Improving transparency in teleoperation by means of cutaneous tactile force feedback

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A study on the role of cutaneous and kinesthetic force feedback in teleoperation is presented. Cutaneous cues provide less transparency than kinesthetic force, but they do not affect the stability of the teleoperation system. On the other hand, kinesthesia provides a compelling illusion of telepresence but affects the stability of the haptic loop. However, when employing common grounded haptic interfaces, it is not possible to independently control the cutaneous and kinesthetic components of the interaction. 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.

We discuss here the feasibility of a novel approach. It aims at improving the realism of the haptic rendering, while preserving its stability, by modulating cutaneous force to compensate for a lack of kinesthesia. We carried out two teleoperation experiments, evaluating (1) the role of cutaneous stimuli when reducing kinesthesia and (2) the extent to which an over-actuation of the cutaneous channel can fully compensate for a lack of kinesthetic force feedback. Results showed that, to some extent, it is possible to compensate for a lack of kinesthesia with the aforementioned technique, without significant performance degradation. Moreover, users showed a high comfort level in using the proposed system.

Categories and Subject Descriptors: H.5.2 [Information interfaces and presentation]: User interfaces—Haptic I/O, Evaluation/methodology; H.1.2 [Models and Principles]: User/Machine Systems—Human Information Processing; H.5.1 [Information Interfaces and Presentation]: Multimedia Information Systems—Artificial, augmented, and virtual realities

General Terms: Experimentation, Human Factors, Performance

Additional Key Words and Phrases: Haptic interfaces, Cutaneous force feedback, Tactile force feedback, Teleoperation

1. INTRODUCTION

A teleoperator is a machine which allows humans to sense and mechanically manipulate objects at a distance, by virtually relocating the operator at a place other than his true location [Sheridan 1995]. It includes artificial sensors in order to perceive the environment, actuators to be able to move into this environment, and network channels to communicate with the human operator. In order to perform complex tasks, teleoperators may also include artificial devices (e.g., arms or hands) to apply forces and perform mechanical work on the environment.
If the user receives sufficient information about the teleoperator and the remote environment he is interacting with, he feels as he is actually present at the remote site. This condition is commonly referred to as telepresence [Draper et al. 1998]. Achieving a good illusion of telepresence is a matter of technology. If the teleoperator transmits sufficient information to the user, displayed in a sufficiently articulated way, the illusion of telepresence can be compelling [Sheridan 1995; Draper et al. 1998]. The primary tool to achieve this objective is providing a transparent implementation of the teleoperation system. Transparency can, in turn, be defined as the correspondence between the master and slave positions and forces [Hashtrudi-Zaad and Salcudean 2002], or as the match between the impedance of the environment and the one perceived by the operator [Lawrence 1993]. A compelling illusion of telepresence can be achieved through different types of information, which flow from the remote scenario to the human operator. Haptic force feedback is one piece of this information flow and it is proved to play an important role in enhancing teleoperation performance in terms of task completion time [Massimino and Sheridan 1994; Moody et al. 2002; Pacchierotti et al. 2012a], accuracy [Moody et al. 2002; Prattichizzo et al. 2012], peak [Hannaford 1987; Wagner et al. 2002], and mean exerted force [Wagner et al. 2002; Pacchierotti et al. 2012a].

Another important goal while designing this type of teleoperation systems is stability [Hashtrudi-Zaad and Salcudean 2002]. It is a key design consideration in haptic systems, since unwanted oscillations may be unsafe for the human operator and the remote environment he is interacting with. The stability of such systems can be significantly affected by communication latency in the teleoperation loop, hard contacts, a relaxed grasp of the user, and many other destabilizing factors which dramatically reduce the effectiveness of haptic force feedback in teleoperation [Franken et al. 2011; Prattichizzo et al. 2012].

This limitation can be addressed by avoiding the use of any actuator on the master side or by designing appropriate control systems. In the former case, no force feedback is provided through the master end-effector, thus making the teleoperation loop intrinsically stable [Prattichizzo et al. 2012]. In the absence of any actuators on the master side, an effective approach to still provide situational awareness is sensory substitution: it consists of replacing a lack of kinesthetic force with other forms of feedback, such as vibrotactile [Schoonmaker and Cao 2006], auditory, and/or visual feedback [Kitagawa et al. 2005]. A similar approach, focusing on enhancing the transparency while preserving the intrinsic stability of the system, has been presented in [Prattichizzo et al. 2012]. The authors substituted haptic force feedback, provided by a common single-contact haptic interface, with tactile stimuli, provided by wearable cutaneous devices, in order to have an intrinsically stable system while outperforming other conventional sensory substitution techniques. They also showed how delays in the communication loop, together with a stiff remote environment, affected the stability of the system while using the grounded device but not while using the cutaneous ones.

However, even though that approach has been successfully tested in many teleoperation scenarios [Pacchierotti et al. 2012a; Prattichizzo et al. 2010b; Prattichizzo et al. 2012], it still provides the user with less transparency than that achieved through kinesthetic force feedback.

For this reason, the design of efficient control systems, able to provide kinesthesia while guaranteeing the stability of the teleoperation loop, has been always considered very important. The major goals while designing these control systems are stability and transparency. Researchers have proposed a great variety of transparency- and stability-optimized bilateral controllers [Hokayem and Spong 2006; Salcudean 1998; Franken et al. 2011] and it has always been a big challenge to find a good trade-off between these two objectives. 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, in [Franken et al. 2011], a dual-layer controller structure is presented. A transparency layer is in charge of computing the ideal forces to be actuated at both the master and the slave, regardless of stability...
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How can we improve the transparency of such systems without affecting their stability?

