In-Vivo and Postmortem Compressive Properties of Porcine Abdominal Organs

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Abstract. In order to provide realistic haptic feedback, simulators must incorporate accurate computational models of the *in-vivo* mechanical behavior of soft tissues. Surgical simulation technology has progressed rapidly but lacks a comprehensive database of soft tissue mechanical properties with which to incorporate. Simulators are often designed purely based on what "feels about right;" quantitative empirical data are lacking. It is important to test tissues *in-vivo* and apply surgically relevant ranges of force, deformation, and duration. A motorized endoscopic grasper was used to test seven porcine abdominal organs *in-vivo*, *in-situ*, and *ex-corpus* with cyclic and static compressive loadings. Elastic and stress relaxation characteristics were examined. Results from liver are presented here. Notable differences were found between successive squeezes and between conditions for elastic and relaxation behaviors.

1 Introduction

Accurate knowledge of biomechanical characteristics of tissues is essential for developing realistic computer-based surgical simulators incorporating haptic feedback. As simulation technologies continue to be capable of modeling more complex behavior, an *in-vivo* tissue property database is needed. However, little is currently known quantitatively regarding the force-deformation behavior of the abdominal organs, particularly in the ranges applied in real surgery. Such knowledge would be useful not only to simulation but also for optimizing surgical tool design, creating "smart" instruments capable of assessing pathology or force-limiting novice surgeons, and understanding tissue injury mechanisms and damage thresholds.

1.1 Background

The biomechanics of soft tissues that are load-bearing during physiological activities (muscles, tendons, intervertebral discs, cartilage, blood vessels) have been well studied. The soft abdominal organs do not bear significant loads except during trauma and surgery. Very little mechanical testing has been done on the abdominal organs rele-

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vant to laparoscopic surgery, and most of that work has been done on excised animal specimens or human cadavers.[1-5]

It has only recently become a thrust of researchers to obtain *in-vivo* measurements of abdominal organ mechanical properties.[6-10] Each research group has taken a different approach to obtaining material properties, each with different boundary conditions and inherent difficulties. Our previous instrument was capable of applying compressive force via a flat-coil actuated grasper.[6] This instrument was used to test several porcine abdominal tissues *in-vivo* to measure their force-deformation response but was only capable of applying up to approximately 100 kPa compressive stress and did not measure force directly.

It is well known that after several loading cycles, soft tissues typically exhibit a phenomenon known as conditioning, which is a steady-state behavior where the elastic (nonlinear) stiffness and hysteresis stabilize.[11] Most researchers precondition their tissue samples before measuring to obtain consistent results; therefore, first-squeeze behavior of tissues has not been frequently reported. However, surgeons do not precondition tissues before operating. Additionally, it is hypothesized that *in-vivo* mechanical behavior of tissues is significantly different from behavior postmortem, thus justifying the added difficulty of *in-vivo* measurement.

2 Methods

The University of Washington Biorobotics Laboratory has developed a motorized endoscopic grasper (MEG) to examine the compressive properties of porcine abdominal organs (see Fig. 1).[12, 13] Briefly, the MEG uses a geared DC motor to drive a Babcock (Karl Storz) grasper using a cable-and-pulley mechanism. The motor is capable of producing the equivalent of 26.5 N of grasping force (470 kPa with the Babcock) by the end effector jaws at up to 3 Hz. Two strain gage force-sensing beams are mounted in the partial pulley to accurately measure applied force. The MEG can be hand-held and can be inserted into the body through standard endoscopic ports. The force sensor and motor encoder do not directly measure jaw force or jaw angle. However, by knowing the mechanism's inherent stiffness and taking into account the kinematics of the grasper mechanism, a reasonable estimation of the force and deformation at the jaw tips can be obtained. This has been validated by compressing linear springs of known stiffness.

In order to determine the forces, deformations, and rates of compressive loadings to apply, we examined data collected from previous experiments.[14] The mean force applied to the tool handles during tissue grasps was 8.52 N ± 2.77 N. Ninety-five percent of the handle angle frequency content was below 1.98 Hz ± 0.98 Hz. Average grasp time was 2.29 s ± 1.65 s, and 95% of all grasps observed were held for less than 8.86 s ± 7.06 s.

The MEG has been approved by the University of Washington Animal Care Committee for use in non-survival animal experiments in an AALAC-accredited surgical research facility. The device has been used in anesthetized pigs with a standard laparoscopic setup to examine the compressive properties of liver, spleen, gallbladder, small bowel, large bowel, stomach, and urinary bladder. This study presents results

from liver (solid organ). *In-vivo* liver data have been collected from a total of twelve different pigs, *in-situ* from seven, and *ex-corpus* from three. The 3 animals tested *ex-corpus* were also tested *in-vivo* and *in-situ* to examine the change in properties after death. (For the purposes of this report, *in-situ* refers to intact but dead tissue within the body proper, while *ex-corpus* is defined as intact but dead tissue removed from the body.) Fourteen different pigs were tested in all. Weight of the pigs averaged 32.2 kg (range: 25.9-47.7 kg) and the gender was female.

