Measurement of Interaction Forces During Abdominal Palpation

by

Andrew Geddes

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Department of Mechanical and Aerospace Engineering
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Abstract

Patient simulators hold much promise for the future of medical education and training. By providing repeatable and safe training scenarios, patient simulators not only improve skills beyond the scope of traditional training methods, but more importantly, do not put real patients at risk. Currently there are no patient simulators in existence, or being developed, that deal with abdominal palpation. Also, there is no published data on the interaction forces occurring during palpation of the abdomen. A testing methodology was validated and used to gather force and position data directly from a doctor's fingertips during abdominal palpation. The resulting stiffness plots, maximum displacements, and forces may be used to construct a prototype abdominal sign simulator capable of mimicking a healthy male abdomen.

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Table of Contents

Abstract	ii
Acknowledgments	iii
Table of Contents	iv
List of Tables	x
List of Figures	xi
1 Introduction	1
1.1 The Problem	1
1.2 The Solution	2
1.3 The First Step	3
2 Patient Simulators	5
2.1 Background Information	5
2.1.1 The Benefits of Simulation	6
2.1.2 Simulator effectiveness	7
2.2 The State of the Art	7
2.2.1 Simulation Example	7
2.2.2 Simulator Response	9
2.2.3 Look, Touch and Feel	11

2.2.4	Skin Colour	13
2.2.5	Skin Temperature	14
2.2.6	Skin Texture	15
2.2.7	Virtual Patient Simulators	16
2.2.8	Abdominal Sign Simulators	17
2.2.9	Why is an Abdominal Sign Simulator Needed?	18
3	Abdominal Palpation and How to Measure it	19
3.1	Background	19
3.2	The Abdominal Examination	20
3.2.1	For What are Doctors Feeling?	21
3.3	Obtaining Abdominal Stiffness	21
3.4	What is Not Being Measured	23
4	Fingertip Force Sensors	24
4.1	Background	24
4.1.1	Force Sensor Desired Characteristics	24
4.1.2	General Characteristics of Force Sensors	25
4.2	Common Technologies used in Tactile Force Sensors	27
4.2.1	Piezoresistive	27
4.2.2	Piezoelectric	27
4.2.3	Capacitive	28
4.2.4	Optical	28
4.2.5	Skin-like Sensor Arrays	29
4.2.6	Fingernail	29
4.3	FlexiForce™ and FSR™	30
4.3.1	FSR™	30
4.3.2	FlexiForce™	32

4.4	Sensor Choice	34
4.5	Sensorized Gloves	36
5	Force Sensor Testing	37
5.1	Experimental Protocol	37
5.1.1	Loading Conditions	38
5.1.2	Pucks	39
5.2	Force Sensor Test Apparatus	42
5.3	Test Set-up	47
5.4	Force Sensor Repeatability	50
5.4.1	Background	50
5.4.2	120N vs. 4.5N Sensor	52
5.5	Force Sensor Test Data Analysis	55
5.6	Force Sensor Test Results	56
5.6.1	Force Distribution	57
5.6.2	Soft Substrates	60
5.6.3	Curved Base	63
5.6.4	Soft Pucks	64
5.6.5	Rigid Pucks	67
5.6.6	Drift	69
5.7	Sensor Damage - Delamination	69
5.8	Sensor Calibration	72
5.9	Precision Error	77
6	Palpation Experiment	79
6.1	Experimental Methods	79
6.1.1	Sensor-Glove Attachment	79
6.1.2	Force Sensor Placement	80

6.2	Motion Tracking	81
6.3	Accuracy Validation	83
6.3.1	Experimental Procedure	84
6.4	Palpation Data Preparation	85
6.4.1	Position Interpolation	85
6.4.2	Coordinate Transformation	85
6.4.3	Voltage to Force Conversion	89
6.4.4	Data Stream Synchronization	89
6.4.5	Zeroing the Data	90
6.5	Data Analysis	90
7	Palpation Results	92
7.1	Subject Data	92
7.2	Abdominal Stiffness	93
7.3	Palpation Velocity	110
7.4	3-Dimensional Movement	110
7.5	Maximum Force	113
7.6	Maximum Displacement	115
7.7	Force at Each Sensor	117
8	Discussion	119
8.1	Abdominal Stiffness	119
8.1.1	Hysteresis	121
8.1.2	Non-linearity	122
8.1.3	Repeatability	123
8.2	Influences of Body Mass Index	124
8.3	Palpation Repeatability	125
8.4	Potential Error Sources	126

8.4.1	Position	126
8.4.2	Force	127
8.5	Simulating Stiffness	128
9	Conclusions and Future Work	130
9.1	Conclusions and Recommendations	130
9.1.1	Data Collection and Sensor Testing	130
9.1.2	Patient Simulation	131
9.2	Future Work	132
9.2.1	Building a Patient Simulator	133
9.2.2	Gathering Additional Data	134
	List of References	135
	Appendix A Matlab Code	139
A.1	Repeatability Analysis	139
A.2	Sensor Calibration	141
A.3	Interpolation	144
A.4	Coordinate Conversion	144
A.5	Force Conversion	146
A.6	Velocity Calculation	147
A.7	Combining Data	147
A.8	Data Synchronization	148
A.9	Master M-File	150
A.10	Zeroing the Data	150
A.11	Polaris Accuracy	151
A.12	Data Analysis	152

Appendix B Test Apparatus Design	158
B.1 Locator	159
B.2 Base	160
B.3 Bracket	161
B.4 Surfaces	162
B.5 Cylinder and Point	164
 Appendix C 3D Motion Plots	 166
 Appendix D Plots of Force, Displacement, and Velocity at the Doctor's Fingertips versus Time	 170
 Appendix E Ethics Approval Notice	 179

List of Tables

7.1	Anatomical measurements and BMI of the volunteers	93
7.2	Averages of the maximum force during each push, and accompanying standard deviations.	115
7.3	Averages of the maximum displacement during each push, and the accompanying standard deviations.	118

List of Figures

2.1	A drawing of endotracheal intubation - <i>aic.cuhk.edu.hk</i>	8
2.2	A side view of the Human Patient Simulator from Medical Education Technologies Inc - <i>meti.com</i>	9
3.1	The four quadrants of the abdomen	20
4.1	FSR™ details from the FSR™ integration guide [1]	32
4.2	The FlexiForce™ Sensor by Tekscan, Inc.	33
4.3	FlexiForce™ details from their Users Guide [2]	34
5.1	The sensing circle at the tip of FlexiForce™ sensors	38
5.2	The fourteen loading conditions used during force sensor testing . . .	39
5.3	The three puck materials. From left to right: cork, rubber, plastic . .	40
5.4	A comparison of the repeatability behavior between rubber and cork pucks placed between the sensor and a hard substrate	41
5.5	A photograph of a cork puck attached to the FlexiForce™ sensor . . .	42
5.6	A photograph of a rubber puck attached to the FlexiForce™ sensor . .	42
5.7	A photograph of the plastic pucks attached to the FlexiForce™ sensor	42
5.8	Force sensor testing device designed by the Carleton University patient simulator group	43
5.9	Non-centered force causes a moment about the testing apparatus which leads to secondary loading	44
5.10	Assembly view of the re-designed FlexiForce™ testing apparatus . . .	45

5.11	A photograph of the redesigned test apparatus	47
5.12	FlexiForce™ drive circuit utilized during testing.	48
5.13	FlexiForce™ driving circuit recommended by Tekscan	48
5.14	Typical repeatability test results for the 120N sensor when applying low loads	51
5.15	Typical repeatability test results for the 4.5N sensors when applying low loads	53
5.16	Typical repeatability behavior of the 4.5N sensors at 3, 7 and 14 New- ton loads	54
5.17	Calibration curves from Vecchi, <i>et al.</i> (2000) on three consecutive days	55
5.18	FlexiForce™ resistance changes with increasing load	58
5.19	Test apparatus stability increases from left to right. Increased sta- bility leads to a more uniform force distribution and therefore lower resistances and higher output.	59
5.20	Test apparatus stability increases from left to right. Increased stability leads to more repeatable load applications and therefore minimizes error.	60
5.21	The effects of various substrates on sensor repeatability. The harder the substrate the better the repeatability.	61
5.22	The effects of various substrates on sensor output. A harder substrate leads to higher output.	62
5.23	Sensor Bending decreases resistance and increases output voltage. The puck serves to inhibit sensor bending.	63
5.24	The curved base of the test apparatus	63
5.25	The effects of a curved base with a soft material placed beneath the sensor on output voltage.	64
5.26	The effects of a soft puck on repeatability.	65

5.27	Repeatability improvement due to the presence of a soft puck. Note that these results are the average of three tests	66
5.28	The effects of pucks on output voltage. Note that these results are the average of three tests	67
5.29	The improved repeatability from using rigid pucks. Note that these results are the average of three tests	68
5.30	The effects of various loading conditions on sensors equipped with rigid pucks. Note that these results are the average of three tests	68
5.31	Effects of delamination of sensors on output voltage	71
5.32	Effects of delamination of sensors on repeatability	71
5.33	Sensor calibration curves	73
5.34	An attempt to calibrate the sensors using only one curve.	74
5.35	Downward shifting in sensor calibration curves	75
5.36	Two sensor calibrations performed thirty minutes apart. In between calibrations the sensor was loaded and unloaded multiple times to simulate experimental conditions.	77
5.37	Sensor response for twenty consecutive applications of a 13N load. . .	78
6.1	Photograph of how the tape secures the sensors to the doctor's fingers. Sensor positions are marked by numbers 1-4.	80
6.2	Force levels at the four sensor positions during one push into an abdomen	81
6.3	Polaris™ position sensor and an optical tool	82
6.4	Polaris™ optical tracking marker located on the fingernail of the doctor's middle finger.	82
6.5	The default coordinate system of the Polaris™ motion tracker	86
6.6	Graphical representation of how the GRS is transformed into the LRS	87
7.1	Abdominal stiffness curves for volunteer #1. Each plot represents one push into the abdomen.	95

7.2	Abdominal stiffness curves for volunteer #2. Each plot represents one push into the abdomen.	96
7.3	Abdominal stiffness curves for volunteer #3. Each plot represents one push into the abdomen.	97
7.4	Abdominal stiffness curves for volunteer #4. Each plot represents one push into the abdomen.	99
7.5	Abdominal stiffness curves for volunteer #5. Each plot represents one push into the abdomen.	101
7.6	Abdominal stiffness curves for volunteer #6. Each plot represents one push into the abdomen.	103
7.7	Abdominal stiffness curves for volunteer #7. Each plot represents one push into the abdomen.	105
7.8	Abdominal stiffness curves for volunteer #8. Each plot represents one push into the abdomen.	107
7.9	Force, displacement (normal to abdominal surface) and velocity aligned with time.	109
7.10	Phase Planes for volunteers 1 through 4.	111
7.11	Phase Planes for volunteers 5 through 8.	112
7.12	3D motion of the doctor's fingertip during the palpation of volunteer #4.	113
7.13	Maximum force applied to the abdomen for each volunteer, ordered by increasing BMI.	114
7.14	Average of the peak force for each push, ordered by increasing BMI. .	114
7.15	Maximum displacement into the abdomen for each volunteer, ordered by increasing BMI.	116
7.16	Average of the peak displacement for each push, ordered by increasing BMI.	117

7.17	Forces occurring at each sensor during a typical palpation.	118
8.1	Initial (solid line), middle (dashed) and final (dashed/dotted) plots of stiffness for volunteer #1.	120
8.2	Stiffness curves for the entire palpation of volunteer #4	123
8.3	Stiffness curve comparison between volunteer #2 (high BMI - solid line) and volunteer #6 (low BMI - dashed line).	125
8.4	Averaged abdominal stiffness curve for volunteer #1.	129
8.5	Averaged abdominal stiffness curve for volunteer #3.	129
B.1	A simple bracket designed to repeatedly locate the loading apparatus with respect to the sensor for testing purposes.	159
B.2	Base of the re-designed Flexiforce testing apparatus. The groove along the top of the base holds the sensor in position.	160
B.3	Vertical component of the bracket to which the point-loading cone is attached, and from which the load is hung.	161
B.4	Horizontal component of the bracket to which the point-loading cone is attached, and from which the load is hung.	162
B.5	Flat disc which rests within the base of the test apparatus.	162
B.6	Curved disc which rests within the base of the test apparatus - designed to simulate the curvature of a finger-tip.	163
B.7	The cylinder which transfers the load to the sensor. The indent at the top of the cylinder accommodates the pointed part of the bracket. . .	164
B.8	Point loading mechanism of the test apparatus. The point bolts to the bracket from which the load is suspended.	165
C.1	3D motion of the doctor's fingertip during the palpation of volunteer #3.	166
C.2	3D motion of the doctor's fingertip during the palpation of volunteer #4.	167

C.3	3D motion of the doctor's fingertip during the palpation of volunteer #3.	167
C.4	3D motion of the doctor's fingertip during the palpation of volunteer #5.	168
C.5	3D motion of the doctor's fingertip during the palpation of volunteer #6.	168
C.6	3D motion of the doctor's fingertip during the palpation of volunteer #7.	169
C.7	3D motion of the doctor's fingertip during the palpation of volunteer #8.	169
D.1	Force, displacement and velocity aligned with time for volunteer #1. .	171
D.2	Force, displacement and velocity aligned with time for volunteer #2. .	172
D.3	Force, displacement and velocity aligned with time for volunteer #3. .	173
D.4	Force, displacement and velocity aligned with time for volunteer #4. .	174
D.5	Force, displacement and velocity aligned with time for volunteer #5. .	175
D.6	Force, displacement and velocity aligned with time for volunteer #6. .	176
D.7	Force, displacement and velocity aligned with time for volunteer #7. .	177
D.8	Force, displacement and velocity aligned with time for volunteer #8. .	178
E.1	. . . . .	180

Chapter 1

Introduction

1.1 The Problem

There is a problem with practical, hands-on training in medical education. Unfortunately, all of us, as potential patients, risk experiencing this problem in a most unpleasant way. I am referring to the lack of experience medical students receive before attempting certain procedures on a real person.

This phenomenon occurs daily in hospitals and doctor's offices around the world when patients unknowingly undergo procedures which young doctors have never before performed. The procedure may be something simple, which poses little or no risk to the patient, for example, a patient history. But what about the first time they try to set a bone, or palpate someone with appendicitis?

For most people, the thought never crosses their mind that Dr. Smith, the handsome and confident-looking young man about to set their daughter's broken arm only graduated from medical school one week ago. He has read plenty of books about what it should feel like to set a broken arm, but has never actually touched one. After all, he is a doctor, and that designation inspires confidence and trust. Fortunately, a senior attending physician will be present to direct Dr. Smith, and take over for him, should things not go according to plan.

The medical schools and the medical industry in general cannot really be blamed for this. There really is no alternative. Since the beginning, they have operated under the “see one, do one, teach one” principle. Every doctor has to do every procedure for the first time. There is no way around it. The question is: how can we better prepare them for this first attempt?

1.2 The Solution

This thesis addresses one potential solution for the aforementioned problem: the use of patient simulators. Patient simulators run the technological gamut from very simple to cutting edge. For example, they may be as basic as a cardiopulmonary resuscitation (CPR) dummy, or as advanced as a modern day computer, coupled with a sophisticated mannequin, running complex modelling software to simulate real medical conditions.

One can begin to see the potential benefits that realistic patient simulators could have on medical education, and, in turn, on patients. For example, medical trainees could practice any procedure on the simulator until they felt confident enough to try it on a real patient. Also, rarely seen cases could be simulated, enabling doctors to keep their skills sharp. Of course, patient simulators bring their own set of problems to the table, the greatest of which is physical realism.

An exhaustive literature review was performed in order to determine the state of the art of patient simulation, and in what areas current simulators were most lacking. A number of areas clearly required attention: physical realism, specifically skin colour, temperature and texture, and abdominal sign simulation. Abdominal signs include diagnostic indicators such as a change in stiffness, which may be detected by palpation. Currently no simulator for abdominal palpation exists, but there is demand for such a device. An abdominal sign simulator mimics the various medical

conditions detectable by palpation of the abdomen. Given the appendicitis example stated earlier, the need for such a device is clearly evident.

1.3 The First Step

This thesis describes the first steps towards the building of a prototype abdominal sign simulator. Force and position measurements were taken at a physician's fingertips during the palpation of eight healthy volunteers. These data allowed for the characterization of the abdominal response to palpation in engineering terms. They also provided mechanical limits, such as maximum force and displacement during palpation. These data are necessary for the construction of an abdominal simulator with realistic response.

The measurements made during these experiments are of the activity of abdominal palpation. The results depend not only on the material characteristics of the abdomen, but also on the movements and choices of the palpator. For example, since the thickness of the abdominal fatty layer differs between individuals, it is possible that the doctor modifies his palpation technique to accommodate this.

The use of multiple volunteers allows for an analysis of the repeatability of the doctor's palpation technique. A better understanding of this technique can only improve simulator design. With the basic abdominal behavior understood, various material and mechanical solutions can be explored in an attempt to recreate this behavior.

Once a suitable prototype capable of realistically simulating a healthy abdomen is developed, the project can be advanced to incorporate the characteristics of certain medical conditions detectable through palpation; appendicitis for example.

Going back to the original problem of young doctors having no alternative but to learn new procedures on real patients, an abdominal sign simulator could provide a

viable alternative. Instead of having to learn what an inflamed appendix feels like by palpating some unfortunate patient whose appendicitis has already been diagnosed, students can take their time and learn on the simulator at their own pace, while causing no discomfort or risk to a real patient.

Chapter 2

Patient Simulators

2.1 Background Information

Simulation is defined as “an imitation of some real device or state of affairs”. Simulation attempts to represent certain features of the behavior of a physical or abstract system by the behavior of another system.

Patient simulators come in two major forms: virtual and physical. A virtual patient simulator may be as simple as an animated patient displayed on a home computer screen, or as advanced as a complete three-dimensional virtual operating room. A physical patient simulator may be a rubber arm on which to practise injections, or a full sized mannequin, capable of breathing in air, exhaling carbon dioxide, bleeding, speaking, and suffering from hundreds of medical conditions.

Patient simulators have the power to significantly change the direction of medical education for the better. Already simulations are becoming standard practice at many medical schools. In Ottawa, medical students run through scenarios in the simulation laboratory at the Ottawa Heart Institute and are graded on their performance. It is not uncommon for students to become very immersed in the simulation, and react with powerful emotion should the simulator “die”.

