Haptics in Teleoperated Medical Interventions: Force Measurement, Haptic Interfaces and Their Influence on User's Performance

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***Abstract*—*Objectives:* Haptics in teleoperated medical interventions enables measurement and transfer of force information to the operator during robot-environment interaction. This paper provides an overview of the current research in this domain and guidelines for future investigations. *Methods:* We review current technologies in force measurement and haptic devices as well as their experimental evaluation and influence on user's performance. *Results:* Force sensing is moving away from the conventional proximal measurement methods to distal sensing and contact-less methods. Wearable devices that deliver haptic feedback on different body parts are increasingly playing an important role. Performance and accuracy improvement are the widely reported benefits of haptic feedback, while there is a debate on its effect on task completion time and exerted force. *Conclusion:* With the surge of new ideas, there is a need for better and more systematic validation of the new sensing and feedback technology, through better user studies and novel methods like validated benchmarks and new taxonomies. *Significance:* This review investigates haptics from sensing to interfaces within the context of user's performance and the validation procedures to highlight salient advances. It provides guidelines to future developments and highlights the shortcomings in the field.**

***Index Terms*—Haptics, Human-robot interaction, Force and tactile feedback, Medical interventions, Performance, Sensing, User interface.**

I. Introduction

As ROBOTS are deployed in challenging industrial, assistive and medical environments, controlling them remotely becomes increasingly important. Teleoperation of robotic platforms plays an important role in i) safety of the operators working in hazardous environments e.g., rescue robots in natural disasters, ii) ensuring ergonomy in Minimally Invasive Surgery (MIS) and iii) providing expert assistance in remote environments otherwise deprived of adequate human resources. A teleoperated robotic platform consists of a (remote) command

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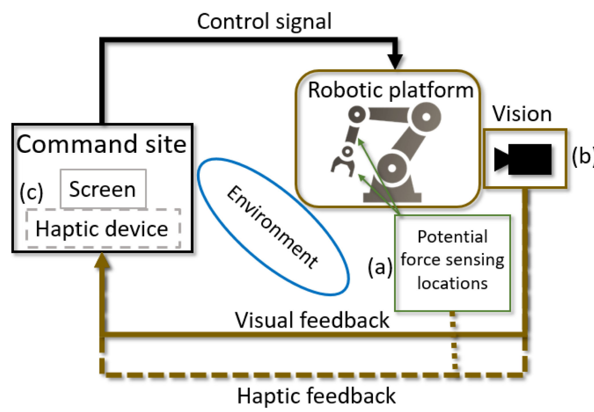

Fig. 1. Main elements in a teleoperated robotic platform.

site, where the operator receives feedback from the robot and generates control commands, and the interaction site, comprising a robot, which interacts with the environment and responds to the commands (Fig. 1). The operator can send commands to the robot through various methods such as joysticks, keyboard and mouse, head movements, voice commands, etc. At the same time, they receive information about what happens in the remote site through visual, audio or haptic feedback.

Early teleoperation systems relied primarily on visual feedback to the operator. However, this usually resulted in high cognitive load for the operator as they had to rely only on visual feedback from the interaction site and did not receive any haptic feedback from the robot-environment interaction [1]. To compensate for this limitation, starting from mid 1990s development of haptic interfaces began first as joysticks in the gaming industry [2] and later on as devices that could deliver either force or tactile feedback but not both [3], [4]. In the following years, haptics in surgical virtual simulators [5] and MIS [6], [7], as well as stability of the interaction [8] started to gain researchers' attention. In recent years, closing the loop by haptic feedback has gained considerable attention especially in medical applications [9]–[11]. It has been reported that haptic feedback reduces errors in remote operations and decreases damage to the environment as well as the operation time [12]. Providing haptic feedback to the surgeon in teleoperated medical applications is particularly important to protect delicate soft internal body tissues from excessive application of force. In addition, haptics can potentially improve the navigation efficiency of surgical instruments inside the human body, by guiding the operator to keep the instrument on the right path and avoid tool-tissue collisions.

A system which provides some type of haptic feedback to the user comprises two main parts: i) a sensing or estimation component for characterizing end effector-environment interaction on the manipulator side, ii) a haptic device on the master side to transfer the gathered information to the user. Each of these parts have received extensive attention in the literature [13]–[16]. Various efforts have been undertaken to characterize the end effector-environment interaction as accurately as possible and to feed it back to the user in the most natural and informative way. Despite the advances in this field, including the development of more accurate force sensing technologies and more user-friendly haptic feedback devices, the majority of existing commercial systems do not yet incorporate haptic feedback. Haptic feedback is still not widely adopted by users, as current systems do not provide sufficient sensing and feedback accuracy, realism and comfort for the user to improve performance while maintaining system stability. In particular, interaction force sensing and estimation at the tool-tip, and the design and selection of the most effective types of feedback for different applications, are still key topics under investigation.

There are a number of reviews of haptics in the literature. Previous surveys have either focused on a specific area such as the effect of haptic feedback on performance [17]–[19], related technologies for sensing [20] and providing feedback [12], or they treat haptics as a part of a global survey on general robotic surgery [21], [22] with emphasis on the results rather than the experimental paradigms and methodologies.

This review paper covers the literature on providing haptic information to the operator in *tele-operated medical interventions* from the year 2000 onward with the goal of highlighting the salient advances, most dominant research trends and potential future paths. It presents a comprehensive overview of tool-tissue interaction force measurement methods and novel haptic interfaces as well as the effect of haptic feedback on user performance. The resources for this literature review were Web of Science and Scopus. The search was carried out with the following keywords: "haptics" plus "medical robotics", "surgical robotics", "safety", "telemanipulation", "teleoperation", "perception", "sensor" and "force measurement". Articles were assessed based on their relevance to the targeted overview, their contribution to the field and their novelty. We provide an overview of the most common approaches to experimental studies, the dominant experimental platforms and methods as well as suggestions and recommendations for future research and development in the field. A summary of articles that included the word haptics but focused on industrial or other applications is presented in Section V, Discussion, to highlight the research overlaps as well as innovations potentially useful in medical interventions.

The rest of the paper is organized as follows: In Section II, the past and ongoing research on measurement of tool-tissue interaction forces, including force sensing and vision-based methods are presented (Fig. 1(a) and (b)). In Section III, the commercially available and novel haptic interfaces in tele-operation are introduced and compared (Fig. 1(c)). The effect of haptic feedback on performance is discussed in Section IV through an overview of the experimental comparative studies among different types of feedback. The paper concludes with a discussion on the achievements in sensing and feedback in surgical teleoperation to date and suggestions for future research, followed by the Conclusion Section.

