A compliant surgical robotic instrument with integrated IPMC sensing and actuation

A.J. McDaid, S.Q. Xie and K.C. Aw*

Mechanical Engineering Department, University of Auckland, Auckland 1142, New Zealand
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Robotic assisted surgery is becoming widely adopted by surgeons for a number of reasons, which include improved instrumentation control and dexterity as well as faster patient recovery times and cosmetic advantages. Robotic assisted surgery is currently one of the fastest growing applications in robotics. Although the traditional robotic actuators which are currently used have advanced performance which can, in some aspects, surpass that of humans, they simply do not have the capabilities and diversity required to meet the demand for new applications in robotic surgery. Novel transducers which have advanced capabilities and which allow safe operation in delicate environments are needed. Ionic polymer–metal composites (IPMCs) have extensive desirable characteristics when compared with traditional actuators and as their transduction mechanisms can mimic biological muscle they have much potential for future advanced biomedical and surgical robotics. In this research, a complete two degree-of-freedom (2DOF) surgical robotic instrument has been developed, which with the attachment of surgical tools (scalpel, etc.) has the ability to undertake surgical procedures. The system integrates an IPMC sensor and actuator at each joint. A gain scheduled (GS) controller, which is tuned with an iterative feedback tuning (IFT) algorithm, has been developed to ensure an accurate and adaptive response. The main advantages of this device over traditional devices are the improved safety through a natural compliance of the joints as well as the mechanical simplicity which ensures ease of miniaturisation for minimally invasive surgery (MIS). The components of the system have been tested and shown to have the capabilities required to operate the device for certain surgical procedures, specifically a device work envelope of 1600 mm², compliance of 0.0668 m/N while still maintaining enough force to cut tissue, IPMC sensor accuracy between 3–22% and a control system which has shown to guarantee zero steady state error.

Keywords: ionic polymer–metal composite; surgical robotics; iterative feedback tuning

1. Introduction

1.1. Robotic surgery

Robotic assisted surgery is currently one of the fastest growing applications in robotics. Advances are being made at a rapid pace with much research in laboratories around the world, while there are also already a number of commercial devices available in the marketplace which are currently being used in operating rooms for surgical procedures. The most established example is the da Vinci® surgical robot from Intuitive Surgical® with...
more than 1450 devices installed worldwide [1]. Robotic assisted surgery is become 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 kinesthetic feedback to surgeon [3], more safety with active constraints through appropriate control algorithms, reductions in surgeon tremor through low pass filtering as well as faster patient recovery times and cosmetic advantages. As a result of increased accuracy, reliability and competence of robotic devices, surgical tasks which were originally considered far too risky for robots are now becoming common practise.

The most common setup for robotic surgery, for example a minimally invasive surgery (MIS) operation, involves using surgical instruments guided by a traditional serial or parallel robot setup. The main concern with this type of setup is the safety when used on humans. The robotic devices can typically be heavy, bulky and have a very high stiffness (to insure precise positioning). As the surgeon has no direct contact with the instruments and patient, it is difficult to have the intuitive ‘feel’ as with traditional surgical techniques. Research is currently being undertaken to overcome this with complex force sensors and haptic feedback through master–slave type teleoperation setups [4]. These can be complex and expensive as well as take long periods of training for the surgeon. On top of this there is the issue of computational errors or loss of power, which can be catastrophic during a surgery with highly powerful and rigid robotic systems. Currently no commercial device has kinesthetic feedback to the surgeon and thus prototypical force feedback systems are currently available only in research laboratories [4]. A range of other devices for a number of surgical procedures have been developed, including active catheter systems [5], laser devices for cutting tissue [6], ultrasonically activated devices [7] and cable driven robots for endoscopic surgery.

1.2. The need for new transducers technology

Although the traditional robotic actuators which are currently used in robotic surgical devices, such as electro-magnetic drives and hydraulic/pneumatic machines, have all been extensively investigated and have advanced performance which can, in some aspects, surpass that of humans, they simply do not have the capabilities and diversity required to meet the demand for new actuation systems in applications of robotic surgery [8–11]. They are stiff, noncompliant and mechanically complex, which makes scalability and miniaturisation difficult. The major restriction now to the progression of future surgical robotic devices is suitable mechanical transducers which can operate in tandem with the human body.

