# Enhancing the performance of passive teleoperation systems via cutaneous feedback

Claudio Pacchierotti, Asad Tirmizi,
Gianni Bianchini, *Member, IEEE,* Domenico Prattichizzo, *Senior Member, IEEE*

**Abstract**—We introduce a novel method to improve the performance of passive teleoperation systems with force reflection. It consists of integrating kinesthetic haptic feedback provided by common grounded haptic interfaces with cutaneous haptic feedback. The proposed approach can be used on top of any time-domain control technique that ensures a stable interaction by scaling down kinesthetic feedback when this is required to satisfy stability conditions (e.g., passivity) at the expense of transparency. Performance is recovered by providing a suitable amount of cutaneous force through custom wearable cutaneous devices. The viability of the proposed approach is demonstrated through an experiment of perceived stiffness and an experiment of teleoperated needle insertion in soft tissue.

**Index Terms**—Telerobotics, Haptics and Haptic Interfaces, Stability, Transparency, Force and tactile sensing, Cutaneous tactile force feedback.

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## 1 INTRODUCTION

TELEOPERATION is widely considered a powerful tool to extend human sensing and manipulation abilities to remote or hazardous environments and to scenarios demanding high precision and accuracy. Teleoperated robotic systems consist of a slave robot, which interacts with the given environment, and of a master system, which is commonly operated by a human. The slave robot is in charge of reproducing the movement of the operator who, in turn, needs to monitor the environment with which the robot is interacting. If the operator receives sufficient information about the slave system and the remote environment, he/she will feel present at the remote site. This condition is commonly referred to as *telepresence* [1] and achieving it is mainly a matter of technology: the more complete the information provided to the operator, the more compelling the illusion of telepresence [2].

The primary tool to achieve this objective is providing a *transparent* implementation of the teleoperation system. Transparency can be defined as the correspondence between the master and the slave positions and forces [3], or as the match between the impedance of the environment and the one perceived by the operator [4]. Achieving telepresence hinges upon conveying realistic information from the remote environment to the human operator. Such information usually consists of a combination of visual and haptic stimuli. Visual feedback is already widely employed in commercial robotic teleoperation systems (e.g., the da Vinci Si Surgical System, Intuitive Surgical, USA), while current systems have very limited haptic feedback. This omission is mainly due to the fact that in certain situations *kinesthetic* haptic feedback can lead to an unstable behavior of the system. Indeed, stability of teleoperation systems with force reflection can be significantly affected by communication latency in the loop, hard contacts, relaxed grasps, and many other destabilizing factors which dramatically reduce the effectiveness of haptics in teleoperation [3] (see Fig. 1a).

Despite stability issues, haptic stimuli play a fundamental role in enhancing the performance of teleoperation systems in terms of completion time of a given task [5], [6], [7], [8], accuracy [6], [9], peak [9], [10], [11] and mean force [7], [8], [11]. Therefore, guaranteeing the stability and transparency of teleoperation systems with haptic feedback has always been a great challenge.

To this aim, researchers have proposed a great variety of transparency- and stability-optimized bilateral controllers [12], [13], and it has always been difficult to find a good trade-off between these two objectives. In this respect, passivity [14] has been exploited as the main tool for providing a sufficient condition for stable teleoperation in several controller design approaches such as the Scattering Algorithm [15], Time Domain Passivity Control [16], Energy Bounding Algorithm [17] and Passive Set Position Modulation [18]. In [15] a coding scheme is applied to the power variables (velocities and forces) to turn the time-delayed communication channel into a passive element. When the controllers at both the master and slave sides are, furthermore, passive, the overall system turns out to be stable. In [18] the authors propose an approach built around a spring-damper controller, where the energy dissipated by the *virtual* damper is stored in an energy tank and jumps in spring potential are limited to the available energy in the tank. More recently, a dual-layer controller structure has been presented in [19]. A transparency layer is in charge of computing the ideal forces to be actuated at both the master and the slave, regardless of stability constraints. Cascaded with the transparency layer, a passivity layer modulates such forces when this is necessary to avoid violations of the

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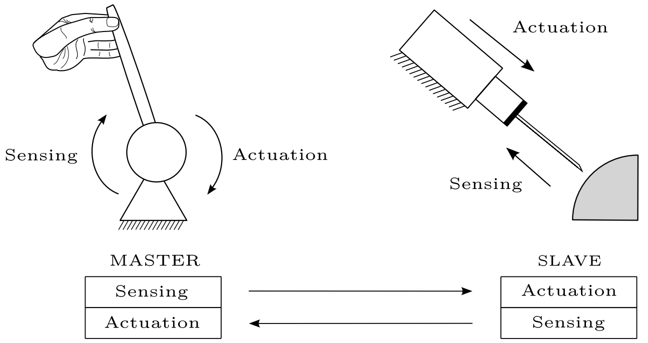
- C. Pacchierotti, A. Tirmizi, G. Bianchini and D. Prattichizzo are with the Dept. of Information Engineering and Mathematics, University of Siena, Via Roma 56, 53100 Siena, Italy.
E-mail: {pacchierotti, tirmizi, bianchini, prattichizzo}@dii.unisi.it
- C. Pacchierotti and D. Prattichizzo are also with the Dept. of Advanced Robotics, Istituto Italiano di Tecnologia, via Morego 30, 16163 Genova, Italy.

passivity condition, thus guaranteeing stability at the price of a temporary loss of performance.

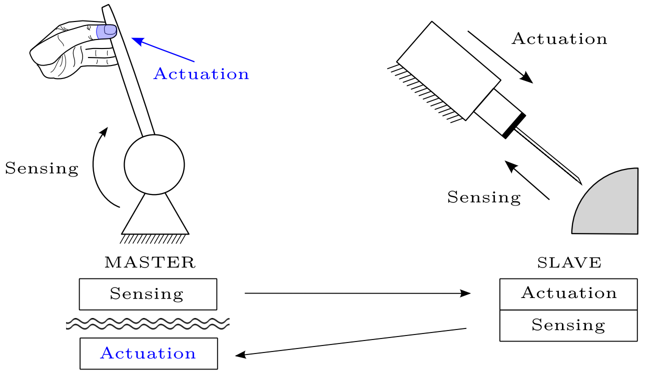
A further approach to stability in teleoperation is *sensory substitution*. It consists of substituting haptic force with alternative forms of feedback, such as vibrotactile [20], auditory, and/or visual feedback [21]. In this case, since no kinesthetic force is fed back to the operator, the haptic loop is intrinsically stable and no bilateral controller is thus needed [9]. The effects of substituting haptic feedback with visual and auditory cues during a teleoperated surgical knot-tying task are evaluated in [21]. Forces applied while using these sensory substitution modalities more closely approximate suture tensions achieved under ideal haptic conditions (i.e., hand ties) than forces applied without such feedback.

Cutaneous feedback has recently received great attention from researchers looking for an alternative to sensory substitution of force feedback; delivering ungrounded haptic cues to the surgeon's skin conveys rich information and does not affect the stability of the teleoperation system [8], [9], [22]. For example, Meli *et al.* [8] found cutaneous feedback provided by a moving platform more effective than sensory substitution via either visual or auditory feedback in a pick-and-place task, and Prattichizzo *et al.* [9] showed that the same type of cutaneous feedback is more effective than sensory substitution via visual feedback in a needle insertion task. Pneumatic balloon-based systems are another popular technique used to provide contact force via cutaneous stimuli. For example, King *et al.* [23] developed a modular pneumatic tactile feedback system to improve the performance of the da Vinci surgical system. The system includes piezoresistive force sensors mounted on the gripping surfaces of a robotic tool and two pneumatic balloon-array tactile displays mounted on the robot's master console. Other lines of research have focused on vibrotactile and skin stretch cutaneous feedback. The system created by McMahan *et al.* [22] for the Intuitive da Vinci robot lets the surgeon feel left and right instrument vibrations in real time without destabilizing the closed-loop controller. 114 surgeons and non-surgeons tested this system and expressed a significant preference for the inclusion of cutaneous feedback of instrument vibrations [24]. Quek *et al.* [25] designed a 3-degrees-of-freedom (3-DoF) skin stretch cutaneous device to substitute full haptic feedback with skin stretch stimuli in teleoperation. Results show that providing cutaneous feedback improved the accuracy of subjects in locating a feature in a 3-DoF virtual environment. Prattichizzo *et al.* [9] call this overall cutaneous-only approach *sensory subtraction*, in contrast to *sensory substitution*, as it subtracts the kinesthetic part of the full haptic interaction - consisting of cutaneous and kinesthetic components - to leave only cutaneous cues (see Fig. 1b). However, although this approach has been effectively employed in complex teleoperation scenarios, it usually provides the user with less transparency than that achieved using full haptic force feedback.

