An IPMC actuated robotic surgery end effector with force sensing

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Growth in patient acceptance of robotic-assisted surgery has led to increased demand and has stimulated research in many new surgical robotic applications. In some cases, the performance of robotic surgery has proven to surpass that of human surgeons alone. A new research area which uses the inherently force-compliant and back-drivable properties of polymers, ionic polymer–metal composite (IPMC) in this case, has shown potential to undertake precise surgical procedures in the delicate environments related to medical practice. This is because IPMCs have similar actuation characteristics to real biological systems, which can help ensure safety. Despite this, little has been done in developing IPMCs for a rotary joint actuator for functional surgical devices. This research proposes and demonstrates the design of a single degree of freedom (1-DOF) robotic surgical instrument with one skeleton-joint mechanism actuated by IPMC with an embedded strain gauge as a feedback unit. The system performance with a developed gain-schedule PI controller is demonstrated. Despite the simplicity of the system, it was proven to be able to cut to the desired depth using the implemented force control (up to 8 gf cutting force).

Keywords: IPMC; integrated sensing; surgical robot; force control

Introduction

Robot-aided surgery and robotic surgical instrumentation are currently some of the fastest growing applications in robotics. Advances are being made at a rapid pace with much research in laboratories around the world. This fundamental research has led to commercial devices available in the marketplace, which are now becoming common place in operating rooms for surgical procedures around the world. Many robotic devices have now surpassed human capabilities in a number of ways. One particular example is the Da Vinci® surgical robotic system from Intuitive surgical, with more than 2585 systems already installed in hospitals worldwide [1].

Robotic-assisted surgery is becoming widely adopted by surgeons for a number of reasons, including advantages such as improved dexterity with advanced instrumentation; undisputed higher precision, which leads to shorter learning curves [2]; embedded force/torque sensors, which have the ability to give direct kinaesthetic feedback to surgeons [3]; more safety with active constraints through appropriate control algorithms; reductions in surgeon tremor through low pass filtering and faster patient recovery times, and cosmetic advantages. Although current advanced surgical devices like the Da Vinci® have many advantages, they require complex mechanisms and highly geared servo motors to ensure intense accuracy. This results in mechanically bulky systems, which carry huge price tags and have large end point impedance [2]. They also rely heavily on force and position

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sensing to maintain safe contact during operation, and patients have to be fixed in position [4]. Force compliance could be a concern with such systems.

Ionic polymer–metal composite (IPMCs), on the other hand, are inherently compliant [5] and therefore have low tip mechanical impedance, which will ensure safety during surgery. IPMCs are also designed to be easily miniaturized and will therefore result in large cost savings. There is currently research with IPMCs for surgical applications, which have proved its viability as a good material for surgical robots, and some of these include a fragment capturer to collect the plaque generated in surgery to remove a thrombus [6], 3D tweezers developed using arbitrary-shaped IPMC [7], an active catheter in [8], and a system for endoscopic surgery [9]. Little has been done using IPMC as actuator in a joint device and in this case as a robotic surgical end effector with embedded sensing for force control in addition to the natural force compliant characteristics of the IPMC. This device has the capability to be mounted on larger robot systems as an end effector, such as a Da Vinci® system or a light weight robotics platform [10] specially made for facilitating a small surgical instrument to undertake the surgery. A larger arm with a bigger working envelope will carry out large (macro) position movement while the IPMC end effector can handle delicate and precise movement (micro). This will combine the advantages of these two systems and give the best performance and properties for a surgical device.

**Robotic end-effector design**

The device will be designed as a robotic surgery end effector to undertake precise and small scale cutting procedures aimed at cutting tissues to a desired depth. Figure 1 illustrates the proposed end effector attached to a larger robot.