To truly operate seamlessly with humans, robotic devices must possess similar actuation characteristics to real biological systems. To achieve this it is clear that a totally new approach must be taken when designing biomedical and surgical robotic devices. Bio-inspired transducers, which have similar properties to human tissue and muscle, in particular mechanical compliance, structural simplicity to allow easy scalability, high power-to-weight and power-to-volume ratios, precise and embedded control capabilities, must be developed. New electroactive polymer (EAP) materials are demonstrating promise in this area and are envisaged by many researchers as the way forward for developing safer and more intelligent robotic systems.

Some electronic EAPs, for example dielectric-based elastomer actuators, have demonstrated high forces and properties nearing human capabilities. The main drawback with this class of EAP is the very high voltages required (>1000 V); this makes their integration and use with humans questionable.
1.3. **IPMCs**

Ionic polymer–metal composites (IPMCs) are a type of ionic EAP which have extensive desirable characteristics when compared with traditional actuators and whose actuation mechanisms can mimic biological muscle and so they have much potential for future advanced biomedical and surgical robotics. With the advancement of materials science research, as well as actuator modelling and control, these IPMC transducers are working towards development of fully integrated intelligent robotic devices.

IPMCs are a bending transducer, typically fabricated in sheets and then cut to size for use in a cantilever beam configuration. The IPMC beam consists of a perfluorinated ionic membrane, for example Nafton® from DuPont, which is sandwiched between two thinly coated conducting electrodes of a noble metal, typically Pt or Au, on either side of the polymer. A beam type actuation greater than 90° can be achieved with small applied voltages across the electrodes; typically less than 5 V. Conversely IPMCs also produce a voltage across the electrodes when mechanically deformed. This passive sensing voltage is usually 1–2 orders of magnitude lower than voltages required for actuation [12].

IPMCs have a number of advantages which make them ideal for use in surgical robotics applications, including low mass, thin, compliant and flexible strips, and low actuation voltages with high displacements, and they are essentially biocompatible and implantable in humans [13]. On the other hand, the transduction mechanisms for IPMCs are very complex and still not fully understood which makes controlling them very difficult. There are still a number of barriers to implementing IPMCs in real world devices including back relaxation, hysteresis and slow response times under DC actuation, some of which are tackled in the development of the device and control system proposed in this research.

Previously, there have been a number of biological applications proposed for IPMCs [11,14–16] and even applications for surgical devices, for example active catheters and systems for endoscopic and endovascular surgery [17–20], have been proposed but little has been done to advance them into real world systems. This paper describes the development of a novel surgical robotic device, with integrated IPMC sensors and actuators, which is aimed for use in real surgical procedures like cutting and grasping tissue. The main motivation for developing this device is that it will be inherently compliant, unlike previous robotic surgery devices, and hence safe for operation in the delicate environments encountered during surgical operations.

Although there have been a number of attempts to control IPMCs using different algorithms and a number of models proposed to predict the IPMCs sensing response there is still very limited research into real-world applications which accurately and reliably integrate the sensing and actuation of IPMCs in one embedded device. The contributions of this research include; a new method for integrating IPMC sensing and actuation in a single device; a novel scalable device design which can lead to easy miniaturization for MIS operations; and a device with guaranteed compliance to ensure safety during surgical procedures.

2. **Surgical robotic system**

A two degree-of-freedom (2DOF) serial configured robotic arm which allows precise and compliant control of a surgical instrument, for example forceps or scalpel, has been designed. The device will be actuated by an IPMC at each joint as well as have feedback from an IPMC joint sensor. It is envisaged that this IPMC surgical robot (IPMCSR) will be attached to the end effector of a traditional robotic device which will undertake the
large-scale (coarse) positioning tasks so the IPMCSR can undertake actual operation by performing the fine movements with micro-precision. This configuration has the advantage of coupling the advantages of a traditional robotic device with the new IPMCSR. The primary robot, which may be non-compliant, for example a da Vinci® system or an active catheter such as those described in [5], will permit a large workspace and high dexterity and will allow the final precise operation to be undertaken with the compliant IPMCSR. This couples the accuracy of a traditional device with the safety of the IPMSR. A schematic of the system is presented in Figure 1, which shows the overall surgical system design and the novel IPMCSR. The system design allows a range of poses to be achieved and different tools to be used, as is required for surgical tasks such as making incisions. If different tasks are required then more DOF can be added to give higher dexterity to the device.

