Tactile sensing technology for minimal access surgery—a review

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Abstract

Minimal access surgery (MAS), also known as keyhole surgery, offers many advantages over the more traditional open surgery. However, it possesses one very significant drawback—the loss, by the surgeon, of the “sense of feel” that is used routinely in open surgery to explore tissue and organs within the operative site. Because of this, important properties such as tissue compliance, viscosity and surface texture, which give indications regarding the health of the tissue, cannot easily be assessed. Restoring this tactile capability to MAS surgeons by artificial means would bring immense benefits in patient welfare and safety.

Artificial tactile sensing systems for MAS are reviewed. The technology is addressed from different viewpoints including those of the basic transduction of tactile data (tactile sensing), the computer processing of the transduced data to obtain useful information (tactile data processing) and the display to the surgeon of this information (tactile display). Applications of tactile sensing in MAS, both to mediate the manipulation of organs and to assess the condition of tissue, are reviewed. Some attempts to add tactile feedback to laparoscopic surgery simulation systems for MAS surgeon training are also described.

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1. Introduction

The state-of-the-art of artificial tactile sensing has been reviewed extensively [1–5]. Different aspects of tactile sensing have been addressed, including transducer principles, design requirements, sensor fabrication, and applications of tactile sensor for
manipulation and exploration. This work has concentrated largely on attempts to endow robots with a tactile sense.

Recently, Lee and Nicholls [6] examined the state-of-the-art of tactile sensing in its mechatronic aspects. They concluded that increased emphasis on understanding tactile sensing and perception issues has opened up the potential for new application areas including rehabilitation robots, service robots, food-processing automation and surgery. Here we focus on tactile sensing technology for minimal access surgery (MAS).

MAS is an operative technique developed to reduce the traumatic effect of surgery; it is also known as minimally invasive surgery, keyhole surgery, endoscopic surgery and laparoscopic surgery. In MAS the surgeon does not have his hands inside the operative field and the manipulations involved in the various procedures are carried out externally and transmitted to the operative site by long slender instruments inserted through small (5 or 10 mm diameter) access wounds. One of the openings is used to introduce a means of viewing the operative site and a light source to illuminate it. Images are displayed on a monitor [7]. Fig. 1 illustrates the essential features of MAS.

The advantages of MAS surgery include—less tissue trauma; less postoperative pain; faster recovery; fewer postoperative complications; and reduced hospital stay. These advantages may be translated into a total health care cost reduction for commercial and governmental institutions [8]. However, MAS has a number of disadvantages including—loss of tactile feedback; the need for increased technical expertise; a possibly longer duration of the surgery; and difficult removal of bulky organs [9].

Enhancing the tactile sensing capability of instruments used in MAS is a prime research area at present. The addition of tactile feedback to MAS simulation systems that are used to improve the practical skills of MAS surgeons is also an important requirement. The integration of tactile sensing with milli-robots for MAS surgery, in tele-surgery [10,11] and with active catheter devices for colonoscopy [12–14], would also confer benefits.

![](image_url)

Fig. 1. Minimal access surgery.
2. Artificial tactile sensing

Nicholls and Lee [6] define a tactile sensor to be ‘a device or system that can measure a given property of an object or contact event through physical contact between the sensor and the object’. According to this definition, a binary contact sensor for detecting simply the presence or absence of the body is a tactile sensor. They also consider other quantities that can take part in the contact process such as temperature, force, moisture content, surface texture, etc.

A tactile sensing system for MAS comprises three basic parts: a tactile sensor part to extract the tactile data through contact, a tactile data processing part that processes the transduced data to obtain useable information and a tactile display part that presents this information to the surgeon. These parts are shown schematically in Fig. 2. They are described in detail in the sections which follow.

2.1. The tactile sensor

2.1.1. Tactile sensor structure

Basically, any tactile sensor comprises four layers—a sensing layer, an electronics layer, a protective layer and a support layer. This is shown in Fig. 3.

