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In Vivo Measurement of Human Tissue Compliance

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ABSTRACT

We present a method for measurement of human tissue compliance in vivo using a commercial haptic interface to apply known step changes in force while recording the resulting displacements. We introduce our system, the Softcometer (SOFT-tissue COmpliance METER). Our motivation was to improve the compliance realism of our Virtual Haptic Back model, but there are many potential applications for this method. We present calibration of the haptic interface, static and dynamic compliance measurement techniques, measurement of contracted muscle compliances, and several important issues affecting our results.

1. INTRODUCTION

The NIH-sponsored Visible Human project (www.nlm.nih.gov/research/visible/visible_human.html) is useful to teach anatomy. We are interested in generating the virtual palpable human, i.e. a virtual reality model of the live human body with high-fidelity graphics like the Visible Human, combined with high-fidelity haptic (force and touch) feedback to the user.

In the Virtual Haptic Back project at Ohio University (Williams et al. 2004), we have a need to measure real, living human tissue compliance properties to ensure maximum realism in our haptic models for manual medicine training. Related fields also require this information: automotive industry, the consumer products industry, physical therapy, and digital human modeling in general. Many biomedical engineering research groups are creating finite-element-based models of live human body components, but are lacking realistic material properties to use in these models.

The problem we are addressing is how to measure real human body tissue properties accurately and quickly, in vivo. The methods should allow for a range of different parts of the body, and a range of humans, including adults, seniors, children, females and males, plus different body types.

In the past, the most common form of human tissue properties measurement has been with cadaver-based measurements. Whether the deceased subject was embalmed or not, this method is inadequate for realistically simulating the behavior of live human tissue.

An exception has been in the dental field where a probe may measure tissue compliance in vivo. Noyes and Solt (1977) presented Bode plots of mobility (peak force/peak velocity) vs. frequency for dental tissue with small forces. More recently, Hartung et al. (2004) have determined soft tissue properties for the human thigh using a special chair, for use in a finite elements model.

The Center for Integration of Medicine and Innovative Technology (CIMIT) has been measuring the properties of organs for virtual physics-based surgery simulation by removing subject organs and exposing them to mechanical displacements and observing the responding forces (www.medicalsim.org). For in vivo measurements there are currently two options: a non-invasive, image-based method examining the strain fields within living tissues subject to force fields; and invasive methods based on measuring the force-displacement responses of tissues (Ottensmeyer, 2002). For invasive methods, laparoscopic methods are common, generally using pigs due to their similarity to human organs. Wang et al. (2000) have developed a sensor for in vivo analysis of multiple-layer buttocks soft tissue to help identify persons subject to pressure ulcers. Edsberg et al. (1999) experimented with human skin in vitro via uniaxial tensile testing, reporting the first compressive-pre-load-induced strain softening of a biological material. EnduraTEC (www.enduratec.com) is involved with all kinds of biological and bioengineering materials studies: teeth, vocal cords, cartilage, artificial heart valves and stents, liver, orthotic heel model, and spinal disc implants. However, most of their materials are

engineered; of the biological tissue studies, all are in vitro or in animal subjects (pigs and cows).

Bruyns and Ottensmeyer (2002) use the TeMPeST 1-D, a voice-coil-motor-actuated machine to measure force/displacement curves in vitro, to determine the mechanical properties of rat organs to support their Virtual Rat Project. Carter et al. (2001) report ex-vivo measurements of pig and sheep liver compliance using a static compliance probe and in vivo measurement of human liver compliance using a hand-held compliance probe during surgery.

Our patent search yielded three related concepts. Randolph (1977) designed a durometer to determine the surface hardness of human tissue for dental and medical use in identifying edema, swelling, puffiness, and distension. Kovacevic (1994) invented a hand-held device for skin compliance measurements in medical and dental cases where tissues must bear loads or swell after treatment. Neurogenic Technologies, Inc., has developed the Myotonometer® (www.neurogenic.com), a hand-held measurement system to determine relative muscle tone, compliance, strength, and spasm.

This article presents experiments to demonstrate our in vivo technique for measuring the compliance of human tissue. Data from this technique can be used 1. To provide realistic haptic properties for the Virtual Haptic Back at Ohio University; 2. To measure the compliance of patients at various points to support clinical diagnosis and treatment; and 3. To measure human body properties for a range of subjects (varying age, gender, and body type) to support industrial and consumer products design. First we present haptic interface details, followed by our static and dynamic compliance measurement techniques and results (including compliance measurement of contracted muscles), and then we discuss and present experiments for several important factors in the effectiveness of our measurements.

2. COMMERCIAL HAPTIC INTERFACE

We have developed a solution for in vivo measurement of the mechanical properties of human tissue compliance in the Virtual Haptic Back Laboratory at Ohio University. The tissue properties required for virtual human models are generally 3D compliance, as defined in (1). Stiffness is the inverse of compliance; we will generally refer only to compliance in this article. The definitions below are general; they may be adapted for specific *X, Y, Z* Cartesian directions, one by one, to obtain the general 3D compliance properties. Units are *mm* for displacement and *N* for force so compliance units are *mm*/*N*. Human tissue is generally nonlinear, non-homogeneous, and non-isotropic, greatly complicating the properties measurement compared to common engineering materials.

$$\text{Compliance} = \frac{\text{Displacement}}{\text{Force}} \quad \left(\frac{mm}{N}\right) \qquad (1)$$

Our method uses two commercial haptic interfaces, both PHANToM® 3.0s (SensAble Technologies, Inc., www.sensable.com), to apply forces and measure displacements in our human subjects at desired compliance measurement points. We can measure the compliance of two points simultaneously with both haptic interfaces and we can also do single point measurement with one haptic interface. We refer to our two haptic interfaces as the 'left' and 'right' PHANToM 3.0s. This section presents the specifications and calibration of our PHANToM® 3.0 haptic interfaces.