The IPMCSR is designed like a skeleton and so the main weight of the device is taken by the rigid arms, hence alleviating the load on the IPMCs. The IPMCs then actuate the system in a controlled manner to carry out some specific desired task. In this way the system takes advantage of the rigid skeleton design, yet still remains passively compliant to ensure safe operation of the system. This is similar to the human body, which utilizes both a rigid skeletal system with flexible muscle actuators.

The two IPMCs at each joint will be sandwiched together but electrically insulated from each other using a thin plastic film. The base will be clamped and the tips will be located in a slot on the robot arm to ensure that the IPMCs can slide, but the actuator can still apply a torque and the sensor bend is always consistent (Figure 1c). This gives the device the ability for closed-loop position control as well as ensuring the system has a natural passive compliance. With the inclusion of force/torque sensors the device also has the ability to control the active compliance of the device through a suitable control algorithm. As IPMCs are material based actuators as oppose to a mechanism based actuator such as an electromechanical motor, they can easily be scaled down and hence miniaturised for use in MIS or endoscopic operations.

The workspace IPMCSR is dependant on the geometry of the arms of the device and on the maximum rotating angle of the joints. The length of the arms can be adjusted depending on the operation required. The maximum angle for the joints is restricted by

Figure 1. Schematic drawings of (a) the overall system level design for the proposed robotic surgery device, (b) the IPMCSR and (c) configuration of the IPMC sensor and actuator joint when no voltage is applied (top) and when actuated (bottom).
the IPMC bending. This is highly dependent on the geometry of the IPMC, especially thickness. As such a combined model for the robotic device and IPMC is needed. This must include the robotic parameters as well as a geometrically scalable IPMC model such as that described in [21,22].

A schematic of the IPMCSR is shown in Figure 2 with all the relevant robot coordinates systems, as defined by the standard Denavit–Hartenberg notation, as well as the robot geometric parameters. $L_{IPMC1}$ and $L_{IPMC2}$ give the distances from the pin joints to the IPMC actuator tip which contacts and applies a force to drive the robot arm.

A mathematical model was developed for the robotic manipulator. The model, in joint space representation, is given by:

$$M(\theta) \ddot{\theta} + C(\theta, \dot{\theta}) + G(\theta) = \tau_{IPMC} - \tau_{frict},$$  

where $M$ is the mass matrix, $C$ is the vector of centrifugal and Coriolis torques, $G$ is the vector of gravitational torques. These effects are balanced by the torque input by the IPMC, $\tau_{IPMC}$, minus the joint friction torques, $\tau_{frict}$; $\ddot{\theta}, \dot{\theta}$and $\theta$ are all vectors made up of the joints angular accelerations, velocities and positions, respectively. The model has been developed to fully describe the mechanical dynamics of the mechanism, including inertia, Coriolis effect, gravity and joint friction of the manipulator so accurate simulations can be undertaken.

The IPMC model in [21] has been used to analyse the performance of the system. As the IPMC model and robot dynamic model are both scalable the system model is extremely useful for design optimisation of the system performance for different applications. The model can also be used for predicting the performance of the device if it is to be scaled down and miniaturised for MIS operations.

### 3. IPMC actuation

IPMCs are extremely useful actuators for developing intrinsically compliant devices. They have a mechanical output in the form of a bending response, which can easily be converted to a rotational motion by the robot configuration described in the previous section. IPMCs...
geometry can be tailored to many requirements of force and displacement. Using the scalable mechanical design based IPMC model developed in [21], a suitable sized actuator to give the required force and displacement can be selected. The force/displacement of the IPMCSR will depend not only on the IPMC force and displacement, but also on the length of each of the robot links. Essentially the links act as levers and so the longer the link the higher the displacement, but less the available force.

