2008-07-01

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Dean Callaghan
Technological University Dublin, dean.callaghan@dit.ie

Mark McGrath
Dublin Institute of Technology, mark.mcgrath@dit.ie

Eugene Coyle
Dublin Institute of Technology, Eugene.Coyle@dit.ie

Recommended Citation

Callaghan, D., McGrath, M., & Coyle, E. (2008) Force Measurement Methods in Telerobotic Surgery: Implications for End-Effector Manufacture. *Proceedings of the 25th International Manufacturing Conference (IMC25),*Dublin Institute of Technology, pp. 389-398. doi:10.21427/D7889V

FORCE MEASUREMENT METHODS IN TELEROBOTIC SURGERY: IMPLICATIONS FOR END-EFFECTOR MANUFACTURE

D.J. Callaghan[1], M.M. McGrath[1] and E. Coyle[2]

1 School of Manufacturing and Design Engineering, Bolton St., Dublin Institute of Technology, Ireland.
2 School of Electrical Engineering Systems, Kevin St., Dublin Institute of Technology, Ireland.

ABSTRACT

Haptic feedback in telesurgical applications refers to the relaying of position and force information from a remote surgical site to the surgeon in real-time during a surgical procedure. This feedback, coupled with visual information via microscopic cameras, has the potential to provide the surgeon with additional ‘feel’ for the manipulations being performed at the instrument-biological tissue interface. This increased sensitivity has many associated benefits which include, but are not limited to; minimal tissue damage, reduced recuperation periods, and less patient trauma. The inclusion of haptic feedback leads to reduction in surgeon fatigue which contributes to enhanced performance during operation.

Commercially available Minimally Invasive Robotic Surgical (MIRS) systems are being widely used, the best-known examples being from the daVinci® by Intuitive Surgical Inc. However, currently these systems do not possess force feedback capability which therefore restricts their use during many delicate and complex procedures. The ideal system would consist of a multi-degree-of-freedom framework which includes end-effector instruments with embedded force sensing included.

A force sensing characterisation platform has been developed by this group which facilitates the evaluation of force sensing technologies. Surgical scissors have been chosen as the instrument and biological tissue phantom specimens have been used during testing. This test-bed provides accurate, repeatable measurements of the forces produced at the interface between the tissue and the scissor blades during cutting using conventional sensing technologies.

The primary focus of this paper is to provide a review of the traditional and developing force sensing technologies with a view to establishing the most appropriate solution for this application. The impact that an appropriate sensing technology has on the manufacturability of the instrument end-effector is considered. Particular attention is given to the issues of embedding the force sensing transducer into the instrument tip.

KEYWORDS: Minimally Invasive Robotic Surgery (MIRS), Haptic, Embedded Manufacture

1. INTRODUCTION

The development of Minimally Invasive Robot Systems (MIRS) has resulted from a need to address shortcomings associated with traditional Minimally Invasive Surgery (MIS). Despite the advantages of laparoscopic surgery there are a number of inherent drawbacks which include limited degrees of freedom, impaired vision, amplified effect of physiological tremor and loss of haptic feedback to the user [1]. MIRS systems are an accurate and reliable means of eliminating the shortcomings associated with traditional laparoscopic surgery. Current commercially available MIRS systems greatly augment the surgeon's ability to carry out an operating procedure effectively but lack the ability to relay haptic (kinesthetic and tactile) feedback to the user. Force feedback is masked and distorted due to friction between the trocar and the instrument, reactionary torques from the abdominal wall, and friction in the grasping mechanism. This indicates that the forces fed back to the user are mechanical forces from the slave robot and not the delicate interaction forces [2]. A typical MIRS system consists of three primary subsystems (Figure 1) arranged as a telerobotic network [3].

- The haptic display is the interface between human and machine which reproduces the measured forces to the user.
- The communication interface is the control module in which the appropriate control scheme is implemented.
- At the slave side is the Teleoperator which mimics the movements of the user and is the location for measurement of interaction forces between the robot end-effector and the tissue during surgical tasks.