II. End Effector-Tissue Interaction Force Measurement

The most popular type of haptic feedback in tele-operated medical tasks is force feedback. In this paradigm, the objective is to measure the tool-tissue interaction forces accurately on the slave side and feed them back to the user in real time with the same intensity as measured. However, the millimeter-scale size of medical instruments, the complexity of the tasks, nonlinear and variable characteristics of soft tissues, sterilization considerations and difficulty in accessing remote areas of the body reduce the measurement accuracy of the interaction forces.

Efforts towards precise measurement of interaction forces in medical applications can be divided into three main categories: sensor-based methods, contact-less methods and force estimation (Fig. 2). Each of these approaches and the related research will be detailed in the following.

A. Sensor-Based Methods

Sensor-based force measurement methods in tool-tissue interaction can be divided into two categories: proximal sensing away from the tool tip and distal sensing close to the contact point. Proximal sensing is achieved mainly through placement of strain gauges and optical sensors at the tool shaft or adding torque sensors near the proximal pulleys. The more challenging approach of distal sensing is addressed through integration of miniature pressure, infra-red or optical sensors at the tool tip. The main contributions are detailed in the following.

1) Proximal Sensing: Remote force sensing was the first approach proposed to measure tool-tissue interaction forces [23]. In this method, the sensors are placed on the proximal sections of the device or integrated into the trocar to avoid size and sterilization limitations imposed by the millimeter-scale tool tip of surgical instruments inserted into the body.

Moving away from the tool tip, the shaft of the instrument is the next closest part to the tool tip-tissue contact point. Many of the surgical laparoscopic and robotic instruments use cable actuation mechanisms passing through the tool shaft. Multiple studies have measured interaction forces using strain gauges placed on the tool shaft and its actuating cables. For example, a conventional laparoscopic tool was redesigned to incorporate sensors for force measurement [24]. A resistive sensor was integrated in one of the jaws of the surgical grasper to measure the normal forces, while four strain gauges were placed around the flexible part of the tool shaft to measure the manipulation forces in horizontal and vertical directions. The tool was held by a Mitsubishi PA-10 robotic arm. The performance of the force measurement system was evaluated in tissue characterization and knot tightening. Preliminary results show the ability of the

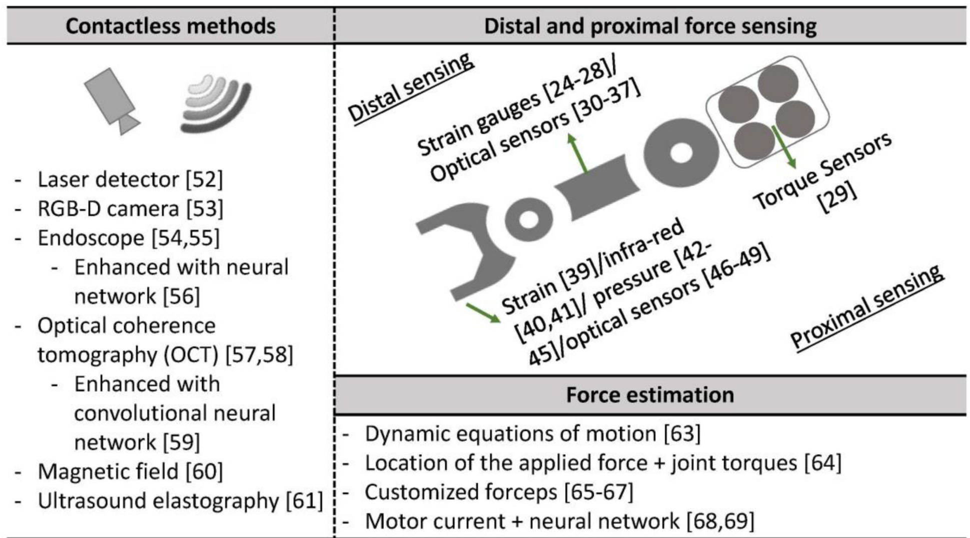

Fig. 2. End effector-tissue interaction force measurement methods discussed in Section II. Distal and proximal sensing, contact-less methods, and force estimation.

grasper to identify soft, medium and hard samples as well as performance improvement in knot tightening. In 2010, another team enhanced a cable-actuated laparoscopic instrument by mounting six strain gauges on three of the actuation cables in the tool shaft [25]. Experimental evaluation was conducted using a seven Degrees Of Freedom (DOF) (modified version of a 5-DOF) Quanser Haptic Wand as the master, while the three DOF surgical tool was mounted on a seven DOF Mitsubishi PA107C robotic arm. Results show a good correspondence between measurements from strain gauges and those from a sensor positioned below the setup. Another example of proximal force measurement in MIS is a test bed made with two seven DOF Quanser Haptic Wands as the master and two Mitsubishi PA10-7C robotic arms as the slave devices [26]. The surgical cable-actuated tool, namely, a da Vinci needle driver, is instrumented using six strain gauges mounted and rigidly glued to three pairs of the cables to capture interaction forces in 3 DOF. In addition, a six DOF ATI Industrial Automation (ATI) force sensor was placed away from the tip behind the surgical tool to measure exerted forces in the other DOF. Recently, [27] proposed an external force estimation strategy based on the cable tension changes in a micromanipulator. The authors of [28] introduced a force sensor with four strain gauges and a measurement error of less than 0.01 N. The resolution and range of this sensor can be customized using a range of materials.

An alternative approach to force sensing on the tool shaft is placement of torque sensors near the proximal pulleys in cable actuated laparoscopic tools. In 2015, the commercially available da Vinci EndoWrist needle driver tool was equipped with four torque sensors placed near the pulleys at the proximal end of the tool [29]. The authors showed that unprocessed torque data could detect the tool-tissue contact at different rotational velocities and shaft loads. However, due to noise and losses tissue contact can be identified only for torques larger than 5 mN/m.

A bio-inspired approach for tactile sensing in flexible surgical tools is proposed in [30]. A flexible sleeve with scattered optic fibers acquires tactile information from the tool-environment interaction. Other solutions for three-axis force sensing in a flexible surgical manipulator are proposed using a ring-shape device augmented with photo interrupters [31] or optic fibers [32]–[34] and integrated into the flexible shaft of the device. The fiber optics technology has a maximum error of $\sim$18% [35]. While this sensor is immune from magnetic and electrical disturbances, a recalibration is required if it is removed and reattached to the device.

For retinal microsurgery, [36] presents a 3 DOF force sensor based on Fiber Bragg Grating (FBG) sensors. The surgical tool includes the new sensor placed right behind its grasper. The sensor can measure axial and transverse forces of up to 21 mN with 3.3 mN and 0.5 mN accuracy, respectively. The same concept has been deployed in [37] for proximal force sensing in flexible endoscopic robots.

A different proximal sensing approach has been presented in [38], where four optical sensors are integrated into the trocar rather than the surgical instrument for measuring interaction forces on the orthogonal plane of the device. The measurement error is less than 12%.