It is envisaged that the procedure used to cut tissues to a desired depth can be a two-step process similar to slicing. Initially the scalpel will be vertically lowered by the robot arm until the scalpel just touches the tissue surface. Since the IPMC is force compliant, it will bend if the robot arm slightly over-travels vertically downwards. Then a controlled

![Figure 1. A proposed IPMC surgical end effector attached to a robot.](image-url)
voltage will be applied to the IPMC to achieve the appropriate blocking force (hence cutting force). Note that at this stage, the tissue has not been cut. When the appropriate force has been reached, the robot arm will move horizontally so that the scalpel will start cutting the tissue. The cutting force will be regulated by the IPMC to maintain a constant cutting depth. A 1-DOF joint will be sufficient for the simple cutting task and so only the end effector will be designed and tested, not the whole robotic system. The end effector consists of a ‘skeleton’ frame with one rotary joint actuated by IPMC, see Figure 2. An IPMC from Environmental Robots Inc based on an XR resin with Pt electrodes was used for all experiments. The IPMC was 27.5 mm long, 10 mm wide with a thickness of 800 µm. XR resin was purchased from DuPont and used to manufacture the thick IPMC. It was first melted to the desired thickness and then hydrolyzed in a solution of KOH and DMSO to convert it into an ion containing cationic polyelectrolyte. It was then put through a REDOX chemical process by first oxidizing it with a metallic salt such as tetra-amino platinum chloride hydrate and then reduced by sodium borohydride to make an active IPMC sensor and actuator. The IPMC has one end clamped at the joint while the other end can slide in a slot. The IPMC will work in air without hydration. This actuation method is proposed and proved to be effective in previous work by McDaid et al. [11]. The advantage of this mechanism is in the vertical direction, the device will be compliant, while in all the other directions, the system is considerably rigid, ensuring all cutting is precise and requires simple sensing and control. As the IPMC is in a rigid frame, it is also free from the influence of the irregular deformation of the IPMC itself and load from any undesired direction. This article aims only to demonstrate the IPMC holding a cutting tool used for tissue cutting, hence a frame was used to hold the cutting tool as shown in Figure 2, although as mentioned earlier this tool can be connected directly to a larger robot, for example, a serial robot as proposed in [11]. This means there is a precise but rigid structure connecting the scalpel, which is actuated by a soft and compliant IPMC actuator. This gives the best performance and properties needed for a surgical device.
A cantilever beam made from printed Acrylonitrile butadiene styrene (ABS) plastic connects the scalpel holder and rigid skeleton frame that houses the IPMC actuator. The ABS cantilever is used to give some small, but measurable deflection when a force is applied to the cutting tip. Using an appropriate sensing circuit, this bending displacement can be measured and used to determine the force being applied when cutting at the tip of the scalpel.

Figure 3 shows a finite element analysis (FEA) simulation using ANSYS® for the overall deformation, when a cutting force generated by the IPMC is 10 gf (through preliminary experimentation it is found that this is the maximum output force achievable with this type of IPMC purchased from Environmental Robots Inc., USA) The scale bar is in millimeters and the major deformation is concentrated at soft beam as intended. The amount of deflection from the simulation will be sufficient to be detected with the use of a common, low cost, foil strain gauge.

**Implementation of a force sensor**

To miniaturize the cutting tool for delicate operations and even key-hole operations, the entire system including sensor must be embedded within the device. Measuring the deflection on the cantilever beam instead of directly measuring the force is simple as a strain gauge can be used instead of a more complex force sensor. This will give high resolution feedback without the need of complex circuitry or large external sensors, for example, load cells or laser sensors.

A strain gauge purchased from RS Components Ltd (5 mm foil strain gauge, RS 632–168) is adhered to the cantilever beam to measure the deflection when a force is applied. The deflection should be proportional to the reaction force at the cutting tip of the scalpel, and this will be verified through simulation and experiments.

According to Euler–Bernoulli equation, the displacement at the location of the strain gauge with respect to applied reaction force is defined as:

$$\delta_s = \frac{P_r}{6EI} x_s^2 (3l - x_s) = \frac{P_r}{K_M} F(x_s)$$  \hspace{1cm} (1)
where $\delta_s$ is the displacement at $x_s$ (position of the strain gauge), $P_r$ is the reaction force, $K_M$ is a constant according to the material properties, and $F(x)$ is a function that depends on position $x_s$. This equation was used to design the thickness of the cantilever to obtain a sizeable deflection, which can be measured by the strain gauge, under a typical IPMC force. The strain gauge deforms together with the soft cantilever beam, and the resulting change in resistance of the strain gauge can be easily measured by a circuit consisting of a bridge and instrumentation amplifier.

