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Development of Tissue Surrogates for Photoelastic Strain Analysis of Needle Insertion

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ABSTRACT:

This paper focuses on the development of full-field experimental methods for validating computational models of needle insertion, and specifically the development of suitable tissue surrogate materials. Gelatine also known as “ballistic gel” is commonly used as a tissue surrogate since the modulus of elasticity matches that of tissue. Its birefringent properties also allow the visualisation of strains in polarised light. However, other characteristics of tissue are not well emulated by gelatine, for example the fibrous network of cells of tissue is not well represented by the granular microstructure of gelatine, which tears easily. A range of birefringent flexible materials were developed and calibrated for photoelastic analysis. The most suitable were then used to explore quantitatively the different strain distributions in tissue when subjected to a range of needles with different tip profiles.

KEY WORDS: Tissue surrogate, birefringent, gelatine, needle insertion

INTRODUCTION

Needle insertion is a common surgical procedure used in everything from drug administration to biopsy extraction. Many such applications would benefit from robust and flexible numerical models of the needle insertion process. For example during a biopsy procedure, the hollow needle, usually with a notch at the tip, will be guided with the aid of ultrasound to cut small tissue samples and remove them from the body. Guidance problems may arise when the targeted area is deep in the body, requiring a relatively large distance to be travelled by the needle tip. Also the tip of the needle is not symmetrical and so forces acting on the tip will be uneven, consequently resulting in bending of the needle in the body during insertion [1,2,3]. The bending can cause inaccuracy in the needle placement and consequently the needle tip may miss cancerous tissue and result in mis-diagnosis [4]. Normally doctors will perform 3 to 6 insertions each session in order to avoid such mis-diagnosis thus making the procedure time quite lengthy for the patient. Additionally if one considers small breast lesions for example, during biopsy the target could move as the needle indents and punctures the skin layer [5]. The target could penetrate deeper into the breast soft tissue, making it more difficult for the needle to reach the lesion. Lesion movements were tracked with ultrasound images and it was found that insertion force and target displacement increase by 90.2% and 275.9% respectively, when the skin thickness is increased from 0 mm to 2.5 mm. Modelling the needle insertion could improve the reliability of needle placement within a body by predicting tissue and needle deflections.

Although computational modelling of complex human tissue systems is now a real possibility due to advancements in mathematical and numerical techniques, such models need validating. Ethical issues clearly limit validation experiments on human tissue, therefore there is a need to develop tissue surrogates. Photoelastic materials have been used for many years in dental and medical applications, offering not only mechanical similarity to tissue, but also visualisation of strains. Photoelastic materials exhibit temporary birefringence, such that when loaded and viewed in polarised light, the double-refraction effect results in the observation of interference fringes which relate directly to the shear strains in the material. Birefringent materials such as epoxy resins have been used to represent stiffer materials such as bone and dentine, and gelatine, also known as “ballistic gel”, is often used as a surrogate to evaluate penetrating impacts or blast loading effects on soft tissues [6,7,8,9]. The use of photoelasticity for tissue analysis has been mainly limited to qualitative analysis since there are many issues which require consideration when developing flexible birefringent surrogate tissue materials for quantitative validation purposes. These include: transparency and birefringent sensitivity; matching the modulus of elasticity to tissue types of different stiffness; physically modelling and quantifying three dimensional systems; modelling viscoelasticity and creep present in tissue materials; representing the fibrous network of tissues’ extracellular matrix, particularly when

punctured with medical instruments; and the stability of mechanical properties of surrogate materials over time and at different temperatures.

This work focuses on the initial attempts to develop full-field experimental methods for validating computational models of needle insertion, and specifically the development of suitable tissue surrogate materials.

MECHANICAL PROPERTIES OF TISSUE

The first step of the development of a surrogate was to investigate mechanical properties of human tissue. The biopsy procedure and tools used are normally similar on all patients, but the mechanical properties of tissue vary according to gender and age group. Also, as the tissue-needle interaction is being investigated, characteristic of each layer of tissue may have to be considered. For example in a breast cancer biopsy, upon puncturing the skin, the needle may pierce layers of fat, glandular tissue and muscles.

