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Steering of Flexible Needles Combining Kinesthetic and Vibratory Force Feedback

Claudio Pacchierotti, Momen Abayazid, Sarthak Misra, and Domenico Prattichizzo

Abstract— **Needle insertion in soft-tissue is a minimally invasive surgical procedure which demands high accuracy. In this respect, robotic systems with autonomous control algorithms have been exploited as the main tool to achieve high accuracy and reliability. However, for reasons of safety and acceptance by the surgical community, autonomous robotic control is not desirable. Thus, it is necessary to focus more on techniques enabling clinicians to directly control the motion of surgical tools. In this work we address that challenge and present a novel teleoperated robotic system able to steer flexible needles. The proposed system tracks the position of the needle using an ultrasound imaging system, and, from that, it computes needle's ideal position and orientation to reach a given target. The master haptic interface then provides mixed kinesthetic-vibratory navigation cues about this ideal position and orientation to the clinician as she steers the needle. Six subjects carried out an experiment of teleoperated needle insertion into a soft-tissue phantom. They showed a mean targeting error of 1.36 mm. An additional experiment of remote teleoperation has been carried out to highlight the passivity-based stability of the proposed system.**

I. Introduction

Teleoperated robotic surgical systems can greatly improve the accuracy and safety of surgical procedures. They can filter out high-frequency signals and surgical tremor [1], or scale down clinician's movements to enhance her accuracy [2]. Moreover, they may also enable expert clinicians to train or assist other colleagues from a distance, or even directly enable operations from a remote location [3]. Teleoperated robotic systems also improve the ergonomics of the operating theatre, since the master interface can be always positioned in a way convenient for the clinician to control [4].

Needle insertion into soft-tissue is a minimally invasive surgical (MIS) procedure used for diagnostic and therapeutic purposes, and it is one of the many surgical procedures which may greatly benefit from the employment of teleoperated robotic systems [5]. Inaccurate placement of the needle tip may, in fact, result in misdiagnosis or unsuccessful treatment

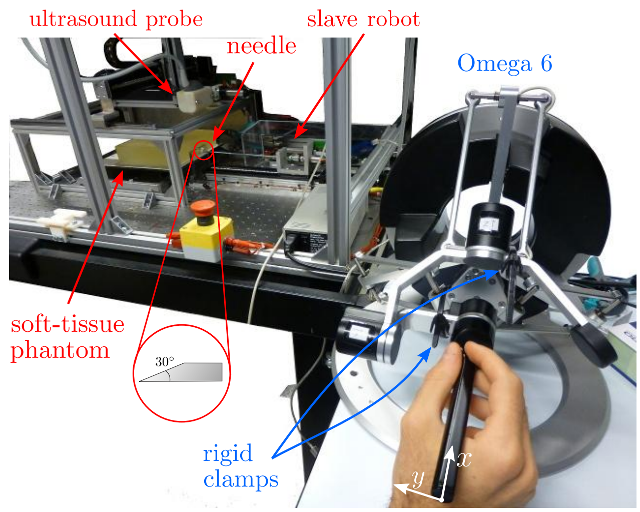

Fig. 1. Teleoperation system. The clinician, through the Omega 6 haptic device, controls the motion of the slave robot and, thus, the needle. The haptic interface also provides the clinician with navigation cues about the ideal position and orientation of the needle tip, evaluated by the steering algorithm.

during, for instance, biopsies or brachytherapies [5], [6]. Hence, researchers have been constantly trying to develop new techniques and systems able to enhance the accuracy of clinicians while performing this type of needle insertions. Flexible needles are one of these technological advancements, introduced to provide enhanced steering capabilities [5]. Several control algorithms have been developed for maneuvering flexible needles in two- and three-dimensional spaces. DiMaio and Salcudean presented a path planning and control algorithm which related needle motion at the base (outside the soft-tissue phantom) to the tip motion inside the tissue [7]. Duindam *et al.* developed a model to describe three-dimensional (3D) deflection of bevel-tipped flexible needles for path planning purposes [8], and Hauser *et al.* developed a 3D feedback controller to steer needles along a helical path [9]. However, results from both Duindam *et al.* and Hauser *et al.* were based solely on simulations, and no experiments in real scenarios were performed. More recently, Abayazid *et al.* presented a two-dimensional (2D) ultrasound image-guided steering algorithm [10] and a 3D needle steering controller for bevel-tipped flexible needles [11], where they used Fiber Bragg Grating sensors to reconstruct the needle shape in real-time.