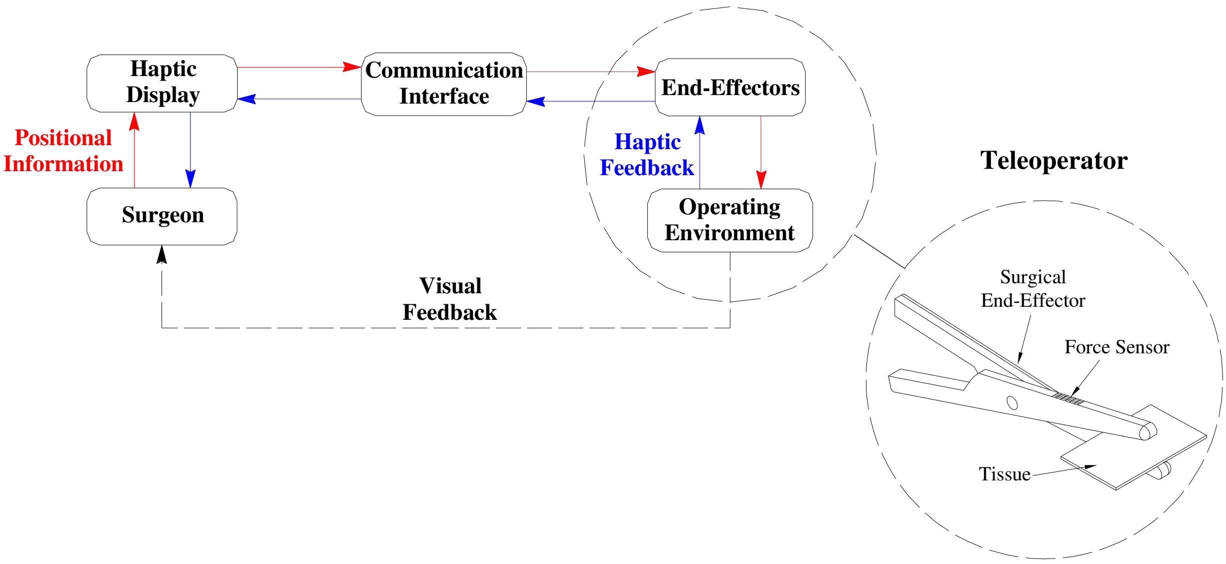

Figure 1. Minimally Invasive Robotic Surgical (MIRS) System

There are a number of feedback modalities which have been considered when attempting to measure the interaction forces between instrument and tissue. These include visual feedback, virtual fixtures, auditory feedback, and haptic feedback. A novel approach by Fischer *et al* [4] simultaneously measured the force applied by a grasper as well as the tissue blood oxygenation saturation as a means of limiting the maximum force being applied to the tissue. A GUI presents the data to the user in the form of coloured circles which change colour in proportion to the applied force. The system is awaiting Institutional Review Board (IRB) of medical devices approval to begin preliminary trials. Abbott *et al* [5] investigated virtual fixtures which is a method of preventing the user applying excessive forces or entering forbidden regions during a surgical procedure by physically restraining the instrument tip. Thorough robotic modelling and control approaches are required for the accurate placement of virtual fixtures. Current research is investigating the uncertainty involving robot position relative to anatomical structures due to unmodelled dynamics [6]. The effects of substituting direct haptic feedback with auditory cues were studied by Kitagawa *et al* [7]. This work concluded that although auditory cues gave additional support to the surgeon, it was suggested that such continuous auditory feedback might be disruptive and confusing in an already noisy operating environment.

In this paper the state of the art in haptic sensing technology currently available for applications in Minimally Invasive Surgery (MIS) is reviewed. The paper is organised as follows; Section 2 outlines the use of surgical scissors in MIRS, followed by a review of existing indirect and direct sensing schemes. Section 3.2 is a review detailing sensor requirements specifically for an effective direct force sensing method. The implications that these specific requirements have on the manufacturability of sensorised end-effectors are discussed in Section 4.

2. SCISSORS IN SURGERY

Scissors have been the traditional tool of surgeons for dissection in conventional surgery and have maintained an active role in laparoscopic surgery. This can be attributed to their precise operator determined action, safety, and low price in comparison with other dissection techniques. Dissection is regarded as a necessary component of many surgical endeavours carried out using open, minimally invasive, or robot-assisted operating techniques. The two-handed scissors-atraumatic forceps technique represents the mainstay of complex laparoscopic surgical dissection [8].