According to the FEA simulation, the strain in this area was approximately 0.1%, and taking into consideration the unevenness caused by manufacturing the real part, the required gain factor should be more than 5000. Experiments were conducted with a gain factor of 5000, and the required voltage was scaled from 0.7 to ~2 V and the output force between 0 and 10 gf. From Figure 4, the relationship between the amplifier output voltage, $V$, versus the force, $F$, is $V = 0.1121F + 0.7125$ and this will be used in the closed-loop control at a later stage.

The final end-effector surgical tool with the IPMC, strain gauge, and scalpel is shown in Figure 5.

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**Figure 4.** Relationship between strain gauge amplifier’s voltage versus the measured force.

**Figure 5.** The prototyped surgical end effector.
Tissue cutting experiments
In this research, two experiments are conducted, the first to determine the relationship between the blocking force and slicing depth of a replica tissue and the second to test the ability of the PI controller to maintain a constant force of the scalpel when slicing across the tissue of varying thickness. Figure 6 shows the set-up to determine the cutting depth and cutting force.

To test the blocking force on a replica, tissue sample made from silicone rubber (ECOFLEX 0030 from SMOOTH-ON) was used, which is slightly harder than human skin. The scalpel was a standard surgery scalpel number 15. A 3 mm bolt was screwed in a threaded hole on the top of the frame to press the arm downward at same position the IPMC will be actuating from during real operation. The desired force is manually provided by screwing the bolt up and down. Finally, the output force over the tip of the scalpel was measured by a ‘gram-force, gf’ scale. When the required force was provided onto the silicone rubber, it was then slowly pulled horizontally so that the scalpel travels across the silicone similar to a slicing action. To measure the resulting cutting depth into the silicon, a cross section view using a calibrated microscope was used. Figure 7 shows the cross section views of the silicone with the cutting depth measured. The depths in Figure 7(a)–(d) are caused by the scalpel with manually exerted cutting forces ranging from 8 to 14 gf with a 2 gf increment.

Figure 8 shows the relationship between the cutting force and cutting depth and can be expressed as a linear relationship of $D = 0.0189F - 0.1132$, where $D$ is the depth, $F$ is the cutting force. From the plot it can also be extrapolated that it will take approximately 6 gf just to start slicing ‘skin’ of the tissue (in this case the silicone rubber), that is, to break the surface.

The relationship between the force and the cutting depth is important when it is implemented with an IPMC. With the cutting tool actuated by an IPMC, a closed-loop system with force feedback from the strain gauge can measure the actual blocking force. Since the IPMC is force compliant, the cutting force will be limited even if the distance between the tissue and the robotic arm changes during the slicing operation due to changes in the tissue thickness.

PI controller design
Previous research in IPMC force control [12,13] has proved that a proportional-integral-derivative (PID) controller has acceptable efficiency for the prototype in this research. A

Figure 6. The (a) front view and (b) side view of a simple set-up to characterize the cutting depth versus cutting force.
model-based method was used to design the force controller, based on the open loop step response (0.5 V to 3 V, 0.5 V step increment) of cutting force. In the experiments, a load cell (Sherborne Sensors Ltd., type SS2) was used to characterize the force response of the IPMC and also to calibrate the strain gauge amplifier’s output voltage to the cutting force provided by the IPMC actuator as shown in Figure 9.

The open loop steady state cutting force versus applied voltage to the IPMC response of the IPMC is shown in Figure 10. This figure shows that the cutting force versus applied voltage can be simplified into a linear system by dividing into two linear regions (applied voltage of ≤1.5 V and >1.5 V).
Here, a gain-schedule PI controller was developed. A PI controller is sufficient because IPMC has a low frequency response. Two different gain sets will be determined. The PI gain-schedule controller is then further tuned with help from the SISO design toolbox and interactive PID tuning tool in Matlab® with 5 different tuning methods: Ziegler–Nichols step response, Skogestad IMC, Chien–Hrones–Reweick, approximate M-constrained integral gain optimization, and robust-response time. The best tuning results are $K_p = 0.154$ and $K_i = 0.054$ when the input voltage range from 0.5 V to 1.5 V and $K_p = 0.445$ and $K_i = 0.108$ when the voltage is >1.5 V to 3 V.