While proximal force measurement facilitates sensor placement, it suffers from several limitations. Although the size factor is less crucial in this method compared to distal force measurement, limits related to the size of the shaft and proximal parts of the tool are still constraining. Also, the addition of high quality small and accurate sensors to conventional tools increases the final price of the instrument.

In summary, force sensing on the tool shaft using strain gauges has been the most common remote force sensing method, and the Mitsubishi robotic arm is widely used as the slave robot holding the surgical instrument. The main advantage of remote force measurement is wider choice of sensors. Being away from the tool tip poses less restrict limitations on the sensor size as well as sterilization requirements. However, due to friction and structural properties of surgical tools, the accuracy of this measurement method is not sufficient in many surgical scenarios. Further, depending on the selected sensors it can be expensive. Finally, the sterilization remains an issue if sensors are placed on the exposed parts of the shaft that enters the abdomen.

2) Distal Sensing: For the precise measurement of the tool-tissue interaction forces, ideally forces should be sensed directly at the tool tip. However, the tips of most laparoscopic and robotic surgical instruments are in the millimeter scale confining the size of any sensor to be placed there. Sterilization of the sensor and its cost are the other two main limitations in the design of tool tip force sensors. Recent efforts have started to overcome these obstacles.

In 2014, a research team at Harvard University produced a three axis force sensor using rapid prototyping and printed-circuit Micro-Electromechanical Systems (MEMS) techniques with potential application in MIS [39]. The sensor size was ten by ten millimeters with 5 N/V sensitivity and sensing range of –500 to 500 mN in x and y axes, and -2.5 to 2.5 N in z axis with $\sim$mN resolution at 20 Hz frequency. Foil-based strain sensors were fabricated in parallel with the mechanical structure, eliminating the need for post-manufacturing assembly. The performance of the sensor was experimentally validated in a simulated tissue palpation task with an in-house built robotic master-slave platform. Results showed that the sensor can detect forces, but, its sensing range is not large enough to cover all the forces that potentially can be applied on the tissue. In 2016, the same team developed a millimeter-scale uniaxial force sensor which can detect small reaction forces at the distal end of a flexible instrument [40]. Their development is especially beneficial for flexible instruments used in robotic surgery to access remote areas of the body. The sensor is fabricated using a composite lamination process. The linkages are pre-machined in the 2D laminate and provide the necessary DOF and fold patterns for self-assembly of the 3D sensor. An infra-red emitter and a receiver are mounted on the sensor. The force is measured through light intensity modulation. The sensor measurement range is 0–200 mN with 0.8 mN resolution and 5 V/N sensitivity. Experimental studies show that the biologically relevant forces can be detected. In a follow on study, they developed a three axis force sensor with integrated temperature and ambient irradiance sensing [41]. Through experimental studies they have shown that the error in force measurement is reduced in their origami-inspired force sensor.

Another research team at Sungkyunkwan University developed a force sensor small enough to fit in the tip of a laparoscopic grasper with approximate dimensions of $4 \times 2 \times 2$ mm^3 [42]. The sensor could measure dual axial forces at the tip of an MIS surgical tool. By implanting one sensor on each jaw of the gripper, three axial tension and single axial grasping forces were measured. Both sensors were capacitive pressure sensor cells based on elastomeric polymer, capable of providing information about normal and shear forces. The final measured values were the result of comparison between the measurements of the two sensors. The sensor was next integrated on the Raven II surgical platform and its performance was experimentally validated [43], [44]. As the sensor was placed at the top part of the tip of the grasper, only forces at this location were measured, however, for tasks like palpation, sensing at the end of the gripper is also required. Due to the limitations of the previous design, in the next trial the sensors were moved further from the tip of the surgical grasper [16]. A miniaturized force sensor was placed at the tool wrist and two torque sensors were planted at the tool base. Using these sensors, three axis Cartesian forces and single axis grasping force were measured. The sensors' performance was validated both individually and as an integrated tool in Raven II. There were some discrepancies in sensing due to the backlash in tendon-driven systems. In their latest work, they sensorized the proximal region of both jaws of the gripper [45]. The two compact three DOF force sensors are capable of measuring the grasping force, 3 DOF manipulation force and the rotational torque. The force range was ± 5 N with a resolution of 3.8, 1.8 and 2.0 mN in 3 DOF and average relative errors of 2.6%, 2.2% and 1.3% of the full-scale output force. Experimental evaluation was conducted via installation of the sensorized instrument on the S-Surge surgical robot and comparison of the measured values with those of a commercial sensor.

An alternative approach is the optical-based sensing technology deployed in a fingertip force sensor [46], [47]. The applied force on the fingertip is measured through projecting and receiving a light beam using two optic fibers closely located on the tip. The sensor can measure 0–4 N with a maximum relative error of 3.1097% [48]. However, this technology has not been tested for distal sensing in tool-tissue interaction. In another attempt, an FBG-based force sensor is introduced to measure interaction forces on the three fingers of an underactuated surgical gripper [49]. The measurement error is less than 5% compared to an ATI sensor. However, the instrument diameter is 15 mm, still large for use in the da Vinci surgical platform as an example of a commonly used setup.

To provide the surgeon with palpation information, as one of the most important tool-tissue interaction scenarios, an inflatable pneumatic balloon which can be placed in a cavity on the instrument shaft close to the end effector is introduced in [50]. Experimental results show the mean of the normalized errors compared to an ATI sensor is 9.81%.

In spite of the advances in development of small sized sensors for tool tip force measurement, further research is required to downsize the current prototypes and improve their sensing range.

B. Contact-Less Methods

The limited application of force sensors due to packaging complexity, size, accuracy and sensing range has motivated research on contact-less vision and range-based methods for force measurement and estimation of tissue characteristics, as well as navigation of medical instruments inside the human body. A survey in 2010 emphasized the importance of further research on visually-sensed deformation as a promising research area for future developments in force estimation for MIS [51]. It argues that if tissue models could be developed during the operation, the displayed force could be model-based rather than estimated from contact sensing.

Various approaches have been introduced in contact-less force measurement ranging from using image data of an endoscope or a Red Green Blue-Depth (RGB-D) camera to laser detectors and optic fibers as well as ultrasound elastography and Optical Coherence Tomography (OCT). Neural networks have been deployed in enhancement of some of these methods. The following paragraphs detail highlights of the advancements in the field.

One of the first investigations on contact-less sensing was conducted in 2005 to measure tool-tissue distance and tissue stiffness in surgical laser cutting [52]. They used the power absorption of the surface to calculate the surface stiffness using a laser detector. The information about stiffness was fed back to the user as vibrations in a Phantom Omni.

