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An Ergonomic, Instrumented Ultrasound Probe for 6-Axis Force/Torque Measurement

Matthew W. Gilbertson and Brian W. Anthony†

Abstract— **An ergonomic, instrumented ultrasound probe has been developed for medical imaging applications. The device, which fits compactly in the hand of sonographers and permits rapid attachment & removal of the ultrasound probe, measures ultrasound probe-to-patient contact forces and torques in all six axes. The device was used to measure contact forces and torques applied by ten professional sonographers on five patients during thirty-six abdominal exams. Of the three contact forces, those applied along the probe axis were found to be largest, averaging 7.0N. Measurement noise was quantified for each axis, and found to be small compared with the axial force. Understanding the range of forces applied during ultrasound imaging enables the design of more accurate robotic imaging systems and could also improve understanding of the correlation between contact force and sonographer fatigue and injury.**

I. INTRODUCTION

Ultrasound is used ubiquitously in medicine to safely and non-invasively examine the internal structure of soft tissues. Common applications include musculoskeletal imaging, tumor detection, fetal imaging, and biopsy needle insertion monitoring. Given the extensive use of ultrasound, improvements to the diagnostic capabilities of ultrasound imaging could lead to significant improvements in medical care.

Numerous electromechanical systems have been developed to enable new methods of imaging or to augment handheld imaging. For example, [1,2] present tele-operated systems that enable a remote sonographer to manipulate an ultrasound probe mounted to a multi-degree of freedom (DOF) robotic arm to scan a patient. References [3-7] present single-DOF handheld systems that control probe contact force.

One unique and significant aspect of ultrasound imaging is that it requires physical contact with the patient. The contact force, controlled by the human operator, is not repeatable and produces non-repeatable tissue compression, especially near the skin surface, resulting in ultrasound images that are difficult to reproduce at a later date [8]. When designing electromechanical devices to control or improve the repeatability of probe contact force, it is important to understand typical ultrasound contact forces to ensure that the devices are capable of applying the appropriate range of forces with appropriate resolution. A lack of understanding of required force range and mean values could lead to devices that are over- or under-designed, too fragile or too large.

Ultrasound probe contact force is also important because it is a significant risk factor for workplace injuries among sonographers [9]. In the USA, some 90% of sonographers currently work in pain [10]. A more thorough understanding of probe contact forces could therefore lead to a better understanding of the correlation between force and musculoskeletal injury.

The literature currently lacks a comprehensive study of ultrasound probe forces across a range of different exam types and different sonographers. Salcudean [1] used an instrumented probe to measure the contact forces applied by one sonographer during several carotid exams. Guérin [11] specifies a general 5N-20N range across cardiac, renal, and abdominal exams. A thorough study is needed to fully characterize the forces across sonographers and exam types.

Chadli [12] and Burcher [8] present handheld systems that measure probe contact forces in one axis and six axes, respectively. In this paper, we present a compact, ergonomic six-axis force/torque measuring system that attaches to an off-the-shelf ultrasound probe. The device, which is not much larger than the probe itself, is shown in Fig. 1. We describe force data gathered from thirty-six abdominal ultrasound exams conducted by five professional sonographers.

Fig. 1. Front, side, and back views of the six-axis force/torque-measuring ultrasound probe. Depicted with a 3D-printed ultrasound probe mock-up.

II. Mechanical Design

A. Functional Requirements

We place the following three functional requirements on the design of the force-measuring ultrasound probe:

1) *Unobtrusive*. The device must be compact and lightweight to ensure that it does not alter the way in which sonographers conduct ultrasound exams.

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†The authors are with the Laboratory for Manufacturing and Productivity, in the Department of Mechanical Engineering, at the Massachusetts Institute of Technology, Cambridge, MA. USA. Contact: `banthony@mit.edu`.

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2) *Rapid attachment/removal*. In hospitals, ultrasound probes are highly utilized pieces of equipment, ranging in price from US$5,000-$10,000. The device, therefore, must permit quick attachment and removal of the ultrasound probe so that the probe can be used for other clinical tasks.
3) *Safe*. Must pose no risk to the sonographer or patient.

