EFFECT OF VELOCITY CONTROL ON KINESTHETIC LUNG TUMOUR LOCALIZATION

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ABSTRACT

Restricted access during minimally invasive surgery precludes manual palpation, making the localization of lung tumours challenging. This paper investigates the force sensing performance that would allow an instrumented kinesthetic probe to localize tumours based on stiffness variations of the lung parenchyma. Agar injected into *ex vivo* porcine lungs produced a model approximating commonly encountered tumours. Using both constant and variable velocity approaches, force-deformation data were collected from multiple sites at various palpation depths, before and after the tumours were injected. Analysis showed an increase in force after the tumours were injected, ranging from 0.07 to 0.16 N at 7 mm. A constant palpation velocity minimized exponential stress decay at constant depths, facilitating easier comparisons between measurements. A sensing range of 0 to 2 N, with 0.01 N resolution should allow a kinesthetic palpation probe to resolve local tissue stiffness changes that suggest an underlying tumour.

Index Terms— Tumour localization, kinesthetic feedback, palpation, minimally invasive surgery

1. INTRODUCTION

Minimally Invasive Surgery (MIS) is a surgical method performed through 5–12 mm incisions, offering numerous advantages over conventional open surgery: lower infection rates, reduced tissue trauma and post-operative pain [1], reduced recovery time and a faster return to work [2]. Unfortunately, the surgeon faces challenges such as reduced dexterity, friction and reversal of instrument motion caused by the port of entry, and the inability to directly manipulate tissue [3].

Tactile perception is an invaluable tool for many surgical procedures since it can provide rich information on the mechanical properties of the tissue being manipulated. Since a malignant tumour will typically be stiffer than the surrounding parenchyma [2], a surgeon can usually localize a tumour via direct palpation when performing open surgery. The common practice is to use standard MIS instruments to probe the surface of the lung and use limited visual or tactile cues to determine the position of the tumour. If available, MIS ultrasound can be used. However, due to the poor image quality and artifacts caused by residual air in the lung, it is not usually possible to find tumours smaller than 1 cm in diameter. If the tumour cannot be located, the surgeon must increase the size of the incision and spread the ribs to allow finger access for direct palpation.

To address these limitations, considerable research effort has sought to perceive haptic cues (i.e., "sense of touch") arising from tissue-instrument interaction to assist the surgeon. Haptic information can be considered in two distinct modes: kinesthetic and tactile information [4]. Tactile information includes the sensation of surface textures or distributed pressures acting across the contacting surface. The use of tactile sensors has been the subject of promising research [5, 6, 7]. In contrast to tactile information, which requires a dense cluster of sensors, kinesthetic information relates to the movement and bulk forces acting at the joints of an arm (human or mechanical) and at the point of contact. Kinesthetic information may be used to assess the contour and stiffness of an object and may be acquired using a simple force/torque sensor. Numerous researchers have proposed using kinesthetic feedback to measure force and displacement during an MIS procedure. In [8], a sterilizable 6 degree-of-freedom (DoF) force/torque sensor based on a strain gauge instrumented Stewart's platform is proposed. A 3 DoF force sensor based on fiber-optic sensing, is presented in [9]. Various instrumented graspers have been proposed in [2, 10, 11]. In [12] tissue interaction was measured using a number of strain gauges and a single-axis load cell integrated into a custom endoscopic instrument.

To the best of our knowledge, the use of kinesthetic feedback from direct probing using a sensorized instrument, rather than grasping, has not been applied to the task of lung tumour localization. It is hypothesized that the force measured will be higher when probing tissue with an underlying tumour than when probing intact tissue at the same depth. The goal of this paper is threefold: 1) to determine with statistical confidence that measurements from indentation testing can indicate the presence of a tumour; 2) to determine the combination of approach velocity and palpation depth that maximizes the measured force increase due to a tumour, while minimizing the number of false negative results; and 3) to determine the sensing range and resolution that is required to realize the approach in actual MIS. These results will be useful in providing target design specifications for systems which aim to perform MIS tumour localization.

2. METHODS

In order to experimentally verify that kinesthetic feedback alone could indicate the presence of a tumour in lung parenchyma, an accurate model was required. *Ex vivo* porcine lungs were used as a biological model for human lung. Several preliminary tests were