Vision-based estimation of the exerted force on a tissue is achieved mainly through processing captured images with a camera and combining the gathered information with a mathematical model of the organ/tissue. For example, [53] used a RGB-D camera to develop a deformation model including the Young's modulus and Poisson's ratio of soft materials. The exerted point force is then estimated based on the model and a registration technique to measure deformation in natural room light. In [54] image processing and photoelasticity were proposed to measure morphological changes during catheter insertion procedures. The deformation of the artery model was measured and compared with the displacement of the catheter. It is then fed back to the user as a force. A demonstration with one participant and the Phantom Omni as the master device was conducted. The paper showed successful measurement of deformations. A year later, interaction forces during MIS were estimated from endoscopic monocular images [55]. In this study, the new concept of the virtual template was introduced, in which the object's surface deformation is modelled without the knowledge of the undeformed shape. An in vitro experimental study on lamb liver was conducted with a surgical tool attached to the PUMA robot. In this method, only a smoothness assumption is sufficient with no need for the exact template of the organ. Also, an initial and a target image are sufficient for estimating the deformation with no need for sequential images. However, similar to many other machine vision methods, this method relies on the correct feature matching process. Also, the experimental study was conducted in natural room light, different from the lighting conditions in laparoscopic surgery.

In 2017, a tool-tissue interaction force measurement system was developed through visual information and sequential images [56]. A recurrent deep neural network-based model was proposed to identify the change in object shape. The network was trained with 600 images. Experiments were conducted on a sponge, plastic bottle, human arm and a tube. The measurements were similar to those obtained from a load cell as the bench force sensor. An average mean squared error of 0.015 N is reported, but the force was measured in one direction only and the method is not suitable for small deformations. Also, the light and camera conditions were not similar to the endoscope view in laparoscopic and robotic surgery.

Later on in 2018, a method was introduced in which the undeformed reference volume and a deformed sample volume of the organ, captured using OCT, were used as input [57]. A novel Siamese 3D Convolutional Neural Network (CNN) architecture was deployed. Using OCT, capturing of the inner structure of the tissue becomes possible. This improves the performance compared to capturing surface deformations only. In this method, the three components of the force vector can be calculated. Experiments were conducted on silicon phantom tissue and porcine tissue. This method is only suitable for surgical procedures involving intraoperative OCT. A similar approach has been deployed in other projects [58]. A semi-supervised deep neural network model was used in [59] for image-based force sensing in robotic surgery. Further research is required to reduce the force estimation errors.

A wireless palpation technique using magnetic fields is presented in [60]. A small probe with a permanent magnet is inserted into the body at the location of interest, while a robotic arm holds a second permanent magnet outside of the body. Experiments in one DOF provided comparable results with traditional indentation techniques. In vivo tests are required to investigate the effects of respiration and vibration on the outcome.

A review on ultrasound elastography indicates that it is currently used for tissue characterization in abdominal organs [61]. Ultrasound elastography can be potentially deployed to estimate force by measuring deformations and using the material-specific force-deformation relationship.

In addition to contact force estimation, cameras can be used for aiding surgical tool navigation inside the body. In 2016, a system was developed to assist navigation of an endoscopic device inside the human colon [62]. The endoscopy is conducted through guiding a magnetically dragged capsule inside the intestine. A Storz GmbH camera was attached to the capsule. A vision navigation module generated local 3D maps and a navigation trajectory from 2D capsule images. The haptic module transformed the map and trajectory information into a guiding virtual force to assist the operator in navigation. Forces were computed using a virtual spring model, as a function of spring stiffness and distance from the desired path. Ex-vivo experimental studies were conducted with the RV-6SDL, Mitsubishi Electric robotic arm as the master robot and Phantom Omni as the slave device.

In summary, contact-less force measurement in medical operations has been demonstrated in specific lighting conditions or with particular imaging facilities. Further improvements are required to make this method suitable in real-life for any type of surgery, endoscopic camera, lighting conditions, and imaging capabilities, with no prior customized model of the abdominal organs and in real-time.

C. Force Estimation

To eliminate the need for in-body sensors, an alternative to sensor-based methods is force estimation. This can be achieved based on known robot and interaction dynamic models, location of the applied force and joint torques, motor currents or output of encoders. Each of these approaches is detailed in the following.

In an effort to overcome the limitations in the sensor size and cost, in 2012, a research team proposed force estimation to replace force measurement in bilateral teleoperation [63]. In this study, the applied force on the slave side was estimated based on the known dynamic equations of the system. The Phantom Premium was used as both the master and slave devices.

Nano25 and Nano17 force sensors were used to provide a bench measurement to validate the estimations. The authors concluded that estimating forces rather than measuring them may result in singularity and system instability at certain points. In another research, it has been shown that the geometric knowledge of the applied force contact point plus the joint torques are sufficient to estimate the exerted force on a robotic arm with less than 5% error [64]. The method is based on the dynamic model of a rigid robot:

$$\boldsymbol{M}(\boldsymbol{q})\,\ddot{\boldsymbol{q}} + \boldsymbol{C}(\boldsymbol{q},\dot{\boldsymbol{q}})\,\dot{\boldsymbol{q}} + \boldsymbol{g}(\boldsymbol{q}) = \boldsymbol{\tau} + \boldsymbol{\tau}_{ext}$$

where $\boldsymbol{q}$ is the generalized coordinate, $\boldsymbol{M}$ is the symmetric, positive definite inertia matrix, $\boldsymbol{C}\dot{\boldsymbol{q}}$ includes the Coriolis and centrifugal terms, $\boldsymbol{g}$ is the gravity term, τ is the control torque, and $\boldsymbol{\tau}_{ext}$ is the torque due to external generalized contact forces acting on the robot. It has been proven to be successful in laboratory conditions, however, locating the contact point in scenarios like tool-tissue interaction in robotic surgery would be a challenging task.

Another effort in indirect measurement of force is a master-slave haptic forceps developed to measure the elasticity of living organs [65]. The grip on the master side permitted grasping of objects through the slave-side forceps. Both master and slave sides included a linear motor and an optical sensor. The stiffness of the materials was calculated using the formula “delta force/displacement” in N/m. Successful performance of the device was illustrated in studies on live rats. In [66] the authors suggest the use of pneumatic actuators and neural network for force estimation in surgical robotic platforms. Simulations provide accurate results; however, the accuracy suffers in the presence of abrupt external force. A simpler approach using a strain gauge in a pneumatically actuated robotic forceps is suggested in [67]. The grasping force is estimated by combining the internal pressure of the actuator and the strain gauge data using the following equation:

$$F = PA - K\,(v_i/\alpha)$$

where F is the external force, P is the air pressure, A is the cross-section of the actuator and K is the spring constant of the actuator, v_i is the output voltage and α is the constant of the strain gauge. The accuracy of the measurements were confirmed in laboratory tests.

