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Development and Preliminary Evaluation of a Motorized Needle Guide Template for MRI-guided Targeted Prostate Biopsy

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Abstract

To overcome the problems of limited needle insertion accuracy and human error in the use of a conventional needle guide template in MRI-guided prostate intervention, we developed a motorized MRI-compatible needle guide template that resembles a TRUS-guided prostate template. The motorized template allows automated, gapless needle guidance in a 3T MRI scanner with minimal changes in the current clinical procedure. To evaluate the impact of the motorized template on MRI, signal-to-noise ratio and distortion were measured under various system configurations. A maximum of 44% signal-to-noise ratio decrease was found when the ultrasonic motors were running, and a maximum of 0.4% image distortion was observed due to the presence of the motorized template. To measure needle insertion accuracy, we performed four sets of five random target needle insertions mimicking four biopsy procedures, which resulted in an average in-plane targeting error of 0.94 mm with a standard deviation of 0.34 mm. The evaluation studies indicated that the presence and operation of the motorized template in the MRI bore creates insignificant image degradation, and provides submillimeter targeting accuracy. The automated needle guide that is directly controlled by navigation software eliminates human error so that the safety of the procedure can be improved.

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Index Terms

Prostate biopsy; MRI-guided procedure; needle guide template; robotic intervention; ultrasonic motor

I. Introduction

The standard method to diagnose prostate cancer is transrectal ultrasound (TRUS)-guided systematic biopsy, where 6–12 cores are sampled from a fixed pattern of the paramedian prostate regions[1, 2]. While TRUS has been accepted as the de facto standard for guiding targeted biopsy due to its ease-of-use, portability, and real-time imaging capability, magnetic resonance imaging (MRI) has been also investigated as an alternative imaging technique to guide biopsies[3–7]. A motivation to use MRI for biopsy guidance comes from MRI's superior soft tissue contrast and multiparametric imaging capability, which may improve detection, localization, and negative predictive value of prostate cancer. With MRI's ability to depict suspicious foci, one can perform a targeted biopsy [8] where MRI helps reliably localize prostate zones and tumor foci and record biopsy sites for future reference.

To date, documented results indicate that an MRI-guided biopsy can detect cancer up to 56% of the time using a 3T scanner (N=844, Hoekset al.[9]) and 52% to 55.5% using a 1.5T scanner (N=27, Anastasiadis et al.[10]; N=52, Roethke et al.[11] respectively), making it a strong alternative to TRUS-guided biopsy, particularly after at least one negative biopsy. In addition, as other researchers have indicated, the inclusion of multiparametric MRI including hydrogen 1 (^1H) MR spectroscopy, diffusion-weighted imaging, and contrast material-enhanced MR imaging, increases the detection rate of prostate cancer[12–15]. Because researchers have found that using advanced diagnostic images during an MR-directed biopsy procedure improves the efficacy of the biopsy[16], the prospects of using MRI-guided biopsy as an alternative diagnostic approach after repeated negative TRUS biopsy is increasing.

Robotics is often discussed as a method to overcome limited access to the confined in-bore space of an MRI scanner and to further improve the accuracy of needle placement in MRI-guided prostate interventions. Some MRI-compatible robotic needle guidance devices have been reported to enable precise needle placement under intraprocedural MRI and demonstrated promising pre-clinical evaluation studies [17–21] and preliminary clinical results [22–25]. However, these robotic needle guidance devices have not been widely deployed clinically, primarily due to their disruption to the established clinical workflow of MRI-guided biopsy. For instance, instead of adapting standard stirrups or tools used in MRI-guided biopsy, these robotic devices often require a new set of special stirrups and workflow changes to accommodate their placement. Therefore, there is a lack of material available to clarify the long-term impact of robotic needle guides in MRI-guided prostate biopsy, including their safety and efficacy.

