On the role of cutaneous force in teleoperation: subtracting kinesthesia from complete haptic feedback

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Abstract

A study on the role of cutaneous and kinesthetic force feedback in teleoperation is presented. Cutaneous cues provide less transparency than kinesthetic force feedback but they do not affect the stability of the teleoperation system. On the other hand, kinesthesia provides a realistic illusion of telepresence but it affects the stability of the haptic loop. Several well-established control techniques ensure a stable interaction by scaling down force feedback as and when required, in order to satisfy the controller stability conditions (e.g., passivity).

We here discuss the feasibility of a novel approach to improve the realism of the haptic rendering while preserving its stability: can cutaneous stimuli be employed to compensate for the lack of kinesthetic feedback required to guarantee the stability of the teleoperation loop? We carried out two experiments to evaluate the role of cutaneous cues in teleoperation and the performance improvement rate when compensating a lack of kinesthesia with cutaneous force. Results showed improved performance while employing the aforementioned compensation technique and a high comfort in using the proposed system.

1 Introduction

Achieving a good illusion of telepresence in teleoperation depends mainly on situational awareness. A teleoperation system needs to make the human operator aware of what the slave system is up to and this can be achieved through different types of information which flow from the remote scenario to the human operator. Haptic feedback is one of these information flows and it has a well-established role in enhancing the performance of teleoperation systems in terms of task completion time [15, 18], accuracy [18, 22], peak [9, 26] and mean exerted force [26]. However, in bilateral teleoperation, stability and transparency can be significantly affected by communication latency in the teleoperation loop, which dramatically reduces the effectiveness of haptic feedback in case of stiff remote environments [10, 22].

This limitation can be addressed by avoiding the use of any actuator on the master side or alleviated by designing proper control systems. In the former case, an interesting approach to still provide situational awareness is sensory substitution. Force feedback is not provided through the kinesthetic channel anymore and this makes the haptic loop intrinsically stable. Sensory substitution techniques replace this lack of kinesthetic feedback with other forms of feedback, such as vibrotactile [25], auditory, and/or visual feedback [14]. A similar approach, focusing on enhancing the transparency while preserving the intrinsic stability of the system, has been presented in [22]. The authors substituted haptic feedback, consisting of cutaneous and kinesthetic components [2, 11], with cutaneous feedback only, in order to make the system intrinsically stable and outperform other conventional sensory substitution techniques. In fact, cutaneous feedback does *not* affect the stability of teleoperation systems and, since it provides the operator with a direct and co-located perception of the contact force, it allows to perform motion tasks in an intuitive way [21, 22]. However, even though the aforementioned approach has been efficiently employed in many teleoperation scenarios [19, 21, 22], it still provides the user with less transparency than that achieved through kinesthetic force feedback.

For the above reason, the design of efficient control systems, able to provide kinesthetic feedback while guaranteeing the stability of the teleoperation loop, has been very challenging. The major goals while designing these systems are stability and transparency. Researchers have proposed a great variety of transparency- and stability-optimized bilateral controllers [12, 23] and it has always been a big challenge to find a good trade-off between these two objectives. In this respect, passivity [24] has been exploited as the main tool for providing a sufficient condition for stable teleoperation in several controller design approaches [8, 13]. However, control techniques guarantee the stability of the system at the price of a temporary loss of transparency, which could lead to degraded performance. For instance, Franken *et al.* presented a dual-layer controller structure [8]. A transparency layer is in charge of computing the ideal forces to be actuated at both the master and the slave, regardless of passivity constraints, while a passivity layer modulates such forces when this is necessary to avoid violations of the passivity condition, thus guaranteeing stability at the price of a temporary loss of transparency.

We can therefore consider two different channels when providing force feedback to humans: cutaneous and kinesthetic [22]. Cutaneous force feedback provides less transparency but does not affect the stability of the system, while kinesthetic feedback provides a realistic illusion of telepresence but it is affected by stability issues. A promising idea consists in controlling cutaneous and kinesthetic cues independently. This would help achieving better transparency while guaranteeing the stability of the teleoperation system: think if we could provide as much force feedback as we want through the cutaneous channel, and as much force feedback as the controller permits through the kinesthetic one. We could then partially compensate a lack of kinesthesia with cutaneous force feedback. However, in order to do so, we need to actuate independently the cutaneous and kinesthetic channels. Moreover, it is necessary to evaluate the performance improvement rate when compensating lack of kinesthesia with cutaneous force.

