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ACHIEVING DEXTEROUS MANIPULATION FOR MINIMALLY INVASIVE SURGICAL ROBOTS THROUGH THE USE OF HYDRAULICS

Devin R. Berg*

Department of Mechanical Engineering
 University of Minnesota - Twin Cities
 Minneapolis, Minnesota 55455
 Email: drberg@umn.edu

Perry Y. Li

Arthur G. Erdman

Department of Mechanical Engineering
 University of Minnesota - Twin Cities
 Minneapolis, Minnesota 55455

ABSTRACT

Existing robotic surgical platforms face limitations which include the balance between the scale of the robot and its capability in terms of range of motion, load capacity, and tool manipulation. These limitations can be overcome by taking advantage of fluid power as an enabling technology with its inherent power density and controllability. As a proof-of-concept for this approach, we are pursuing the design of a novel, dexterous robotic surgical tool targeted towards transgastric natural orifice surgery.

The design for this hydraulic surgical platform and the corresponding analysis are presented to demonstrate the theoretical system performance in terms of tool positioning and input requirements. The design involves a combination of a novel 3D valve, hydraulic artificial muscles, and multi-segmented flexible manipulator arms that fit in the lumen of an endoscope. A dynamic model of the system is created. Numerical simulations show that a hydraulic endoscopic surgical robot can produce the desired performance without using large external manipulators such as those employed by conventional surgical robots. They also provide insight into the component interactions and input response of the system. Future work will include manufacturing a prototype to validate the concept and the numerical models.

NOMENCLATURE

A_0 The orifice area of the control valve inlet orifice.
 A_f The curtain area at the controlled valve outlet.

b The length of each strand within the braided sheath of an artificial muscle actuator.
 C_{d0} The discharge coefficient of the control valve inlet orifice.
 C_{df} The discharge coefficient of the valve outlet.
 D The diameter of the artificial muscle actuator.
 D_N The fixed diameter of the valve outlet.
 E Elastic modulus of the manipulator backbone material.
 F_a The contractile force produced by the artificial muscle actuator.
 F_b Force load at the distal end of the manipulator due to beam deformation.
 F_L External force load applied at the distal end of the manipulator.
 F_x Total load in the transverse direction at the distal end of the manipulator.
 F_z Total load in the longitudinal direction at the distal end of the manipulator.
 g The gap between the orifice face and the flapper within the control valve.
 γ The angle of each strand of the braided sheath measured relative to the longitudinal axis of the actuator.
 I Second moment of area of the manipulator backbone.
 I_b Moment of inertia at the distal end of the manipulator.
 L The length of the manipulator backbone.
 L_a The overall length of the artificial muscle actuator.
 M Total moment at the distal end of the manipulator.
 m Mass at the distal end of the manipulator.
 M_b Internal moment of the beam.

*Address all correspondence to this author.

- M_L External moment load applied at the distal end of the manipulator.
- n The number of turns completed by one strand within the braided sheath of an artificial muscle actuator.
- P_a The internal pressure of the artificial muscle actuator measured relative to the external pressure.
- PE Potential energy within the beam for any arbitrary beam shape.
- P_s The supply pressure provided to the control valve.
- Q_a Flow rate of fluid entering or exiting the actuator.
- Q_1 Flow rate of fluid entering the control valve.
- Q_2 Flow rate of fluid passing through the controlled flow restriction and exiting the control valve.
- r Radius of the routing discs on the manipulator backbone.
- ρ Density of the fluid.
- s Position coordinate along the manipulator backbone. Varies from 0 to L .
- θ Angular orientation of the beam cross-section.
- V The internal volume of the artificial muscle actuator.
- x Beam position in the transverse direction.
- z Beam position in the longitudinal direction.

INTRODUCTION

Robots have been used in surgery since approximately 1985 when Shao et al. used a robot designed for industrial applications to perform a neurosurgical biopsy [1, 2]. Since that time, the field of surgical robotics has undergone many changes as the procedures performed with the aid of robots have become less invasive and more complex. Current developments in surgical practice are leaning increasingly towards endoscopic tools and techniques [3]. For surgical robots to adapt to these developments, it is necessary to overcome significant challenges in terms of tool manipulability and precision. Most systems rely on a large external robot manipulating cable driven tools. Due to hydraulics' advantages in power density and relative incompressibility [4], hydraulics may allow surgical robotics to be embedded endoscopically. Yet, the use of hydraulics in this field has been under-explored due to difficulties associated with the integration of hydraulic components into the distal end of the robotic manipulator due to size constraints [5].

For hydraulic actuation to be employed advantageously in an intuitive robotic minimally invasive surgical platform, some barriers need to be overcome: small size, realization of multiple degrees of freedom, and achieving the range of force and motions necessary to perform procedures effectively. To take advantage of the existing infrastructure of endoscopic devices, the new surgical tools must be constrained to an overall outer diameter of 5-6 mm based upon existing endoscopic tool channel sizes. Fluid power components of this scale are not available

commercially. Work is currently in progress exploring the use of hydraulic fluid power at the meso-scale (mm to cm) for use in prosthetics [6] and at the micro-scale (μm to mm) for which the research is ongoing in the field known as microfluidics [7]. However, as pointed out by Love et al., there exists a gap between conventional fluidics and microfluidics that includes the type of high pressure-low flow rate fluid power components that would be necessary for the application of hydraulic power to the field of surgical robotics [6].

In addition to the size constraints placed upon the surgical device, it is also necessary to consider the force requirements of typical procedures encountered in the operating room. Preliminary evaluations of these constraints based upon the measurement of force requirements for organ manipulation and suturing tasks have revealed that a tool load capacity in the range of five Newtons is sufficient for the majority of common usage cases [8]. Further, it is desirable that the robotic manipulator be able to achieve full retroflexion to enable the level of dexterity necessary for complex surgical procedures.