To immediately benefit from the advantages of intraprocedural MRI guidance and to deploy a long-term study of MRI-compatible needle placement devices, we developed an MRI-compatible robotic needle guide template that can be adapted to a transperineal biopsy based on a standard template that has a grid of needle guide holes in 5 mm intervals. Our newly developed robotic device enables gapless needle insertion guidance in a 3T MRI scanner, and potentially increases the accuracy of needle placement due to its higher control resolution of needle placement. Our emphasis in developing such an MRI-compatible robotic device is to leverage the clinical workflow established for conventional template-

guided transperineal biopsies with minimal disruption to the existing clinical workflow. The robotic device we developed was tested to quantify its impact on MR images and targeting accuracy in the same clinical setting of the MRI-guided transperineal prostate biopsy.

II. A Motorized Needle Guide Template

A. Current Intervention and Design Specifications

Fig. 1 shows a 3T MRI-guided prostate intervention system using a conventional needle guide template clinically used at our institution[26, 27]. A similar system has been used in MRI-guided biopsy[28] and brachytherapy [4] in 0.5T MRI. To minimize changes to the current clinical workflow, we designed and fabricated a two-degrees-of-freedom (2-DOF) motorized needle guide template that is interchangeable with the conventional template on the current intervention system. Table 1 shows a comparison of the specifications of the motorized template and a conventional template with needle guide holes in 5mm intervals. A larger vertical range was preferred to cover needle insertion positions required for both normal patients and those who had the rectum removed.

B. Motorized Template Design and Fabrication

Two ultrasonic motors with integrated optical encoders (URS60-E3N, Shinsei Corp., Tokyo, Japan) were selected for manipulation. The main advantages of the non-magnetic ultrasonic motors over pneumatic actuators that are used in other needle guidance systems[18, 19, 21, 24, 29] are high resolution controllability and non-backdrivability for safety. A lead-screw-and-nut mechanism transmits the two motor rotations to vertical and horizontal prismatic motion of each crossbar. The crosspoint of the bars creates a needle insertion position on the axial plane of the template. Since the mechanism is non-backdrivable, a detectable position is required for the initiation of each bar position. Hence, two fiber optic limiter switches (E32-D25XR fiber optic sensor, E3XSD6 amplifier, Omron Corp.) are installed at the end of each bar's moving range and provide the home position of each bar. The repeatability of the homing using the optic switches seems sufficient. However, periodic calibration, which can be performed by measuring physical position of the bars from the template frame, may be required to ensure the kinematic accuracy.

An 18-gauge automatic biopsy gun(InVivo Germany GmbH, Schwerin, Germany) that provides high speed double fire motion to ensure intact core specimens is used in the current intervention. The biopsy gun is inserted manually by a clinician and the insertion depth that is determined by the navigation software is manually achieved by using 10 mm tick markers on the biopsy needle. This yields a relatively large control resolution compared to the submillimeter axial positioning. However, the core biopsy samples tissue for a length of over 15 mm along the needle axis so that the manual depth control is widely accepted in clinical practice.

The motorized template can be installed on the prostate intervention table with leg support that holds a patient in lithotomy position on the patient table of the MRI scanner. The needle and perineum-contacting parts were designed to be detachable for sterilization. To provide stable perineum contact and to avoid possible skin pinching by the moving crossbar, the device was designed so that the front surface of the template frame is mostly in contact with the perineum, and the front bar is offset a few millimeters back from the contacting surface.

The structural body of the motorized template is made of Ultem resin (Polyetherimide), aluminum alloys and brass alloys. The non-ferrous transmission parts include timing-belts (MXL type, trapezoidal teeth, urethane body, Kevlar core, 1/8 width, 0.08 pitch), pulleys (MXL type, 1/8 width, 0.08 pitch, aluminum body, brass set screw), and a miter gear set (48 pitch, 15 teeth, 20 degree pressure angle, brass body, custom-made brass setscrew).