However, for reasons of safety and acceptance by the surgical community, it is often necessary to disregard autonomous approaches and focus more on techniques enabling clinicians to directly control the motion of the surgical tools [12]. In

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C. Pacchierotti and M. Abayazid have equally contributed towards the results presented in this study.

C. Pacchierotti and D. Prattichizzo are affiliated with the Dept. of Information Engineering and Mathematics, University of Siena, and with the Dept. of Advanced Robotics, Istituto Italiano di Tecnologia, Italy. pacchierotti,prattichizzo@dii.unisi.it

M. Abayazid and S. Misra are affiliated with MIRA - Institute for Biomedical Technology and Technical Medicine, University of Twente, Enschede, The Netherlands.
m.abayazid,s.misra@utwente.nl

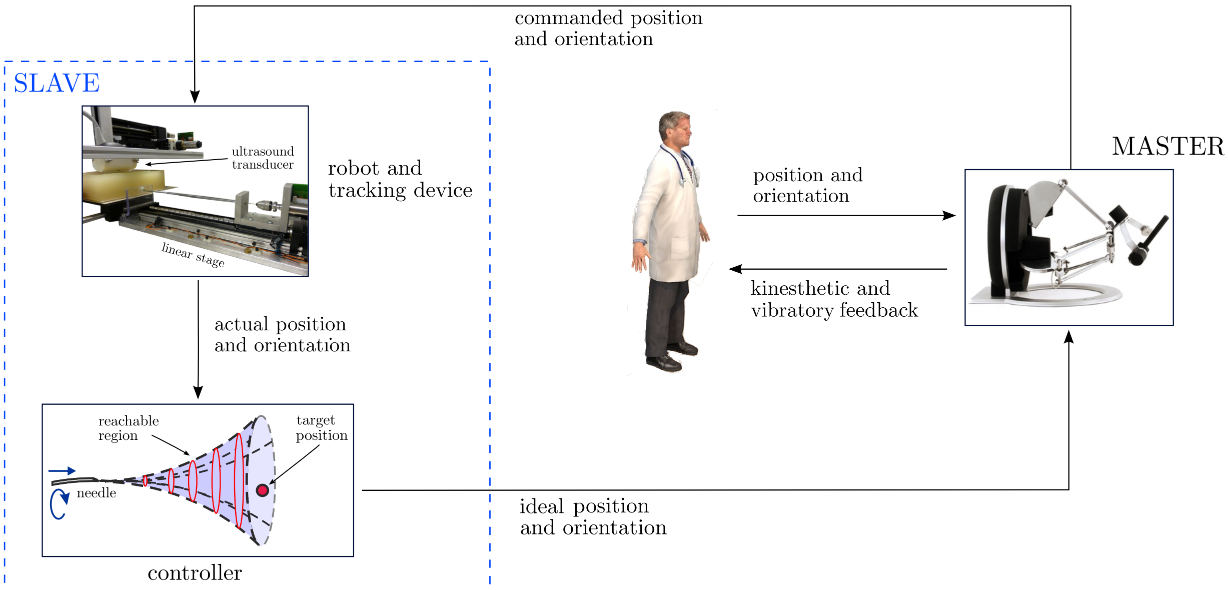

Fig. 2. Teleoperation system overview. The ultrasound-guided steering algorithm, presented in Sec. II-A, computes the *ideal* position and orientation of the needle. The haptic interface provides this information to the clinician through a mix of kinesthetic and vibratory forces, as described in Sec. II-B. The human operator then controls the motion of the slave robot from the master interface.

such a case, the clinician needs to observe, from the master side, the environment the needle is interacting with. This is possible through different types of information, which flow from the remote scenario to the human operator. They are usually a combination of visual, auditory and haptic stimuli. Visual and auditory feedback are already employed in commercial robotic surgery systems (e.g., the da Vinci Si Surgical System, Intuitive Surgical, Sunnyvale, CA, USA) while it is not common to find commercially-available devices implementing haptic force feedback: two of the few examples are the DLR MiroSurge [13] and the Sensei (Hansen Medical, Mountain View, CA, USA) robotic catheter system.