Towards this challenging objective let us recall that haptic force feedback provided by common grounded haptic devices is sensed by the operator through two different channels: cutaneous and kinesthetic [Birznieks et al. 2001; Hayward et al. 2004; Prattichizzo et al. 2012]. Cutaneous stimuli are sensed by pressure receptors in the skin, and they are useful to recognize the local properties of objects such as shape, edges, embossings and recessed features. This is possible, principally, thanks to a direct measure of intensity and direction of contact forces, and to the encoding of the force spatial distribution over the fingertip [Birznieks et al. 2001; Johnson 2001]. On the other hand, kinesthesia provides the user with information about the relative position of neighbouring parts of the body, mainly by means of sensory organs in muscles [Edin and Johansson 1995] and joints [Hayward et al. 2004]. As extensively discussed in [Prattichizzo et al. 2012; Tirmizi et al. 2013], cutaneous force feedback provides, in general, less transparency than kinesthesia but it does not affect the stability of the system. Kinesthetic force feedback, on the contrary, provides a compelling illusion of telepresence but it is affected by the aforementioned stability issues.

Single-contact grounded haptic devices, such as the Omega\textsuperscript{1} and Phantom\textsuperscript{2} interfaces, are commonly employed in teleoperation to control the movement of the slave system and provide force feedback to the human operator. They provide both kinesthetic and cutaneous stimuli but it is not possible to decouple them: the force they provide, referred here as haptic force [Prattichizzo et al. 2012], is applied at their end-effector and exerted both at the user’s fingertips (i.e., the cutaneous component) and joints (i.e., the kinesthetic component). That is why, when control algorithms regulate kinesthesia to guarantee the stability of teleoperation system, cutaneous force is affected as well, although this is not necessary: only kinesthesia needs to be modulated to guarantee the stability of such systems [Pacchierotti et al. 2012a; Prattichizzo et al. 2012].

Is thus possible to improve the transparency of teleoperation systems by developing control techniques which regulate kinesthesia only and leave the cutaneous component of the interaction unaltered?

In this work we discuss how and to what extent acting on the cutaneous channel can improve the transparency of teleoperation systems. In Sec. 2 we discuss how to independently control the cutaneous and kinesthetic components and, towards this objective, we present a novel cutaneous haptic device able to apply three dimensional force vectors at the fingertip. Then, in Sec. 3, we carried out two experiments which quantitatively evaluate to what extent acting on the cutaneous component can compensate for a lack of kinesthesia in teleoperation, improving performances without affecting the stability of the teleoperation loop. Sec. 4 addresses concluding remarks and perspectives of the work.

2. HAPTIC FORCE FEEDBACK: CUTANEOUS AND KINESTHETIC CHANNELS

With the aim of providing the user with a full cutaneous feedback while modulating kinesthesia, we need to independently control the forces provided through the kinesthetic and cutaneous channels. A promising idea consists in using a cutaneous device, i.e., a haptic device able to provide solely cutaneous force feedback, together with a common single-contact haptic device.

In this way we are able to provide

• cutaneous force feedback with the cutaneous device and

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Fig. 1: Prototypes of three wearable cutaneous devices able to provide force vectors at one contact point. The interface shown in Fig. 1a has been presented in [Minamizawa et al. 2007], it has 2-DoF and open-loop control. The one developed in [Chinello et al. 2012] is shown in Fig. 1b, it has 3-DoF and three force sensors on the mobile platform for closed-loop force control. The last one, showed in Fig. 1c, is the one we are going to employ in this work. It is similar to the one depicted in the middle but has improved performance, wearability and closed-loop force and position control.

- haptic force feedback (i.e., kinesthetic and cutaneous stimuli) with the single-contact haptic interface (e.g. a Phantom device).

However, before presenting them to the users, it is necessary to integrate stimuli coming from the cutaneous device with that being provided by the single-contact haptic device. We thus require a cutaneous interface which provides the user with a cutaneous force feedback as similar as possible to the one provided by single-contact haptic interfaces. The interfaces which are the closest, in terms of interaction modality, to grounded haptic devices are the ones able to apply three dimensional force vectors to the fingertip, since both are capable of providing 3-DoF forces at one contact point [Tirmizi et al. 2013; Prattichizzo et al. 2012]. Moreover, in addition to that, it is also crucial to employ a cutaneous device which leaves the operator free to interact with the end-effector of the given single-contact haptic device, i.e. we need a wearable cutaneous interface. A wearable device is thereby intended as a device which, once worn, still make the user able to use the end-effector of single-contact haptic devices in a natural way. Further information about the concept of wearability in haptics can be found in [Chinello et al. 2012; Prattichizzo et al. 2012].

2.1 Wearable cutaneous devices

Literature on cutaneous technologies is quite rich but most of the devices presented are either not suitable to be used in conjunction with a commercial haptic device (i.e., they are not wearable enough) or they do not provide three dimensional force vectors at the fingertip (i.e., their interaction modality is different with respect to that of single-contact haptic interfaces [Prattichizzo et al. 2012]).

Glove-type haptic displays, such as the CyberGrasp4, can provide force vectors to all five fingers of the hand simultaneously. However, the mechanics of such displays is very complex, thus compromising their wearability. A similar device has been developed by Solazzi et al., in [Solazzi et al. 2010], where the authors presented a wearable 3-DoF cutaneous display. However, its wearability is still limited by the mechanical structure. The motors are placed on the forearm and two cables for each actuated

3Pictures 1a and 1b are adapted from the data reported in [Minamizawa et al. 2007] and [Chinello et al. 2012], respectively.