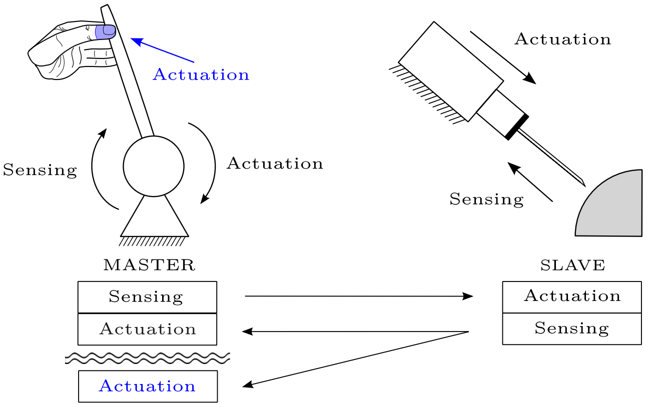
In this paper we present a novel technique based on the combination of kinesthetic and cutaneous force feedback. It mixes the promising cutaneous-only approach of sensory subtraction [9] with the time-domain passivity control algorithm of [19], with the goal of preserving performance when kinesthetic feedback needs to be modulated to guarantee stability. In our technique, the ideal force feedback computed by the transparency layer is actuated via a grounded haptic device as long as the passivity condition is not violated. As the passivity layer detects a violation, kinesthetic feedback is modulated according to the algorithm in [19] while a cutaneous device conveys a suitable amount of cutaneous force in order to recover performance (see Fig. 1c).



(a) The common approach for teleoperation systems. The force fed back to the user is applied directly on the end-effector of the master device, which is also in charge of steering the slave robot. A control action is needed to avoid instability.



(b) Teleoperation system employing cutaneous feedback *only*. Force feedback is applied to the fingertips of the operator and the loop is intrinsically stable.



(c) Enhanced cutaneous-kineshetic approach proposed in this work. Force feedback on the master side is computed according to [19] and actuated via the grounded haptic device, as long as the passivity condition is not violated. As the passivity layer detects a violation, a cutaneous interface conveys a suitable amount of cutaneous force in order to recover transparency.

Fig. 1. Kinesthetic and cutaneous force feedback in teleoperation. Our approach aims at compensating any lack of kinesthetic feedback by providing cutaneous force through a couple of cutaneous interfaces.

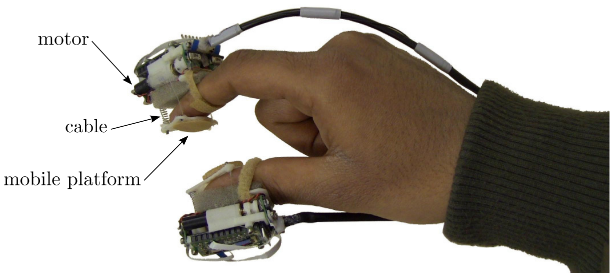


Fig. 2. The fingertip cutaneous devices used in the experimental section of this work.

The proposed strategy yields a teleoperation system which is stable due to passivity control, but with improved realism, since cutaneous feedback conveys force information that cannot be provided through the haptic interface. The control algorithm of [19] is used in this paper only for illustrative purposes, since our technique may in principle be used on top of several other time-domain control methods.

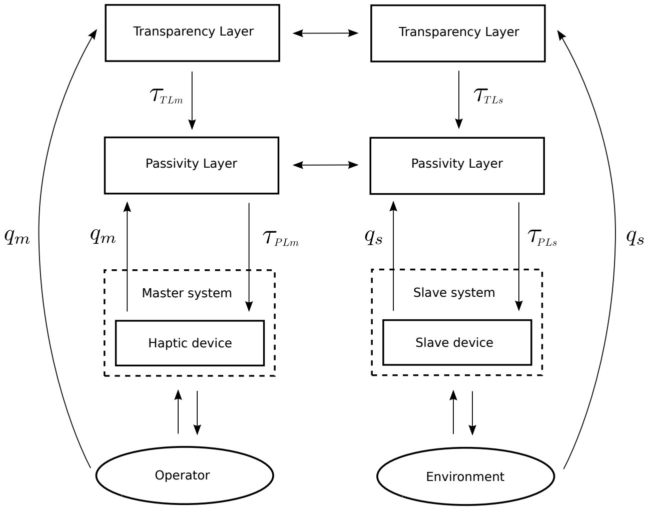
The proposed approach is evaluated in two benchmark scenarios. In the first scenario, we test the performance in terms of perceived stiffness of a virtual hard constraint using full haptic feedback and the proposed cutaneous-kinesthetic approach. The second scenario involves a teleoperated needle insertion in soft tissue. Task performance is compared for the following cases: haptic feedback computed according to [19], cutaneous feedback only (sensory subtraction approach), and the proposed mixed method.

The rest of the paper is organized as follows. Sec. 2 introduces the cutaneous device employed in this work, Sec. 3 describes the proposed approach, Sec. 4 and Sec. 5 illustrate the experimental results, while Sec. 6 discusses them. Lastly, Sec. 7 addresses concluding remarks and perspectives of the work.

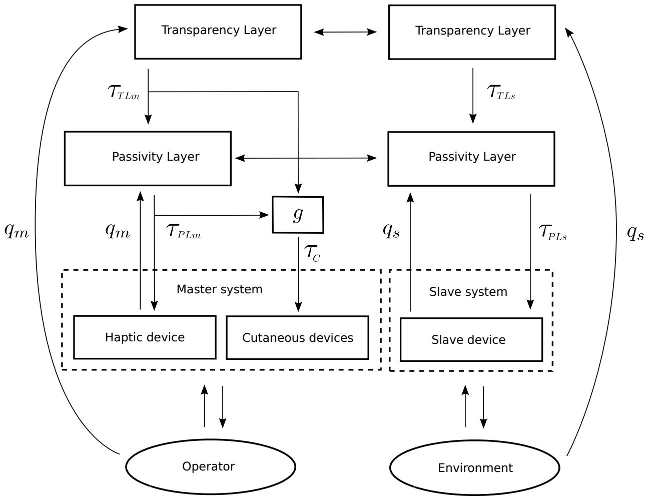
## 2 HAPTIC FORCE FEEDBACK: KINESTHETIC AND CUTANEOUS CUES

Most of the well-known grounded haptic devices, such as the Omega (Force Dimension, CH) or the Phantom (3D Systems, USA) interfaces, provide kinesthetic force feedback to the users [26]. However, these devices also provide *cutaneous* stimuli to the fingertips, if we assume that the interaction with the remote environment is mediated by a stylus, a ball, or by any other tool mounted on the end-effector of the device [26], [27], [28]. As mentioned before, cutaneous feedback does not affect the stability of teleoperation systems as long as the actuators are suitably designed so as to minimize their effect on the position of the master device [9]. Nevertheless, cutaneous feedback often provides less realism than kinesthetic force. Kinesthetic feedback, on the contrary, provides a compelling illusion of telepresence, but it is affected by stability issues.

In order to improve the performance of teleperation systems with force reflection, in this paper we propose to provide cutaneous stimuli combined with full haptic feedback - cutaneous and kinesthetic - provided by grounded haptic interfaces. To this purpose, the operator makes use of the end-effector of the grounded haptic device in combination with a wearable interface that provides additional cutaneous force.



(a) Original approach presented by Franken *et al.* [19].



(b) The mixed kinesthetic-cutaneous feedback approach presented in this work.

Fig. 3. Our approach modifies the control strategy in [19] by adding the opportunity of providing cutaneous feedback when the required force cannot be conveyed using kinesthetic feedback.

The literature on cutaneous technologies is quite rich, but most of the proposed devices are not suitable to be used while operating with a grounded haptic device. A suitable interface has been developed in [29], where the authors presented a wearable and portable ungrounded haptic display that applies cutaneous forces to simulate the weight of virtual objects. It consists of two motors that move a belt in contact with the fingertip. When the motors spin in opposite directions, the belt applies a cutaneous force perpendicular to the user's fingertip, while when the motors spin in the same direction, the belt applies a cutaneous force tangential to the skin. However, this device cannot render forces in all directions, it has only two motors, and the force control is open-loop. Moreover, its control accuracy largely depends on the visco-elastic parameters of the finger pad, which change with different subjects. Performance of this type of devices has been improved with the 3-DoF wearable cutaneous device presented in [30]. It consists of a static platform that houses three DC motors above the user's fingernail and a mobile platform that applies the requested stimuli to the fingertip. Three cables connect the two platforms. By controlling the cable lengths, the motors can orient and translate the mobile platform in three-dimensional space.

