An MRI compatible needle manipulator for prostate cancer therapy

Jean-Sébastien Plante\textsuperscript{1}, Lauren Devita\textsuperscript{2}, Kenjiro Tadakuma\textsuperscript{2}, Yan Shaoze\textsuperscript{2}, Steven Dubowsky\textsuperscript{2}

\textsuperscript{1}Université de Sherbrooke, Canada
\textsuperscript{2}Massachusetts Institute of Technology, United States

1. Introduction

The ability to perform cancer therapies within Magnetic Resonance Imaging (MRI) systems has the potential to greatly improve cancer survival rates and quality of life after treatment [1]. For example, a real-time image and a MRI compatible manipulator would allow surgeons to accurately guide a needle to a malignant tumor while avoiding vital structures. Very small tumors (<3 mm) can be localized and reached enabling early cancer detection and local treatment. Health risks, treatment costs, and side effects are minimized. Applications of MRI guided robotic surgery extend beyond cancer therapy and could have significant impact in neurology, orthopaedics, and cardiology.

There has been substantial research in the development of in-bore MRI manipulation for diagnostic and surgical procedures. These manipulator systems must be compatible with the very high magnetic fields (~ 3 Teslas) used in MRI. Non-magnetic materials and low electrical currents must be used to assure safety and prevent
interference with the imaging process. Conventional robotic actuators, which are typically electromagnetic devices, must be more than 1m away from the center of the magnet [2]. Hence, they require complex, expensive, and cumbersome transmissions making this approach unpractical.

Dielectric Elastomer Actuators (DEAs) are a promising alternative for MRI manipulation because they can be actuated directly within the bore of the MRI without affecting image quality [3]. Unlike other MRI compatible actuators such as piezoelectrics and pneumatics, DEAs are an attractive solution because they offer good mechanical performance while being simple and inexpensive (see Figure 1). However, using DEAs in practical robotics and mechatronics systems is not easy due to their time-dependant failure modes that limit their performance and reliability. Recent studies have shown that using DEAs intermittently at high speeds can significantly increase their performance and reliability [4].

Binary actuation is a design paradigm that exploits the advantages of using DEAs intermittently. Binary systems are the mechanical equivalent of digital electronics [5]. A binary system as intended here is driven by bistable actuators that switch between two possible states, extended or retracted. Overall system architecture is greatly simplified compared to conventional robotic systems since very few sensors are needed [6]. Further, in the context of MRI applications, bistable DEAs can advantageously be turned off before an image is taken, further reducing the chances of affecting the image.
This chapter presents a binary manipulator for MRI guided prostate cancer biopsies and brachytherapies using a transperineal approach (see Figure 2). The problematic of prostate cancer detection and treatment is exposed along with a review of alternative approaches. The proposed manipulator concept is presented and an analytical model of its performance is developed. Model predictions are compared with experimental results. This study shows that the proposed binary manipulator concept has the potential to meet the clinical requirements of transperineal needle insertion. The proposed concept is effective and could easily extend to general manipulation problems.

2. Prostate Cancer Therapy

Prostate cancer is the most frequently diagnosed cancer in men and the number two cause of cancer related death in men after lung cancer [7]. In the United States alone, 218,890 men will be diagnosed with prostate cancer in 2007 while 30,000 men die from the disease each year [7,8]. The number of deaths has been declining since the early 1990s due to better detection and treatment methods.

2.1 Prostate Cancer Detection

To date, needle biopsy is the best method to confirm if a malignant prostate tumor is present. Every year, about 800,000 men have a prostate biopsy, of which 600,000 have negative results [9]. Studies show that current biopsy techniques miss up to 20% of all prostate cancers, leaving 50,000 men with “false negative” results [10,11]. In
these cases, the tumor will go undetected. Left untreated, it may be too late to successfully treat, especially if the cancer has spread beyond the prostate.

Accurate biopsies are vital to prostate cancer detection and, hence, treatment. Early stage detection requires accurate imaging systems and precise needle insertion techniques to probe small potential tumors. Currently, ultrasound imaging is used to manually insert a biopsy needle to the prostate. Unfortunately, the biopsy needle often misses the tumor because ultrasound images are too low resolution to see small, early stage, tumors. Tumors smaller than 5mm are not detected by ultrasound imaging [12]. Only about 20% of tumors between 5 and 10mm are detected by ultrasound. Even large tumors, on the order of 20 to 25mm, are only detected 79% of the time.