2.1 PHANToM® 3.0 Haptic Interface Specifications

From the manufacturer's information, the PHANToM® 3.0 specifications are reported below. This device is capable of exerting forces in *X, Y*, *Z* and measuring displacements in *X, Y*, *Z*. It is capable of covering the points of interest on the subject's back without moving the subject, and it is capable of the forces and displacement resolution we need.

Table I. PHANToM® 3.0 Specifications

Translational workspace	838 x 584 x 406 *mm*
Displacement resolution	0.02 *mm*
Maximum force	22 *N*
Continuous force	3 *N*
Compliance	1 *mm*/N
Backdrive friction	0.2 *N*
Apparent tip inertia	< 159 *g*
Footprint	203 x 203 *mm*

2.2 PHANToM® 3.0 Haptic Interface Calibration

We need reliable *X*, *Y*, *Z* displacement measurements from the PHANToM® 3.0 with sufficient resolution. Since our displacement measurements are taken relative to the initial tip placement on the human body surface, we do not need absolute accuracy in position measurements. The manufacturer reports a 0.02 *mm* displacement resolution for the PHANToM® 3.0 (Table I), which is adequate for our purposes.

Our in vivo compliance measurement methods include exerting force step inputs via the PHANToM® 3.0 in steps of 0.5, 1, 2, 3, 4, 5, and 6 *N*. Our force calibration technique prior to each experiment is to command the PHANToM® 3.0 to exert these levels of force on an external force transducer and ensure that the desired force levels are achieved. This force transducer is the ultra precision mini load cell MDB-2.5 from Transducer Techniques, Temecula, CA. The

resolution of the force transducer is 0.006 *N*. All data reported in this article passed this force calibration test within 0.05 *N* of the desired absolute force, at all force levels directly prior to data collection in each case.

We also need to calibrate the compliance of the PHANToM® 3.0 itself because it is not rigid. Since we are measuring the compliance of the human body, we need to know the compliance of the measuring device since it could affect our results. The less compliant the measuring device relative to the human body compliance, the better. Figure 1 shows the results of a calibration experiment wherein the left PHANToM® 3.0 was commanded to exert the step inputs of force (0.5, 1, 2, 3, 4, 5, and 6 *N*), increasing the force level every 1.5 seconds while pushing on a rigid surface. We expect zero displacement since the surface is rigid; the displacements evident in Figure 1 are due to the compliance of our left PHANToM® 3.0. A linear fit is made to these data resulting in a compliance of 0.39 *mm*/*N* (the slope) with a small *y*-intercept. Averaging four such calibration experiments for the left and also right PHANToM® 3.0s yield average compliance values of 0.37 *mm*/*N* for our left and 0.44 *mm*/*N* for our right PHANToM® 3.0s. From Table I, the manufacturer states that the compliance is 1 *mm*/*N*. The manufacturer must be quoting worst-case compliance results since our measurements, taken near the middle of the workspace, indicate that the PHANToM® 3.0s are significantly less compliant, which benefits our measurements.

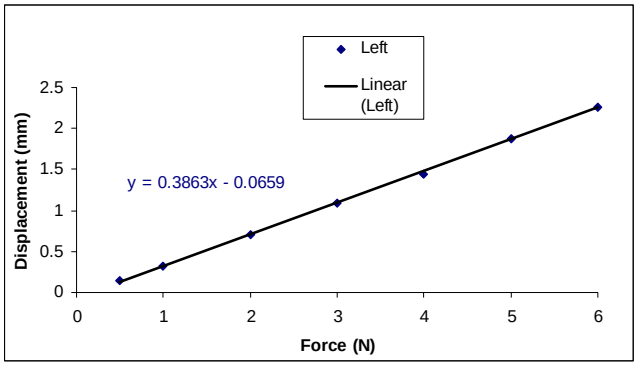


Figure 1. Left PHANToM® Compliance

If the PHANToM® 3.0 is significantly less compliant than the human tissue measured, there will be little error due to this internal measuring device compliance. Assuming a simple series spring model with the applied force acting through the PHANToM® 3.0 in series with the human tissue, the overall equivalent compliance is

$$C_{EQ} = C_P + C_H \tag{2}$$

We can find the human tissue compliance C_H from $C_H = C_{EQ} - C_P$, where the equivalent compliance C_{EQ} is measured (see methods below) and the PHANToM® 3.0 compliances C_P were stated above, for our left and right PHANToM® 3.0s.

3. COMPLIANCE MEASUREMENT METHODS

To date we have used this in vivo human tissue compliance measurement technique for the back, the abdomen, and various points measured for clinical muscle tension studies. In this article we will focus on back compliance measurements.

3.1 Static Compliance Measurement Methods

For our method, the first step is to mark the landmarks at which we wish to measure tissue properties of the subject. The tissue properties measurement method is shown in Figure 2. The subject is prone in this case and we are measuring surface properties of the back at vertebra T7 (this article uses the standard notation of T*n* for the n^{th} thoracic vertebra, plus C for cervical and L for lumbar vertebrae). The seated operator has placed the tip of the PHANToM® 3.0, fitted with a rounded probe the size of a finger pad (partial sphere, 10 *mm* diameter), at the desired location. The haptic interface is commanded to exert seven increasing step levels of force (0.5, 1, 2, 3, 4, 5, and 6 *N* exerted every 1.5 sec). For each force, the displacement into the back is measured by the haptic interface encoders and forward displacement kinematics and recorded by the system automatically. For static compliance measurements we take a single displacement value near the end of each 1.5 sec application time, prior to increasing the input force to another step and repeating the process, while the subject holds her breath. The resulting displacement data are plotted on the vertical axis vs. the force on the horizontal axis. If the result is linear, the slope of this line is the compliance of the back at this point on the subject. If the result is nonlinear, the compliance changes, defined by the slope of the curve at each point. The compliances at this point in the remaining Cartesian directions (in the plane of the back, normal to the direction being measured in Figure 2) are measured in a similar manner.