The cutaneous device employed for the experiments in this work is a wearable 3-DoF cutaneous device, shown in Fig. 2 and presented in [28]. It is similar to the one in [30] but it has higher accuracy, higher wearability, and both closed-loop force and position control. It is also composed of two platforms: one fixed on the back of the finger and one in contact with the fingertip. These two platforms are connected by three cables made of ultra-high-molecular-weight polyethylene. Three small electrical motors, equipped with position encoders, control the length of the cables, thus being able to move the platform toward the fingertip. The actuators we used are 0615S motors (Dr. Fritz Faulhaber GmbH & Co. KG, Germany), with planetary gearheads having 16:1 reduction ratio. The maximum stall torque of the motors, after the gearbox, is 3.52 mNm. One force sensor (400 FSR, Interlink Electronics, USA) is placed at the center of the platform and in contact with the finger, so that it can measure the component of the cutaneous force perpendicular to the volar skin surface of the fingertip. It has a diameter of 5 mm and a thickness of 0.3 mm, making it very transparent for the user and easy to integrate with the device. The mobile platform and the mechanical support for the actuators are made with a special type of acrylonitrile butadiene styrene, called ABSPlus (Stratasys, USA). The device is overall light weight, around 35 g, and the small dimension of the mobile platform makes this cutaneous device suitable to be used together with common grounded haptic interfaces [9], [28]. Although this device is capable of orienting and translating the mobile platform in three-dimensional space, in this work we used it as a 1-DoF system (all motors pulled the cables together), so that only forces in the sagittal plane of the finger are actuated, roughly normal to the longitudinal axis of the distal phalanx.

# 3 INTEGRATING KINESTHETIC AND CUTANEOUS FORCE FEEDBACK

In this section we discuss how our approach integrates the sensory subtraction method of [9] with the passivity-based controller of [19].

## 3.1 Time-domain passivity control for haptic force feedback

We briefly review the passivity-based time-domain control scheme of [19], which guarantees a stable behavior of bilateral telemanipulation systems in the presence of time-varying destabilizing factors, such as hard contacts, relaxed user grasps, stiff control settings, and/or communication delays. The 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 (see Fig. 3a). The operator and the environment impress a movement $q_m$ and $q_s$ to the master and slave systems, respectively. The Transparency Layer displays the desired behavior to obtain transparency by computing the torques $\tau_{TLm}$ and $\tau_{TLs}$ to be applied to the operator and to the environment, respectively. The Passivity Layer checks how the action planned by the Transparency Layer influences the energy balance of the system. If the passivity condition is not violated, the planned action $\tau_{TL*}$ can be directly applied to both sides of the system. However, if loss of passivity is detected, a scaled control action $\tau_{PL*}$ is applied to preserve stability, resulting in a loss of transparency. Separate communication channels connect the layers at the slave and master levels so that information related to exchanged energy is separated from information about the desired behavior.

## 3.2 Force compensation via cutaneous stimuli

Although we already introduced the general idea of compensating a lack of haptic feedback through cutaneous stimuli, it is necessary to evaluate the amount of cutaneous force that should be provided to compensate for a given lack of haptic feedback, and to what extent cutaneous stimuli can actually compensate for this loss. The experimental work done in [28] provides an insight into these problems from a perceptual point of view. A cutaneous actuator was there used *together* with a grounded haptic device: users wore one cutaneous device on the index finger while grasping the Omega's end-effector. The task consisted in teleoperating a virtual tool along one direction until a stiff constraint was perceived. A spring modeled the contact force between the tool and the stiff constraint. Users were asked to move the remote tool across the virtual environment and stop as soon as the stiff constraint was perceived. The average penetration inside the stiff constraint provided a measure of accuracy [9]. A null value in the metrics denoted the best performance, while a positive value indicated that the participant overran the target.

Task performance (penetration inside the stiff constraint) was evaluated while progressively scaling down the haptic force provided by the grounded haptic interface and the consequent performance degradation was analyzed. Indeed, less force feedback leads to a higher penetration inside the stiff constraint. As the haptic feedback was scaled down, cutaneous force was progressively increased, until the performance obtained with full haptic feedback (i.e., same penetration inside the stiff constraint) was recovered. No stability or passivity issues were there considered. The objective of the experiment was to understand, from a mere perceptual point of view, how much cutaneous force was necessary to compensate, in terms of performance, for a predetermined reduction of the haptic feedback provided by the grounded haptic interface.

Denoting as $\tau_{stc}$ the (full) force to be rendered due to the contact with the stiff constraint, let $\tau_h$ be the scaled haptic force feedback provided by the grounded interface (with $|\tau_h| \leq |\tau_{stc}|$). The additional cutaneous force for which the performance with cutaneous compensation was statistically equivalent to the one registered when using only the grounded device was found to be

$$\tau_c = g\left(\frac{\tau_h}{\tau_{stc}}\right)\tau_{stc}, \tag{1}$$

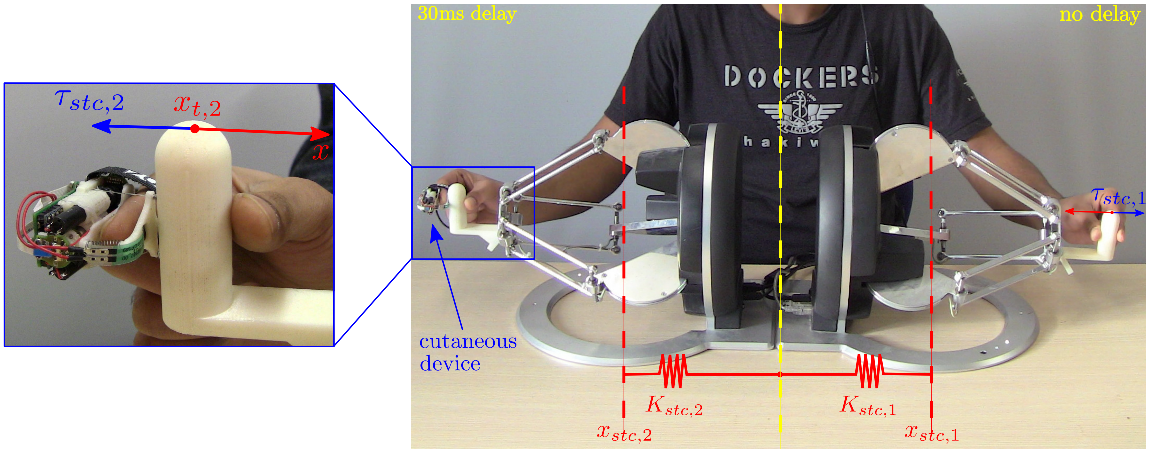


Fig. 4. Experiment #1. The master system is composed of two Omega haptic interfaces $n = 1, 2$ and one cutaneous device. Each interface interacts with a virtual stiff constraint, modeled with a spring of elastic constant $K_{stc,n}$. A simulated master-slave communication delay of 30 ms was simulated between the 2nd Omega and its virtual environment *(left)*. This delay brings the system close to instability as stiffness increases. On the contrary, no delay was introduced between the 1st Omega and its virtual environment *(right)*.

where $g(\cdot) : [0, 1] \to \mathbb{R}$ is a suitable scalar mapping. This means that providing $\tau_{stc}$ through the grounded haptic interface showed statistically equivalent performance as providing $\tau_h$ through the same interface and $\tau_c$ through the cutaneous actuator. The function $g(\cdot)$ was evaluated by means of repeated experiments and polynomial fitting. Details on the method can be found in [28].

Using such experimental protocol, a proper $g(\cdot)$ can be evaluated for any teleoperation scenario. Note that $g(\cdot)$ is task- as well as device-dependent. In all the experiments conducted, however, it turned out that $g(\cdot)$ is strictly monotonic: the more the force provided by the grounded interface is reduced, the more cutaneous force is necessary to achieve comparable performance. Moreover, $g(\alpha)$ was found to be always greater or equal to $1 - \alpha$, regardless of the particular scenario considered. Note that evaluating a proper $g(\cdot)$ for a given scenario may require a long experimental process. In [28], data was gathered from 16 participants, each of whom performed 60 trials. A quick-and-dirty choice for $g(\cdot)$ may be $g(\alpha) = 1 - \alpha$. This approach provides worse performance than properly evaluating $g(\cdot)$, but it still yields better performance than using no cutaneous compensation at all [31].