Surgical dissection and transection using round-tip, monopolar and bipolar scissors are integral parts of many of the most frequently performed surgical interventions such as adrenalectomy, cholecytectomy, gastric bypass, heller myotomy [9], and prostatectomy [10]. Sharp dissection implies the use of concentrated effort/energy on a relatively small area of tissue to achieve separation with little disruption to surrounding tissue. This is most readily achieved by dividing homogenous tissue at right angles to lines of tension. Blunt dissection on the other hand separates bulk tissues ideally between tissue cleavage planes [8].

Various researchers have investigated the measurement of interaction and gripping forces between a grasper and tissue [11-14] by attaching the force sensing transducer directly onto the grasper. To date little research has been undertaken to investigate the interaction between a pair of scissor blades and the tissue being cut through the attachment of a force/strain transducer on to or into the blades.

3. FORCE-SENSING METHODS FOR HAPTIC FEEDBACK

Full dexterity inside the patient as well as the decoupled determination of grasping/cutting forces and interaction forces is deemed necessary for appropriate immersion of the surgeon [15]. Achieving haptic feedback through the measurement of the interaction forces between a surgical end-effector and the tissue can be accomplished in a number of different ways. Placing the force sensing transducer at the distal end (direct force sensing) of the laparoscopic instrument is regarded by many researchers as being the ideal sensor location (Figure 2). This more accurately reflects the instrument-tissue interaction forces and reduces errors as a result of friction between the laparoscopic instrument and the point of incision [12]. However, the limited size of the blades or jaws places severe restrictions on the placement and integration of force sensors on the end-effector.

Alternatively, the distal instrument-tissue interaction can be estimated with a force sensor placed outside the patient's abdominal wall [16] as highlighted in Figure 2. This method overcomes miniaturisation constraints as well as sterilisation issues associated with direct force sensing. However, other problems are encountered such as disturbance forces at the incision point. This results in noisy force feedback which ultimately affects the quality of the force feedback particularly during delicate surgical procedures.

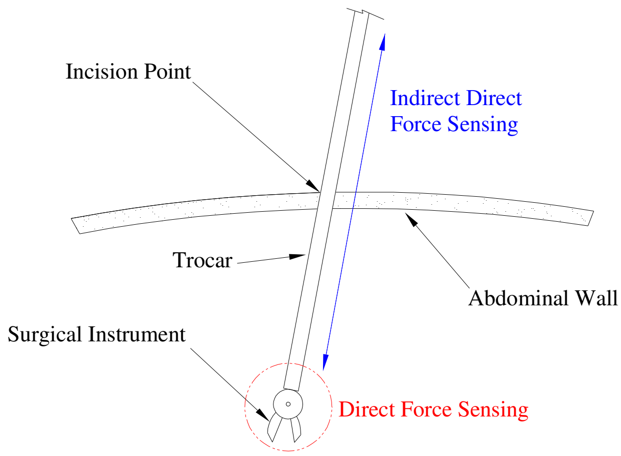

Figure 2. Indirect and Direct Sensor Locations

3.1 Indirect Force Sensing

A novel approach of overcoming the adverse frictional effects associated with indirect force sensing is being investigated by a number of groups. This method involves the use of what is commonly termed "the overcoat method". This is a double barrel arrangement allowing the trocar to be fed into the abdominal cavity unimpeded by the fulcrum effect at the entry point. Shimachi *et al* [17] developed a system using the overcoat method which can be integrated into the daVinci® robotic surgical system. However, the total force sensing error is estimated to be 0.5 N as a result of the deformation of the adaptor frame supporting the motion drivers. Studies were also carried out by this group which used accelerometer measurements to cancel out the adverse inertial and gravitational effects of the motion drivers/instrument mass along the axis of the trocar. This method does not consider the grip forces at the jaws of the instrument.

Zemeti *et al* [16] have also investigated the overcoat method; a 6-axis force/torque sensor, having a force resolution of 0.002 N in three dimensions and torque resolution of 25 μNm about x,y and z-axes, is mounted outside of the abdominal cavity. Experimental results highlight the robots potential for the measurement of contact forces at the distal end without being corrupted by the friction between the instrument trocar and the passive guide. Grasping forces are not measured or controlled as the robot currently consists of a manually controlled grasper.