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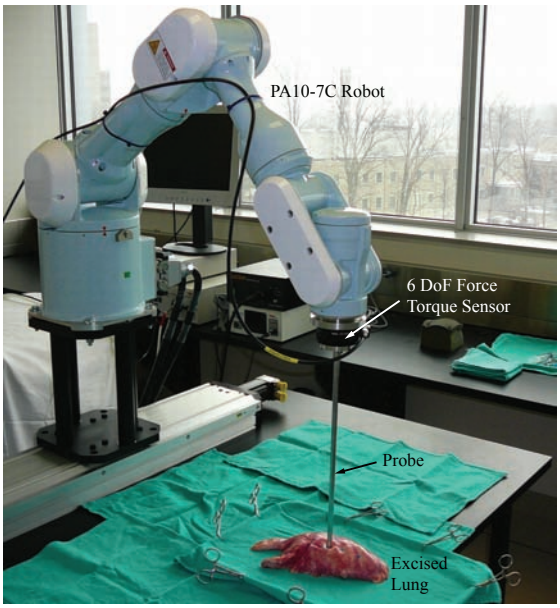


Fig. 1. The Mitsubishi PA10-7C palpating an *ex vivo* porcine lung.

undertaken to determine the most realistic method of simulating tumours [13]. It was discovered that any method which required a major incision in the tissue changed the force-deformation behaviour of the otherwise intact lung. Therefore, injecting material provided the most realistic alternative. Sigma Gelrite Gellan Gum (agar) was prepared in a ratio of 225 mL water to 7.5 g agar. 1.5 mL of the boiled mixture was injected into cold lung to form approximately 1 cm diameter tumours; the extra volume was lost through the puncture prior to solidification. The agar tumours were qualitatively verified by an experienced thoracic surgeon to feel like a typical tumour.

The kinesthetic probe was realized using an aluminum rod approximately 50 cm long with a diameter of 9 mm and a hemispherical distal tip [13]. A Gamma Force/Torque sensor from ATI Industrial Automation, Inc. was placed in series between the proximal end of the probe and the mounting plate of a Mitsubishi PA10-7C robot, Fig. 1. The Gamma sensor was factory calibrated for a 100 N measurement range in the z-direction (axial direction of the probe), and 32 N in the x- and y-directions. The resolution is 0.003 N in the z-direction and 0.002 N in the x- and y-directions. Force data for the three orthogonal vectors was used to calculate the magnitude of the resultant force vector, from which all of the analysis was conducted.

Following the recommendations in [14], stiffness changes were detected by indenting the tissue to a specified depth from the surface and recording the force. The probe was moved in an axial direction towards the surface of the lung until a threshold of 0.04 N was detected in a repeatable manner. Once in contact with the tissue surface, continuous force measurements in three orthogonal directions were initiated. The probe was then advanced into the lung tissue to a pre-programmed depth: either 5, 7 or 9 mm, after which it was held stationary, while sampling at 250 Hz for 10 s. The probe was then retracted and moved to a point on the lung 30 mm away from any previous testing site and the palpation process was repeated. Preplanning of the testing grid ensured that only the lower lobe was sampled, and that the main bronchial branches were avoided. After the testing was completed for each lung, agar was injected into each palpation site, to approximately the mid-thickness of the lung, and the palpation tests were repeated. To avoid pre-conditioning [15], only one sample was collected for the intact tissue at each test site, followed by a single sample after the agar had been injected.

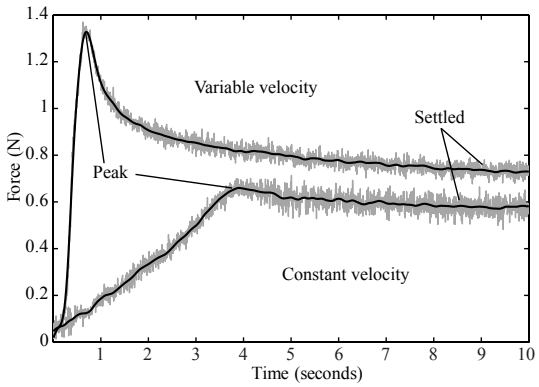


Fig. 2. Raw and filtered data for 9 mm palpations.

Since the indentation velocity will affect the force-deformation behavior of a visco-elastic material, the above mentioned experiments were performed with two variations of velocity control. In the variable-velocity approach, the maximum velocity of the robot was 40 mm/s; however, this velocity was not achieved due to the short translations required for palpation. With this approach, the peak velocity increases with palpation depth. For the constant-velocity approach, the maximum velocity was set to 2 mm/s, allowing direct performance comparisons for different depths.

The raw data were post-processed in MATLAB® using the *filtfilt* function and a second order Butterworth filter with a cutoff frequency of approximately 2.9 Hz. Fig. 2 shows the raw and filtered data from two palpation tests at 9 mm, using both the variable- and constant-velocity approaches. Since lung tissue tends to exhibit a well known decaying stress with constant strain, as can be seen in Fig. 2, it is important that any comparisons between measurements be made at the same point on the curve. Two points were considered in this study: the peak force measured, and the force after the response had settled, taken at a constant number of data points from the peak.

