Optical Measurement of Needle Insertion Depth*

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***Abstract*— We present an optical encoder to report the insertion depth of surgical needles with sub-millimeter resolution. The work pertains to robot-assisted percutaneous needle placement applications. The motivating application is prostate brachytherapy, a procedure to permanently implant radioactive seeds into the prostate to eradicate cancer. Independently, we developed multiple robotic needle placement systems involving a motorized needle driver with axially loaded friction transmission and also a fully manual needle driver. Both needle driving approaches require encoding the depth of needle insertion. The implant needles have a pattern with reflective and non-reflective stripes of 5 mm in length. To achieve sub-millimeter encoding accuracy, a method based on a linear sensor array has been developed. It yielded a compact and robust optical encoder that is low-cost and potentially disposable. Custom hardware was prototyped, including light source, linear array sensor, needle guide, microcontroller, and USB interface.**

***Index Terms*— medical robotics, optical encoder, needle insertion**

I. Introduction

A. Motivating application

Our motivating application for measurement of the needle insertion depth is prostate brachytherapy.

In contemporary brachytherapy, seeds are delivered transperineally by needles inserted through the holes of a template guide under transrectal ultrasound (TRUS) image guidance. This process has numerous technical and procedural shortcomings, perhaps most importantly the inability to achieve arbitrary needle trajectories. Several researchers have developed robotic assistants to improve on transperineal prostate implants [1], [2].

The needle placement robot system developed by Stoianovici et al. [3] comprised a 3 degree-of-freedom (DOF) Cartesian bridge over the patient, 2 DOF remote center of motion, and 1 DOF needle insertion with a motorized driver using an axially loaded friction transmission. While all stages had redundant encoders, the needle driver lacked precise encoding of the depth of needle insertion.

The needle placement robot system designed by Kronreif et al. [4], [5] comprised two offset X-Y stages which allow positioning and orientation of the needles over the perineum in four degrees-of-freedom (two translational and two rotational), as shown in Fig. 1. The final act of needle insertion is performed through the needle guide by the physician, in order to preserve full control over the act and provide natural haptic feedback during needle insertion. This approach also requires a needle depth encoder.

While successful brachytherapy requires precise insertion depth measurement, neither the contemporary template-based technique, nor the aforementioned two needle placement robots, provide this essential functionality. *We have accomplished this task, by developing a compact optical encoder with custom hardware including light source, linear array sensor, needle guide, microcontroller, and USB interface.*

The structure of the remainder of this paper is as follows: The design requirements are discussed in Section II. The design approach, including alternatives and related work, is presented in Section III. Section IV describes the experimental system that was created to prove the concept and to refine certain key design parameters. These results led to the design and construction of a custom prototype hardware described in Section V.

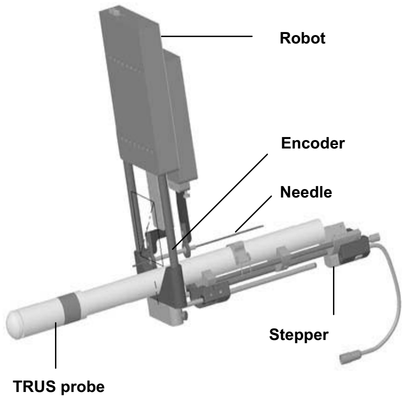

Fig. 1. CAD Model of Brachytherapy Robot (courtesy of Gernot Kronreif, Ph.D., Austrian Research Laboratories)

II. Design Requirements

The objective was to develop an encoder, primarily for the robot depicted in Fig. 1, that would be integrated with the needle guide through which the needle is inserted into the body. Since the final act of needle insertion is left to the surgeon, the robot does not control the insertion axis. Therefore, the encoder allows for precise seed placement in this axis according to the treatment plan. Our objective was to minimize the error due to the sensor. Of course, there are other sources of error, such as needle deflection and organ motion. But accurate encoding allows the system to track

*This work was supported by NIH grant 5R44CA088139-04 and NSF grant EEC-9731478.

the position of the needle tip, and therefore the deposited seeds, with respect to the anatomy by local search in the image, and use this information to update the implantation plan. Since the robot is small and is mounted on the posts that previously supported the template, it is also necessary for the encoder to be small and light weight. The encoder requirements can be summarized as follows:

- The required accuracy is 0.5 mm or better.
- The maximum weight shall be about 50 grams.
- The maximum size must be such as to not interfere with the implant procedure.
- It must meet requirements for electrical safety.
- It should be compatible with the existing needles, which are 18 gauge x 20 cm and have a pattern of black stripes of 5 mm length (see Fig. 2).
- The part of the device in proximity to the needle must be sterile.
- The sterile part of the device should be inexpensive, so that it can be a disposable item (we believe this is most compatible with the surgical workflow for prostate brachytherapy).