Ultrasound images, therefore, cannot be used to detect or treat millimeter size prostate tumors. Figure 3 shows ultrasound and MR images of the same prostate [13]. Even to the untrained eye, it is clear that MR images are far more detailed than ultrasound images. Given the unique imaging capabilities of MRI, it is possible to diagnose and treat small, millimeter size tumors that cannot be detected otherwise [14,15].

2.2 Prostate Cancer Treatment

In the past, radical prostatectomy has been the only option to treat prostate cancer. Important undesirable side effects such as incontinence and impotence combined with long treatment duration and recovery time has caused many men to turn to new
treatments such as radiation therapy. Brachytherapy is an interesting form of radiation therapy because it is an outpatient procedure needing a recovery time of only a few days. During brachytherapy, similar to needle biopsy, a needle is inserted into the tumor in the prostate, typically using an ultrasound image for guidance, see Figure 4. The needle deposits small radioactive pellets in the tumorous region. Brachytherapy is analogue to cryotherapy where very low temperature substances are injected to “freeze” a tumor. Figure 4 shows a transperineal approach which, compared to the other alternative of transrectal approach, has lower risks of hitting vital structures and does not require antibiotics because the insertion happens in a clean environment.

Transperineal brachytherapy is growing in popularity over other treatment options because of a significantly smaller chance of incontinence and impotence, quick treatment and recovery and high success rates [16]. As brachytherapy becomes the method of choice, the procedure must become more reliable. Early detection through accurate biopsy needle placement is crucial to treat prostate cancer. To increase prostate cancer survival rates, it follows that needle placement methods must be improved with MRI guided methods.

2.3 Needle Placement in MRI Systems

Robotic needle manipulators have been developed for assisted interventions in open-bore MRI [17,18]. Open-bore MRI uses split magnets that leave room for large remote manipulators to be used. Compared to closed-bore MRI, open-bore MRI operate at lower magnetic fields (1.5 versus 3.0 Teslas) and have significantly lower
resolution. Image quality of open-bore MRI is not detailed enough or suitable for accurately treating a tumor on the order of 5mm [19].

Closed-bore MRI manipulation requires actuators with excellent MRI-compatibility because of their proximity of the image center. Piezoelectric motors have been reported to cause image distortion in closed-bore MRI [20]. Piezoelectric motors are also relatively slow, complex, and expensive. Manipulators using pneumatics have had limited success in conjunction with closed-bore MRI [21,22]. The main drawbacks to this approach are control issues due to the imprecision and compliance of pneumatics. Recent advances have been made using pneumatic stepper motors to eliminate these effects [23]. However, pneumatic systems, using continuous cylinders or step motors, tend to be complex and expensive.

Compared to other actuation alternatives, DEAs have a huge potential for closed-bore MRI manipulation. They have excellent MRI compatibility, offer good performance, and are simple and low cost. The binary manipulator system presented in the next section is an illustration of what a practical MRI manipulator using DEAs could look like.

3. Elastically Averaged Parallel Manipulator Using Dielectric Elastomer Actuators

The manipulator is intended to be used inside the bore of a closed-bore MRI machine to perform transperineal prostate cancer needle biopsy, brachytherapy, and
cryotherapy (all can be performed by inserting a needle through the perineum). This section discusses design requirements, presents the manipulator concept, and proposes an analytical model of the manipulator.

3.1 Design Requirements

The following design requirements for system size, workspace, precision, and forces have been developed in collaboration with doctors and researchers at Brigham and Women’s Hospital in Boston Massachusetts.

Size and Workspace:

The device must be MRI compatible and able to fit between the patient’s legs while inside the bore of the MRI machine, see Figure 5. The bore of a typical MRI machine is 550mm in diameter. The patient’s legs are propped up to provide access to the perineum. This leaves a small space for the device to reside. This requires that the device be no larger than a 200mm diameter cylinder, 500mm deep.

Under the control of a doctor using a real-time MR image, the proposed device is required to reach a target (a tumor) in the prostate by penetrating the perineum as shown in Figure 6. The average prostate is located 60 to 80mm from the perineum, and the size of the average prostate is 30 to 50mm in the z-direction [24]. The needle must be able to travel between 60 and 130mm in the z-direction. The required workspace is an elliptic cylinder with a major axis of 80mm and a minor axis of 70mm, 70mm in the z-direction, as shown Figure 5 and Figure 6. This workspace is larger than the average
normal prostate, but is necessary to accommodate the frequently enlarged prostates of cancer patients as well as position differences from patient to patient.