1.1 Cutaneous stimuli and kinesthesia

The idea behind the aforementioned approach of mixing cutaneous and kinesthetic cues originates from the observation that the stimuli received by a user while holding a haptic handle consist of a cutaneous and a kinesthetic component [22]. Cutaneous sensations are produced by pressure receptors in the skin and they are useful to recognize the local properties of objects such as shape, edges,

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embossings and recessed features, thanks to a direct measure of the intensity and direction of the contact forces [2]. On the other hand, kinesthesia provides the user with information about the relative position of neighbouring parts of the body, by means of sensory organs in muscles and joints [11].

Single-contact grounded haptic devices, such as the Omega[1] and Phantom[2] devices, are commonly employed in teleoperation systems to provide force feedback to the users. They provide both kinesthetic and cutaneous stimuli but it is *not* possible to decouple them. While using this type of haptic devices the force is applied at the end-effector and exerted both at the user's fingertips (i.e., the cutaneous component) and joints (i.e., the kinesthetic component). These two stimuli cannot be controlled independently. Moreover, the cutaneous component of the haptic interaction fed back by single-contact haptic devices depends not only on the shape of the object being virtually touched, but also on the shape of the haptic device end-effector and the way the user grasps it. Therefore, it is not trivial to perceive the shape of an object, in the absence of any exploratory movements, while using these kind of interfaces [7].

As already mentioned before, decoupling cutaneous and kinesthetic force feedback would permit to design more efficient control algorithms and provide a better illusion of telepresence in teleoperation. However, it remains unclear which is the best approach to decouple their control, how this affects the performance of teleoperation systems and if users feel comfortable with the proposed solutions. Towards this objective, let us define *haptic* force as the force being composed of cutaneous and kinesthetic components [22], i.e. the one provided by *haptic* interfaces such as the Omega. Cutaneous devices can then be defined as haptic devices able to provide solely cutaneous force feedback and thus not able of providing any kind of kinesthesia to the users. If an operator was able to wear one of these cutaneous devices and still use single-contact haptic devices in a natural way, we would attain the ability to easily decouple the control of cutaneous and kinesthetic components: we could provide cutaneous force feedback with cutaneous devices and haptic force feedback (i.e., both kinesthetic and cutaneous components) with the single-contact haptic interface.

The development of cutaneous devices with such characteristics has been very challenging. We need a cutaneous device able to provide force feedback at the fingertips and still allow the operator to use the end-effector of single-contact haptic devices in a natural way, i.e. we need a *wearable* cutaneous device. Moreover, it is important to provide the user with a cutaneous force feedback as similar as possible to the one provided by single-contact haptic interfaces. The cutaneous devices which are the closest, in terms of interaction modality, to grounded haptic devices are those able to apply three dimensional force vectors to the fingertip, since both are capable of providing forces at one contact point.

Literature on wearable haptic technologies is quite rich but most of the devices presented are not suitable to be used while operating with a commercial haptic device (i.e., they are not wearable enough) or they do not provide three dimensional force vectors at the fingertip. Glove-type haptic displays applying forces, such as the Rutgers Master II or the CyberGrasp, have been presented in [3, 16]. They provide force vectors to all five fingers of the hand simultaneously. However, the mechanics of such displays is complex, and thus compromises their wearability and portability. Wearability of force feedback devices has been dramatically improved in [17], where the authors presented a wearable and portable ungrounded haptic display able to apply cutaneous forces to simulate weight sensations of virtual objects. The approach was based on the novel insight that cutaneous sensations make a reliable weight illusion, even when the proprioceptive information is absent. The device consisted of two motors and a belt able to deform the fingertip. When motors spin in opposite directions the belt applies a force perpendicular to the contact surface on the user's fingertip while, if motors spin in the same direction, the belt applies a tangential force to the skin. This device was used in [21, 22] to validate a sensory substitution technique in teleoperation and in [20] for experiences of remote tactile interaction. However, the device proposed in [17] cannot render forces in all directions, it has only two motors, the force control is open-loop and it is not very accurate. The main issue is that its control accuracy largely depends on the visco-elastic parameters of the finger pad, which change with different subjects.

