A hardware and software protocol for the evaluation of electromagnetic tracker accuracy in the clinical environment: a multi-center study

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Abstract

This paper proposes an assessment protocol that incorporates both hardware and analysis methods for evaluation of electromagnetic tracker accuracy in different clinical environments. The susceptibility of electromagnetic tracker measurement accuracy is both highly dependent on nearby ferromagnetic interference sources and non-isotropic. These inherent limitations combined with the various hardware components and assessment techniques used within different studies makes the direct comparison of measurement accuracy between studies difficult. This paper presents a multi-center study to evaluate electromagnetic devices in different clinical environments using a common hardware phantom and assessment techniques so that results are directly comparable. Measurement accuracy has been shown to be in the range of 0.79-6.67mm within a 180mm^3 sub-volume of the Aurora measurement space in five different clinical environments.

Keywords: electromagnetic tracking, accuracy analysis, evaluation protocol, clinical environment

1. Introduction

Spatial measurement systems play an integral role in many image-guided surgery (IGS) systems. Of the several types of spatial measurement systems, optical tracking systems were the first to make inroads into clinical practice[1,2]. Over the years, as computation capabilities increased, research and deployment of IGS systems has expanded across many medical disciplines. From a clinical standpoint, this change is attributable to the proliferation of minimally-invasive surgery techniques. As incisions become smaller, the use of flexible surgical tools such as guidewires, catheters and small-bore needles has become more prevalent in standard practice. Optical trackers have two key requirements that limit their use: reliance on direct line-of-sight between the optical tracking system and tracked tool and the ability to only track rigid tools. For these reasons, interest in electromagnetic tracking as a technology that can potentially overcome these limitations has begun to grow[3,4].

For all their benefits, electromagnetic tracking systems have not yet found the same widespread acceptance as optical systems. To a large extent this is because of the susceptibility of electromagnetic tracker measurements to distortion by metal and electromagnetic interference sources[5,6]. The quantification of these errors has been a focus of several groups developing IGS systems. Direct comparison of results obtained by different groups is difficult because of the differing approaches taken to devise a hardware phantom and methods to analyze and quantify the data[7-12]. This paper presents initial work towards defining both a hardware phantom and assessment methods to quantify the accuracy of the Aurora

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electromagnetic tracker (Northern Digital Inc., Waterloo, Ontario, Canada). We present results from four institutes. The work is seen as ongoing, with the main intention being to present initial work at defining a standard phantom and method of analysis for the benefit of a wider audience.

At our institute (Georgetown University Medical Center), we have gained extensive experience in the targeting of liver tumors for needle biopsy in swine models using electromagnetically navigated guidance[13-15]. Based on our experience through this work, a set of primary requirements were devised for the design of the hardware phantom and are enumerated below:

1. The phantom must be affordable and easily replicated.
2. The phantom must account for the entire volume of interest, yet not be unwieldy.
3. Electromagnetic device and phantom setup must be simple and efficient.
4. Data collection time should be kept to a minimum.

Based on the body of literature available on accuracy analysis of electromagnetic devices and our experience, a set of requirements for the analysis of the data and quantification of errors were devised and are listed below:

1. Statistics used to define the accuracy must be representative of the underlying nature of electromagnetic measurement errors.
2. The result must be concise yet representative of errors within a volume.
3. Software data collection and analysis tools must be cross platform (portable across operating systems).

Given these requirements, we designed the hardware phantom and analysis protocol described in the following section.

2. Materials and Methods

2.1 Measurement device

This multi-center study was conducted solely on the NDI Aurora electromagnetic tracking system (NDI Inc, Waterloo, Canada). The device consists of a flat field generator, a system control unit that interfaces with a PC, and tracked sensor coils. The device computes the position and orientation of the sensor coils at a maximal rate of 40 Hz based on the relative strength of pulses from the field generator induced within the sensor coil. A single five degree-of-freedom MagTrax sensor needles (Traxtal Technologies, Bellaire, TX, USA) was used for the study. The NDI Aurora device has a manufacturer quoted RMS positional accuracy between 0.7 - 0.8mm and RMS orientation accuracy of 0.3° within a 200 – 400mm radial distance from the field generator[16].