4CyberGlove Systems LLC, San Jose, CA, USA.

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finger are necessary to transmit the motor torque to the fingertip. Tsagarakis et al. in [Tsagarakis et al. 2005], presented a cutaneous device where two motors were employed to feed back tangential forces to the user, in order to simulate relative lateral motion, in terms of direction and velocity, to the fingertip. In [Gleeson et al. 2010] the authors proposed a fingertip worn device with two degrees of freedom. The device used two RC servo motors and a compliant flexure stage to create planar motion. The servos can operate simultaneously, allowing motion along any path in a plane. Another interesting device has been developed in [Kuchenbecker et al. 2008], where the authors presented a fingertip device which provided the user with the cutaneous sensation of making and breaking contact with virtual surfaces. However, the thimble there presented has no actuation and relies on the kinesthetic force feedback provided by the haptic device it is attached to.

The wearability of this type of cutaneous devices has been significantly improved in [Minamizawa et al. 2007], where the authors presented a wearable ungrounded haptic display able to apply cutaneous forces to simulate weight sensations of virtual objects (see Fig. 1a). The device consisted of two motors and a belt able to deform the fingertip. When motors span in opposite directions the belt applied a force perpendicular to the contact surface on the user's fingertip while, if motors span in the same direction, the belt applied a tangential force to the skin. This device was also used in [Prattichizzo et al. 2010b; Prattichizzo et al. 2012] to validate a sensory substitution technique in teleoperation and in [Prattichizzo et al. 2010a] for remote tactile interaction. However, the device proposed in [Minamizawa et al. 2007] 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 being the control open-loop, its accuracy depends on the visco-elastic parameters of the finger pad, which change with different subjects.

Performance of this type of interfaces has been greatly improved in the 3-DoF wearable cutaneous device presented in [Chinello et al. 2012] and shown in Fig. 1b. 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 arbitrary surface. Three force sensors are placed near to the platform vertices, in contact with the finger, so that they measured the three components of the cutaneous force applied to the fingertip. The most relevant improvement with respect to the interface presented in [Minamizawa et al. 2007] is the capability of this device to be force-controlled.

The cutaneous device proposed in this work is a wearable 3-DoF cutaneous interface similar to the one presented in [Chinello et al. 2012] but showing higher accuracy and wearability. The device is shown in Fig. 1c and consists of two platforms: one fixed to 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 [Stein 1988]. Three small electrical motors, equipped with position encoders, control the length of the cables, thus being able to move the platform towards the fingertip. One force sensor is placed at the platform’s centre, 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 to the user and easy to integrate with the mobile platform. Its overall light weight, 35 g, and the small dimension of the mobile platform, make this cutaneous device suitable to be used together with common grounded haptic interfaces (see Fig. 2a and [Prattichizzo et al. 2012]). The actuators used in our prototype are three 0615S DC micromotors\(^5\), with planetary gear-heads having 16:1 reduction ratio. The maximum stall torque of the motor is

\(^5\)Dr. Fritz Faulhaber GmbH & Co., Schönaich, Germany.
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(a) User performing the 1-DoF teleoperation task. (b) Front view of the handle. (c) Top view of the handle.

Fig. 2: Experimental setup. Users were asked to wear one cutaneous device on the index finger and grasp the handle, positioned with its longitudinal axis at 90 deg from the Omega $x$-axis.

3.52 mNm. The encoders are 10-bit rotary position sensors named AS5040$^6$ and the force sensors is a piezoresistive 400 FSR$^7$. With respect to the interface presented in [Chinello et al. 2012], this device can be controlled both in force and position. Moreover, the smaller size improves its wearable.

Note that, since the experiments described in Sec. 3 consist of a 1-DoF telemanipulation task, this wearable device was there used 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.

2.2 Decoupling of cutaneous and kinesthetic channels

Consider a teleoperation scenario, as shown in Fig. 2a, where an Omega 3 and a cutaneous haptic device are used simultaneously to provide force feedback to the user. Without affecting the generality of the discussion, for the sake of simplicity, let us assume that the teleoperation task is performed along a single DoF. The motion of the Omega has been constrained along this single direction (i.e. the $x$-direction, as shown in Fig. 2a) by three rigid clamps fixed to its parallel structure. The cutaneous device is the one described in Sec. 2.1 and shown in Fig. 1c. Users wear it on the index finger, and grasp the handle as shown in Fig. 2b. Subject's hand is thus positioned with its longitudinal axis at 90° from the Omega’s $x$-axis$^8$. Consider the case of the hand moving towards the positive direction of the $x$-axis, and thus the force to be directed towards the index finger only (see Fig. 2).

As already discussed before, the perceived haptic force, fed back by single-contact haptic devices such as the Omega interfaces, consists of a cutaneous component $F_{c,h}$ and a kinesthetic component $F_{k,h}$. The complete haptic force feedback provided by these devices can be thus defined as $F_h = [F_{c,h} F_{k,h}]^T$, where superscript $T$ denotes the transpose. However, the main issue regarding grounded haptic interfaces is that the two components $F_{c,h}$ and $F_{k,h}$ cannot be independently controlled. For this reason, in order to decouple the control of cutaneous and kinesthetic stimuli, we employed a cutaneous device, which is able to provide cutaneous force feedback only $F_c = [F_{c,c} 0]^T$.