One very important consideration which must be taken into account is the available force output of the IPMC at varying displacements. As the IPMCs bend further from the equilibrium the available force output reduces. One way the available force of an IPMC may be increased, which is proposed in [21] is by pre-bending the IPMC which will take advantage of the passive stiffness plus the electrically induced force.

Implementing a system with IPMC actuators presents a unique feature into a robotic system in that as well as an active compliance which can be implemented with force feedback and a suitable control algorithm (as in traditional robot systems), each joint of the system will have a natural passive compliance due to the flexibility of the polymer actuator. This is extremely useful for systems where safety is an extremely important factor as in surgical robotics. IPMCs have been shown to be able to be controlled precisely and robustly when activated with a power source and due to the passive compliance are safe if power is lost. This is a very useful feature over traditional robotic surgery devices where if the power is surged or lost the robot behaviour can be potentially fatal for a patient.

The passive compliance of an IPMC is highly dependent on its geometry, in particular its thickness and the polymer membrane material. The equation relating the IPMC geometry and material properties to the stiffness or compliance is given below in Equation (2). Using this relationship the IPMC actuator can be sized for the particular application needed. The passive compliance of a beam type IPMC manipulator (using the standard Euler–Bernoulli equation), which is the type concentrated on in this research, is defined as

$$S_{\text{manip}} = \frac{1}{k_{\text{manip}}} = \frac{L^3}{3EI},$$

where $s_{\text{manip}}$ is the mechanical compliance and $k_{\text{manip}}$ is the stiffness of the manipulator. The compliance depends on the IPMC properties, free length, $L$, modulus of elasticity, $E$ and moment of inertia, $I$. This is very useful in designing a manipulator for a specific application as the IPMC dimensions and material properties can be tailored to meet the manipulator requirements for force, displacement and compliance. Passively compliant IPMC manipulators, as used in this research, do have advantages over an actively controlled compliant system including guaranteed compliance throughout operation, no need for complex control algorithms, safety achieved without loss of positioning performance and no expensive force/torque feedback sensors are needed.

4. IPMC sensing for feedback

IPMCs sensing characteristics have been widely reported by a number of researchers. Work has been undertaken to explain the fundamental sensing characteristics as well as developing models to predict their sensing behaviour [23,24]. A number of methods for IPMC sensing have been proposed including measuring the resistance change of the electrodes [25], the charge in the IPMC itself [26] and making hybrid structures with other materials for example PVDF [27]. Although much work has been done in IPMC
sensing, like actuation phenomena, there is still no widely accepted model for the sensing properties of IPMCs.

In this research we have employed a simple sensing model, which was first reported in [28], to predict the motion of a rotary joint using experimental results. In this way the model can be used to decode an electrical signal from the IPMC sensor to give a predicted angular displacement. As there is still no standardised manufacturing technique for IPMCs their behaviour is still highly sample dependant and so this model has the major advantage of ensuring accuracy for the specific IPMC sensor used. The down side to this is the model parameters must be tuned individually for each sensor, although as the model structure is relatively simple the tuning process is not very labour intensive.

The sensor mode used in this research is based on the charge generated by the IPMC itself and thus has the advantage of being a passive device and so no external power source is needed. The IPMC is to be used as a rotary sensor and is placed in a testing apparatus, as shown in Figure 3. The most notable characteristics of the voltage response due to bending are the initial spike followed by a voltage recovery or decay to a steady state value. The polarity of the voltage spike corresponds to the bending direction. In the development of this model it was assumed, based on observations that the time \( t_{peak} \) to reach the voltage peak \( V_{peak} \) is equal to the time the rotary joint is actually bending. Taking this fact the bending speed of the joint, in \(^\circ\) s\(^{-1}\), is calculated by:

\[
\dot{\theta} = \frac{\theta}{t_{peak}},
\]

where \( \theta \) is the joint bending angle in degrees.