2.2 Hardware phantom

The hardware phantom used for this study is a plexiglass cube of 180mm sides with 225 precisely machined holes from the top face, spaced uniformly apart with 10mm spacing. The holes range in depth from 10mm to 150mm in random order. All holes were machined to a diameter of 1.4mm so as to provide a snug channel for a 5DOF MagTrax needle with an outer diameter of 1.27mm. This design addresses all the requirements enumerated above.

The simple cube design results in reduced machining costs and is easily replicated at other locations as it requires only basic machining capabilities to machine the holes.

The phantom's spatial extent, $180mm^3$, encompasses a sufficient volumetric region for most interventional procedures. This is based on our observations of clinical needle biopsy and tumor ablation procedures conducted within the thoracic-abdominal cavity of human patients. At the same time this size is sufficiently small as to not make the phantom unwieldy.

The dimensions of the cube make it portable and easy to setup within the clinical environment, with no additional hardware components required for phantom placement.

Finally, the choice of 225 spatial samples minimizes the data collection time to approximately 30-45 minutes while still providing a consistent description of the underlying errors. This spatial distribution and sample size were deemed sufficient based on our previous work described in[11]. In that work we evaluated the dependence of various statistics on the number of sampling points uniformly distributed in the volumetric region of interest. The dataset consisted of 2548 data nodes collected within a 240mm^3 volume. Various sample sizes were selected from the larger dataset and the same analysis methods proposed here were used on each subset. The statistic measure for each subset, summarized in Table 1, provides a quantitative outlook on how each statistic varies with the number of sample points. Based upon these results, it was decided that 225 nodes are sufficient to characterize the underlying error distribution, while providing feasible data collection time.

Number of Nodes	2500	1000	500	250	225	100	50
RMS Error	2.38	2.16	2.42	2.38	**2.39**	2.08	1.94
Median Error	1.79	1.59	1.79	1.79	**1.86**	1.63	1.26
Standard Deviation	1.24	1.12	1.33	1.27	**1.24**	1.09	1.13
Max Error	6.97	6.00	6.96	6.08	**5.97**	4.59	5.09
Error Range	6.93	5.88	6.91	6.00	**5.93**	4.56	4.93

Table 1: Summary of effect on statistic measures based on number of nodes selected.

Due to unforeseen mechanical limitations of machining 1.4mm diameter holes to depths larger than 30mm, it was decided to construct the cube from 36 layers of cast acrylic of 3/16 inch thickness, each stacked upon one another to form the cube. Being cast acrylic, the sheets have a width that is variable in the range of + 0.8mm to – 1.5mm. The holes were machined using a EuroLaser M-1200 laser cutting system (eurolaser GmbH, Seevetal, Germany) to a hole diameter accuracy of 0.0025mm. Accuracy of hole position along the X-Y plane is assumed to be the same as the positioning device used during laser cutting with an accuracy of 0.005mm. The depth of each hole is variable because of the large width tolerance of the cast acrylic sheets. Therefore, a Mitutoyo mechanical depth gauge with a resolution of 0.01mm and instrumentation error under 0.03mm (Mitutoyo Corporation, Japan) was used to record the depth of all holes. Using the recorded depth, the end point of each hole was assigned a Cartesian coordinate based on an arbitrarily chosen axes with origin at the vertex closest to hole number 1 (Figure 1). Holes are numbered 1 to 225 and it is assumed that data collection is performed in sequential order as registration between the measured block coordinates and Aurora data is done on similar datasets.

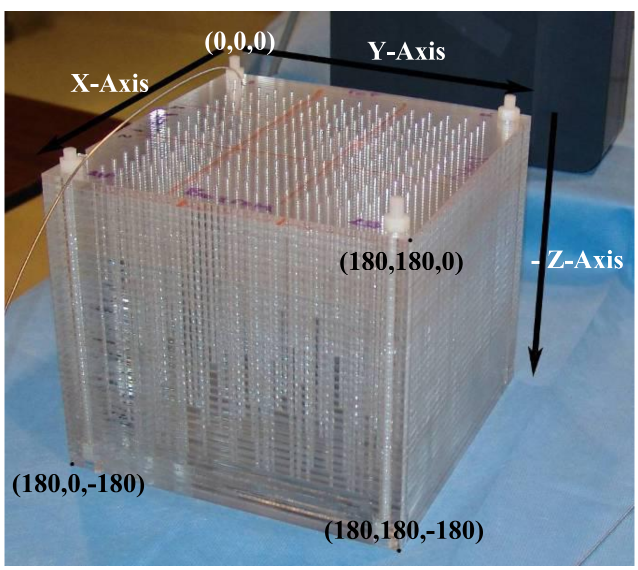


Figure 1: Hardware phantom and its relative coordinate.