[1]Force Dimension, Nyon, Switzerland.

[2]Sensable group, Geomagic, 3D Systems, Rock Hill, SC, USA.

(a) (b)

Figure 1: The 3-DoFs wearable haptic display. The motors, acting on the lengths of the three wires, provide the requested force F_p to the user's fingertip (red arrow). This force is balanced by another one, F_r, supported by the structure of the device on the back of the finger (blue arrow).

The cutaneous device employed in this work is a wearable 3-DoFs device, described in Sec. 2 and shown in Fig. 1. It is similar to the one presented in [17] but it has one additional degree-of-freedom, close-loop control, and higher accuracy [4].

1.2 Contribution

In this work we discuss how and to what extent cutaneous force feedback can compensate a lack of kinesthesia in teleoperation without significantly affecting users' performance.
Towards this objective, in order to independently control the cutaneous and kinesthetic channels, we present a suitable (i.e., efficient and wearable) cutaneous device. Then we discuss how to employ a couple of the aforementioned cutaneous devices to fully actuate the cutaneous channel when kinesthesia is scaled down. Finally, we present two experiments, carried out to evaluate the proposed idea and the performance improvement rate when compensating a lack of kinesthesia with cutaneous force. Sec. 2 introduces the wearable cutaneous devices employed, while the two experiments are presented and discussed in Sec. 3. Sec. 4 addresses concluding remarks and perspectives of the work. It is also worth highlighting that, with respect to the works presented in [19, 22, 21], we employ cutaneous stimuli here, provided by cutaneous devices, *together* with haptic force feedback, provided by single-contact grounded interfaces, thus expecting improved performance [6].

2 A 3-DoFs wearable cutaneous device

The cutaneous device used in this work has been presented in [4] and it is shown in Fig. 1. It consists of two main parts: the first one is placed on the back of the finger and supports three small electrical motors; the other one is a mobile platform in contact with the volar skin surface of the fingertip. These two parts are connected by three cables. The motors, by controlling the length of the cables, are able to move the platform towards the user's fingertip. As a result, a force is generated, simulating the contact of the fingertip with an

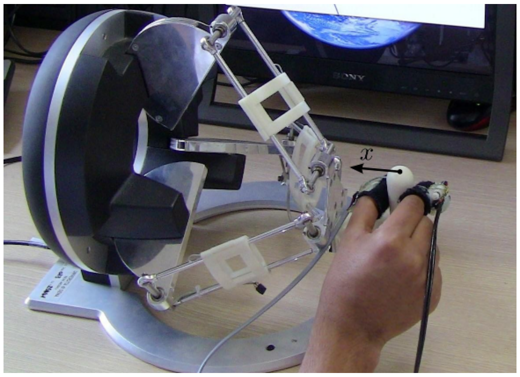


Figure 2: Experimental setup. The users were asked to wear two cutaneous devices, one on the thumb and one on the index finger, and grasp the handle with their right hand, positioned with its longitudinal axis at 90 deg from the Omega x-axis.

arbitrary surface. The direction and amount of the force reflected to the user is changed by properly controlling the cable lengths. The actuators used for the device prototype are three 0615S Falhauber motors, with planetary gear-heads having 16:1 reduction ratio. The maximum stall torque of the motor is 3.52 mNm. Three piezoresistive force sensors (400 FSR^{TM} Interlink Electronics) are placed near to the platform vertices, in contact with the finger, so that they measure the components of the cutaneous force applied to the fingertips. They have a diameter of 5 mm and a thickness of 0.3 mm, making them very transparent for the user and thus easily integrated with the device. The total weight of the device, including sensors, actuators, wires, and the mechanical support is about 30 g.