The sensing layer might be a single element or a 1- or 2-dimensional array of elements. A tactile array might consist of a number of discrete sensor elements connected together by wiring. Alternatively, it might consist of a distribution of
sensor material in the form of a continuous sheet. The former poses difficulties in manufacture due to the need to accommodate the large number of interconnecting wires. In the ideal case, these arrays should be sensitive to both normal force and shear force. The main transducer technologies used are piezoelectric [15], piezoresistive [16,17], capacitive [18], optical [19], and mechanical [3]. Some transducers are based on elastomer [19] or silicon micromachining technology [20].

The electronics layer comprises the hardware necessary for signal conditioning. In this layer may be found multiplexors, bridges, amplifiers and other functional circuitry [20–24]. One major design trend is to integrate this layer with the sensing layer in the same chip as might be realised using integrated circuit (IC) compatible technology such as micromachining [25].

A thin layer of elastic material for protection, both of the sensor and, in MAS, the patient, covers the sensor. This layer also provides the important functionality of grasping. The presence of this layer may, however, complicate the analysis of the sensor output [26–28].

The support layer might be rigid or flexible [29]. The method of attaching this layer to the gripper greatly affects the performance of the overall sensor system.

The most common tactile sensors available today are static and passive. They can be used to provide only a static perception of object shape. Less common are dynamic and active sensors. These can be used in conjunction with relative motion between sensor and contact body to provide a dynamic perception of high frequency quantities such as surface texture, sharp of edges etc. Although these latter, dynamic, sensors are less far advanced than the simpler static type, they will be essential in providing a base for intelligent dextrous manipulation by robot [30–32].

2.1.2. Design considerations for tactile sensor for MAS

Since any tactile sensor measures, or at least detects, contact quantities between itself and the touched body, the design of such a sensor must start from a description of the working environment. This will determine the type, the number and the arrangement of the sensing elements forming the tactile array [33]. The environment for manipulation and tactile exploration comprises various kinds of objects to be handled and their mechanical properties such as weight, compliance, viscosity etc. The variable parameters of the environment are the locations and orientations of bodies together with their physical conditions (e.g. wetness, temperature, etc.)

In MAS the working environment is a closed system containing soft tissue, living organs and body fluids as well as other instruments deployed by the surgeon.

Since a tactile sensor for MAS is used inside the body, it must be reliable, biocompatible and waterproof and packaged in an appropriate useful manner. It must also be miniature and might need to be disposable. Thus, issues relating to cost and ease of assembly/disassembly must be addressed.

Some of these specifications can be met by the use of elastomer-based tactile sensors, but there remain some inherent limitations. Silicon-based tactile sensors have the potential to meet all of the specifications. In the following the two methods are investigated.
2.1.3. Elastomer-based tactile sensors

In the piezoresistive devices, the sensing layer is carbon- or silver-impregnated rubber. Contact forces between the sensor and the touched body cause a local increase in the concentration of conductive particles, and thus a local increase in conductivity [16]. Other devices measure changes in resistance between the conductive rubber layer and contact electrodes [17].

One major problem with this kind of sensor is the high degree of crosstalk between adjacent tactile elements due to the lateral stiffness of the rubber. This limits the spatial resolution to a few mm. Also, the random nature of the contact distribution between the conductive particles may cause significant deviation of the measured resistance from the nominal.

In one form of capacitive tactile sensor [34], the tactile array is composed of two layers of copper strips separated by thin strips of silicone rubber. As a force is applied to the surface above the point where two strips cross, the distance between the strips decreases, increasing the capacitance between the strips. By measuring the capacitance at each crossing point, the spatial distribution of pressure across the sensor can be determined.

The creep, hysteresis and lateral crosstalk properties of the elastic layer limit the size of tactile element in a rubber-based tactile sensor. Further, the viscoelastic properties of a rubber layer limit the dynamic range, and electrical connections with a conductive rubber are sources of noise.