$^6$AMS AG, Unterpremstaetten, Austria.
$^7$Interlink Electronics, Camarillo, CA, USA.
$^8$A short video about the experimental evaluation presented in Sec. 3 can be downloaded at http://goo.g1/tHK2g. It may also help to better understand the experimental setup and teleoperation task considered here.
Since the cutaneous stimuli applied by the cutaneous device are similar, in terms of interaction modality, area of application and magnitude, to that applied by the Omega interface (see Sec. 2.1), we can consider the two cutaneous components $F_{c,c}$ and $F_{c,h}$, even if provided through two different devices, as *additive* components. In other words, the cutaneous force $F_{c,c}$, provided by the cutaneous devices and sensed by the user, can be summed up with the one provided through the Omega haptic interface, $F_{c,h}$. The validity of this assumption is supported by the studies carried out in [Prattichizzo et al. 2012; Pacchierotti et al. 2012a; Prattichizzo et al. 2010b; Tirmizi et al. 2013], where cutaneous force feedback provided by cutaneous devices was successfully employed in teleoperation to partially substitute haptic force feedback provided by an Omega interface.

The total amount of force $F_t$, fed back to the user by both the Omega and the cutaneous interface, can thus be expressed as

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

where $F_c$ is the force provided by the wearable cutaneous devices, $F_h$ the one provided by the Omega haptic interface, and variables $\alpha$ and $\beta$ indicate the fraction of force fed back through the two devices.

We can then distinguish between the cutaneous and kinesthetic force feedback

$$F_t = \begin{bmatrix}
\alpha F_{c,h} + \beta F_{c,c} \\
\alpha F_{k,h}
\end{bmatrix}^T = [F_{c,t} \quad F_{k,t}]^T,$$

where $F_{c,t}$ and $F_{k,t}$ represent, respectively, the cutaneous and kinesthetic component of the complete haptic interaction. This clearly shows how acting on $\alpha$ and $\beta$ makes possible to independently modulate the force provided through the two perceptive channels, even though with some limitations: the system, for example, cannot provide more force through the kinesthetic channel than through the cutaneous one. However, that scenario is not relevant towards our objectives, since we are willing to reduce kinesthesia in favour of cutaneous cues.

It is important to underline that eq. 1 and 2 hold true not only in our simplified 1-DoF scenario, but also in the case of any other multi-DoF interaction. The only thing that matters is using a cutaneous device providing cutaneous stimuli similar, in terms of interaction modality, area of application and magnitude, to the ones provided by the single-contact haptic device. However, for the sake of clarity and simplicity, in this work we deal with 1-DoF scenarios only. Further validation of this approach for 2- and 3-DoF systems will be considered in the near future.

From eq. 2 it is also possible to define the terms *under-actuation*, *over-actuation* and *(full) actuation* of the kinesthetic and cutaneous channels. These definitions will be useful towards the evaluation of our approach, presented in Sec. 3.

The cutaneous component of the interaction $F_{c,t}$ is considered

(i) under-actuated if $\alpha + \beta < 1$,
(ii) (fully) actuated if $\alpha + \beta = 1$, and
(iii) over-actuated if $\alpha + \beta > 1$.

On the other hand, the kinesthetic component $F_{k,t}$ is considered

(i) under-actuated if $\alpha < 1$,
(ii) (fully) actuated if $\alpha = 1$, and
(iii) over-actuated if $\alpha > 1$,

where $\alpha$ and $\beta$ represent the fraction of force provided by the Omega and the cutaneous device, respectively (see eq. 2). A simple application of these concepts is shown below.

**EXAMPLE.** Consider the situation where we want the cutaneous channel to be (fully) actuated, regardless of what happens to the kinesthetic one, i.e. $\alpha + \beta = 1$. In order to do so, when the haptic force feedback is scaled down ($\alpha < 1$), we can compensate for this lack of haptic force through the cutaneous device, increasing $\beta$ (i.e., providing additional force with the cutaneous display). Of course, this only affects the cutaneous channel: the kinesthetic component of the interaction, $F_{k,t} = \alpha F_{k,h}$, is left unaltered.

3. EXPERIMENTAL EVALUATION

In order to validate the proposed compensation approach and evaluate as to what extent acting on the cutaneous channel can improve the transparency of teleoperation systems, we carried out two experiments. The experimental setup is the same as described in Sec. 2.2. Users were asked to wear one cutaneous device on the index finger, and grasp the Omega’s end-effector as shown in Fig. 2.

The task consisted in teleoperating a virtual tool along one direction, until a stiff constraint was perceived. The stiff constraint played the role of an active constraint, i.e. a software function used in assistive robotic systems to regulate the motion of remote implements [Abbott et al. 2007]. In these experiments, we consider the case of a forbidden-region active constraint, which is in charge of preventing the teleoperated implement from entering a specific region of the workspace. This scenario has been chosen since it is a simple but relevant example of teleoperation task [Prattichizzo et al. 2012]. When performing keyhole neurosurgery, for instance, the surgical tool can be steered using a haptic device such as the Omega, and the motion of the tool will be along one direction only [De Lorenzo et al. 2011].

A virtual environment simulating the considered 1-DoF teleoperation task has been implemented. A spring $K_{sc} = 1800$ N/m was used to model the contact force, $F_{sc}$, between the tool and the stiff constraint. When the operator steered the remote tool towards the unsafe area delimited by the stiff constraint, located at $x_{sc}$, a force was fed back to him, in order to prevent the penetration of the tool:

$$F_{sc} = -K_{sc} (x_t - x_{sc}),$$

where $x_t$ represents the position of the tool.