3.2 Direct Force Sensing

An alternative means of sensing the interaction forces at the instrument-tissue interface is to locate the force transducers at the distal end either close to, or on, the instrument tip. There are two types of force that require measurement at this point, interaction/manipulation forces and grasping/cutting forces. Kuebler *et al* [18] have developed a six-axis resistive-based force/torque sensor utilising a Stewart Platform arrangement which is placed between the gripper and the cardanic joint. Results have shown that the sensor can provide realistic kinesthetic feedback of the remote interaction forces. The sensor can handle manipulation forces up to 20 N with a resolution of 0.25 N in the z-direction and 0.05 N in the x and y-directions. The force-torque sensor (FTS) does not cater for the measurement of the gripping forces, this being facilitated independently through the use of a uniaxial sensor.

The most suitable location for the force transducer is on the instrument tip allowing for direct measurement of the grasping forces and the interaction forces, without frictional and transmission disturbances. This is technically the most challenging location for placement of a force transducer owing to size restrictions. Other issues include the cost of the sensor as well as preservation of the design so as to ensure functionality and performances are not compromised [19].

A number of different sensing technologies are currently being investigated for suitability as direct force sensing transducers. Tholey *et al* [11] attached a flexible resistive element on to a grasper for the measurement of forces normal to the gripper surface. During calibration, the arrangement exhibited nonlinear characteristics as well as significant hysteresis up to an applied force of 13 N.

A grasper catering for the measurement of forces being applied in 3-DOFs has been developed by Fischer *et al* [4]. The gripper was manufactured from 2024-0 aluminium instead of stainless steel to increase sensitivity, while strain gauges were bonded on to the gripper for force measurement. The measured forces were displayed to the user via a haptic interface using visual indicators as a measure of the applied force. All electrical components in the device were coated with an appropriate silicon epoxy to ensure biocompatibility as well as sterility. This method does not allow a standard autoclave sterilisation protocol to be used and hence the instrument is sterilised using hydrogen peroxide.

A force measurement evaluation test-bed has been developed by Callaghan *et al* [20] which caters for the measurement of the range of forces experienced by a pair of scissor blades during a typical cutting procedure. A range of synthetic tissue samples of differing elastic properties were cut, while the magnitude and distribution of the forces along the length of the blades were recorded. The maximum forces experienced by the blades ranged from 2 N for the softest tissue sample to 14 N for the stiffest of the five samples. This apparatus will facilitate the characterisation and evaluation of new direct sensing transducers in the measurement of scissor-tissue cutting forces.

3.3 Direct Force Sensing Selection

It has been noted in Section 3.1 that for indirect force measurement schemes the quality of the estimated forces at the distal end is degraded as a result of frictional effects, gravity and the inertia of mechanical components. Many research groups have indicated that the ideal location for force sensor placement is as close as possible to the site of interaction [12, 19, 21-23], which for MIS is at the instrument tip. This implies that augmented instruments employing the direct force sensing method are the most appropriate for accurate measurement of complex instrument-tissue manipulation and gripping/cutting forces. Another benefit of accurate real-time direct force measurement is that the data collected from these instruments is expected to yield more accurate models for surgical simulators used in surgeon training. A general scheme showing the collection of such data is illustrated in Figure 3.