C. Device Control and Navigation

The motors are controlled by a Linux-based control computer with digital I/O, analog output, and encoder counter boards. The major functions of the controller are (1) to process incoming signals from all sensors and encoders, (2) to output command to the two motors of the robot, (3) to calculate proportional-integral-derivative (PID) control, (4) to coordinate translation, (5) to provide a graphical user interface for an operator, and (6) to provide Transmission Control Protocol/Internet Protocol (TCP/IP) socket interfaces to communicate with other systems.

Fig. 3 illustrates the architecture of the motorized template control system. The control computer and the motorized template are connected by 15 m electrical cables, so the control computer can be outside the scanner room. We are in progress of updating the directly connected cable with “through patch panel” wiring for safety and to minimize possible EM destruction. For the motorized template position registration, an MR marker-based registration block (Z-frame)[30] currently used in clinical procedures to register the conventional template to the MRI coordinate system can be affixed to the motorized template.

A customized graphical user interface (GUI) was developed to provide (1) robot monitoring: sensor reading/motor output, current needle position, (2) robot control: target setting, fine adjustment, emergency switch, and (3) manual Z-frame registration, which is achieved by manually entering the positions of the MRI-visible markers of the Z-frame read from the scanner console. We also customized 3D Slicer[31], an open-source surgical navigation software, with the Prostate-Nav module[32], to use it as the navigation software to calibrate the robot to images, plan needle placement to targets, monitor position of inserted needles, and finally, to remotely control the motorized template.

The navigation software runs on a Linux-based workstation connected to the hospital’s network to import images from the MRI scanner using the Digital Imaging and Communication in Medicine (DICOM) protocol, and communicates with the control computer using the OpenIGTLink protocol[33]. The navigation software provides a graphical user interface that allows a radiologist to define targets on either preprocedural or intraoperative MR images. If preprocedural MRI is used, it can be deformably registered to an intraoperative image using 3D Slicer’s deformable image registration function. After the motorized template is registered to the image coordinate system and targets are defined, a linear transformation representing the geometric correlation between the motorized template, the image coordinate system, and the list of the targets is transferred to the control computer.

III. Impact on MRI

A. Experimental Setup

To evaluate the impact of the motorized template on MRI, signal-to-noise ratio (SNR) and distortion tests were conducted in a 3T scanner (MAGNETOM Verio, Siemens Healthcare, Erlangen, Germany). A saline water bottle phantom (0.375% NiSO₄ and 0.5% NaCl, 1900 ml, Model No. 8624186, Siemens Healthcare, Erlangen, Germany) was placed in the approximate location of the prostate and subsequently near the center of the scanner’s imaging volume. A Body Matrix Coil (Siemens Healthcare, Erlangen, Germany) was used to cover the imaging volume.

Six system configurations that can be situated in the clinical procedure were configured to measure the impact of the motorized template on MRI: (1) *Baseline*, where only the phantom is placed in the MRI scanner, (2) *Device in Place*, where the motorized template is

placed in the MRI scanner but not connected to the control computer, (3) *Cable in Place*, where the motorized template and the wires are placed but not connected to the controller, (4) *System Connected*, where the motorized template is placed and connected to the control computer, (5) *System Ready*, where the motorized template is placed, connected, and powered on, and (6) *Motors Running*, where the ultrasonic motors are in constant motion during imaging.

After MRI acquisition, the image data were exported to a workstation (Mac Pro with Mac OS X 10.7.5, Dual 6-Core Intel Xeon 2.66 GHz, 24 GB memory, Apple Inc., Cupertino, CA) using DICOM format. The images were then loaded onto medical image computing and visualization software, 3D Slicer [31] version 4.2.

B. SNR Measurements

There are several methods to determine the SNR of MR images suggested in the National Electrical Manufacturers Association (NEMA) standards and the literature. The most commonly used method among the studies on MRI-compatible robotics is a single image method, where SNR is measured based on two regions of interest (ROIs) on a single image assuming the statistically and spatially uniform distribution of noise [34–36]. Usually, one ROI is defined in the tissue of interest, and the other is in the image background, i.e. air. Although the method takes a practical approach requiring only a single image, it is known that the method does not yield accurate SNR measurements in a phased-array surface coil system and/or sum-of-square reconstruction due to the statistical spatial variation of noise. Alternatively, one can measure SNR without assuming the statistically and spatially uniform distribution of noise with the following methods.