2.3 Analysis method

Various error statistics have been used in previous evaluation studies to express the measurement accuracy of the electromagnetic tracking devices. The goal of our analysis method was to provide a concise set of statistics that describe the errors within the volume of interest. In addition to ferromagnetic distortion sources, accuracy of the Aurora device is further hampered by separation between the field generator and sensor coils, such that position errors are non-isotropic and follow a lognormal distribution. This has been established in previous work and position errors have been shown to most closely represent a lognormal distribution[6,11,17]. Therefore, measures such as mean and standard deviation alone are not adequate representations of the error. For this reason, the geometric mean and geometric standard deviation were chosen along with the median, maximal error and RMS error measures. The geometric standard deviation and geometric mean together define the lognormal probability density function in the same way as mean and standard deviation do for a Gaussian distribution. Formulas for all of the statistics used in our analysis are given in Table 2.

(a)
$$\mathrm{E}_m = \sum_{i=1}^{n} \varepsilon_i$$

(b)
$$\sigma = \sqrt{\frac{1}{n}\sum_{i=1}^{n}(\varepsilon_i - \mathrm{E}_m)^2}$$

(c)
$$\mathrm{E}_{rms} = \sqrt{\frac{1}{n}\sum_{i=1}^{n}\varepsilon_i^2}$$

(d)
$$\Pi = \sqrt[n]{\varepsilon_1 \bullet \varepsilon_2 \bullet \varepsilon_3 \bullet \ldots \bullet \varepsilon_n}$$

(e)
$$\sigma_g = \exp\left(\sqrt{\frac{\sum_{i=1}^{n}\left(\ln \varepsilon_i - \ln \Pi\right)^2}{n}}\right)$$

(f)
$$p(\varepsilon) = \frac{1}{\varepsilon \ln \sigma_g \sqrt{2\pi}} e^{\frac{-(\ln \varepsilon - \ln \Pi)^2}{2(\ln \sigma_g)^2}}$$

Table 2: Statistics used in data analysis, (a) mean, (b) standard deviation, (c) RMS, (d) geometric mean, (e) geometric standard deviation, (f) lognormal probability density function.

To minimize the effect of measurement noise, at each collection node, 100 data samples are logged. Using the standard deviation between samples as a marker, 100 samples at each node were found to be sufficient. Additional samples did not provide a substantial improvement in spatial measurement to counter the lengthened collection time, as shown in Table 3.

	No. of Data Samples (at each node)						
	10	*50*	*100*	*150*	*200*	*250*	*300*
Mean StdDev	0.186	0.405	**0.412**	0.410	0.409	0.410	0.409
Min StdDev	0.013	0.015	**0.014**	0.013	0.012	0.011	0.011
Max StdDev	1.413	2.647	**2.641**	2.554	2.495	2.487	2.466

Table 3: Summary of variation in standard deviation based on choice of number of samples.

Absolute position error is evaluated using Horn's point based registration algorithm[18]. Transforming the measured Aurora readings to the arbitrary phantom coordinates and computing the difference between the two provide a representation of position error at each node. Registration nodes are selected in a pseudo-arbitrary manner. A set of eight to ten registration nodes that fall within the "optimal" measurement sub-volume of the Aurora device are selected. It is assumed that these points suffer from minimal distortion and can be used for registration purposes (this is validated during the analysis). Note that the "optimal" location will vary depending on device setup.

In addition to these statistics, a graph depicting the position errors as a function of distance separation between the field generator and sensor and a histogram plot of position errors at all nodes are provided for each test, examples of which can be seen for the clinical suite results in Section 3.2. These plots provide visual confirmation of the nature and distribution of the errors across the volume.

2.4 Data collection and setup method

A software data-logging utility that is portable between different operating system environments was developed using the open-source IGSTK framework [19]. The goal in the longer term is to make the software code and hardware design freely available for download so that other groups wishing to test the accuracy of electromagnetic devices can do so using the protocol set forth here using the same data collection methods.