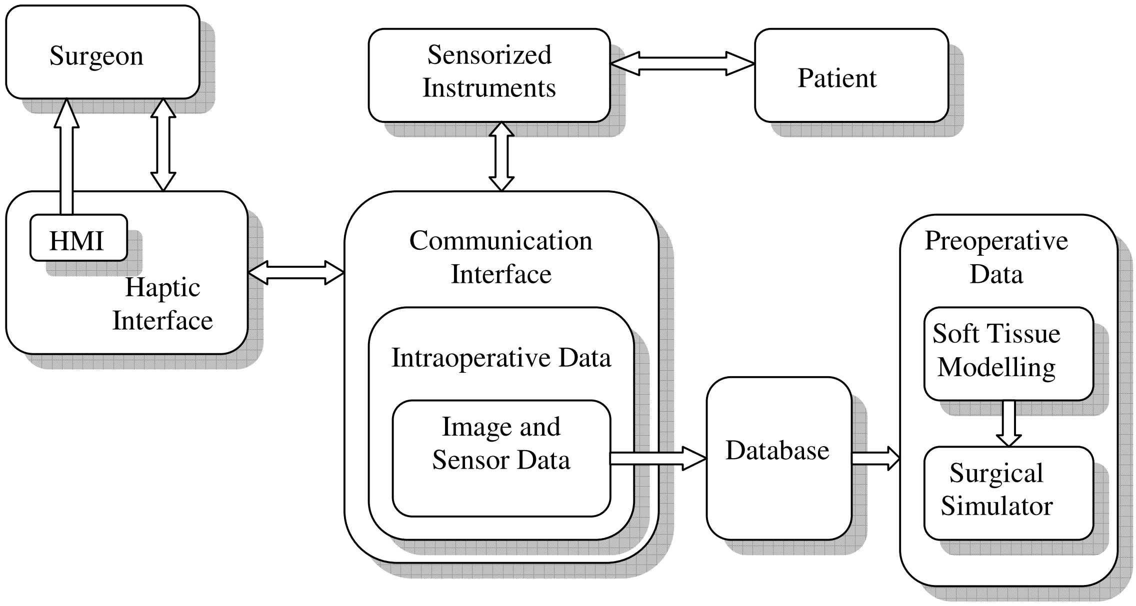

Figure 3. Data Transmission Scheme

3.3.1 Sensor Requirements

A number of important issues which are specific to a direct force-sensing scheme for the measurement of delicate interaction forces require consideration during the design and manufacturing phases.

- **Space Requirements** – Placing sensors directly on to the grasper or scissor blades requires that the sensing element be sufficiently small so as to maintain the integrity of the instrument tip. Advances in Micromachining technology has allowed Micromechanical systems (MEMS) to be successfully attached onto, or embedded into, surgical instruments [24]. Geometrical constraints, biocompatible material requirements, and assembly complexities of surgical MEMS can make device fabrication quite challenging [25]. An additional constraint with scissor blades is that the blade width is generally narrower than that of a typical grasper reducing the area on to which a sensor can be attached.
- **Distributed sensing -** A sensor with distributed sensing capabilities is required to allow the user to feel texture, blade slippage and subtle tissue anomalies. Haptic feedback is a combination of both kinesthetic (form and shape) feedback as well as tactile (texture and fine detail) feedback. The indirect and direct methods previously outlined give the user a sense of kinesthetic feedback in the form of interaction/manipulation and grasping forces. Tactile sensing is required to enable the user to feel the magnitude and position of the forces which are generated during grasping or dissection procedures. The physical construction of tactile sensors differs from that of force sensors in that the sensing element is distributed over the contact area of a grasper, scissor blade or tip probe.

  A grasper developed by Dargahi *et al* [13] using the piezoelectric polymer, PVDF, demonstrated that for a concentrated load, the entire surface of the sensor can be used to sense the magnitude of applied force as well as its position. The sensor has a measuring range up to 2 N and exhibits good sensitivity and linearity. One of the limitations of the reported sensor relates to the nature of the response from the PVDF film. The charge generated on loading slowly drains off over time following initial sensor displacement.

Optical Fibre sensors also exhibit potential for use in the area of tactile sensing for this application area. Two prototype tactile sensors were developed by Heo *et al* [26] using optical fibre sensors which exhibited no hysteresis, good repeatability, high accuracy and resolution. The drawback, however, with these sensors is their sensitivity to both strain and temperature, which need decoupling to allow effective strain/force measurement.