Since with any needle insertion, the tip must first puncture the skin, the mechanical properties of skin were considered first. A study done by Agache *et al* shows that the Young's modulus, E , of human skin increases with age [10]. In the study, mechanical properties of the skin were measured on 138 persons, aged from 3 to 89 years old. It was found that elasticity and stretchability of skin deteriorates with age with an average E of 0.42MPa below 30 years and 0.85MPa above 30 years old. However, in research by Edwards and Marks[11], the Young's modulus found range from 15MPa to about 150MPa. Looking at the mean value, it peaks at about 70MPa at 11 years old and slowly declines to about 60MPa at 95 years old. The vast difference in numerical results from these two studies is probably due to the difference in experimental techniques. Agache performed in vivo testing by applying a fixed torque on test subjects and measuring the resulting angle of twist then calculating the modulus of elasticity using a set of equations. Whereas Edwards and Marks quoted the work of Vogel where in vitro testing was performed on different human skin samples. Tensile tests to failure were performed on specimens, and the corresponding stress-strain curves plotted. To compare both experiments, the in vitro test by Vogel would be considered more accurate as a test to failure interprets more data compared to a small section done by Agache *et al*. While this information is useful, the compressive and shearing behaviour of skin was not discussed. Compression, or more precisely, of the puncturing of skin would be more relevant to our study.

Considering the soft tissue layers, Saraf *et al* [12] performed experiments on four types of soft tissue: stomach, liver, heart and lung. The tissues were put under hydrostatic compression and simple shear test to get the dynamic response. Their study concluded that the tangent shearing moduli of the four tissues range from 0.008MPa to 0.34MPa. In terms of shearing, the liver tissue was found to be stiffest while the lung tissue was the softest. However, the dynamic bulk moduli vary from 150MPa to 500MPa, as the stomach being the stiffest and the lung again was the softest. In addition, it is stated that these tissues do not possess linear elastic behaviour and the results are just an approximation. Nevertheless, the study fulfilled its primary goal to determine the average mechanical properties of the four tissues.

In a study by Davis *et al* [13], the force required to insert a micro-needle into living skin was measured and it was found that the insertion force depends heavily on the sharpness and size of the needle but as the needle punctures the tissue, the fracture toughness could affect the overall strength of the tissue. Taylor *et al* [14] found that the value of fracture toughness of skin was highly variable, and depends strongly on the crack growth. They found that specimen size was important since stress and failure energy were seen to be constant for larger specimens with no dependence on crack size. Furthermore, it is of interest that soft tissues were found to be highly tolerant to defects as they could withstand cracks up to several millimetres without losing much strength.

DEVELOPMENT OF A SURROGATE TISSUE MATERIAL

The review of mechanical properties of tissue showed that the modulus is highly variable depending on its location and depth, is not linear-elastic and shows some resistance to tearing. A range of materials were tested as potential surrogates, considering: i) transparency and birefringence; ii) resistance to tearing when punctured; and iii) performance under load. The materials considered were gelatine, gelatine with additives, konjac, agar, and a range of silicone rubbers.

Gelatin- based materials are commonly used to represent soft tissues and several studies have been performed to characterise the mechanical properties of such materials [6-9]. The work by Kwon and Subhash[7] shows that 11% concentrated gelatine results in a Young's modulus of 10.9kPa. Their experiment focuses on the strain rate sensitivity of the gelatine and different responses were obtained under different loading conditions. Gelatine concentrations of 5%, 7% and 14% yielded average Young's moduli of 40kPa, 63kPa and 110kPa respectively [8]. The difference in stiffnesses between these two studies might be due to the types of raw gelatine powder used. [7] used engineering gelatine powder with a known Bloom strength of 250, while [8] uses gelatine powder for food preparation. Gelatine powder used for food preparation has lower Bloom strength and

expected to have a value of around 60 to 80. The Bloom level is a measurement of toughness, therefore, higher Bloom value gelatine are more resistive to tearing. Glycerol and sorbitol have been added to gelatine mixtures, where glycerol acts as a plasticizer which increases the molecular weight [9]. Different ratios of glycerol and sorbitol were mixed together with gelatine and it was found that an increase in the proportion of glycerol caused an increase in flexibility but the material suffers an overall loss in strength.