The haptic device measured the position of the operator’s hand (with a resolution of 0.01 mm) and sent it to the controller. Then the virtual environment computed the force to be fed back and transmitted it to the user through the grounded haptic interface and cutaneous device, according to the considered feedback modality. Similarly to Sec. 2.2, since the motion of the hand is again directed towards the positive direction of the $x$-axis, the force is applied towards the index finger only (see Fig. 2b). To have a wider workspace, a scale factor of 3 between the position of the tool in the virtual environment and the operator’s hand is used.

The subject’s hand was positioned with its longitudinal axis at 90 deg from the Omega $x$-axis. Before the beginning of each experiment, the position of the subject’s hand with respect to the handle, the correct functioning of the devices and virtual environment were carefully checked. 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 the natural way of grasping the handle for the given 1-DoF task. As for the haptic rendering, the interaction was
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Table I. : Experiment #1. Variables $\alpha$ and $\beta$ represent the fraction of force provided, respectively, by the single-contact grounded haptic interface and the cutaneous device. In condition $HC_0.2$, for example, the force to be applied at the master side, $F_{sc}$, is provided by the Omega 3 and the cutaneous device scaled to the 20% and 80%, respectively. Therefore, the cutaneous channel is fully actuated ($\alpha + \beta = 1$) while the kinesthetic one is under-actuated ($\alpha = 0.2 < 1$). On the other hand, in condition $H_0.2$, the force $F_{sc}$ is provided through the Omega only, scaled by the 20%. This time both the cutaneous and kinesthetic channel are under-actuated ($\alpha = \alpha + \beta = 0.2$).

Users were asked to move the remote tool across the virtual environment and stop as soon as the stiff constraint was perceived (i.e., the users had to remain in contact with the stiff constraint). After 3 seconds of continuous contact with the constraint, the system played a sound. Users were instructed to move the tool back 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.

Sixteen participants (12 males, 4 females, age range 21 – 33) 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 and they were naïve as to the purpose of the study.

3.1 Experiment #1: full actuation of the cutaneous channel
The first experiment aimed at evaluating the performance reduction rate when reducing kinesthesia, and discussed to what extent a fully actuated cutaneous channel could mitigate the consequent performance degradation.
Each participant made sixty trials of the aforementioned teleoperation task:

- thirty trials with force feedback provided by both the cutaneous device and the Omega, with three randomized repetitions for each feedback modality $HC_0$, indicated in Table Ia, and

\[ \alpha + \beta = 1 \]

\[ \beta = 0 \]

This case has been also analysed in [Prattichizzo et al. 2012], where all the force feedback is provided through the cutaneous device and the Omega 3 is only employed to track hand's position.

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Fig. 3: Experiment #1. Average penetration beyond the stiff constraint (mean and standard deviation) and performance improvement from HC_α to H_α, in terms of penetration beyond the stiff constraint. Filled markers in (a) represent the modalities found statistically different. Lines represent the cubic approximation to the data sets.

- thirty trials with force feedback provided by the Omega only, with three randomized repetitions for each feedback modality H_α indicated in Table Ib.

Subjects were asked to wear the cutaneous device, grasp the end-effector of the Omega with their right hand, as shown in Fig. 2, and complete the sixty trials of the teleoperation task, i.e. repetitions with modalities HC_α and H_α were mixed together. Each participant was informed about the procedure before the beginning of the experiment and a 10-minutes familiarization period was provided with the Omega 3 alone as well as while using it together with the cutaneous device, in order to make the subjects acquainted with the experimental setup.

The force was provided to the users according to eq. 2, where α and β were set according to the feedback modality being tested (see Table I). In conditions HC_α the force was fed back by both the cutaneous device and the Omega, and the cutaneous component of the haptic interaction, F_{c,t}, was always fully actuated, i.e. α + β = 1 (this is also the case discussed in the example at the end of Sec. 2.2). In conditions H_α the force was fed back through the Omega device only, i.e. β = 0. Condition H_1, when all force was fed back through the Omega interface and the cutaneous device was switched off, is considered our control condition, i.e. the one from which we expect the best performance.

Since 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 [Franken et al. 2011; Prattichizzo et al. 2012]. A null value in the metrics denoted the best performance, while a positive value indicated that the subject overran the target.

Fig. 3a shows the average penetrations, beyond the stiff constraint, for each feedback condition HC_α (blue) and H_α (red). Fig. 3b shows the improvement, in terms of penetration beyond the stiff constraint, from H_α to HC_α with respect to α. Please note that this difference makes only sense for α ≤ 0.6, i.e. when H_α and HC_α were found statistically different.

The collected data of each modality passed the D’Agostino-Pearson omnibus K2 normality test. Comparison of the means among the feedback modalities (H_α vs HC_α), and among the kinesthetic feedback provided, α, was tested using a two-way ANOVA. The means differed significantly among both the feedback modalities (F_{1,270} = 1243.92, p < 0.001) and the kinesthetic feedback provided (F_{8,270} = 488.55, p < 0.001). Also the interaction between the factors was found significant (F_{8,270} = 42.950, p < 0.001). Post-hoc analysis (pair-wise post-hoc t-tests corrected according to Bonferroni) revealed statistically
It is also worth noting that in conditions HC α decrement of similar forbidden-region active constraint. They found an average penetration inside the constraint of devices (see Table Ia).

Humans perceive stiffness when dealing with cutaneous and kinesthetic stimuli.