However, force feedback is widely considered to be a valuable navigation tool during teleoperated surgical procedures [14], [15]. It enhances clinicians' performance in terms of completion time of a given task [16], [17], accuracy [14], [18], peak and mean applied force [18], [19]. Force feedback improves performances in fine microneedle positioning [15], telerobotic catheter insertion [20], suturing simulation [21], cardiothoracic procedures [22], and cell injection systems [23]. In addition to these approaches, which mostly involve kinesthetic force feedback, there is a growing interest in vibratory feedback: Schoonmaker and Cao [24] demonstrated that vibratory stimulation is a viable substitute for force feedback in minimally invasive surgery, enhancing clinicians' ability to control the forces applied to the tissue and differentiate its softness in a simulated tissue probing task. More recently, Kuchenbecker *et al.* presented the VerroTouch system [25], which measures the vibrations at the tip of the surgical tool and recreates them on the master handle.

In this study we present a novel approach to robotic teleoperation of flexible needles. It enables clinicians to directly maneuver the surgical tool in the 3-dimensional space while providing them with navigation cues through kinesthetic and vibratory force feedback. These cues are computed by the steering algorithm of Abayazid *et al.* [11], which can track needle's position during the insertion thanks to the ultrasound-guided tracking algorithm of Vrooijink *et al.* [26].

The clinician has thus full control on the motion of the needle, and haptic feedback - kinesthetic and vibratory - provides the necessary guiding information, as evaluated by the steering algorithm. The complexity of the flexible needle kinematics and surgical scenario make haptic feedback a valuable support tool for guidance. A picture of the teleoperation system is reported in Fig. 1. Moreover, Fig. 2 shows how the master and slave systems are inter-connected.

The paper is organized as follows: Sec. II introduces the teleoperation system, describing in details the robotic systems employed at the master and slave sides. Then, in Sec. III we carried out two experiments of teleoperated needle insertion in soft tissue to evaluate the effectiveness of the system. Sec. IV addresses concluding remarks and perspectives of the work.

II. TELEOPERATION SYSTEM

The slave system consists of a bevel-tipped nitinol needle, mounted on a two degrees-of-freedom (DOF) robotic device (see Fig. 3). The robot allows the needle to move along the direction of insertion and rotate about its axis. Moreover, an ultrasound-guided tracking system is used to determine needle tip position during the insertion. The steering and tracking algorithms have been presented by in [11] and [26], respectively.

The master system consists of the single-contact grounded haptic interface Omega 6 (Force Dimension, Nyon, Switzerland), shown in Fig. 4. Two rigid clamps prevent the wrist of the haptic device from moving. The actuators then block two additional DOF, resulting in a haptic interface with 2 DOF,

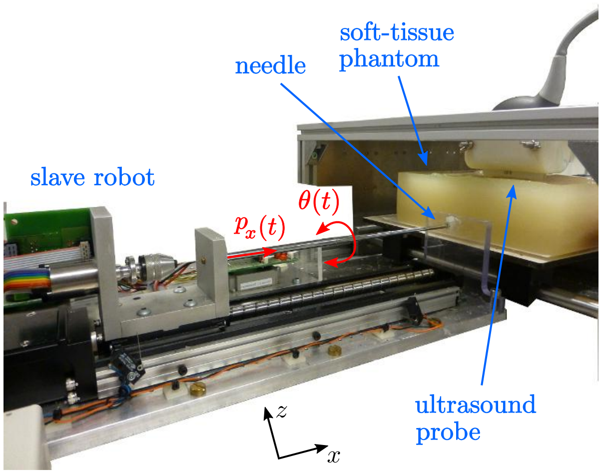

Fig. 3. Slave system. The two degrees-of-freedom robotic device steers the bevel-tipped needle according to the commanded position $p_x(t)$ and commanded orientation $\theta(t)$, which come from the master device. The ultrasound probe allowed needle tracking during the insertion.