In order to calibrate a relationship for the bending response a number of experiments were undertaken, first with a constant bending angle and varying speeds and then with a constant speed and varying angles. The two variables, speed and displacement were adjusted separately in order to eliminate any cross coupling between the variables. The bending speeds in the experiments were kept constant, and so the acceleration is assumed to occur over a very short period of time. As the time constant of the IPMC is slower than the time for acceleration the effect is assume negligible and so the model does not consider any effects of acceleration.

Figure 3. IPMC rotary joint test rig and corresponding sensing voltage response [26].
The results of the tests using a 22 mm long, 8 mm wide IPMC with a thickness of 400 µm, with a constant bend radius (ensured through the test rig) are shown in Figure 4. Note the raw voltage signal was of the order of 1–2 mV and this has been amplified and filtered through sensor interface circuits.

It can be seen from Figure 4a that with varying joint speed (with a set displacement of 86.4°) there is a clear relationship between the peak voltages. From Figure 4b it can be seen that with the set speed of 450° s⁻¹ different travel angles produce different peak voltages, but the shape of the voltage spike is the same. As a next step to extracting a useful correlation for the model the data for different speeds in Figure 4a was normalised with respect to the peak voltage for each speed and plotted against the joint bending angle; see Figure 5a. From this it can be seen that bending angle relationships are consistent for different speeds. Previous researchers have proposed that the bending angle is proportional to the voltage peak [29], although from the results observed here it is clear this relationship is more complex than linear. To account for this a simple method of creating a piecewise
A linear function consisting of 4 distinct regions has been developed, as shown in Figure 5a, which very accurately represents the mean data for different speeds for displacements up to 86.4°. Utilising this relationship with the data from Figure 4b for varying displacements a region plot can be formed, as shown in Figure 5b, which determines which of the four linear piecewise functions to use depending on the location of the voltage peak. From this the bending displacement and speed can be found. Full data analysis for this sensor model can be found in the thesis by van den Kurk in [30].

5. Control system

As IPMC actuators exhibit highly nonlinear and time varying behaviour most traditional control schemes will not be able to handle the actuation response for high displacements over a long period of time. As such, for this research an adaptive controller, tuned using an iterative feedback tuning (IFT) technique for varying target displacements, was employed. This control scheme was proposed in [31] and has been shown to be extremely useful for controlling high IPMC displacements due to their nonlinear nature.

The basic control structure has its origins in simple linear control, as per Figure 6, where $G_c$ is a PID controller, but has a gain schedule, $f_{GS}$ to adapt the controller parameters, $\rho$, as a function of the reference input, $r$, in order to give the system a good response over a large operating range. The schedule is developed by optimally tuning the system for a number of target inputs, using the adaptive IFT algorithm, then interpolating the optimal gains to get a relationship for adjusting the control parameters based on the reference input.

The IFT algorithm operates by seeking to optimise an objective function, $J$, which is based on the system performance, in this case a least squares fit of the tracking error, $e(t,p)$ as per Equation (4), in which $N$ is the number of samples in a given experiment:

$$J(\rho) = \frac{1}{2N} \sum_{t=1}^{N} (e(t,p))^2$$

The optimal is achieved by calculating the differential of the objective function, $\frac{\partial J(\rho)}{\partial \rho}$ and equating this to zero. Using $\frac{\partial J(\rho)}{\partial \rho}$ a gradient search is used to find the optimal parameters. $\frac{\partial J(\rho)}{\partial \rho}$ is found through experiments on the actual IPMC system and so the IFT algorithm is model free and can be implementing online as an adaptive tuning scheme as shown in [32]. This makes the IFT algorithm ideal for tuning the response of the highly complex IPMC system as no knowledge of the system is needed prior to operation; the algorithm simply seeks to optimize the PID gains for the system.

![Figure 6. Control diagram of the GS nonlinear controller.](image-url)
6. Results and discussion

The IPMC transducer materials used in this research were obtained from Environmental Robots Inc [13]. They were fabricated in sheet form and then cut to the desired geometries for the application. The results and discussions for the elements in the proposed robotic system are outlined below.