The robotic surgical platform proposed here is being designed to meet these requirements while maintaining a smaller footprint than conventional surgical robots. By meeting these requirements through the use of hydraulics, it is possible to develop a more capable endoscopic surgical robot which can be packaged in a compact way. This technology would allow the robotic manipulator to exhibit greater freedom of motion, have better controllability, be more cost effective, and have superior ease of mobility. Hydraulic actuation typically displays greater stiffness and faster response time than the comparable electromechanical or pneumatic counterparts. When compared with a pneumatic approach, hydraulic power provides less susceptibility to leakage and is a safer choice if leakage were to occur. Thus hydraulic power can be safely operated at a higher pressure than a similar pneumatic system. Further, if a compatible hydraulic fluid is selected, such as saline, leakage is not likely to result in contamination of the abdominal cavity. Lastly, the low costs of the components used in this design would permit the device to be one time use and thus loosen requirements for sterilization.

Similar work in this area has explored the use of continuum manipulators ranging from pre-curved tubes or electromechanical, tendon-driven manipulators for surgical applications to pneumatic McKibben actuators for large-scale object manipulation [9–11]. One particularly interesting example is the combination of pneumatic actuation with a precurved concentric tube manipulator for surgical treatment of epilepsy [12]. This work demonstrates the feasibility of a fluid power approach; however, the robotic control unit is too large for internal abdominal access with an endoscope. Additionally, there is a long history of hyperredundant robots consisting of a multitude of rigid links providing similar yet distinct operating characteristics [13, 14].

Our proposed design builds from these similar solutions in order to provide improved performance and miniaturization for surgical robotics. In particular, we propose a combination of a novel 3D valve, hydraulic (e.g. saline) actuated McKibben muscles, and flexible manipulators to take advantage of the possible high power density and low compressibility of hydraulics.

The remainder of this paper first describes the proposed device design including particular details of the arrangement of the device components and an overview of the novel control valve design. Next, analysis of each of the components is presented as well as a description of the theoretical behavior of the device based upon this analysis. Finally, some conclusions are discussed as well as a proposal for future work.

PLATFORM DESIGN AND ANALYSIS

The proposed surgical robotic platform consists of three basic units: manipulators, actuators, and control valves. All components are designed to be located at the distal end of the endoscope and contained within its working channel whose diameter is $\sim 6\text{mm}$. Figure 1 shows a simplified version with two manipulators. Each manipulator can be divided into multiple serially connected segments with a single flexible backbone and is actuated by a set of tendons/cables, similar to other serpentine like robots [15–17]. The bending of each manipulator segment is controlled by a group of three tendons antagonistically operated by hydraulic artificial muscle actuators at the base of the platform. The tendons, which are arranged circumferentially, pass through intermediate discs and terminate at the end of the given segment. The tensions in the tendons control the bending of the segment in 2-degrees of freedom. Unlike cable driven surgical tools that are actuated via long cables by external motors or actuators, here, the tendons are kept relatively short since the muscle actuators in our design are small enough to be stacked serially and placed distally within the lumen of the endoscope. Hydraulic flow/pressure to the muscle actuators are modulated by valves which are also contained within the endoscope behind the muscle actuators (Fig. 2). Thus the only connections required externally are lines for the hydraulic supply (assumed to be a constant pressure supply) and return, and electronic signal wires.

In the following sections, the individual system components are presented and modeled. They are then assembled into a system model. The model can be simulated to evaluate the design. For simplicity of presentation, in the rest of the paper, instead of the general three dimensional situation, we consider only the planar case in which each manipulator segment has one DOF of bending and requires only two tendons and actuators for each segment. A similar methodology can be easily extended to the 3D case.

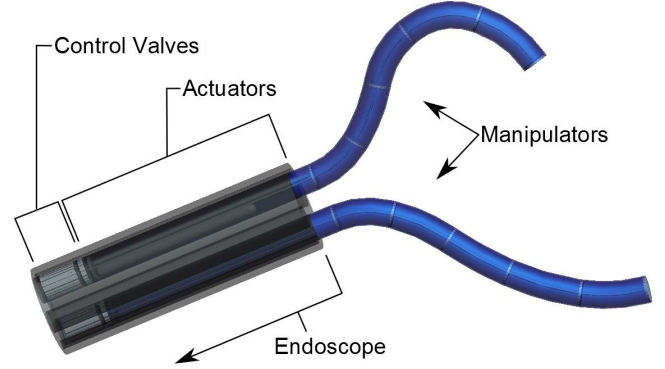


Figure 1. MODEL OF THE ROBOTIC SURGICAL TOOL DESIGN.

Backbone Mechanics

The use of a continuous backbone for cable driven robots has been well documented in the literature [10, 14–17]. Here, the backbone of each manipulator segment is modeled as an Euler-Bernoulli beam which is cantilevered at the base. Assuming that the position and orientation of the distal end is given, we wish to determine the shape and the beam reaction force and moment acting on the end-effector at the distal end of the beam. We assume that the number of discs along the actuator tendons are sufficiently large so that each tendon can be considered to be parallel to the beam at a distance r . This way, the tendons present loading on the beam only via the terminal disc but not via the intermediate discs.

One method of modeling the deformation of a robot with a continuous backbone is to use an energy formulation [18]. The basis of this method is that of all the shapes that satisfy the end point configuration, the continuous backbone with negligible mass will assume the shape with the least potential energy. Knowing this shape for the specified location and orientation of the distal end of the beam, it is then possible to calculate the resulting load applied at the distal end of the beam. This inverse kinematics model assumes planar bending and neglects the effects of the mass of the beam itself [15, 17]. Further, it is assumed that the beam stiffness is such that only bending occurs and the beam does not shear or stretch.