**Precision:**

Binary systems have discrete workspaces consisting in all possible end-effector locations. Here, precision is defined as the minimum distance between a random point in the required workspace from the closest possible needle tip location. Manual needle insertion methods currently in use have a perforated template with holes 5mm apart to guide the needle. Statistical studies showed that the average distance of these templates is 1.9 mm [25]. The target for the manipulator is to match the same level of precision or do better than current methods. It worth mentioning that the manipulator will be guided by a surgeon using real-time MR images. In this case, the absolute precision becomes less critical since corrections can be applied during insertion.

**Forces:**

Tests were performed on beef muscle tissue to establish the magnitude of the external forces applied on the needle during insertion [25]. Beef muscle has similar properties to those of the flesh between the perineum and the prostate. In these tests, forces are considered to be applied at the needle point of entry. Combining these results with those of the literature give a maximum axial force of 14N and a maximum radial force of 1.6 N for a beveled needle tip [26]. A trihedral needle tip would lower the axial force to ~8.25N and the radial force to ~0.5N. Hence trihedral needles are desirable.
3.2 Manipulator Concept

The manipulator concept is shown in Figure 2. A schematic representation is shown in Figure 7. It is an elastically averaged parallel manipulator having 2 planes, each containing a circular array of 6 bistable actuators and 6 spring elements. A biopsy needle runs through a rigid tube attached at the center point of each plane and advances from Plane 1, through Plane 2 to the target. The planes are separated by a distance \( p \) and the needle tip is at a distance \( z_d \) from the first plane. Needle manipulation is achieved by changing the location of the center point of each plane when the static equilibrium of the springs is modulated by the bistable modules. Compared to conventional serial chain manipulators, the proposed parallel architecture provides higher system stiffness and higher precision.

The bistable module concept is shown in Figure 8. Bistability is achieved by flipping a bistable truss with two independent cone-shaped DEAs. The module output has a stroke \( \zeta \) between its “extended” and “retracted” states (Figure 7 shows all bistable assemblies in the “retracted” position). Using DEAs in a bistable fashion has a considerable advantage over continuous operation because it increases actuator performance and life [4]. Moreover, power is not needed to hold a state which reduces energy consumption and the risk of interferences with the MRI process.

3.3 Manipulator Analytical Model

A binary manipulator inputs \( Q \) can be represented in a binary sequence of 0 and 1:
\[ Q = [a_1 \ a_2 \ \ldots \ a_n] \] (1)

where \( a_i \) = 1 or 0 is the state (extended or retracted) of the \( i \)th bistable module and \( n \) is the total number of bistable modules. Binary systems can only reach a finite set of discrete points (\( 2^n \) points) covering the system’s workspace. In most systems, these discrete points are not evenly distributed in the workspace. Hence, an analytical model that maps the manipulator workspace for any given inputs is essential to optimize the manipulator design for a given task.

For a given set of inputs, the spatial configuration of the elastically averaged parallel manipulator is found by resolving the static equilibrium in each plane of the device. Typical spring deformations and applied forces in a single plane of the device are represented schematically in Figure 9. Only 3 springs are shown for clarity. The model uses the following input parameters:

- \( k_i \) stiffness of \( i \)th spring
- \( l_{0i} = l_{0i}w_i \) position vector of the \( i \)th undeformed spring
- \( \beta_i = \beta_iw_i \) pre-stretch vector of the \( i \)th spring.
- \( \delta_i = \delta_iw_i \) stroke vector of \( i \)th bistable module. The magnitude of the stroke vector is given by \( \delta_i = \zeta a_i \) with \( a_i \) (0 or 1) the state of the bistable module and \( \zeta \) the stroke.
- \( f_{ext} \) external force vector at center point
- \( W \) weight vector at center point

to compute the following output parameters:
\( u = ue \) displacement vector of the center point

\( l_i = l_i v_i \) position vector of the \( i \)th deformed spring

\( F_i = F_i v_i \) internal force vector of the \( i \)th spring

In the preceding list, \( w_i, v_i \), and \( e \) are the respective unit vectors of the undeformed direction of the \( i \)th spring, the deformed direction of the \( i \)th spring, and the displacement of the center point.