6.1. System design and compliance

An overview of the system is shown schematically in Figure 7, where \( r \) is the reference signal, \( u \) is the controller \( (G_c) \) output, \( f_{GS} \) is the gain scheduler, \( \tau \) is the torque applied by the IPMC actuator onto the robotic device, \( \theta, \dot{\theta}, \ddot{\theta} \) are the device angular displacements, velocities and accelerations respectively, \( v \) is the electrical potential output from the IPMC sensor and \( \tilde{\theta} \) is the predicted angular displacement feedback from the decoder.

The design of the system has been chosen with arm lengths of 100 mm, and with IPMC actuators of 30 mm long (free deflection) by 10 mm wide. The above configuration will result in a nominal maximum deflection of approximately 11° and a work envelope of 1600 mm². There is little definitive literature on the forces required to cut human tissue as there are many variables including the sharpness of the scalpel blade, type of tissue speed of cut etc, but based on the experiments undertaken on soft tissue samples in [33] the IPMC device will be able to successfully carry out cutting tasks.

In each joint of the system there are two IPMCs, one for sensing and one for actuation. The passive IPMC sensor will add some resistance to the device and hence increase the compliance of the joint. In this way the IPMC sensor geometry needs to be chosen with dimensions which will ensure adequate sensing response but not add too much stiffness to the system. The compliance of each joint is defined in Equation (5). The IPMCs have been chosen with dimensions of \( 35 \times 10 \times 0.8 \) mm for the actuator and \( 22 \times 8 \times 0.2 \) mm for the sensor (both 5 mm clamped section), this results in the following compliance of the system. This compliance relates a deflection of 3.3 mm with a force of 50 mN:

\[
S_{\text{manip}} = \frac{1}{k_{\text{act}} + k_{\text{sensor}}} = 0.0668 \text{m/N} \tag{5}
\]

It has been determined from simulations that the IPMC can in fact operate the surgical device and so real world experiments were then undertaken to prove the system does work. In order to test the feasibility of the entire system a systematic approach was taken to test

![Figure 7. Schematic drawing of the proposed IPMCSR system.](image-url)
the system by ensuring the performance of each component separately. In this way the capabilities of each part will be ensured individually which is useful to help troubleshoot any issues in the system. This was done first as IPMCs exhibit highly complex behaviour in real world situations and hence testing the entire system at once and then attempting to troubleshoot is likely to become problematic.

6.2. Actuators

In order to determine the actuators operating performance in the surgical robotic device the joint design was tested on a single degree-of-freedom joint first. The test rig for the actuators is shown in Figure 8. This system will represent the final joint design and is used to test performance.

A 35 mm long, 10 mm wide IPMC with an average thickness of 800 µm was used to test the device. The open loop results are shown in Figure 9. It can be seen that the IPMC can indeed move the rotary joint as required and the displacement is acceptable with a low voltage of 3 V.

![Figure 8. The single DOF test rig to test actuator operating performance of the system.](image)

![Figure 9. Open loop experimentation of the IPMC actuators for a 1DOF joint.](image)
6.3. Sensors

In order to test the sensor decoder an algorithm was developed to convert the sensor voltage into a predicted joint motion. This involved determining the start and end of a specific voltage spike or peak and extracting the relevant variables to determine the joint velocity and displacement. To verify the decoder algorithm thirty experiments were conducted which consisted of varying speeds and varying bending angles. The raw data, i.e. the induced electrical potential from the IPMC, was input to the decoder routine and the results are shown in Table 1.

The results in Table 1 show that the error is less than 3% in all bending angles except at low deflections. Some of these errors may be put down to the simplification of the voltage response as well as assuming that the peak corresponds exactly to the time of bending, as per Equation (3). Also, at the small deflections the value for $V_{peak}$ and $t_{peak}$ are much smaller and so errors in calculating this may result in larger corresponding deflection predictions. Other contributing factors may be that at low deflections the electrical noise can become a problem as the signal to noise ratio is decreased and also no allowance was made for the acceleration of the joint in the decoder. This will have a larger influence on the results at smaller deflections.