Let $s \in [0, L]$ denote the distance along the beam of length L , and $\theta(s)$ be the angular orientation at location s . $\theta(\cdot)$ is described by a smooth parameterized curve with sufficient degrees of freedom. The potential energy stored within the beam can be expressed as

$$PE(\theta(s)) = \frac{1}{2} \int_0^L EI \theta'(s)^2 ds \quad (1)$$

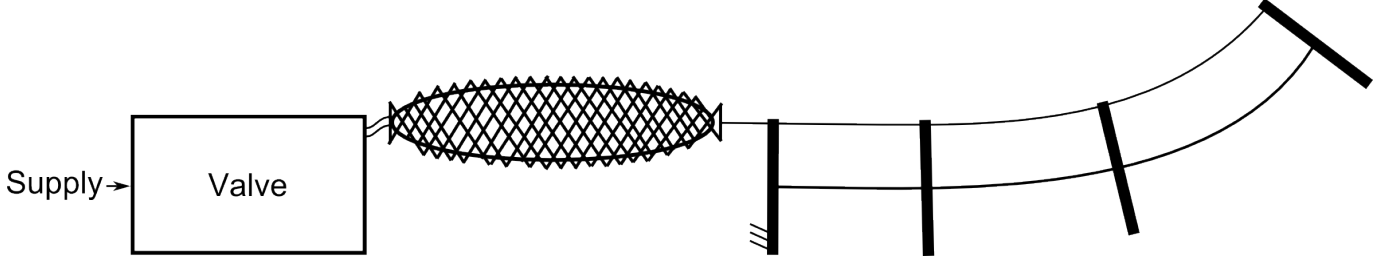


Figure 2. LINE DRAWING SHOWING HOW THE VALVE, ACTUATOR, AND TENDON COMPONENTS ARE CONNECTED.

where I is the second moment of area, E is the elastic modulus, and L is the length of the beam segment. The beam shape is specified as a n -degree polynomial parameterization of $\theta(s)$. Assuming that the location and orientation of the distal end of the beam are specified as inputs such that x_{end} , z_{end} , and θ_{end} are known. As axial and transverse strain are neglected, $x(s)$ and $z(s)$ can be specified as functions of $\theta(s)$ such that

$$x(s) = \int_0^s \cos(\theta(s)) ds, \quad s \in [0, L] \quad (2)$$

$$z(s) = \int_0^s \sin(\theta(s)) ds, \quad s \in [0, L] \quad (3)$$

Beam shape is then given by the constrained optimization of the potential energy of the beam subject to boundary conditions at $s = L$,

$$x(L) = x_{end}, \quad z(L) = z_{end}, \quad \theta(L) = \theta_{end} \quad (4)$$

and at $s = 0$

$$\theta(0) = 0 \quad (5)$$

An example of the output from this model is shown in Fig. 3. Inputs to the model were 30 mm for both the tangential and longitudinal tip positions and an angle of 10° for the tip orientation. Here, a high order polynomial is used to parameterize $\theta(s)$. The optimization then produces a result that meets those end conditions and minimizes the strain energy in the beam.

Once the potential energy of the beam has been minimized using an optimization algorithm, the beam shape is known and the end load can be calculated. This is done by approximating the internal moment of the beam as a function of $\theta(s)$ as

$$M_b(s) = EI\theta'(s) \quad (6)$$

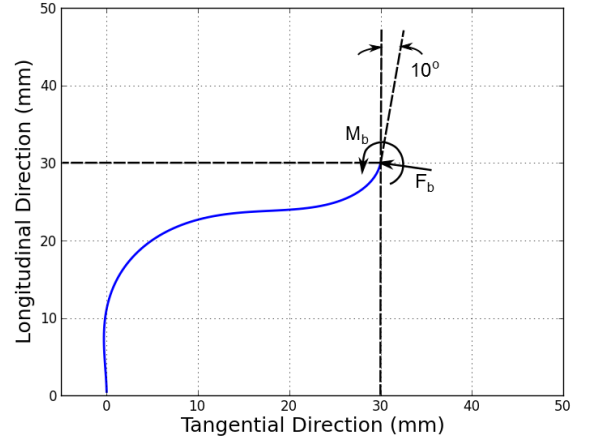


Figure 3. EXAMPLE BACKBONE MODEL OUTPUT FOR ARBITRARY INPUT PARAMETERS.

Then the moment load at the distal end of the beam is just Eq. (6) evaluated at $s = L$. By taking moments at two locations on the beam (e.g. $s = 0$ and $s = L/2$), the resulting force ($F_{b,x}, F_{b,z}$) can be calculated as

$$\begin{bmatrix} M_b(0) - M_b(L) \\ M_b(s) - M_b(L) \end{bmatrix} = \begin{bmatrix} -(z(L) - z(0)) & x(L) - x(0) \\ -(z(L) - z(s)) & x(L) - x(s) \end{bmatrix} \begin{bmatrix} F_{b,x} \\ F_{b,z} \end{bmatrix} \quad (7)$$

Alternately, $(F_{b,x}, F_{b,z}, M_b(L))$ can be computed from the Lagrange multipliers corresponding to the constraints in Eqs.(4).