The system can be modelled as a three DoFs parallel mechanism, where the static part is fixed on the back side of the finger, and the mobile platform, or end-effector, is in contact with the finger pad. The mobile platform is moved acting on three cables connecting its vertices to three actuators, making the system able to arbitrary orient the platform on the user's fingertip. The force applied by the device to the finger pad is balanced by a force supported by the structure of the device on the back of the finger. This structure has a larger contact surface with respect to the mobile platform so that the local pressure is much lower, and the contact is mainly perceived on the finger pad and not on the back side of the finger (see Fig. 1). The overall light weight of the device and the small dimension of the mobile platform makes this cutaneous device suitable to be used together with common grounded haptic interfaces (see Fig. 2 and [22]).

Note that, since the experiments described in Sec. 3 consist of a 1-DoF telemanipulation task, this wearable device was used there as a 1-DoF cutaneous device (all motors pulled the cables together), so that only the forces in the sagittal plane of the finger were actuated, roughly normal to the longitudinal axis of the distal phalanx.

3 Evaluation of the proposed approach

In order to better understand the role of cutaneous and kinesthetic feedback in teleoperation, two experiments were carried out. The scenario considered is a teleoperated needle insertion in soft tissue, along one direction [5, 22]. This scenario has been chosen since it is a simple but meaningful example of teleoperation task. When performing keyhole neurosurgery, for instance, the needle can be steered using a haptic device such as the Omega, and the motion of the needle will be along one direction only [5].

The experimental setup is shown in Fig. 2. The master side consisted of a handle fixed to the end-effector of an Omega device, whose motion was constrained along one direction by three rigid clamps fixed to its parallel structure. The users were asked to wear two cutaneous devices, one on the thumb and one on the index finger, and grasp the handle as shown in Fig. 2. The subject's hand was positioned with its longitudinal axis at 90 deg from the Omega x-axis. The position of the subject's hand with respect to the joystick was checked before the beginning of each experiment. To prevent changes in the perceived direction of the force feedback generated by the Omega, the users were instructed to move the forearm rather than the wrist while moving the device. During the experiments, the subjects maintained the initial orientation of the fingers with respect to the end-effector, which was a natural way of grasping the handle for the given 1-DoF task.

The task consisted in inserting the needle into a soft tissue and stopping its motion as soon as a virtual stiff constraint was perceived. The stiff constraint played the role of a virtual fixture [1], i.e. a software function used in assistive robotic systems to regulate the motion of surgical implements, the needle in our case. Its motion is still controlled by the surgeon, but the system constantly monitors it and takes action if the surgical tool fails to follow a predetermined procedure. Virtual fixtures play two main roles: they can either guide the motion or strictly forbid the surgeon from reaching certain regions. In this experiment, we consider an example of forbidden-region virtual fixture, in charge of preventing the needle from entering a specific area of the workspace.

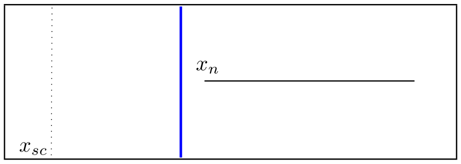
A virtual environment simulating a needle insertion in soft tissue has been implemented. Its model is described in [22]. The operator steering the needle remotely felt a resistive force while penetrating the tissue, due to its visco-elastic properties, and an opposite force while trying to pull the needle out. In real scenarios, these forces are either measured by force sensors or estimated from other parameters. A spring $K_t = 1.8$ N/m and a damper $B_t = 4.5$ N s/m were used to model the contact force between the needle and the tissue, while a spring $K_{sc} = 1800$ N/m was used to model the contact force F_{sc} between the needle and the stiff constraint. For the sake of simplicity, we assumed that the mass of the tissue $M_t = 1$ kg was concentrated at the contact point. The viscosity coefficient of the body beneath the tissue was $V_t = 0.8$ N s/m.