In our initial experiments, a mixture only consisting of gelatine powder and water was used. 20 g of gelatine powder was mixed with 200 ml water for 1:10 ratio and refrigerated (5°C) overnight. Leaving the mixture overnight allows proper blooming of the gelatine and also means the gelatine crystals have enough time to absorb liquid. The blooming process is very important as it ensures smooth texture of the finished product. After leaving it overnight, the gelatine mixture crystallized and the grains can be seen to have enlarged. The cluster of gelatine grains were then heated to about 60°C to 70°C, taking care not to let the mixture boil as this introduces bubbles. Once the mixture was fully liquefied, it was poured into the cylindrical mould and cooled to room temperature. The cooled mixture is then refrigerated overnight again before testing.

A similar procedure was used with different ratios of gelatine to water, and with additives of glycerin and/or sorbitol as indicated in Table 1. An attempt was made using a systematic process to fabricate usable specimens at highest concentration possible, but which would still solidify. Only one composition of gelatine powder, glycerin and sorbitol (4:10:10) was successful because the nature of the substances made it difficult for the gelatine powder to absorb. Hence, the blooming period was extended to 3 days for the mixtures with additives, to allow the gelatine to crystalize fully.

The konjac gel is derived from the konjac plant and is commonly known in Asia as “Konnyaku”. It was considered that fibrous microstructure of the konjac may provide superior toughness over gelatine. To fabricate, 10 g of konjac powder was progressively mixed with 800 ml of cold water while stirring gently to ensure even mixing. The mixture was then heated to 70°C, stirring the mixture until all the granules dissolved. Precautions were taken not to overheat the mixture. The mixture was poured into the mould and cooled to room temperature before setting in the refrigerator overnight. Higher concentrations of the mixture were attempted but it was found that the mixture tended to become too thick and solidified too quickly causing residual stresses.

Agar is derived from algae and is used as a culinary thickening agent, growth medium for bacteria in science labs and a dental impression material. It was supplied in a solid gel form, so was heated gently until liquid then was poured into the mould and cooled to room temperature before leaving to re-set in the refrigerator overnight.

Table 1 – Composition, photoelastic fringe constant and Young’s modulus of the gelatine specimens.

Specimens	Composition	Fringe Constant, f (N/m. fringe)	Young’s Modulus, E (kPa)
Gelatine: Water	1 : 10	36	27.3
	2 : 10	33	56.9
	3 : 10	28	84.1
Gelatine: Water: Glycerin	2 : 10 : 10	82	47.5
	3 : 10 : 10	85	75.5
	3 : 5 : 10	115	112.5
Gelatine: Glycerin: Sorbitol	4 : 10 : 10	152	120.6

Transparency and birefringence

Disc specimens of all the materials were loaded in compression in a circular polariscope. Silicon rubbers were discounted immediately as they exhibited poor photoelastic response. Figure 1 shows the birefringent response of the gelatine, konjac and agar. All materials were birefringent but the gelatine displayed the greatest sensitivity. The konjac possessed poor optical properties as the fibrous material tended to refract the light. The gelatine possessed the best optical and birefringent properties and therefore a standard disc in diametral compression calibration test [15] was performed to determine the photoelastic fringe constant, f, for the different mixtures of gelatine. The results are presented in Table 1 and show that the highest fringe constant was achieved by the mixture containing both glycerin and sorbitol.