2.1.1 The Benefits of Simulation

Some of the major benefits that patient simulation brings to medical education are as follows:

- Patients are not exposed to any unnecessary danger or discomfort. Abdominal palpation is a prime example of this. Teaching a young doctor what it feels like to palpate an inflamed appendix is not a particularly appealing educational practice for the patient. A much better solution would be to teach the young doctor on a patient simulator capable of simulating appendicitis.
- Simulations can be repeated until the trainee feels comfortable and confident enough to try the procedure on a real patient. Many procedures have a small margin of error and repetitive practice would be very beneficial for experienced doctors as well as new ones. Patient simulators provide the option of repeatedly practicing the procedure without putting a real person at risk.
- Rare cases, or those that are difficult to detect or dangerous to treat, can be simulated. There are some medical conditions which are life-threatening, but may only present themselves to a doctor once or twice a year. It is challenging for a doctor to ready herself to treat such a condition when her experiences with it are few and far between. Patient simulators could be used to recreate the condition whenever, and wherever the educator saw fit, ensuring that when a patient eventually does present with the condition, the physicians are aptly prepared to treat it.
- Training on a simulator is less stressful for the trainee as there is no need to worry about injuring the patient. This provides for an environment more conducive to learning. Doctors must learn to operate under high stress conditions. However, this does not mean that their educational environment should mimic

these high stress situations when they are attempting to understand the intricacies of a particular therapy. Working with a patient simulator allows a person to build confidence and understanding, so that when they are exposed to the real thing, they have the mental and physical know-how to perform the procedure safely.

2.1.2 Simulator effectiveness

Many studies have been released about the effectiveness of simulators in the training of medical students, doctors, nurses and paramedics, references [3] [4] [5] [6] [7] [8] [9] [10] [11] and [12] provide some of the many examples. These studies not only demonstrate that the participants enjoyed the simulation experience and found it to be a valuable one, but also showed statistical evidence that medical personnel whose training involved simulation exercises performed better in real life situations than those whose training did not include simulation.

2.2 The State of the Art

Before it was decided to work on abdominal sign simulators, a literature review was performed in an attempt to understand the state of the art in the patient simulator industry and to determine how to best contribute to the field. This section of the report will highlight some of the major issues in patient simulation that were discovered, and propose some potential solutions.

2.2.1 Simulation Example

As an introduction to patient simulation, consider endotracheal intubation. Patient simulators are already commonly used to train medical personnel in this procedure

(figure 2.1). Patients are intubated when they are unable to breathe on their own (during surgery for example), or if there is a risk of their airway being obstructed. The procedure involves passing a tube through the mouth and down into the trachea, providing clear passage for air to reach the lungs. Intubation is particularly applicable to simulation because it is used frequently, and is necessary for saving lives. It is also difficult to perform and poses a serious risk to the patient.

Figure 2.1: A drawing of endotracheal intubation - *aic.cuhk.edu.hk*

There are a large number of complications that can arise during an intubation, so practicing is important. Modern physical simulators employ a variety of scenarios to make the intubation as easy or as difficult as the simulation operator wishes. For example, figure 2.2 shows the Human Patient Simulator (HPS) developed by Medical Education Technologies, Inc (METI). The HPS is an excellent example of a modern, high-fidelity, mannequin-based patient simulator. The HPS is able to make intubation more difficult by activating the following features: tongue swelling, pharyngeal obstruction, pharyngospasm and bronchospasm.

Figure 2.2: A side view of the Human Patient Simulator from Medical Education Technologies Inc - *meti.com*

2.2.2 Simulator Response

State of the art simulator response is very impressive in high-fidelity models. These simulators cost hundreds of thousands of dollars, require no input from a human operator, and employ complex physiological models and mechanical systems in their simulation arsenal.

There are two mechanisms by which a patient simulator responds to the actions of the user; both are utilized by the HPS. The first method, called script-driven simulation, involves following a script which spells out the appropriate reactions to specific user actions. Script-driven simulations must anticipate the actions of the user, and the proper response must already be programmed into the physiological system. For example, if a painful condition is being simulated and the user injected morphine to counter the pain, the software running the simulation will simply follow the script for a morphine injection and produce the proper, pre-programmed response. The disadvantage of this method is that it is virtually impossible to anticipate every potential action taken by the user and every potential response from the affected physiological systems.

The second approach to automatic physiological system response is model-driven simulation. This approach utilizes mathematical models of systems to continuously change the physiological attributes of the simulator in response to user actions. Mathematical models can calculate the variables that define circulation, ventilation, gas

uptake and distribution and pharmacodynamics, all in real time [10]. The advantage of this approach is that simulations are no longer limited to the responses specified by the script-driven approach.

This advantage can be better understood by going back to the earlier morphine injection example. Whereas the script-driven approach has a finite number of responses to morphine injections of various volumes, a model-driven simulation has an infinite number of responses, based on attributes of the mathematical model. In a model-driven approach, the volume of injected morphine is a variable in an equation which influences physiological parameters such as respiratory rate and blood pressure. The use of a model-driven approach also allows for one physiological system to influence another, creating a network of interacting systems and sub-systems. For example, the systemic uptake and distribution model influences the cardiovascular model through adjustments made to oxygen saturation levels in arterial blood [13].

The HPS is a script-controlled, model-driven simulator. That is, it utilizes the simplicity of script-based simulations, combined with the power of mathematical modelling to represent physiological systems and responses. In some HPS systems the responses computed by the mathematical model are manifested in a mechanical model. For example, the cardiovascular and pharmacokinetic models are purely mathematical, while the pulmonary gas exchange model is a hybrid mathematical/mechanical system. This hybrid system uses a computer to calculate the uptake and delivery of respiratory and anaesthetic gasses which is then outputted to a mechanical bellows which simulates breathing. The mathematical cardiovascular model in the HPS contains two sub-models: one determines various stages of ischemia using oxygen supply/demand ratios, the other takes this ischemia information and generates appropriate dysrhythmias, contractile activity and ECG outputs.

These physiological models are interfaced with the trainee, and with various medical equipment through either artificial or physico-chemical interfaces. Examples of a

physico-chemical interface are the HPS's five electrodes beneath the skin on its chest which generate current for ECG's, or the inhalation of oxygen and expulsion of CO₂ from its lungs. The cardiovascular model generates current for ECG's in synchrony with blood pressure wave forms, heart sounds, and palpable pulses.

An example of an artificial interface used by the HPS is the pulse oximeter. In this case the oximeter does not actually monitor the absorption of light in blood, instead it is stimulated using an artificially generated electronic signal. Physico-chemical interfaces are clearly superior in terms of simulation realism, but are often more costly, and the perceived educational benefit does not always justify the necessary engineering resources [13].

2.2.3 Look, Touch and Feel

Compared to the complexity of the technology discussed in the previous section on simulator response, it comes as somewhat of a surprise that the physical appearance of patient simulators is severely lacking. While the internal technical attributes have become very advanced, appearance has remained rudimentary at best, and the simulation experience is suffering as a result.

First impressions are very powerful. If a doctor is planning to train with a patient simulator, and she walks into the room, looks at the patient simulator, and immediately notices that its skin is an unnatural colour and has no characteristic markings, a significant portion of the simulation experience has already been compromised. Instead of thinking "What's wrong with this patient?", the doctor will think "Look at the colour of that skin, it looks completely fake". In the doctor's mind, this first impression has already firmly established that this is not a real person; consequently, she can relax and not worry.

Earlier it was stated that one of the main advantages of using simulators is that they provide for a stress free learning environment. While this is certainly true,

simulators can also be used to run emergency scenarios, specifically designed to test people under pressure. Therefore it is desirable to have simulators which are ultra-realistic.

In reviewing many papers on patient simulator effectiveness as a teaching and training device, a number of comments from users repeatedly appeared: “The skin colour is not realistic and does not change as the patient’s condition is changing”, “The skin temperature does not change as the patient’s condition is changing”, and that the physical appearance in general is clearly fake [7] [14]. Students in the Patient Simulator design project at Carleton University re-iterated the comments found in these journal articles after viewing the simulators located at the Ottawa Heart Institute. There is absolutely no question that from an end-user standpoint (nurses, doctors, medical students) that physical realism is one of the areas most lacking in state of the art patient simulators.

Currently, patient simulator skin is simply a layer of hard plastic or rubber of a predetermined, constant colour. The skin is malleable, and it is softer or harder at the appropriate places. However, there are no colour or temperature changing abilities in place, and as a whole the appearance is clearly artificial.

There are three major areas of exterior realism that require attention: skin colour changing, skin temperature changing, and skin texture. Each of these areas is important and plays a key role in simulation realism from both an aesthetic and physiological standpoint. Skin colour and temperature are important indications of patient condition and are used for diagnostic purposes. Skin texture is important for overall realism.

There are hundreds of changes that occur in skin as a result of various medical conditions. Obviously, it is not possible to address all of them at once. In an attempt to determine the single most important symptom displayed by the skin, a doctor, a nurse, a medical student and an osteopath were contacted. The following sections on

colour, temperature and texture review the information gathered from these informal discussions.

2.2.4 Skin Colour

The three colours most commonly associated with human skin and medicine are red, white (blanching) and blue (cyanosis). Some common causes of these colours are as follows:

- Red
  - Inflammation
  - Fever
  - Infection
  - Allergic reaction
- White (blanching)
  - Poor circulation
  - Shock
  - Poor perfusion
- Blue (cyanosis)
  - Low temperature
  - Vasoconstriction
  - Asphyxiation

Red appeared to be the most important colour change from a diagnostic standpoint, along with colour (and temperature) differentials. For instance, to help determine if something is wrong with the upper-left quadrant of the abdomen they will compare its colour and temperature to that of the other three quadrants.

The medical professionals also stated that testing capillary refill is important. To test capillary refill the doctor will push on the skin with a finger, forcing blood out of the local capillaries which blanches the skin. The health of the capillaries and of

the overall blood supply to that area can be determined by the time it takes for the colour to return (the capillaries to refill). Anything longer than two seconds means that there might be a problem. Also, some medical conditions can make it very difficult, or even impossible, to cause the skin to blanch.

2.2.5 Skin Temperature

The goals of temperature controlled skin are twofold: first, to maintain a normal, healthy body temperature and secondly, to increase/decrease the temperature to simulate a medical condition.

The temperature of the skin, like its colour, is determined by blood flow and therefore vasoconstriction and vasodilation. Skin temperature varies with age and location, for example, the chest is warmer than the feet.

Some medical conditions causing temperature change in the skin:

- Temperature Increase
  - Inflammation
  - Fever
  - Allergic reactions
  - Infection
- Temperature Decrease
  - Blood loss
  - Poor circulation
  - Shock

The doctor had a difficult time choosing whether simulating colour or temperature change is more important, as the two are intrinsically related. For example, simulating inflammation with an increase in temperature, but no change in colour may not be beneficial at all, as it is not indicative of a real life situation. In fact, if the temperature of the skin changed, but the colour did not, the user would probably not even think to feel that part of the skin.

2.2.6 Skin Texture

The look and the feel of the skin are also important attributes worthy of consideration. The look gives the first visual impression, and the feel, the first tactile impression. Realistic-looking and realistic-feeling skin would go a long way in furthering the credibility of a simulation.

Real skin is quite complex as it displays viscoelastic, inhomogeneous and anisotropic behavior. However, it is only necessary to mimic these properties to the degree that they are perceptible by the user. As long as the skin feels and responds realistically enough that the user does not take notice, no further improvement is required. The exception to this may be in the area of the abdomen if the simulator was being used for abdominal palpation. In that case, the skin's response would need to be very accurate as precise differences in feel could indicate different medical conditions. Also, doctors specializing in palpation would be far more sensitive to tissue response than most users.

The thickness of the skin in relation to its location on the body is also important. For instance, the skin on the palm of the hand is thicker than the skin on the back of the hand. It is also coarser and tougher. This translates into different types of artificial skin being used for different areas of the body. This could potentially lead to problems with seams joining the skin sections together.

Real skin has three definite layers: epidermis, dermis and sub-cutaneous. Many artificial skins mimic this three layered approach (Livingskin, for example - www.livingskin.com). The outer layer (epidermis) contains pores, hairs, and other details. Livingskin's outer layer also contains pores and fine details. The dermal layer contains translucent, specialized skin cells, which allow the colour from the sub-cutaneous layer to show through. The sub-cutaneous layer contains fat cells and the skin's blood supply; it is the source of the majority of the skin's colour.

One area in desperate need of improvement is the joints (specifically knee and shoulder). Right now there are large, clearly fake, seams in the skin at these areas. The seams are there to facilitate joint mobility, and perhaps for mechanical ease when replacing damaged components or damaged skin. Making skin that allows for joint movement without the need for any seams is highly desirable.

Most of the high end prosthetic devices utilize skin with hair (Livingskin's Derma-Hair for example). Body hair is something that adds significantly to a realistic appearance, especially on the head. Right now, even the most advanced simulator, METI's Human Patient Simulator, simply uses molded plastic for hair.

Bleeding and sweating are also features to consider, and both were brought up as important by the doctor. Before implementing either of these features a study would need to be performed in order to ensure that the benefits justified the engineering efforts, as both features would be difficult to do well.

Although skin will not be discussed beyond this point in the report, this information should prove useful at a later point in the abdominal sign simulator project.

2.2.7 Virtual Patient Simulators

Virtual reality (VR) based simulators have made substantial advances over the last few years as technology catches up with the raw processing power requirements of complex physiological models. It used to be the case that computer processing power requirements were holding back VR simulation because computer resources were simply too expensive [15]. New advances in computer technology and the use of parallel processing has opened the door to less expensive, yet more powerful VR simulators. Haptic feedback is now the greatest challenge that simulator manufacturers and researchers are facing.

Haptics, the study of touch, incorporates tactile stimuli into virtual reality (VR) systems, allowing the user to experience the 'feel' of a procedure [16]. Creating a

realistic sense of touch in VR simulation is proving to be extremely challenging for researchers, but it is absolutely necessary for advancing simulator effectiveness.

If a person was giving someone an injection in a virtual simulation and there was zero resistance as the needle came into contact with the skin, the person would not have experienced what it is really like to give someone an injection. To address this, state of the art simulators employ haptic feedback devices to simulate various aspects of touch. An injection device would attempt to recreate the “pop” one feels as the resistance of the skin is overcome by the tip of the needle [16].

There are already a few areas of VR simulation where haptic response is at an acceptably realistic level, needle simulation being one of these areas. Virtual endoscopes have also been advanced to a level where they are suitably realistic [16]. Procedures such as endoscopy and minimally invasive surgery are particularly conducive with VR simulation as the doctor is usually looking at a computer screen to see what he or she is doing during the real procedure.

2.2.8 Abdominal Sign Simulators

Currently there are no abdominal sign simulators on the market. This is a gaping hole in the patient simulator industry, as abdominal palpation is a common practice in the diagnosis of many medical conditions.

In order to design and build an abdominal sign simulator some key information is required. First, the designer must possess a basic understanding of what palpation is, and why it is used. This information is covered in the next chapter. Once the medical side of the procedure is understood, the technical aspects must be considered. When a doctor palpates an abdomen, she is looking for specific indicators of good, or bad health. One of the most common indicators is the abdomen’s stiffness. As chapter 3 will explain, a change in stiffness is a strong diagnostic indicator of a problem in that area.

Before a simulator can be built with a realistic response to palpation, the stiffness characteristics of a real abdomen must be understood. To do this, one must measure the force applied to the abdomen during palpation, and the resulting abdominal displacement. With this information, stiffness can be plotted, and mechanisms for artificially replicating the stiffness can be researched.

2.2.9 Why is an Abdominal Sign Simulator Needed?

In his report on the physical examination of the abdomen [17], Dr. James Christensen states that knowing the difference between normal and abnormal tissues of certain organs is very difficult, and that one needs to feel many normal and abnormal tissues to feel secure in their distinction. The body is designed to protect its important organs, making it difficult to simply palpate and assess them.

Currently there is no effective way for doctors or medical students to practice palpation, except on real patients or volunteers. This is of course, not an ideal situation. Most people do not enjoy being palpated, especially if it is by a nervous medical student and they are suffering from a medical condition which makes the palpation painful. Doctors also have to take what they can get in terms of training on specific medical conditions. If no patient presents with appendicitis, then there may be no practical appendicitis training.

Chapter 3

Abdominal Palpation and How to Measure it

3.1 Background

Palpation is a widely used diagnostic and assessment tool employed by doctors in almost every specialty. Abdominal palpation is the process of using touch to detect the presence of any abnormalities or characteristics indicative of disease in a patient's abdominal area. Using this technique, doctors can feel the size and shape of organs, gauge the stiffness and tenderness of tissues and discover organs normally not palpable. Palpation can be used to detect many medical conditions ranging from appendicitis and hernias to simple gas.

The palpating physician views the abdomen as a rectangle composed of four quadrants with a horizontal and vertical line transecting the umbilicus. The sternum forms the superior border and the pubis forms the inferior border (figure 3.1).

Figure 3.1: The four quadrants of the abdomen

3.2 The Abdominal Examination

A typical abdominal examination involves four stages of which abdominal palpation is the most critical: inspection, auscultation, palpation and percussion [17].

Inspection is a purely visual activity. During this phase the doctor observes the abdomen, looking for characteristic signs of disease such as: skin colour, symmetry, distension, and visible peristalsis.

Auscultation and percussion are audio inspections of the abdomen. Auscultation is simply listening to the abdomen using a stethoscope. Percussion involves tapping the abdomen with one or two fingers and listening for sounds characteristic of certain medical conditions. For example, most of the abdomen resonates because our intestines are filled with gas. If a hyper-resonant area, higher in pitch, is discovered, it could be an indication of gaseous bowel distension [17]. Percussion can be used to determine the size and density of organs in an abdomen that is too tender to be palpated. While not directly applicable to this report, the addition of abdominal sounds would be a great asset to any abdominal simulation.

3.2.1 For What are Doctors Feeling?

There are two types of abdominal palpation: survey and deep. Survey palpation is relatively light and generally the fingers are only pushed about 1cm into the abdomen. As the name suggests, this technique is used to get a general feel for the abdomen and to make mental notes of areas to inspect under greater scrutiny. Using survey palpation doctors can detect areas of tenderness through communication with the patient and through involuntary abdominal muscle contractions. These contractions, called guarding, are a response to pain, as the body attempts to protect itself from harm. A healthy abdomen exhibits a uniform spongy resistance to pressure in all areas except over the liver [17].

When performing deep palpation both hands are usually used. The dominant hand is used to feel and the subordinate hand, which is on top of the dominant one, to exert pressure [17]. This is the standard position, unless the subordinate hand is moving an organ into a position where it can be more easily palpated. This process is called ballottement. With deep palpation, doctors feel for the location, depth, size, consistency, tenderness and stiffness of a tissue or organ. Stiffness is important because often tumors and organs enlarged by inflammation will not be any more tender than normal tissue [17].