Artificial Muscle Actuator Modeling

Although hydraulic actuation offers high power density, mechanical rigidity and high dynamic response, the force capability of a conventional linear hydraulic actuator is limited by the piston area, $F = P * A$. In view of the stringent diametric requirement of the present application, we explore the use of hydraulic artificial muscles [19] which can be shown to have better peak

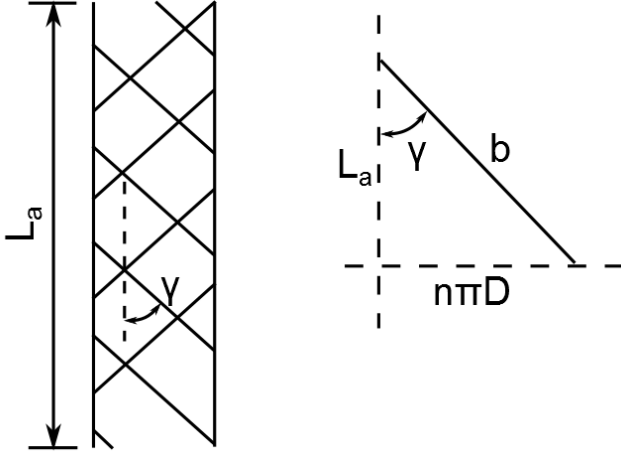


Figure 4. ARTIFICIAL MUSCLE ACTUATOR GEOMETRY (LEFT) AND MAGNIFIED VIEW OF THE BRAID GEOMETRY (RIGHT).

force capability than conventional linear actuators for the same pressure and diameter.

Artificial muscle actuators consist of a contained internal bladder, surrounded by a flexible, braided outer sheath. It is the geometry of this outer sheath that transmits the radial expansion of the internal bladder due to applied pressure to contractile force along the longitudinal axis of the muscle actuator rather than elongation as in a traditional hydraulic actuator. The load capacity of the artificial muscle actuator is then a function of the geometry and orientation of the outer sheath and the pressure applied to the internal bladder. This makes it possible to achieve greater peak load capacity for the same cross sectional area [20].

Typically, artificial muscles are actuated pneumatically with compressed gas [11, 12, 21, 22]. In our approach, the artificial muscles are actuated hydraulically with a pressurized liquid. There are two important advantages: 1) a higher pressure can be applied so that force and power density can be further increased and the actuator diameter can be decreased; 2) liquid has a much lower compressibility and therefore better rigidity than compressed gas.

As shown in Fig. 4, adapted from Chou and Hannaford, it is first necessary to define the geometric parameters of the braided sheath [20]. These parameters include the length of the individual braided strands, b , the number of turns each strand makes over the length of the actuator, n , and the angle between the strands and the longitudinal axis of the actuator, $\gamma(t)$.

The overall length of the actuator, $L_a(t)$, and the actuator diameter, $D(t)$, can then be represented in terms of the constants, n and b , and as functions of the variable $\gamma(t)$, as seen in Eq. (8)

and Eq. (9).

$$L_a(t) = b \cdot \cos \gamma(t) \quad (8)$$

$$D(t) = \frac{b \cdot \sin \gamma(t)}{n\pi} \quad (9)$$

Then, calculating the volume of a cylinder and substituting in the functions for L_a and D ,

$$V(t) = \frac{\pi D(t)^2 L_a(t)}{4} = \frac{b^3}{4\pi n^2} \cdot \sin^2 \gamma(t) \cos \gamma(t). \quad (10)$$

The first derivatives of $L_a(t)$ and $V(t)$ with respect to $\gamma(t)$ are calculated as

$$\frac{dL_a(t)}{d\gamma(t)} = -b \cdot \sin \gamma(t) \quad (11)$$

$$\frac{dV(t)}{d\gamma(t)} = \frac{b^3 \sin \gamma(t)}{2\pi n^2} \cdot (\cos^2 \gamma(t) - \frac{1}{2} \sin^2 \gamma(t)) \quad (12)$$

From Eq. (11) and Eq. (12), the first derivative of $V(t)$ with respect to $L_a(t)$ is given as

$$\frac{dV(t)}{dL_a(t)} = \frac{dV(t)/d\gamma(t)}{dL_a(t)/d\gamma(t)} = \frac{-b^2}{4\pi n^2} \cdot (3 \cos^2 \gamma(t) - 1). \quad (13)$$

From the principle of virtual work we have:

$$F_a(t) \cdot dL_a(t) = P_a(t) \cdot dV(t) \quad (14)$$

and solving Eq. (14) for the force output with Eq. (8 and 13) results in

$$F_a(t) = P_a(t) \frac{b^2}{4\pi n^2} \cdot \left(3 \frac{L_a(t)^2}{b^2} - 1 \right) \quad (15)$$

$$= P_a(t) \frac{\pi D(t)^2}{4} \cdot \left[\frac{3 \cos^2(\gamma(t)) - 1}{\sin^2(\gamma(t))} \right] \quad (16)$$

where F_a is the contractile force and P_a is the pressure differential across the bladder wall. Because the term within the brackets in

Eq. (16) can be greater than 1, the force capability of an artificial muscle can be greater than that of a hydraulic piston actuator for the same area and pressure. Notice, however, that this advantage comes with the price that the force/pressure relationship varies with $\gamma(t)$ or actuator length $L_a(t)$.

Because the actuator is connected to the terminal disc of the backbone via the tendon, the length of the actuator, $L_a(t)$ can be determined by the change in the tendon length as the manipulator bends. The actuator length is given by:

$$L_a(t) = L_{a0} - r \cdot \theta(L) \quad (17)$$

where L_{a0} is the length of the actuator when the manipulator is straight (i.e. $\theta(s) = 0$ for all $s \in [0, L]$), and r is the radius of the routing disc.

Refinements to this basic model have been presented in the literature; however, given the actuator size and pressure used in this application, the basic model provides sufficient accuracy to gain an understanding of the theoretical system performance [20–22]. This analysis suggests that it is possible to achieve the design requirement of five Newtons at the distal end of the manipulator using a relatively modest operating pressure of 700 kilopascals while staying within the size constraints of the endoscope working channel.