2.1.4. Silicon tactile sensors

In contrast to these rubber-based sensing layers, silicon possesses excellent mechanical properties [20]. Mechanical deformation may remain linear over a considerable stress range and there may be negligible creep and hysteresis. These properties make silicon layers preferable in many applications. Moreover, microelectronic integration of interconnections within the silicon supporting substrate can allow a large number of tactile elements to be implemented.

To realise the desired silicon sensor structure, bulk micromachining, which etches bulk parts of the silicon from the rear, or surface micromachining, which deposits and etches structural layers on the top of the wafer, can both be used. These techniques can be used to construct a silicon diaphragm in the wafer, whose deformation by force or pressure forms the basis of the sensor action. This deformation can be measured capacitively or by means of microengineered versions of strain gauges.

Several attempts to realise a silicon-based tactile sensor using capacitive measurements have been reported. Chun and Wise [35,36] describe an $8 \times 8$ array tactile imager based on silicon capacitive transduction. The basic cell is formed between a thin, selectively etched boron-doped silicon diaphragm and a metallised pattern on an opposing glass substrate to which the silicon substrate is electrostatically bonded. Silicon dioxide is used to isolate the transducer plates on the silicon from the substrate and allow them to function as isolated row lines. In this structure, thick silicon rims support the silicon diaphragm so the cell is not easily scaled to smaller size for high-density application. Suzuki et al. [37] have modified this tactile imager by using a double supported bridge structure. The bridge structure occupies a smaller area for
a given sensitivity than would the diaphragm structure, allowing a $32 \times 32$ array. In addition, the bridge structure is supported on only two sides, requiring a smaller die area. Wolffenbuttel and Regtien [38], use surface micromachining to realise a tactile sensor using a bridge microstructure. The bridge is built on a composite silicon dioxide and silicon nitride film on top of a silicon wafer. A capacitive cell is formed between a doped polysilicon on silicon wafer and a free-standing doped polysilicon bridge on top of the wafer. The applied force acts on a force transfer beam formed from a polynitride layer. Polysilicon hubs at the border of the bridge allow the plate to be free-standing with two clamped edges. For tactile sensing applications, the maximum force that can be sensed is that at which the two opposing plates touch each other. However, the fabrication of polysilicon bridges requires a limited layer thickness, which implies a limited beam length and hence a reduced operating capacitance range. To overcome this problem a multi-bridge tactile element, composed of several bridges in a small array, has been investigated. Each bridge is very thin, but in combination they can withstand a substantial force.

Chu et al. [39] describe a three-axis tactile sensor based on a differential capacitive principle. The sensor is fabricated using IC processing and bulk-micromachining technology. Normal and shear forces are detected by capacitor arrays arranged as mesas. The bulk micromachining process is used to make the mesa structure. Three micrometer recesses are etched through a glass substrate. An aluminium layer is sputtered and patterned for the electrodes and interconnections. Anodic bonding is used to bond together the glass wafer and the silicon wafer. The sensor chip is supported on a polymer layer and covered with an elastomer layer for protection.

Gray and Fearing [40] report an $8 \times 8$ capacitive fabricated array intended for medical applications involving small manipulators and endoscopic surgery. The main problem, as yet unsolved, is the severe hysteresis to which the elements are subject. A foundry surface micromachining process is used. This process produces two structural layers of polysilicon (poly1 and poly2) and two layers of sacrificial phosphosilicate glass (PSG1 and PSG2). One or both layers of “sacrificial” glass are etched away to leave the poly1 and poly2 structure layers free standing.

Leineweber et al. [41] have developed a linear array tactile sensor comprising eight force-sensitive elements. Each element consists of two sensor capacitors and two reference capacitors. The latter are made insensitive to pressure by using a thicker membrane. This is used to eliminate parasitic capacitance and temperature effects. The sensor is fabricated using a surface micromachining process. Each pressure-sensing element consists of a capacitor whose top plate is a polysilicon membrane. The bottom electrode is implanted in the silicon substrate. The cavity, which is 80 $\mu$m in diameter, is realised by employing a sacrificial layer of silicon dioxide that is removed later by etching with hydrofluoric acid. The process is compatible with the standard CMOS process so that integration of the sensing elements and electronics for signal conditioning and data transfer on the sensor chip is possible.