- **Modular design –** Commercially available MIRS systems have instruments that are modular in design and allow the instrument tip to be disposed of after approximately 12 to 20 uses. Integrating a force sensor on to the instrument tip would increase its complexity and cost. This requires consideration at the design stage so as to create an instrument that has an extended life and is thus reusable or alternatively can be manufactured at a cost that is acceptable enough to allow disposal of the instrument after one use.
- **Sterilisation and Biocompatibility** – Surgical instruments that are used inside the body need to be thoroughly sterilised to ensure complete destruction of microorganisms (e.g. spores). Steam sterilisation in an autoclave, at approximately 121°C and 205 kPa absolute pressure, is the standard sterilisation protocol used for most surgical instruments. Steam autoclaving between 132°C and 134°C is recommended by Intuitive Surgical® for sterilisation of the daVinci® surgical instruments. Therefore, it is imperative that a transducer placed at the instrument tip is capable of withstanding these elevated temperatures for between 4 and 15 minutes to ensure all spores are eliminated. It is also important to note that if the sensor is to be bonded to the instrument then the bonding agent should also be able to withstand these temperatures and pressures. All electrical components placed on to the instrument tip must be appropriately sealed and protected. This is achieved through application of an appropriate epoxy that is again sterilisable as well as compatible with the environment in which it is in contact. If suitable adhesives and epoxies are not available an alternative sterilisation protocol using Ethylene Oxide, Hydrogen Peroxide or other chemical agents may be more suitable. Ortmaier *et al* has suggested that as a result of unanswered questions surrounding sterilisability of electrical components alternatives such as optical methods may have to be used for measuring and transmitting information [27].
- **Sensor integration** – The majority of previous research efforts which focused on direct force sensing have attached the sensor *onto* the surface of the instrument. A viable alternative may be to embed the force measurement transducer *into* the instrument during the manufacturing process. This method ensures no contact between sensor and tissue, and as a result eliminates issues involving sterilisation and compatibility. Verimetra Inc. has successfully embedded MEMS sensors into microgippers. This eliminates the need for glue and adhesion layers which improves sensitivity and reduces errors due to creep [25]. A miniature polymeric gripper developed by Dollar *et al* [28] has six strain gauges embedded within its structure enabling three-axis force measurement. The shape deposition manufacturing (SDM) technique was used to manufacture the gripper arms which incorporate the embedded sensors. This technique proved to be a quick, inexpensive and robust manufacturing method. The gripper output is sensitive to temperature variations with a temperature drift of 0.15 N over a 5 minute period in open-air; indicating that temperature compensation may need to be considered during embedding of force sensors using SDM.
- **High speed dynamic sensing –** Realistic modelling and simulation of tissue deformation is an ongoing area of research as a result of the complexity of human organs and the challenges associated with the acquisition of tissue parameters [29]. The real-time collection of *in vivo* instrument-tissue interaction data can be used to validate the accuracy

and realism of these models. It is proposed that sensors used in the collection of this interaction information should have high-speed dynamic sensing capabilities enabling the measurement of tactile as well as kinesthetic force information.

4. MANUFACTURING IMPLICATIONS

There are a number of important issues which require consideration in relation to the manufacture of an effective sensorised surgical instrument. Embedding of the force sensor is a requirement if the sensorised instrument is to be sterilised using a conventional autoclaving technique. Embedded sensors are also unobtrusive resulting in an instrument tip that maintains its original shape, integrity and functionality. This can be achieved using a layered manufacturing technique such as laser assisted SDM technology. This method of manufacture has been proven to be a viable means of successfully embedding optical strain sensors into stainless steel components [30]. However, the sensors could be damaged during the manufacturing phase owing to the high melting point of stainless steel in comparison to the sensor material. A possible solution to this problem would be to coat the sensor using a sputtering or electroplating technique. This will form a metallic layer on the sensor surface facilitating; a) protection from the high temperature stainless steel deposit and b) a layer for the stainless steel to adhere to. Embedding thin-film thermomechanical microsensors requires that the sensor be fabricated on an insulating layer for protection before and after embedding [31]. This is crucial for stability and survival of the sensor as well as electrical and thermal isolation from the final embedding layer.

5. CONCLUSIONS

Current commercially available MIRS systems are unable to relay haptic sensory information to the user. One of the reasons for this shortcoming is the inability of the system to provide for the reliable measurement of interaction as well as grasping/cutting forces. It is proposed that a direct force method is the most appropriate means of accurately measuring delicate interaction and cutting forces for a dissection end-effector. In addition to realistic force measurement, the real-time *in vivo* data collected during surgical procedures can be used in the modelling of instrument-tissue interaction. These models can subsequently be applied in surgical simulators to augment the sense of realism felt by the user.

Placing the sensor directly onto the scissor blade allows force to be measured without interference from disturbance and frictional effects generated at the incision point. Embedding the sensor into the instrument structure is being proposed as a possible solution to sterilisation issues as well as addressing issues in relation to the restrictions with sensor placement. Manufacturability of a sensor-integrated end-effector has been discussed with particular emphasis on manufacturing techniques suited to embedding of the sensor. SDM, sputtering and electroplating are a few of the options available for instrument manufacture, the choice of which depends on the sensor type and the material into which it is to be embedded.

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