A recent study suggests that although the previously developed customized sensors have successfully measured the grip force in surgical tools, the same level of accuracy is achievable through the existing sensors in surgical robots [68]. A da Vinci surgical tool sensitized on the proximal side for ground truth is used in the experimental studies. The results show that the commanded motor current and measured motor current provide accurate force estimates. The root mean square error for the distal-end torque was 4.42 *m*Nm. In a similar study, [69] used motor current and neural network to estimate interaction forces in robot-assisted MIS. Results are promising with a maximum mean error of 0.19 ± 0.18 N with applied forces ranging between 0–20 N, but the execution time should be improved for online force estimation.

The tool-tissue interaction force sensing in medical applications requires further effort to provide precise measurements addressing sterilization and cost issues. The focus of various proposed solutions is mainly on the development of technology rather than evaluation of its performance. Many studies lack quantitative sensor accuracy estimation and benchmarking, important in performance evaluation and comparison among multiple solutions. In addition, further investigation is required for determining the desired accuracy and bandwidth in these sensors to support haptic applications, which might be dependent on the type of surgery among other factors.

III. Haptic Interfaces

Providing haptic feedback during tele-operation requires an interface to transfer this information to the operator. Ideally, a haptic feedback device should deliver realistic haptic sensation through a durable, low inertia and comfortable user interface. The resolution and bandwidth should be sufficient to retain the gathered information about tool-environment interaction.

Various commercially available haptic devices have been used in different experimental studies as the master device for tele-operation. In [70] researchers conducted a comparative experimental study with two surgeons and seven engineers to evaluate the performance of three commercially available haptic hand-controllers: Sigma 7, HD2 and Phantom Premium in two peg-in-hole tasks. The KUKA robotic arm was used as the slave manipulator. The objective measures of effort, speed, accuracy, performance improvement and applied force were used in this evaluation. The authors concluded that Phantom Premium has the best performance, mainly due to its similar mechanical structure to the human arm. Another study compared the Phantom device with a game joystick in manipulating Computed Tomography (CT) based rendering of femur fracture in a virtual environment [11]. An experimental study with 20 participants was carried out. Results showed that accuracy is almost the same while the joystick has an easier learning curve. Accounting for the higher price of phantom compared to a game joystick, its use is not justified in this application.

The commercially available haptic devices are mainly desktop-based and dedicated to providing the user with force feedback. These are mostly designed to meet general needs in tool-environment interaction and are not tailored for specific applications. To compensate for this, various devices have been introduced in the literature for targeted tasks and capable of generating force/tactile feedback or both (Fig. 3). In this section, they are categorized into devices designed for force feedback and those dedicated to vibrotactile feedback.

A. Force Feedback

Research into haptic interfaces has been active for over 20 years with developments in general purpose force feedback. A three DOF planar pantograph haptic interface was one of the first developments for haptic interaction with a virtual environment as a means of implementing varying slave behavior [71]. In 2003, a novel universal force controlled manipulator was introduced [72]. The main body of the robot had three DOF. Depending

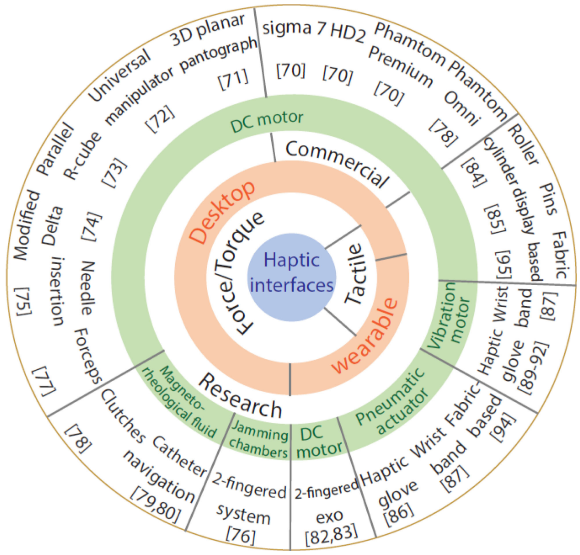

Fig. 3. Haptic interfaces categorized.

on the application, various end effectors could be attached to the device. To compare this development with industrial haptic devices, three haptic performance indicators were defined: haptic resolution, force depth, and impedance ratio. The new system had a better performance compared to industrial devices. However, the proposed three DOF is not sufficient for many applications.

To improve precision in teleoperation, a novel desktop device was developed for force feedback [73]. The design of this parallel mechanism was based on twisted R-Cube to achieve a larger workspace. The performance of the device was not tested experimentally. As another example of a parallel mechanism, a seven DOF haptic interface based on the modified delta mechanism was developed recently [74]. The device could deliver up to 6 N of force along each axis.

A number of interfaces have been designed for specific medical interventions. Many of these devices are actuated via Direct Current (DC) motors. In 2013, a force feedback system was proposed to detect soft tissue layers while inserting a needle into the tissue [75]. In this robotic coaxial needle insertion assistant, the tip and shaft forces (friction while inserting the needle) were separated from each other through two coaxial inner and outer needles. The inner needle only experienced the forces on the tip and the shaft frictional force was experienced by the outer needle. The force was fed back to the user handle through a DC motor. The performance of the system was evaluated in an experimental study with eleven participants on an artificial brain vessel tissue. Results indicate that providing the user with only tip force feedback increases the rate of success. Another example is a haptic device developed for sensing palpation in teleoperation using a pneumatic and a granular jamming actuator [76]. The setup comprised of a hemispherical indenter driven with a Maxon motor (EC-30). An ATI force sensor was used with a resolution of 0.003 N. In [77] novel forceps were designed and developed to be mounted on a hand controller on the master side. The forceps could provide force feedback for grasping of a blood vessel or hard objects.

Some recent developments have moved away from DC motors as actuators. In 2016, a two DOF haptic interface was developed using Magneto-Rheological Fluid (MRF)-based clutches for medical percutaneous intervention and soft tissue palpation [78]. The performance of the developed device was compared with Phantom Omni and Quanser Haptic Wand in experimental studies on a phantom tissue. The applied forces were measured using an ATI Nano43 sensor attached to the end effector. A five DOF in-house robot was used as the slave. In the same year, another MRF-based haptic interface for navigation of catheter in surgical operations was developed [79], [80]. The catheter went through a container filled with MRF. The viscosity of the fluid was adjustable by energizing the electromagnetic coils. In 2018, an experimental study was conducted in Virtual Reality (VR) on the slave side with ten participants [81]. The catheter was navigated through a virtual cerebral vessel by the developed haptic interface. The method was shown to be beneficial in avoiding collisions and improving safety of endovascular tele-surgery.

Recently, a two-fingered wearable exoskeleton is developed to provide force feedback on thumb and index fingers using linear actuators and a cable system. The application of the device has been demonstrated in interaction with virtual environment [82] as well as in robot-assisted MIS [83].