The first part of the setup process is placement of the phantom such that the occupied volume coincides with a clinical volume of interest. In the following we give an example setup to assess electromagnetic tracker accuracy in a region of space within our Interventional Radiology suite that coincides with the usual location of the liver when a patient lies supine on the couch. This is followed by placing the field generator, affixed onto the Aurora mounting arm, at a location that remains clinically viable. In our experience, in a standard Interventional Radiology suite, we have found a separation of 330mm between front face of the field generator and center of the phantom and field generator at a height of 280mm above the Interventional Radiology patient table to be optimal in terms of minimal distortion from the metal components of the patient table while remaining clinically feasible for liver biopsy procedures. A typical setup of the field generator and phantom within a DynaCT environment at Georgetown University Hospital is shown in Figure 2, where (a) denotes the 280mm distance between the field generator and the table and (b) denotes the distance between field generator face and center of the phantom. The exact positioning of the phantom and field generator will be dependent on the intended clinical application and these distances are intended only as a suggestion. Field distortion from nearby metal components can be greatly different in another clinical environment and it is quite likely that alternate setup positions exist even in our clinical environment that are both clinically feasible and possibly result in reduced distortion artifacts.

Once the phantom and field generator are satisfactorily placed, the MagTrax needle is fully inserted into each hole in the numbered order. The data-logger collects 100 samples at each node and outputs it to individual text files. One data collection pass is completed in approximately 30-45 minutes.

Figure 2: Typical device placement for tests in DynaCT suite at Georgetown University Medical Center.

3. Results

Electromagnetic tracker accuracy evaluation was carried out in two types of environments: 1) baseline tests in a ferromagnetically clean environment and 2) clinical environments. The clinical environment varied between groups and as expected, so did the accuracy of the device due to varying types and amount of unavoidable metallic distortion components specific to each environment.

3.1 Baseline results

Results from tests done in a ferromagnetically clean laboratory environment are summarized in Table 4 (all distance and error measures are provided in millimeters). As it has been established through previous work that electromagnetic tracker accuracy is highly dependent on the separation between the field generator and sensor, the distance between the field generator and center of the phantom is presented within the second column of the table as an indicator of the severity of the measurement errors. The larger errors within the Georgetown environment are due to the tests being conducted in a laboratory with several computers, monitors, fluorescent lighting and other possible sources of interference. While every attempt was made to eliminate metal objects within a one meter radial distance from the field generator, the larger errors within this environment are indicative of unavoidable distortion sources.

Test location	FG-phantom separation	RMS error	Median error	Standard deviation	Max error	Range	Geometric mean	Geometric std dev
Georgetown University - lab	190	1.54	1.06	0.9	6.67	6.5	1.06	1.75
Philips Research	190	0.74	0.49	0.44	4.17	4.12	0.49	1.87
GE Research	150	0.38	0.26	0.23	1.97	1.92	0.25	1.82
GE Research	285	0.32	0.24	0.17	1.8	1.78	0.23	1.8
GE Research	440	1.27	1.02	0.63	3.42	3.33	0.92	1.9

Table 4: Summary of baseline tests conducted at three of the four institutes involved in this study. Second column provides the distance between the Field Generator (FG) and the face of the phantom. Baseline tests were not conducted at one site (CSTAR). All measurements are in millimeters.