Despite the errors at smaller deflections the IPMC sensor has demonstrated its ability to reasonably accurately predict the bending displacement of a rotary joint which is what is needed for the IPMCSR system.

6.4. Control system

In order to validate the performance of the control system testing was first undertaken on an IPMC actuator with displacement feedback from a traditional laser sensor with a 10 µm resolution. This conventional sensor was used so the control system can be tested independent of sensor feedback.

First the IFT algorithm was used to tune the PID gains for a number of different displacements in order to optimise the controller for a large range of target displacements. This was done successfully and so then a relationship was found between the optimally tuned gains and the target displacement:

$$K_p = -839.7 \ln(r) - 3757.6,$$

$$K_i = -808 \ln(r) - 3560.2,$$

Table 1. Actual and decoded bending angle for varying joint speeds of the IPMC sensor.

| Actual bending angle (°) | Decoded bending angle at various joint speeds (° s⁻¹) | Mean error (%) |
|-------------------------|-----------------------------------------------------|----------------|
| 120                     | 6.72, 6.72, 6.52, 6.40, 6.49                        | 21.67          |
| 150                     | 17.78, 17.54, 17.51, 17.86, 17.10                    | 2.55           |
| 200                     | 29.53, 29.52, 29.89, 30.03, 29.51                    | 2.97           |
| 257                     | 48.96, 49.23, 49.15, 49.58, 49.25                    | 2.32           |
| 360                     | 62.00, 61.53, 62.48, 62.63, 61.99                    | 1.38           |
| 50.4                    | 76.01, 76.42, 77.04, 76.67, 76.66                    | 0.83           |
Figure 10. Random stair-step response for an IPMC actuator for PID controlled and gain schedule controlled.

\[ K_d = 84.73 \ln(r) + 1038.1. \]  

(8)

The relationships in Equations (6)–(8) were then used for \( f_{GS} \) in the GS controller. The full results can be found in [31].

Once the controller was tuned a random stair-step input was entered as the target for the IPMC to track and the comparison of an optimally tuned PID controller and an optimally tuned IFT controller are shown in Figure 10, where the dotted line represents the target reference displacement. It can be seen that in all cases the GS controller outperformed the linear PID controller. This demonstrates that the developed nonlinear controller is better for controlling the IPMC over a range of displacements.

Once the GS controller had been verified, the IFT algorithm was implemented on the rotary mechanism which is driven by the IPMC, as shown in Figure 8. The results presented in Figure 11 show that the IFT algorithm can iteratively tune the response of the mechanism to track a target (dotted line) from an over-damped, an under-damped or even a manually tuned response. This has verified the use of the IFT algorithm with a mechanism which has real world phenomena, e.g. friction and gravitational effects.

Note, however, that these experiments were undertaken on the mechanism as in Figure 8 and so the IPMC did not have to work against the gravity of the mechanism itself. This replicates the first DOF in the IPMCSR system, which mainly has to act against friction in the joint. The second DOF has to act against gravity so the performance may be different, however we predict that the IPMC will still be able to handle this as is seen in [21] even when some pre bend due to a load is applied on the IPMC the dynamic characteristics due to the electrically induced force remain the same i.e. if the IPMC is pre-loaded it should still have the same dynamic characteristics but simply operate around a different equilibrium point.

7. Conclusions and future work

In summary, a surgical device has been proposed and it has been designed to be able to carry out certain operations with accuracy as well as failsafe compliance, not seen in traditional robotic systems. All of the components in the system have been tested and have been proven to have the necessary performance. The compliance of the designed system
ensures its safety and the geometry of the robotic arms ensures that it has enough force to carry out a number of operations including cutting and moving tissue.

All of the components of the IPMCSR system have been shown to work in isolation but the integrated system must be implemented and tested to ensure the integrated system performance. Future work on IPMC sensing will include the coupling effect of speed and displacement as well as the effect of acceleration should be examined further to make the model more accurate and robust. Experiments should be carried out using real biological tissue in a number of scenarios to ensure its performance is stable. Also the system can be integrated with a traditional positioning system for robotic surgery, for example the da Vinci® robot.

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