Development of task specific haptic devices is an increasingly popular practice. From needle insertion and palpation to catheter navigation, the new solutions move towards platforms with similar user interfaces as conventional scenarios. Price, durability, bandwidth and resolution are among the factors that require further improvement in future developments.

B. Tactile Feedback

Many developments have focused on delivering tactile feedback. These devices can be divided into three main categories: contact-based, wearable and fabric-based with more publications in the two later categories in recent years.

An example of a contact based device is a one DOF new tactile contact location display system [84]. This thimble-sized display was designed to be mounted on a Phantom haptic force feedback device. A small cylinder suspended beneath the fingertip could display sliding or rolling contact. Experimental studies in VR showed that participants could determine the curvature of objects. In 2016, a novel device was developed to generate the pre-slide sensation in tele-operation and robotic surgery to prevent slippage of soft objects from the robotic gripers on the slave side [85]. The haptic interface was made up of multiple pins with a single actuator. The pre-slide sensation was generated through vertical and tangential pin movements. No experimental studies with humans have been conducted yet.

Multiple research projects have focused on wearable devices mainly in the form of a glove or a wristband with pneumatic actuators and/or vibration motors. These actuators have been used to deliver force or torque on user's hand [86] or wrist [87] in exploration and path following tasks. The maximum force delivered on fingers is reported to be 16 N. Users' performance was reported to improve with the haptic feedback [88]. In 2017, a wearable haptic glove was developed to provide kinesthetic and cutaneous feedback to the user with a combination of pneumatic and vibrotactile actuators [89]. The final prototype was able to resist forces up to 7 N with 5 mm displacement with a considerable time lag of 0.5 s before reaching its maximum stiffness. Vibrotactile feedback on the hand and wrist is addressed in multiple studies mainly to provide haptic information about the interaction of a tele-operated robotic hand with environment. Feedback is provided on four [90], [91] or six zones [92] on the hand or eight zones [87] on the wrist. Results back these developments through superior performance of the users.

Fabric-based actuators have gained more attention in the past few years. These are a cost effective solution with the potential of "transparent" future developments through which the wearer is not aware of the device, similar to wearing clothing [93]. In 2016, a fabric-based tactile actuator and sensor system with potential use in haptic gloves was developed [94]. The actuator was made of multi-layers of paper and fabric and used pneumatic actuation. It could insert forces up to 2.20 ± 0.017 N. The sensor was made of layers of piezoresistive fabric and conductive fabric. In 2018, a wearable haptic device was proposed to provide variable degrees of softness [95]. In this study, a wearable fabric yielding device was introduced that could deliver adjustable stiffness using two DC motors that stretched the fabric to make it feel stiffer. The performance of the device was evaluated through an experimental study with twelve participants. Future work on wearability and miniaturization of the system is suggested. To provide an objective measure of comparison among various cutaneous displays, [96] suggested measuring the space of tactile sensation in a specific task namely pinching palpation of four materials with different stiffness. They concluded that producing the sensation of hard material is more difficult compared to soft material mainly due to weak servo motors.

In summary, force feedback devices are designed mainly for palpation, needle insertion, and instrument navigation. The conventional devices that rely on motors for providing force feedback are well established and used in various applications. On the other hand, the newly developed feedback strategies such as MRF-based devices still are under development and require further improvements. Unlike established application of force feedback in medical tasks, tactile feedback has a larger set of targeted applications. These range from surgery and navigation to teleoperation, exploration and human-robot interaction. In addition, many of the tactile feedback interfaces are wearable gloves and wrist bands as opposed to the common standalone force feedback devices. Finally, fabric-based interfaces are suitable for integration in wearable devices due to their flexible nature. They are mainly used for providing feedback about the stiffness of objects. Although they can apply forces, there is a limitation in their maximum possible applied force due to the intrinsic compliance of fabrics.

IV. Effect of Haptic Feedback on Performance

Various studies have been conducted on the effect of different types of feedback on user performance in tele-operation. These studies cover visual, auditory, force, vibrotactile/cutaneous and pseudo-haptic feedback. The dominant theme in the following are comparative experimental studies between two or more types of feedback, conducted in either a virtual environment or using real setups, with typically a dozen participants.

Visual feedback is the main type of feedback present in tele-operation. The user can see the movements of the telecommanded platform, but in the absence of haptic feedback, they do not feel the contact forces between the robot and the environment. Most of the studies investigating the effect of haptic feedback on performance are based on the comparison of the visual feedback only with shared force and visual feedback.

The effectiveness of haptic feedback is reviewed in a survey published in 2009 [17]. A decade ago, they reported that, although in most of the experimental studies some improvement was observed using haptic feedback, results were mixed and not conclusive. In a 2010 review, the authors summarized the evidence for force feedback benefits in MIS, and reported that the results are favorable for force feedback [20]. Another survey in 2011 [21] reported that lack of haptic feedback is a major safety issue; however, cost-benefit studies should be conducted to justify its implementation. In another study, the authors concluded that force feedback has a large effect on average and peak forces, while it has only a moderate effect on task accuracy and no effect on task completion time [18]. A review of the benefits of haptic feedback in tele-operation and VR systems in 2015 [19] concluded that force feedback has a significant positive effect on operator performance. On the other hand, the effect of vibrotactile feedback is much lower and it is mainly helpful only in avoiding exaggerated force insertion.

The higher cost and lack of haptic feedback are the main limitations of robotic surgery [22]. Some newly developed robotic surgery platforms have integrated haptic feedback including Surgeon's Operating Force-feedback Interface Eindhoven (SOFIE, Eindhoven University of Technology, Netherlands) and DLR-MIRO (Institute for Robotics and Mechatronics, Germany), however, they are not commercially available yet. In this section, we focus on recent research not covered in the earlier review articles.

Multiple studies have looked into the effect of visual versus force feedback on performance. Experimental tasks vary from cannulation in MIS and contactless laser surgery to palpation and grasping of artificial materials, navigation and tele-manipulation in VR. Participating group sizes range from one to 20 novice or experienced participants. Results are mixed. Applied force is reported to decrease [97] or increase [98], [99] in different studies. Execution time is similarly affected, with reduction in some scenarios [98], [100] and increase in the others [97]. Efficiency is more consistently reported to improve in the presence of force feedback [100]– [103]. This result is confirmed in another recent study on the effect of force and tactile feedback on users' performance while palpating a soft tissue phantom [104]. In an experimental study with 19 participants, it was demonstrated that the addition of vibrotactile and pneumatic feedback significantly improves correct localization attempts. Similar results were obtained in [24]. Comparison of direct force feedback with visual representation of force or no feedback at all in suturing porcine tissue shows that less force is applied in the presence of direct force feedback, compared to other conditions [105], [106].