3.3 Obtaining Abdominal Stiffness

Before building a simulator which accurately reproduces the stiffness of an abdomen, that stiffness must first be measured. The two components of stiffness are force and displacement, which must be obtained in the static case. However, in this experiment the measurements were made dynamically in order to preserve palpation realism. This means that the resulting relationship may be dependent on velocity, side to side motion and shear forces on the sensor. Despite the fact that stiffness was not

measured in the strictest sense, the force versus displacement relationship will still be referred to as such in this report. Once the force/displacement data have been acquired, stiffness can be plotted and analyzed.

Currently, there exists no data on the forces applied to the abdomen during palpation, with the exception of one unpublished paper from a 1987 conference [18]. This paper looked at the pain threshold in children with appendicitis. The experiment essentially involved measuring the force at which the patient started to experience a great deal of discomfort. It is impossible to infer any information on abdominal stiffness from the paper.

While the data on palpation force are minimal, Cutkostky *et al.* (1999) stated that the maximum force that a human finger can exert is 50N for a short period of time, and 15N for a prolonged period of time [19]. This at least provided an upper-bound for force sensor requirements.

An apparatus has been designed to simultaneously collect both force and displacement data directly from a doctor's finger-tips during real palpation. At the moment, these data are limited to healthy male patients, but the apparatus could easily be used on sick patients, women, and children in the future.

The apparatus works by monitoring four separate thin-film force sensors located on the doctor's middle, index and ring fingers. It is at these locations that force is applied to the abdomen during palpation. On the back of the hand, attached to the fingernail is a retro-reflective optical marker, whose location is tracked via an infra-red camera.

The sensors and optical marker are attached to a standard medical grade latex glove. Data were collected from 8 volunteers during palpations performed by Dr. Juan Bass, a paediatric abdominal surgeon at the Children's Hospital of Eastern Ontario. The test procedure and apparatus are discussed in greater detail in Chapters 6 and 7.

3.4 What is Not Being Measured

To date the data collection and analysis has been limited to one area (the lower right quadrant) of healthy male abdomens. This information provides a baseline for later experiments on patients suffering from abdominal conditions such as appendicitis.

The next chapter covers the availability of suitable force sensors and their respective advantages and disadvantages.

Chapter 4

Fingertip Force Sensors

4.1 Background

Tactile fingertip sensors are an active field of research. This is especially true for sensors designed for robotic limbs [20] [21]. Therefore, many of the sensors described here were initially developed for use in robotic tactile feedback. However, there are also sensors being developed for medical research. For example, Vecchi *et al.* (2001, 2003) is developing sensors to be used on prosthetic hands. Sensors are also being researched for neuromuscular control, prosthesis and orthosis for handicapped people, hand biomechanics research, clinical evaluations of hand function and for hand rehabilitation devices [22].

Some of the earlier methods of measuring finger and grip forces include: subjective magnitude, which is low resolution and depends on the objectivity of the participant [23]; integrated electromyography, which is limited to static forces [24]; and strain gauges, or other transducers on objects [25].

4.1.1 Force Sensor Desired Characteristics

Crago *et al.* (1986) recommends that a force sensor possess the following characteristics to adequately evaluate hand function:

- 0.1 to 80N force range
- 0.1N resolution for light touch
- 1N resolution for normal touch force
- Low mass, small size
- Avoid limiting range of motion
- Avoid limiting degrees of freedom
- Minimal wiring to avoid breakage

Many different methods of force sensing have been researched, and no technology has yet to emerge as a clear winner for tactile force sensing. Each method shows promise, but most have one or more disadvantages which make them impractical for use outside of a research setting [22]. Common disadvantages include: size, lack of over-force protection, difficult fabrication, limited force sensing range, fragility and lack of accuracy. For sensors designed to measure human touch force, perhaps the greatest problem is the fact that they must be located between the fingertip and the object, which will always result in a loss in tactile feedback to the wearer. This can cause numerous problems, such as the application of excessive force [26].

Current research is tending towards sensors that integrate numerous sensing modalities into one unit; for example normal and shear force, or hardness [27]. The reason for this is that without being able to gauge hardness along with force, robots will have a difficult time handling fragile materials [20]. At the moment these types of sensors are of relatively little interest as this project is only concerned with normal forces.

4.1.2 General Characteristics of Force Sensors

There are a number of characteristics that apply to most force sensing technologies, and that can affect the performance of the sensor.

- Repeatability - The ability of the sensor to consistently output the same response for an identical applied force. For some sensors, it is possible to improve sensor repeatability by conditioning the sensor before calibration and use [2].
- Linearity - A measure of the sensor's response over the entire force sensing range. A completely linear response is ideal.
- Hysteresis - During dynamic loading, it is the difference in response between loading and unloading for a given force [2].
- Drift - The gradual change in sensor output, caused by a constant applied force.
- Temperature Sensitivity - Sensor response can vary at high temperatures and high loads for some sensors [2]. Proper calibration at the expected temperature can minimize this problem.
- Durability - sensor robustness. This usually depends on how they're handled. Many sensors considered here are only meant for normal forces. Exposure to shear forces can rapidly diminish sensor life.

It is important to make sure that the size of the sensor matches the size of the fingertip. If the sensor is too small, it can sink into the tissue and some of the applied force will bypass it, resulting in measurement errors. It is also important to test the sensors on tissue (or whatever it is they will be used on) as well as a test bench. Test benches usually provide ideal conditions, such as a rigid base. When mounted on a finger the sensor characteristics can change due to the softness and curvature of the tissue/bone substrate [28]. These complexities can be dealt with by the use of a puck, which will be discussed shortly.

4.2 Common Technologies used in Tactile Force Sensors

4.2.1 Piezoresistive

These sensors operate via the piezoresistive effect, which refers to the change in electrical resistance of a material due to an applied mechanical stress. To utilize this effect in sensor technology, piezoresistors are usually integrated into a membrane which is deflected under an applied load [29]. The sensors are incorporated into an electrical circuit, and as the membrane deflects, the resistance in the piezoresistor decreases, causing a measurable change in output voltage. The disadvantage of this sensor technology is its high sensitivity to temperature changes.

Sorab *et al.* (1988) used piezoresistive sensors to measure the forces delivered to newborns by doctors during birth. The sensors were attached directly to the physician's hands and were able to measure forces in the range of 0 - 45 N with a resolution of 0.5N, low hysteresis and high repeatability.

4.2.2 Piezoelectric

It is possible to take advantage of the piezoelectric effect by measuring the electrical charge generated when polyvinylidene fluoride (PVDF) film is deformed [30]. Piezoelectric films are anisotropic, this means that the response to an applied force is dependent on the direction that it is applied. For an application such as abdominal stiffness measurement, where the film would be affixed to a soft substrate (a fingertip in this case), the measured voltage would be affected by the degree of sensor bending. This problem could be eliminated through the use of some sort of a rigid puck. Nevertheless, this is an undesirable solution as it essentially negates this type of sensor's most attractive attribute of extreme thinness, and would be intrusive to the wearer.

4.2.3 Capacitive

Two metal plates are mounted opposite one another on membranes. As the membranes deform, the distance between the two plates changes, resulting in a change in capacitance. Capacitive sensors provide high linearity and sensitivity [29]. They also have superior hysteresis and stability characteristics to piezoresistive sensors. However, they have significantly higher production costs and signal processing is more complex [29].

A German company called Novel makes finger-tip and glove-shaped sensors based on capacitive sensing technology. The glove shaped sensor consists of 135 individual sensors, comprising a total sensor area of 12.896 mm^2 . The sensor was initially designed for measurement of pressure distribution at a hand-machine interface. The individual sensing elements are elastic, and are arranged in a matrix, such that the sensor is able to conform to the shape of the hand, allowing its use in various grips and positions. Unfortunately, Novel sensors are not sold separately from their accompanying electronic hardware and software. This brings the price of a basic system to \$25,000.

4.2.4 Optical

It is possible to use light sent through fiber optic cables to specialized sensors to measure force. A Mach-Zehnder interferometer has been used for such a purpose [29]. When the sensor is deformed, the light experiences a phase shift which is detected by photodiodes. Such a sensor has been designed at a size of $0.3\text{mm} \times 5\text{mm}$. [29].

Optical sensors have been incorporated into sensing arrays and show some promising advantages over other types of sensors in terms of immunity to electromagnetic fields, simplicity of design, corrosion resistance and flexibility [21]. However, this sensor is very new and still in the early phases of testing and development.

4.2.5 Skin-like Sensor Arrays

There is a general lack of commercially available sensors which provide feedback for torques, forces, contact shape and slip for robots [31]. “Smart skin” sensors try to combine many, or even all these sensors into one unit, essentially imitating the sensing abilities of human skin. They are often comprised of multiple layers, each with its own sensing modality. These sensors are aimed at detecting mechanical stimuli as they would be perceived by a human fingertip.

4.2.6 Fingernail

As mentioned earlier, the problem with tactile sensing for medical research is the degradation of haptic feedback due to the sensors being located on the palmar surface of the fingertip. In order to avoid this problem, Asada *et al.* (1999, 2001, 2004) have designed a sensor that is capable of measuring not only normal force, but shear force and finger posture. This unique sensor is mounted on the back of the fingernail, and works by measuring the change in colour of the tissue beneath the nail.

When a finger is used to apply a force, the volume of blood beneath the fingernail changes [32]. Asada *et al.* designed a sensor to measure this change by monitoring finger nail colour with a technique termed reflective photoplethysmography. Colour detection is accomplished by an array of micro-LED’s and photodetectors. So far, the system has proved capable of measuring shear forces and finger posture (-10° to 90°) with a root mean square error of 10%. Unfortunately the measurements of normal forces were not nearly as accurate. Over a range of 0 to 3N the sensors had 1N error.

While this method of sensing holds great promise for future applications, unfortunately the technology is not yet capable of making the measurements required by this project.

4.3 FlexiForce^{TM} and FSR^{TM}

The two main commercially available force sensors which have been used for past experiments similar to this one are the Force Sensing Resistor (FSR^{TM}) made by Interlink Electronics, and the FlexiForce^{TM} made by Tekscan Inc. In an experiment comparing the two sensors for use in a glove designed for measuring grasping forces (3-28N), Vecchi *et al.* (2000) found that the Tekscan FlexiForce^{TM} sensors were superior to the FSR^{TM} sensors in terms of repeatability, linearity, time drift and dynamic accuracy. However, the FSR^{TM} sensors proved to be far more robust. Both sensors are only capable of measuring normal force, and excessive shear force may damage them.

4.3.1 FSR^{TM}

The FSR^{TM} consists of two conducting interdigitated patterns deposited on a thermoplastic sheet facing another sheet containing a conductive polyetherimide film. The sensors have a virtually infinite impedance until a force is applied. Force causes the two sheets to come together, with the size of the contact area being relative to the applied force. This decreases the electrical resistance of the sensor [33].

It is also possible to use conductance ($\frac{1}{R}$) as a method to measure force with FSR^{TM} sensors. Measuring conductance allows for interpretation on a linear scale [1]. A simple current-to-voltage converter gives a voltage output directly proportional to conductance.

FSR^{TM} sensors exhibit a turn-on threshold, or “break force” that marks the beginning of the dynamic sensing range. The force required to exceed the break point varies with the type of substrate used, the flexibility, the size and shape of the sensor and the size of the gap between the two conductive elements [1]. Typically, for forces above 100N, the sensors saturate and additional applied force will yield no change in resistance.

The FSR™ sensors boast a resolution of $\pm 0.5\%$ of full scale force. However, they are not particularly accurate. The sensors are capable of $\pm 5\%$ to $\pm 25\%$ accuracy depending on the consistency of the measurement and actuation system, the repeatability tolerance held in manufacturing, and the use of part calibration [1]. This implies that FSR™ sensors are designed for a qualitative, rather than quantitative measurement. However, in 1991 Jensen *et al.* were able to utilize these sensors to measure finger forces between 0-30N with an average error of only 1N. Figure 4.1 provides some of the technical details of the FSR™ sensor.

General FSR Characteristics

These are typical parameters. The FSR is a custom device and can be made for use outside these characteristics. Consult Sales Engineering with your specific requirements.

Simple FSR Devices and Arrays

PARAMETER	VALUE	NOTES
Size Range	Max = 20" x 24" (51 x 61 cm) Min = 0.2" x 0.2" (0.5 x 0.5 cm)	Any shape
Device thickness	0.008" to 0.050" (0.20 to 1.25 mm)	Dependent on materials
Force Sensitivity Range	< 100 g to > 10 kg	Dependent on mechanics
Pressure Sensitivity Range	< 1.5 psi to > 150 psi (< 0.1 kg/cm ² to > 10 kg/cm ²)	Dependent on mechanics
Part-to-Part Force Repeatability	± 15% to ± 25% of established nominal resistance	With a repeatable actuation system
Single Part Force Repeatability	± 2% to ± 5% of established nominal resistance	With a repeatable actuation system
Force Resolution	Better than 0.5% full scale	
Break Force (Turn-on Force)	20 g to 100 g (0.7 oz to 3.5 oz)	Dependent on mechanics and FSR build
Stand-Off Resistance	> 1MΩ	Unloaded, unbent
Switch Characteristic	Essentially zero travel	
Device Rise Time	1-2 msec (mechanical)	
Lifetime	> 10 million actuations	
Temperature Range	-30°C to +70°C	Dependent on materials
Maximum Current	1 mA/cm ² of applied force	
Sensitivity to Noise/Vibration	Not significantly affected	
EMI / ESD	Passive device	
Lead Attachment	Standard flex circuit techniques	

Figure 4.1: FSR™ details from the FSR™ integration guide [1]

4.3.2 FlexiForce™

The FlexiForce™ (figure 4.2) is a low cost piezoresistive sensor, and direct competitor to Interlink's FSR™. This sensor has already proven useful in an experiment similar to the one presented in this report [34]. Smit et al. (2003) successfully utilized these

sensors to measure the finger tip forces of physiotherapists during a spinal mobilization technique. This technique, which is designed to treat and assess cervical pain, essentially involves applying a force with your thumbs on either side of the patient's spine.

Figure 4.2: The FlexiForce™ Sensor by Tekscan, Inc.

The sensor consists of a polyester film bilayer, with silver deposited on each layer as the conductive material. A layer of pressure-sensitive ink defines the active sensing area which is 9.53mm in diameter. As this is smaller than the fingertip area applying the force, some of the force will bypass the sensor. This issue is not an uncommon problem, and Tekscan suggests that a rigid material, slightly smaller than the sensing area, be placed over it to ensure that all forces are accounted for.

As with the FSR™, FlexiForce™ sensors experience a saturation effect. The force at which the sensor saturates depends on its sensitivity, and the reference resistance in the sensors circuit. Figure 4.3.2 details some of the sensor's characteristics. These

sensors are capable of handling a large range of forces, the limits of which are determined by the pressure sensitivity of the conductive inks, which must be between 1 and 10,000 psi.

Sensor Properties	Model A201
Operating Range	15°F (-9°C) to 140°F (60°C)
Linearity (Error)	<+/- 5%
Repeatability	<+/- 2.5% of full scale (conditioned sensor, 80% force applied)
Hysteresis	<4.5% of full scale (conditioned sensor, 80% force applied)
Drift	<3% per logarithmic time scale (constant load of 90% sensor rating)
Temperature Sensitivity	Output variance up to 0.2% per degree F (approximately 0.36% per degree C). For loads >10 lbs., operating temperature can be increased to 165°F (74°C).

Figure 4.3: FlexiForce™ details from their Users Guide [2]

Chester and Watson (2000) [35] and Petty and Messenger (1996) [36] stated that pressure sensors placed between the fingers and the surface will interfere with the palpation of soft tissues, especially at the beginning of the pressure range. However, when questioned about this afterwards, 72.5% of the physiotherapists in the Smit (2003) experiment reported no tactile interference by the sensors, while 27.5% reported a slight disturbance, without problematic influence on the technique [34].

4.4 Sensor Choice

The choice was between the FSR™ and FlexiForce™ sensors because of their commercial availability and their previous use in applications similar to the one presented

here. While some of the other sensor technologies look promising, they are either still in the development stage, or have no documented use in situations similar to those in this experiment.

Tekscan's FlexiForce™ sensors were chosen because of their superior performance with respect to repeatability, drift and accuracy. The only potential downside to these sensors is robustness, although they are rated for 1,000,000 loading cycles. In this application, robustness did in fact prove to be an issue.

The extreme thinness, and the flexibility of these sensors is perhaps their most attractive feature. Initially it was a requirement that the sensors be capable of conforming to the surface of the doctor's fingertips, in an effort to minimize obstruction of tactile feedback. FlexiForce™ sensors, appeared to shine in this respect because of their previous uses as fingertip sensors in medical procedures [34]

Cost-wise, the FlexiForce™ sensors are very appealing. The sensors are purchased in packs of eight, with each sensor costing approximately ten dollars.

FlexiForce™ sensors are designed to be used with Tekscan's Economical Load and Force (ELF) system, which is capable of receiving data from the sensor through wireless transmission. The system utilizes a Microsoft Windows™ based software application which assists with sensor calibration, and allows for the graphical interpretation of applied force in real time. Purchasing the ELF system was deemed unnecessary, as National Instruments LabView software can do many of the same things and has already been acquired.

These sensors are simple to use and set up. They act as a variable resistor and can be incorporated into any number of circuits. In this case they were incorporated into a basic voltage divider. When unloaded, the sensor produces a resistance in the 20M Ω range. At full load this resistance drops to less than 10k Ω . The sensors are designed to handle voltages between 0.1V and 18V. Initially a 12V DC source was used in order to utilize the entire 0-10V range of the data acquisition card. For the

final palpation experiment a 5V source was used.

4.5 Sensorized Gloves

One of the easiest ways to use these sensors for palpation measurements is to affix them to a glove. This minimizes set-up time as the sensors do not need to be detached and re-attached to the doctor's finger-tips between tests. The glove is simply be taken off and then put back on. The use of a glove also helps to ensure that the location of the sensors with respect to the doctor's fingertips remains constant from test to test. While not ideal, because a glove with sensors on the fingertips still impedes direct tactile contact, this is a promising route to go, and it has been successfully used in the past to make similar measurements to the ones used in this project [37].

Vecchi *et al.* (2001) designed a glove equipped with eight piezoresistive sensors, with the goal of acquiring data on grasping activities. The glove made use of FSR™ technology and was shown to be feasible for measuring forces in the range of 0 to 10N.

Leung *et al.* (2003) used a sensorized glove to measure the forces applied by physicians performing external cephalic versions (ECV). An ECV is a procedure designed to move a fetus to a headfirst position by pushing on the mother's abdomen. The gloves were standard latex with 16 nylon reinforced neoprene rubber piezoresistors which were 6.4mm in diameter and 0.1mm thick. The sensors had a measurement error of less than 10% and were capable of handling a maximum pressure of 133kPa.