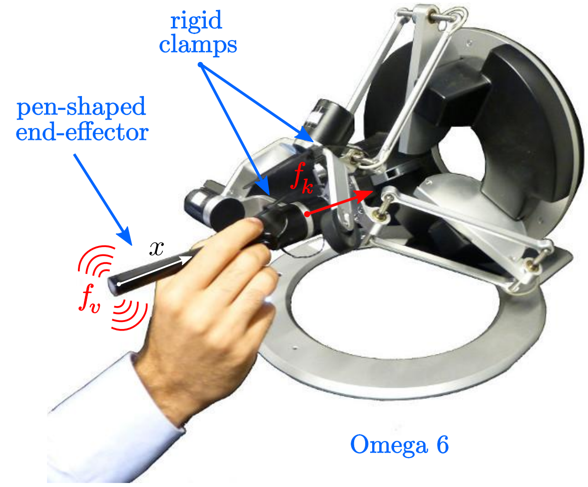

Fig. 4. Master system. The Omega 6 haptic interface enables the clinician to directly steer the needle while being provided with kinesthetic force $\mathbf{f_k}$ and vibratory force $\mathbf{f_v}$ about needle's ideal position and orientation, respectively. The motion of the haptic device is constrained along its x-axis.

one active (translation in the x direction) and one passive (rotation of the pen-shaped end-effector about the x-axis). The master interface allows the clinician to steer the needle and provides her with navigation cues through kinesthetic and vibratory force feedback. The teleoperation system is interconnected as depicted in Fig. 2.

Communication between the slave and the master systems is set up through a User Datagram Protocol over IP (UDP/IP) socket connection on an Ethernet Local Area Network (LAN). The stability of the teleoperation system is guaranteed by the passivity-based approach presented by Franken *et al.* [27]. The control algorithm is able to guarantee stable behaviour of bilateral telemanipulation systems in the presence of time-varying destabilizing factors, such as stiff control settings, relaxed user grasps, and/or communication delays. The control architecture is split into two separate layers. The hierarchical top layer, named *Transparency Layer*, aims at achieving the desired transparency, while the lower layer, named *Passivity Layer*, ensures the passivity of the system.

Further information about this passivity-based control algorithm can be found in [27], while the slave and master systems presented in this work are detailed in Sec. II-A and II-B, respectively.

A. SLAVE SYSTEM

The slave system is composed of a 3D ultrasound tracking device and a two DOF robot, as shown in Fig. 3. The robot allows the needle to be translated and rotated about its axis. This permits the needle to reach any point in the 3-D space. The needle tip is tracked by controlling a 18 MHz ultrasound transducer via a three DOF Cartesian robot. The transducer is connected to a Siemens Acuson S2000 ultrasound machine (Siemens AG, Erlangen, Germany). Further information about the tracking device is presented in [26].

As discussed in Sec. I, the controller needs to evaluate ideal position and orientation of the needle in order to provide this information to the clinician. Towards this objective, we assume the bevel-tipped needle to move along a circular path during insertion [28], [29], and the soft-tissue phantom to be stiff enough to support the needle shaft to follow the path created. The direction of the circular path then depends on the orientation of the bevel tip [11], and this orientation can be controlled by rotating the needle at the base about its insertion axis. The control algorithm, during the insertion, defines the region the needle can reach from its current position. As the distance between the needle and the target decreases, the volume of the reachable region decreases as well. The control algorithm then defines ideal position and orientation of the needle in order to always keep the needle tip in its reachable region, until the needle reaches the center of the target. Further information about this controller can be found in [11].

The main difference between the approach presented by Abayazid *et al.* [11] and the one presented here is the role of the clinician. In the work of Abayazid *et al.* the controller has full control on the motion of the slave robot and it applies the requested translation and rotation directly to the needle. No human is involved in the control loop. However, as discussed earlier, for reasons of safety and acceptance by the surgical community, autonomous robotic control is not desirable. For this reason, in our work, only the clinician can act on the motion of the needle. The controller first evaluates the ideal orientation and position of the needle. Then sends this information to the master interface, which presents it to the clinician, who, in turn, commands the slave robot and steers the needle towards its target point.

B. MASTER SYSTEM

The master system is responsible for both steering the slave robot and displaying navigation cues. It has to face the challenging problem of conveying *two* pieces of information, i.e., ideal position and orientation of the needle, through the same sensory channel. In order to avoid confusion and consequent possible errors in the surgical procedure, the meaning of such cues must be easy to understand. In this paper we propose to provide the human operator with two