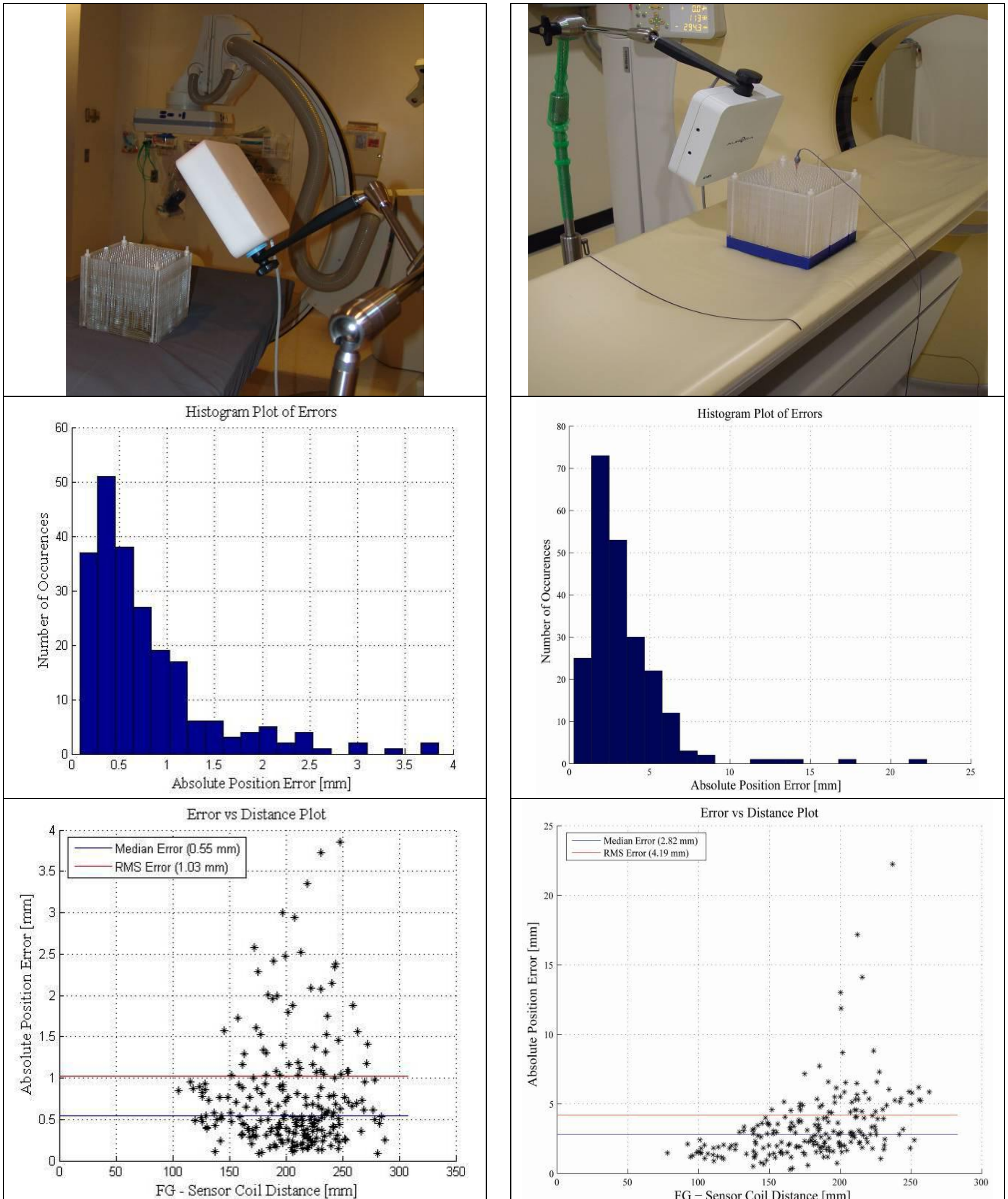
3.2 Clinical suite results

A greater variability in the measurement accuracy of the electromagnetic device is visible within the clinical environment. Generally speaking, the errors were consistently greater within the clinical environment, much as would be expected, except at Georgetown University. The relatively high accuracy in the DynaCT suite at Georgetown University may be attributable to the fact that there are minimal metal artifacts in the vicinity of the measurement volume. The test location is the patient table which is composed of carbon fiber and is suspended further away from the electronic components that drive the patient table stage.

The entire set of results is summarized in Table 5, where all distance and error measures are provided in millimeters. The error-vs-distance and histogram plots and picture of experiment setup of one test from each group are presented in Figures 3 and 4. While it is hard to draw succinct conclusions about the source of measurement errors, these results provide a quantitative outline of the errors under the specific test conditions. In addition, the results also provide insight into the relationship between errors and the distance between field generator and sensor, from which one can better assess what the limitations are for any clinical application that may rely on electromagnetic tracker data within that particular environment.