Before attempting to use the gloves, they tried using a pressure sensing mat on the mother's abdomen. This is an approach that was also considered for this palpation experiment. They found that the mat was difficult to affix to the abdomen, and that it complicated the procedure. They also found that wearing the sensorized gloves had no negative effects on the success rate of the ECV procedure.

Chapter 5

Force Sensor Testing

5.1 Experimental Protocol

A series of experiments were performed in order to accurately characterize the behaviour of Tekscan's FlexiForce™ piezoresistive force sensors. It was necessary to determine how the sensors would perform under the expected atypical loading conditions before the palpation experiment took place.

FlexiForce™ sensors were chosen for this experiment because of their mechanically flexible nature. However, it is important to note that the area of the sensor that actually does the force sensing is affected by bending (figure 5.1). That is, the sensor will not necessarily register the same resistance when a specific load is applied while the sensor is on a solid flat substrate, or a flexible curved substrate. A variety of experiments were performed in an attempt to determine the affects of various loading and substrate conditions. The end goal of these experiments was to determine the most accurate method of calibrating the sensors for the abdominal palpation experiment.

Figure 5.1: The sensing circle at the tip of FlexiForce™ sensors

Prior to each test series the sensor was conditioned per Tekscan's recommendations. 110% of the test load was applied to the sensor five times. The load remained on the sensor until the signal stabilized.

A total of seven test series were performed, each of which contained at least 40 individual tests. In each test, an identical load, ranging from 3 to 14N was applied to the sensor a specific number of times. All data were collected using LabVIEW software and a National Instruments data acquisition board. Output voltage was sampled at 30Hz.

5.1.1 Loading Conditions

For the palpation experiment, it was initially expected that the sensors would be attached directly to the doctor's fingertips - a curved, flexible substrate. In order to recreate this situation as accurately as possible during testing, a soft substrate and a curved surface were used.

Many different tests were performed, the results of which will be discussed shortly. However, they all centered on fourteen loading conditions (figure 5.2) designed to illuminate the effects that soft substrates, curved surfaces and the presence of pucks have on sensor behavior:

Figure 5.2: The fourteen loading conditions used during force sensor testing

5.1.2 Pucks

Pucks serve to improve sensor performance in three ways:

1. Discourage secondary loading
2. Discourage sensor bending
3. Encourage uniform force distribution

Both Tekscan, and other researchers using FlexiForce™ sensors [38] [39] recommend the use of pucks to ensure accurate and repeatable force application to the sensor surface. Pucks should be used when the area of loading exceeds that of the sensitive area of the sensor, or when the area of loading is much smaller than the sensing surface and needs to be spread out to prevent damage to the sensor.

Tekscan recommends either a hard or soft puck depending on the intended application. While a hard puck results in improved sensor performance, it also eliminates sensor flexibility which obstructs the doctor's haptic feedback. The flexibility of the sensors is one of the prime reasons that FlexiForce™ sensors were chosen for this project.

In order to determine what type of puck would best serve the palpation experiment, three different materials were tested (figure 5.3). Initially, only a cork and rubber pucks were considered as they are both flexible and will conform to a fingertip. Figure 5.5 is a photograph of a cork puck attached to the sensor.

Figure 5.3: The three puck materials. From left to right: cork, rubber, plastic

Figure 5.4 shows the differences in repeatability (standard deviation divided by the mean) when using a cork and rubber puck. Clearly the cork puck is the superior performer. The poor repeatability of the rubber puck could be attributed to its soft, spongy nature which may have introduced shear forces and non-uniform loading at

the sensor's surface, resulting in erratic and unpredictable behaviour.

Figure 5.4: A comparison of the repeatability behavior between rubber and cork pucks placed between the sensor and a hard substrate

In the end, both of these materials proved to be inadequate from a repeatability perspective, and rigid plastic pucks had to be used to boost sensor performance to an acceptable level. Rigid pucks provide the added advantage of ensuring a uniform force distribution across the sensor surface, but unfortunately prevent any sensor flexibility in the process.

The pucks were attached to the sensor with double sided carpet tape. At first various adhesives were used (re-usable glue stick and Krylon Easytack™), but they proved to be too weak. Figures 5.5, 5.6 and 5.7 are photographs showing the cork, rubber and plastic pucks attached to the sensors respectively.

Figure 5.5: A photograph of a cork puck attached to the FlexiForce™ sensor

Figure 5.6: A photograph of a rubber puck attached to the FlexiForce™ sensor

Figure 5.7: A photograph of the plastic pucks attached to the FlexiForce™ sensor

5.2 Force Sensor Test Apparatus

Initially sensor testing was to be performed using an apparatus designed by the 2005-06 Carleton University Patient Simulator Project (figure 5.8). Unfortunately, this device proved to suffer from a number of fundamental design flaws that rendered it

unacceptable.

Figure 5.8: Force sensor testing device designed by the Carleton University patient simulator group

First, it is not possible to center the plunger above the sensor. Second, the apparatus itself most likely inhibits complete load transfer to the sensor. Figure 5.9 shows how a moment caused by an improperly placed load would result in secondary loading through the test apparatus. The combination of these two deficiencies prompted the design of a new testing device.

Figure 5.9: Non-centered force causes a moment about the testing apparatus which leads to secondary loading

The new apparatus allows for a great deal of freedom when choosing different loading mechanisms, or substrates. It was designed with flexibility in mind, and minimizes any potential measurement error introduced by the apparatus itself. A point loading mechanism eliminates the need for a support structure to balance the load above the sensor. An assembly view of the new apparatus can be seen in figure 5.10.

Figure 5.10: Assembly view of the re-designed FlexiForce™ testing apparatus

The new test apparatus is designed to make testing as repeatable as possible. The sensor and the cylinder through which the force is applied are accurately located via a bracket (see Appendix B.1) to ensure that testing conditions and load applications are repeatable. The bracket locates the cylinder directly above the sensitive portion of the sensor prior to each test. The groove in the top of the base-plate has been machined to the specific dimensions of the sensor, as provided by Tekscan (see Appendix B.2). The entire apparatus was clamped to the edge of a desk using a “C” clamp.

The bracket from which the load is suspended was bolted together from four separate pieces for ease of machining. The four parts consist of two vertical elements and two horizontal elements (see Appendix B.3).

The big advantage of this design, besides the point-loading mechanism, is its flexibility in testing many different loading conditions. For example, the hole in the

base is deep enough to accommodate a flat or curved surface, as well as a soft, tissue-like material to simulate a finger-tip. The design of the soft and curved base can be seen in Appendix B.4. The design also allows for ample soft material to be placed above the sensor to simulate the abdomen.

The point loading device bolts onto the bracket and rests in the cylinder (Appendix B.5) as seen in the assembly drawing in figure 5.10. The point is placed within a cone-shaped divot in the top of a cylinder which in-turn rests on-top of the sensor. The cylinder has a diameter slightly smaller than that of the circular, force-transducing, area of the sensor. This point loading set-up helps ensure that loads which are not perfectly centered beneath the sensor do not cause uneven force application at the sensors surface. This helps improve inter-test repeatability.

Figure 5.11 is a photograph of the assembled test apparatus.

Figure 5.11: A photograph of the redesigned test apparatus

5.3 Test Set-up

A voltage dividing circuit was used to measure the sensor's response (figure 5.12). This circuit was chosen because it is simple to construct and modify. Voltage dividers have been successfully used in the past to test other thin film sensors by Jensen *et al.* [39]. Tekscan provides the specifications for a recommended circuit (figure 5.13), but it is more complex, utilizing an inverting operational amplifier, while not providing any real benefits. Tekscan engineers stated that a voltage dividing circuit will work

equally as well as their officially recommended one.

Figure 5.12: FlexiForce™ drive circuit utilized during testing.

Figure 5.13: FlexiForce™ driving circuit recommended by Tekscan

Initially a 9V battery was used to drive the circuit, however when the battery voltage output was monitored during sensor testing, it dropped significantly. Therefore, a regulated 12V DC power supply was purchased and used instead.

The effective sensitivity of the sensor can be adjusted by changing the reference resistance R_{ref} . Large resistances result in increased sensitivity, while small ones

decrease sensitivity. The idea is to chose a resistor such that the output reaches the maximum allowable voltage, in this case 10V, at the top end of the desired force scale. This provides for maximum resolution.

It is possible to roughly calculate the ideal reference resistance using the input voltage and the sensor's resistance at the desired full scale force; 20N in this case. Using a multi-meter the sensor's resistance with a 20N load was found to be approximately $8K\Omega$. Ohms law (equation 5.1) was used to derive an equation for the current passing through the circuit (I). Once an expression for the current is found, it can be substituted into an equation expressing the voltage across the reference resistor. It is this voltage which is being monitored by the data acquisition card.

$$V = IR \quad (5.1)$$

$$\begin{aligned} I &= \frac{V_{drive}}{R} \\ &= \frac{12}{R_s + R_{ref}} \end{aligned}$$

$$\begin{aligned} V_{out} &= IR_{ref} \\ 10 &= \frac{12R_{ref}}{5000 + R_{ref}} \end{aligned}$$

$$R_{ref} = 25000\Omega$$

where:

- V_{drive} is the driving voltage from the power supply
- R_s is the sensor resistance

- R_{ref} is the reference resistance

As seen in the above equation, a reference resistance of $25K\Omega$ theoretically provides for the optimal circuit set-up. However, in case the estimate of a 20N maximum is incorrect, and to allow for in-congruencies between sensors, a $15K\Omega$ resistor was used instead. Once again utilizing Ohms law it is possible to calculate the expected output voltage for a 20N load using a $15K\Omega$ reference resistance:

$$\begin{aligned} V_{out} &= \frac{V_{drive} R_{ref}}{5000 + R_{ref}} \\ &= \frac{(12)(15000)}{5000 + 15000} \\ &= 9V \end{aligned}$$

5.4 Force Sensor Repeatability

5.4.1 Background

Repeatability proved to be a source of frustration early on in testing. Repeatability refers to the ability of the sensor to consistently respond in the same way for a specific load. Initially, Tekscan's 25lb (120N) FlexiForce™ sensors were used, however measurements fluctuated wildly at low loads. Figure 5.14 demonstrates the repeatability with a 9N load. At this stage of the testing it was impossible to state the error in terms of force, as the sensors were not yet calibrated, but this is unnecessary as the measurement deviations are clearly far too large to be of any practical value.

Tekscan states the repeatability of their FlexiForce™ sensors as “less than 2.5% of full scale (conditioned sensor, 80% force applied)”. This is a rather cryptic explanation, and it was only after extensive testing that the repeatability behavior was truly understood.

The stated full scale force of the sensor is 120N. This means that the repeatability

is $\pm 3\text{N}$ (2.5% of 120N). This is constant throughout the force sensing range. While a 3N error could be considered acceptable towards the upper-end of the scale, it effectively renders the sensor useless at the lower end. This was a critical problem as accurate measurement in this range are required to adequately monitor palpation.

Repeatability Test - 9N

Figure 5.14: Typical repeatability test results for the 120N sensor when applying low loads

While attempting to determine the source of the poor repeatability, a great deal of electromagnetic interference was discovered in the laboratory where the testing was being performed. The sensor was effectively acting as an antenna. A number of different attempts were made to try and compensate for the interference. First, a very large resistor ($2M\Omega$) was placed in parallel with the sensor. This helped, but did not completely eliminate the random oscillations. Shielding the sensor by placing it inside of a Faraday cage was effective, but impractical. The most effective method proved to be collecting data via a differential input. This way the signal oscillations

due to the electromagnetic interference are irrelevant because both the positive and negative inputs are affected by the same interference and oscillate together. Unfortunately, even with the differential input negating the electrical interference, the repeatability of the sensors was still very poor.

Once it was determined that the sensor itself was the cause of the poor repeatability, Tekscan was consulted, and they recommended using the 1lb (4.5N) version of their FlexiForce™ sensor. Although the sensor is only rated to 4.5N, Tekscan states that predictable resistance changes occur up to ten times this value; although they are small. This effectively increases the sensors range to 45N. As long as the driving circuit is designed in such a way that these smaller resistance changes can be resolved, the sensor is still reliable. The trade off is that the smaller resistance changes that occur above 4.5N result in decreased resolution.

By Tekscan's statement about sensor performance, the repeatability of the new sensor should be better than $\pm 0.5\text{N}$ (2% of 20N). While not ideal, this is quite good for a thin, flexible sensor.

At the time of the testing, the sensors were not calibrated. This is of course because the point of the testing was to determine how to best perform the calibration. As a result, it was not possible to state the repeatability findings in the same terms as Tekscan. That is, it was not possible to state it in terms of a percentage of full scale force. However, meaningful comparisons may still be made between different sensors and loading conditions by comparing standard deviation magnitude with the mean. This provides a good qualitative assessment of sensor performance.

5.4.2 120N vs. 4.5N Sensor

Figure 5.14 demonstrated the poor repeatability of the 120N sensor at 9N. Compare this to figure 5.15 which shows the same test performed with the 4.5N sensor.

Figure 5.15: Typical repeatability test results for the 4.5N sensors when applying low loads

At 9N, the 4.5N sensor is almost two orders of magnitude better. This made the choice between these two sensors straightforward and all testing from this point forward was performed with the 4.5N sensor. Figure 5.16 demonstrates the repeatability behavior of these sensors at 3N, 7N and 14N. This figure shows how the output voltage stabilizes as the load on the sensor increases. Remember that this stabilizing is occurring because at higher loads the resistance changes are much smaller. So while repeatability appears to increase dramatically at heavier loads, in reality the actual error in terms of force remains the same. It is only the standard deviation which is decreasing relative to the mean.

Figure 5.16: Typical repeatability behavior of the 4.5N sensors at 3, 7 and 14 Newton loads

The repeatability of this type of sensor was previously tested by Vecchi, *et al.* [38], over a period of three days. They found that the sensors output changed 1.6% of the full scale load.

Figure 5.17: Calibration curves from Vecchi, *et al.* (2000) on three consecutive days

5.5 Force Sensor Test Data Analysis

A Matlab™ program was written to analyze the mean voltage output and the standard deviation for each test. The program works by accepting a threshold voltage value, above which the program assumes that a load has been applied to the sensor. All data below this value is omitted from the analysis. The program scans through the data and calculates the mean value for each load application. Once all of the data has been analyzed, the overall mean and standard deviations are returned. With these two numbers it is possible to analyze repeatability by calculating the standard deviation as a percentage of the mean.

The program allows for the analysis of multiple tests at one time. As the program runs, figures of the data from each test are plotted and displayed so that the appropriate threshold value can be entered by the user.

The Matlab™ code for this application can be found Appendix A.1

5.6 Force Sensor Test Results

There were a number of preconceived notions with respect to loading condition effects on repeatability, namely that:

- **Attaching a puck to the sensor will improve repeatability** - A puck should cause applied loads to distribute themselves in a similar manner upon each load application.
- **Attaching a puck to the sensor will increase output voltage** - A puck will eliminate any load from bypassing the sensor due to the fact that a finger-tip is larger than the sensor itself.
- **A curved surface will worsen repeatability** - A curved surface should cause the sensor to bend. When the sensor is bent, its resistance decreases. As various areas of the sensor are bent during loading, the resistance response should change in an unpredictable manner.
- **A curved surface will increase output voltage** - As mentioned above, sensor bending leads to a decrease in resistance, and therefore an increase in output voltage.
- **Soft substrates will worsen repeatability** - Soft substrates were expected to have the same effect on sensor performance as a curved surface. When loaded, the sensors are free to sink into the soft material and conform to it, resulting in unpredictable resistances.
- **Soft substrates will increase output voltage** - Soft substrates should result in sensor bending during loading, and therefore a decrease in resistance and increase in output voltage.

The following sections discuss the results of the force sensor tests, and why many of the above assumptions were incorrect.

5.6.1 Force Distribution

During testing, it was discovered that force distribution had a strong effect on sensor response. Force distribution is directly related to the stability of the test apparatus. An interesting correlation was found between how difficult the test apparatus was to stabilize, and the measured repeatability and signal output. Stability essentially refers to how “tipsy” the test was. For example, a test with only a flat hard base was very stable. However a test with a curved base, a puck, and soft material was far less stable, causing the cylinder to tip and the load to shift around.

The reason force distribution plays an important role is because of the resistance behavior of the sensor. A 10N force on half of the sensor will result in a much smaller resistance change than a 10N force spread out over the entire sensing surface. Figure 5.18 shows how the load vs. resistance behavior is highly non-linear.

Flexiforce Response - Resistance

Figure 5.18: FlexiForce™ resistance changes with increasing load

Due to the fact that these sensors were designed for forces up to 4.5N, only very small loads result in resistances in the steep part of the curve. Much of the repeatability testing was performed at heavier loads which push the resistance down to the flatter region.

A non-uniform load results in a varying resistance across the sensor surface. The areas experiencing most of the load will exhibit a very low resistance, while the unloaded, or less loaded areas will have a much higher resistance. This situation could result in one part of the sensor having a resistance way up in the steep area of the force versus resistance curve. The overall effect of this is a lower output voltage than expected for that specific load.

The results from a non-uniform force distribution can be demonstrated when the loading conditions are ordered from what is assumed to be the least, to most, uniform

loading configuration. A plot of the mean voltage output (figure 5.19) demonstrates a clear and definite trend relating uniformity to a higher output.

Figure 5.19: Test apparatus stability increases from left to right. Increased stability leads to a more uniform force distribution and therefore lower resistances and higher output.

In stable tests it is very easy to locate the cylinder directly above the sensitive part of the sensor using the locator bracket. However, in the less stable tests, such as those with a soft material between the sensor and the cylinder, the bracket does not help much as the cylinder is free to sink into the material and shift around the moment the bracket is removed and the load applied.

Figure 5.20 shows sensor repeatability from the least to most stable loading condition. There is a clear trend demonstrating that as stability increases, error decreases.

Figure 5.20: Test apparatus stability increases from left to right. Increased stability leads to more repeatable load applications and therefore minimizes error.

5.6.2 Soft Substrates

Repeatability

The presence of a soft material both above and below the sensor was expected to worsen repeatability. Soft material above the sensor results in the cylinder shifting about with each load application, varying the force distribution in the process. The same effect is expected with soft material placed beneath the sensor as the sensor is free to settle into the material in different ways. The material used beneath the sensor was much firmer than that used above the sensor, therefore its detrimental effects on repeatability were less severe.

Figure 5.21 shows the negative effects that soft substrates have on repeatability.

Figure 5.21: The effects of various substrates on sensor repeatability. The harder the substrate the better the repeatability.

Output Voltage

Soft materials placed above the sensor are the biggest contributors to destabilizing the test apparatus. Therefore, in tests where little or no sensor bending is expected, the resulting non-uniformity of the loading decreases the output voltages (figure 5.22). Also, without the placement of a puck and with the cylinder not directly resting on the sensor, secondary loading through the non-sensitive portions of the sensor contributes to the voltage decrease.