Control Valve

In order to achieve precise manipulation, it is necessary to obtain a control valve that will provide a controllable fluid flow rate and pressure while occupying as little space as possible within the device. A review of the available commercial valves and common micro-fluidic valve solutions was conducted and no existing valve design was found suitable for the present application due to the limitations of high supply pressure, proportional control, and method of activation. Therefore, development of a novel control valve (see model in Fig. 5) capable of manipulating high pressure flows on a small scale is in progress. The basic concept being employed is a flapper-style valve which acts as a pressure divider to control the proportion of the supply pressure that is applied to the internal bladder of the actuator. For size comparison, a first generation prototype of the valve is shown in Fig. 6.

Each manipulator segment is driven by a hydraulic circuit of the form shown in Fig. 5. It consists essentially of three flapper nozzle valves [23] controlled by one single tetrahedral valve plug that plays the role of the flapper. The valve is normally open thus permitting flow to bypass the actuator and dump to the return line. The tetrahedral shaped valve plug can move both in plane and out of plane. When the valve plug is activated in a particular direction in the plane, the orifices are activated differentially:

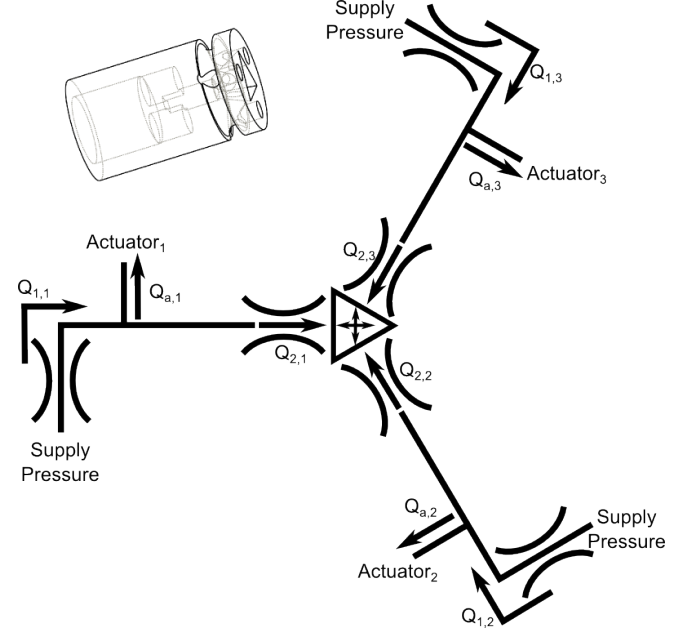


Figure 5. HYDRAULIC CIRCUIT AND MODEL OF PHYSICAL REALIZATION OF VALVE DESIGN.

the orifice(s) that the plug gets closer to become more restrictive, and the corresponding actuator pressures increase; the orifice(s) that the plug get further again become less restrictive, and the corresponding actuator pressures decrease. This in turn realizes a directional actuation of the manipulator to bend in the direction where the orifices become more restrictive. Out of plane movement of the valve plug pressurizes or de-pressurizes the actuators simultaneously. This has the effect of increasing or decreasing the manipulator stiffness.

This style of valve has the advantages of low complexity and insensitivity to contaminants in the fluid [23]. However, since it is difficult to achieve complete closure of the orifice, the system must be able to accommodate some leakage. This presents little concern as the flow rate required to inflate the actuators is significantly smaller than the available flow rate of the supply, five liters per hour based on actuator fill time versus 55 liters per hour based on available flow rate for endoscopic irrigation pumps, respectively [23,24]. The valve consists of a single supply pressure input which feeds three downstream orifices. The central plug is manipulated electromagnetically by three independent control signals. The plug can be manipulated to block each downstream orifice individually, in pairs of two, or all three depending on how the control signals are applied. When an orifice is blocked, pressure increases at the feed point for the corresponding actuator. Conversely, when the orifice is unblocked, flow passes to the

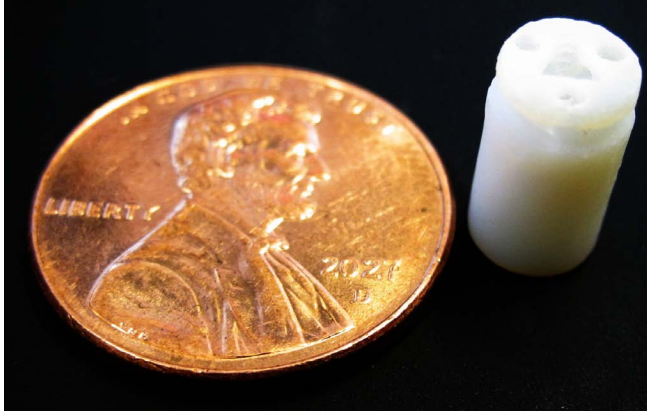


Figure 6. PHOTOGRAPH OF FIRST GENERATION VALVE PROTOTYPE.

return line and pressure at the actuator feed point decreases. Figure 7 shows one third of the flow path when the variable orifice is open (top) and when the orifice is blocked (bottom). In the open position, flow passes through the valve following the least resistance path and exits to the return line. When the variable orifice is blocked, the path to the return line has a high flow resistance and thus the fluid flow is directed into the actuator.

In the steady-state condition when no flow passes to the actuator, the upstream and downstream flow rates become equal and the equations simplify to

$$\frac{P_a}{P_s} = \left[1 + \left(\frac{C_{df} A_f}{C_{d0} A_0} \right)^2 \right]^{-1} \quad (18)$$

This relationship is shown in Fig. 8 [23]. This plot shows that there is large region of approximately linear response between 0.4 and 1.6 for the orifice ratio. Further, it can be seen that the pressure response flattens out an orifice ratio of approximately 3.0, meaning that this is a reasonable upper limit placed on the size of the flapper valve exit area relative to the upstream flow restriction.