For static equilibrium, the sum of the forces at the center point must equal zero:

\[
\sum \text{forces} = \sum_i F_i + f_{ext} + W = 0
\]  

(2)

The internal force of the \( i \)th spring generated by the spring elongation is given by:

\[
F_i = k_i (l_i - l_{0i}) v_i = k_i \left[ (l_{0i} + \beta_i + \delta_i) w_i - ue - l_{0i} v_i \right]
\]  

(3)

Equations (2) and (3) are solved iteratively for the displacement vector of the center point: \( u \). The method is used to find the center point locations of each plane, \( u_1 = (x_1, y_1, z_1) \) and \( u_2 = (x_2, y_2, z_2) \), from which the needle tip (end effector) location can be found by:

\[
u_d = (x_d, y_d, z_d) = u_1 + \frac{z_d}{p} (u_2 - u_1)
\]  

(4)

where \( p \) and \( z_d \) are geometric parameters defined in Figure 7. In these calculations, the origin is taken as the center point of plane 1 before any perturbation. The model is used to compute the full manipulator workspace corresponding to all combinations of the bistable modules states.
The manipulator system stiffness represents its ability to resist external forces without major deflections. System stiffness is defined by the equivalent stiffness at the tip of the manipulator:

\[
K_{\text{eq, tip}} = \frac{k_{\text{eq1}}k_{\text{eq2}}p^2}{k_{\text{eq1}}(z_d)^2 + k_{\text{eq2}}(z_d - p)^2}
\]  \hspace{1cm} (5)

where \( k_{\text{eq1}} \) and \( k_{\text{eq2}} \) are the equivalent stiffness in planes 1 and 2, \( p \) and \( z_d \) are geometrical parameters defined in Figure 7 [25]. Equation (7) is defined in a radial plane of the device and must be evaluated from 0 to 360° to evaluate the manipulator stiffness in all directions.

If the effects of external forces, \( \mathbf{f}_{\text{ext}} \), and gravity, \( \mathbf{W} \), can be neglected, the displacement vector of the center point of each plane can be obtained from Eqs. (2) and (3) as:

\[
\mathbf{u} = \frac{1}{\sum k_i} \sum_i k_i \left[ \left( \mathbf{l}_{0i} + \beta_i + \delta_i \right) \mathbf{w}_i - l_{0i} \mathbf{v}_i \right]
\]  \hspace{1cm} (6)

Defining a dimensionless ratio of the spring stiffness, \( \mu_i = k_i/k_1, \) yields:

\[
\mathbf{u} = \frac{1}{\sum \mu_i} \sum_i \mu_i \left[ \left( \mathbf{l}_{0i} + \beta_i + \delta_i \right) \mathbf{w}_i - l_{0i} \mathbf{v}_i \right]
\]  \hspace{1cm} (7)

Equation (7) shows that when there are no external forces or gravity, the displacement vectors of the center points are functions the ratio of the spring stiffness, not their absolute value. Hence, the manipulator’s workspace is independent of its overall stiffness and a very “soft” manipulator can have the same workspace as a very “stiff” one.
4. Results

The laboratory prototype of Figure 2(b) was developed to evaluate the effectiveness of the manipulator concept. The key design parameters are listed in Table 1. The system has an outer diameter of 400mm and a length of 450mm. It uses 12 functional bistable modules, each using a pair of cone-shaped DEAs. These cone actuators are hand fabricated using 2 active polymer layers made from 1.5mm thick films of 3M’s VHB4905/4910. The cone actuators have an outer diameter of 100mm, develop a maximum force of 6N, and have a stroke of 12mm.

The prototype was designed using the performance specifications of current hand-fabricated DEAs. Hand-fabrication limits the number of active film layers in each actuator resulting in relatively large actuators with limited forces. Consequently, the prototype had to be designed at twice the scale than the clinical device and the springs stiffness had to be lowered to prevent actuator saturation. Developing smaller and stronger DEAs using many active layers is feasible through the development of appropriate actuator manufacturing techniques (currently under development).