Finally, it is important to also point out that, in general, it is not possible to compensate for any arbitrary lack of haptic force through this technique. This is mainly due to the limited capability of cutaneous stimulation and to the technological limitations of the cutaneous actuator employed. Under a certain value of $\frac{\tau_h}{\tau_{stc}}$ (when the force to compensate is too high), it is not possible to fully compensate for the loss, but only to mitigate any degradation of performance by conveying as much force as possible through the cutaneous actuator.

In this work, proper mapping functions for the two experimental scenarios in Secs. 4 and 5 were evaluated following the aforementioned protocol.

### 3.3 Combined cutaneous-kinesthetic control algorithm

In the previous subsection we discussed how cutaneous stimuli can effectively compensate for a given lack of haptic force. We now exploit such findings in order to improve the transparency of passive teleoperation systems. As already mentioned, our idea is to combine the time-domain passivity control approach of [19] with cutaneous force feedback.

With reference to Fig. 3b, the Transparency Layer evaluates the desired force feedback $\tau_{TLm}$ to be provided at the master side, while the Passivity Layer checks how the planned action influences the energy balance of the system. If the passivity condition is not violated, then $\tau_{TLm}$ can be fully applied to the operator through the grounded haptic interface. However, if loss of passivity is detected, only a scaled control action $\tau_{PLm}$, such that $|\tau_{PLm}| < |\tau_{TLm}|$, can be applied through the grounded interface, in order to guarantee stability. In this case, we provide an amount of cutaneous force $\tau_c$ according to the method discussed in Sec. 3.2, that is

$$\tau_c = g\left(\frac{\tau_{PLm}}{\tau_{TLm}}\right) \tau_{TLm}.$$

Forces $\tau_{PLm}$ and $\tau_c$ are provided through the grounded haptic device and the cutaneous actuator, respectively. If no violation of the passivity conditions is detected, we have $\tau_c = 0$. In this condition force feedback is provided through the grounded device only, which is the ideal condition. We remark that $g(\cdot)$ is a task-dependent function that can be evaluated experimentally according to the guidelines in [28].

## 4 EXPERIMENT #1: PERCEIVED STIFFNESS

In order to demonstrate the feasibility and effectiveness of our method, two experiments have been carried out. The first experiment evaluates our system from a perceptual point of view. It is inspired by the work of [32], and it involves the evaluation of the perceived stiffness of a virtual environment. We compared the performance of the unaltered algorithm of [19] and of our cutaneous-kinesthetic approach.

### 4.1 Participants

Fifteen participants (13 males, 2 females, age range 20 - 29 years) took part in the experiment, all of whom were

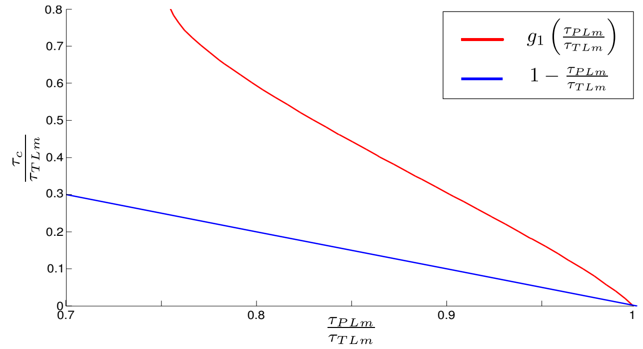


Fig. 5. Experiment #1. Function $g_1(\cdot)$ indicates the level of cutaneous stimuli needed to compensate for a certain reduction of haptic force.

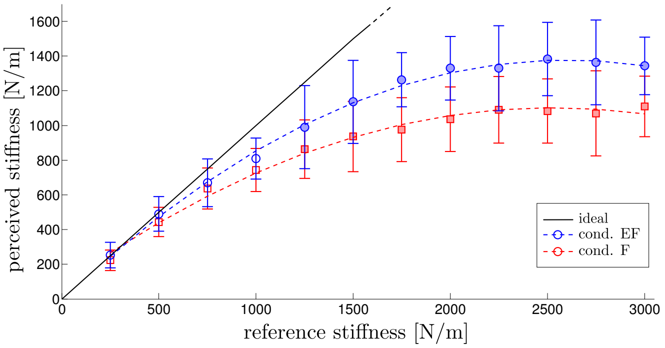


Fig. 6. Experiment #1. Average stiffness ± standard deviation perceived by the participants for the two feedback conditions and the twelve reference stiffness values. In condition F, force feedback is provided by the Omega only, as computed by the unaltered algorithm of [19]. In condition EF, force feedback is provided by both the Omega and the cutaneous device, as computed by the method discussed in Sec. 3.3. Filled markers represent the modalities found statistically different. Dashed lines represent the quadratic approximation to the data sets. The black line represents the ideal perceived stiffness.

right-handed. Eight of them had previous experience with haptic interfaces. None reported any deficiencies in their perception abilities. Before the beginning of the experiment, a 10-minute familiarization period was provided to acquaint them with the experimental setup.

### 4.2 Experimental apparatus and procedure

The experimental setup is shown in Fig. 4. The master system is composed of two Omega 3 haptic interfaces and one prototype of the cutaneous device presented in Sec. 2. Participants wear one cutaneous device on the right index finger, and grasp the Omega's end-effectors as shown in Fig. 4. The motion of the Omega interfaces is constrained along the $x$-axis. Each interface interacts with a virtual stiff constraint, which behaves like a virtual wall. When participants steer one of the haptic interfaces toward the workspace area delimited by its stiff constraint, the system computes the respective ideal force to be fed back

$$\tau_{stc,n} = K_{stc,n}(x_{t,n} - x_{stc,n}), \quad n = 1, 2 \tag{2}$$

where $x_{t,n}$ indicates the position of the $n$-th interface, while $x_{stc,n}$ and $K_{stc,n}$ indicate the position and stiffness of the $n$-th constraint, respectively.

In order to highlight the role of our cutaneous compensation technique, a simulated master-slave communication delay of 30 ms was introduced between the 2nd Omega and its virtual environment. This delay brings the system close to instability as stiffness increases. On the contrary, no delay was introduced between the 1st Omega and its virtual environment. This fact, combined with a high sampling rate ($\sim$ 7 kHz), prevents the 1st Omega from showing any unstable behavior for the employed values of the stiffness.

The 1st Omega (on the right in Fig. 4), when the operator is in contact with the stiff constraint, always feeds back the ideal force $\tau_{stc,1}$. The 2nd Omega (on the left in Fig. 4) is equipped with a cutaneous device and can operate according to one of the two following feedback conditions:

(F) force feedback provided by the Omega only, as computed by the unaltered algorithm of [19],

(EF) force feedback provided by the Omega and the cutaneous device, as computed by the method in Sec. 3.3.

In condition F, if the passivity condition is not violated, then the planned force $\tau_{PLm} = \tau_{TLm} = \tau_{stc,2}$ is directly fed back to the human participant via the Omega. If loss of passivity is detected, a scaled action $\tau_{PLm}$ is applied. Since we designed the virtual environment so that the interaction between the virtual tool and the environment is passive, in this experiment we enforced only the left-hand side of the passivity controller (master side, see Fig. 3). Stability issues can in fact arise only from the master side of the system and from the communication between the master and slave sides.

In condition EF, if the passivity condition is not violated, the planned force $\tau_{PLm} = \tau_{TLm} = \tau_{stc,2}$ is directly fed back to the human participant via the Omega, as in condition F. However, when loss of passivity is detected, the scaled control action $\tau_{PLm}$ is applied via the Omega, and the cutaneous device provides the cutaneous force

$$\tau_c = g_1\left(\frac{\tau_{PLm}}{\tau_{TLm}}\right)\tau_{TLm}, \tag{3}$$

where $g_1(\cdot)$ is the mapping function indicating the level of cutaneous stimuli needed to compensate for a reduction of haptic force during the considered task. Function $g_1(\cdot)$, evaluated for this task according to the guidelines in [28], is reported in Fig. 5.

We tested the perceived stiffness of the virtual environment for reference values of stiffness $K_{stc,ref}$ between 250 N/m and 3000 N/m, with a step size of 250 N/m (12 values in total, see Fig. 6). During the experiment, the motors of the Omega interfaces never reached their saturation limits and never showed an unstable behavior.