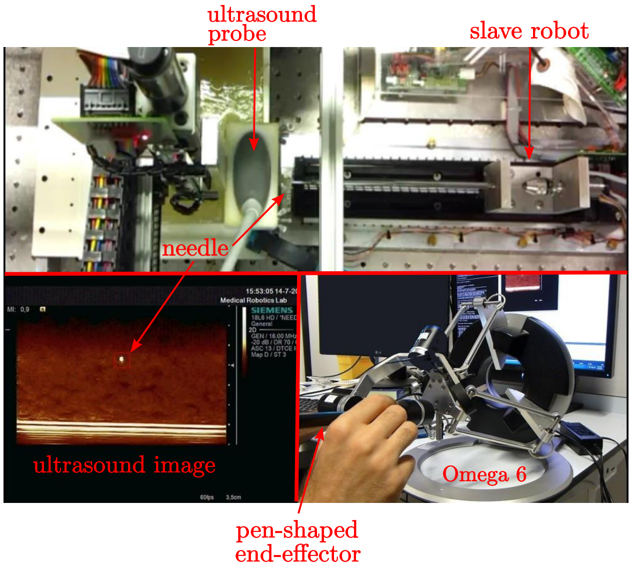

Fig. 5. Experimental test. The human operator, through the haptic device (bottom-right), steers the needle inside the soft-tissue phantom (top). The controller, through the ultrasound imaging system (bottom-left), tracks the needle and evaluates the ideal position and orientation of the needle tip to be fed back to the clinician via kinesthetic and vibratory force feedback. No visual feedback is provided.

different stimuli, in order to better differentiate between translations and rotations:

(i) kinesthetic force $\mathbf{f_k}$ to convey information about the ideal *position* of the needle tip,

(ii) vibratory force $\mathbf{f_v}$ to convey information about the ideal *orientation* of the needle tip,

as depicted in Fig. 4. Ideal position $p_{i,x}(t) \in \Re$ and ideal orientation $\theta_i(t) \in \Re$ at time t are computed by the slave controller as described in Sec. II-A.

Kinesthetic force feedback along the x-axis is controlled by a penalty function based on the distance between the position of the haptic probe $\mathbf{p(t)} = [p_x(t) \quad p_y(t) \quad p_z(t)\ ]^{\mathrm{T}} \in \Re^{3\text{x}1}$ and the current ideal position $p_{i,x}(t)$, while the motion along the y and z axes is blocked:

$$\mathbf{f_k} = \mathbf{K}\ \mathbf{D(t)} - \mathbf{B}\ \dot{\mathbf{D}}\mathbf{(t)}, \tag{1}$$

where $\mathbf{B} = 1.5\ \mathbf{I_3}$ Ns/m, $\mathbf{K} = \mathrm{diag}\,[1 \quad 4 \quad 4]$ N/mm, and $\mathbf{D} = \mathbf{p_i(t)} - \mathbf{p(t)}$ is the distance between the ideal position $\mathbf{p_i(t)} = [p_{i,x}(t) \;\; 0.10 \;\; 0.08]^{\mathrm{T}}$ m and the current position of the haptic probe. The motion is thus limited along the x-axis and a kinesthetic force guides the clinician towards $p_{i,x}(t)$ (see Fig. 4).

On the other hand, information concerning the orientation of the needle tip is provided through vibratory feedback. It is controlled by a penalty function based on the difference between the current orientation of the haptic probe $\theta(t) \in \Re$ and the current ideal orientation $\theta_i(t)$:

$$\mathbf{f_v} = \mathbf{A}\ |\theta(t) - \theta_i(t)|\ \ \mathrm{sgn}(\sin(\omega t)), \tag{2}$$

where $\mathbf{A} = \dfrac{3}{\pi}\ \mathbf{I_{3\text{x}1}}$ N/rad and

$$\omega = \begin{cases} 200\ \text{Hz} & \text{if } \theta(t) - \theta_i(t) \geq 0, \\ 150\ \text{Hz} & \text{if } \theta(t) - \theta_i(t) < 0. \end{cases}$$

Vibrations thus provide information about the ideal orientation $\theta_i(t)$, indicating in which direction and how much the clinician should rotate the pen-shaped haptic probe. Frequency ω indicates in which direction the clinician should rotate the pen-shaped haptic probe: clockwise for $\omega = 200$ Hz and counter-clockwise for $\omega = 150$ Hz. Frequency values are chosen in order to maximally stimulate the Pacinian corpuscle receptors [30], be easy to distinguish [31] and fit the master device specifications. On the other hand, the amplitude of these vibrations indicates how much the clinician should rotate the haptic probe.