The valve design is modeled as a flapper style valve and thus the flow through the upstream orifice for each orifice (Q_1) and the downstream valve orifice (Q_2), shown in Fig. 5, is calculated in Eqs. (19) and (20).

$$Q_1 = A_0 C_{d0} \left[\frac{2}{\rho} (P_s - P_a) \right]^{\frac{1}{2}} \quad (19)$$

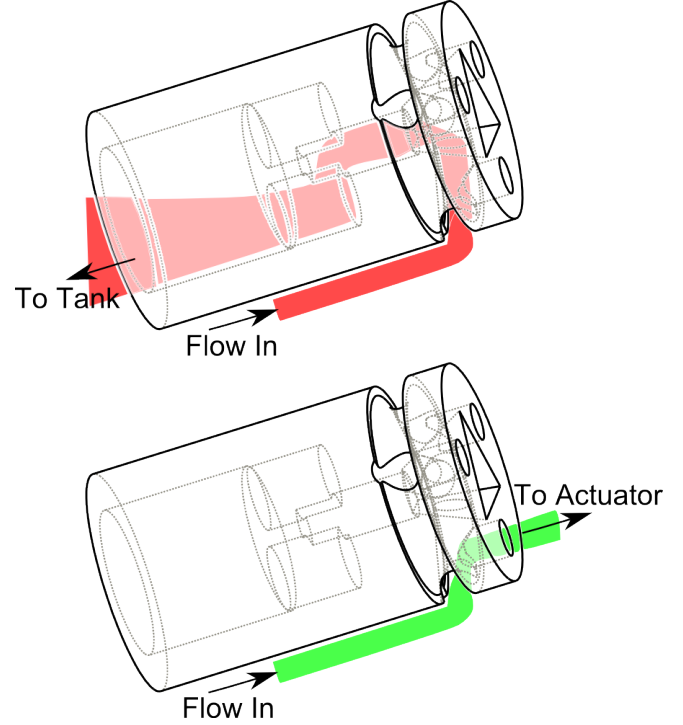


Figure 7. MODEL OF CONTROL VALVE SHOWING PART OF THE FLOW PATH FOR BOTH OPEN (TOP) AND BLOCKED (BOTTOM) VARIABLE ORIFICE CONDITIONS.

$$Q_2 = A_f C_{df} \left[\frac{2}{\rho} P_a \right]^{\frac{1}{2}} = \pi D_N g C_{df} \left[\frac{2}{\rho} P_a \right]^{\frac{1}{2}} \quad (20)$$

Here it is shown that the upstream flow rate is consistent with the standard orifice equation while the downstream flow exiting the downstream orifice is restricted by the flapper. In this case the curtain area (A_f) rather than the nozzle area is used in the flow rate calculation. The curtain area is equivalent to $\pi D_N g$, where D_N is the orifice diameter and g is the gap between the orifice and the flapper. Further, the discharge coefficient for the flapper orifice (C_{df}) is typically on the order of $0.8 \cdot C_{d0}$ [23].

Additionally, the flow rate to or from the actuator must satisfy the difference between the upstream and downstream flow rates as in Eq. (21).

$$Q_a = Q_1 - Q_2 \quad (21)$$

Thus, since Q_a is equal to $\dot{V}(t)$ under the incompressibility assumption, actuator pressure, $P_a(t)$, can be found by combining Eq. (19, 20, and 21) with Eq. (10).

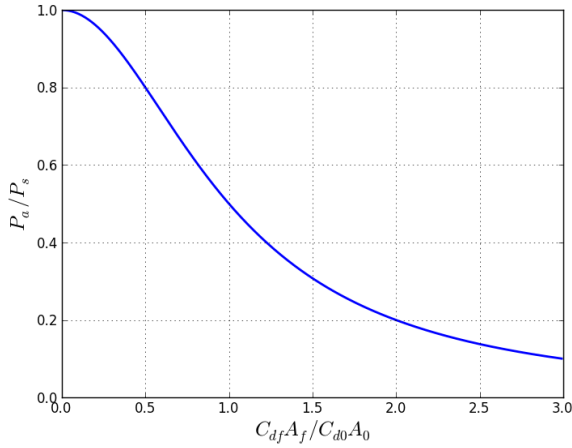


Figure 8. PLOT OF PRESSURE RATIO VERSUS ORIFICE RATIO FOR A FLAPPER VALVE.

System Modeling

A model of the system was developed in MATLAB/SIMULINK[®] (The MathWorks Inc.) as a combination of the models described in the previous sections. The current model assumes planar bending and doesn't take into account deformation due to beam mass. It is assumed that an end-effector is placed at the distal end of the manipulator. A force and moment balance is applied to the end-mass treated as a free body as shown in Fig. 9. Loads applied to the end mass originate from the actuators, the reaction load from the backbone, and from any external loading due to tissue manipulation or collision as shown in Eq. (22-24). The system components are coupled together through both hydraulic and geometric relationships. The control valve supplies pressure directly to the actuators. Each actuator is coupled to a tendon which in turn is routed along the manipulator backbone. Backbone shape directly determines relative length of each actuator. Thus, actuator length is combined with pressure from the control valve to calculate load applied to the distal end of the manipulator.

$$\sum F_x = m\ddot{x} = F_{b,x} + F_{L,x} + F_{a1,x} + F_{a2,x} \quad (22)$$

$$\sum F_z = m\ddot{z} = F_{b,z} + F_{L,z} + F_{a1,z} + F_{a2,z} \quad (23)$$

$$\sum M = I_b\ddot{\theta} = F_{a1} \cdot r - F_{a2} \cdot r - M_b - M_L \quad (24)$$

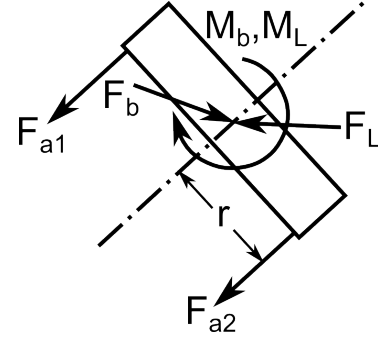


Figure 9. FREE BODY DIAGRAM FOR MASS AT DISTAL END OF THE MANIPULATOR.

