointestinal soft tissues exhibit different impedances and with a higher force bandwidth the device is capable of rendering a wider range of soft tissue behavior. As long as the impedance controller compensates for the high damping at lower frequencies of motion, the obtained damping of the haptic device is a desirable characteristic. In comparison to the translational DOF, the rotational DOF possesses a much lower force bandwidth. This is in particular due to the flexible nature of the drive system coupling, unlike the translational DOF, which is a rigid coupling. With the test data collected from the ex-vivo experiments, we did not observe any rotational manipulation changes greater than 2 Hz. If there were simulation scenarios requiring rotational DOF impedance change of greater than 5.7 Hz we might encounter an issue, but have not seen so far from our ex vivo studies.

In order to assess the performance of the VTEST©, a comparison of existing research and commercial simulators was done. Table 2 shows the rotational and translational force feedback performance of multiple devices. Despite the differences in their actuation mechanisms, the final force feedback output capability dictates whether the device is well suited for simulating a certain procedure or not. It is to be noted that many devices have active and passive force feedback mechanisms; a comparison of the two would be unfair thus we will be comparing only active force feedback performance specifications. The NOTES procedure requires, rotational and translational force feedback with a nominal force feedback of 5.5 N (25 N peak) translationally and 0.026 Nm of torque (0.2 Nm peak). Table 2 shows that only one device can render translational forces up to 20N with two devices rendering up to 8N and 15N. However the device by Korner and Manner [17] does not possess rotational force feedback, in effect disqualifying the device to serve the purpose of a NOTES simulation. Of the remaining devices, the only two systems that perform better in

the translational and rotationally DOF are the haptic devices by Ikuta et al. [14] and the Surgical Science EndoSim [21]. The VES-IV by Ikuta et al. utilizes a decoupled ball drive mechanism which provides seamless force feedback although, however the VTEST[©] proposed here in this work employs a uniquely different omni-wheel actuation mechanism. The existing commercial devices, such as the EndoSim may be able to render greater forces since they are designed for colonoscopy and other endoscopy simulations; however their use for NOTES procedures would entail a gross underutilization of the force feedback capabilities, potentially relegating the systems dynamics to only a small portion of its dynamic range. In particular, Kilchenman et al. [24] showed that for a given size identification task, above a certain maximum force output there was no measured improvement in task performance. The fact that there is no improvement in user perception of the virtual object size, in addition to the known drawbacks of high-output force feedback devices of high inertia and friction, using existing higher force-output devices may not serve as ideal candidates for a NOTES simulation. Regardless, there are commercial and research devices that could be modified to serve the purpose of a NOTES simulation, however it is the use of the omni-directional wheel drive-mechanism in conjunction with the flexible drive shaft that has resulted in a novel mechanism haptic device for NOTES procedures developed using NOTES specific benchmarking studies. Unfortunately due to proprietary information concerns, some manufacturers did not make the haptic device specifications available.

The goal of developing the system model and describing the device dynamics was to be able to control it in efforts to improve the transparency of the system. The implementation of the feedforward friction compensator was aimed at improving the transparency of the system by compensating for the device's passive impedance caused by friction and inertia. The results of the friction compensation indicate substantial reduction of stiction and viscous friction. However there are regions of over compensation, specifically when the user 'pulls' the scope. Another aspect potentially contributing to the overcompensation is the missing component of 'human impedance' in the controller. Since the device dynamics were described void of human impedance, the controller is unable to fully compensate. As future work, we intend on improving upon this issue by implementing direction dependent friction compensation with the fully developed simulation software. Additionally we will be focusing on implementing realistic haptic feedback in the control dials of the endoscopic handle interface.