The first method is to measure pixel-by-pixel SNR through the repeated acquisition [37–40]. Because the method relies less on the assumption of statistical and spatial noise distribution than any other method, it is often used as the gold standard SNR measurement. A drawback of this method, however, is that it typically requires a few hundred image acquisitions, making it difficult to use with a slow imaging sequence.

The second method is the “noise-only image” method [36, 39, 41]. The method calculates SNR based on a single or multiple noise-only images, which are acquired without radio frequency (RF) excitation. The noise-only image method was further generalized, where images are reconstructed in “SNR units” based on noise-only images, allowing direct pixel-by-pixel SNR measurements [39]. A disadvantage of the noise-only image method is that most of the clinical MRI system does not allow acquiring noise-only images with standard pulse sequences.

The third method is the “difference image” method, where noise is estimated by calculating pixel-by-pixel differences between two images acquired under exactly the same conditions [36, 42, 43]. Despite several drawbacks of the method, e.g. requirement for stationary tissue and limited precision due to small sample size, the method allows measuring local SNR with a standard clinical sequence. Therefore, we decided to use the difference image method for our SNR measurement.

Based on the method, the following two image sequences used in our clinical trial of MRI-guided prostate biopsy [26, 27] were used for the SNR measurement: 2-dimensional Turbo Spin Echo (2D TSE) (TR/TE = 3000/106 ms; acquisition matrix = 320×204 ; flip angle = 48° ; field of view = 280×224 mm; slice thickness = 3 mm; number of slices = 20; receiver bandwidth = 252 Hz/pixel) and 3-dimensional Fast Low Angle Shot (3D FLASH) (TR/TE = 4.7/2.46 ms; acquisition matrix = 320×168 ; flip angle = 10° ; field of view = 250×188 mm; slice thickness = 4 mm; number of slices = 26; receiver bandwidth = 601 Hz/pixel).

Image acquisitions were repeated twice for each imaging sequence under identical conditions. A total of 24 image data sets (two images for each imaging sequence under six configurations) were acquired from the area where the prostate is located during the clinical procedure. We selected circular ROIs with a diameter of 40 mm on axial slices at 70 mm and 120 mm from the motorized template for all 24 images. Those ROIs represent the nearest and the farthest slices in the prostate gland in the clinical setting. The ROIs are within the region of the phantom. We calculated local SNR within each ROI using:

$$\text{SNR} = \frac{\frac{1}{2} \text{mean}_{\mathbf{r} \in \text{ROI}}(S_1(\mathbf{r}) + S_2(\mathbf{r}))}{\frac{1}{\sqrt{2}} \text{stddev}_{\mathbf{r} \in \text{ROI}}(S_1(\mathbf{r}) - S_2(\mathbf{r}))} \quad (1)$$

where $S_1(\mathbf{r})$ and $S_2(\mathbf{r})$ are signal intensity in a pixel at position $\mathbf{r}$ on the first and second images acquired with the identical imaging sequence and conditions.

The resulting SNR values (Fig. 4) show a noticeable decrease from the *Device in Place* configuration to the *Cable in Place* configuration, and from the *System Ready* configuration to the *Motor Running* configuration. The 3D FLASH sequence showed higher SNR than the 2D TSE sequence. For both sequences, the ROI closer to the motorized template consistently showed lower SNR. Overall, the SNR values decreased 44% and 42% in the *Motors Running* configuration from the *Baseline* in 2D TSE and 3D FLASH images, respectively. Fig. 5 shows representative phantom images from the SNR measurements.