We call this system the Softcometer (SOFT-tissue COmpliance METER). The measurement tool (PHANToM® 3.0) is accurate and calibrated in *mm* and *N*. Our system is sensitive enough to pick up the heartbeat. Breathing can interfere with the compliance measurements. Therefore, the subject is asked to take three deep breaths in succession, then take half a breath and hold it in, closing the glottis and relaxing all muscles. Then the force is applied and the corresponding displacement recorded. We command the haptic interface to exert the 7 force levels every 1.5 *sec*, and the data are recorded automatically during one breath cycle. Each of these

specifications is considered in more detail later in this article.

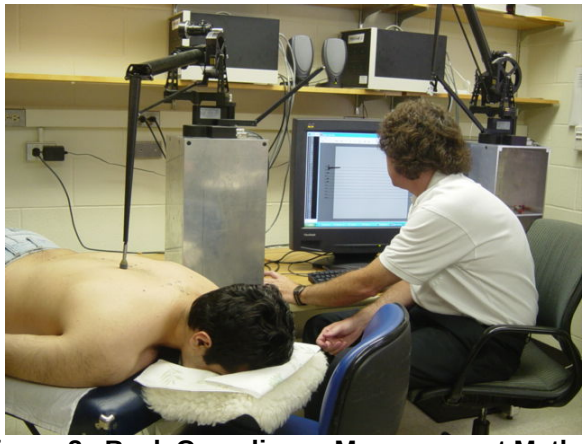


Figure 2. Back Compliance Measurement Method

Figure 3 shows a representative data collection result. Measured displacement is the dependent variable, plotted vs. the independent variable time. The effect of the changing force steps every 1.5 *sec* is evident in Figure 3. At each change in force input, a first-order rise is evident. Why does tissue behave this way? We suspect subdermis fluid moving away from the measurement probe is primarily responsible for this dynamic effect. To generate compliance curves, we record the displacement near the end of each 1.5 *sec* period, just prior to increasing the force for the next step.

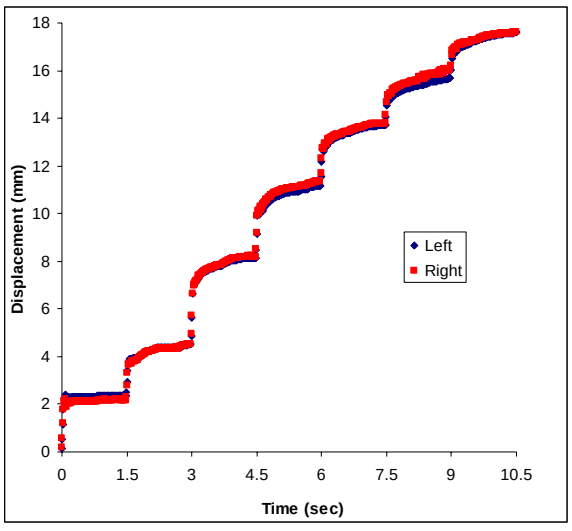


Figure 3. Cervical Compliance Measurement Data

Since backs are 3D surfaces and not flat planes, we have developed a method to command the PHANToM® to exert force in the normal direction to the back at each measurement point, rather than only along a global vertical direction that is not necessarily perpendicular to the back. At each measurement point of interest we use an angle measuring device to ascertain the angles (in two orthogonal directions) of the surface relative to absolute vertical. Then these numbers are entered into the program and the forces are exerted in the desired direction, normal to the back.

Now we present sample data from experiments with the in vivo measurement of back compliance properties using the commercial haptic interfaces. Figure 4 shows the compliance curves (dependent measurement displacement vs. independent applied force) for vertebra L3, including the center (S, for spinous process), 4 *cm* left of center, and 4 *cm* right of center. Figure 5 shows the compliance curves for vertebra T10, including the center (S), 2 *cm* left, and 2 *cm* right.

Both graphs are for compliance normal to the subject's back and include best-fit lines for the data. The compliance with linear fit is the slope of each line. We see in all cases that compliance over the spinous process (S) is fairly linear, while the compliance over the sides are less linear. The L3 compliance (Figure 4) is approximately 1.21 *mm*/*N* over the spinous process and 2.22 *mm*/*N* 4 *cm* to the left and right. The T10 compliance (Figure 5) is approximately 0.53 *mm*/*N* over the spinous process and 1.19 *mm*/*N* 2 *cm* to the left and right. The compliance lines left and right of the spine in Figure 5 are not identical to each other, due to natural asymmetries in the subject's back, but the slopes are very similar. From Figures 4 and 5 we see that L3 is much more compliant than T10, which is expected from anatomy. Also, the boney spinous processes in each case are less compliant than the left and right regions, which are muscular.