B. *System Description*

The device is shown in Fig. 1. The sonographer grasps the blue portion of the device and places the ultrasound probe (white) in contact with the patient. A load cell measures the 6-DOF relative forces and torques applied between the sonographer's hand and the ultrasound probe. In Section IV-A, we show that the probe is moved quasi-statically during ultrasound exams. Therefore, the relative contact forces and torques between the ultrasound probe and the patient can be calculated.

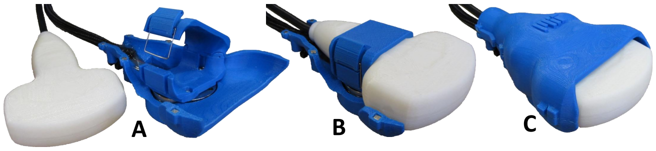

Fig. 2. The ultrasound probe is rapidly attached to or removed from the device by hand, without requiring tools, with mean attachment and removal times of 13 sec and 16 sec, respectively. To attach, the top half of the clamp is lifted (A), probe is inserted and latch is closed (B), the top shell is brought near the bottom shell, magnetically snapping into place (C).

Fig. 2 demonstrates the procedure for attaching the ultrasound probe to the device, and Fig. 3 shows an exploded view of the assembly. The device contains six total fasteners, which secure the hinged, locking probe clamp (**5** and **3**) and the bottom shell **7** to the six-axis Mini40 load cell **6** (ATI Industrial Automation). Twelve high-strength neodymium magnets **2** embedded in the top **1** and bottom **7** shells hold the two shells together with 5 N of force. Ridges present on the top shell prevent shell-to-shell movement in the Y and X directions (refer to Fig. 4, left), while the magnets prevent motion in the Z direction.

A three-axis analog-output accelerometer **9** (Analog Devices ADXL 335, mounted to an Adafruit PCB) is mounted to a recessed shelf within the bottom shell. Cable ties (Fig. 1, top-right) secure the load cell and accelerometer cables to the bottom shell, providing strain relief. The accelerometer is used to measure the orientation of the device with respect to gravity, as discussed in Section III.

This device is custom-designed to fit the GE C1-5-D ultrasound probe **4**, commonly-used for abdominal exams. The probe was 3D-scanned with a NextEngine Desktop 3D scanner at Massachusetts General Hospital (MGH) in Boston, MA. SolidWorks was used to design the clamps and shells (average thickness 2.5 mm), ensuring a shell-probe air gap of no less than 3.6 mm. The five parts were 3D-printed from ABS plastic with a Dimension 3D printer.

Two National Instruments USB-6009 DAQ boards, housed in a shielded electronics enclosure (**4** in Fig. 5), read the load cell and accelerometer voltages. A LabVIEW virtual instrument running on a laptop records data at a rate of 60Hz.

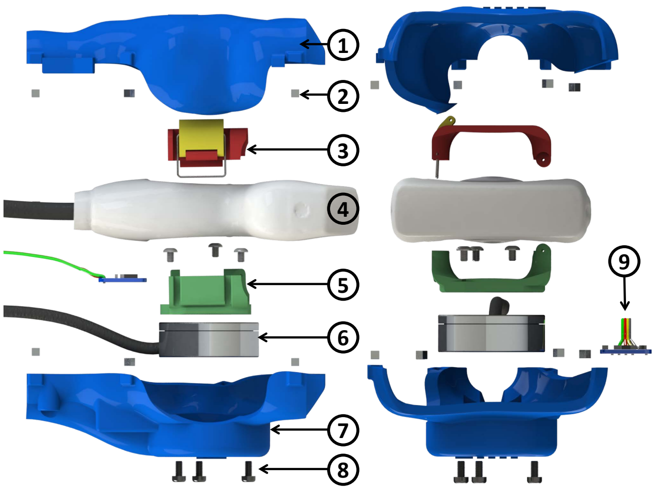

Fig. 3. Exploded view of the device.

The goal of this study is to measure the forces F_x, F_y, and F_z and torques τ_x, τ_y and τ_z that the ultrasound probe applies to the patient's body, based on the coordinate system defined in Fig. 4.