C. Distortion Measurement

Image distortion was quantified by calculating the absolute percentage change in the phantom's diameter from the respective *Baseline* diameter per imaging sequence based on the NEMA standard for 2D geometric distortion[44]. The reported measurements are the absolute maximum percentage difference in each diameter orientation for each configuration setup. As specified in the NEMA standard, four evenly-spaced diameters at 45 degree intervals are used to determine distortion. Because the boundary of the phantom is not explicitly visible in the MR images, the area of the phantom was segmented on the original MR images by thresholding to identify the diameters.

The distortion value was calculated by comparing the absolute percentage difference of a given diameter obtained with the *Device in Place* configuration with the corresponding *Baseline* diameter. For example, in the axial Spin Echo image sequence in the frequency encoding direction, the vertical diameter when the motorized template presents (d_T) is compared to the vertical diameter of the *Baseline* (d_B). The absolute percentage difference was defined as the absolute value of $(d_T - d_B)/d_B$.

Table 2 shows the results of the image distortion measurement. A maximum of 0.4% distortion was calculated in the right-to-left diagonal direction. Overall, no noticeable distortion was observed as shown in Fig. 5.

IV. Targeting Accuracy

A. Experimental Setup

To evaluate the targeting accuracy of the motorized template, we conducted a phantom study in a 3T clinical MRI scanner. A custom-made agar phantom (approximately 100 mm × 100 mm × 100 mm) was placed as a target volume on the prostate intervention table installed on the patient table of the MRI scanner. A Body Matrix Coil (Siemens Healthcare, Erlangen, Germany) was placed to cover the volume of the phantom and the Z-frame. Fig. 7 shows the targeting accuracy evaluation setup.

The Z-frame was attached to the motorized template for calibration using 2D TSE (TR/TE = 3000/116 ms; acquisition matrix = 256×256 ; flip angle = 140° ; field of view = 160×160 mm; slice thickness = 2 mm; receiver bandwidth = 250 Hz/pixel). The image of the Z-frame was transferred to the navigation software and the position of the template in the scanner coordinate system was determined.

The phantom was imaged using a 2D TSE sequence (TR/TE = 5250/100 ms; acquisition matrix = 320×224 ; flip angle = 150° ; field of view = 160×160 mm; slice thickness = 3 mm; receiver bandwidth = 205 Hz/pixel) to plan the placement of the needles. From the phantom image, we generated 20 random targets within the area reachable by the motorized template. The random targets were grouped in four sets of five targets, mimicking four biopsy procedures requiring five needle insertions each.

B. Targeting Accuracy Calculation

We inserted an 18-gauge needle into the phantom through the guide hole created by the motorized template. The part that provides a needle guide hole is detachable so various hole sizes can be provided and changed during a procedure. After insertion of the needles, a confirmation image was obtained using 2D TSE (TR/TE = 2700/106 ms; acquisition matrix = 280×280 ; flip angle = 48° ; field of view = 200×200 mm; slice thickness = 3 mm; receiver bandwidth = 252 Hz/pixel). From the confirmation image, a needle susceptibility artifact (signal void) was manually identified to localize the needle position in the phantom and assess if the needle accurately reached the target. Targeting accuracy was calculated as the distance between the center of the needle artifact and the target in the images. We calculated this distance along anteroposterior and mediolateral axes separately.

C. Targeting Accuracy Evaluation Result

All needle insertions were successfully performed using the motorized template. Fig. 8 shows a screenshot of the 3D Slicer ProstateNav module showing a representative needle confirmation image and the planned target position.

Fig. 9 (a) illustrates the x- (mediolateral) and y- (anteroposterior) direction error results measured from the confirmation images. The overall targeting errors of the motorized template are shown in Fig. 9 (b). An overall average in-plane targeting error of 0.94 mm with a standard deviation of 0.34 mm was measured. The mean targeting error of each registration of the four phantom procedures resulted in a small deviation of 0.17 mm.

V. Discussion

We developed a new motorized needle guide template for MRI-guided transperineal prostate biopsy. It allows gapless needle placement in the MRI scanner without altering the clinical prostate biopsy workflow. The idea behind this device is similar to an actuated needle-holding device with X-Y stages for TRUS-guided prostate interventions developed by Song[45].