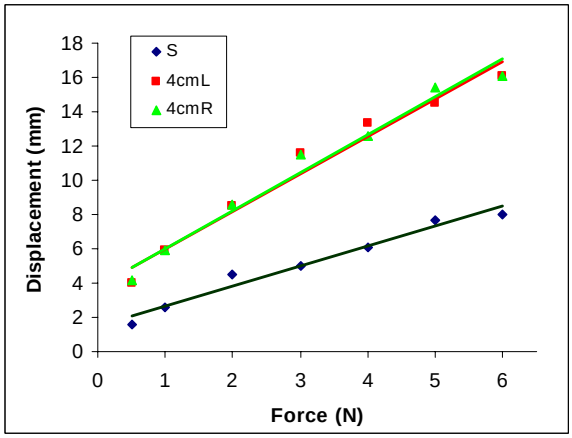


Figure 4. L3 Compliance Results

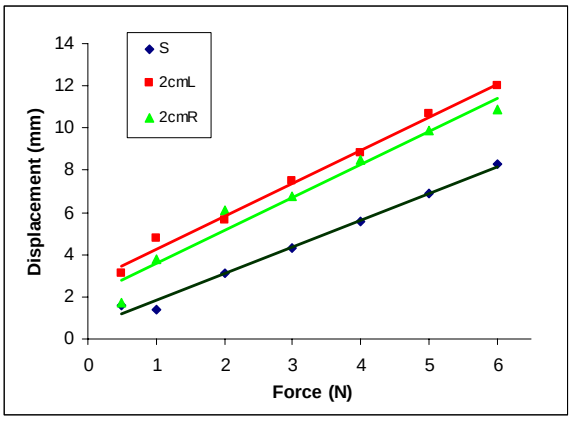


Figure 5. T10 Compliance Results

In Figures 4 and 5 we see that each of the best-fit straight lines is only for seven data points, one for each force step, i.e. we did not include the implied data point of (0,0). Since the data is non-linear this means that the best-fit line does not pass near the origin as it must in the real world. We have three methods to deal with this problem, demonstrated in Figure 6 for the L3, 4 cm L case of Figure 4 only, for clarity: 1. We may simply keep the result of Figure 4 but artificially draw a second line from (0,0) to the left end of the best-fit line, to handle displacements at low force values (less than 0.5 *N*) with a steeper slope (higher compliance); 2. We may include the data point (0,0) and re-derive a new best-fit straight line; and 3. We may fit a non-linear curve to the data, including (0,0) – here we demonstrate a quadratic curve fit.

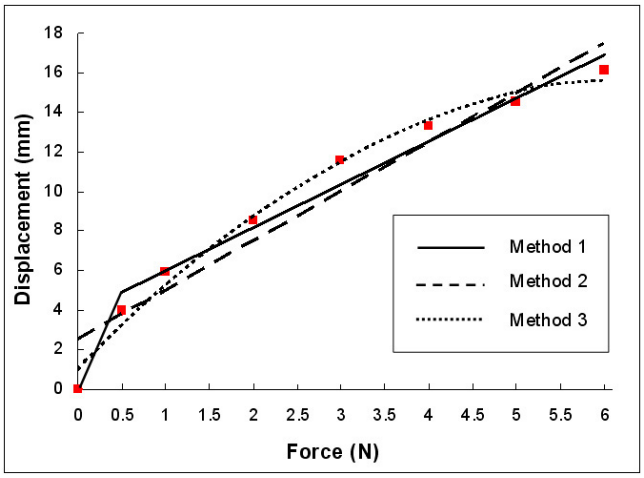


Figure 6. Improved Compliance Curves, L3 4 cm L

Table II summarizes the results from the improved compliance curves shown in Figure 6. For our Virtual Haptic Back purposes, Method 1 is the best, because the best-fit line that does not pass through the origin captures the main compliance behavior in a linear manner, in the force range we need most. This is the method selected in the results presented in this article. However, clearly from Figure 6, the non-linear fit to the data is best for non-linear tissue. The application should dictate the best-curve fitting method.

Table II. Improved Compliance Results

Method	Displacement Function *d*	Compliance	r^2
1	$d = 9.90 f \quad 0 \le f < 0.5$ $d = 2.18 f + 3.86 \quad f \ge 0.5$	9.90 2.18	NA 0.97
2	$d = 2.49 f + 2.54$	2.49	0.94
3	$d = -0.35 f^2 + 4.55 f + 1.02$	$-0.70 f + 4.55$	0.99

Figure 7 shows the compliances associated with Figure 6, for the three methods discussed above.

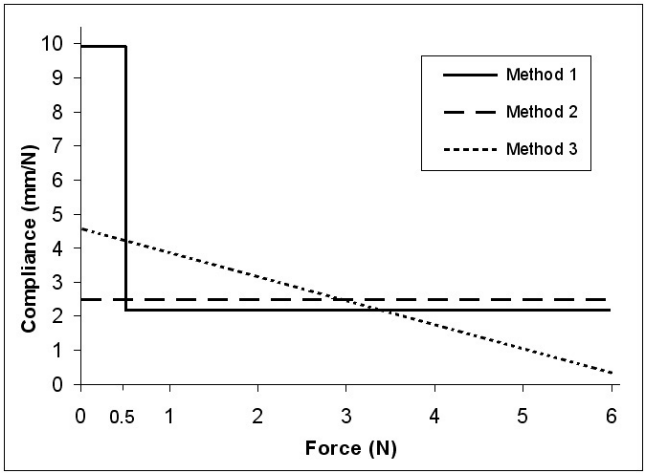
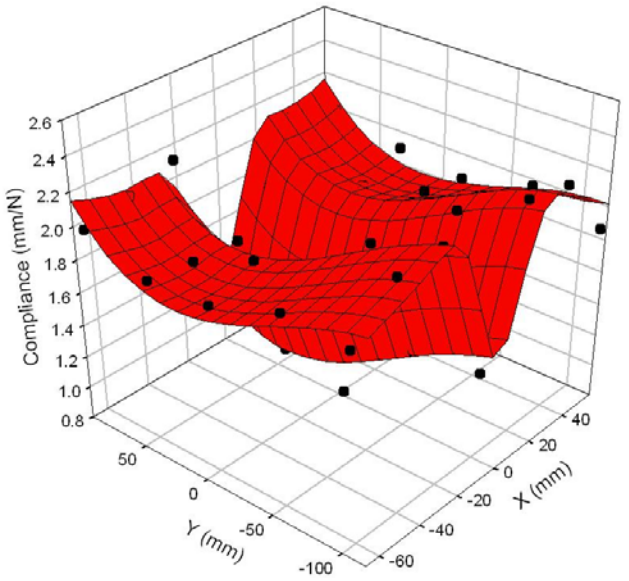


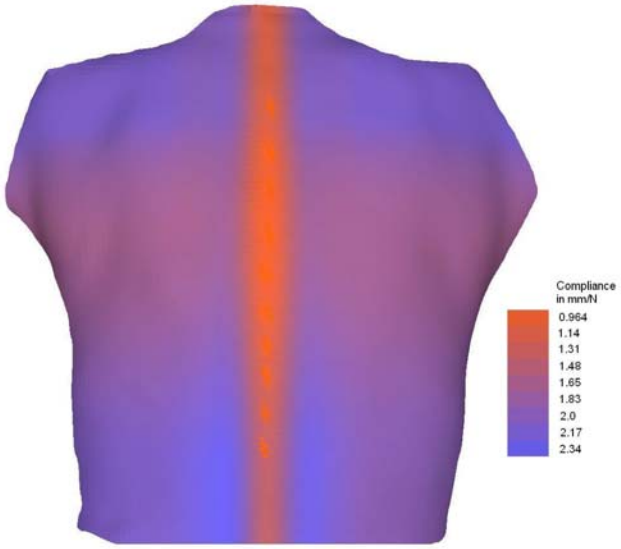
Figure 7. Compliances from Figure 6, L3 4 cm L

Clearly with Method 1 there is a potential problem where the two compliance values change by a step, i.e. we should include a function to smoothly change the compliance in the neighborhood of $f = 0.5$ *N*.