The observed performances are in agreement with previous results. In [Prattichizzo et al. 2012] the authors carried out an experiment of simulated 1-DoF needle insertion in the presence of a similar forbidden-region active constraint. They found an average penetration inside the constraint of ∼ 1.7 mm for the feedback modality called here H₁, and of ∼ 3.9 mm for HC₀. In [Pacchierotti et al. 2012b] a novel 1-DoF cutaneous device was employed for a similar task, showing comparable results but, again, only cases H₁ and HC₀ were taken into account. In this paper we deeper studied the contribution of cutaneous feedback in teleoperation, analysing its involvement at different levels of actuation.

From these results it is also interesting, although maybe obvious, to notice how users, during tasks H₂, tended to stop when the force exerted by the Omega interface reached a certain reference value (∼ 2.5 N), regardless of the penetration inside the stiff constraint. During tasks HCα, as expected, this reference force exerted by the grounded haptic interface decreased, thanks to the additional cutaneous stimuli being provided. This may open interesting perspectives towards a better understanding of how humans perceive stiffness when dealing with cutaneous and kinesthetic stimuli.

<table>
<thead>
<tr>
<th>Feedback condition</th>
<th>α</th>
<th>β</th>
<th>α + β</th>
</tr>
</thead>
<tbody>
<tr>
<td>HCO₉.4.0.8</td>
<td>0.4</td>
<td>0.8</td>
<td>1.2</td>
</tr>
<tr>
<td>HCO₉.4.1.0</td>
<td>0.4</td>
<td>1.0</td>
<td>1.4</td>
</tr>
<tr>
<td>HCO₉.4.1.2</td>
<td>0.4</td>
<td>1.2</td>
<td>1.6</td>
</tr>
<tr>
<td>HCO₉.5.0.7</td>
<td>0.5</td>
<td>0.7</td>
<td>1.2</td>
</tr>
<tr>
<td>HCO₉.5.0.9</td>
<td>0.5</td>
<td>0.9</td>
<td>1.4</td>
</tr>
<tr>
<td>HCO₉.5.1.1</td>
<td>0.5</td>
<td>1.1</td>
<td>1.6</td>
</tr>
<tr>
<td>HCO₉.6.0.6</td>
<td>0.6</td>
<td>0.6</td>
<td>1.2</td>
</tr>
<tr>
<td>HCO₉.6.0.8</td>
<td>0.6</td>
<td>0.8</td>
<td>1.4</td>
</tr>
<tr>
<td>HCO₉.6.1.0</td>
<td>0.6</td>
<td>1.0</td>
<td>1.6</td>
</tr>
</tbody>
</table>

Table II: Experiment #2. Each participant made eighteen randomized trials of the teleoperation task, in addition to the one already performed for Experiment #1, with two randomized repetitions for each feedback modality shown in this table. Again, variables α and β represent the fraction of force provided, respectively, by the single-contact grounded haptic interface and the cutaneous device. In these modalities the cutaneous channel was always over-actuated, i.e. α + β > 1.

significant difference for conditions with α ≤ 0.6 (depicted as filled markers in Fig. 3a). However, it is worth noticing that also during modalities whose results were not found significantly different (α > 0.6), subjects still showed better performances when receiving additional cutaneous force feedback by the cutaneous device.

The results of this experiment indicate that the subjects, while receiving force feedback through the Omega only (conditions Hα), reached a greater average penetration in the stiff constraint (worse performance) as compared to that obtained while receiving feedback also from the cutaneous devices (conditions HCα). This is, of course, more evident as the value of α decreases.

It is also worth noting that in conditions HCα, since the cutaneous channel was always fully actuated, a decrement of α led only to a decrement of kinesthetic feedback. The correspondent lack of the cutaneous stimuli coming from the Omega was compensated by an increment of the force exerted by the cutaneous devices (see Table Ia).

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

The observed performances are in agreement with previous results. In [Prattichizzo et al. 2012] the authors carried out an experiment of simulated 1-DoF needle insertion in the presence of a similar forbidden-region active constraint. They found an average penetration inside the constraint of ∼ 1.7 mm for the feedback modality called here H₁, and of ∼ 3.9 mm for HC₀. In [Pacchierotti et al. 2012b] a novel 1-DoF cutaneous device was employed for a similar task, showing comparable results but, again, only cases H₁ and HC₀ were taken into account. In this paper we deeper studied the contribution of cutaneous feedback in teleoperation, analysing its involvement at different levels of actuation.

From these results it is also interesting, although maybe obvious, to notice how users, during tasks H₂, tended to stop when the force exerted by the Omega interface reached a certain reference value (∼ 2.5 N), regardless of the penetration inside the stiff constraint. During tasks HCα, as expected, this reference force exerted by the grounded haptic interface decreased, thanks to the additional cutaneous stimuli being provided. This may open interesting perspectives towards a better understanding of how humans perceive stiffness when dealing with cutaneous and kinesthetic stimuli.
3.2 Experiment #2: over-actuation of the cutaneous channel

The second experiment aimed at evaluating the extent to which an over-actuation of the cutaneous channel could improve the performances registered in Experiment #1. We expect to find cutaneous stimuli able to compensate for a certain lack of kinesthesia with no significant performance degradation.

Each participant made eighteen trials of the same 1-DoF teleoperation task presented in Sec. 3.1. Force feedback was provided by both the cutaneous device and the Omega, with three randomized repetitions for each feedback modality $H_{CO_{\alpha,\beta}}$ shown in Table II. The cutaneous channel was now over-actuated (i.e., $\alpha + \beta > 1$) in order to compensate for the progressive lack of kinesthetic force feedback. These trials were performed right after the ones described in Sec. 3.1, and the participants were not aware that they were performing a different experiment. The average penetration inside the stiff constraint provided again a measure of accuracy in reaching the target depth and transparency of the system.