There are MRI-guided transrectal biopsy devices also available. The major benefits of the transperineal approach could be summarized in: (1) transperineal biopsy can be performed for those who cannot undergo transrectal biopsy due to rectum removal, (2) prostate cancer tends to occur in the peripheral zone (PZ), which can be considered roughly as the posterior outer layer of the prostate. Therefore, transperineal needles have a higher likelihood of encountering cancer, and require fewer needle insertions to cover a given target volume, (3) prostate therapies are often performed transperineally and it is highly desirable to provide both biopsy and therapy in the same setting.

The motorized template is directly controlled by navigation software, eliminating human error that can be caused by miscommunication between the radiologist performing needle insertion in the MRI room and the navigation operator at the MRI console. It also eliminates the chance of miscounting template hole positions, leading to improved safety for the patient and less stress for the performing radiologist. Since the motorized template is designed to replace the conventional template and its function without radical changes, the current intervention workflow remains, allowing rapid clinical deployment.

In the initial template design stage, we discussed an alternative manually driven device that can provide the same guidance function. However, we concluded that there are two major drawbacks to this approach. First, a manually driven template will still require verbal/written communication of target information, subject to human error. Second, target aiming accuracy will be limited. Both also lead to prolonged procedural time. We believe that motorized guidance overcomes the limitations of a manually driven template and its impact should be clinically evaluated.

A similar purpose device (Utrecht robot) developed by [21] also aims to deliver accurate needle placement for transperineal prostate interventions. Technically, the main difference between the Utrecht robot and our device is that we use ultrasonic motors to drive our device, which provides high accuracy and non-backdrivable stability, whereas pneumatic power is used in the Utrecht robot. Functionally, our device is a still needle guide template that needs fully manual needle insertion and biopsy triggering, whereas the Utrecht robot provides a tapping mechanism for needle insertion.

The evaluation of the impact of the motorized template on MRI was performed under a controlled environment, i.e. without a patient. Because the presence of a patient would greatly affect MR images, the resulting SNR values and image distortion can be different in a clinical environment. Therefore, it is necessary to repeat the measurements with a patient to obtain more accurate values. The use of an imaging coil with the motorized template could also affect safety and image quality. In the current intervention design, we assumed the effect would be minimal because the motors are outside the coil coverage volume. However, further investigation would be required to evaluate any effect.

Comparatively, in [46] the authors reported an SNR study of an ultrasonic motor-actuated transrectal prostate robot in 3T MRI. The results suggested no measurable reduction of SNR in the motor off configuration and a 40% to 60% reduction in SNR with the motors on. A similar study for a pneumatically actuated robotic needle guide for transperineal prostate intervention reported no more than 5% loss in SNR in a Philips 3T scanner [29]. We also found that the SNR under the control condition (*Baseline*) is even lower than the *Device in Place* condition. Although contrary to common sense, this finding is also observed in the similar studies [29, 46].

The accuracy study demonstrates that the motorized template's gapless needle placement achieved 1 mm 2D root-mean-square (RMS) error, which is a higher targeting accuracy than the conventional needle template with needle guide holes in 5 mm intervals. The RMS error and the standard deviations of targeting errors in the x- and y-directions were equivalent or less than the pixel size of the confirmation image (0.875 mm), indicating that the error was hardly noticeable on the confirmation image that is used in the ongoing MRI-guided prostate biopsy. In the clinical procedure, however, needle deflection due to non-uniform tissue stiffness, and post re-registration prostate deformation and rotation caused by needle insertion and patient movement could degrade the targeting accuracy [47].

The diameter of the 18-gauge needle is 1.27 mm and the needle guidance hole of the device is 1.4 mm in diameter. The tolerance allows smooth sliding insertion and minimum friction

when firing the biopsy gun. Manual templates, however, seem to have slightly larger holes, e.g. 1.5 mm diameter for an 18-gauge needle due to possible shrinkage of the plastic material by repeated cleaning and sterilization, i.e. moisture absorbance and heat deformation. The larger tolerance and deformation of the manual templates would result in poor needle insertion accuracy.