4.1 Analytical Results

The analytical model developed in section 3.3 is used to study the prototype workspace, precision, and stiffness properties. Figure 10(a) shows the needle workspace at penetrations from 60 to 130mm. Figure 10(b) shows the cross-section of the workspace at a penetration of 110mm. As discussed above, a binary system’s
Manipulator precision is evaluated from 1000 random target points within the workspace. The minimum distance between these points and the closest possible needle tip (end-effector) position was calculated. The manipulator must be able to reach within a distance of 1.9mm of any point in the prostate. Figure 11(a) shows the error distribution. The average distance and standard deviation, for a random point to the nearest possible end-effector point in the required workspace are 0.68mm and 0.51mm, respectively. The same values for a prostate sized workspace are 0.41 mm and 0.21 mm. Clearly the manipulator design meets its precision requirements with only 12 binary inputs. This is not surprising since the resolution of binary systems increases drastically with the number of inputs. For example, doubling the number of binary inputs to 24, which is technically and economically feasible with DEAs, would drop the minimum distance in the micrometer range.
Figure 11(b) shows the distribution of system stiffness in the radial direction (perpendicular to the needle) at the needle tip. The values in the other directions perpendicular to the needle are similar. This figure shows that the stiffness of the system is quite uniform. The radial stiffness for any point in workspace is within 10% of the average stiffness of the system. Recall that system stiffness is limited by the low forces of our hand fabricated DEAs. With a minimum stiffness of 0.03N/mm, the transversal force of 1.6N caused by a beveled needle would result in an unacceptable deflection of 50 mm. Developing higher stiffness requires actuators with higher force output. To meet the clinical requirements on needle insertion forces, it is estimated that the number of active layers must increase from 2 to 30, which is technically feasible with proper actuator manufacturing techniques [25].

4.2 Experimental Results

The motion of the needle tip was measured experimentally and compared with predictions of the analytical model. The tip motion is measured with a magnetic position sensor (miniBIRD model 80 from Ascension Technology). This electromagnetic tracking system measures the 3 positions and 3 orientations of a small sensor (1.3 mm in diameter x 6.5 mm long) with RMS accuracies of 1.4 mm and 0.5°. The sensor is small compared to the manipulator and does not affect its performance.

Figure 12(a) shows the measured location of 54 random inputs over the system’s 4096 possible inputs. The general size and shape of the workspace is essentially the same as that predicted by the analysis, see Figure 10(b). Figure 12(b) compares
analytical predictions of the needle tip location with experimental measurements. The circle shows the size of an average prostate. In this area, the average distance between the simulated and experimental points is about 3 mm, which is smaller than the required precision of +/- 5mm. It should be noted that the sensor itself has an RMS error of about 1.4 mm suggesting that more precise laser measurements and calibration would significantly reduce the experimental errors. Also, the prototype is a laboratory device containing many geometric error sources that would be eliminated in a production grade system.

Finally, the MRI compatibility of the prototype was verified in a 3 Teslas MRI at Harvard’s Brigham and Woman’s Hospital [25]. These tests confirmed previous results on individual actuators showing that DEA technology is MRI compatible [3]. Specifically, the manipulator was not degraded by the high magnetic fields and the MR images were not affected by the manipulator using conventional filtering.

5. Conclusions

This chapter presented an MRI compatible manipulator concept based on elastically averaged binary DEAs for prostrate cancer detection and treatment. A manipulator design using 12 binary degrees-of-freedom was proposed along with an analytical model of its performance. Experimental and analytical validations suggest the approach to have sufficient precision and workspace range to meet the medical requirements. However, system size and stiffness are currently limited by the low forces of the hand-fabricated DEAs available in this study. Appropriate manufacturing
techniques for high force DEAs are currently under development. Tests performed in a 3 Teslas MRI machine confirmed the excellent MRI compatibility of the technology.

Results presented in this chapter suggest that the proposed manipulation approach based on elastically averaged parallel manipulation is an effective way of using DEAs in practical robotics and mechatronics systems. Other medical procedures requiring precise robotic needle placement in MRI environments could benefit from this work such as breast cancer detection and treatment, endovascular surgeries, and spinal procedures. Possible applications of elastically averaged parallel manipulation using bistable DEAs extend to general robotic tasks, inside or outside MRI systems, providing precision, simplicity, and low costs.

6. Acknowledgements

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Table 1. Manipulator Parameters

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<tr>
<td>d_l</td>
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Spring Constants, $k = 0.05$ N/mm

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Figure 1. Dielectric elastomer actuator under 100% extension.

Figure 2. Elastically-averaged parallel manipulator for MRI guided prostate cancer therapy.
Figure 3: Ultrasound image (left) and MR image (right) of same prostate [13].

Figure 4. Prostate cancer transperineal brachytherapy treatment.

Figure 5: Size and workspace requirements.
Figure 6. Needle Path.

Figure 7. Schematic of the proposed manipulator.

Figure 8. Bistable module.
Figure 9. Analytical model variable definition: (a) node and center point displacements, (b) free body diagram.

Figure 10. Analytically predicted workspace: (a) full needle workspace, (b) workspace at an insertion depth of 110mm.
Figure 11. Analytically predict ions: (a) minimum distance from needle tip to target, (b) system stiffness in N/mm at the needle tip.

Figure 12. Experimental measurements of needle tip position (mm): (a) 54 random points, (b) comparison with analytical model.