Figure 8 shows a sample result for experimental in vivo back compliance measurements over the entire back of one subject. The same data are shown in two manners, a 3D surface plot (Figure 8a) and a color map (Figure 8b). In Figure 8a *X* and *Y* are the independent back coordinates, while the *Z* data presents the dependent compliance measurements. The dots represent actual data points while the surface is a best-fit surface to these points. As expected, the compliance is lowest along the spinal column and then it varies symmetrically as shown for this particular subject. As shown in Figure 8b, the next lowest compliance regions are along the ribcage. The highest compliances are at the shoulder muscles and in the lower back to the left and right of the spine.



a. Best-Fit Surface



b. Associated Compliance Color Map
Figure 8. In Vivo Back Compliance Results

3.2 Dynamic Measurement Methods

For our Virtual Haptic Back purposes we are primarily interested in static compliance measurements, to improve the haptic realism of our models. For future realism improvements and for other applications we may be interested in the dynamics of tissue, i.e. including a time element in addition to the displacement and force as previously discussed. We can experimentally determine viscoelastic models to represent the dynamic compliance of tissue.

As described above, in our static compliance measurements we apply a given step change in force while measuring the displacement with the haptic interface. Figure 3 shows a sample data result from our static compliance measurements for a given subject. For static compliance we take a single displacement value near the end of the 1.5 *sec* force application time interval, prior to increasing the input force and repeating the process. If we analyze all of the data we can get a dynamic viscoelastic model.

In Figure 3 it appears that the 1.5 *sec* interval is not sufficient, even for the relatively low-compliance area of the cervical vertebrae, especially for higher step input force levels. That is, the displacement curves are still rising at the end of each of the later 1.5 *sec* intervals. However, there is a tradeoff as the subject must hold her breath for the entire 7 force step applications. We will explore this issue further in Section 4.4.

From Figure 3, which is typical of most of our data runs, it appears that a first-order system will capture the dynamic tissue behavior. Though tissue properties are generally nonlinear, we can try a linear viscoelastic model $b\dot{x}(t)+\frac{1}{c}x(t)=f(t)$ for each of the 7 force step level time intervals of Figure 3, where $x(t)$ is the displacement, $\dot{x}(t)$ is the velocity, $f(t)$ is the applied input force magnitude, and b and c are the lumped, constant viscous damping and spring compliance coefficients, respectively. From the experimental displacement vs. time data we can determine the time constants for each time interval. Recall that in first-order systems after three time constants, the displacement rises to within 5% of the final step change displacement value. Thus, measuring the time constant and taking the dynamic spring compliance to be the same as the static spring compliance, we can calculate the viscous damping coefficient b:

$$\tau = bc \qquad\qquad b = \frac{\tau}{c} \tag{3}$$

From Figure 3, assuming the 1.5 *sec* time intervals are sufficient, the seven time constants are approximately 0.21, 0.75, 0.96, 0.96, 0.96, 0.96, and 0.86 *sec*. Other than the first interval with the light 0.5 *N* force, these time constants are similar; ignoring the first, the average time constant is 0.91 *sec*. The average compliance in this case is 2.92 *mm*/*N*. Therefore, the approximate viscous damping coefficient in this example is 0.31 *Ns*/*mm*, a reasonable linear model for Figure 3 beyond 1.5 *sec*.

In some cases we have noted a classical underdamped step response in the experimental data, rendering the first-order model insufficient. In those cases we can instead fit the experimental data to a standard second-order linear system model.

3.3 Contracted Muscle Compliance Measurement

In order to demonstrate that our in vivo tissue compliance measurement is effective for determining

reduced compliance of muscles in various clinical applications, we conducted the following experiment. Using the same basic methods outlined above, we included EMG leads for voluntary contraction feedback to the subject. We asked our expert subject (the third author) to perform various levels of voluntary contraction of muscles (in the lumbar, cervical, and trapezius regions, separately). The subject used the EMG display to hold various levels of voluntary contraction while the haptic interface performed the compliance measurements (all while the subject held his breath). This process is pictured in Figure 9 (the oscilloscope for EMG readings is not clearly visible under the subject's head).

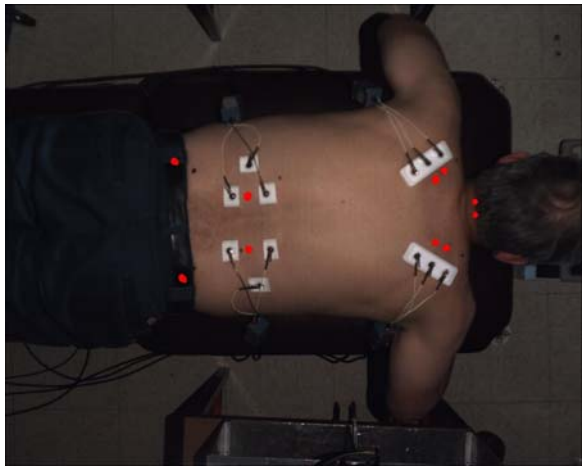


Figure 9. Measurement Points plus EMG Leads

Figure 10 shows the left and right compliance plots for the lumbar measurement region, for a voluntary contraction equivalent to 100 *mV*. We see that the data are nonlinear but may be represented by a best-fit line in the force range 0.5 – 6 *N*. Though the displacements allowed in the subject's lumbar region are significantly different (note the *y*-intercepts of Figure 10), the compliance, i.e. the slopes of the lines in Figure 10, are similar: 1.35 *mm*/*N* for the right and 1.27 *mm*/*N* for the left.