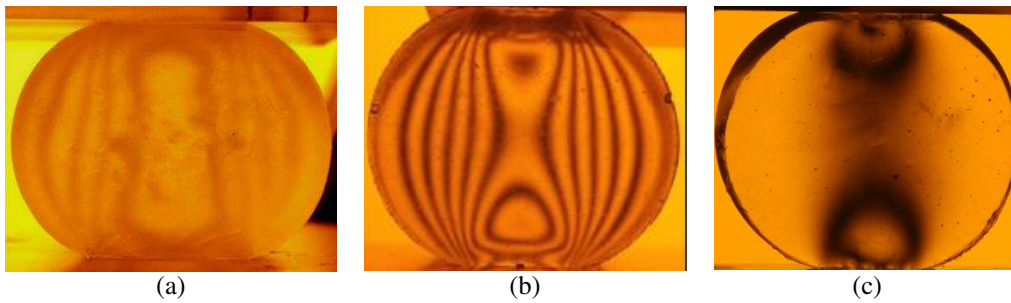


Fig. 1 Birefringence of (a) konjac gel, b) gelatine and (c) agar

Resistance to tearing when punctured

Syringes or needles come in many sizes and shapes. The size, or thickness, is marked with “gauge number”. The smaller the gauge value, the thicker the needle. For example for breast lesion localization, a needle of gauge 20 (nominal outer diameter of 0.9 mm) is normally used, which is considered fine-needle biopsy while a core-needle biopsy means using a thicker and longer hollow needle to extract tissue samples. The “needles” used in our experiments were scaled up models made from 5mm diameter steel rod. The different tip shapes of the needles are displayed in Fig. 2: Westcott, Chiba and Turner [3] [16]; and as a comparison, a symmetrically tipped cone “needle” was also manufactured.



Fig. 2 Scale models of the types of needle tip: (a) Westcott, (b) Chiba (25° bevel), (c) Turner (45° bevel) and (d) 60° Cone.

Fig. 3 shows the 45° bevel needle inserted into each material and shows that the fibrous nature of the konjac does give it greater resistance to tearing than the gelatine or the agar. The konjac clings to the needle as it is inserted, (Fig. 3a), but the surface of the gelatine and agar splits, forming a tear (Figs. 3b and c). Since the agar performed no better than the gelatine under the tearing test, it was discounted, since the gelatine was a more established surrogate material.

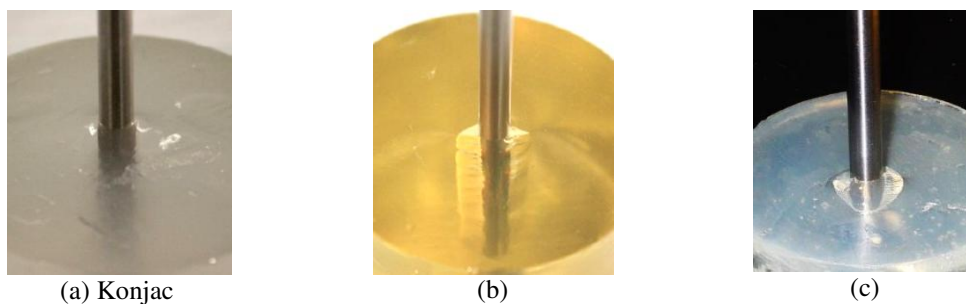


Fig.3 Demonstrating that (a) konjac gel has a greater resistance to tearing than (b) gelatine or (c) agar

Performance under load

A cylinder sample of each gelatine material was loaded in compression in a Tinius Olsen 5kN single column H5kS benchtop test machine with a step increment of 1mm, starting from 0 to 10 mm and its corresponding reaction force recorded. The tests were repeated three times for each specimen and the values averaged out before plotting the stress strain curves and determining the Young’s modulus (Table 1) from the slope of the graph. Due to the barrelling effect, the stress was calculated

from the load using an average area, with consideration of constant volume [17]. The effect of barrelling is difficult to overcome and the only solution was to perform multiple tests to obtain the average value. A similar procedure was attempted with the konjac specimen, but the material proved to be very non-linear and showed signs of creep, (Fig 4a) and viscoelasticity (Fig. 4b).