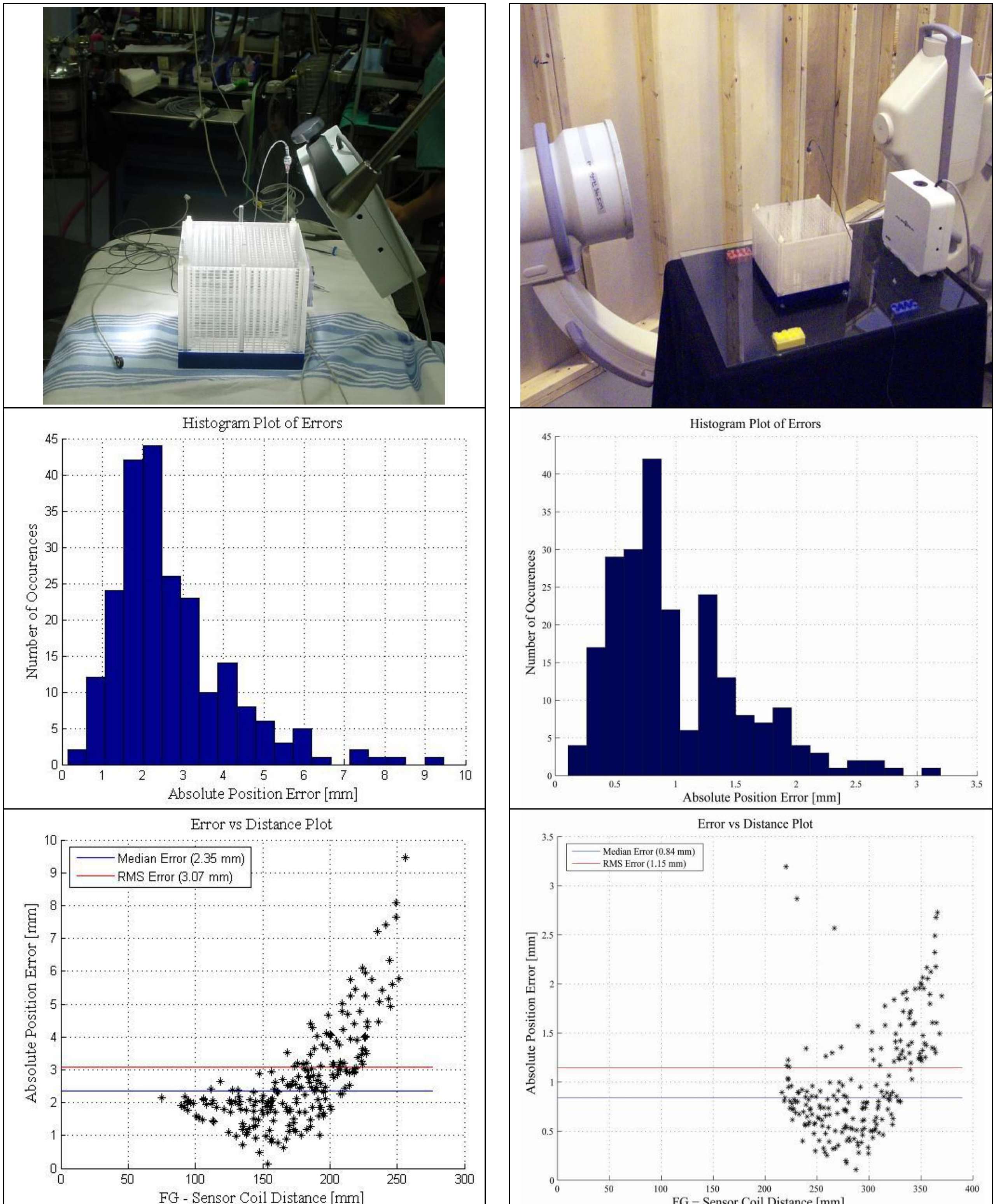
Test location	FG-phantom separation	RMS error	Median error	Standard deviation	Max Error	Range	Geometric mean	Geometric std dev
IR Suite – DynaCT (Georgetown)	245	0.79	0.53	0.43	2.71	2.67	0.54	1.91
Bronchoscopy Suite (Georgetown)	270	6.67	4.98	3.57	20.86	20.31	4.65	1.9
CT Suite (Philips)	205	4.19	2.82	2.53	22.24	21.95	2.74	1.87
OR Suite – no lights (CSTAR Animal OR)	195	3.14	2.39	1.46	9.5	9.37	2.45	1.7
OR Suite – w/ lights (CSTAR Animal OR)	195	3.07	2.35	1.46	9.45	9.31	2.37	1.71
OR Suite – w/ lights and surgical tools (CSTAR Animal OR)	195	2.92	2.29	1.3	8.2	8.01	2.33	1.65
C-Arm (GE)	285	1.15	0.84	0.55	3.19	3.09	0.87	1.75

Table 5: Summary of clinical accuracy evaluation tests conducted at various institutes within different environments. Second column provides the distance between the Field Generator (FG) and the face of the phantom. All measurements are in millimeters.