Figure 11 shows the various cutting forces versus time as provided by the IPMC. It is clear that the simple gain-scheduled PI controller can be used to provide various cutting forces with reasonable accuracy. Here, the actual force is being read by the load cell (blue), while the strain gauge (brown) is used as the feedback to the PI controller. From Figure 11, the load cell and strain gauge data track each other and are closely matched to
the requested force (magenta). The output (green) plot is the voltage from the controller to the IPMC actuator. Hence, the gain-schedule PI controller has been shown to be sufficient to control the IPMC.

Figure 12(a) shows the experimental set-up to demonstrate the cutting of a replica silicone tissue with a scalpel actuated with an IPMC and controlled with a gain-schedule PI force controller. The arm is lowered until the scalpel touches the silicone and then the controller takes over when the demanded force is input into the controller. The scalpel is then moved horizontally in a constant speed (slicing speed), which cuts the tissue. This preliminary work aims to demonstrate that the cutting force can be maintained when cutting in two typical cutting conditions, that is, (a) the scalpel will cut across a silicone tissue of varying thickness but at constant slicing speed and (b) a constant tissue thickness but at higher slicing speed. Here, the silicone rubber was made in such a manner that the thickness is almost constant from D to B and increases from 3 mm at one edge (C) to 9 mm (A) at the other end as in Figure 12(b). This is used to mimic slicing speed with level conditions and slower slicing speed but with unlevelled cutting conditions.

Figure 13(a) shows the output voltage from the PI controller to the IPMC (dark blue) gradually decreasing when the scalpel travels horizontally across the silicone tissue that increases from 3 mm (C) to 9 mm (A) with a demanded cutting force of 8 gf (green) and in the process cuts the tissue. The measured force by the strain gauge varies between 8 gf.

Figure 11. Cutting force controlled by gain-scheduled PI controller feedback from the strain gauge.

Figure 12. (a) Experiment set-up to demonstrate the force control when cutting across a silicone tissue, (b) of constant thickness (D to B) but slicing speed of 2.1 mm/s, and of increasing thickness (C to A) but slicing speed of 1.3 mm/s.
and 10 gf (light blue) as the scalpel slices across from C to A. In an attempt to regulate the cutting force to the demanded 8 gf, the voltage to the IPMC (dark blue) varies to compensate for this uneven condition. In Figure 13(b), the slicing speed of the scalpel from D to B is slightly faster (2.1 mm/s) from the previous cut from C to A (1.3 mm/s). As the scalpel slicing speed increases from 1.3 mm/s to 2.1 mm/s at $t = 3$ s, the cutting force (light blue) increases from the demanded 8 gf (green) and the voltage to the IPMC (brown) starts to drop in order to reduce the cutting force, as in Figure 13(b), but only manages to reduce the cutting force to 9 gf at $t = 10$ s.

**Discussion**

The strain gauge gives sufficiently accurate feedback and is easy to integrate into the surgical device; however, it is also sensitive to temperature and needs to be compensated if used in a real application. In the cutting experiments it can be seen that the force does overshoot from the desired cutting force even when the voltage applied to the IPMC starts to reduce, and this could be due to the fact that the IPMC is a very low frequency response actuator system. Nevertheless, a simple gain-schedule PI controller has been implemented and shows its potential to be used for force control in an IPMC actuated surgical system. However, there are still many issues to be solved, as in the actual biomedical application, some form of electrical insulation should be considered to prevent the short-circuiting between the two clamping electrodes, how much does the force compliant be reduced with the use of thicker or multiple IPMC to increase the cutting force, and so on.
Conclusions
A one degree of freedom robotic surgical end effector with an embedded force feedback sensor implemented using a simple strain gauge has been designed. The experimental results show reasonable performance of this mechanism. The rotary joint of the system was performed as intended, and the IPMC actuator was used to actuate the surgical tool mechanism.

The embedded force sensing comprised of a stain gauge and cantilever beam, which provides a measurable bending displacement when a cutting force is applied by the IPMC. Feedback measurements from the strain gauge closely match those measured by the load cell, hence confirming the accuracy of the strain sensor as a force sensor.

A gain-scheduled PI controller proved to be sufficient at controlling the cutting force from 2 gf to 8 gf while cutting a replica silicone rubber to mimic human tissue.

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