As for the haptic rendering, the interaction was designed according to the god-object model [27] and the position of the Omega handle was linked to the needle position x_n moving in the virtual environment. The tissue position changed according to the interaction with the needle, which was able to penetrate the surface only when the exerted force F_t was larger than a predetermined threshold ($F_p = 0.1$ N). To have a wider workspace, a scale factor of 3 between the position of the needle in the virtual environment and the operator's hand was used.

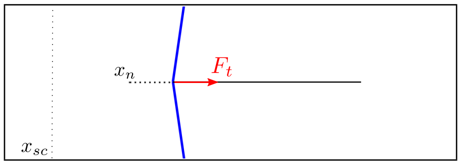
It was thus possible to discriminate four different operating conditions for the needle-tissue interaction model presented here: no contact (see Fig. 3a), contact without penetration, penetration within the safe area (see Fig. 3b), and penetration and contact with the stiff constraint (see Fig. 3c). In the first case, since the needle was out of the tissue, the model was designed to feed back no force to the operator and the surface of the tissue tended to return to its predetermined initial position. When the needle touched the tissue, but the force F_t was not yet sufficient to penetrate it, the tissue surface was deformed by the movement of the needle. As soon as $F_t > F_p$, the needle penetrated the surface. If the operator steered the needle towards the unsafe area delimited by the stiff constraint, located at x_{sc}, a force was fed back to the operator in order to prevent the penetration of the needle:

$$F_{sc} = -K_{sc}\ (x_n - x_{sc}).$$

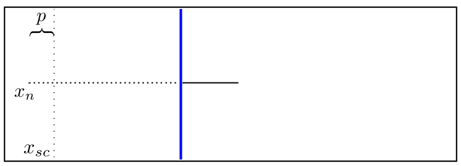
The haptic device measured the position of the operator's hand (with a resolution of 0.01 mm), sent it to the controller and then the virtual environment computed the force feedback and the dynamics of the tissue. The controller thereon sent the force back to the user



(a) no contact with the soft tissue



(b) needle penetrating the soft tissue



(c) penetration inside the stiff constraint

Figure 3: The virtual environment was composed of the needle (black), driven by the operator, the deformable tissue (blue), and the stiff constraint (red). The position of the needle x_n was linked to the position of the haptic device end-effector, the position of the stiff constraint was fixed to x_{sc}, and p represented the penetration of the needle inside the stiff constraint. Dashed parts were not visible by the subjects.

through the grounded haptic interface and cutaneous devices. The operator could see the part of the needle outside the tissue and the tissue surface, while the position of the stiff constraint and the part of the needle inside the tissue were not visible (see Fig. 3, dashed parts were not visible).

Ten participants (8 males, 2 females, age range 23 – 29) took part in the experiment, all of whom were right-handed. Seven of them had previous experience with haptic interfaces. None of the participants reported any deficiencies in their perception abilities.

The task consisted in inserting the needle into the soft tissue and stopping the motion as soon as the stiff constraint was perceived (i.e., the users had to remain in contact with the stiff constraint). After 5 s of continuous contact with the constraint, the system played a sound. The users were instructed to pull the needle out of the tissue as soon as the auditory signal was heard. The position of the stiff constraint x_{sc} was chosen randomly at the beginning of each trial.

3.1 Decoupling of cutaneous and kinesthetic channels

As already discussed in Sec. 1.1, single-contact haptic devices, such as the Omega interface, provide haptic force feedback composed of a cutaneous part $F_{c,h}$ and a kinesthetic part $F_{k,h}$ [22]. The complete haptic force feedback provided by these devices can be thus defined as $F_h = [F_{c,h} \;\; F_{k,h}]^T$. As discussed before, the main issue is that these two components cannot be independently controlled. For this reason, in order to decouple cutaneous and kinesthetic stimuli, we employed the cutaneous devices described in Sec. 2, which are able to provide cutaneous feedback only $F_c = [F_{c,c} \;\; 0]^T$.