In order to isolate the changes that were caused by the presence of the artificial tumour from changes caused by tissue autolysis and other environmental factors, tissue controls were incorporated into the testing. When the lungs were excised, the right and left lung remained connected by the intact portion of the bronchi. In all cases, the left lung had tumours injected, while the right lung was used as the control. Since the lungs came from the same animal and remained connected throughout the experiments, the effect of factors such as time *post mortem*, temperature cycles, hydration, etc., could be isolated from the effect of the artificial tumour.

3. RESULTS

For the purpose of defining an appropriate sensor range for a kinesthetic lung tumour localization system, it is necessary to look at the maximum forces occurring in each of the experimental approaches. This information may seen in Fig. 3.

Each of the experiments was performed in a paired manner, assessing the response of the lung tissue before and after a tumour was introduced. For both the artificial tumour tests and control tests, the change in force feedback given by $\Delta F = F_f - F_i$ was determined for each pair of samples, where F_f is the measurement recorded with the tumour present, or the second test of the control, and F_i is the force measured during the initial palpation of either the test or control lung. These ΔF values were grouped into sets according to depth, velocity scheme and peak or settled values.

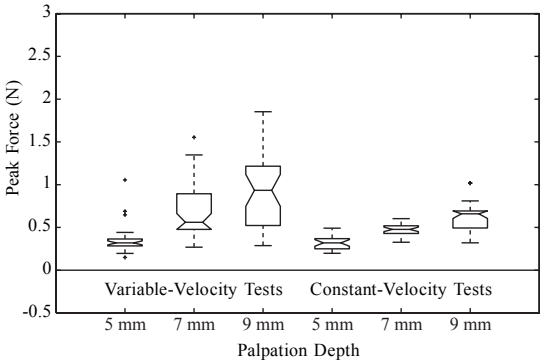


Fig. 3. Peak force measurements with a tumour present.

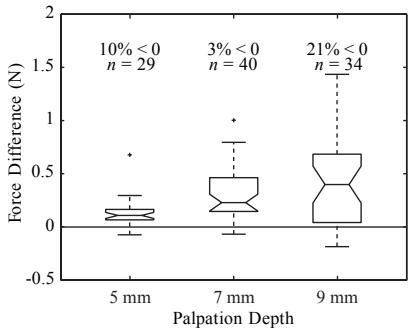
Before the results from the artificial tumour tests can be interpreted, any changes that can be attributed to factors other than the presence of a tumour must be accounted for. For each test-control pair, the value of the upper quartile measurement of the control was used to offset the tumour test data to account for these changes. The resulting tumour test data, corrected for the controls, are shown for variable-velocity peak and settled and constant-velocity peak and settled sets in Fig. 4. Within each subfigure, the percentage of *negative* differences (i.e., the second force measurement was less than the initial measurement) and the total sample size are indicated for each sample set. The upper and lower bounds of the box plot represent the first and third quartile of the measurements. The line through the box represents the median, while the notch represents the range of the 95% confidence interval of the median of the samples. If the notches of two sample sets do not overlap, it can be concluded with 95% confidence that the true medians do differ. The whiskers represent the range of measurements not including outliers, which are defined as measurements deviating from the median by more than 1.5 times the interquartile range and are indicated by a '+'.

For this analysis, *p*-values were determined using the *Wilcoxon Paired-Sample Test*, and found to be less than 10^{-4} in all cases. This non-parametric test is analogous to the *paired-sample t-test*, but is applicable when assumptions of normal distribution and equal variance cannot be made [16]. All statistical analysis was performed using the MATLAB® Statistical Toolbox, and verified using SPSS.

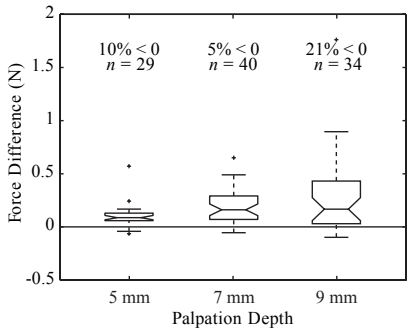
4. DISCUSSION

It is clear from the corrected test results in Fig. 4, that each experimental approach performed reasonably well. The analysis indicates that each approach produced a statistically significant change in the force-deformation behaviour in the presence of a tumour. This provides a statistical indication that kinesthetic feedback alone could be used to detect tumours within lung tissue. However, some methods appear to be consistently better than others.