Figure 5.22: The effects of various substrates on sensor output. A harder substrate leads to higher output.

In tests where the sensor is free to bend, for example when a soft material is placed above the sensor, over a curved surface, the addition of the soft material causes an increase in voltage (figure 5.23). These results indicate that sensor bending has a greater effect on sensor behavior than non-uniform loading.

Figure 5.23: Sensor Bending decreases resistance and increases output voltage. The puck serves to inhibit sensor bending.

5.6.3 Curved Base

Repeatability

The effects of a curved base (figure 5.24) on repeatability are highly dependent on whether a puck or soft substrate is present. Overall, curvature had a relatively minor impression on repeatability when compared to the other loading variables. If anything, the curved base is a contributor to destabilizing the test and worsening repeatability.

Figure 5.24: The curved base of the test apparatus

Output Voltage

It should be mentioned that a curved base did not necessarily bend the sensor. With a soft surface beneath the sensor, it is easy to see that the sensor is not deforming much because the aluminum cylinder applying the load holds it flat. The cylinder is effectively acting as a puck. However, with a soft surface above the sensor, it is free to deform, and therefore changes in output voltage are expected (figure 5.23).

When a curved base is used with a soft material beneath the sensor the output voltage experiences a slight decrease. The most likely reason for this is that the curved surface is concentrating the force onto a smaller part of the sensor, resulting in less of a resistance drop than when the same force is applied with a flat base (figure 5.25).

Figure 5.25: The effects of a curved base with a soft material placed beneath the sensor on output voltage.

5.6.4 Soft Pucks

Repeatability

Initially rubber and cork pucks were used for sensor testing. Unfortunately, they did not have as great an impact as expected. It was hoped that these pucks would

provide improved repeatability and complete load transfer through the sensor, while still permitting the sensor to contour to the doctor's fingertip. Overall, the tests showed that they would prove useful in the more complex loading conditions, but the benefits were minimal.

When loading conditions are at their most basic, with only a flat base and no soft material above or below the sensor, there is no question that flexible pucks worsen repeatability (figure 5.26). It is difficult to say why the effect is so pronounced in these simple loading conditions. One could hypothesize that because the puck is softer than the aluminum base, the sensor is deforming in different ways each time a load is applied, resulting in a non-uniform force distribution.

Figure 5.26: The effects of a soft puck on repeatability.

At more complex loading conditions flexible pucks begin to improve repeatability, although not by a substantial margin. The degree to which they help is most likely limited by instability issues with the test apparatus. In less stable conditions, the

puck serves to further decrease stability, and load uniformity follows suit. However, the positive effect of discouraging sensor bending compensates for this. For example, when a puck is placed beneath the sensor while a soft material is present above the sensor, there is a noticeable improvement in repeatability, although it is minor (figure 5.27).

Figure 5.27: Repeatability improvement due to the presence of a soft puck. Note that these results are the average of three tests

Output Voltage

With a puck present the output voltage decreases in every case. The natural conclusion would be that the puck is helping to keep the sensor from deforming during loading. Sensor bending results in a decrease in resistance and thus an increase in output voltage. It is also known that pucks contribute to instability and therefore non-uniform force distribution in more complex loading conditions. This also decreases the output voltage. The effects of this are clearly demonstrated in figure 5.28.

Figure 5.28: The effects of pucks on output voltage. Note that these results are the average of three tests

5.6.5 Rigid Pucks

After the extensive testing with soft, flexible pucks, it was deemed necessary to try something more rigid. With the effects of the various loading conditions understood, a method to counter-act their negative effects was needed. In the tests which best simulated the experimental environment, the effects of sensor bending and non-uniform load distribution were having a large negative effect on sensor performance.

When the same loading configuration tests were performed with rigid pucks attached to both the top and bottom of the sensor, the difference was immediately apparent. In loading conditions where the repeatability was the worst, rigid pucks improved it by as much as five times (figure 5.29). Also, non-uniform loading issues were abolished and the same voltage output was measured across all tests (figure 5.30).

Figure 5.29: The improved repeatability from using rigid pucks. Note that these results are the average of three tests

Figure 5.30: The effects of various loading conditions on sensors equipped with rigid pucks. Note that these results are the average of three tests

Although the use of one rigid puck eliminates bending, it does not guarantee an

even force distribution, or abolish secondary loading. Therefore, a puck must be placed on both sides of the sensor for optimal performance.

Unfortunately, the use of rigid pucks inhibits tactile feedback, but for these first measurements of force and displacement of a healthy abdomen it is a worthwhile compromise to improve the quality of the data.

5.6.6 Drift

The drift performance of FlexiForce™ sensors was assessed in an earlier experiment by Vecchi, *et al.* [38]. They subjected the sensors to loads of 3,5,10, and 15N for ten minutes, while sampling voltage at 20Hz. Depending on the load, the sensors showed a drift of 5% to 10% relative to the initial load. The 5N and 10N loads tended to decrease in signal strength over the 10 minute periods, while the 15N load increased the signal. This error is not a concern for this project as loads will only be applied to the sensor for a few seconds at a time. The repeatability tests discussed in the previous section also serve to demonstrate that drift is not an issue over the time periods that force was applied to the sensor.

5.7 Sensor Damage - Delamination

At one point during testing the output voltages suddenly dropped significantly between test series. It turned out that this particular sensor had suffered some damage which permanently altered its behavior. With a 3N load the sensor's original resistance was in the $60k\Omega$ range. After the decrease in output voltage was noticed, the resistance at 3N was tested again and found to be $200k\Omega$. This is a significant change and an additional 2.3 Newtons of force was required to bring the resistance back down to $60k\Omega$ again.

The same observation was made after the rigid pucks were removed from a sensor.

The pucks are attached quite firmly with double-sided carpet tape, and pulling them off most likely led to some delamination. Vecchi *et al.* [38] mentioned that delamination was a concern with FlexiForce™ sensors during their comparison with FSR™ sensors.

To determine if delamination was really a problem, a series of four consecutive tests were performed where a 3 Newton load was applied to the sensor ten times. In between each test the rigid pucks were attached and removed from the sensor a few times. Assuming that delamination was the cause of the sudden drop in sensor performance and output magnitude, the same trend should be observed during these tests. The results of the tests can be seen in figures 5.31 and 5.32. As expected, repeatability worsens and the output magnitude decreases. This indicates that the sensors need to be handled with much more care than originally thought. From this point forward pucks were permanently attached to sensors to minimize any chance of delamination.

Figure 5.31: Effects of delamination of sensors on output voltage

Figure 5.32: Effects of delamination of sensors on repeatability

The fact that sensor damage manifests itself in the form of a change in resistance behavior provides for a method of damage detection. Monitoring the sensor's resistance for a specific load before and after testing can be used to assess the validity of the data.

5.8 Sensor Calibration

Each sensor was independently calibrated using ten discrete loads. Each load was applied five times, and the average value was taken. Matlab™ (see code in Appendix A.2) was used to fit two polynomial equations to the force versus voltage curve (figure 5.33). Using two equations to fit one curve was necessary as the quality of fit with a single curve was unacceptable. Finding a curve which accurately fit both the higher and lower sections of the data at the same time proved to be very difficult, typically resulting in curves such as the one shown in figure (5.34). Each sensor was independently calibrated with its own two third degree calibration curves.

Figure 5.33: Sensor calibration curves

Figure 5.34: An attempt to calibrate the sensors using only one curve.

Initially the calibration experiments were to be performed in the loading conditions which most accurately represented the expected experimental conditions, specifically, a curved base and puck, with a soft material both above and below the sensor. However, with the use of rigid pucks, the different loading conditions had little or no effect on sensor behavior. Therefore, calibration was performed on a hard flat surface.

In the past, this type of calibration has worked well with piezoresistive finger-tip sensors. Brimacombe *et al.* [40] experimented with various calibration techniques, for a similar model of Tekscan force sensor, and found that a user defined cubic calibration was best.

Jensen *et al.* 1991 [39] performed a calibration experiment on sensors very similar

to Tekscan's FlexiForce™ sensors. Jensen applied 10 different loads to the sensor for a period of 2 seconds each, while sampling at 40Hz. The 10 points were successfully fitted to a second-order polynomial equation.

During calibration a serious problem became evident with respect to sensor performance. Once calibrated, all of the sensors exhibited a decreasing response upon subsequent use. That is, the sensors resistance for a specific load increased between calibrations, resulting in a decreased output voltage. This problem is immediately apparent as a downward shift in the calibration curve (figure 5.35). Unfortunately, this shift was significant over the expected time period of the palpation experiment.

Figure 5.35: Downward shifting in sensor calibration curves

After extensive testing it was determined that the only way to solve the problem was to completely isolate the sensor, circuit, and data acquisition board from all power sources, save that of the battery powering the laptop to which the DAQ was connected. Once this was done, the calibration issues disappeared and the curves

lined up nicely.

Before this solution was found, it was believed that the power source itself might have been the problem. As a test, the relatively inexpensive 12V supply was replaced with a high-end device, but no improvement was noted. Temperature variations were also considered. Perhaps even the presence of a warm body could be augmenting the problem, but the addition of a temperature compensated resistor did not help either.

The next-most obvious solution was that the testing environment was to blame. The electrical situation in the building appeared to be unsteady, so tests were performed in two alternate locations; one of which was the Children's Hospital of Eastern Ontario, the location where the palpation experiment was to take place. Unfortunately, this too did not yield any improvements in sensor performance.

Using the laptop battery as the excitation voltage for the circuit is possible through a 5V source located on the USB-powered data acquisition board. The drop in excitation voltage from 12V to 5V can be compensated for by increasing the reference resistance.

As was done for the 12V source, Ohm's law (equation 5.1) can be used to calculate the ideal reference resistance. In this case, the desired output voltage at the maximum predicted load (20N) is 4.7V. To achieve this value a $125K\Omega$ resistance was required. In practice, a $100k\Omega$ resistance was used to accommodate any higher than expected forces and variations in sensor behavior.

Calibration tests were performed thirty minutes apart to mimic experimental conditions, and also twenty four hours apart. In both cases there was no significant change in sensor response (figure 5.36)

Figure 5.36: Two sensor calibrations performed thirty minutes apart. In between calibrations the sensor was loaded and unloaded multiple times to simulate experimental conditions.

5.9 Precision Error

With the sensors correctly calibrated it was finally possible to convert the voltage values to forces (see appendix A.5), and to test the sensor's error, or repeatability, in terms of force. The test consisted of repeatedly loading the sensors with the same force, twenty times. The test was performed with two different loads; one which utilized the lower calibration curve (5.5N), and the other the upper curve (13N).

Using a double sided t-test, the smaller load generated a precision error of $\pm 0.09\text{N}$ with a confidence of 99%. The larger generated a precision error of $\pm 0.13\text{N}$ with a confidence of 99%.

This is excellent, and is in fact far superior to Tekscan's stated repeatability of

$\pm 2.5\%$ of full scale force. For lower loads the repeatability is $\pm 0.45\%$ of full scale force, and for higher loads it is $\pm 0.65\%$ of full scale force. The superior performance is most likely due to the ideal loading conditions provided by the rigid pucks. Figure 5.37 shows the results of this test.

Figure 5.37: Sensor response for twenty consecutive applications of a 13N load.

Chapter 6

Palpation Experiment

The experiment consists of monitoring both force and displacement at a doctor's fingertips during abdominal palpation. Using these data, a basic understanding of the abdomen's response to palpation can be gained, as well as information on the palpation technique itself. The experiment provides guidelines for the development of a simulated abdomen.

6.1 Experimental Methods

This experiment was approved by the ethics committee at the Children's Hospital of Eastern Ontario. The approval notice can be found in Appendix E.

6.1.1 Sensor-Glove Attachment

It was necessary to attach the force sensors to the latex gloves in such a way that haptic feedback obstruction for the doctor was minimal. Initially a light tack, spray adhesive (Krylon Easytack™) was used to fix the sensors in place. A light tack adhesive was preferable over a stronger adhesive to ensure that sensor attachment was not permanent. Unfortunately the adhesive was too weak, and a better option proved to be taping the sensors at the proximal end of the finger. A second glove placed overtop

of the first secured the sensors in place at the fingertip (figure 6.1).

Figure 6.1: Photograph of how the tape secures the sensors to the doctor's fingers. Sensor positions are marked by numbers 1-4.

6.1.2 Force Sensor Placement

Four sensors were used to capture the palpation force. Initially the doctor recommended that as many as eleven sensors be used so that almost the entire surface of each finger is covered. However, upon further discussion it was decided that four sensors would be sufficient. It was assumed that most, if not all, of the force would be transmitted through four contact points. Figure 6.1 shows the location of each sensor. This arrangement becomes somewhat obvious when one tries to palpate someone, or

even themselves.

During a trial test the doctor agreed that the sensor placement was excellent. Data from that trial indicated that any additional sensors would be largely useless. Figure 6.2 shows the various forces at each of the four sensor positions during a single palpation. Most of the force occurred at positions 1 and 2, with a mere 6% of the total force occurring at positions 3 and 4. Adding additional sensors even farther away from the area of concentrated force would serve only to further inhibit the doctors tactile feedback and complicate data collection.

Figure 6.2: Force levels at the four sensor positions during one push into an abdomen

6.2 Motion Tracking

A Polaris™ hybrid position sensing system was used to track the motion of the doctor's finger-tip during palpation (figure 6.3). The system works by strobing infra-red (IR)

light (60Hz) to track the position of passive optical markers. The markers are retro-reflective which means they reflect light directly back to its source, as opposed to scattering it. One such marker was taped to the surface of the latex glove, directly above the finger nail of the doctor's middle finger (figure 6.4).

Figure 6.3: Polaris™ position sensor and an optical tool

Figure 6.4: Polaris™ optical tracking marker located on the fingernail of the doctor's middle finger.

Despite the fact that the IR light pulses at 60Hz, the maximum sampling rate was limited to less than 30Hz, which is still adequate. This is due to limitations in data transfer speed by the serial connection between the computer acquiring the data and the Polaris™ device.

6.3 Accuracy Validation

The Polaris™ system is designed to track the movement of specific tools. The tools generally consist of a pattern of three to six passive optical markers (figure 6.3). The use of multiple markers in a unique geometry allows the system to identify the tool by the marker configuration. Also, multiple markers means that the rotation of the tool can be monitored along with the position.

The error associated with tracking the position of a tool is calculated automatically as a 3D root mean squared error by the position sensor. If the volumetric accuracy of the sensor is above 0.35mm, a warning will come up in the Polaris™ software. The derivation of this error calculation can be found in technical bulletin NDI-TB-0005, “Polaris™ Calibration Performance and Methodology”.

For the palpation experiment the use of a tool was undesirable as they are large in size, and would be intrusive to the physician. Instead, a circular piece of retro-reflective tape made by 3M was used.

A simple test was performed in order to determine the accuracy at which the position of a stray marker could be tracked. The stray marker was attached to the base-plate of a tool and randomly moved around inside the tracking volume. Ideally, the difference in position between the tool and the tape would stay constant throughout the test. Matlab™ was used to compare the positions of the stray marker and the tool at every data point, and it was found that the difference in position remained constant throughout the test (see appendix A.11). This indicates that the

Polaris system is equally capable of tracking a single stray marker as it is an official tool.

6.3.1 Experimental Procedure

Data were collected at the right lower quadrant of the abdomen from 8 male volunteers. It was necessary to use either all male, or all female subjects due to differences in abdominal anatomy between the two sexes. This area of the abdomen houses the cecum, appendix, right ureter and lower kidney pole. Also, in males the right spermatic cord is located here, and in females, the right ovary and fallopian tube.

As mentioned in Chapter 3 there are two major types of palpation, survey and deep. For this experiment the doctor was palpating quite deeply, but he did not use his other hand to lend additional force to the palpation.

1. Each volunteer had his height, weight, and girth measured so that a body mass index (BMI) calculation could be made.
2. The volunteers were asked to lie down on an examination table and lift up their shirt so that the doctor could easily access the right lower quadrant of their abdomen.
3. A small mark was drawn in pen on the volunteer's abdomen to mark the point of palpation.
4. Using a calibrated optical tool, and the Polaris™ system, the doctor mapped out 3 points surrounding the palpation point. These three points were used to designate the abdominal plane and local coordinate system.
5. a small metal plate was placed on the volunteer's abdomen and the doctor tapped it three times. This tapping motion provides an obvious event in both the position and force data streams and was used for data synchronization.

6. The volunteers were instructed to relax and to place their arms at their sides. The doctor then began the palpation, being careful to palpate the same anatomical region for each volunteer. He pushed into their abdomens a total of ten times.

6.4 Palpation Data Preparation

6.4.1 Position Interpolation

In practice, the rate at which the Polaris™ system captures data is random, generally falling in the range of 23-27 Hz. The force data were being gathered by a separate computer at exactly 30Hz. These different sampling rates cause a problem when attempting to synchronize the two data streams for analysis.

In order to accurately and easily generate plots of the data it is necessary that both the force and position data consist of the same number of data points. A program was written to automatically re-sample the position data up to 30Hz using Matlab's spline interpolation technology (See appendix A.3).

6.4.2 Coordinate Transformation

The use of a stray marker necessitates the use of the default Polaris™ coordinate system. This system has its origin within the position sensor, and is not adjustable (figure 6.5).

Figure 6.5: The default coordinate system of the Polaris^{TM} motion tracker

While tracking position in the default coordinate system (henceforth referred to as the global reference system (GRS)) is not optional, in post processing it is possible to transform the data into a more useful and intuitive coordinate system that is local to the abdomen. This coordinate system will be referred to as the local reference system (LRS) and is unique to each volunteer.

Because abdomens are not perfectly flat, the LRS is defined through a series of three points surrounding the point of palpation. This results in a z-axis which is normal to the abdominal surface at the palpation point. The positive z-axis extends into the abdomen. A graphical representation of the transformation can be seen in figure 6.6.