Fig. 2. Sample Needle: Mick TP Prostate Seeding Needle, No. MTP-1820-C (Mick Radio-Nuclear Instruments)

III. Design Approach

A. Possible Solutions

We considered two basic approaches for sensing the needle depth: i) attaching an external encoder to the needle, and ii) directly sensing the needle position or motion (i.e., using the needle itself as part of the encoder). We chose the second approach because it promised less interference with the surgical procedure (e.g., there is no need to attach an external sensor to the sterile needle). There are many different physical principles, such as resistive, magnetoresistive and optical properties, that can be used to build an encoder. Due to our requirement for compatibility with the existing needle, an optical solution was the only feasible choice. This has the added advantage of being a non-contact sensor.

B. Different Optical Sensors

The accuracy of measurements using an incremental encoder with quadrature design is completely constrained by the fineness of the pattern (often expressed as the number of lines). As described above, the needles have a pattern of black stripes. Since the stripe width is 5 mm, the achievable resolution with a quadrature encoder is 2.5 mm, which does not meet the required accuracy specification of 0.5 mm.

The solution is to place many sensor elements within one stripe to increase the resolution on the sensor side. In this way, it is possible to get a measurement precision that is better than the stripe width. We considered two approaches for increasing the resolution in this manner:

1) Building a sensor array from a number of discrete quadrature encoders, such as the HEDR-8000/8100 and AEDR-8300 series of reflective optical encoder modules from Agilent.
2) Using a linear array of photodiodes, such as the TSL201R and TSL202R (see Table I) from TAOS (Texas Advanced Optoelectronic Solutions) Inc., in conjunction with a light source.

The first approach had the advantage that the encoder modules include an LED light source, lenses for focusing the light, a photodetector IC that senses the reflected light, and comparators that produce a quadrature output. At least five of these quadrature encoders are needed to get the required resolution of 0.5 mm. The problem is that the encoders are relatively large (about 4 mm in length) and therefore it is difficult to physically place 5 or more of them within one stripe, especially since they would only fit if arranged in a helical pattern around the needle.

We therefore focused on the second approach, which easily provides the high resolution that we require, but necessitates more work in the optical design and signal processing, as described in the following sections.

TABLE I
Specifications of TAOS TSL201R (TSL202R)

Description:	64 x 1 (128 x 1) array of photodiodes
Sensor array length:	8 (16) mm (125 μm per pixel)
Resolution:	200 dpi
Best light sensitivity:	650 - 750 nm (red)
Clock frequency:	5 kHz - 5 MHz
Analog output:	0 V (no light), 3.4 V (saturation), multiplexed

C. Related Work

We are not aware of any previous work in the area of accurate needle depth measurement that satisfies our requirements. There is, however, a large body of work in the area of optical encoders, and some that concerns linear displacement measurement with high resolution. The work that is most closely related to our own is that of Leviton at NASA Goddard Space Flight Center [7], [8] and Tullis et al. at Hewlett-Packard [9]. Leviton designed an absolute encoder that consisted of a two-dimensional array of photodetector elements (CCD sensor) that detected a sophisticated pattern composed of absolute and periodic fiducial markers. This encoder achieved an accuracy of 10 nm for lengths up to 33 cm. Our work differs in that we do not require a sophisticated pattern to be attached to the object (in our case, needle). Tullis et al. extended the concept to the design of linear and rotary incremental optical encoders for injet printers using photosensor arrays and non-patterned targets. In contrast, our approach is to use the existing (at least partly known) pattern on the target and to produce absolute, rather than incremental, position measurements.

IV. EXPERIMENTAL ENCODER SYSTEM

A. Requirements

First, it was necessary to prove the concept by designing and implementing an experimental system that included the mechanical adapter, the signal processing and decoding. An additional goal was to determine the most suitable clock frequency for the TAOS sensor (in a range of 5 kHz to 5 MHz) and lighting conditions. These two parameters influence the sensor integration time. These goals led to the following requirements for the experimental system:

- Flexible hardware that can generate timing signals of different frequencies.
- An A/D converter to digitize the sensor output for subsequent analysis.
- A PC interface and software to acquire data from the hardware.
- Minimize cost by using existing hardware as much as possible.