Table 1 summarises the various attempts to realise a capacitive tactile sensor based on micromachining technology.

Several piezoresistive micromachined tactile sensors have also been reported. Petersen et al. [42] have developed a silicon-based force sensor. It is composed of two
parts—a micromachined silicon chip is bonded to a micromachined glass substrate by anodic bonding. The silicon chip contains transduction piezoresistors and addressing circuitry using commercial IC processes on one side. The other side is mechanically machined with a force-concentrating mesa. The glass substrate contains patterned bonding-pad metallisation. The developed chip can be operated either as a single force-sensing element or assembled into dense, tactile sensor arrays. This chip has been modified by Sorab et al. [43] to be used in a tactile sensing system for measuring fingertip-applied forces. It has been used to measure forces applied by the clinician during childbirth. The miniature sensors are here encapsulated in a compliant casing of silicone gel.

Sugiyama et al. [25] have developed a 32×32-element silicon pressure imager with CMOS processing circuits using an IC process. Polysilicon piezoresistors are arranged to form a full bridge on each individual silicon diaphragm. A micro-diaphragm pressure sensor is formed by single-side processing in each individual cell. Undercut etching of the interlayer between the silicon nitride diaphragm and the silicon substrate forms the reference pressure chamber. Each diaphragm is centrally supported on a narrow column.

Beebe et al. [44] have developed a silicon-based force sensor contained in a flexible polyimide-based package. The fabrication process is compatible with standard integrated circuit processes and produces a flexible package that sandwiches the metal leads between protective polyimide layers. Silicon direct bonding and bulk micromachining are utilised to fabricate the silicon-sensing element, which consists of a circular diaphragm over a 10 μm deep sealed cavity. The diaphragm is instrumented with piezoresistors in a bridge configuration. Sensitivity to force is realised via the addition of solid dome over the silicon diaphragm. The dome transmits the applied force to the diaphragm and protects it from damage. By an appropriate choice of diaphragm and cavity dimensions it can be ensured that the diaphragm bottoms out before breaking. Beebe et al. [29] have used this force sensor for finger mounted applications.

<table>
<thead>
<tr>
<th>Research group</th>
<th>Number of elements</th>
<th>Resolution (inter-element spacing)</th>
<th>Sensitivity (each element)</th>
<th>Load range (each element)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Chun, Michigan (1985, 1987)</td>
<td>8×8</td>
<td>2 mm</td>
<td>10 V/N (at max load)</td>
<td>0.5 N</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>1.5 V/N (at low load)</td>
<td></td>
</tr>
<tr>
<td>Suzuki, Michigan (1990)</td>
<td>32×32</td>
<td>0.5 mm</td>
<td>27 pF/N</td>
<td>0.01 N</td>
</tr>
<tr>
<td>Chu, Delft (1996)</td>
<td>3×3</td>
<td>2.2 mm</td>
<td>13 pF/N (normal)</td>
<td>0.01 N</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>32 pF/N (shear)</td>
<td></td>
</tr>
<tr>
<td>Gray, Berkeley (1996)</td>
<td>8×8</td>
<td>0.160 mm</td>
<td>5000 (%ΔC/C₀)/N</td>
<td>0.002 N</td>
</tr>
<tr>
<td>Leineweber, Duisburg (2000)</td>
<td>1×8</td>
<td>0.24 mm</td>
<td>200–250 V/N</td>
<td>0.02 N</td>
</tr>
</tbody>
</table>
Kane et al. [45] have developed a microstructure, consisting of a centre chattel plate suspended over an undercut etched pit by four bridge elements. This is capable of resolving the three independent components of a point traction stress. The sensor is realised using a fabrication process that is fully CMOS compatible to allow for the future integration of local processing and control circuitry. While contact pressure sensing is important, shear stress information is of critical importance for full grasp—force determination and slip detection as well as contact patch torque calculation.