The total force provided to the clinician through the Omega 6 haptic interface is then evaluated by combining eq. 1 and 2,

$$\mathbf{f_t} = \mathbf{f_k} + \mathbf{f_v}. \tag{3}$$

The effectiveness of mixing kinesthetic and vibratory feedbacks to convey multiple pieces of information has been also discussed in [32]. No visual feedback is provided. The ultrasound image is only used by the tracking algorithm, and it is not shown to the clinician.

III. EXPERIMENTAL EVALUATION

This section presents the experimental validation of the integrated teleoperation system. The experimental setup is shown in Figs. 1 and 5. It is composed of the slave and master robots described in Sec. II-A and II-B, respectively.

A. *Teleoperation of flexible needles with navigation cues*

The flexible needle, made of nitinol alloy, has a diameter of 0.5 mm, with a bevel angle (at the tip) of 30°. It is inserted into a soft-tissue phantom made of gelatine mixture, to which silica powder is added to mimic the acoustic scattering of human tissue [11]. The motion of the haptic device, as mentioned in Sec. II-B, is constrained to its x-axis only (rotation and translation, see also eq. 1 and Fig. 4).

The task consists of steering the needle toward a given target point, located at $\mathbf{o_t} = [85 \;\; -10 \;\; 5]^{\mathrm{T}}$ mm with respect to the initial position of the needle. The control algorithm calculates the ideal position and orientation of the needle tip, as discussed in Sec. II-A, and the haptic interface presents

TABLE I

TELEOPERATED NEEDLE INSERTION RESULTS: THE MEAN ERROR IN REACHING THE TARGET POINT e_t AND THE MEAN ERRORS IN FOLLOWING THE IDEAL POSITION AND ORIENTATION SIGNALS, e_p AND e_r, RESPECTIVELY.

User	Age [years]	Sex	e_t [mm]	e_p [mm]	e_r [deg]
1	30	F	2.56	3.04	12.38
2	26	M	0.90	2.93	20.03
3	28	M	1.46	2.84	18.92
4	28	M	1.65	2.75	19.55
5	24	F	0.65	2.92	26.71
6	26	M	0.92	2.06	11.95
mean			**1.36**	**2.76**	**18.26**
σ			0.70	0.36	5.50

these two pieces of information via kinesthetic and vibratory feedback, as discussed in Sec. II-B.

Six participants (4 males, 2 females, age range 24 - 30 years) took part in the experiment, all of whom were right-handed. None of them had previous experience with haptic interfaces. Each participant conducted one trial of the aforementioned teleoperation task and was asked to follow both haptic cues, kinesthetic and vibratory, being as precise as possible. Participants were informed about the procedure before the beginning of the experiment and a 10-minute familiarization period was provided to make them acquainted with the experimental setup[1]. The average error in reaching the target point e_t, and the average errors in following the ideal position and orientation signals, e_p and e_r, provided a measure of accuracy. Error e_t is calculated as $e_t = \|\mathbf{n_f} - \mathbf{o_t}\|$, where $\mathbf{n_f} \in \Re^{3x1}$ represents needle tip position at the end of the task. Errors on the ideal signals, e_p and e_r, are computed as the mean over time of $\|p_x(t) - p_{i,x}(t)\|$ and $\|\theta(t) - \theta_i(t)\|$, respectively. A null value of these three metrics denotes the best performance.

Participants showed an average targeting error e_t = 1.36 mm ($\sigma_t = 0.70$ mm), an average errors in following the ideal position $e_p = 2.76$ mm ($\sigma_p = 0.36$ mm), and an average error in following the ideal orientation e_r = 18.26 deg ($\sigma_r = 5.50$ deg). Results are summarized in Table I. Moreover, Fig. 6 shows commanded and ideal orientation for a representative run of the experiment.

Participants performed worse with respect to employing the autonomous approach presented in [11], where the steering algorithm directly controlled the slave robot, achieving a mean targeting error e_t of 0.25 mm. However, the targeting accuracy found in our work is still sufficient to reach the smallest lesions detectable using state-of-art ultrasound imaging systems (ϕ 2 mm). Moreover, our results outperform MRI-guided biopsies carried out directly by humans. El Khouli *et al.* [33], in fact, found a mean 3D biopsy targeting error of 4.4 ± 2.9 mm for biopsies of phantoms, and a mean 3D localization error of 5.7 ± 3.0 mm for breast biopsies performed in patients.