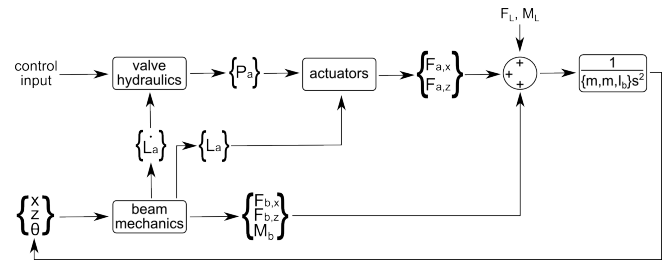


Figure 10. SIMPLIFIED BLOCK DIAGRAM OF SYSTEM MODEL.

As shown in Fig. 10, translational and rotational inertial dynamics form the basis of the model and allow for calculation of the location and orientation of the distal end of the manipulator due to applied control. From this model, it was possible to determine system response for a single segment manipulated in planar bending by a single actuator. The result is shown in Fig. 11, where actuator pressure denotes the pressure at the feed point to the activated actuator and tip orientation is the angular orientation of the distal tip of the manipulator. The tip orientation is zero when the manipulator is straight and has no deformation. The plot shows that under the ideal unloaded conditions, the tip orientation approached 180°, corresponding to full retroflexion as specified in the design requirements, as the pressure at the actuator feed point approaches 200 kilopascals. This shows that the desired beam deflection can be achieved with approximately 30 percent of the proposed supply pressure when no load is applied to the manipulator. It was found that the dynamics of this model was limited by the flow rate of the valve rather than the inertia of the manipulator due to low mass and low moment of inertia at the distal end, on the order of 10.2 milligrams and 16.2 $mg \cdot mm^2$, respectively. The valve orifices are on the order of 0.3 mm in diameter for both the upstream and downstream orifices sized to achieve both sufficient flow rate and appropriate valve control. The entire motion of the manipulator was completed in

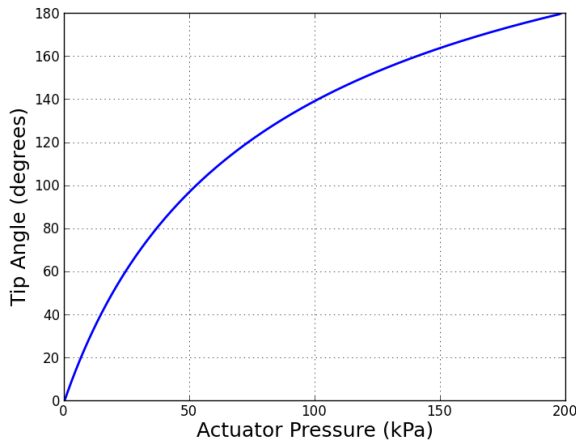


Figure 11. PLOT OF TOOL TIP ORIENTATION VERSUS ACTUATOR PRESSURE.

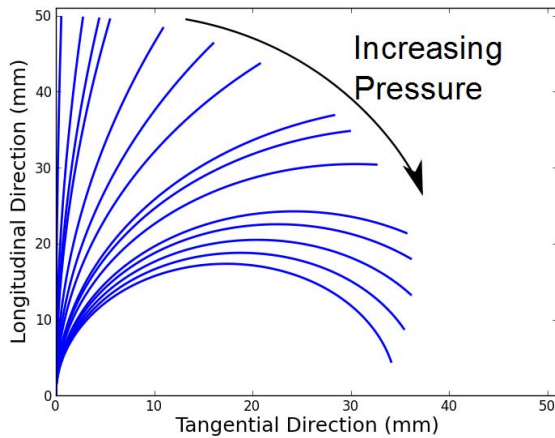


Figure 12. PLOT OF PLANAR BEAM DEFLECTION FOR INCREASING ACTUATOR PRESSURE WITH NO EXTERNAL LOAD.

less than half of one second. This time was determined to be sufficient for most applications, especially true when considering that most applications will not require full retroflexion.

The range of motion for a single segment manipulator with no external loading is then defined by the sweep between the undeformed configuration and the fully deformed configuration as shown in Fig. 12. This sweep range can be rotated around the longitudinal axis of the undeformed manipulator in order to project the workspace into the third dimension.

CONCLUSIONS AND FUTURE WORK

This paper provides a first look at our current work on the development of a hydraulic platform for minimally invasive surgical procedures. The primary benefit of this approach is that the control and actuation components are all located within the distal end of the endoscope while meeting requirements of common surgical procedures. To achieve this, a novel design for a flow control valve has been introduced. A basic approach to the modeling of each of the components of this surgical manipulator has been presented and assembled into a system model in order to assess the performance of the system as a unit.

Future work will include the development of a more complete model which allows for three dimensional deformation. Further, a prototype of this design is being produced in order to validate the model and demonstrate the feasibility of fluid power for driving a serpentine manipulator for minimally invasive surgery. Development of an appropriate control approach will be necessary to achieve precise positioning of the manipulator. While the present work uses transgastric natural orifice surgery as a case study to define the design constraints, it is expected that the device will be suitable for a variety of surgical access methodologies including laparoscopies, laparotomies, and single port access surgeries.