The total amount of force F_t fed back to a user wearing the wearable cutaneous devices while interacting with the Omega handle can be thus expressed as

$$\begin{cases} F_t = \alpha\, F_h + \beta\, F_c = \alpha\, [F_{c,h} \;\; F_{k,h}]^T + \beta\, [F_{c,c} \;\; 0]^T \\ \quad\;\, = [\alpha\, F_{c,h} + \beta\, F_{c,c} \;\;\; \alpha\, F_{k,h}]^T \\ \alpha, \beta \geq 0 \end{cases}$$

where F_c is the force provided by the wearable cutaneous devices and F_h is the one provided by the Omega haptic interface. The variables α and β indicate the force to be fed back through, respectively, the Omega and the cutaneous devices.

We can then define

$$F_t = [\alpha\, F_{c,h} + \beta\, F_{c,c} \;\;\; \alpha\, F_{k,h}]^T = [F_{c,t} \;\; F_{k,t}]^T, \tag{1}$$

where $F_{c,t}$ and $F_{k,t}$ represent, respectively, the cutaneous and kinesthetic components of the full interaction. Since we are considering $F_{c,t}$ as a combination of $F_{c,h}$ and $F_{c,c}$, it is now clear why it is important to employ cutaneous devices which are very close, in terms of interaction modality, to grounded haptic devices. It is also worth highlighting the importance of the subject's fingertip position with respect to the haptic interface's end-effector, as already mentioned at the beginning of Sec. 3 and shown in Fig. 2. Equation 1 shows that the proposed approach makes it possible to control independently the cutaneous and kinesthetic channels, even if with some limitations[3].

As already discussed, since cutaneous force feedback does not affect the stability of teleoperation systems while kinesthetia does, we want to evaluate the performance improvement rate while compensating a lack of kinesthetic feedback through the cutaneous channel. Towards this objective, we are going to *fully* actuate the cutaneous channel, i.e. $\alpha + \beta = 1$, while reducing the force provided through the kinesthetic one (tasks HC_*). We will then evaluate the performance improvements with respect to employing a single-contact haptic interface only, i.e. considering $\beta = 0$ (tasks H_*).

3.2 Experimental results

Each participant made eighteen repetitions of the needle insertion task with force feedback provided by *both* the cutaneous devices and the Omega, with three randomized trials for each feedback modality HC_*:

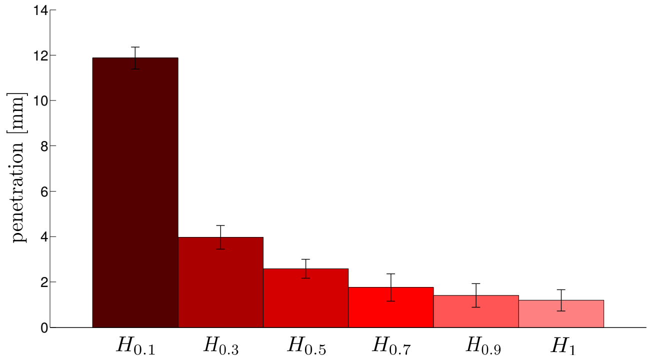
- task HC_0, $\alpha = 0$ and $\beta = 1$ (i.e., cutaneous feedback only),
- task $HC_{0.1}$, $\alpha = 0.1$ and $\beta = 0.9$,
- task $HC_{0.3}$, $\alpha = 0.3$ and $\beta = 0.7$,
- task $HC_{0.5}$, $\alpha = 0.5$ and $\beta = 0.5$,
- task $HC_{0.7}$, $\alpha = 0.7$ and $\beta = 0.3$,
- task $HC_{0.9}$, $\alpha = 0.9$ and $\beta = 0.1$,

and eighteen repetitions of the needle insertion task with force feedback by the Omega only, with three randomized trials for each feedback modality H_*:

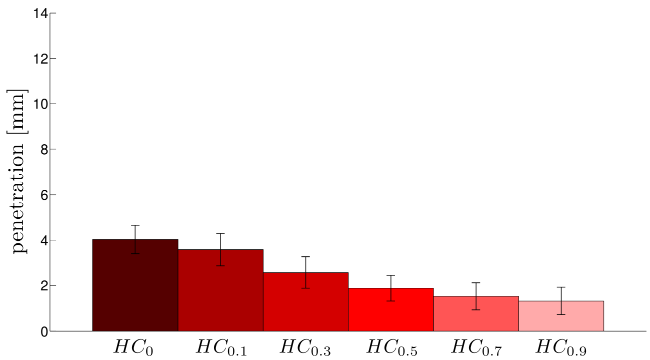
- task $H_{0.1}$, $\alpha = 0.1$ and $\beta = 0$,
- task $H_{0.3}$, $\alpha = 0.3$ and $\beta = 0$,
- task $H_{0.5}$, $\alpha = 0.5$ and $\beta = 0$,
- task $H_{0.7}$, $\alpha = 0.7$ and $\beta = 0$,
- task $H_{0.9}$, $\alpha = 0.9$ and $\beta = 0$,
- task H_1, $\alpha = 1$ and $\beta = 0$ (i.e., full Omega feedback).

The trials above were randomly presented to the subjects. They were asked to wear the two cutaneous devices, grasp the end-effector of the Omega with their right hand, as shown in Fig. 2,

[3] The system, for example, cannot provide more force through the kinesthetic channel than what it can through the cutaneous one. However, that scenario, since we are willing to reduce kinesthesia in favour of cutaneous cues, is not relevant towards our objectives.

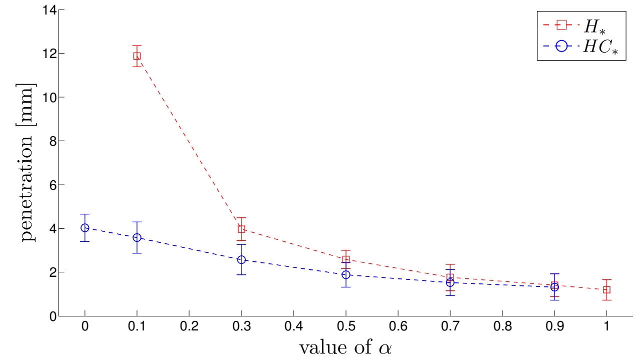


(a) tasks H_*

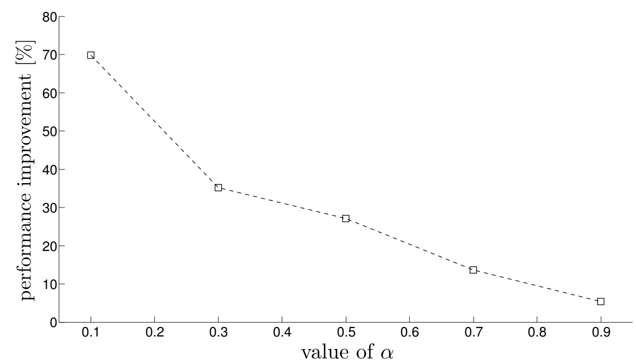


(b) tasks HC_*

Figure 4: Experimental results. Average penetration beyond the stiff constraint (mean and standard deviation), for tasks H_* and HC_*. A null value of this metric indicates high accuracy in reaching the target depth.



(a) Average penetration beyond the stiff constraint (mean and std)



(b) Performance improvement from H_* to HC_*

Figure 5: Experimental results. Average penetration beyond the stiff constraint (mean and standard deviation) and performance improvement from H_* to HC_* in terms of penetration beyond the stiff constraint.

and complete the thirty-six randomized repetitions of the needle insertion task, i.e. trials from tasks HC_* and H_* were mixed together. The force fed back to the users is expressed in eq. 1, where α and β were set accordingly to the task being performed (see the list above). In tasks HC_* the force was fed back by both the cutaneous devices and the Omega, and the cutaneous component of the haptic interaction $F_{c,t}$ was always fully actuated, i.e. $\alpha + \beta = 1$. In tasks H_* the force was fed back only by the Omega device, i.e. $\beta = 0$.

Since the subjects were asked to stop as soon as the virtual stiff constraint was perceived, the average penetration inside the stiff constraint provided a measure of accuracy in reaching the target depth. A null value in the metrics denotes the best performance, while a positive value indicates that the subject overran the target.