(a) **(b)**

Figure 3: Summary of results (a) IR suite – DynaCT, Georgetown University and (b) CT suite, Philips Research.



(a) **(b)**

Figure 4: Summary of results within (a) Animal OR suite, CSTAR and (b) C-arm test suite, GE Research.

4. Discussion and Conclusions

It is hoped that this work brings to the fore the need for a common hardware phantom and analysis methodology for evaluating the accuracy of electromagnetic tracking in the clinical environment. We have presented both a hardware phantom and analysis methods whose design is based on our experience in the use of electromagnetic tracking in the clinical environment. Both the hardware and an implementation of the analysis methods were used to assess the accuracy of electromagnetic tracking at four institutes in various laboratory and clinical settings.

A curious difference was the variability in baseline results at different institutes. From the tests at Georgetown University it is clear that while every effort was made to remove any visible metallic objects from a one meter radial distance from the test area, comparison of baseline results indicate that distortion from unavoidable sources such as computers and monitors, overhead lights and wiring conduits within the building, while further than two meters away, still induce a significant amount of distortion within the electromagnetic measurement field. Surprisingly, the DynaCT suite, a clinical environment, yielded the best accuracy in the tests at Georgetown. This may be because the patient table is made of carbon fiber and is suspended further away from the electronic and metal components used to position the couch. In conventional CT and OR suites, the couch positioning components typically rest directly below the patient. Therefore, raising the measurement volume and field generator further up is one of the few changes that can be made. However, the diameter of the CT gantry will place an upper limit on how high the patient location can be raised while still being clinically feasible.

As was expected from the onset of this study, the accuracy results vary widely between clinical environments. However, it is to be noted that the accuracy of electromagnetic tracking systems appears to be within acceptable bounds for a large variety of clinical applications. With a better understanding of the distortion factors in each environment, applications that rely on electromagnetic trackers for navigation can be optimized to work efficiently provided necessary steps are taken. In our experience these necessary step include positioning the field generator and/or measurement volume further away from the distortion source and in rare cases might involve making a change in the clinical protocol such that it is commensurate with the change in location of measurement volume and field generator placement. Given the continued improvement in the measurement accuracy and resilience of electromagnetic tracking systems we expect their applicability within the clinical environment to increase over the coming years. As electromagnetic tracking systems improve in accuracy and better address some of these distortion issues, it will become less important to modify and shape the clinical protocol to facilitate incorporation of electromagnetic trackers.

A significant shortcoming of this iteration of the hardware phantom is the use of several layers to construct the cube. The non-uniform thickness of each layer necessitated the measurement of each hole depth using a depth gauge. In addition, despite the four corner rods that secure the layers, they have a tendency to slide under impact. Some of these mechanical limitations of the initial iteration of the hardware phantom will be addressed and assuaged with the next iteration that will use as few as four layers, wherein each layer is extruded from a thicker Acrylic sheet. Thus uniform layer thickness can be assured and layer slippage can be minimized. Another issue to be addressed as part of this ongoing study is the selection of registration nodes. The ideal solution would be to compute statistics for all possible choices of registration nodes and then select the eight nodes that together provide the lowest error statistic. This however, is far from feasible due to the inordinate length of time taken to cycle through all possible combinations. A possible solution is to select a subset of the entire node set and recursively search through this smaller subset for best registration node choices.

Another shortcoming of the protocol and phantom is the lack of testing of the rotational accuracy of the five degrees of freedom sensors. Here the analysis has been limited to the positional accuracy but the 5D sensors do provide two degrees of freedom for the sensor's orientation. In future work, a set of directional statistics[17] will be included that reflect the rotational accuracy of the sensor. Further, we will investigate phantom designs in which the set of needle holes are not parallel to one another.

Based on these results and results form previous studies at our institute (Georgetown University Medical Center), we feel confident that electromagnetic tracking systems provide sufficient accuracy to be considered for clinical trials. To this end, we are currently pursuing an Institutional Review Board (IRB) human trial to use electromagnetic tracking for image-guided biopsy procedures within the abdominal cavity.

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