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REFERENCES

- [1] Shao, H., Chen, J., Truong, T., Reed, I., and Kwoh, Y., 1985. "A New CT-Aided Robotic Stereotaxis System". In *Proceedings of the Annual Symposium on Computer Application in Medical Care*, American Medical Informatics Association, p. 668.
- [2] Kwoh, Y., Hou, J., Jonckheere, E., and Hayati, S., 1988. "A Robot with Improved Absolute Positioning Accuracy for CT Guided Stereotactic Brain Surgery". *IEEE Transactions on Biomedical Engineering*, **35**(2), pp. 153–160.
- [3] Rothstein, R., Rosen, J., and Young, J., 2004. "Improving Efficiency in Endoscopy with Robotic Technology". *Gastrointestinal Endoscopy Clinics of North America*, **14**(4), pp. 679–696.
- [4] Granosik, G., and Borenstein, J., 2006. "Pneumatic Actuators for Serpentine Robot". In *Climbing and Walking Robots*. Springer Berlin Heidelberg, pp. 719–726.
- [5] Peirs, J., Reynaerts, D., and Van Brussel, H., 2000. "Design of Miniature Parallel Manipulators for Integration in

- a Self-Propelling Endoscope". *Sensors and Actuators A: Physical*, **85**(1-3), pp. 409–417.
- [6] Love, L., Lind, R., and Jansen, J., 2009. "Mesofluidic Actuation for Articulated Finger and Hand Prosthetics". In Proceedings of the 2009 IEEE/RSJ International Conference on Intelligent Robots and Systems, IEEE, pp. 2586–2591.
 - [7] Whitesides, G., 2006. "The Origins and the Future of Microfluidics". *Nature*, **442**(7101), pp. 368–373.
 - [8] Berg, D., Kinney, T., Li, P., and Erdman, A., 2011. "Determination of Surgical Robot Tool Force Requirements Through Tissue Manipulation and Suture Force Measurement". In Proceedings of the 2011 Design of Medical Devices Conference, ASME. DMD2011-5257.
 - [9] Webster, R., Okamura, A., and Cowan, N., 2006. "Toward active cannulas: Miniature snake-like surgical robots". In Proceedings of the 2006 IEEE/RSJ International Conference on Intelligent Robots and Systems, IEEE, pp. 2857–2863.
 - [10] Camarillo, D., Milne, C., Carlson, C., Zinn, M., and Salisbury, J., 2008. "Mechanics Modeling of Tendon-Driven Continuum Manipulators". *IEEE Transactions on Robotics*, **24**(6), pp. 1262–1273.
 - [11] Trivedi, D., Lotfi, A., and Rahn, C., 2008. "Geometrically Exact Models for Soft Robotic Manipulators". *IEEE Transactions on Robotics*, **24**(4), pp. 773–780.
 - [12] Cardona, D., 2012. "A MRI Compatible Concentric Tube Continuum Robot With Pneumatic Actuation". Master's thesis, Vanderbilt University.
 - [13] Anderson, V., and Horn, R., 1967. "Tensor Arm Manipulator Design". *ASME Transactions*, **67-DE-57**, pp. 1–12.
 - [14] Webster III, R., and Jones, B., 2010. "Design and Kinematic Modeling of Constant Curvature Continuum Robots: A Review". *The International Journal of Robotics Research*, **29**(13), pp. 1661–1683.
 - [15] Gravagne, I., and Walker, I., 2000. "Kinematic Transformations for Remotely-Actuated Planar Continuum Robots". In Proceedings of the IEEE International Conference on Robotics and Automation, Vol. 1, IEEE, pp. 19–26.
 - [16] Li, C., and Rahn, C., 2002. "Design of Continuous Backbone, Cable-Driven Robots". *Journal of Mechanical Design*, **124**, p. 265.
 - [17] Simaan, N., Taylor, R., and Flint, P., 2004. "A Dexterous System for Laryngeal Surgery". In Proceedings of the 2004 IEEE International Conference on Robotics and Automation, Vol. 1, IEEE, pp. 351–357.
 - [18] Gravagne, I., Rahn, C., and Walker, I., 2003. "Large Deflection Dynamics and Control for Planar Continuum Robots". *IEEE/ASME Transactions on Mechatronics*, **8**(2), pp. 299–307.
 - [19] Schulte, H., 1961. "The Characteristics of the McKibben Artificial Muscle". In *The Application of External Power in Prosthetics and Orthotics*. National Academy of Sciences National Research Council, pp. 94–115.
 - [20] Chou, C., and Hannaford, B., 1996. "Measurement and Modeling of McKibben Pneumatic Artificial Muscles". *IEEE Transactions on Robotics and Automation*, **12**(1), pp. 90–102.
 - [21] Klute, G., and Hannaford, B., 2000. "Accounting for Elastic Energy Storage in McKibben Artificial Muscle Actuators". *Journal of Dynamic Systems, Measurement, and Control*, **122**(2), pp. 386–388.
 - [22] Tsagarakis, N., and Caldwell, D., 2000. "Improved Modelling and Assessment of Pneumatic Muscle Actuators". In Proceedings of the 2000 IEEE International Conference on Robotics and Automation, Vol. 4, IEEE, pp. 3641–3646.
 - [23] Merritt, H., 1967. *Hydraulic Control Systems*. John Wiley & Sons Inc.
 - [24] Cheng, L., 2007. "Development of an Endoscopic Irrigation Pump: Experience with Byrne Medical, Inc.". PhD thesis, Texas A&M University.