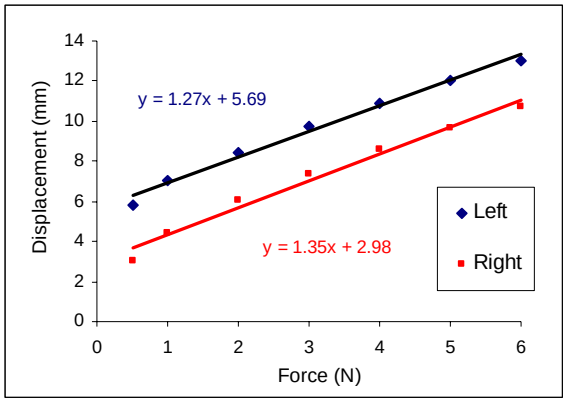


Figure 10. Lumbar Compliance, 100 mV

From the calibration section we found experimentally that the compliances of the measuring devices (PHANToM® 3.0 haptic interfaces) are 0.44 *mm*/*N* for our right and 0.37 *mm*/*N* for our left PHANToMs®, a significant fraction of the overall compliance measured in Figure 10. If the measured compliance is significantly greater from the PHANToM® compliance, the latter may be ignored. If the PHANToM® compliance is a significant fraction of the equivalent measured compliance, then we may apply the correction of (2): the corrected compliance values are 1.35 – 0.44 = 0.91 *mm*/*N* for the right and 1.27 – 0.37 = 0.90 *mm*/*N* for the left. The true results are less compliant than the measured results, due to the PHANToM® compliance. Taking into account the (different) compliances of the right and left PHANToMs®, the (true) measured right and left side back compliances are nearly identical.

Figure 11 presents a typical compliance result in the cervical region (the lumbar and trapezius results are similar) with right and left measurement points and voluntary contractions to create progressively less compliant tissue. In Figure 11, the percentage numbers indicate the percent contraction at each level. In this experiment 400 *mV* corresponded to the maximum voluntary contraction. We see that increased voluntary contractions, leading to tenser tissue, can be measured by our system as reduced compliance.

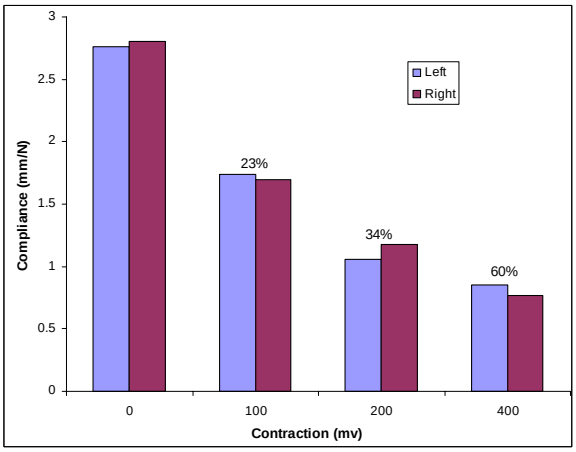


Figure 11. Cervical Compliance with Contraction

4. COMPLIANCE MEASUREMENT ISSUES

This section presents some important issues relating to our compliance measurement methods: reproducibility, seated vs. prone measurements, the effect of thoracic (lung) volume, and the effect of different time intervals for the step changes in force.

4.1 Reproducibility

A crucial aspect of our measurement system is to ascertain if the measurements are reproducible, i.e. if

we measure the compliance at the same point on the same person in the same manner, will we get the same answer (within reasonable limits)? This is complex since the subject may change from day to day and even by time of day so any changes in compliance measurement could be due to non-repeatable measurements, changing tissue in the subject, or a combination.

For the same subject, this compliance test was repeated thrice at different times and on three consecutive days as shown in the legend of Figure 12, for 8 back points (4 on the left and 4 on the right). For the 1st-day test we just did one trial at each point, so there is no standard error bar for that data. Figure 12 shows that there is little compliance measurement differences on different days or different times of day. Our back compliance measurements are thus shown to be reproducible, at least for one subject. The differences in Figure 12 are possibly equally due to subtle changes in the subject as due to measurement inaccuracies.

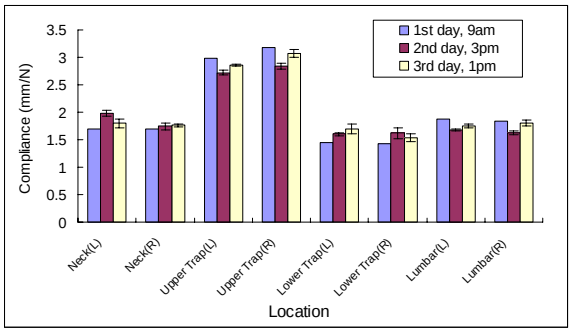


Figure 12. Reproducibility Results

4.2 Seated vs. Prone Compliance Measurements

We are also interested in how the compliance might change for measurements of the same point of seated (Figure 13) vs. prone (Figure 2) subjects. We made an adjustable chair for the seated measurements (Figure 13). Twelve subjects were involved in this experiment, six female and six male. The order of the seated and prone measurements of each subject was chosen randomly. Three points T3, T7, and L3 (all offset 2 *cm* to the right of the spine) on the back of each subject were tested. The compliance at each point was tested four times and averaged. Since the spine curvature is generally different seated vs. prone, the relative angles of T3, T7, and L3 are also different. We adjusted the chair and used pillows to make the subjects' spines as similar as possible seated and prone.