Figure 6.6: Graphical representation of how the GRS is transformed into the LRS

in figure 6.6:

- j is the point of palpation
- R_j is the position of the palpation point in GRS coordinates. This is the point being tracked by the Polaris™ system.
- R_i is the origin of the LRS in GRS coordinates
- r_{ij} is the position of the palpation point in LRS (x', y', z') coordinates
- Points k and l make up the abdominal plane along with R_i

These variables are related via the following equation (6.1):

$$R_j = R_i + [\phi]r_{ij} \quad (6.1)$$

where $[\phi]$ is a 3X3 transformation matrix comprised of the orthogonal unit vectors defining the LRS

Equation 6.1 can be rearranged so that it solves for r_{ij} :

$$r_{ij} = [\phi]^{-1}[R_j - R_i] \quad (6.2)$$

Or in expanded form:

$$\begin{pmatrix} r_{ij,x'} \\ r_{ij,y'} \\ r_{ij,z'} \end{pmatrix} = \left[\begin{pmatrix} \hat{x} \\ \vdots \\ \vdots \end{pmatrix} \begin{pmatrix} \hat{y} \\ \vdots \\ \vdots \end{pmatrix} \begin{pmatrix} \hat{z} \\ \vdots \\ \vdots \end{pmatrix} \right]^{-1} \left[\begin{pmatrix} R_{j,X} \\ R_{j,Y} \\ R_{j,Z} \end{pmatrix} - \begin{pmatrix} R_{i,X} \\ R_{i,Y} \\ R_{i,Z} \end{pmatrix} \right]$$

The first step to completing the transformation is to collect the position of the three points on the surface of the abdomen, surrounding the point of palpation (j). These three points (i, k, l) form a plane in 3D space which designates the basis for the LRS. One of these three points (i) will be chosen as the origin of the LRS. All three of the points will be used to find the unit vectors which make up the transformation matrix $[\phi]$.

Taking the cross product of vectors $i\vec{k}$ and $i\vec{l}$ results in a vector normal to the plane. Next, taking the cross product of the normal and either of the two original vectors ($i\vec{k}$ or $i\vec{l}$) results in a third orthogonal vector. These three vectors define the

LRS. The only remaining step is to normalize the vectors by dividing them by their length. We now have the three unit vectors required to make up $[\phi]$.

Using Matlab™ all of the position data collected by the Polaris™ system were transformed into the appropriate LRS coordinates via the code found in Appendix A.4

6.4.3 Voltage to Force Conversion

The sensors were calibrated approximately 30 minutes before testing began. The conversion from Volts to Newtons was performed by the Matlab™ script found in Appendix A.5. The program accepts the polynomial coefficients determined by the calibration code and applies the correct conversion based on the sensor number, and on the voltage level. The program automatically determines whether the upper or lower calibration curve should be used.

6.4.4 Data Stream Synchronization

Once the position and force data have been converted they combined into one array using the code found in Appendix A.7. Matlab™ was then used to synchronize the force and position data streams. (see Appendix A.8). As mentioned in the experimental procedure section, prior to each palpation, the doctor tapped on a metal plate with his middle finger (the finger to which the optical position marker was attached). The purpose of the tap was to create an event which could be easily detected in both the force and position data streams.

The synchronization code utilizes the fingertip velocity (found via the code in Appendix A.6) and the applied force at position 1 during tapping. The program scans through the data until a tap is detected as a sudden increase in velocity. With the tap identified the program looks for the next point at which the velocity is closest

to zero. This is the point at which the sensor on that fingertip should be registering the maximum force occurring during the tap. The program logs this velocity, and the accompanying displacement data point and then searches for the corresponding force. Once the peak force is found the force data is shifted so that the two streams align.

6.4.5 Zeroing the Data

The final step in the data preparation is to zero both the force and position starting values. The first step is to eliminate the offset from the x,y and z position data. The palpation point was in the center of the plane designated by 3 points on the volunteer's abdomen. However, during the conversion from the GRS, the origin is placed at one of the three defining points, not at the palpation point. Although this step is not necessary, zeroing the position makes visualizing the palpation movement a little easier.

Prior to beginning each palpation, and directly after tapping, the doctor rested his fingers lightly on the palpation point for a few seconds. This left an easily discernable point in the force and displacement data. Matlab™ was used to apply the offset value to the synchronized data (see Appendix A.10). The program plots each of the 4 data streams requiring zeroing and then prompts the user for an offset value. Once the values have been entered, the offset is applied to all of the data points in the stream.

6.5 Data Analysis

With all of the necessary conversions made, and the data synchronized and zeroed, analysis could begin.

The experiment was designed to examine three major palpation attributes:

1. The apparent stiffness of an abdomen during palpation, and what affects this stiffness
2. Maximum force applied to the abdomen during palpation
3. Maximum displacement into the abdomen during palpation

In addition to these main goals, the experiment also shed some light on how repeatable the doctor's palpation technique was, both on the same volunteer, and between volunteers. Also, relations between BMI and abdominal response were explored.

A Matlab™ program was written to generate 6 plots for each volunteer (Appendix A.12):

1. Force vs. displacement for each of the ten pushes into the volunteer's abdomen
2. Displacement vs. velocity
3. 3-Dimensional motion
4. A combined plot of force, displacement and velocity vs time
5. The first, last and middle stiffness curve plotted on a single chart
6. All of the stiffness curves plotted on a single chart

The next chapter demonstrates these plots and explains them in detail.

Chapter 7

Palpation Results

This Chapter presents the results of the palpation experiment as a series of tables and charts. Stiffness, velocity, force, and displacement are presented and key features of each are explored. Anatomical measurements and the body mass index's (BMI) of the volunteers are listed in tabular form, and plots are made attempting to correlate the characteristics of abdominal stiffness with increasing BMI.

When discussing these results, a single loading and unloading cycle will be referred to as a “push”, as in a single push into the abdomen. The term “palpation” will refer to the entire series of pushes.

7.1 Subject Data

Table 7.1 shows the anatomical measurements for each of the volunteers. BMI was calculated by dividing the subject's weight by his height squared:

$$BMI = \frac{Weight(kg)}{Height^2(m)} \quad (7.1)$$

Volunteer	Weight (kg)	Height (cm)	Girth (cm)	BMI
1	80.9	174.5	94.5	26.4
2	88.2	177	97.5	28.2
3	61.7	162	77.5	23.5
4	68.2	164.8	82.5	25.5
5	89.3	187.7	86	25.0
6	71.0	174	77	23.5
7	84.9	176	98	27.4
8	85.9	194	90.5	22.7

Table 7.1: Anatomical measurements and BMI of the volunteers

7.2 Abdominal Stiffness

Abdominal stiffness was analyzed via plots of total palpation force versus displacement into the abdomen, along the z-axis of the LRS. It may be recalled from the previous Chapter that this reference system was established through a series of points on each of the volunteer's abdomens. Figures 7.1 through 7.8 show the stiffness plots for each push into the abdomen for every volunteer.

Some palpations, such as those for volunteer #5 (figure 7.5) appear fairly linear in nature, but the majority show definite non-linearity. The stiffness curves seem to consist of two separate areas. The first area is relatively linear and has a modest slope. This area makes up the majority of the curve. The second part of the curve is defined by an abrupt increase in slope. This sudden increase always occurs very close to the maximum displacement. The characteristics of these two parts of the stiffness curve will be discussed in detail in the next chapter.

Hysteresis is present in nearly all of the palpations, to varying degrees. For volunteer #3 the hysteresis was minimal (figure 7.3), while for volunteer #4 it was excessive (figure 7.4).

Note that due to a software error in the program responsible for force data collection, the results for a number of the volunteers were limited to less than 10 pushes. The exact cause of the error is unknown, but may be due to a software buffer becoming full during the collection of longer data sets (there is a delay between the computer collecting the data and writing that measurement to disk). This error was not experienced during the testing of the force sensors, or during the trial run of the palpation experiment.

Figure 7.1: Abdominal stiffness curves for volunteer #1. Each plot represents one push into the abdomen.

Figure 7.2: Abdominal stiffness curves for volunteer #2. Each plot represents one push into the abdomen.

Figure 7.3: Abdominal stiffness curves for volunteer #3. Each plot represents one push into the abdomen.

Figure 7.3: Abdominal stiffness curves for volunteer #3. Each plot represents one push into the abdomen.

Figure 7.4: Abdominal stiffness curves for volunteer #4. Each plot represents one push into the abdomen.

Figure 7.4: Abdominal stiffness curves for volunteer #4. Each plot represents one push into the abdomen.

Figure 7.5: Abdominal stiffness curves for volunteer #5. Each plot represents one push into the abdomen.

Figure 7.5: Abdominal stiffness curves for volunteer #5. Each plot represents one push into the abdomen.

Figure 7.6: Abdominal stiffness curves for volunteer #6. Each plot represents one push into the abdomen.

Figure 7.6: Abdominal stiffness curves for volunteer #6. Each plot represents one push into the abdomen.

Figure 7.7: Abdominal stiffness curves for volunteer #7. Each plot represents one push into the abdomen.

Figure 7.7: Abdominal stiffness for volunteer #7. Each plot represents one push into the abdomen.

Figure 7.8: Abdominal stiffness curves for volunteer #8. Each plot represents one push into the abdomen.

Figure 7.8: Abdominal stiffness curves for volunteer #8. Each plot represents one push into the abdomen.

It may be easier to envision the various aspects of the palpation using figure 7.9 which shows force, displacement and velocity as three plots in a vertical column, aligned by time. With this plot it is easy to line up events, such as maximum force with displacement and velocity values at the same instant. The plots for the other seven volunteers may be found in Appendix D.

Figure 7.9: Force, displacement (normal to abdominal surface) and velocity aligned with time.

7.3 Palpation Velocity

Figures 7.10 and 7.11 show the complete velocity versus z-displacement charts for the eight volunteers. The charts show that the doctor's velocity while pushing into the abdomen is often two to three times slower than the unloading phase. This pattern is the same for all palpations, with the maximum velocities being reached about half way between the abdomen's surface and the largest displacement.

Shallower palpations tend to be accompanied by lower velocities (figure 7.10 (c)). However, deeper palpations are not necessarily linked to higher velocities (figure 7.10 (b)).

Velocity was calculated by dividing the difference between subsequent displacement values by the sampling period (one 30th of a second).

7.4 3-Dimensional Movement

Figure C.2 is an example of the motion of the doctor's fingertip during an entire palpation session, on a single volunteer. The 3D motion plots for the other 7 volunteers can be found in Appendix C.

During the palpation, side to side movement is minor, generally not exceeding 10mm in any one direction. The fingertip does not follow a straight path, but waivers back and forth as the doctor pushes in and lifts out. Note that on the 3D motion charts the z-displacements have been changed to negative values to make for more intuitive plot. The magnitudes remain the same.

(a) Volunteer #1

(b) Volunteer #2

(c) Volunteer #3

(d) Volunteer #4

Figure 7.10: Phase Planes for volunteers 1 through 4.

(a) Volunteer #5

(b) Volunteer #6

(c) Volunteer #7

(d) Volunteer #8

Figure 7.11: Phase Planes for volunteers 5 through 8.

Figure 7.12: 3D motion of the doctor's fingertip during the palpation of volunteer #4.

7.5 Maximum Force

There was a large difference in maximum force between volunteers, with the lowest being 8.7N and the highest 15.3N. Figure 7.13 shows the maximum palpation forces plotted in order of increasing volunteer body mass index (BMI). BMI is simply a measure of a person's weight relative to their height. It is commonly used by physicians to classify people as over or under-weight. A rating of 18 - 25 is considered optimal, less than 18 is under-weight, greater than 25 is over-weight and 30 suggests that the person is obese.

In general the palpations which showed large maximum forces also had higher

average forces (figure 7.14). Overall, the average peak force for all of the volunteers was 9.45N.

Figure 7.13: Maximum force applied to the abdomen for each volunteer, ordered by increasing BMI.

Figure 7.14: Average of the peak force for each push, ordered by increasing BMI.

Volunteer	Avg Max Force (N)	Standard Deviation
1	9.67	1.23
2	8.30	0.81
3	11.27	2.04
4	12.35	1.01
5	8.51	0.31
6	8.70	1.07
7	10.0	1.51
8	8.94	0.82

Table 7.2: Averages of the maximum force during each push, and accompanying standard deviations.

Table 7.5 lists the average peak push force for each volunteer, and its accompanying standard deviation. The average standard deviation for the entire experiment was 1.07N.

7.6 Maximum Displacement

When maximum displacement is plotted with increasing BMI, there is a trend towards deeper displacement (figure 7.15). The same trend is observed in a plot of the average maximum displacement for each push (figure 7.16). Overall, the average peak palpation depth was 54.3mm.

Figure 7.15: Maximum displacement into the abdomen for each volunteer, ordered by increasing BMI.

Figure 7.16: Average of the peak displacement for each push, ordered by increasing BMI.

Table 7.6 lists the average peak displacement for each volunteer, and its accompanying standard deviation. The average standard deviation for the entire experiment was 2.33mm

7.7 Force at Each Sensor

Figure 7.17 shows a typical force vs. time profile for one push, at each sensor position. In the previous chapter, figure 6.1 showed the numerical position of each sensor and figure 6.2 showed the force measured at that location. Here we can see that the greatest force is still applied at positions 1 and 2. However, positions 3 and 4 played a larger role than anticipated. During the trial run of the experiment the forces at these sensors were very low.

Volunteer	Avg Max Displacement (mm)	Standard Deviation
1	55.60	1.76
2	66.69	2.35
3	47.55	2.81
4	56.17	0.074
5	62.03	1.49
6	54.45	3.54
7	53.82	2.58
8	46.3	2.73

Table 7.3: Averages of the maximum displacement during each push, and the accompanying standard deviations.

Figure 7.17: Forces occurring at each sensor during a typical palpation.

Chapter 8

Discussion

This chapter reviews the plots presented in the results section, and discusses the attributes which have relevance to patient simulation. The response of the abdomen is explored, with emphasis on the non-linear and hysteresis behavior seen in the stiffness curves. BMI and its influences on stiffness, force and displacement are discussed. Finally, the doctor's palpation technique itself will be analyzed from a viewpoint of repeatability. Potential sources of error are mentioned, along with suggestions of how to minimize them.

Very little data on abdominal stiffness exists. The results of this experiment make two useful contributions. First, they provide the basic data on the stiffness characteristics of the abdomen in general, and second, they establish a goal for realism in abdominal sign simulation.

8.1 Abdominal Stiffness

The abdomen is comprised of multiple layers of soft tissue, namely skin, fat, muscle and internal organs. While the material properties of the abdomen in general have not been explored, the behavior of the various soft tissues is well documented. Soft tissues are viscoelastic in nature, meaning that they exhibit the behavior of both

elastic and viscous materials. That is, the shape of the stiffness curve depends on the rate of loading and unloading.

Usually when the stiffness of a soft tissue is tested, the tissue is first pre-conditioned until the stiffness curve remains constant for a specific loading condition. This takes about 10 cycles, and is due to fluid extrusion during periods of inactivity. During this experiment, no pre-conditioning took place and the results show that none was necessary. Figure 8.1 shows three stiffness plots for volunteer #1. The solid line represents the first push, the dashed line represents the middle, and the dashed/dotted line the final push. This plot shows none of the characteristic decreases in stiffness that signal the need for preconditioning.

Figure 8.1: Initial (solid line), middle (dashed) and final (dashed/dotted) plots of stiffness for volunteer #1.

8.1.1 Hysteresis

Hysteresis is common in the loading and unloading of soft tissues, due to their viscoelastic nature. If the abdomen was purely elastic, the stiffness curve would be linear and Hooke's law would apply ($F = kx$). Where F is the applied force, x is the displacement and k is the stiffness. Linear elastic materials obeying this equation have the same loading and unloading path. However, the viscous nature of soft tissue causes energy loss in the form of heat, between loading and unloading. This energy loss is characteristically visible as a loop in plots of stress versus strain and is called hysteresis.

This theory is best supported by the stiffness plots for volunteers 3 and 4. Volunteer 3 shows a minimal amount of hysteresis, while volunteer 4 shows a great deal of it. Exploring the other plots for these two people, it was noted that the unloading velocity for volunteer 4 was much higher than that for volunteer 3 (approximately 125mm/s vs. 80mm/s). This indicates that the force on the sensors is decreased due to rapid unloading. The internal friction caused by the viscous nature of the abdomen's tissue is limiting the force of the elastic response.

With a clear indication that unloading velocity is effecting hysteresis magnitude, it follows naturally that higher loading velocities would cause steeper stiffness curves. However, this does not seem to be the case. For volunteer 7, the first push shows a far steeper curve than the fifth push, but there is no obvious difference in their velocities. In fact, if anything, the fifth push had the higher velocity.

In general the slopes of the stiffness curves are the same, despite differences in loading velocity. It would have been possible to test this theory with greater accuracy by asking the doctor to palpate unnaturally quickly or slowly, but this was not the purpose of the experiment, and a natural palpation motion was imperative.

8.1.2 Non-linearity

Although a number of palpations suggest otherwise, the overall experiment demonstrates that the human abdomen does not behave like a linear-elastic structure during palpation.

A non-linear stiffness curve is expected due to the material properties of the soft tissue layers. Because the abdomen is a layered structure, the deeper the doctor palpates, the more tissue types effect the stiffness. Therefore, it is reasonable to hypothesize that the early, more linear portion of the curve represents the part of the palpation where the doctor's fingertips are moving through the fatty portion of the abdomen. This layer would offer much less resistance to palpation than the muscle that lies beneath it.

Evidence that the layered structure of the abdomen is responsible for the non-linearity is provided by the Young's modulus of the various abdominal tissues. Tissues deeper in the abdomen, such as the muscle, and fat surrounding the internal organs, have higher Young's modulus values than the fatty tissue near the surface [41]:

- Fatty layer of the human abdominal wall: 0.1Mpa
- Human abdominal muscle (oblique abdominis externus and rectus abdominus): 0.3Mpa
- Human abdominal muscle with fascia:
  - Oblique abdominis externus: 0.78 - 3.92Mpa
  - Oblique abdominis internus: 4.9 - 16.67Mpa
- Encapsulated human visceral fat (fat surrounding internal organs): 1.08 - 1.96Mpa

To test this hypothesis one can compare stiffness curves between the slimmer and fitter volunteers with those of the not-so-slim volunteers. Volunteer #2 had the highest BMI, indicating that he was the most overweight. Volunteer #8 had the lowest BMI, but due to him being much taller than volunteer #2, the comparison

is not as valuable. Therefore it is more appropriate to compare with volunteer #6, whose height is close to that of #2, but whose BMI is much lower. The stiffness curve experiences a rapid increase in slope at approximately 60mm for volunteer #2 and 50mm for volunteer #6, supporting the hypothesis that the thickness of the fatty tissue layer plays a large role in the length of the first portion of the stiffness curve.

8.1.3 Repeatability

The expected result for the palpation of a single person's abdomen is that the stiffness curves will be very similar for each push. The material properties of the abdomen are not likely to change during the palpation, therefore the most likely cause of differences in the curves is changes in palpation depth and velocity by the doctor. Figure 8.2 shows a typical stiffness plot for every push into the abdomen on one chart.

Figure 8.2: Stiffness curves for the entire palpation of volunteer #4

8.2 Influences of Body Mass Index

It should be noted that BMI values can be misleading depending on a person's body type and fitness level. For example, someone of the ectomorph body type (very tall) may register a BMI in the healthy weight range, when in fact they are overweight. This is due to the added height of this body type skewing the calculation. Also, a person who is very fit, with little body fat but a lot of muscle mass may incorrectly register as overweight.