In the targeting experiment, an MRI-compatible needle tested for 1.5T MRI was used because no 3T MRI-compatible biopsy needles are commercially available. In clinical practice, researchers often use 1.5T tested needles in 3T MRI interventions [11, 22, 27] because there is no significant MR image degradation observed in 3T imaging. Furthermore, targets are identified from diagnostic MRI without needles, and intra-procedural MRI mainly provides remapping of the targets and confirmation of the needle positions using anatomical marks. Therefore, using a 1.5T tested needle in a 3T MRI-guided intervention seems to have minimal effects.

In conclusion, we developed a motorized MRI-compatible needle guide template that resembles a TRUS-guided prostate template to overcome the problems of limited needle insertion accuracy and human error in the use of a conventional needle guide template in MRI-guided prostate intervention. The evaluation studies indicated that the presence and operation of the motorized template in the MRI bore creates insignificant image degradation and provides submillimeter targeting accuracy. The automated needle guide that is directly controlled by navigation software eliminates human error so the safety of the procedure can be improved.

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Biographies

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Fig. 1. Current prostate intervention system in clinical setup: (a) patient preparation view at the front of the scanner showing the intervention board and leg support, and (b) the template assembly view during the procedure (depicted from rear of the scanner). The patient is in the feet-first supine position.

Fig. 2. (a) CAD model of the motorized template and (b) a detailed mechanism view, showing ultrasonic motors, timing-belt and miter-gear transmission, crossbar mechanism, fiber optic limiter switches.

Fig. 3. Control architecture of the motorized template. The robot control PC, sensor/encoder unit, and motor driver unit are integrated into a mobile cart outside the MR room.

Fig. 4. The plot of the SNR of 2D TSE and 3D FLASH MR images acquired under the 6 different configurations. Circular ROIs with a 40 mm diameter were selected at the position of the prostate gland on the slices at 70 mm and 120 mm from the motorized template (MT).

Fig. 5. Phantom images from the SNR measurements show the impact of the motorized template under various system configurations: 1) *Baseline*, 2) *Device in Place*, 3) *Cable in Place*, 4) *System Connected*, 5) *System Ready*, and 6) *Motors Running*.

Fig. 7. Targeting accuracy evaluation setup in the 3T clinical MRI scanner, showing the motorized needle guide template.

Fig. 8. Screenshot of a needle position confirmation in 3D Slicer ProstateNav module, showing the planned target and the inserted-needle artifact.

Fig. 9.

(a) Plot of x- (mediolateral) and y- (anteroposterior) direction error results of the targeting accuracy evaluation, and (b) boxplots of the in-plane targeting error results as 2D error distances between target position and needle artifact position.

TABLE 1

Specifications of the current conventional template used in clinical procedures and the design specifications of the motorized template.

	Conventional Template	Motorized Template
Horizontal range	70 mm	77 mm
Vertical range	80 mm	102 mm
Control resolution	5 mm	0.001 mm *
Needle guide	Manual	Automatic
Brake	Screw fastening	Non-backdrivability **

* Derived from the encoder resolution and the pitch of the lead screw.

** Non-backdrivability is provided by 1) the high stall torque of the ultrasonic motors and 2) the low pitch of the lead-screw and nut mechanism.

Distortion results showing the absolute diameter changes of phantom images when the motorized template (MT) presents.

TABLE 2

Sequence		2D TSE		3D FLASH	
Imaging plane (distance from MT)		70 mm	120 mm	70 mm	120 mm
Distortion (%)	Horizontal	0.0	0.0	0.0	0.0
	Vertical	0.0	0.0	0.0	0.0
	D1(L-R)	-0.1	-0.2	-0.2	-0.1
	D2(R-L)	0.4	-0.4	0.0	-0.4