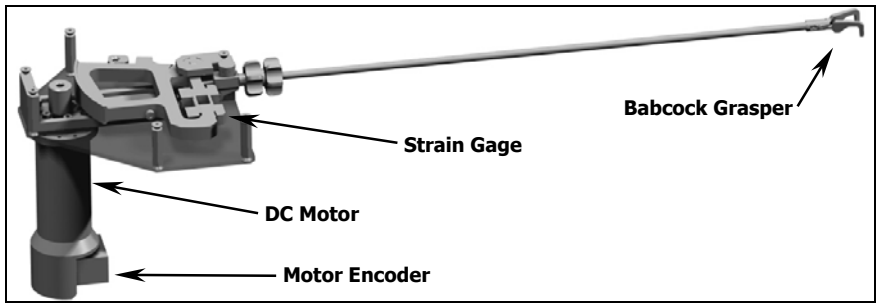

Fig. 1. Motorized Endoscopic Grasper (MEG) (rendered CAD drawing; protective top cover not shown)

While under general anesthesia, organs were grasped with the MEG in various locations with various loading profiles, using a new site for each test regime to ensure the natural (unconditioned) state of the tissue was measured. To emphasize, *no preconditioning was performed on these tissues*. When tests were conducted *in-vivo* and then repeated *in-situ*, effort was made to use different locations for both conditions. Because intact organs were being tested, initial tissue thickness was not controlled. Two types of loads were applied: cyclic and step strains. The cyclic testing consisted of constant velocity squeezes that varied in frequency from 0.1 Hz (loading rate of approximately 5 mm/s) to 2 Hz (100 mm/s), in different tests. Two different types of step strains were applied. "Single" step strains were held for 60 sec at 3 different strain levels. "Periodic" step strains were always held for 10 sec, and then released for times varying from 2.5 to 30 sec (*i.e.*, duty cycles of 80%, 50%, and 25%). These tests were also done at 3 different strain levels. After *in-vivo* testing was completed, the animal was euthanized and time of death recorded, and the protocol was repeated to obtain *in-situ* data. All *in-situ* data were typically collected within 2 hrs postmortem. After *in-situ* data were collected, the abdomen was opened and the organs were removed. Vessels to the organs were cut, so fluids were free to drain. Hollow organs were stapled and then cut to ensure contents remained intact. No other changes to the organs were made. Organs were kept moist with 0.9% saline solution and stored in an ice chest with ice packs. The *in-situ* protocol was repeated e*x-corpus* at three intervals, roughly 4-8, 20-23, and 24-28 hrs postmortem. Organs were kept moist with regular sprays of saline solution during testing.

3 Results

Representative plots of testing results are shown in Figures 2-7. Figure 2 shows the stress-strain behavior of all seven organs tested *in-vivo* with five successive 5 mm/s constant velocity squeezes. Only the fifth squeezes are plotted, which is near the conditioned state (conditioning was observed to occur after 7-10 cycles).

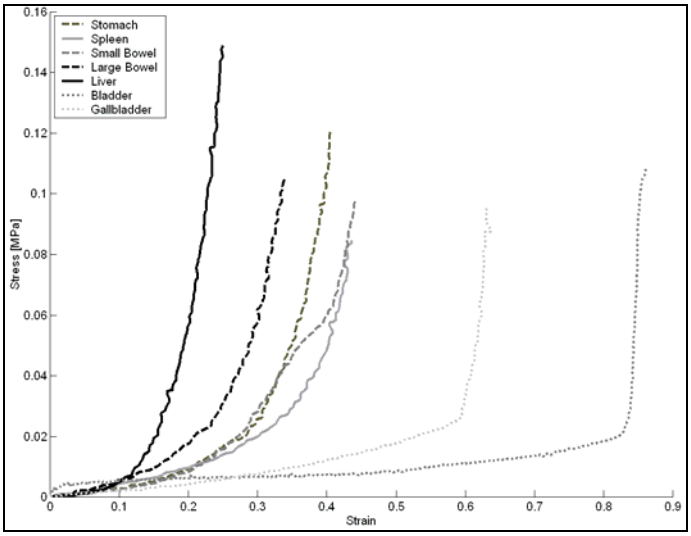

Fig. 2. Representative stress-strain behavior for all organs (bladder, gallbladder, large bowel, liver, small bowel, spleen, stomach) tested *in-vivo* with five successive 5 mm/s compressive loadings: fifth squeeze only is shown