Despite these shortcomings, influences of BMI on the test results have potential significance for patient simulation, as an ideal patient simulator would realistically simulate many different body types. Therefore it would be desirable to see no real differences in the stiffness curves between those with a low BMI and those with a high BMI. However, this is not the case and BMI does have an effect.

The most obvious difference between those with low and high BMI is maximum palpation depth. The maximum depth for volunteer #2 was 69.2mm, while volunteer #6's was only 59.2mm. Note that the force required to achieve these two depths differed only by 0.6N. This difference in displacement is most likely due to the increased thickness of the fatty layer on volunteer #2. The fatty layer offers less resistance than the muscles which lie beneath it. The results indicate that the higher a person's BMI, the longer the initial phase of the stiffness curve.

The slope of the stiffness curve is also affected by BMI. Figure 8.3 shows typical stiffness curves for both volunteers #2 and #6 on the same plot. The difference in slope in the first portion of the curve clearly shows that volunteer #2 with his higher BMI has a lower slope. This makes sense as one would expect fatty tissue to offer far less resistance during palpation than muscle or organs.

The amount of force that the doctor applies appears to have no relation to BMI. Figure 7.14 in the previous chapter indicates that doctor palpates with a certain

amount of force regardless of the patient's body type. BMI also has no notable effect on hysteresis.

Figure 8.3: Stiffness curve comparison between volunteer #2 (high BMI - solid line) and volunteer #6 (low BMI - dashed line).

8.3 Palpation Repeatability

The doctor who performed the palpation is an experienced paediatric abdominal surgeon. He uses abdominal palpation regularly as a diagnostic tool. It is reasonable to assume that his palpation technique is representative of experienced doctors as a whole.

When analyzing the data one volunteer at a time, it is surprising how inconsistent the doctor is with respect to applied force. This is neither good nor bad, but one

would expect that the average maximum force for each push to be roughly the same for each volunteer. That result would indicate that for a basic diagnostic palpation the doctor stops pushing when he experiences a characteristic resistance. However, for the eight volunteers, the average maximum force ranged from a low of 8.3N to a high of 12.3N.

Tables 7.6 and 7.5 in the previous chapter list the average force and displacements for each volunteer, along with the standard deviations. On average, the standard deviations between peak force were 1.1N.

This is useful information for patient simulator engineers because it shows just how much palpation forces and displacements can vary, not just between people, but during the palpation itself.

8.4 Potential Error Sources

8.4.1 Position

There were a number of potential error sources for both the force and position data. First, for the position data there is the problem of skewed coordinate systems. While efforts were made to ensure that the LRS was properly aligned so that the z-axis was orthogonal to the abdominal surface at the point of palpation, some of the 3D motion plots show that a perfect LRS was not always achieved. The potential implications of this on data quality are an underestimation of palpation depth and an overestimation of side to side movement. Changing the palpation depth would also change the shape of the stiffness curve, giving it an artificially steep slope. However, in reviewing the 3D motion plots there is no LRS that was so highly skewed as to jeopardize the usefulness of the data.

Another source of error for position data is breathing. Although the palpation

was short, the volunteers were not asked to hold their breath as that would have been unrealistic. When a person breathes, their abdomen moves up and down on its own, not in reaction to the palpation force.

The displacement measurements in these tests were made from the doctor's fingertip, not the surface of the abdomen. This is likely not a source of error, as the doctor's fingertip should have remained in the same position relative to the abdominal surface throughout the palpation.

The last problem that may have affected the position data is momentary obstructions to the Polaris™ line of sight. This was happening during some of the palpations, at their maximum depths. It is possible to tell when this occurs by watching the data collection software on the computer. When this happened the camera was repositioned and the palpation for that volunteer had to be redone. However, it is possible that some of the tracker losses were not noticed and that there are gaps in the position data. That being said, no gaps were apparent when viewing the data, and gaps that may have occurred would have been very small, and most likely not an issue as the up-sampling process would have filled them in.

8.4.2 Force

The performance of the force sensors has already been discussed in great detail, so this will be a brief summary of how the FlexiForce™ sensors may have introduced error into the data. The sensors had a tendency to drift between calibrations. However this problem seems to have been solved by isolating the power supply. The precision error of the sensors was calculated as $\pm 0.09N$.

The final potential source of force error involves the placement of the sensors. It is possible that additional sensors should have been used in order to ensure that all of the force applied to the abdomen was accounted for. If there was any force missed during testing, it was most likely very minor, and would not have lead to any

significant changes in the results. The position of the fingers during palpation clearly demonstrates that the three fingers to which sensors are attached are doing all of the pushing. That, combined with the doctor's comments and the force distribution graphs indicate that no additional sensors are required.

8.5 Simulating Stiffness

To simulate the stiffness revealed through these experiments, a generic curve which best represents the overall response should be used. Figures 8.4 and 8.5 show the average curves of all of the stiffness curves for volunteers #1 and #3 respectively. Volunteers #1 and #3 have very little hysteresis in their force versus displacement curves, despite the fact that their loading and unloading velocities differed. This implies that their abdomens are less viscous than most of the other volunteers. Simulating one of these curves is a logical start for a simulator engineering team. Once this basic behaviour is successfully replicated, more complex features, such as hysteresis, may be integrated in the simulator.

Figure 8.4: Averaged abdominal stiffness curve for volunteer #1.

Figure 8.5: Averaged abdominal stiffness curve for volunteer #3.

Chapter 9

Conclusions and Future Work

This Chapter draws together the discussions in the previous chapter and applies them to recommendations for developing an abdominal sign simulator. Suggestions on how to improve data quality in future experiments are also made.

9.1 Conclusions and Recommendations

9.1.1 Data Collection and Sensor Testing

The newly designed force sensor test apparatus proved to be adequate for testing FlexiForce™ sensors. However, the sensors themselves caused a never-ending stream of problems. Although they worked quite well in the end, they were solely responsible for setting the project back many months. A better alternative should be found for future data collection attempts in order to avoid more problems. It is recommended that future researchers acquire sensors demonstrating improved robustness and performance without the presence of rigid pucks.

The Polaris device worked admirably. However, for future palpation experiments the use of retro-reflective tape is not recommended. The camera had difficulty tracking the tape during the deepest parts of the palpation as the doctor's fingers often

pushed so far into the abdomen that the camera's line of sight was obstructed. This was a source of frustration during data collection as the camera needed to be repositioned multiple times to ensure that it would not lose sight of the marker.

9.1.2 Patient Simulation

The purpose of this experiment was to discover exactly what the doctor was doing and feeling during the abdominal palpation. The hope was that this information could then be used to develop a realistic abdominal sign simulator. The experiment was a success and the first steps towards building an accurate simulation have been taken.

The stiffness curves presented in this report indicate that body type strongly influences abdominal response and palpation depth. Therefore the first decision that the simulator engineers will have to make is what body type, or types, they wish to simulate.

The ultimate goal of a patient simulator is to be able to accurately replicate not simply a healthy individual, but one suffering from many potential medical conditions. These different medical conditions often present themselves as changes in perceived stiffness from the palpators point of view. The engineering team may wish to start designing with variable stiffness capabilities in mind. This not only leaves the door open for simulating various medical conditions in the future, but also different body types.

The results of the experiment provide a number of guidelines to which an abdominal simulator should adhere:

- When simulating a slim, fit person, the simulator should have a maximum palpable depth of no less than 65mm. When simulating an overweight individual

the maximum palpable depth should be no less than 75mm. Average palpation depths of $54\text{mm} \pm 2\text{mm}$ should be expected.

- Maximum palpation depths should be achieved with forces ranging from 9N to 14N. Average palpation forces can be expected to be $9.5\text{N} \pm 1\text{N}$
- The simulation area should match the physical dimensions of a real abdomen. This allows the doctor to rest his or her hands in a realistic manner on the abdominal surface. Also along these lines, the doctor was using the volunteer's anatomy to locate the proper palpation spot (belly button and hip bone for example). Therefore it is recommended that realistic anatomical structures be included in the simulator.
- The simulator should have a way of varying its apparent stiffness. Ideally the simulator would be able to alter its response in such a way as to vary the slope and length of the first part of the stiffness curve. Both increasing the length of this section, and decreasing the slope represent an increasing BMI.
- If possible the simulator should mimic the viscoelastic effects of real soft tissues. Care should be taken so that the hysteresis behavior demonstrated in the stiffness plots is replicated by the simulator, and that the abdomen does not end up feeling like a spring. While loading velocity did not have much effect, unloading velocity did. Therefore the stiffness curves for palpations at various velocities should be studied so that the proper hysteresis amounts can be implemented.

9.2 Future Work

Continuation of the work begun here splits off into two directions. First the current data set can be used to engineer a device capable of simulating a realistic, healthy

abdomen. Second, the techniques developed to collect the current data set can be augmented and used to collect more data for advancing later versions of the simulator.

9.2.1 Building a Patient Simulator

With the stiffness characteristics of the abdomen understood, and the mechanics of a basic diagnostic palpation measured, a prototype simulator can now be built. The guidelines set out in the conclusions section above should be followed.

The greatest challenge for the engineers responsible for designing the simulator will be discovering the materials and methods necessary to achieve an abdominal stiffness that not only mimics the curves presented in this report, but that is variable so that different medical states and body types can be simulated. This is a more daunting task than it initially appears to be as stiffness changes will need to be localized to discrete areas of the abdomen.

Many potential solutions for a variable stiffness abdomen have already been presented by the Carleton University Patient Simulator Team. Examples include the use of electro-rheological fluids, electro-active polymers, linear actuators and stepper motors.

Once the simulator has been built, its efficacy can be tested by having the same doctor palpate the simulator who palpated the volunteers. This will provide a new data-set for comparison, and also verbal feedback from the doctor.

Eventually the simulator could be taken beyond just palpation, and incorporate the other three abdominal examination techniques: inspection, auscultation and percussion. Perhaps even balottement (moving an organ into place so that it can be palpated more easily) could be included. Inspection would most likely be the first and easiest technique to include in the simulation arsenal, starting with realistic skin colour, temperature and texture, as discussed in Chapter 2.

9.2.2 Gathering Additional Data

The data collected for this report provide an excellent start for the simulator team, but there is much more to be done. So far, data collection has been limited to the lower right quadrant of healthy male abdomens. The next step would be to collect stiffness data from the other three abdominal quadrants and then to gather data from healthy female volunteers. The stiffness in each of the four quadrants will most likely differ due to differences in the underlying anatomy. Once the stiffness for all four healthy quadrants has been analyzed, the next logical step would be to gather data from volunteers suffering from various medical conditions. It is also recommended that palpation data be collected from more than one doctor to determine if there are any differences in their palpation techniques.

Because this experiment was limited to defining the apparent stiffness of the abdomen, it may be desirable at some point to determine the real stiffness. This could be accomplished by redesigning the experiment so that it is optimized for parameter identification. This would help determine exactly what influence velocity and displacement have on the tissue's response to loading.

It should be noted that palpation is not a procedure limited to the abdomen. It is used on many areas of the body as a diagnostic technique. Eventually palpation data could be taken from other parts of the body and new simulators for that area could be built.

In conclusion, this report has captured sufficient information to enable construction of a prototype abdominal sign simulator capable of accurately simulating the healthy state of a male abdomen. The validity of the testing methodology has been demonstrated and could be useful for the analysis of other body types and medical conditions in future experiments.

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Appendix A

Matlab Code

A.1 Repeatability Analysis

```
filename: rep_analysis.m
%this m-file uses voltage data from LabView to calculate the mean, standard
%deviation, and repeatability of load applications.

clear load
for filecount = (1:3)
    clear data threshold x k;

    switch (filecount)
        case 1
            data=load('file1.lvm');
        case 2
            data=load('file2.lvm');
        case 3
            data=load('file3.lvm');
    end %switch

    plot(data(:,1),data(:,4))
    threshold=input('Threshold value: '); %the threshold value is inputted
    %by the user sets the voltage above with the program is to assume
    %that a load has been applied

    i=1; %increments rows in data

    k=0;
```

```

while (i+10) <= length(data)
    if (data(i,4) >= threshold) && (data((i+5),4) >= threshold)
        k=k+1; %increments columns of x
        j=1; %increments rows in x which stores the load applications
        i=i+5; %skips ahead 5 data points to neglect the values leading
            %up to the measurement.
        while (data((i+2),4)>threshold) && ((i+10)<length(data))
            %writes the data points to array x while the data points
            %remain above the threshold value.
            %The i+2 is there to eliminate the datapoints that occur
            %after the load has been removed, but that are still above
            %the threshold.
            %the i+10 accounts for data sets that end while the data
            %points are still above the threshold.
            x(j,k)=data(i,4);
            j=j+1;
            i=i+1;
        end %while
    else i=i+1;
    end %if
end %while

```

%The next block of code runs through each column of array x storing each value so that the mean value can be calculated. This code is only necessary so that the zero's in the array (due to differnt loading time intervals) can be eliminated from the mean and standard deviation calculation.

```

clear temparray means std_dev;

[w,k]=size(x);
for m=(1:k)
    i=1;
    while (x(i,m)~= 0) && (i < w)
        temparray(i,1)=x(i,m);
        i=i+1;
    end
    means(m,1)=mean(temparray); %stores the mean value for each load
        %application
    std_dev(m,1)=std(temparray); %stores the standard deviation for each load
        %application
    clear temparray;
end

```

```

overall_std_dev=std(means); %calculates the overall standard deviation
overall_mean=mean(means); %calculates the overall mean
repeatability=overall_std_dev/overall_mean;

output(filecount,1)=overall_mean;
output(filecount,2)=overall_std_dev;
output(filecount,3)=repeatability;
end
output

```

A.2 Sensor Calibration

```

filename: calibration.m
%This program requires a LabView file with voltage data collected from all
%4 sensors at once.

%The code assumes that the each load was applied to the sensor 5 times.

clear all
global voltage voltage_low voltage_high coef_high coef_low

data=load ('calibration_file.lvm');

%-----
%The first section of this file scans through the data and calculates an
%average voltage for each calibration load.

load = [0; .8826; 2.2261; 3.0891; 5.4329; 7.4531; 9.7968; 12.1112;...
        14.4550; 16.4752; 20.3488]; %these are the forces used
        %for calibration

k=0;%columns for avgload

for column=2:2:8
    plot(data(:,1),data(:,column))
    threshold=input('Threshold value: ');
    count=0; %keeps track of the loop number
    i=1; %increments rows in data
    while (i+20) <= length(data)
        count=count+1;
        loadappcount = 0;
    end
end

```

```

i=i+1;

while (loadappcount < 5 && (i+20) <= length(data))
    if (data(i,column) >= threshold) && (data((i+5),column) >= ...
        threshold)
        loadappcount = loadappcount+1;
        i=i+5; %skips ahead 5 data points to neglect the values
            %leading up to the measurement.
        x=1; %resets row for datapoint
        clear datapoint;
        while (data((i+2),column) >= threshold)
            %writes the data points to array x while the data
            %points remain above the threshold value.
            %the i+2 is there to eliminate the datapoints that
            %occur after the load has been removed, but that
            %are still above the threshold.
            datapoint(x,1)=data(i,column);
            i=i+1;
            x=x+1;
        end %while
        singleloadapps(loadappcount,count)=mean(datapoint);
        i=i+6;% skips the last points from the previous load app
    else i=i+1;
    end %if

end

end %while
k=k+1; %incrments columns for avgload
avgload(:,k)=mean(singleloadapps);

voltage(1,k)= data(10,column); %inserts the zero load output

for i=1:10
    voltage((i+1),k)=(avgload(i,k));
end

end

%--END load averaging section----
%This next block of code determines two equations to fit the two different
%sections of the voltage vs. force data.

%First the voltages and loads to either the high or low calibration curve.

```

```

%The high and low voltages and loads are overlapped to help smooth out the
%transition between the two calibration curves

for i=1:4 %this increments through the columns of "voltage"
    voltage_low(:,i) = voltage(1:1:7,i);
    voltage_high(:,i) = voltage(5:1:11,i);
end

load_low = load(1:1:7,1); %specifies which loads fall in the range of the
    %lower curve
load_high = load(5:1:11,1); %specifies which loads fall in the range of the
    %higher curve

lower=0:.1:4; %used to display the output from polyfit (polyval)
upper=3:.1:5;

for i=1:4
    %fits the lower curve
    coef_low(i,:)=polyfit(voltage_low(:,i),load_low,3);
    output_lower(i,:)=polyval(coef_low(i,:),lower);

    %fits the higher curve
    coef_high(i,:)=polyfit(voltage_high(:,i),load_high,3);
    output_high(i,:)=polyval(coef_high(i,:),upper);
end

%transposes the matrices
output_lower = output_lower';
output_high = output_high';

for i=1:4
    subplot(2,2,i)
    plot(load,voltage(:,i),'*',output_lower(:,i),lower,output_high(:,i)...
        ,upper);
end

%plots the resulting calibration curves for all four sensors
figure(2)
plot(load,voltage(:,4),'*',output_lower(:,4),lower,output_high(:,4),upper);
title('Sensor Calibration')
xlabel('Force (N)')
ylabel('Voltage (V)')

```


A.3 Interpolation

filename: interpolate.m

%This script takes the displacement data from the polaris system which is
%generally sampled at around 25Hz, and interpolates it up to 30Hz so that
%it can be easily synced with the force data.