B. Design and Implementation

1) Light Source: The ideal light source for the encoder has compact dimensions and forms a parallel, narrow beam of rays that spans the width of the sensor array. Furthermore, the light wavelength should be in the range of 650 through 750 nm (visible red light), where the TAOS sensor has the highest responsivity.

Although a linear array of small, precise light sources would have been convenient, such a light source was not commercially available. The available linear light bars were too big and the beams were not sufficiently bright or parallel. We therefore created an array of light sources using discrete LEDs. We tested eight different high-brightness LEDs and a laser diode, and ultimately built an array of 3 Lite-on LTL2H3SEK LEDs (for the TAOS TSL201R), which are red (639 nm) and have a light intensity of 6350 mcd.

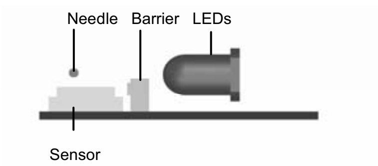

Fig. 3. Barrier

2) Light Guidance: Although a cylindrical lens could form the required ray beam, inclusion of a lens would increase the cost and size of the encoder. Therefore, the experimental system was used to verify that the encoder could work without a lens. It was, however, necessary to introduce a barrier (Fig. 3) to block light from directly impinging on the sensor. The light is guided by the barrier to the needle where it is reflected at an angle of 90 degrees by the curved needle surface onto the sensor array. The black stripes reflect less light than the unmarked needle surface; these differences in intensity are detected by the sensor.

3) Mechanical Adapter: The encoder requires a mechanical adapter to accurately align the sensor, LEDs and needle (Fig. 3). This adapter also integrates the barrier between the LEDs and sensor described above. The adapter was initially constructed from a rapid prototype polymer, but this was not accurate enough. The second version for the experimental system was made of aluminum as shown in Fig. 4.

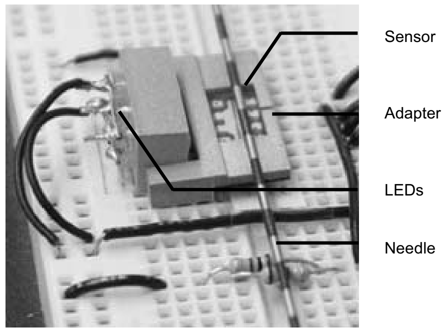

Fig. 4. Experimental Encoder Setup

4) Sterile Interface: The original idea was to use a disposable sleeve around the needle, so that only the sleeve would require sterilization. In addition, the sleeve would protect the encoder from dirt, blood and other contaminants. Several plastic and glass sleeves with rectangular and round profiles were obtained with thin wall thicknesses and dimensions close to the needle diameter of 1.27 mm. The influences on the optical system were tested. The results are described in Subsection IV-C.

5) Hardware: The hardware interface and signal processing were performed by adapting a motor controller board that had previously been developed for a different project [6]. This ISA bus board contains an Altera FPGA of the FLEX 10KE family that provides the ISA interface to different I/O components such as an A/D converter (Maxim MAX125). The board satisfied our requirements for signal generation and processing, A/D conversion and PC interfacing. The board and the FPGA software were adapted for the experimental encoder system.

The FPGA was programmed to provide the following:

1) Exact clocking of the TAOS sensor and the A/D converter.
2) An interface between the encoder and host computer.
3) Storage of the sensor output for later (offline) analysis.
4) Real-time processing of the sensor data.

As shown in Fig. 5, the sensor output signal is amplified and digitized by the A/D converter. Since the conversion time (3 μs) of the A/D converter on the board is not fast enough, it was necessary to skip pixels, where the number of skipped pixels depended on the selected TAOS clock frequency.

Two different designs are implemented in the FPGA:

1) The pixel data are read from the A/D and stored in a FIFO implemented on the FPGA. After all

conversions are complete, the values are transfered to the host computer.

2) The digitized pixel values are directly processed and decoded on the FPGA. The calculated depth is transfered to the host computer. This design is illustrated in Fig. 5.