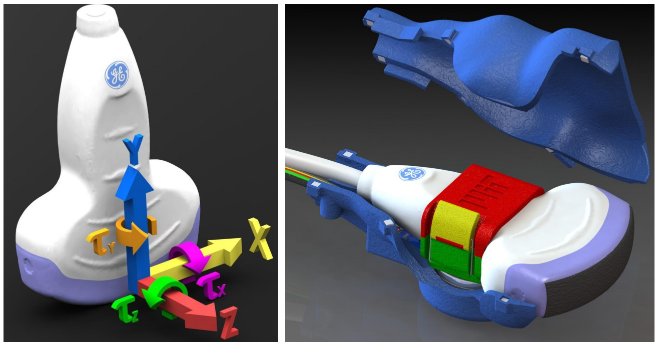

Fig. 4. Left: probe-tip coordinate system. Right: solid model

III. GRAVITY COMPENSATION

The tri-axial analog accelerometer is used to measure the orientation of the device with respect to gravity in order to perform gravity compensation. As the sonographer rotates the device through different angles with respect to gravity, the weight of the ultrasound probe will appear in the measured forces and torques. We compensate for the effect of gravity by subtracting off the weight of the ultrasound probe based on the angle of orientation in order to improve the accuracy of the system.

The three accelerometer output voltages, g_x, g_y, and g_z, are the projections of the gravitational acceleration vector $\overrightarrow{G}$ onto accelerometer X, Y, & Z axes. Assuming negligible inertial effects due to hand tremors (Section IV-A), the equation relating the probe-tip forces and torques $\mathbf{F_P}$ to the load cell readings $\mathbf{F_{LC}}$ and accelerometer readings $\mathbf{A}$ is thus

$$\mathbf{F_P} = \mathbf{F_{LC}} + \frac{m_2}{S}\mathbf{A} + \mathbf{N_T} \tag{1}$$

where sensitivity $S = 300mV/(9.8m/s^2)$, $m_2 = 146g$, and

$$\mathbf{F_P} = \begin{pmatrix} F_x^p \\ F_y^p \\ F_z^p \\ \tau_x^p \\ \tau_y^p \\ \tau_z^p \end{pmatrix}, \mathbf{F_{LC}} = \begin{pmatrix} F_x^m \\ F_y^m \\ F_z^m \\ \tau_x^m \\ \tau_y^m \\ \tau_z^m \end{pmatrix} + \begin{pmatrix} 0 \\ 0 \\ 0 \\ r_y F_z^m - r_z F_y^m \\ r_z F_x^m \\ r_y F_x^m \end{pmatrix}$$

$$\mathbf{A} = [g_x, g_y, g_z, -g_y r_z + g_z r_y, g_x r_z, g_x r_y]^T$$
$$+ [0, 0, 0, -g_y c_z + g_z c_y, g_x c_z, g_x c_y]^T$$

where $\mathbf{N_T}$ represents the contributions from the noise sources, as discussed in Section IV. Quantities with superscript p are probe-tip forces and torques; superscript m denotes measured forces and torques. c_y and r_y are the Y-distances from the load cell origin to the center of mass and probe tip, respectively; similarly, c_z and r_z are the Z-distances. Trigonometry is used to calculate the probe roll and pitch angles from the accelerometer voltages.

IV. MEASUREMENT ACCURACY AND PRECISION

In the system, there exist numerous sources of noise which degrade the accuracy and precision of the force, torque, and angle measurements. These noise sources include involuntary hand tremors, cable tug induced by the ultrasound probe cable, and sensor noise. In this section, we discuss the different sources of noise and evaluate their magnitudes.

A. Inertial Forces Induced by Involuntary Hand Tremors

Here we demonstrate that involuntary sonographer hand tremors contribute negligibly to the measured contact forces and accelerations. As the sonographer grasps the probe, his or her hand will tremor sinusoidally in the X, Y, and Z directions with frequencies $\omega = 5 - 9Hz$ [13] and amplitude A, inducing forces on the ultrasound probe through the load cell and accelerations in the accelerometer. For the moment, consider tremors in the Z-direction only, with the patient as reference ground position. The relative position between the hand-side of the load cell and the patient, $z_1(t)$, is thus $z_1(t) = A\sin(\omega t)$. Compliance k and damping b within the load cell and mounting clamp cause the ultrasound probe to move quasi-independently, with trajectory $z_2(t)$ relative to the patient. The force $f(t)$ measured by the load cell is