Each evaluation started by setting $K_{stc,1} \ll K_{stc,2} = K_{stc,ref}$. Participants were asked to interact simultaneously with the two stiff constraints and tell the experimenter which one *felt* stiffer. In this first interaction all the participants reported $K_{stc,2}$ to feel stiffer than $K_{stc,1}$. We then increased $K_{stc,1}$ by a fixed step size of 50 N/m and asked the participant again. After that, we kept increasing $K_{stc,1}$ by 50 N/m until the participant reported $K_{stc,1}$ to feel stiffer than $K_{stc,2}$. At that point, we took the average between the two last values of $K_{stc,1}$ as the perceived stiffness for the considered participant, reference stiffness, and feedback condition. In an ideal scenario (no stability issues), both Omega interfaces would accurately render the stiffness of the respective constraints and, therefore, the perceived stiffness would always be very close to $K_{stc,ref}$. On the other hand, when the Passivity Layer reduces the force feedback given by the 2nd Omega, the object feels less stiff than it should. In this latter case, the perceived stiffness will turn out to be lower than $K_{stc,ref}$. The cutaneous force conveyed by the cutaneous device in condition EF aims at recovering this lack of haptic force. We expect participants to perceive the constraint *stiffer* when employing the mixed cutaneous-kinesthetic control approach with respect to the unaltered algorithm of [19]. For the sake of clarity, the experimental protocol has been summarized below.

**Algorithm 1:** Perceived stiffness experiment

```
foreach participant do
    foreach feedback condition do
        foreach reference value of stiffness $K_{stc,ref}$ do
            set $K_{stc,1} \ll K_{stc,2} = K_{stc,ref}$;
            repeat
                $K_{stc,1} = K_{stc,1} + 50$ N/m;
                participant interacts w/ stiff constraints;
                participant tells which one feels stiffer;
            until ($K_{stc,1}$ feels stiffer than $K_{stc,2}$);
            $K_{stc,1} - 25$ N/m is the perceived stiffness;
        end
    end
end
```

Participants were not aware of how the stiffness changed over time and between the two Omega interfaces.

### 4.3 Results

In order to compare the performance of the two feedback conditions considered, we evaluated the perceived stiffness for twelve reference values. A perceived stiffness lower than the ideal one indicated a loss of transparency in the system. Data resulting from different repetitions of the same condition, performed by the same participant, were averaged before comparison with other conditions.

Fig. 6 shows the average stiffness perceived by the participants for the two feedback conditions and the twelve reference stiffness values. In order to determine whether the registered data differ between the two feedback conditions, we ran twelve Wilcoxon signed-rank tests [33] (significance level alpha = 0.05), one for each reference stiffness, i.e., F vs EF for $K_{stc,ref} = 250$ N/m, 500 N/m, 750 N/m,. . ., 3000 N/m. The Wilcoxon signed-rank test is the non-parametric equivalent of the more popular paired t-test. The latter is not appropriate here since the dependent variable was measured at the ordinal level. The analysis revealed significant statistical difference between conditions F and EF for $K_{stc,ref} \geq 1250$ N/m (depicted as filled markers in Fig. 6). However, also when results were not found significantly different ($K_{stc,ref} < 1250$ N/m), participants still showed better performance when receiving additional cutaneous force feedback by the cutaneous device. Details on the statistical analysis are reported in Table 1.

| **Wilcoxon signed-rank test (EF - F, alpha = 0.05)** | | |
|---|---|---|
| **$K_{stc,ref}$ (N/m)** | **Z statistic** | **p-values** |
| 250 | -1.179 | .238 |
| 500 | -1.941 | .052 |
| 750 | -1.232 | .218 |
| 1000 | -1.854 | .064 |
| 1250 | -2.150 | .032 |
| 1500 | -2.868 | .004 |
| 1750 | -3.425 | .001 |
| 2000 | -3.346 | .001 |
| 2250 | -3.098 | .002 |
| 2500 | -3.279 | .001 |
| 2750 | -3.279 | .001 |
| 3000 | -3.140 | .002 |

TABLE 1
Statistical analysis results for Experiment #1. Z statistics are based on negative ranks. Red p-values indicate significant difference.

## 5 EXPERIMENT #2: TELEOPERATED NEEDLE INSERTION IN SOFT TISSUE

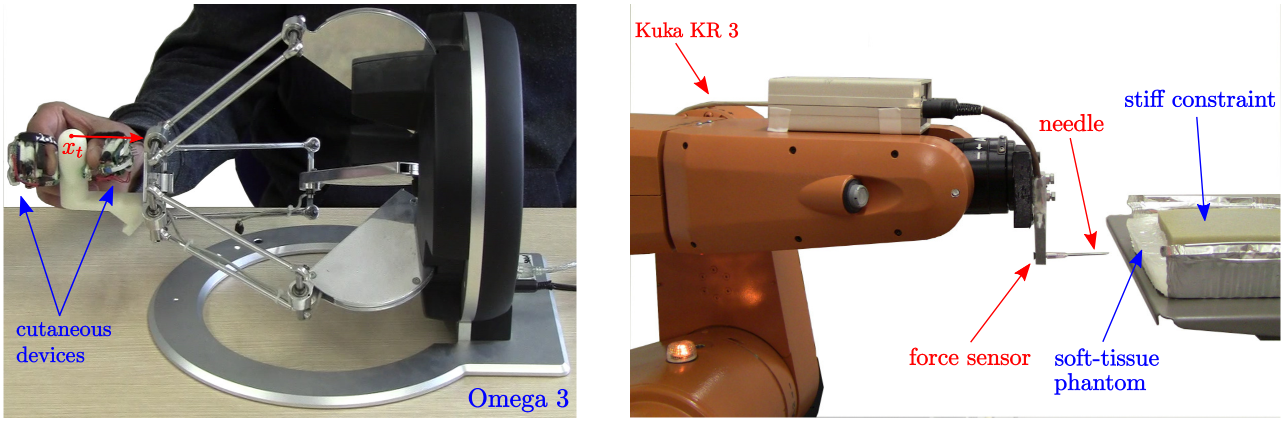
The second experiment aims at evaluating the performance of the mixed cutaneous-kinesthetic approach in a paradigmatic 1-DoF teleoperation experiment of needle insertion in soft tissue. This scenario has been chosen since it is a simple but relevant example of teleoperation task [9], [28]. When performing keyhole neurosurgery, for example, the surgical tool can be steered using a haptic device such as the Omega, and the motion of the tool is along one direction only [34]. In this experiment, we compare the performance while employing the unaltered algorithm of [19], the cutaneous-only sensory subtraction approach of [9], and the proposed cutaneous-kinesthetic method.

### 5.1 Participants

Twenty participants (16 males, 4 females, age range 23 - 32 years) took part in the experiment, all of whom were right-handed. Four of them had previous experience with haptic interfaces. None reported any deficiencies in their perception abilities and they were all naïve as to the purpose of the study. Participants were informed about the procedure before the beginning of the experiment, and a 10-minute familiarization period was provided to acquaint them with the experimental setup.

### 5.2 Experimental apparatus and procedure

The experimental setup is shown in Fig. 7. The master system is composed of one Omega 3 haptic interface and two prototypes of the cutaneous device presented in Sec. 2. Participants wear one cutaneous device on the index finger, one cutaneous device on the thumb, and grasp the Omega's end-effectors as shown in Fig. 7a. The motion of the Omega is constrained along its $x$-axis. The slave system is composed of a 6 DoF manipulator KUKA KR3 (KUKA Roboter GmbH, Germany), a 1-DoF force sensor, and a hypodermic needle, as shown in Fig. 7b. The needle is attached to the force sensor that, in turn, is fixed to the end-effector of the



(a) Master system. (b) Slave system and environment.

Fig. 7. Experiment #2. The master system is composed of one Omega haptic interface and two prototypes of the cutaneous device presented in Sec. 2. The motion of the Omega was constrained along is $x$-axis. The slave system is composed of a 6 DoF manipulator KUKA KR3, a 1-DoF force sensor, and a hypodermic needle. The needle is attached to a force sensor that, in turn, is fixed to the end-effector of the robotic manipulator. The environment is composed of a soft-tissue phantom made of gelatine mixture. A stiff object is placed $2$ cm away from the insertion point.

KUKA manipulator. The needle, made of stainless-steel, has a diameter of 1 mm and a bevel angle (at the tip) of 30°. The teleoperation system is managed by a GNU/Linux machine, equipped with a real-time scheduler, that communicates via Eth.RSIXML (KUKA Roboter GmbH, Germany) with the telemanipulator at 80 Hz and with the Omega interface at 1 kHz. No delay was introduced between the Omega haptic interface and the KUKA manipulator. The environment is composed of a soft-tissue phantom made of gelatine mixture. A stiff object, made of polystyrene foam, is placed 2 cm from the insertion point.

Participants control the motion of the slave robot through the haptic interface. The force sensor registers the force $\tau_s$ exerted by the remote environment on the needle. According to the feedback condition being considered, the Omega 3 and the cutaneous devices feed back a suitable amount of force to the human participant. The task consists of inserting the needle into the soft-tissue phantom and stopping the motion as soon as the stiff object is perceived. After 3 s of continuous contact with the object, the system plays a beep sound. Participants are instructed to pull the needle out of the soft-tissue phantom when the sound is heard. A video of the experiment can be downloaded at http://goo.gl/YY1Uai.