Fig. 4a shows the average penetrations, beyond the stiff constraint, for the feedback modalities $H_{CO_{\alpha,\beta}}$ with respect to the (over-)actuation of the cutaneous channel, $\alpha + \beta$. The lines indicate six different levels of kinesthetic actuation. Fig. 4b shows the average penetrations for the same feedback modalities, but with respect to the actuation of the kinesthetic channel. The four lines take into account different levels of (over-)actuation of the cutaneous one.

The collected data of each modality passed the D’Agostino-Pearson omnibus K2 normality test. In order to determine whether the performances observed can be considered as statistically equivalent to our control condition $H_1$, i.e. when all force was fed back through the Omega device, we performed a two one-sided t-test (TOST). The null hypothesis of the TOST states that the mean values of two groups are different by a certain amount $\theta$ (or larger). Then, in order to test for equivalence, the 90% confidence intervals for the difference between the two groups are evaluated. The null hypothesis that the groups differ by at least $\theta$ is rejected if the limits of the interval fall outside the $\pm \theta$ bounds. Conversely, comparability is demonstrated when the bounds of the 90% confidence interval of the mean difference fall entirely within the $\pm \theta$ bounds [Chen et al. 2010; Limentani et al. 2005]. The design of equivalence tests can be quite tricky due to the fact that the acceptance criterion $\theta$ has to be defined on the basis
of prior knowledge of the measurement. For a sample data set of $n$ independent measurements with standard deviation $s$, for instance, $\theta$ must be for sure greater than $s/\sqrt{n}$, otherwise the test may fail simply because of imprecision, rather than because of a true difference. However, it must also be less than any specifications or standards that the testing is challenging, or the test becomes too easy and will not thus adequately discriminate. In this work we evaluated $\theta$ as suggested in [Limentani et al. 2005], where the authors provided a useful step-by-step process for performing equivalence testing with commonly available computational software packages.

A two one-sided t-test was then performed between $H_1$ and each $HCO_{\alpha,\beta}$ condition shown in Table II. In order to avoid raising the family-wise error rate, i.e. the probability of at least one incorrectly rejected null in a family of tests, the simple correction discussed in [Lauzon and Caffo 2009] has been taken into account. The tests revealed statistical equivalence between $H_1$ and $H_{0.7,0.9}$, $H_{0.8,0.8}$, $H_{0.8,0.6}$, and $H_{0.9,0.3}$ (depicted as filled markers in Fig. 4a). Moreover, we also tested the means among the actuation of kinesthetic and cutaneous channels, i.e. $\alpha$ and $\alpha + \gamma$, using a two-way ANOVA. The means differed significantly among the actuation of both the kinesthetic ($F_{1,360} = 104.25$, $p < 0.001$) and cutaneous channel ($F_{1,360} = 7.605$, $p < 0.001$). Also the interaction between the factors was found significant ($F_{1,360} = 2.210$, $p = 0.006$). Post-hoc analysis (pair-wise post-hoc t-tests corrected according to Bonferroni) revealed statistically significant difference for all the conditions but for $\alpha = 1.4$ vs $\alpha = 0$. This may be due to the limited actuation capabilities of the cutaneous device employed (see also Sec. 4).

The results of this experiment indicate that the subjects, while over-actuating the cutaneous channel, reached smaller average penetration in the stiff constraint (better performance) in comparison to that obtained by a (fully) actuated cutaneous channel ($\alpha + \beta = 1$). Moreover, this over-actuation approach lead, in some of the considered modalities, to performance comparable to that obtained while employing the force from the grounded haptic interface only ($\alpha = 1$ and $\beta = 0$, i.e. our control condition $H_1$). Results therefore indicate that, to a certain extent, it is possible to compensate a lack of kinesthesia with the over-actuation of the cutaneous channel, expecting no significant performance degradation. However, it is worth noting that the performance of the proposed approach is directly related to the effectiveness of the cutaneous devices employed. The research on the development of such devices is therefore of paramount importance for further development of this work.

3.2.1 A more general law. Until this point we have analysed the performance of the system while over-actuating the cutaneous channel at certain levels, i.e. $\alpha + \beta = 1.2, 1.4, 1.6$. However, it may be useful to find a more general law to indicate the level of cutaneous stimuli needed to compensate for any given reduction of kinesthesia. From data shown in Fig. 4 it is indeed possible to evaluate a polynomial fitting of performance vs actuation of cutaneous and kinesthetic channels, and then find the intersection between this curve and the reference value $H_1$. From that, we can evaluate the function

$$\beta = g(\alpha),$$

which gives the amount of cutaneous force to be provided through the cutaneous interface, $\beta$, to fully compensate for a given lack of kinesthesia. For example, in the case where kinesthesia is scaled down to its 90%, $g(0.9)$ will provide the amount of force to be applied through the cutaneous device.

However, it is important to notice that it is not possible to compensate for any lack of kinesthetic force through this technique. This is mainly due to the limited capability of the cutaneous channel in conveying sufficient information and to the technological limitations of the cutaneous devices employed. In fact, our $g(\alpha)$ is found to be defined only for $1 \geq \alpha \geq \alpha_{\min} \approx 0.75$. Under this range of values, it is not possible to fully recover transparency, but only to mitigate the degradation of performance by conveying as much force as possible through the cutaneous device.

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### Table III: 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.