$$f(t) = k\,(z_1(t) - z_2(t)) + b\,(\dot{z}_1(t) - \dot{z}_2(t)) \qquad (2)$$

And the probe equation of motion is $m_2\ddot{z}_2(t) = f(t)$. Converting to the frequency domain via the Laplace Transform, (2) becomes

$$F(s) = \frac{m_2 s^2(k + bs)}{m_2 s^2 + bs + k} Z_1(s) = C(s)Z_1(s)$$

For a given frequency $s = \omega$, the maximum value of the measured force is thus $f_{max}(\omega) = c(\omega)A$. Using the results from [13], the worst-case tremor-induced forces occur for $\omega = 8.0Hz$, with amplitude $A = 75\mu m$, giving the result $f(8.0Hz) \approx 0.03N$. This is insignificant compared to the measured forces in Table I, and can therefore be ignored.

B. Accelerations Induced by Involuntary Hand Tremors

As the hand tremors, the accelerometer, which ideally measures only the orientation of the device with respect to gravity, also measures tremor-induced accelerations. To evaluate the magnitude of these accelerations, we take the second derivative of the hand position to obtain $\ddot{z}_1(t) = -A\omega^2\cos(\omega t)$. The maximum acceleration is $\ddot{z}_{1,max} = A\omega^2$. Based on the worst-case values of ω and A from [13], $\ddot{z}_{1,max} \approx 0.20m/s^2$, which is less than 5% of the gravity-induced accelerations (up to $9.8m/s^2$) and can therefore be ignored. Thus, the probe can be assumed to move quasi-statically (relative to the applied forces) during ultrasound exams.

C. Noise From Cable Pull Force

Another source of signal noise is produced by the ultrasound probe cable. Although the ultrasound cable is strain-relieved by Velcro-strapping it to the device cable as shown in Fig. 5, as the probe is moved and rotated through different orientations the cable flexes slightly, resulting in small, non-repeatable forces and torques applied to load cell.

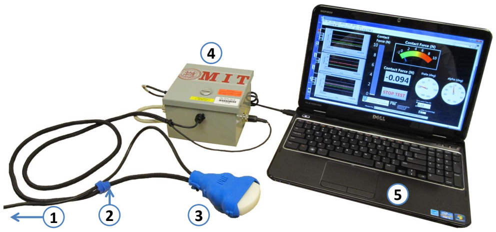

Fig. 5. Photograph of the complete system. The ultrasound probe cable **1** is Velcro-strapped **2** to the device cable to provide strain relief.

D. Noise Sources

The 6x1 vector of the signal noise, $\mathbf{N_T}$, from (1) is comprised of the noise from the load cell, accelerometer, hand tremors, and the ultrasound cable, by

$$N_T = N_{l.c.}^{sensor} + N_{l.c.}^{tremor} + N_{accel}^{sensor} + N_{accel}^{tremor} + N_{cable} \qquad (3)$$

The quantity $N_{l.c.}^{sensor}$, for example, represents the contribution of the load cell ('l.c.') sensor noise to the total noise. The relative contributions from each of the five noise sources were measured and are shown in Fig. 6 for F_y and τ_x.

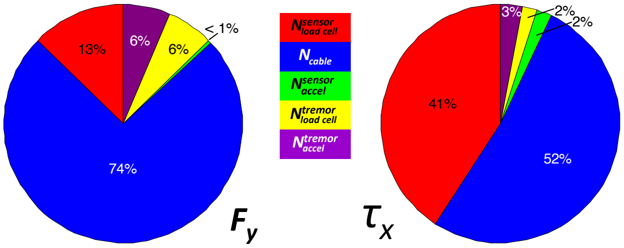

Fig. 6. Contributions from each noise source to noise in F_y and τ_x. F_x and F_z (not shown) are similar to F_y; τ_y and τ_z are similar to τ_x. The non-repeatable tug of the cable comprises the majority of total signal noise, followed by the load cell sensor noise.

V. SONOGRAPHER STUDIES

To evaluate the performance of the device and to collect early use data, 10 professional sonographers used the device to conduct 36 total abdominal ultrasound exams on 5 patients at MGH. Forces, torques, and accelerometer readings were recorded. A plastic bag was placed over the device to ensure sterility (Fig. 7). Mean exam duration was 419 sec. Example force/torque data from the first 9 runs are presented in Fig. 8.