Fig. 4a and 4b shows, respectively, the average penetrations, beyond the stiff constraint, for each feedback modality in tasks H_* and HC_* (means and standard deviations are plotted). Moreover, in Fig. 5a penetrations during tasks H_* and HC_* are plotted together. Fig. 5b shows the improvement, in terms of penetration beyond the stiff constraint, from tasks H_* to HC_*, with respect to α.

The collected data of each task passed the D'Agostino-Pearson omnibus K2 normality test. Then a parametric two-tailed paired t-test (a = 0.05) was performed to evaluate the statistical significance of the differences between tasks H_* and HC_* providing the same kinesthetic force feedback, i.e. tasks with the same value for α (e.g. $H_{0.1}$ and $HC_{0.1}$). The p-values revealed a statistically significant difference between $H_{0.5}$ and $HC_{0.5}$, $H_{0.3}$ and $HC_{0.3}$, and $H_{0.1}$ and $HC_{0.1}$. The results of this experiment indicate that the subjects, while receiving force feedback through the Omega only (tasks H_*), reached a greater average penetration in the stiff constraint (worst performance) in comparison with that obtained while receiving feedback also from the cutaneous devices (tasks HC_*). This is, of course, more evident as the value of α decreases. Moreover, no difference between groups was observed in terms of task completion time. However, note that the performance of the proposed approach is directly related to the efficiency of the wearable cutaneous devices employed. The research on the development of devices with such characteristics is therefore paramount for the prospectives of this work.

In addition to the quantitative evaluation of the system performance, we also measured users' experience in using the proposed wearable cutaneous devices in the aforementioned teleoperation scenario. Users' experience was measured by a questionnaire using bipolar Likert-type seven-point scales. The questionnaire considered the perceived comfort, i.e. the *lack* of comfort due to wearing the external cutaneous devices with respect to using the Omega device without any cutaneous device. An answer of 7 meant a very good comfort, while an answer of 1 meant a very bad comfort. The questionnaire consisted of 4 questions. The mean value for the answers was 5.9.

4 Conclusion and Future Works

A study on the role of cutaneous and kinesthetic force feedback in teleoperation has been presented. The idea behind this works originated from the observation that the stimuli received by a user while holding a haptic handle consist of a cutaneous and a kinesthetic component. Cutaneous feedback provides less transparency than kinesthesia but it does not affect the stability of the teleoperation system, while kinesthesia provides a realistic illusion of telepresence but it affects the stability of the haptic loop. However, when employing common haptic interfaces, such as the Omega devices, it is not possible to decouple the cutaneous and kinesthetic components of the haptic interaction, i.e. it is not possible to control them independently. For this reason, many control techniques ensure a stable interaction by scaling down both kinesthetic and cutaneous force feedback, even though acting on the cutaneous channel is not necessary.

In order to decouple the control of the two components of the haptic interaction, we employed a couple of wearable cutaneous devices. They are capable of providing cutaneous force feedback only and can be easily worn while using the end-effector of an Omega device. In this way we could control independently the cutaneous and kinesthetic channels and cutaneous stimuli could thus be employed to compensate for the lack of kinesthetic feedback required to guarantee the stability of the teleoperation loop.

We carried out two experiments comparing performance while providing solely haptic feedback by the Omega interface and while providing both haptic and cutaneous stimuli through the Omega and the wearable cutaneous devices. We scaled down the force fed back through the Omega, compensating for the lack of force with the cutaneous devices. In this way the cutaneous channel was always fully actuated while kinesthesia was scaled down. Results showed improved performance while employing the aforementioned compensation technique.

The main drawback was that the performance of the proposed approach highly depended on the efficiency of the cutaneous devices employed. For this reason, work is in progress to design new cutaneous displays with better dynamic performance and wearability, in order to improve the results hereby registered. Moreover, we are planning to evaluate the effects of over-actuation of the cutaneous channel, i.e. having $\alpha+\beta>1$. Work is also in progress to validate the approach with more subjects and in a more challenging scenario (i.e., needle insertion in 3-DoFs).

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