<table>
<thead>
<tr>
<th>Questions</th>
<th>Mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Q1 It has been difficult to wear and use the cutaneous device.</td>
<td>1.69</td>
<td>0.79</td>
</tr>
<tr>
<td>Q2 It has been easy to use the end-effector of the Omega 3 together with the cutaneous device.</td>
<td>5.93</td>
<td>1.06</td>
</tr>
<tr>
<td>Q3 I felt hampered by the cutaneous device.</td>
<td>2.19</td>
<td>1.28</td>
</tr>
<tr>
<td>Q4 I was well-isolated from external noises.</td>
<td>6.44</td>
<td>0.89</td>
</tr>
<tr>
<td>Q5 I was able to hear the sounds made by the actuators of the cutaneous device.</td>
<td>1.31</td>
<td>0.48</td>
</tr>
<tr>
<td>Q6 It was easy to feel the presence of the stiff constraint.</td>
<td>6.56</td>
<td>0.81</td>
</tr>
<tr>
<td>Q7 Sometimes I did not feel the presence of the stiff constraint.</td>
<td>1.06</td>
<td>0.25</td>
</tr>
<tr>
<td>Q8 I had the feeling of performing better while receiving force feedback by the cutaneous device.</td>
<td>4.94</td>
<td>1.77</td>
</tr>
<tr>
<td>Q9 The force provided by the cutaneous device on the fingertip felt strange.</td>
<td>2.18</td>
<td>1.11</td>
</tr>
<tr>
<td>Q10 The force fed back by the Omega 3 felt similar to the one provided by the cutaneous device.</td>
<td>3.46</td>
<td>2.37</td>
</tr>
<tr>
<td>Q11 I felt the force provided by the cutaneous device only on the fingertip.</td>
<td>5.68</td>
<td>1.08</td>
</tr>
<tr>
<td>Q12 At the end of the experiment I felt tired.</td>
<td>2.06</td>
<td>0.85</td>
</tr>
</tbody>
</table>

3.2.2 Users’ experience evaluation. In addition to the quantitative evaluation of the proposed approach discussed before, we measured user’s experience in using the cutaneous device in the given teleoperation scenario. It was evaluated by a questionnaire, considering three dimensions:

(i) comfort in using the given system,
(ii) users’ evaluation of their own performance,
(iii) preferred modality.

After having performed the experiments described above, participants were asked to fill in a 12-item questionnaire. It contained a set of assertions and was scored according to a 7-point Likert scale, where a score of 7 was described as “completely agree” and a score of 1 as “completely disagree” with the assertion. To avoid acquiescence bias, i.e. the tendency to agree with statements as presented, we designed a scale with balanced keying [Welkenhuysen-Gybels et al. 2003]. The evaluation of each question is reported in Table III.

Results showed that participants felt positive about the proposed approach. Most of them were convinced they were performing better while being provided with force feedback through the cutaneous devices even though, as shown in Sec. 3.1, this was not always the case. Subjects also stated that using the Omega together with the cutaneous device was very easy and that the force provided by the grounded haptic device was somehow similar to that provided by the cutaneous device. As discussed in Sec. 2.2, this is very important when it comes to independently control the cutaneous and kinesthetic channels. Moreover, results showed a reasonably high comfort level while using the proposed cutaneous interface and assessed its overall wearability. This also validates the capability of such device to be used in scenarios where the wearability and portability of the master system is paramount [Prattichizzo et al. 2010a; Chinello et al. 2012].

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 work originated from the observation that stimuli received by a user while holding a haptic handle consist of a cutaneous and a kinesthetic component. Cutaneous feedback provides less realism in the interaction than kinesthesia but it does not affect the stability of the teleoperation
Improving transparency in teleoperation by means of cutaneous tactile force feedback

system, while kinesthetic force provides a realistic illusion of telepresence but it may affect the stability of the haptic loop. However, while employing common haptic interfaces, such as the Omega devices, it is not possible to decouple these components, i.e. it is not possible to control them independently. For this reason, many control techniques ensure a stable interaction by scaling down kinesthetic feedback, which in turn scales down the cutaneous stimuli component as well, even though acting on this cutaneous channel does not affect stability.

In this work we discussed how and to what extent acting on the cutaneous channel could improve the transparency of teleoperation systems. In order to decouple the control of the two components of the haptic interaction, we employed a wearable cutaneous device in conjunction with a grounded haptic device. The cutaneous device provides cutaneous force feedback only and can be easily worn without affecting the ability of interacting with the grounded haptic interface. The combination of cutaneous stimuli, provided by the cutaneous interface, and haptic feedback, provided by the grounded display, made possible, though with some limitations, to independently control the two components of the interaction.

Then we carried out two experiments, comparing performance while providing solely haptic feedback by an Omega interface and while providing both haptic and cutaneous stimuli through the Omega and a wearable cutaneous device. We scaled down the force fed back through the grounded interface, compensating for the lack of kinesthesia with the cutaneous device. Results showed improved performance using the aforementioned compensation technique. Moreover, in some of the considered conditions, over-actuating the cutaneous channel even led to performance comparable to the one registered while using the grounded interface alone.

The main drawback of the work is that the performance of the proposed approach is dependent on the given task and on the effectiveness of the cutaneous interface employed. For this reason, the results presented here cannot be considered valid for any teleoperation system. However, the experimental protocol hereby presented can be easily replicated to assess the right compensation technique for any teleoperation setup. In other words, researchers should always re-evaluate the mapping function $g(·)$, presented in eq. 3, according to their particular task and teleoperation system.

REFERENCES


Improving transparency in teleoperation by means of cutaneous tactile force feedback