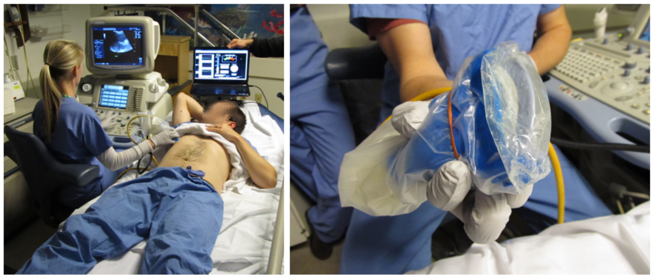

Fig. 7. One of the thirty-six ultrasound exams conducted at MGH - Boston. The device was approved for use by the MGH Internal Review Board.

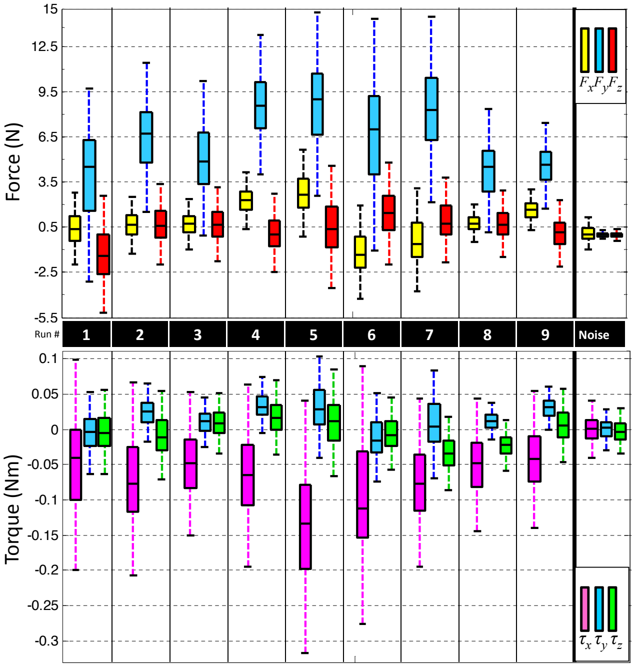

Fig. 8. Box and whisker plot of the forces (top) and torques (bottom) for Runs 1 - 9. Solid horizontal black line within each box indicates median value; boxes enclose 50% of the data. Whiskers extend $\pm 2.7\sigma$ from the median value; assuming a normal distribution, 99.3% of the data fall within the whisker bounds. A noise-measurement run is shown on the right.

Table I summarizes the force/torque data for all thirty-six runs. $\bar{\sigma}$ is the mean standard deviation of the nine runs. σ_{noise} is the standard deviation of the noise. Forces are expressed in N and torques in mNm. The data demonstrate that the mean force along the axis of the probe (Y-axis) is much greater than the forces in the X and Z directions. F_z, τ_y, and τ_z are so small that they lie within the measurement noise. F_x, F_y, and τ_x lie outside of the measurement noise.

TABLE I

SUMMARY OF FORCE [N] AND TORQUE [MNM] DATA FOR THE 36 RUNS.

	F_x	F_y	F_z	τ_x	τ_y	τ_z
Mean	0.3	7.0	-0.01	-190	1	11
Max	5.7	27.3	10.4	460	190	570
$\bar{\sigma}$	1.6	3.0	1.6	110	37	68
σ_{noise}	0.25	0.35	0.45	32	20	20

VI. CONCLUSIONS

This paper presented an ergonomic, instrumented ultrasound probe that measures contact forces and torques in all six axes. The system was used to measure contact forces and torques during 36 abdominal ultrasound exams conducted by 5 professional sonographers. The mean axial force (7.0N) was found to be in close agreement with that measured by Salcudean [1] in carotid exams (6.4N), and within the 5N-20N range specified by Guérin [11] in abdominal, cardiac, and renal exams.

Future work includes using the device in a much greater number of exams of different types with more sonographers in order to develop a more comprehensive understanding of the forces applied during ultrasound imaging.

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