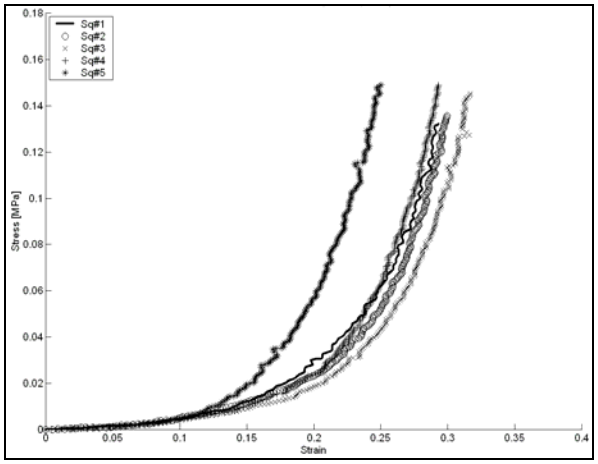

Fig. 3. Stress-strain behavior for one liver tested *in-vivo* (5 cycles, 5 mm/s)

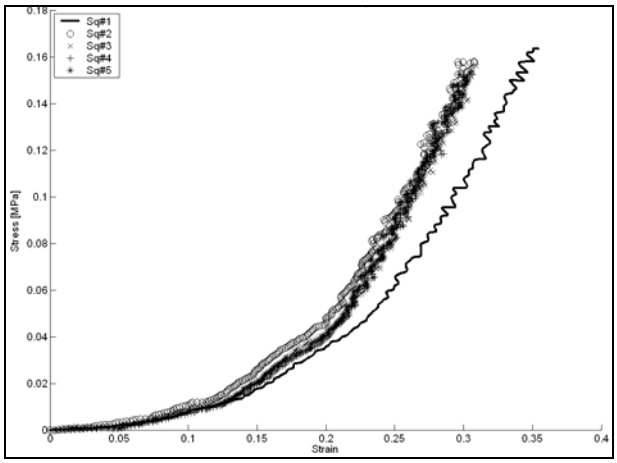

Fig. 4. Stress-strain behavior for one liver tested *ex-corpus* (5 cycles, 5 mm/s)

Figure 3 shows the behavior of one liver tested *in-vivo* with five successive 5 mm/s squeezes, while Figure 4 shows the same liver tested *ex-corpus* (~25 hours postmortem). Overall stiffness of the liver appeared to be similar between conditions, but there is clearly more inter-squeeze variability *in-vivo*; only the first squeeze appeared significantly different from subsequent squeezes *ex-corpus*.

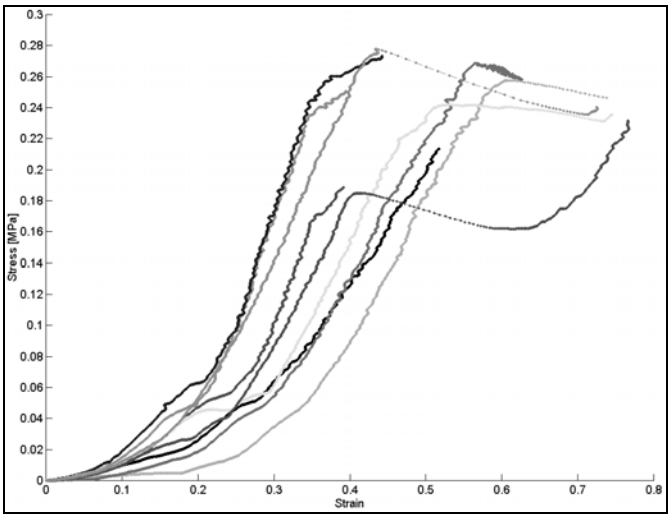

Fig. 5. *Ex-corpus* failure behavior for liver at nine different sites

Liver was also loaded to failure with the MEG during *ex-corpus* testing. Figure 5 shows the failure behavior of liver tested at nine different sites with loading rates of slightly less than 5 mm/s. The failure point (defined as a sudden decrease in stress)

varied between locations. Ultimate strain was between 33% and 60%, while ultimate stress was between 170 kPa and 280 kPa.

Stress relaxation behavior for one liver due to single step strains appears in Figure 6. All organs tested exhibited the well-known decaying exponential normalized stress over time with constant applied strain. The amount of relaxation varied between *in-vivo* and postmortem conditions. Three key observations can be made from Figure 6: 1) greater applied strain resulted in less relaxation, 2) there was typically more relaxation postmortem than *in-vivo*, and 3) steady-state was not reached even by 60 s.