Force, vibrotactile, auditory and visual feedbacks are compared in placement, insertion and needle steering tasks with one to 20 participants. Visual feedback only is reported as the least effective type of feedback [107]–[110]. Its combination with force and tactile feedback is preferred over auditory feedback [108], [109].

Commercially available surgical systems usually do not provide force feedback due to stability issues. In these systems, haptic feedback consists of cutaneous information and feedback. Starting from 2014, some research groups decomposed haptic feedback into these two components and looked at user performance and system stability in the presence of each of them separately. The tasks vary from manipulation of a brain phantom, needle insertion and navigation to tele-operated pegboard tasks in VR with three to 25 novice and expert participants. Although cutaneous feedback is reported to be sufficient for better performance [111], maximum improvement happens in the combined presence of all types of feedback [112]–[115]. However, in a wearable vibrotactile device for interaction in VR [116], it has been shown that in some scenarios tactile only feedback performs better than combination of tactile and visual [117]. Also, keeping interaction force and reducing motion-generating forces in knot tying and pegboard transfer tasks result in better performance compared to direct force feedback [118]. In addition, magnified haptic feedback in needle insertion improves performance compared to direct hand interaction [119].

In VR, haptic feedback is provided to the user by the perceptual combination of the force applied to the user and the visual feedback from the virtual environment [120]. "Pseudo-haptic" feedback occurs due to incongruity between haptic sensation and visual information [121]. An experimental study was conducted with ten participants. The experimental setup is a one DOF end effector applying force on a beam. Maximum displacement of the end effector is manipulated according to the desired object stiffness. Results show a fair performance of pseudo-haptic feedback in identification of compliance.

Most studies are focused on either technical aspects of implementation and display of haptics or cognitive mechanisms for users. The authors of [122] suggest to combine the two approaches to make the commercialization of applications more feasible. Although research has shown there is a need for force and tactile feedback in robotic surgery, the need is not yet addressed in any commercially available solution or products [12].

In summary, the average number of participants in each experiment are twelve and apart from three in VR, the rest are conducted on animal or artificial tissues. The tasks range from palpation and probing to suturing, knot tying, navigation and needle insertion. While various robotic platforms such as KUKA, Mitsubishi and da Vinci are used as the slave manipulator, the most popular master haptic devices in these experimental studies are Phantom, Omega and Haption Virtuose.

Based on this review, visual and auditory feedback are the least effective types of feedback compared to others. It is better to have visual feedback than no feedback at all; however, complete force, vibrotactile and cutaneous feedback are more effective in improving performance. Berger *et al*. talk about the "uncanny valley of haptics", arguing that providing the user with additional information including haptic feedback is beneficial only if these information channels are fused and presented in congruence [123]. Otherwise, haptic feedback could even degrade performance.

V. Discussion

A decade ago, haptic feedback devices were at the very early stages of development and mostly provided force feedback only [15]. Sensing of tool-tissue interaction forces was reported as the most challenging issue. A few tactile feedback devices were developed but they had not been tested in clinical situations.

Providing the operator with some type of information from the interaction between the robot end effector and its environment is believed to be beneficial in improving the efficiency and reducing the errors in telemanipulation. This is especially challenging in medical applications, as collecting data from the medical tool-living tissue interaction has some practical limitations, and the most efficient interface for providing this information to the user depends on the application.

In this article the main research streams in measurement of tool-tissue interaction forces, namely proximal sensing, distal sensing and contact-less methods are presented. Proximal sensing is known as the easiest approach. Placing one or more force and/or torque sensors on the tool shaft or actuation pulleys, makes approximate measurement of applied forces possible. The main drawback of this method is lack of accuracy due to unmodeled interaction forces between the tool tip and the sensor. However, it is widely preferred to have some haptic feedback even if it is not precise, rather than no feedback at all. The research in this domain is quite mature with various research teams contributing to the field.

Distal sensing is much more challenging due to the strict size limitations associated with the small tool tip of medical instruments, sterilization and cost considerations, but it can potentially provide more accurate measurements. This is a young field of research and only a few teams around the world are working on developing dedicated sensors with adequate sensing range and size. With further advancements in MEMS and material sciences, distal sensing can turn into one of the preferred choices for measurement of tool-tissue interaction forces. If the limitations of current technology can be overcome, it will provide a firsthand onsite report from interaction forces with no losses involved. Our expectation is for these sensors to reach the market within the next decade. However, in many cases it will require replacement of conventional instruments by those with embedded sensors on the tool tip, resulting in additional equipment cost.

Finally, contact-less methods have recently been deployed for force measurement in teleoperated medical applications. More attention has been devoted to this field in the past eight years with studies on measurement of tissue deformation through endoscopic images. However, current advancements are confined

TABLE I
SUMMARY OF DIFFERENT TYPES OF FEEDBACK IN TELEOPERATION AND THEIR EFFECT ON USER'S PERFORMANCE

Feedback	Year	Task	participants	Result
Force vs. visual	2007	Cannulation in MIS	6 students & 6 surgeons	Reduced applied force in presence of force feedback and increased trial time for students [97]
Force vs. visual	2007	Probing on silicon gels	8 novices & 6 experienced surgeons	Larger applied forces by surgeons but quicker detection of tissue contact [98]
Force and visual	2013	Palpation on sample sponges with varying stiffness	20	Improvement of palpation quality and characterization of soft tissue with force feedback [101]
Force vs. visual	2014	Grasping a rubber band and stretching it	1 experienced surgeon	Larger applied grip forces in the presence of force feedback [99]
Force vs. visual	2015	Tele-manipulation in virtual environment	14	Performance affected by inaccuracies in feedback [102]
Force vs. visual	2017	Navigation in tele-operated interventional radiology	20	Reduction in positioning error and execution time in shared control plus improvement in accuracy and efficiency [100]
Force vs. Visual	2018	Performing laser incisions	17 surgeons	Improvement of performance accuracy and task completion time with force feedback [103]
Direct force vs. visual representation of force	2017	Suturing on procine tissue	3 surgeons & 9 students	Less force and damage with direct force feedback [106]
Direct force vs. visual representation of force vs. no force feedback	2017	Suturing on procine tissue	3 surgeons & 9 students	Less force in the presence of either type of feedback compared to no force feedback [105]
Force & vibrotactile	2013	Placement of total hip arthroplasty	1 experienced surgeon	More accurate placement in the presence of haptic feedback [107]
Vibrotactile, auditory, visual	2013	Insertion of a virtual tool and retraction at the hitting point	8	Lower rejection time in presence of vibratory feedback. Vibratory feedback subjectively assessed as more pleasant [108]
Visual vs. vibrotactile & auditory	2015	Dry-lab tasks with the da Vinci robot	78 surgeons & 36 non-surgeons	Strong preference for having auditory and especially vibrotactile feedback over visual feedback only [109]
Visual vs. vibrotactile	2016	Flexible needle steering in soft tissue phantom	20	Vibrotactile feedback more effective than visual feedback in conveying navigation cues [110]
Visual vs. force vs. force & vibrotactile	2019	Soft tissue phantom palpation	19	Superior performance in the presence of both force and vibrotactile feedback [104]
Interaction force & reduced motion generating force	2014	Knot tying, pegboard transfer, object manipulation	9	Improvement of user's performance compared to direct force feedback [118]
Cutaneous only	2016	Manipulating a brain phantom	3 surgeons	Cutaneous feedback on surgeon's fingertips is enough in teleoperated robotic neurosurgery [111].
Cutaneous only, visual & auditory	2014	Bimanual teleoperated pegboard transfer in VR	10	Improved completion time and more accurate force exertion and displacement compared to auditory and visual feedback, but less efficient compared to complete haptic feedback [112]
Visual vs. cutaneous	2016	Palpation of a heart model	18	Reduced task completion time, exerted pressure and absolute error in the presence of cutaneous deformation feedback [115]
Kinesthetic vs. cutaneous & kinesthetic	2015	Perceived stiffness & needle insertion in soft tissue phantom	15 & 20	Improved performance in presence of both types of feedback compared to kinesthetic only [113]
Visual vs. Vibrotactile	2017	Perceived stiffness in VR	10	Improved performance in tactile only compared to visual only and combination of both [117]
Visual vs. vibrotactile vs. kinesthetic	2016	Navigation of laparoscopic instrument tip towards a target	23+11	Improved performance in presence of all types of feedback. Reduced execution speed with visual and vibrotactile feedback, but improved with kinesthetic feedback [114]
Magnified haptic vs. direct hand interaction	2017	Needle insertion in Bovine gelatin	13	Improved performance in presence of magnified haptic feedback compared to direct hand interaction [119]