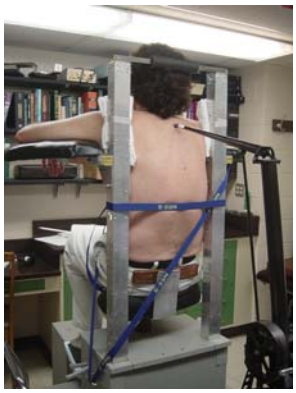


Figure 13. Chair for Seated Back Measurements

Figure 14 shows the seated vs. prone compliance results. We averaged results over all subjects since there was no statistical difference between male and female subject compliances (with a 0.05 significance level).

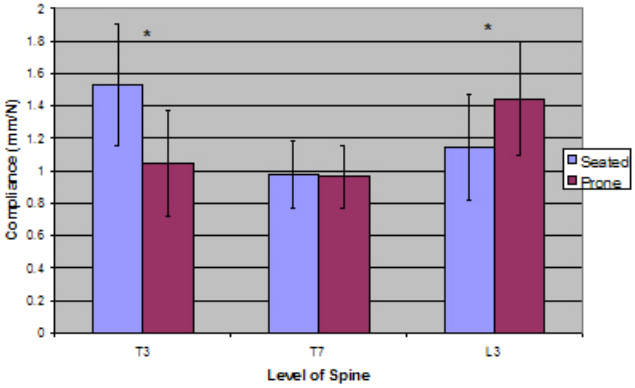


Figure 14. Prone/Seated Compliance Results

Figure 14 is a comparison of paraspinal tissue compliance measurements at the 3 sites in both seated and prone positions with standard deviations bars shown. The asterisk indicates a significant difference ($p<0.05$). Table III summarizes the average compliance results for all 12 subjects for the six conditions.

Table III. Average Prone/Seated Compliance (*mm/N*; standard deviations in parentheses)

	T3	T7	L3
Seated	1.528 (0.372)	0.977 (0.210)	1.143 (0.323)
Prone	1.044 (0.328)	0.964 (0.195)	1.441 (0.351)

The compliance of the upper back (T3) measured prone is less than that seated. The compliances of the middle back (T7) are about the same seated and prone because there is not much muscle change in this area going from seated to prone. The compliance of the lower back (L3) measured prone is greater than that seated.

4.3 Thoracic Volume Effect

Another question we need to address in making reliable tissue compliance measurements is: what is the effect of thoracic volume on the measured compliance? That is, our subjects must hold their breath during all static and dynamic compliance measurements, otherwise the respiration motion interferes with the displacement measurements. Is there an effect of how much breath is held (i.e. thoracic volume) on the resulting compliance measurement?

There were ten subjects in this experiment, five female and five male. Each subject lay facedown on a table and controlled the level of his/her breath by watching a scope to which a chest respiration sensor was connected. Subjects were instructed to reach normal and maximum inhalation levels and two intermediate levels (2x and 3x) were identified. The our static compliance measurements were made 2 cm to the right of vertebrae T3, T7, and L2.

Figure 15 shows average compliance results over all subjects to demonstrate the compliance trends with different breath levels. Generally increased thoracic volume (more breath held) means decreased measured compliance for most subjects, but the effect is very slight and not borne out for the maximum breath level. We did not find any significant gender differences.

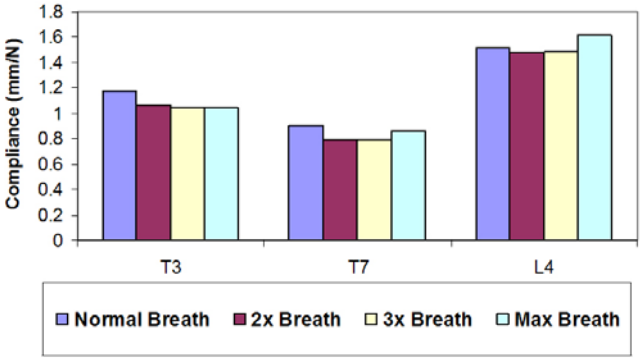


Figure 15. Thoracic Volume Compliance Results

Multivariate tests were used to analyze data from all trials at T3, T7, and L4 to determine if changes in respiratory volume made a significant difference in compliance. No significant differences were noted in the data at T3 ($P=0.444$), T7 ($P=0.518$), or L4 ($P=0.892$) between levels of respiration. The compliance of T3, T7 and L4 were all significantly different ($P < 0.05$).

In the interest of subject comfort, and since there are no significant compliance differences over thoracic volume, we conclude that the normal comfortable breath level should be held for all compliance measurements. All other results presented in this article used the normal breath level.

4.4 Force Step Change Time Interval

As mentioned previously, our compliance measurement technique at a given point involves automatically changing the force command in steps and recording the displacement seven times while the subject holds her breath. This subsection presents the effect of different time intervals of force step changes.

There were ten subjects in this experiment, five female and five male. We tested 5 points (all offset 2 *cm* to the right of the spine) on the back of each subject: T3, the midpoint between T3 and T7, T7, the midpoint between T7and L3, and L3. At each point, the compliance test was repeated with different time intervals of 0.5, 1, 1.5, 2, 2.5 and 3 *sec*. Figure 16 shows a typical result of the experiment at one test point (L3) for one subject with the six different force step time intervals.

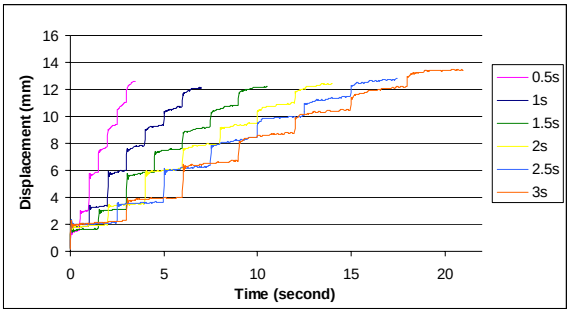


Figure 16. Data with Different Force Time Intervals

The data of Figure 16 were analyzed to generate the best-fit static compliance lines of Figure 17, using displacement values near the end of each force step time interval. The slope of each best-fit line is the compliance determined for that particular time interval. Though some of the line intercepts vary, the slopes are very similar, indicating that there is not a strong effect of force time interval on compliance.