Mei et al. [46] have developed a three-dimensional tactile sensor for robotic applications. The sensor comprises CMOS integrated piezoresistive sensing elements and on-chip data-reading circuitry together with a force concentrating structure and overload protection.

In the sensing array, there are $4 \times 8$ sensing cells that can detect three-dimensional forces. The sensing cell has an E-shaped square membrane structure fabricated by silicon bulk-micromachining. Finite element analysis was used to determine the optimal positions for the piezoresistors for measuring $F_x$, $F_y$, and $F_z$.

Table 2 summarises the various attempts to develop a piezoresistive tactile sensor based on micromachining technology.

As shown in Tables 1 and 2, arrays with configurations up to $32 \times 32$ have been achieved, and a spatial resolution less than 1 mm is possible. Different load ranges are available (from 0.002 to 10 N) and this can be varied without the need to modify the mask design. For example, increasing the thickness of the diaphragm fabricated by bulk micromachining will increase the load range and this could be done by

<table>
<thead>
<tr>
<th>Research group</th>
<th>Number of elements</th>
<th>Resolution (inter-element spacing)</th>
<th>Sensitivity (each element)</th>
<th>Load range (each element)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Petersen, Transsensory devices (1985)</td>
<td>Single element (can be assembled into arrays)</td>
<td>3 mm (in array form)</td>
<td>0.01 V/N</td>
<td>10 N</td>
</tr>
<tr>
<td>Sugiyama, Toyota (1987)</td>
<td>$32 \times 32$</td>
<td>0.35 mm</td>
<td>100 V/N</td>
<td>Not given</td>
</tr>
<tr>
<td>Beebe, Louisiana (1995)</td>
<td>Single element (can be assembled into arrays)</td>
<td>6 mm (in array form)</td>
<td>0.0015 V/N</td>
<td>10 N</td>
</tr>
<tr>
<td>Kan, Stanford (1996)</td>
<td>$5 \times 5$</td>
<td>0.6 mm</td>
<td>566 V/N (normal)</td>
<td>0.0162 N (normal)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>133 V/N (shear)</td>
</tr>
<tr>
<td>MEI, Hefet, China (2000)</td>
<td>$4 \times 8$</td>
<td>4 mm</td>
<td>0.013 V/N (normal)</td>
<td>50 N (normal)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.0023 V/N (shear)</td>
</tr>
</tbody>
</table>
controlling the etching process. Adjusting the diaphragm thickness can also be used to change the sensitivity but there is a trade off between the load range and the sensitivity. A recent trend is to develop sensors that have a capability to measure shear force as well as normal force.

The sensors listed in Tables 1 and 2 are presently all being applied in experimental setups and are not commercially available yet. However, the design of these sensors could be adapted and used in MAS applications. For example an array of $8 \times 8$ elements with a spatial resolution of 1 mm and overall dimension less than the trocar diameter (10 mm) could be applied in MAS. Such an array could be attached to a probe to give a tactile image representing a distribution of pressure when pressed against soft tissue and this could be used along with a neural network classifier to detect abnormalities embedded in the tissue.

2.2. Tactile data processing

Tactile data processing is an important aspect of any tactile sensing system. A variety of algorithms for processing the sensory data may be used. Data interpretation using model-based approaches equivalent to those used in visual image processing techniques and using neural networks are common.

Solid mechanics and finite element methods have been used for the construction of a forward model of a tactile sensor to obtain subsurface stress/strain data from the surface stress and material properties of the protective layer which covers the sensor [28]. The inverse model has also been investigated, where the contact parameters are calculated from the subsurface strain measured by the tactile sensor [47].

Muthukrishnan et al. [48] have adapted some edge detection methods used in vision systems. They have investigated two methods, the first for detecting the contours of an object smaller than the tactile sensor with a single probing operation; the second to approximate the contour of an object larger than the sensor area by sequential touching and integrating the straight lines so extracted.

Neural networks have also been used in tactile image analysis to identify contact shape [49], to discriminate fine form [50], and to perform classification [51].