to certain applications, lighting and imaging facilities. Future research should work on eliminating these restrictions by devising generalized image processing algorithms and tissue models that would fit various scenarios. Although quite challenging, vision-based force measurement methods offer a software platform with no hardware changes in the conventional platforms and consequently no sterilization concerns. This advantage makes them a cost-effective solution suitable for any platform that matches the required technical specifications for software implementation. A common drawback in many of the studies on measurement of tool-tissue interaction forces is lack of quantitative evaluation of the developed sensing technologies. In addition, investigation into the required bandwidth and accuracy of the measurement methods for various types of surgery, tasks and users is required. This will serve as a benchmark for performance evaluation of new and existing solutions for force sensing.

After measuring the tool-tissue interaction force, the next step is to feed it back to the user. Among the various commercial devices available, the Phantom and Omega are the most widely used devices in the research studies to date. Apart from the conventional desktop devices, wearable interfaces that provide vibrotactile feedback have also gained attention. These devices can be combined with desktop devices or used standalone. Moving from rigid platforms to flexible ones, fabric-based solutions are the most recent developments in this field. These interfaces are mainly used to provide the user with information about the stiffness of an object. In spite of some advancements, these are yet to find their way into commercial devices. This is mainly due to their limitation in applying large amounts of force, their limited applications and lifecycle. All of the three main types of haptic interface have their cons and pros making each of them suitable for certain types of applications.

Apart from medical interventions, haptics is deployed in rehabilitation [124]–[127], teleoperation of mobile robots [128], [129] and Unmanned Aerial Vehicles (UAV) [130] among others. Wearable haptic interfaces for teleoperation of robotic arms and UAVs in complicated manipulation tasks are widely addressed in recent years. Some examples of developments with application in teleoperation of robotic arms are wrist devices for kinesthetic feedback [131], [132], a pneumatically actuated glove for force feedback [133], a haptic electromyography perception mechanism [134], a wearable haptic fingertip device for human-robot team interaction [135], and a passive force feedback glove with layer jamming on finger joints [136]. In control of UAVs, recent developments include a cable-driven haptics guidance system integrated into an exosuit [137], haptic feedback on the torso by compressing closed air pouches against the skin [138] as well as a novel desktop device [139]. Thus, wearable haptic interfaces are becoming the dominant technology in both medical and industrial applications due to their flexibility, versatility and intuitiveness. Further investigation into the best type of device and the most suitable type of feedback for each application is required.

It is widely believed that haptic feedback improves the performance and accuracy in medical applications, however, results are mixed regarding the task completion time and exerted force. One of the most recent developments in this field is separate analysis of kinesthetic and cutaneous feedback on performance. Since the risk of system instability in closed loop force feedback arises from the kinesthetic part, omitting it would guarantee system stability. Cutaneous feedback is shown to improve completion time and accuracy, while kinesthetic feedback improves execution speed. However, complete force feedback would result in better outcomes compared to any of the two components individually. In spite of the positive evidence for using force feedback in medical applications, its commercial use is currently very limited. This is mainly due to the risk of instability in the system, thus further developments are required.

This study highlights a large variety in sensing and feedback mechanisms and devices as well as validation procedures. The selection of the best method for experimental validation of the new developments is utterly subjective. In these studies, the number of participants, task type, involved instruments, duration of the study and measures of performance are mainly chosen based on the available capacities in the research center rather than a widely accepted benchmark. Results point toward a need for a systematic approach to validation to make the new ideas comparable. This can be achieved via simulation and validated benchmarks, or development of better taxonomies characterizing the tasks in terms of what haptic solution would be suitable.

VI. Conclusion

Haptic feedback has been shown to have a positive effect on performance. However, there is an ongoing debate on the best form of haptic feedback and the aspects of performance most affected. While some areas such as proximal force sensing and desktop force feedback devices are quite advanced, research in other areas is yet to achieve maturity. Distal sensor-based and vision-based force sensing are promising solutions, which require further improvement. With advancements in fabrication techniques such as 3D printing with sterile materials, we expect distal sensing to attract further attention resulting in the development of more reliable sensors. More powerful processors and endoscopes pave the path for future developments in image-based sensing, reducing its current dependence on specific lighting conditions and tasks. Wearable haptic feedback devices and fabric-based solutions are gaining increasing attention in medical interventions among other applications. These devices offer unconventional solutions for feeding back haptic information on different body parts. We envision a surge in the fabrication of novel wearable devices suitable for use on various body parts and able to provide different types of feedback for a wide range of applications. Wearable haptic devices offer potential beyond medical applications achievable through more reliable, durable and powerful developments. Future advancements in this rapidly growing field will enhance safety, reliability and efficiency in medical tele-operation. However, an explosion in new sensing and feedback devices and mechanisms on one hand and lack of a widely accepted systematic validation procedures on the other hand, might impede the introduction of new devices and techniques in clinical settings. There is a need for better and more systematic validation, both through better user studies, but also through novel methods like validated benchmarks and new taxonomies that reduce the need for human trials, at least during early system development.

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