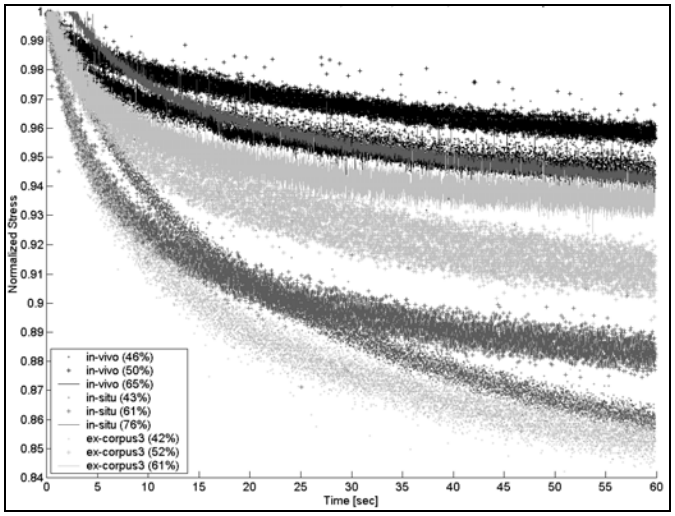

Fig. 6. Stress relaxation behavior for one liver tested *in-vivo* (*black*), *in-situ* (*dark gray*), and *ex-corpus* (*light gray*) each at three different strain levels: low (*circles*), medium (*plusses*), and high (*solid line*)

4 Discussion

Stress and strain were reported in this study. Stress was defined as force normalized to the contact area of the jaw paddles (constant). Strain was defined as 1-λ, where λ is the compression ratio, or deformed thickness normalized to initial thickness. The terms "stress" and "strain" are used loosely, because purely uniaxial stress and strain are not being applied to the tissues. Force and deformation may have been more appropriate measures, but lack of normalization – especially in deformation – makes comparisons between tests impossible, because tissue thickness was not controlled. This study examined *structural* properties of tissues, not *material* properties.

Some organs exhibited drastic differences between successive squeezes, particularly between first and second cycles. For the hollow organs, this is most likely due to compression of movable material within the hollow structure, such as feces or gas or liquid, and is marked by a sudden change in stiffness when the opposing walls touch.

The dependence of the amount of relaxation on the magnitude of strain applied is likely an artifact from the inherent compliancy of the MEG mechanism. Algorithms compensate for the compliancy during off-line analysis, but the PD position controller was not designed to compensate for tissue relaxation. Because of this, as the tissue relaxes, the mechanism unloads and the tissue is strained slightly greater. Therefore, a step strain was never truly applied. This caused some variability in the relaxation results, but the fact that more relaxation was observed postmortem compared to *in-vivo* is likely valid, since there were no pressurized fluids perfusing the tissue postmortem.

Perhaps the most difficult aspect of testing biological materials is the large degree of variability (difference between animals, heterogeneity in the organs, strain history-dependence, strain rate-dependence, etc.). This particular study compounded this problem by testing intact organs *in-vivo*. Testing *in-vivo* introduces potential noise, such as movement artifacts from beating heart and respiration, varying rates of tissue re-perfusion, tremor from holding the MEG by hand, etc. This was evident from the greater variability between squeezes in the cyclic loading *in-vivo*. This greater variability may have been actual tissue behavior (reperfusion between cycles) or simply motion artifacts (squeezing slightly different sites with each cycle). Unfortunately, this variability may mask effects from other factors, such as loading rate. The variability might have been quantified by repeated measures of the same site, but the fact that the tissues exhibit strain history-dependence makes this impractical; the sites would have to be allowed to fully recover to their natural state before subsequent testing, requiring the animal to be anesthetized for extended amounts of time.

While this variability makes finding statistical significance in the data difficult, it does not render the data useless. For the scope of surgical simulation, it is worthwhile to determine ranges of tissue properties. With this information, simulators can realistically change the organs' virtual mechanical behavior so that the virtual liver operated on one day feels different from the next. We are interested in quantifying the forces surgeons feel when grasping organs during actual surgery as a first step toward more realistic surgical simulators.

5 Conclusions

Simulators should include computational models of tissues' response to loads actually applied by surgeons *in-vivo*. Surgically relevant levels of force and deformation can be applied with the MEG to abdominal tissues while measuring the resulting mechanical behavior. Because tissues are not preconditioned during surgery, first-squeeze behavior is important to quantify, as well as how the behavior changes with subsequent squeezing.

We recorded both *in-vivo* and postmortem data in animal experiments using the MEG. Results show nonlinear stress-strain behavior for all tissues tested. Tests included cyclic loadings of varying frequency to observe elastic response, as well as constant and periodic step strains to observe stress relaxation. Notable differences were observed between *in-vivo* and postmortem behavior, making the added difficulty of obtaining *in-vivo* data worthwhile. In future studies, elastic and relaxation data will

be fit with different constitutive models in order to quantify the differences in behavior and for inclusion in surgical simulators.

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