Each participant is supposed to perform twelve randomized trials of the needle insertion task, with four repetitions for each of the following feedback conditions:

(F) force feedback provided by the Omega only, as computed by the unaltered algorithm of [19],
(C) force feedback provided by the cutaneous devices only, as in the sensory subtraction approach of [9],
(EF) force feedback provided by the Omega 3 and the cutaneous devices, as computed by the mixed cutaneous-kinesthetic method detailed in Sec. 3.3.

Condition F is the same as condition F already described in Sec. 4. The Transparency Layer is in charge of evaluating the ideal force to be provided, i.e., the force $\tau_s$ registered by the force sensor at the slave side, hence $\tau_{TLm} = \tau_s$. If the passivity condition is not violated, then the planned force $\tau_{PLm} = \tau_{TLm} = \tau_s$ is applied to the master via the Omega device, otherwise a scaled $\tau_{PLm}$ is applied. The cutaneous actuators are not active.

In condition C, the force $\tau_s$ registered by the force sensor is all fed back through the cutaneous devices. The Omega interface only tracks the position of the fingers and does not provide any force.

Condition EF is similar to condition EF described in Sec. 4. In case of violation of the passivity condition, the scaled force $\tau_{PLm}$ is provided through the Omega, while the cutaneous actuators provide the force feedback

$$\tau_c = g_2\left(\frac{\tau_{PLm}}{\tau_{TLm}}\right)\tau_{TLm}, \qquad (4)$$

where $g_2(\cdot)$, computed again according to [28], is reported in Fig. 8.

In conditions C and EF, a positive cutaneous force directed toward the negative direction of the $x$-axis (see Fig. 7a) is provided by applying a normal stress to the index finger. Conversely, a negative cutaneous force, directed toward the positive direction of the $x$-axis is provided by applying a normal stress to the thumb. In all the considered conditions no visual feedback on the needle is provided. When the motors of the cutaneous device were commanded to provide more force than they could, they were instructed to provide the maximum applicable force (3.5 N). The motors of the cutaneous device never reached their saturation point in condition EF, while they did during trials in condition C.

### 5.3 Results

With the aim of comparing the performance of the three different feedback conditions, we evaluated the average needle penetration inside the stiff constraint, the maximum needle penetration inside the stiff constraint, and the average force reduction due to passivity constraints, computed as the mean over time of $\tau_{TLm} - \tau_{PLm}$. Data resulting from different repetitions of the same condition, performed by the same participant, were averaged before comparison

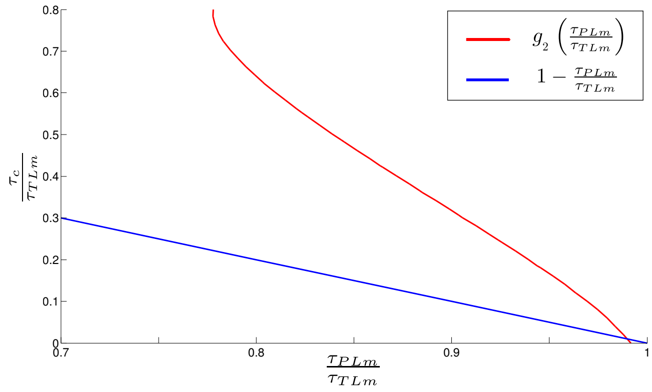


Fig. 8. Experiment #2. Function $g_2(\cdot)$ indicates the level of cutaneous stimuli needed to compensate for a certain reduction of haptic force.

with other conditions. Such metrics provide a measure of accuracy (average penetration) [9], [28], overshoot (maximum penetration) [9], and force reduction [8] for the given task. Penetration measures can be considered particularly relevant to the medical scenario, as an excessive penetration of the needle can result in permanent damage of tissues. Moreover, a high force reduction severely compromises the realism of the haptic interaction.

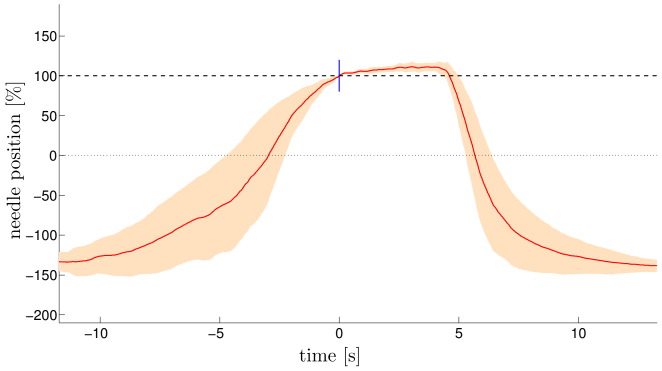
Fig. 9 shows the trajectory of the needle (solid red line) versus time. The time bases of different trials are synchronized at the time the needle enters the stiff constraint (t = 0, solid blue line). Trajectories are averaged among participants for each feedback modality, and average trajectories plus/minus standard deviations are shown. The position of the stiff constraint (dashed black line, 100 percent) and of the soft tissue phantom surface (dotted black line, 0 percent) are shown as well.

Fig. 10 shows the force registered by the force sensor (solid blue line) and the one applied to the participant (solid green line) versus time. In condition C the force sensed and applied is the same, since no passivity constraints are enforced. The difference between the blue and green line is a measure of loss of transparency. The time bases of different trials are again synchronized at the time the needle enters the stiff constraint ($t = 0$, solid red line). Forces are averaged among participants for each feedback modality, and average forces plus/minus standard deviations are shown. Note that a stable rendering of this virtual environment without any stability control would not be possible. Indeed, if the desired force $\tau_{TLm} = \tau_s$ is fully actuated through the Omega interface (i.e., the passivity layer is bypassed), unstable behavior arises, as it is clear from the representative run shown in Fig. 12.

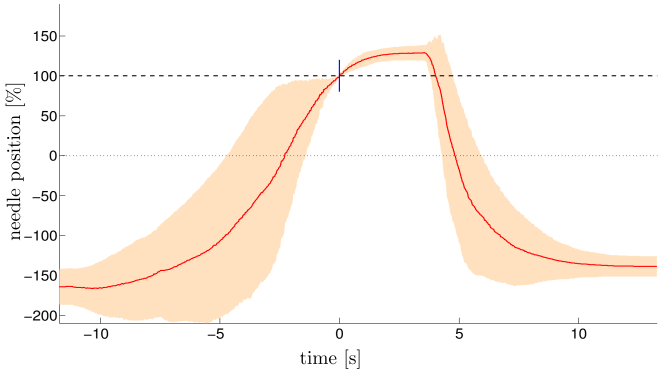
Fig. 11a shows the mean penetration inside the stiff constraint for the three experimental conditions. The collected data passed the Shapiro-Wilk normality test, but Mauchly's test indicated that the assumption of sphericity had been violated. A repeated measures ANOVA with a Greenhouse-Geisser correction [35] determined that mean penetration inside the stiff constraint differed statistically significantly between feedback conditions (F(1.384, 26.289) = 72.874, $p < 0.001$). Post hoc tests using Bonferroni correction revealed statistically significant difference between all the groups.

Fig. 11b shows the maximum penetration inside the stiff constraint for the three experimental conditions. The collected data passed Shapiro-Wilk normality test, and Mauchly's test indicated that the assumption of sphericity had not been violated. A repeated measures ANOVA determined that maximum penetration inside the stiff constraint differed statistically significantly between feedback conditions (F(2, 38) = 26.128, $p < 0.001$). Post hoc tests using Bonferroni correction revealed statistically significant difference between all the groups.

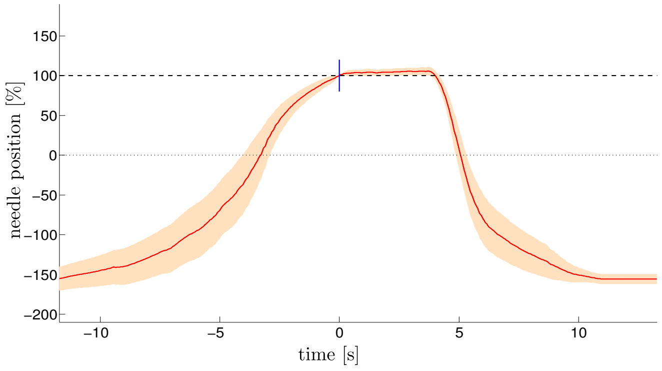
Fig. 11c shows the average force reduction at the master side due to passivity constraints, for experimental conditions F and EF. We did not consider data from feedback conditions C, since it was not subject to any force reduction. The collected data passed Shapiro-Wilk normality test. A paired-



(a) Condition F.