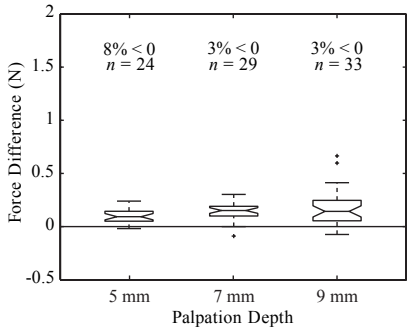
The percentage of *negative* force differences provides an indication of the likelihood of arriving at a false negative conclusion (i.e., not seeing a force *increase* in the presence of a tumour). The 7 mm palpation tests consistently produced the least number of false negatives when testing the variable-velocity approach. The 7 and 9 mm palpations both resulted in 3% negative force differences when the peak force of the constant-velocity approach was analyzed, whereas the false negatives resulting from the 7 mm palpation increased to 7% when considering the settled values. However, it should be noted that the negative value causing this change was extremely close to zero, on the order of -10^{-3}. If this value were assumed to be zero, the percentage of false negatives for the 7 mm test would again fall to



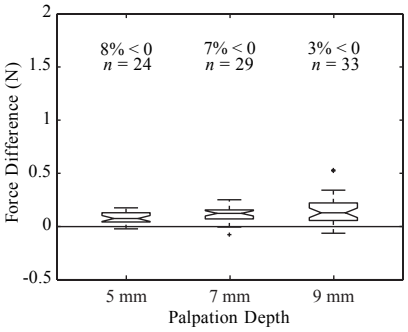
(a) Peak: variable-velocity



(b) Settled: variable-velocity



(c) Peak: constant-velocity



(d) Settled: constant-velocity

Fig. 4. ΔF of corrected palpation measurements.

3%. Furthermore, in an operative setting, multiple palpations by the surgeon may mitigate the effect of false negative results.

Finally, the 7 and 9 mm constant-velocity palpation tests produced very similar performance in terms of the median force difference measured, ranging from 0.12 to 0.13 N. While the medians were very similar, the data spread tended to be tighter for the 7 mm tests and the false negatives were a result of outlier measurements.

The greatest median ΔF overall was a result of the 9 mm variable-velocity test. However, this approach was also associated with the highest incidence of false negative results.

The median force difference for the 5 mm palpation tests was lower than the 7 mm and 9 mm tests regardless of measurement or analysis method. This indicates that while the 5 mm palpation tests produced a reasonable test statistic, it was not optimal in any of the experimental approaches. It is interesting to note that the data spread is minimal for all 5 mm tests compared to the other depths. Presumably, the shallower palpation depth would not be influenced by a rigid support medium to the same extent as deeper palpations. This is especially significant in thinner regions of the lung, where deeper palpation depths might exhibit an altered response due to the underlying rigid support surface.

If the deviations observed with differing palpation velocity are due to the visco-elastic nature of the tissue, it would follow that once the tissue had settled, there should be little difference between the response of the variable- and constant-velocity tests, at the same palpation depth. By comparing Figs. 4(b) and 4(d), it is clear that the median of the settled responses for the variable- and constant-velocity tests cannot be concluded to be different at a 95% confidence level. This suggests that any difference in the measured peak response that is attributed to a higher palpation velocity tends to diminish as the sample settles. Furthermore, comparing the constant-velocity peak and settled values indicates that reasonably accurate comparisons could be made from readings taken after different settling times have elapsed. The nearly-constant stress resulting from the constant-velocity approach shown in Fig. 2 illustrates this well. This may be advantageous when considered in a clinical setting, where maintaining a constant strain for an extended period may not be possible.

The peak values shown in Fig. 3 indicate the maximum forces that can be expected during the use of a kinesthetic instrument to locate lung tumours. With the exception of a few outliers, the peak measurements are less than 2 N and 1 N for the variable-velocity and constant-velocity tests, respectively. Thus, it would be reasonable to define the full scale range of a sensor system as 0 to 2 N.

The required resolution can be determined from the corrected ΔF values in Fig. 4. Since the 7 mm constant-velocity approach appeared to perform slightly better than the 9 mm, the values from this method, considering both the peak and settled analysis, will be considered. Considering the upper and lower quartile bounds from both the peak and settled analysis, a significant portion of the ΔF values range from 0.07 to 0.16 N. Therefore, if an MIS kinesthetic sensing system is unable to resolve force differences within this range, it would not likely be able to differentiate the majority of these underlying tumours. To avoid quantization error, the resolution of the instrument should be one order of magnitude less than this value: 0.007 N, or approximately 0.01 N.

5. CONCLUSIONS

This study used *ex vivo* porcine lung with simulated tumours to determine whether a constant- or variable-velocity palpation method would provide suitable kinesthetic feedback for the localization of a lesion. Of the palpation methods tested, the 7 mm constant-velocity tests tended to perform the best, in terms of statistical confidence in the observed force increase, limited data spread, and the number of false negative results observed. The results suggest that a full scale sensing range of 0 to 2 N and a minimum resolution of 0.01 N would be appropriate for a kinesthetic feedback instrument intended for use in lung tumour localization.

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