Charlton et al. [52] have used neural networks to reconstruct contact parameters such as normal force, shear force and indenter width and position from tactile data. Their training data was obtained using finite element methods to construct and solve the model of the tactile sensor.

Brett and Stone [53] have developed a technique for measuring contact force distribution in MAS. They use a small number of sensory outputs to detect the bending of surface of known behaviour. Neural networks are used to interpret the contacting forces from the sensory data.

2.3. Tactile display

In principle, a tactile system should sense tactile stimuli in a remote environment, and present them to the user with the best possible fidelity. Tactile display is used to present to the user, information about texture, local shape, and local compliance.
The human skin responds to a number of distributed physical quantities, the most important of which are vibration, small-scale shape or pressure distribution, thermal effects and electrorheological effects.

Tactile display has received particular attention especially since new actuator technologies have been introduced. These technologies include shape memory alloy (SMA), piezoelectrics, and electrostatics. Actuators employing these technologies have been used in tactile display for some applications such as tele-manipulation and remote palpation during minimally access surgery. They can also be used to add tactile feedback within virtual environments.

Typically, tactile displays control either displacement or force. In displacement display, an array of pins is shaped into a contour. In force display, the pin array produces a surface stress distribution representing the data. The spatial resolution of a tactile display is limited by actuator size. Currently, the spacing between the centres of pins is around 2 mm [54,55]. Fast development of micro-fabrication technologies and the micro-actuators produced by them, opens up new possibilities to meet the requirements of high-resolution graphic tactile display.

Cohn et al. [55] have developed a 5 × 5 tactile display of pneumatic actuators driven by solenoid valves. This has the advantages of economy as well as binary operation allowing for uniform open-loop response from actuator to actuator. Also, continuously variable force output is possible using pulse-width modulation. The maximum force is 0.34 N and the bandwidth is 7 Hz.

Fisher et al. [56] have developed a tactile actuator array for use in MAS. Each actuator is made of an SMA spring which works against a bias spring, both are mounted on a ceramic pin with 0.8 mm external diameter. The bias spring produces the force for resetting the actuator during the passive cooling time. The maximum force is 2.5 N.

Howe et al. [34] have developed a tactile shape display for deployment on the handles of surgical instruments. Here, shape memory alloy is selected as the actuation method because of their very high power to volume, power to weight and force to weight ratios. Their non-linear behaviour, hysteresis and slow time response can be overcome by appropriate control schemes.

A length of SMA wire is attached to a rigid frame at one end and to a small lever at the other. A spring connected between the lever and the frame keeps the wire in tension and provides a restoring force. Heating with an electric current actuates the SMA wires—the elevated temperature results in a material phase change, which increases the tension and shortens the length between the ends of the wire. This causes the lever to rotate about a fixed shaft. The other end of the lever then forces a pin upward against the tip of the operator’s finger. The levers provide a 3:1 reduction in force and amplification in displacement.

Moy et al. [54] have developed a compliant tactile display for tele-taction. It comprises a one-piece pneumatically actuated tactile display moulded from silicon rubber. Instead of actuated pins, they use an array of pressurised chambers to provide the stimuli. The advantages of this compliant tactile display over other pneumatically actuated tactile displays include comfortable use, absence of leakage, and negligible pin friction.
Tactile displays using electro-rheological gel are under investigation [57] but these have yet to be extensively assessed and they also pose a possible safety problem due to the relatively large voltages involved.

3. Applications of tactile sensing in MAS

Howe et al. [34] have investigated the remote palpation technique for surgical applications. A tactile array sensor, in the remote tip of an instrument or probe measures the distribution of pressure across the tissue contact. The resulting signal is displayed using a tactile display device mounted in the finger tip contact area of the surgeon’s interface. This prototype system involves an 8 × 8 array at 2 mm spacing in each direction, providing 64 force sensitive elements. This system has been used for tumour localisation in a simulated tissue consists of a hard rubber cylinder inside a block of foam rubber.