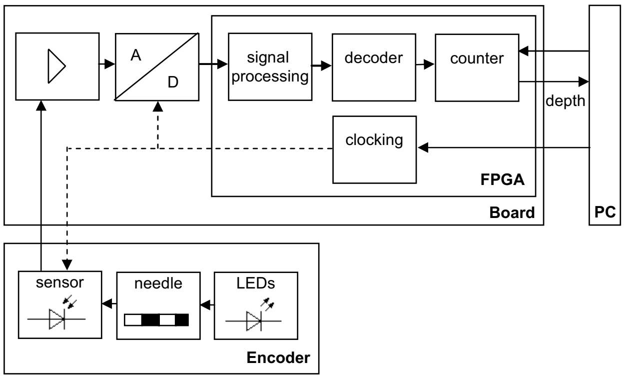

Fig. 5. Block Diagram for Experimental System

6) Signal Processing and Decoding: The experimental system was realized as an incremental encoder that uses the number of pixels of the linear sensor that fit in one stripe. Each of these pixels provides one output channel. This is a generalization of a two-channel quadrature encoder. The decode logic of a quadrature encoder can be described and implemented easily by a state machine. Depending on the sequence of states, the counter stays constant or is incremented or decremented. The state machine becomes more complex as the number of channels increases. Therefore, the developed decode logic for an arbitrary number of channels uses the following two equations:

$$cnt\_en = A \oplus A' \oplus B \oplus B' \oplus \ldots \oplus X \oplus X' \quad (1)$$

$$cnt\_dir = \bar{A}A'\bar{B}' \vee A\bar{A}'B' \vee \bar{B}B'\bar{C}' \vee B\bar{B}'C' \vee \ldots \vee \bar{X}X'A' \vee X\bar{X}'\bar{B}' \quad (2)$$

$$A \,..\, X \; - \; current\ state, \;\; A' \,..\, X' \; - \; previous\ state$$

where cnt_en indicates when the counter must be changed and cnt_dir indicates the direction (increment or decrement). The equation for the direction only describes the states when the needle moves in one particular direction because it is assumed that no states are bypassed. The decode logic was implemented on the FPGA with VHDL for an arbitrary number of channels.

Since the sensor output values have to be converted to binary values, this method is appropriate when a comparator can be used and the detected stripe lengths stay constant so that no further signal processing is needed. We developed an edge detection method based on discrete derivatives. This method, implemented with VHDL on the FPGA, uses the knowledge of the stripe length and does not depend on changes of the level of the black and white stripes and therefore does not require a fixed threshold. However, this analysis of the signal shape can be used directly for the depth calculation as was done for the prototype described in Section V.

C. Experiments and Results

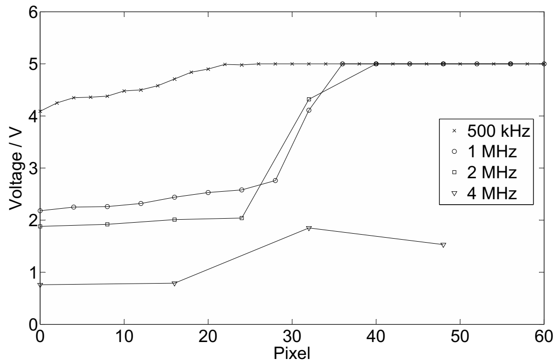

Fig. 6. Output of TAOS sensor TSL201R vs. Sensor Frequency (edge of needle stripe is at pixel 30)

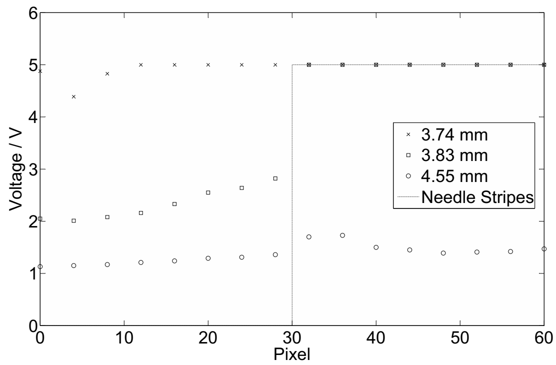

Fig. 7. Output of TAOS sensor vs. Barrier Height

The first experiments demonstrated that the TAOS sensors, illuminated by an array of Lite-on LTL2H3SEK LEDs, are appropriate to detect the needle pattern and to satisfy the accuracy requirement of 0.5 mm.