%Load displacement data, which will be using frames (each frame is 1/60th
%of a second.

```
clear displacement_data load
displacement_data=load('displacement.txt');
```

%Interpolate the data so that we have the same number of data
%points as we would if we were collecting at 30Hz

```
frames=displacement_data(:,1);
x_points = displacement_data(:,2); %x-position
y_points = displacement_data(:,3); %y-position
z_points = displacement_data(:,4); %z-position
```

%Each frame is 1/60th a second. Therefore we are interested in the value of
%every second frame (1/30th of a second). The variable x provides the
%correct number of data points such that a position value can be
%obtained for every 30th of a second

```
x = frames(1,1):2:displacement_data(length(displacement_data),1);
```

%interpolate the data

```
interpolated_x=interp1(frames,x_points,x,'spline');
interpolated_y=interp1(frames,y_points,x,'spline');
interpolated_z=interp1(frames,z_points,x,'spline');
```

%transpose data

```
interpolated=[interpolated_x' interpolated_y' interpolated_z'];
```

A.4 Coordinate Conversion

filename: coord_conversion_new.m

%Transforms data from camera (GRS) to local abdomen (LRS) coordinates

%p1,p2 and p3 are points measured by the polaris system which designate the
%abdominal plane.

```
p1=[179;125.5;-1279]; %p1 is also the origin of the LRS
```

```

p2=[200.5;95.4;-1289.3];
p3=[146.5;108.3;-1318.3];
palp_point=[171.8;107.6;-1294.3];

%Algebra used to generate the rotation matrix
v1=p2-p1; %vector in plane
v2=p3-p1; %vector in plane
n=cross(v1,v2); %normal to plane
v3=cross(v1,n); %3rd vector in plane, but orthonormal to v1 and n

%make orthogonal vectors unit vectors
unitx = v1/norm(v1);
unity = v3/norm(v3);
unitz = n/norm(n);

rot_matrix = [unitx unity unitz]; %rotation matrix

%Transformation equation:  $R_j = R_i + [\text{rot\_matrix}] * r_{ij}$ 

Ri = p1; %origin or LRS in GRS coords

for i = 1:1:length(interpolated)
    Rj = [interpolated(i,1) ; interpolated(i,2) ; ...
          interpolated(i,3)];
    rij(:,i)=inv(rot_matrix)*(Rj-Ri);
end

coords = transpose(rij);

figure(1)
%X-axis
subplot(3,1,1);
plot(coords(:,1))
title('X-Axis')
xlabel('Time (s)')
ylabel('Displacement (mm)')

%Y-axis
subplot(3,1,2);
plot(coords(:,2))
title('Y-Axis')
xlabel('Time (s)')
ylabel('Displacement (mm)')

```

```
%Z-axis
subplot(3,1,3);
plot(coords(:,3))
title('Z-Axis')
xlabel('Time (s)')
ylabel('Displacement (mm)')
```

A.5 Force Conversion

filename: VoltageToForce.m

%Uses the data from LabView collected from the 4 flexiforce sensors. The %script skips the 1,3,5,7th columns because they all contain the same time %stamps. The script takes the voltage values at each time and converts them %into forces. the forces are then ouputed in an array.

%CALIBRATE.M must be run prior to running this script

```
clear force_data load;
force_data=load('voltages.lvm'); %file to be converted
```

```
forcecolumn=0;
for column = (2:2:8)
    row=1;
    forcecolumn=forcecolumn+1;
    while row <= length(force_data)
        if column == 2; %sensor 1
            if force_data(row,column) > voltage_low(6,1) %this is the
                %middle point where the two calibration curves overlap
                coeffs = coef_high(1,:);
            else
                coeffs = coef_low(1,:);
            end
        end
        if column == 4; %sensor 2
            if force_data(row,column) > voltage_low(6,1)
                coeffs = coef_high(2,:);
            else
                coeffs = coef_low(2,:);
            end
        end
        row=row+1;
    end
end
```

```

        if column == 6; %sensor 3
            if force_data(row,column) > voltage_low(6,1)
                coeffs = coef_high(3,:);
            else
                coeffs = coef_low(3,:);
            end
        end
        if column == 8; %sensor 4
            if force_data(row,column) > voltage_low(6,1)
                coeffs = coef_high(4,:);
            else
                coeffs = coef_low(4,:);
            end
        end
        force(row,forcecolumn)=polyval(coeffs,force_data(row,column));
        row=row+1;
    end
end

%-----
%Force Summation

for i=1:1:length(force)
    total_force(i,1) = sum(force(i,:),2); %sums the rows for the force
    %matrix
end

```

A.6 Velocity Calculation

filename: velocity.m

%Calculates finger-tip velocity in the z-direction

```

zvelocity(1,1)=0;
for i=2:1:length(coords)
    zvelocity(i,1) = (coords(i,3)-coords(i-1,3))/(1/30);
end

```

A.7 Combining Data

filename: Combine.m

```

%combines all data for easy export

for i = 1:1:length(coords)
    combined_data(i,1)=coords(i,1); %x-axis displacement
    combined_data(i,2)=coords(i,2); %y-axis displacement
    combined_data(i,3)=coords(i,3); %z-axis displacement
    combined_data(i,4)=zvelocity(i,1); %z direction velocity
end

for i = 1:1:length(force)
    combined_data(i,5)=force(i,1); %force at position 1
    combined_data(i,6)=force(i,2); %force at position 2
    combined_data(i,7)=force(i,3); %force at position 3
    combined_data(i,8)=force(i,4); %force at position 4
    combined_data(i,9)=total_force(i,1); %total force
end

```

A.8 Data Synchronization

filename: sync.m

%This program automatically syncs up the force and displacement data

%The doctor must have his finger resting on the tapping surface prior to
 %starting the data collection. The less movement prior to tapping the
 %better.

%Once the spike in velocity associated with the tap is found, we look for
 %the next value which is closest to zero and assume that the maximum force
 %applied during the tap occurs at that point.

clear synced velocity summed_force

%First scan velocity and find the spikes that indicate tapping

```

velocity=combined_data(:,4);
row=1;
velocity_found = 0;
force_found = 0;

```

```

while velocity_found == 0;
    if velocity(row+1,1) > (velocity(row,1)+100)

```

```

        %the beginning of the tapping motion has been identified
        while velocity(row+1,1) > 0
            row = row + 1;
        end
        % We've now found the velocity at which the maximum force during
        % the tap occurs
        velocity_row = row;
        velocity_found = 1;
    else row = row + 1;
    end
end

%Next we must locate the maximum force that occurs during the firs tap
row = 1;
summed_force=combined_data(:,9);

tapping_threshold = input('Tapping Threshold: '); %this is the value which
    %all taps surpass

while force_found == 0;
    if summed_force(row,1) >= tapping_threshold
        while summed_force(row+1,1) > summed_force(row,1)
            row = row + 1;
        end
        %We've now found the max force during the tap
        force_found = 1;
        force_row = row;
    else row = row + 1;
    end
end

%Now that the two points we need to sync up have been found, we will output
%a new matrix with the aligned data

for i=velocity_row:1:length(velocity) % makes a new matrix and adds in the
    %displacement and velocity data
    synced(i-velocity_row+1,1)=combined_data(i,1);
    synced(i-velocity_row+1,2)=combined_data(i,2);
    synced(i-velocity_row+1,3)=combined_data(i,3);
    synced(i-velocity_row+1,4)=combined_data(i,4);
end

for i = force_row:1:length(summed_force) %adds in force values

```

```

    synced(i-force_row+1,5)=combined_data(i,5);
    synced(i-force_row+1,6)=combined_data(i,6);
    synced(i-force_row+1,7)=combined_data(i,7);
    synced(i-force_row+1,8)=combined_data(i,8);
    synced(i-force_row+1,9)=combined_data(i,9);
end

```

A.9 Master M-File

filename: DoIt.m
 %This M-file calls all of the other M-files required to prepare the force
 %and palpation data for analysis.

%The end result is a file called output.txt with the following columns:

```

%1 - x-axis displacement
%2 - y-axis displacement
%3 - z-axis displacement
%4 - z direction velocity
%5 - force at position 1
%6 - force at position 2
%7 - force at position 3
%8 - force at position 4
%9 - summed force

```

```

clear %clear all variables
calibrate %calibrate sensors
interpolate %up-sample position data to 30Hz
coord_conversion_new %convert position data to a local coordinate system
voltageToForce %convert voltage values to Newtons
velocity %Calculate velocity at each position
combine %Combine data
sync %Synchronize force and position data

```

```

save output.txt synced -ascii

```

A.10 Zeroing the Data

filename: zero_data.m
 %This program accepts inputs from the user to zero the data for force and
 %x, y and z displacement.

```

%The desired data is plotted so that the user can determine the required
%offset amount.
plot(synched(:,9)) %plot force

force_offset = input('Enter force offset value: ')

for i = 1:length(synched)
    synched(i,9) = synched(i,9)-force_offset;
end

plot(synched(:,1)) %plot x-displacement
xdisp_offset = input('Enter x - displacement offset value: ')

for i = 1:length(synched)
    synched(i,1) = synched(i,1)-xdisp_offset;
end

plot(synched(:,1)) %plot y-displacement
ydisp_offset = input('Enter y - displacement offset value: ')

for i = 1:length(synched)
    synched(i,2) = synched(i,2)-ydisp_offset;
end

plot(synched(:,1)) %plot z-displacement
zdisp_offset = input('Enter z - displacement offset value: ')

for i = 1:length(synched)
    synched(i,3) = synched(i,3)-zdisp_offset;
end

```

A.11 Polaris Accuracy

```
filename: polaris_accuracy.m
```

```
%Calculates the difference between the position of a piece of
%retro-reflective tape and a calibrated optical tool.
```

```
stray=load('stray.txt');
tool=load('tool.txt');
```

```
for i=1:1:length(tool)
```



```

        x_diff = tool(i,1) - stray(i,1);
        y_diff = tool(i,2) - stray(i,2);
        z_diff = tool(i,3) - stray(i,3);
    end

    i=1:1:length(tool);

    figure(1)
    plot(i,x_diff)

    figure(2)
    plot(i,y_diff)

    figure(3)
    plot(i,z_diff)

    mean_x=mean(x_diff);
    mean_y=mean(y_diff);
    mean_z=mean(z_diff);

    std_x=std(x_diff)
    std_y=std(y_diff)
    std_z=std(z_diff)

```

A.12 Data Analysis

filename: autoplot.m

%Generates plots for data analysis. Must be run seperately for each
%volunteer.

```
clear all
```

```
data = load('output1.txt'); %load data file
```

```

x_disp = data(:,1);
y_disp = data(:,2);
z_disp = data(:,3);
total_force = data(:,9);
velocity = data(:,4);

```

```
x=1:1:length(data);
```

```

plot(x,z_disp,x,total_force)

%Enter the point at which the tapping has ended and palpation is about to
%begin

start_point =input('enter start point: ')
end_point =input('enter end point: ')

push_count = 0;
k = 0;
max_force = 0;
max_disp = 0;

%locates and singles out each palpation event (push into abdomen)
for i=start_point:1:end_point
    if (z_disp(i,1) > z_disp(i+1,1)) && (z_disp(i,1) > z_disp(i-1,1))...
        && (z_disp(i,1) > 30) %displacement peak found.
        peak_point = i;
        backup = peak_point -10;
        forward = peak_point +10;
        vel = velocity(i,1);
        push_count = push_count + 1;
        while vel > 0 %backs up from peak and finds the start of the push
            vel = velocity(backup,1);
            backup = backup - 1;
        end
        while vel < 0 %advances to locate the end of the push
            vel = velocity(forward,1);
            forward = forward + 1;
        end
        j = 1;
        k = k + 1;
        for o = backup:1:forward % stores data for each push in a
            %separate column
            %-- x-displacement
            single_xdisp(j,k) = x_disp(o,1);
            %-- y-displacement
            single_ydisp(j,k) = y_disp(o,1);
            %-- z-displacement
            single_zdisp(j,k) = z_disp(o,1);
            %-- force
            single_force(j,k) = total_force(o,1);
            %-- velocity

```

```

        single_velocity(j,k) = velocity(o,1);
        j = j+1;
    end
end
end

%Find max force
for i=start_point:1:end_point
    if total_force(i,1) > max_force
        max_force = total_force(i,1);
    end
end
max_force

%Find max displacement
for i=start_point:1:end_point
    if z_disp(i,1) > max_disp
        max_disp = z_disp(i,1);
    end
end
max_disp

%Plots-----

figure(1) % force vs. displacement
title('Stiffness')
xlabel('Z-Displacement (mm)')
ylabel('Force (N)')
hold on
for i=1:1:1%push_count
    clear temp
    row = 1;
    while (single_zdisp(row,i) ~= 0) && (row < (length(single_zdisp)))
        %a temporary storage array (temp) is necessary to avoid including
        %the zeroes into the shorter columns in the plots.
        temp(row,1)=single_zdisp(row,i);
        temp(row,2)=single_force(row,i);
        row = row+1;
    end
    plot(temp(:,1),temp(:,2))
end

```

```

figure(2) % z-displacment vs. z-velocity (phase)
title('z-displacment vs. z-velocity (Phase Plane)')
xlabel('Z-Displacement (mm)')
ylabel('Velocity (mm/s)')
hold on
for i=1:1:push_count
    clear temp
    row = 1;
    while (single_zdisp(row,i) ~= 0) && (row < (length(single_zdisp)))
        temp(row,1)=single_zdisp(row,i);
        temp(row,2)=single_velocity(row,i);
        row = row+1;
    end
    plot(temp(:,1),temp(:,2))
end

figure(3) %Force, displacement and velocity vs. time
subplot(3,1,1), plot(total_force(start_point:end_point,1))
xlabel('Time (s)')
ylabel('Force (N)')
subplot(3,1,2), plot(z_disp(start_point:end_point,1))
xlabel('Time (s)')
ylabel('Z-displacement (mm)')
subplot(3,1,3), plot(velocity(start_point:end_point,1))
xlabel('Time (s)')
ylabel('Velocity')

%Invert the z-coordinate values so that this 3d chart is more intuitive.
inverted_zdisp(:,1) = single_zdisp(:,1).*-1
figure(5) %3-Dimensional motion
plot3(single_xdisp(:,1),single_ydisp(:,1),inverted_zdisp(:,1))
grid on;
axis image;
title('3-Dimensional Motion')
xlabel('X')
ylabel('Y')
zlabel('Z')

%Plot the stiffness for each push on one plot
figure(7)
hold on
plot(single_zdisp(:,1),single_force(:,1))

```

```

plot(single_zdisp(:,2),single_force(:,2))
plot(single_zdisp(:,3),single_force(:,3))
plot(single_zdisp(:,4),single_force(:,4))
plot(single_zdisp(:,5),single_force(:,5))
plot(single_zdisp(:,6),single_force(:,6))
plot(single_zdisp(:,7),single_force(:,7))
plot(single_zdisp(:,8),single_force(:,8))
plot(single_zdisp(:,9),single_force(:,9))
plot(single_zdisp(:,10),single_force(:,10))
title('Stiffness')
xlabel('Z-Displacement (mm)')
ylabel('Force (N)')

```

%calculate and plot an average stiffness curve for one volunteer. The average is

```

row=0;
for point=5:5:50
    row=row+1;
    col=0;
    for i=start_point:1:end_point
        if (z_disp(i,1) <= (point + 2.5)) && (z_disp(i,1) >= (point - 2.5))
            col = col + 1;
            disps(row,col)=z_disp(i,1);
            forces(row,col)=total_force(i,1);
        end
    end
end

forces = forces';
disps = disps';
for cols=2:1:10;
    clear tempdisp tempforce
    row = 1;
    while (disps(row,cols) ~= 0) && ( row < (length(disps)))
        tempdisp(row,1)=disps(row,cols);
        tempforce(row,1)=forces(row,cols);
        row=row+1;
    end
    average_disp(cols,1)=mean(tempdisp);
    average_force(cols,1)=mean(tempforce);
end

```

```
average_disp(1,1)=0;  
average_force(1,1)=0;  
  
figure(9)  
plot(average_disp,average_force)  
hold on  
plot(z_disp,total_force)  
title('Average Stiffness - Volunteer #3')  
ylabel('Force (N)')  
xlabel('Displacement (mm)')
```


Appendix B

Test Apparatus Design

B.1 Locator

Figure B.1: A simple bracket designed to repeatedly locate the loading apparatus with respect to the sensor for testing purposes.

B.2 Base

Figure B.2: Base of the re-designed Flexiforce testing apparatus. The groove along the top of the base holds the sensor in position.

B.3 Bracket

Figure B.3: Vertical component of the bracket to which the point-loading cone is attached, and from which the load is hung.

Figure B.4: Horizontal component of the bracket to which the point-loading cone is attached, and from which the load is hung.

B.4 Surfaces

Figure B.5: Flat disc which rests within the base of the test apparatus.

Figure B.6: Curved disc which rests within the base of the test apparatus - designed to simulate the curvature of a finger-tip.

B.5 Cylinder and Point

Figure B.7: The cylinder which transfers the load to the sensor. The indent at the top of the cylinder accommodates the pointed part of the bracket.

Figure B.8: Point loading mechanism of the test apparatus. The point bolts to the bracket from which the load is suspended.

Appendix C

3D Motion Plots

Figure C.1: 3D motion of the doctor's fingertip during the palpation of volunteer #3.

Figure C.2: 3D motion of the doctor's fingertip during the palpation of volunteer #4.

Figure C.3: 3D motion of the doctor's fingertip during the palpation of volunteer #3.

Figure C.4: 3D motion of the doctor's fingertip during the palpation of volunteer #5.

Figure C.5: 3D motion of the doctor's fingertip during the palpation of volunteer #6.

Figure C.6: 3D motion of the doctor's fingertip during the palpation of volunteer #7.

Figure C.7: 3D motion of the doctor's fingertip during the palpation of volunteer #8.

Appendix D

Plots of Force, Displacement, and Velocity at the Doctor's Fingertips versus Time

Figure D.1: Force, displacement and velocity aligned with time for volunteer #1.

Figure D.2: Force, displacement and velocity aligned with time for volunteer #2.

Figure D.3: Force, displacement and velocity aligned with time for volunteer #3.

Figure D.4: Force, displacement and velocity aligned with time for volunteer #4.

Figure D.5: Force, displacement and velocity aligned with time for volunteer #5.

Figure D.6: Force, displacement and velocity aligned with time for volunteer #6.

Figure D.7: Force, displacement and velocity aligned with time for volunteer #7.

Figure D.8: Force, displacement and velocity aligned with time for volunteer #8.

Appendix E

Ethics Approval Notice

Children's Hospital of Eastern Ontario

Centre hospitalier pour enfants de l'est de l'Ontario

CHEO RESEARCH ETHICS BOARD APPROVAL – EXPEDITED REVIEW

Principal Investigator: Dr. Juan Bass

Proposal Number: #06/49X

Protocol Title: Abdominal Signs Simulator – Applied Force during Abdominal Palpation

Department or PSU: Surgery

Approval date: July 24, 2007

Valid Until: July 23, 2008

Documents reviewed and approved: Research protocol submitted July 14, 2006

This is to notify you that the Children's Hospital of Eastern Ontario Research Ethics Board has granted approval to the above named research study on the date noted above. Your project was reviewed under the expedited stream, which is reserved for projects that involve no more than minimal risk to human subjects.

During the course of the research, no deviations from, or changes to, the protocol or consent form may be initiated without prior written approval from the REB. Further, investigators are asked to report the following to the REB:

- Proposed changes to the study procedures (including the recruitment strategy, inclusion criteria, etc.);
- Concerns or issues that arise in conducting the research;
- Changes to the consent documents and advertisement notices;
- Changes to the investigators who assume responsibility for the study; &
- An annual report.

Wishing you success in your project.

Regards,

Dr. Carole Gentile, C.Psych.
Chair, Research Ethics Board

CG smch 24/07/07
c.c. Pat Brazeau, Manager, CHEO REB

This is an official document. Please retain the original for your file

version 11/2003