B. Teleoperation in a remote scenario

As mentioned earlier, communication between the master and slave systems is set through a UDP/IP socket connection. For this reason, although the experiment presented in Sec. III-A considers a scenario in which the robotic systems are connected to the same LAN, master and slave could be easily placed in different LANs and then communicate through a common internet connection.

The use of internet as a means of communication in bilateral teleoperation has lately gained increasing attention due to its cost-effective and flexible applications [34]. However, this type of digital transmission exchanges data packets through a network characterized by significant variations in time delays. Such a network may also cause unreliable communication due to the loss of packets associated with the considered channel congestion [34], [35]. As a result, bilateral teleoperation performance may severely degrade, and

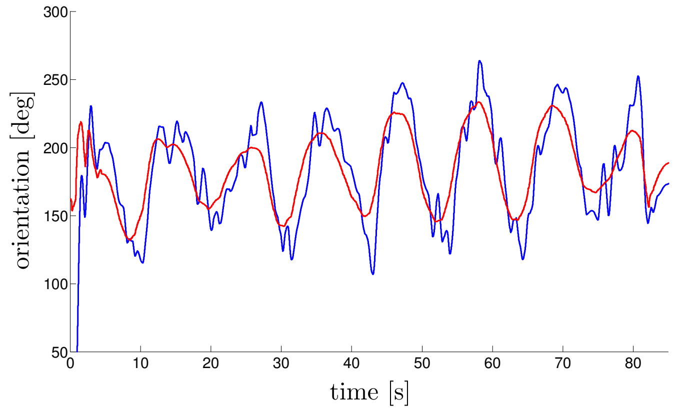

Fig. 6. Experimental evaluation. Commanded and ideal orientation for a representative run are shown in blue and red, respectively.

unstable behaviors may arise. However, the passivity-based controller employed in this work is able to guarantee the stability of our teleoperation system even in such cases (see Sec. II).

In order to evaluate the performance of our system when connected through a common internet connection, we carried out one additional repetition of the needle insertion experiment. The participant was 26 years old, male, right-handed and had previous experience with haptic interfaces. The same setup and protocol presented in Sec. III were employed. The only difference was the master and slave systems being connected through a Digital Subscriber Line (DSL) internet network. The master robot was placed in Genova, Italy, while the slave system was in Enschede, The Netherlands[2]. Results showed a targeting error $e_t = 0.71$ mm, an error in following the ideal position e_p = 2.87 mm, and an average error in following the ideal orientation e_r = 13.26 deg. An average round-trip time of 56.3 ms was registered during the needle insertion experiment. Packet loss was negligible.

IV. CONCLUSIONS AND FUTURE WORKS

In this work we presented a novel robotic teleoperation system to steer flexible needles. It is composed of a slave and a master system. The slave is a two DOF robotic device which enables translation and rotation of the needle. The master is an Omega 6 haptic interface, in charge of tracking the position of the human hand while providing the operator with navigation cues, composed by kinesthetic and vibratory forces. In order to evaluate the performance of the proposed system, six participants carried out an experiment of teleoperated needle insertion in a soft-tissue phantom, relying only on the haptic information provided by the master interface. Results showed worse performance with respect to autonomous insertions, i.e., where the steering algorithm controls directly the slave robot [11]. However, the registered targeting accuracy is still sufficient to reach the smallest lesions detectable using state-of-art ultrasound imaging systems. Moreover, conveying information solely through the haptic channel leaves other sensory channels free. For example, a clinician teleoperating a needle with our system may also be provided with additional

[1]A video of the experiment can be downloaded at http://goo.gl/fmknbI.

[2]A video of the experiment can be downloaded at http://goo.gl/JeeyIj.

visual information, e.g., an ultrasound image of the needle. In order to highlight the stability properties of our system, we also carried out an experiment of remote teleoperation, in which the master and slave systems were connected through a common internet connection. Results were comparable to the one registered in the first experiment and no unstable behaviour arose.

Work is in progress to evaluate the proposed teleoperation system in different clinically-relevant scenarios. We plan to test the system with different target points, introducing obstacles to avoid, and using biological tissue. Moreover, we will validate the system with more subjects and compare the mixed kinesthetic-vibratory approach with other feedback techniques, i.e., sensory substitution through visual, cutaneous or auditory feedback [36].

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