The experiments also indicated that the best signals (with the sharpest edges) were produced with a clock frequency of 1 MHz and 2 MHz for the TAOS TSL201R and TSL202R, respectively, as shown in Fig. 6. We concluded that the final design should use the TAOS sensor TSL202R (128 pixels), illuminated by an array of 4 LEDs, because it is long enough to guarantee that the output will have a rising and falling edge.

The tested sleeves did not result in satisfactory output signals. We therefore decided to omit the sleeve and instead make the entire encoder a sterile, disposable device. This led to an additional requirement for a low-cost design, as discussed in Section V.

We tested different distances and relations between the sensor, needle, and LEDs to determine the combination that produced the best output signal and therefore defined the optimal adapter dimensions. For example, Fig. 6 shows the sensor output corresponding to different barrier heights. If the barrier is too high (4.55 mm), all the light is blocked whereas if it is too low (3.74 mm), the barrier is ineffective. We chose 4 mm for the prototype design.

Furthermore, our results indicated that for detecting the edges some analysis of the signal was necessary to account for variations in lighting. Therefore, for the prototype, we decided to use this analysis in the decode method, rather than using the equations (1, 2) derived above.

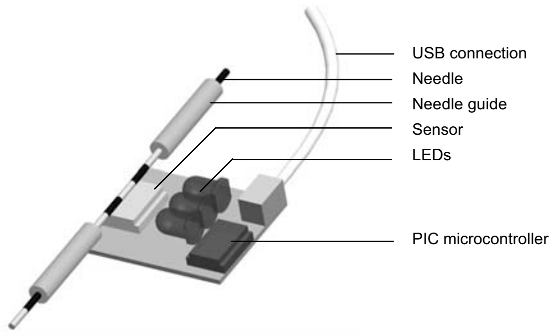

Fig. 8. Prototype Design

V. Custom Prototype Hardware

A. Requirements

After the experimental system proved the concept of an optical encoder for measuring the needle insertion depth, the goal was to design and prototype a custom hardware solution. The following specifications for the prototype were developed. They complete, and render more precisely, the original requirements for the needle insertion measurement described in II. In particular, the prototype shall consist of a small board that includes the encoder, the clocking and the data processing, with the following specific requirements (Fig. 8):

- The whole board shall be disposable, therefore the device must be low cost.
- A microcontroller shall be used for generating the control signals and processing the data. A microcontroller is preferred over an FPGA because it is less expensive and more flexible.
- The board shall contain an USB connection to transfer the measured needle depth to a host PC.
- The board design shall use the adapter dimensions from the experimental system, thereby achieving the tested alignment of sensor, light source and needle.
- The board shall use the linear sensor array TAOS TSL202R, which was shown to have an appropriate length and resolution.
- The board shall include an A/D converter of sufficient speed to sample all pixel values.
- The device shall be designed to allow the use of a comparator instead of the A/D converter. This would reduce cost even further.

B. Design and Implementation

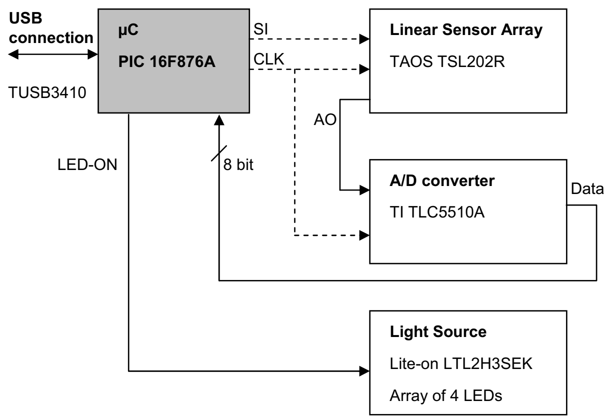

Fig. 9. Block Diagram for Prototype

As shown in Fig. 9, the microcontroller is the central unit of the measurement device:

1) It generates the clock signals and ensures an exact clocking of the TAOS sensor and the A/D converter.
2) It processes and decodes the sensor data and calculates the needle insertion depth.
3) It communicates with the host computer over a USB connection.
4) It controls the state of the LEDs so that the device can be powered up with the LEDs off (this reduces inrush current as required by the USB specification).

The microcontroller PIC 16F876A and the USB interface including the serial-to-USB converter TUSB3410 were obtained from an existing design in our laboratory. Based on the results of the experimental system, an array of 4 Lite-on LTL2H3SEK LEDs and the linear sensor array TAOS TSL202R are used. The flash A/D converter TI TLC5510A was selected – it has a conversion speed of 20 MSPS, which is more than adequate. The design was implemented as a 4-layer printed circuit board (PCB) shown in Fig. 10.