6 Conclusion

In this work we have described the design, development and evaluation of a 2 DOF haptic device for flexible endoscopy based NOTES procedures. Flexible NOTES procedures are different in their actuation requirements when compared to endoscopy or colonoscopy procedures as described in the introduction section. A variety of the previously mentioned commercial and research haptic devices may possess the capability to serve as NOTES based haptic devices if appropriately altered and modified. However, given the difference in actuation requirements between NOTES and other flexible endoscopy procedures such as colonoscopy, and the limitations of existing flexible endoscopy haptic devices, the development of a novel NOTES specific mechanism was undertaken. Since the design and actuation specifications were obtained from NOTES specific studies, this resulted in a

NOTES procedure-specific haptic device. The device was capable of rendering up to 5.62 N and 0.190Nm of continuous force and torque in the translational and rotational DOF, respectively. The shortcomings in the translational force feedback capabilities warrant further development and improvement. In conclusion, the decoupled DOF allow for natural motion during use, provide a wide range of force feedback with favorable force bandwidth and frequency dependent impedance response. The haptic device operates with a simulation PC, rendering in real-time the graphics to be viewed on an OR-like flat panel display along with the force feedback response, eventually felt by the user via the haptic device.

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Fig. 1.
A CAD model of the VTEST[®] simulator with a soft silicone sheet covering the abdomen

Fig. 2.
The simulation interface with the silicone foam abdomen removed along with the surgical simulation environment being rendered on a monitor

Fig. 3.
Schematic of the endoscope handle and the placement of the two optical discs and encoders for sensing dial manipulation

Fig. 4.
The frequency histogram showing the range of forces along the insertion/retraction axis experimentally identified during the ex-vivo simulator studies

Fig. 5.
The frequency histogram showing the range of torques about the insertion/retraction axis experimentally identified during the ex-vivo simulator studies

Fig. 6.

The Haptic Interface with the silicone foam abdomen cover removed to reveal the embedded actuation and control electronics

Fig. 7.
The translational force feedback unit with a schematic diagram of the method of force application

Fig. 8.
Schematic of the rotational force feedback mechanism

Fig. 9.
Schematics of the three boundary conditions used in the experiments

Fig. 10. Input-Output curve of the translational DOF. Motors were commanded a ramp-up and ramp-down input force (+6N to -4 N)

Fig. 11. Input-Output curve of the rotational DOF. Motors were commanded a ramp-up and ramp-down input torque of (+190 N-mm to -190 N-mm)

Fig. 12.
Translational DOF velocity response to an approximate impulse input

Fig. 13.
Translational DOF acceleration response to an approximate impulse input

Fig. 14.

A Bode magnitude plot of $F(s)$ for the translational DOF. A 2nd order model was fit to the data to estimate the transfer function (85% fit to experimental data)

Fig. 15.

A Bode magnitude plot of $F(s)$ for the rotational DOF. A 2nd order model was fit to the data to estimate the transfer function (61% fit to model data)

Figure 16.

A Bode magnitude plot of $F(s)$ for the translational DOF with a human in the loop.

Figure 17.

A Bode magnitude plot of $F(s)$ for the rotational DOF with a human in the loop

Fig. 18.

The 1 DOF 2nd order modified mass-spring-damper dynamic system model

Fig. 19.

A Bode magnitude plot of $Z_{\text{device}}(s)$ for the rotational DOF. The comparison of the experimental data vs. the 2nd order model fit highlights the operational frequency range

Fig. 20.
Graphical representation of the implemented friction model.

Fig. 21.

Block diagram of the device-dynamics dependent friction compensation controller

Fig. 22.
Friction-position map for user induced periodic motion

Fig. 23.
Friction-velocity map for user induced periodic motion

Fig. 24.

Device performance during a simulated virtual wall with a stiffness $K=0.34$ N/mm

Table 1

The approximation of the 2nd order device dynamics

System Parameter	Rotational	Translational
Mass (Inertia/Mass)	0.00121 kg.m ²	0.2 kg
Stiffness (K)	1.006 Nm/rad	778 N/m
Damping (B)	0.0446 Nm.s/rad	9.31 Ns/m

Table 2

Force Feedback capability comparison of existing endoscopic haptic devices

	Force Feedback		Force Feedback Decoupling
	Rotational	Translational	
Samur et al. [13]	0.085 Nm	5.4 N	Yes
VES-IV by Ikuta et al. [14]	0.2 Nm	8N	Yes
KAIST-Ewha [16]/Yi et al. [15]	0.04 Nm	7.15 N	Yes
Korner and Manner [17]	N/A	20 N	Yes
CAE EndoVR [18]	-	-	-
Simbionix GI mentor [19]	-	-	-
Surgical Science EndoSim [21]	0.2 Nm	15 N	Yes
VTEST© (proposed system)	0.19 Nm	5.62 N	Yes