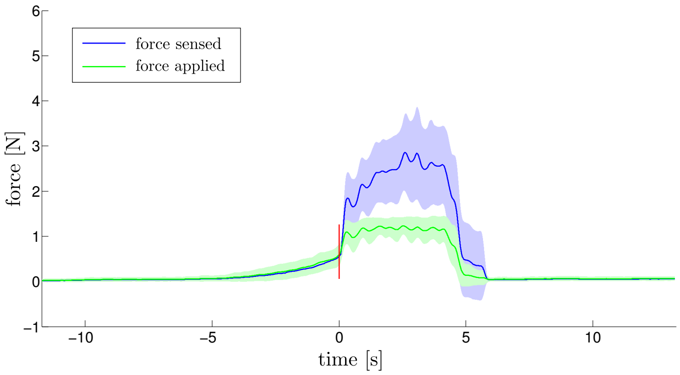


(b) Condition C.

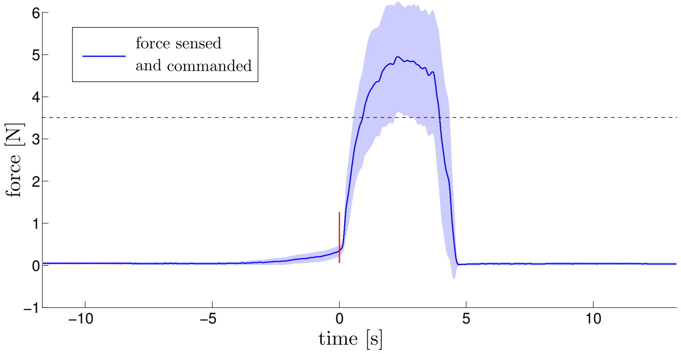


(c) Condition EF.

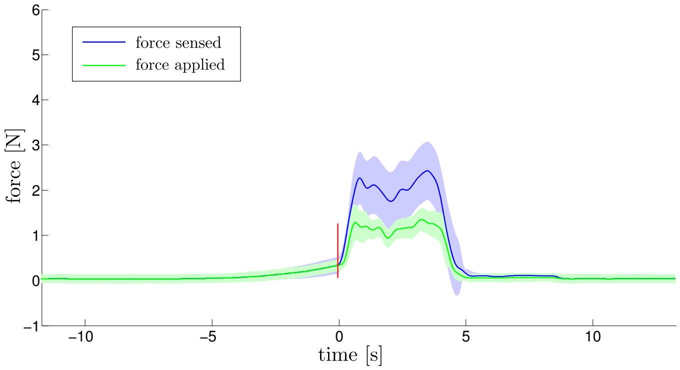
Fig. 9. Experiment #2. Average needle trajectory (solid red line) and its standard deviation (orange patch) are plotted. The position of the stiff constraint (dashed black line) and the position of soft tissue phantom surface (dotted black line) are shown as well. The blue line represents the instant when the needle enters the stiff constraint.



(a) Condition F.



(b) Condition C.



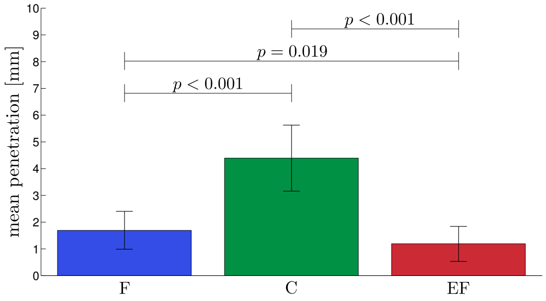
(c) Condition EF.

Fig. 10. Experiment #2. Teleoperated needle insertion in soft tissue. Average force sensed at the needle tip (solid blue line) and force provided to the participant (solid green line), together with their standard deviations (light patches). The red line represents the instant when the needle enters the stiff constraint. The horizontal dashed line in (b) indicates the saturation point of the cutaneous device. The maximum force the cutaneous device was able to provide is in fact 3.5 N.

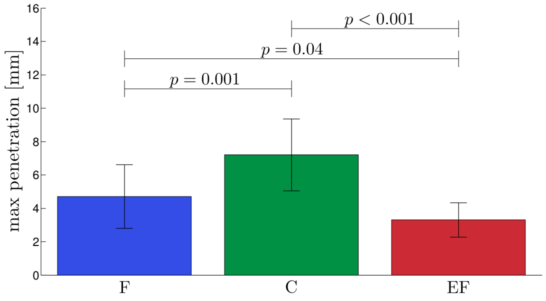
samples t-test determined that the average force reduction at the master side differed statistically significantly between feedback conditions (t(19) = 2.414, $p = 0.026$).

No significant difference between the conditions was observed in terms of task completion time.

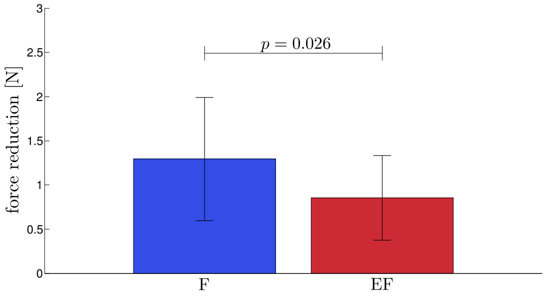
In addition to the quantitative evaluation discussed above, we also measured the users' experience. Immediately after the experiment, participants were asked to fill in a 11-item questionnaire using bipolar Likert-type seven-point scales. It contained a set of assertions, where a score of 7 was described as "completely agree" and a score of 1 as "completely disagree" with the assertion. The evaluation of each question is reported in Table 2.



(a) Mean penetration inside the stiff constraint



(b) Maximum penetration inside the stiff constraint



(c) Force reduction

Fig. 11. Experiment #2. Teleoperated needle insertion in soft tissue. Mean penetration, maximum penetration and force reduction (mean and standard deviation) for the unaltered method of [19] (F), the cutaneous-only sensory subtraction approach of [9] (C), and the mixed cutaneous-kinesthetic method (EF) are shown. A null value of these metrics indicates the best performance.

## 6 DISCUSSION

Two experiments have been carried out. In the first one, the perceived stiffness of a virtual environment was evaluated, employing the unaltered algorithm in [19] (condition F) and the proposed cutaneous-kinesthetic approach (condition EF). Results are reported in Sec. 4.3 and Fig. 6. The stiffness perceived during repetitions with condition EF was closer to the ideal stiffness than that registered under condition F. The proposed cutaneous-kinesthetic approach was thus more effective in rendering the properties of the virtual environment than the unaltered algorithm of [19]. Moreover, since the two feedback conditions share the same underlying passivity controller, they guarantee the same stability properties.

It is worth noticing that in this first experiment we did not take into account the effects of handedness and delay in the perception of the stiffness of the virtual constraints. Cutaneous stimuli were in fact always provided on the right

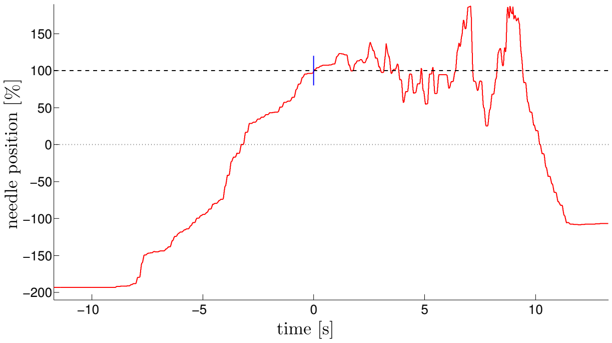


Fig. 12. Experiment #2. Teleoperated needle insertion in soft tissue with no passivity control. Position of the needle versus time, for a representative run. Desired force $\tau_{TLm}$ is fully rendered through the Omega device (Passivity Layer bypassed). Unstable behavior arises. The position of the stiff constraint (dashed black line) and the position of soft tissue phantom surface (dotted black line) are shown as well. The blue line represents the instant when the needle enters the stiff constraint.

hand, which was also the dominant hand of all participants. However, in the second experiment, participants used their right hand to test all the three feedback conditions. Regarding the effect of force delay in the perception of stiffness, Pressman *et al.* [36] presented the results of a forced choice paradigm in which participants were asked to identify the stiffer of two virtual spring-like surfaces based on manipulation without visual feedback. Virtual surfaces were obtained by generating an elastic force proportional to the penetration of the master handle inside a virtual boundary, similarly to what we did in Sec. 4.3. Results show that when force lagged the penetration, surfaces were perceived as stiffer. Conversely, when the force led the penetration, surfaces were perceived as softer. On the other hand, Knörlein *et al.* [37] studied the influence of visual and haptic delays on stiffness perception in augmented reality scenarios. They found delays in force feedback to result in a decrease of perceived stiffness. However, haptic delays smaller than 30 ms were not perceived by the users. For all these reasons, we claimed the difference between conditions EF and F in the first experiment to be due to the effect of our cutaneous compensation technique.