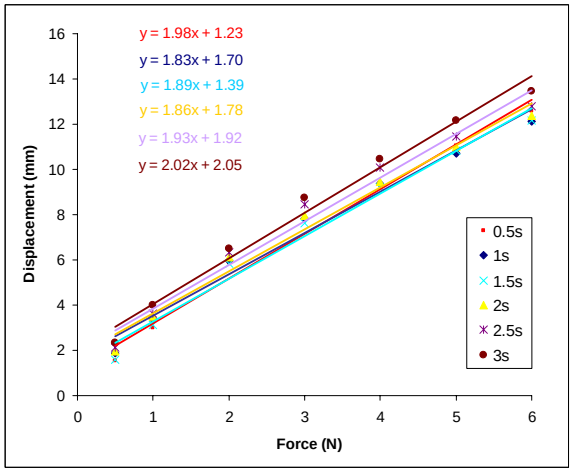


Figure 17. Compliance Lines for Figure 15

The compliance values of Figure 17 are plotted in Figure 18 vs. the six force step time intervals. We do not present any composite data in this experiment due to variation. The case shown is typical.

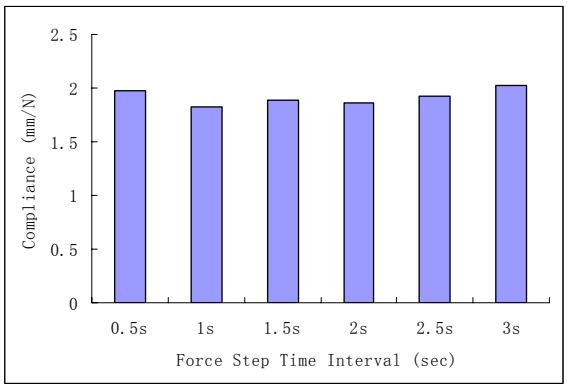


Figure 18. Effect of Time Interval on Compliance

Since there is no strong effect of force time interval on measured compliance we can choose any convenient time interval. The shorter the time interval, the more comfortable for the breath-holding subjects and the more data we can obtain in the same laboratory time. However, the longer the time interval, the more certain we are that tissue dynamics does not interfere and the recorded displacement value is the proper one (i.e. not increasing any longer). Therefore, we choose a time interval in the middle of the range considered, 1.5 *sec*. This is the value used in all other results presented in this article. As we saw in Figure 3, a force step time interval of 1.5 *sec* can be borderline in terms of the displacement settling to a final value in time.

4.5 Resting Time between Compliance Tests

This test, with four subjects, determined the effect of different resting times between successive compliance measurements (as opposed to the time interval between force step changes used in one compliance measurement, considered in the previous subsection). Three test points were chosen on the subject back (neck, lower trapezius, lumbar), each offset 2 *cm* to the right of the spine center. At each point, the compliance test was repeated 4 times (trials) with the same resting time interval between compliance measurements. We used the average of the last three trials as the result at each point because the first trial did not have any waiting interval. Then we repeated this procedure at the second and third back points. After testing the three points in this manner, the waiting time interval was increased. We used waiting time intervals of 5, 10, 20, 40, and 60 *sec*. Figure 19 shows a typical result for one subject, with standard error bars over the trials. Each group of columns displays the compliances of three back points with the same waiting time interval.

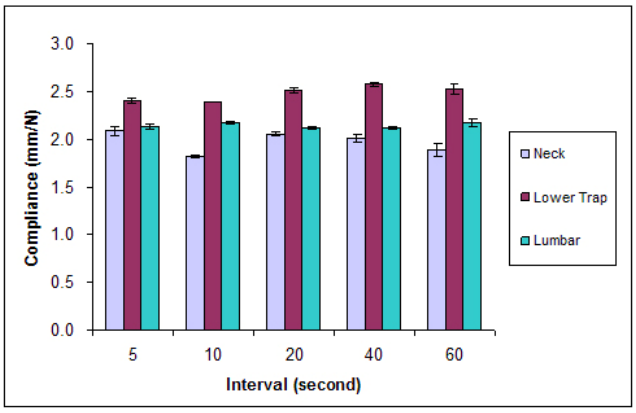


Figure 19. Compliances vs. Waiting Time Intervals

From the data of Figure 19, typical of all subjects, we do not see significant differences in measured compliance over the waiting time interval. Thus, we may use whatever waiting time interval is convenient in the laboratory for each measurement. All other results presented in this article were obtained without controlling the waiting time interval.

5. SUMMARY

This article has presented our methods for in vivo measurement of human tissue compliance using our Softcometer (SOFT-tissue COmpliance METER). We use PHANToM® 3.0 haptic interfaces to exert a series of known force levels at each point of interest while the subject is immobile and holding her breath while relaxed. The PHANToM® measures the associated displacements, from which compliance curves are automatically generated by the computer. We use this information to improve the haptic realism of our Virtual Haptic Back model (used for training medical students in palpatory diagnosis at Ohio University), but this type of information is useful in various industries and applications.

We presented our static and dynamic compliance measurement techniques, with sample results including with voluntary muscle contractions to simulate compliance measurements of contracted muscles. We demonstrated that our method can measure different voluntary muscle contraction levels, indicating it will also be effective for clinicians measuring muscle tone clinically where muscle compliance is a concern.

We also discussed several important issues related to our in vivo measurement techniques. Our method was shown to be reproducible over different days and times of the day. Compliance characteristics vary for different back points in seated vs. prone subjects. The thoracic volume effect was shown to decrease compliance as more breath was held; therefore we use only the normal breath level. The effect of time intervals between applied force

steps was shown and we compromised on an intermediate value of 1.5 *sec*. There was no effect of waiting time interval on successive compliance measurements.

Our in vivo human tissue compliance measuring method may be extended to other parts of the human anatomy in addition to the back. This method can be used by biomedical researchers, industrial ergonomic designers, and clinical medical personnel.

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