Cohn et al. [10] describe a laparoscopic manipulator with hand-like end-effector. The manipulator fingers are constructed as moulded rubber balloons. They suggest using a strain sensor array on the end effector surface to provide tactile feedback. One potential application of this manipulator is the palpation of the lung for detection of tumours [55].

Omata and Terunuma [58] use a different approach for stiffness detection of soft tissue, which involves the use of a piezoelectric ceramic as a transducer. This is caused to vibrate at its resonant frequency. When the free end of the probe touches a material the resonant frequency shifts due to acoustic impedance. The shift in resonant frequency depends on the stiffness of the material. Matsumoto et al. [59] have used this sensor to ascertain the presence of a gallstone in the gallbladder or common bile duct during laparoscopic surgery.

Brett and Stone [53] have investigated a method for obtaining force and tactile information. Their approach is to determine a distribution of contact force using a small number of sensory elements distributed across the surface of a finger (of known bending behaviour). The bending of the finger surface is used to assess the contact forces. The output of the sensor elements in contact with the soft tissue is used in conjunction with the behaviour of the finger surface to compute surface shape using algorithmic or neural methods.

Eltaib and Hewit [60,61] have described an approach to tissue condition assessment. This involves the application, to the tissue surface, of a sinusoidal displacement, and measurement of the resulting contact force. This used to identify the dynamic properties of the tissue. A tactile probe for MAS based on this principle has been used for detecting abnormalities in a simulated tissue made from gelatine containing balls made of stiffer material.

Attempts to use tactile sensing for manipulation have been reported. Active catheters with multiple tactile sensors mounted on the tip have been designed and fabricated by a Japanese group [12,13]. The catheter incorporates two MAS wire bending actuators. The tactile sensors are fabricated monolithically on flexible film
using thinned integrated circuits. The system includes three tactile sensors, one passive sensor for temperature compensation and aluminium connection wires.

Virtual reality (VR) is a powerful tool for training, simulation and computer aided design. In medical training, VR has the potential to reproduce the natural characteristics of human anatomy, requiring only a re-setting of the host computer once a particular training session has been completed. The lack of tactile feedback in VR for MAS training systems limits their usefulness.

In order to provide tactile feedback to the users, force/tactile interfaces have developed in recent years Burdea [62]. These interfaces enable the user to interact with virtual environments by receiving motor action commands from the user by displaying tactile images to the users. Massie and Salisbury [63] have developed a device (the PHAToM) comprises a serial three degree of freedom interface that provides a transitional workspace in three dimension. The device provides a computationally and mechanically tractable way to enable haptic interaction with complex virtual objects. Benali et al. [64] have developed a six degree of freedom, haptic interface system for surgical virtual reality training. The system can apply forces to the user’s hand and can also regulate the sensation felt when contact occurs. Different types of contact and movement can be simulated.

4. Conclusions

Artificial tactile sensing has been investigated extensively last twenty years. Different aspects of tactile sensing have been addressed, including transducer principles, design requirements, sensor fabrication, and applications of tactile sensor for manipulation and exploration. This work has concentrated largely on attempts to endow robots with a tactile sense. Recently, other applications of tactile sensing in medicine, food-processing automation and surgery have been addressed.

This paper has focussed on the development of artificial tactile sensing systems for minimal access surgery. The technology is addressed from different viewpoints including those of the basic transduction of tactile data (tactile sensing), the computer processing of the transduced data to obtain useful information (tactile data processing) and the display to the surgeon of this information (tactile display). These three aspects are of equal importance to the production of an effective tactile sensing system but they have been investigated so far with different emphasis. The design and fabrication of tactile sensors have received the greatest attention to date and a number of reasonably effective arrays are becoming available. Tactile data processing is still fairly rudimentary and much effort is needed to provide surgeons with meaningful and useable information. Tactile display is even less well established and the display devices available at present are generally too large, too imprecise, too awkward to use and too expensive to make much impression on MAS surgeons.

The field of tactile sensing is likely to be a fruitful one for a number of years while the difficult problems of trying to emulate the human tactile facility are tackled. However, the benefits to minimal access surgeons of regaining their lost sense of touch are immense.
References