In the second experiment, we compared the performance of a 1-DoF teleoperation experiment of needle insertion in soft tissue employing the unaltered algorithm of [19] (condition F), the cutaneous-only sensory subtraction approach of [9] (condition C), and the mixed approach (condition EF). Results are reported in Sec. 5.3 and Fig. 11. The cutaneous-kinesthetic algorithm outperformed the other two feedback conditions for all the metrics considered. As expected, the cutaneous-only sensory subtraction approach performed the worst. However, even under condition C, all the participants were able to perceive the presence of the stiff constraint and stop the motion of the hand right after the penetration. No difference between the conditions was observed in terms of task completion time. We may read this result by saying that the participants became equally confident with all the feedback modalities proposed. Regarding users' experience, participants felt confident with the system and not hampered by the cutaneous devices. Even if results prove differently, participants did not have the feeling of performing better while receiving additional force feedback from the cutaneous devices.

Although this second experiment serves a different purpose than the first one, i.e., showing a change in performance rather than a change in perception, it is still interesting to notice that in condition F (Omega only and no cutaneous devices), subjects tended to stop the motion of their hand when the force exerted by the Omega interface reached a certain reference value ($\sim$2.5 N), regardless of the penetration inside the stiff constraint. During condition EF (both Omega and cutaneous devices), as expected, this reference force provided by the grounded haptic interface decreases, thanks to the supplementary cutaneous stimuli being provided. This means that the change in stiffness between the soft tissue phantom and the stiff constraint was better perceived in condition EF with respect to condition F. For this reason, the results of experiment #2 can be also evaluated from a perceptual point of view. Similarly to experiment #1, in fact, providing cutaneous feedback through our cutaneous devices results in a better perception of the mechanical properties of the environment.

From the above results, it can be concluded that the proposed method introduces an improvement in the performance of the considered teleoperation system and in the perception of the remote environment with respect to the unaltered algorithm of [19]. The cutaneous-only sensory subtraction approach performs worse than the other two feedback conditions, but still provides a reasonable awareness about the presence of the stiff constraint. These results are also in agreement with previous findings in the literature, e.g., [9] and [38].

## 7 CONCLUSIONS AND FUTURE WORK

In this work we presented a novel control method to improve transparency of passive teleoperation systems with force reflection, which is based on complementing haptic feedback with a suitable amount of additional force through cutaneous interfaces when a reduction of kinesthetic feedback is required to satisfy stability constraints. The viability of this approach was demonstrated via one experiment of perceived stiffness and one experiment of teleoperated needle insertion in soft tissue. Results showed improved performance with respect to common control techniques not using cutaneous compensation.

The method is rather general and applicable to a wide range of teleoperation systems provided that each scenario is characterized using perceptual considerations by a suitable mapping function.

Work is in progress to evaluate the proposed control algorithm in more challenging teleoperation scenarios (e.g., 3-D needle insertion, peg-in-hole tasks). Moreover, we plan to evaluate the difference in the perception of surface stiffness between our mixed cutaneous-kinesthetic method vs. kinesthetic-only and cutaneous-only approaches. We will there also consider possible effects of handedness, learning, delay, experience, and presence of additional sensory stimuli, using appropriate statistical methods and tools. Work is also in progress to design new cutaneous displays with better dynamic performance and wearability, in order to improve the results hereby registered. The validation

| Questions | | Mean | $\sigma$ |
|---|---|---|---|
| Q1 | I was well-isolated from external noises. | 6.40 | 0.60 |
| Q2 | I needed to learn a lot of things before I could get going with this system. | 1.95 | 1.00 |
| Q3 | At the end of the experiment I felt tired. | 1.45 | 0.51 |
| Q4 | I felt confident using the system. | 5.85 | 0.99 |
| Q5 | I think that I would need the support of a technical person to be able to use this system. | 2.35 | 1.04 |
| Q6 | I thought the system was easy to use. | 5.65 | 0.75 |
| Q7 | I would imagine that most people would quickly learn how to use this system. | 6.15 | 0.67 |
| Q8 | It has been easy to wear and use the cutaneous devices. | 6.40 | 0.68 |
| Q9 | It has been easy to use the Omega 3 together with the cutaneous devices. | 6.50 | 0.51 |
| Q10 | I had the feeling of performing better while receiving force feedback by the cutaneous devices. | 4.05 | 1.27 |
| Q11 | I felt hampered by the cutaneous device. | 1.50 | 0.76 |

TABLE 2
Experiment #2. Users' experience evaluation. Participants rated these statements, presented in random order, using a 7-point Likert scale (1 = completely disagree, 7 = completely agree). Means and standard deviations are reported.

of the proposed approach on top of other energy-based control strategies, as well as the design of ad-hoc controllers for optimal exploitation of joint kinesthetic and cutaneous feedback, are the subject of current research. Moreover, we plan to compare the proposed method with different feedback techniques, e.g., sensory substitution through visual, vibrotactile, or auditory feedback.

## ACKNOWLEDGMENTS

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**Claudio Pacchierotti** (S'12) received the B.S., M.S., and Ph.D. degrees from the University of Siena, Italy in 2009, 2011, and 2014, respectively. He was an exchange student at the Karlstad University, Sweden in 2010. He spent the first seven months of 2014 visiting the Penn Haptics Group at the University of Pennsylvania, Philadelphia, USA, which is part of the General Robotics, Automation, Sensing, and Perception (GRASP) Laboratory. He also visited the Dept. of Innovation in Mechanics and Management of the University of Padua and the Institute for Biomedical Technology and Technical Medicine (MIRA) of the University of Twente in 2013 and 2014, respectively. He received the 2014 EuroHaptics Best PhD Thesis Award for the best doctoral thesis in the field of haptics. He is currently a postdoctoral researcher at the Dept. of Advanced Robotics of the Italian Institute of Technology, Genova, Italy. His research deals with robotics and haptics, focusing on cutaneous force feedback techniques, wearable devices, and haptics for robotic surgery.

**Asad Tirmizi** received his B.S. degree in Aeronautical Engineering in 2008 from National University of Sciences and Technology, Pakistan. From 2008 to 2010 he worked as Research Assistant in Air University, Pakistan. He got his M.S. degree in 2012 from Karlstad University, Sweden. He was an exchange student at University of Siena, Italy in 2012 for his M.S. Thesis. He is currently a PhD student at the Department of Information Engineering of the University of Siena. His research interests include robotics and haptics, focusing on cutaneous force feedback techniques, stability of controllers and human-robot interfaces for medical applications.

**Gianni Bianchini** received the M.S. degree in Electronic Engineering from the University of Firenze, Italy in 1997, and the Ph.D. degree in Control Systems Engineering from the University of Bologna, Italy, in 2001. In 2000 he was a visiting student at the Center for Control Engineering and Computation, University of California at Santa Barbara, USA. In 2001-2002 he has been a Research Associate at the University of Siena and since 2003 he is Assistant Professor of Control Systems at the same University. He is interested in robust and nonlinear control, analysis and control of hybrid systems, passivity methods in haptics, and energy systems. He is currently Associate Editor of the IEEE Trans. on Circuits and Systems-II and a member of the Conference Editorial Board of the IEEE Control System Society.

**Domenico Prattichizzo** (S'93 – M'95) eceived the Ph.D. degree in Robotics and Automation from the University of Pisa in 1995. Since 2002 he is an Associate Professor of Robotics at the University of Siena and since 2009 he is a Scientific Consultant at Istituto Italiano di Tecnoloogia. In 1994, he was a Visiting Scientist at the MIT AI Lab. Since 2014, he is Associate Editor of Frontiers of Biomedical Robotics. From 2007 to 2013 he has been Associate Editor in Chief of the IEEE Transactions on Haptics. From 2003 to 2007, he has been Associate Editor of the IEEE Transactions on Robotics and IEEE Transactions on Control Systems Technologies. He has been Chair of the Italian Chapter of the IEEE RAS (2006-2010), awarded with the IEEE 2009 Chapter of the Year Award. Research interests are in haptics, grasping, visual servoing, mobile robotics and geometric control. He is currently the Coordinator of the IP collaborative project "WEARable HAPtics for Humans and Robots" (WEARHAP).