The code for the PIC microcontroller is programmed in the C language. The timing, including the trigger pulses for the sensor output (SI pulses) and the reading of pixel values, as well as the signal processing and depth calculation are implemented as inline assembler because an exact timing and efficient coding are absolutely essential. The TAOS clock of 1.6 MHz is generated by a pulse width modulation (PWM) module in the microcontroller, which was programmed to generate a 50% duty cycle. This frequency is close enough to the optimal value of 2 MHz determined in the experimental system.

The program is structured in two parts:

- The *foreground program* contains the Interrupt Service Routine (ISR) that generates the SI pulses and reads the pixel values. The ISR is started by a timer and guarantees an exact timing.

- The main program contains the *background program* which is activated when a complete dataset is acquired from the A/D converter. It executes the signal processing and calculates the depth. The main program also contains the initialization and communication to the host computer.

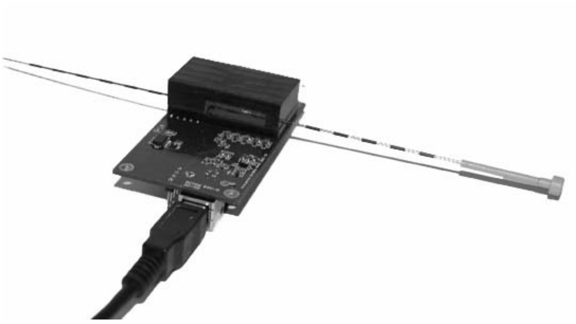

Fig. 10. Prototype for needle insertion depth measurement

C. Signal Processing and Decoding

Since the signal difference between the black and the white stripes is clear, the edges can be detected using a threshold as shown in Fig. 11.

In contrast to the experimental system, the linear sensor array of the prototype is processed as both an incremental and an absolute encoder. For the absolute encoder part, the linear array is used as a fixed scale with sub-millimeter resolution over a length of one stripe. Thereby, the fine position of a detected feature of the needle can be measured. In our implementation, we always detect two edges and use a central position as our feature. The incremental encoder part provides the coarse measurement. In our case, the pattern stripe is constant and known. Therefore, the stripes are counted as they pass through the sensor array. If the length of the pattern stripe varied, the stripe length could be measured by the sensor array and used as an increment.

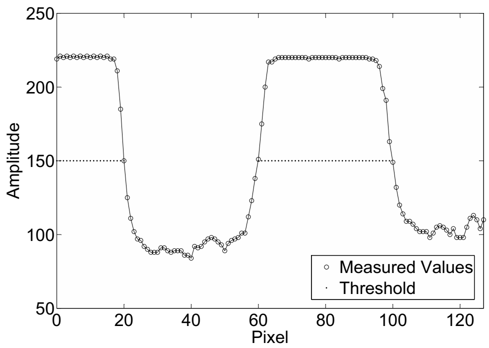

Fig. 11. Sensor Output Signal of Prototype

D. System Validation

The correct function of the absolute and incremental processing of the encoder was validated. To test the absolute encoder part, the needle was moved in steps of $125\,\mu$m over a length of 2 stripes using a micrometer stage. The results indicated that the maximum deviation was $250\,\mu$m. The incremental encoder part was validated by moving the needle back and forth between two distant points for 100 times. The values at each point were constant, which indicates that no stripes were lost.

VI. Conclusion

This paper presented the development of a compact and robust optical encoder that is low-cost and potentially disposable. It allows measurement of the needle insertion depth with sub-millimeter resolution ($\pm 250\,\mu$m) and is compatible with existing needles. The developed measurement method is based on a linear sensor array. Custom hardware was prototyped including the light source, linear array sensor, needle guide and microcontroller. The prototype provides a USB interface to transfer the measured needle depth to a host computer.

Further size and cost reductions of the prototype board are possible. The design of the current board already allows the use of the comparators integrated in the microcontroller instead of the A/D converter. Furthermore, the microcontroller could be programmed via the USB connection, which would eliminate the need for a separate programming connector.

We are currently integrating the encoder with the brachytherapy robot. We plan to use a small sponge to clean the blood from the needle.

Acknowledgment

We thank Hamid Wasti for his assistance with the PCB design and Dr. Iulian Iordachita for his help with the design and fabrication of the mechanical components.

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