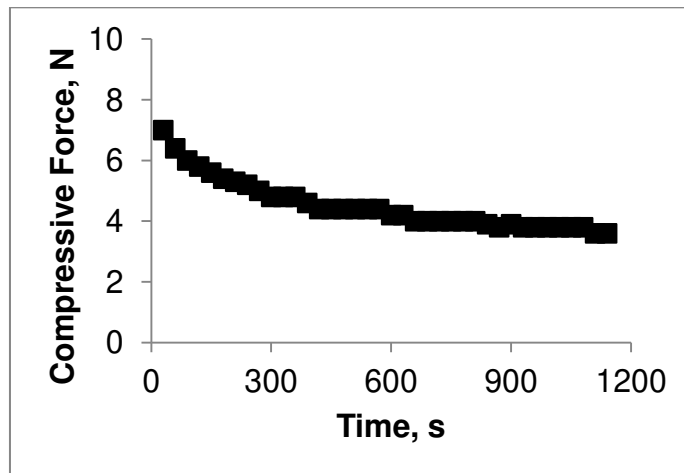


Fig. 4(a) Load response of Konjac specimen with 10 mm displacement over 20 minutes.

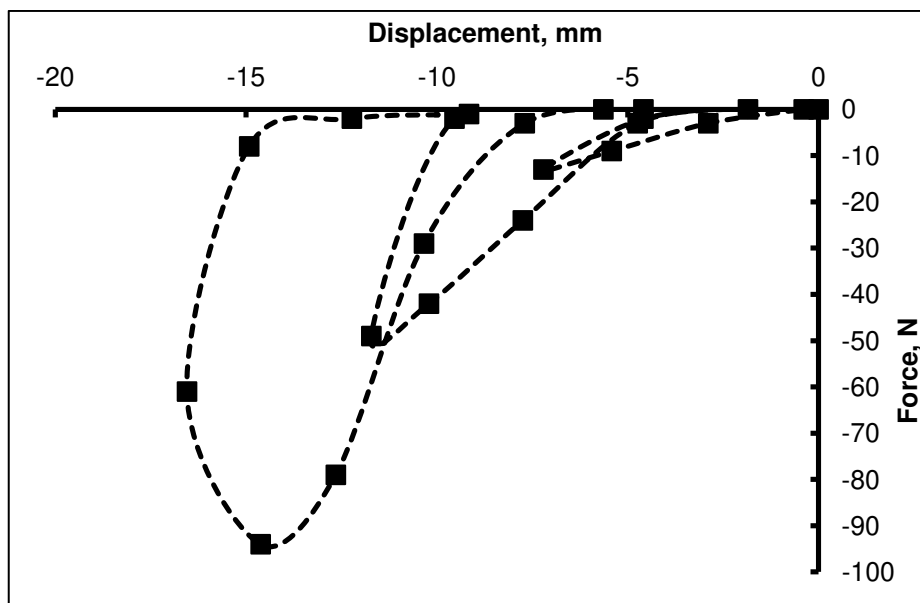


Fig. 4(b) Force-displacement of konjac when loaded and unloaded three times.

Discussion of surrogate material experiments

At this stage in the research the gelatine was considered to be the most suitable selection for photoelasticity tests on the different shaped needles, because, of the materials tested, it had the best transparency and birefringent sensitivity, and the modulus can be varied with the use of additives in order to represent the varying stiffness of the human body. The photoelastic calibration shows that the concentration of gelatine and additives directly affect the fringe constant. While the specimen with added sorbitol produced the highest fringe constant, this was not a significant increase compared to mixture with glycerine only. As a result, the range of gelatine with glycerin as an additive was selected for this research. The lower stiffness specimen (2:10:10) could represent soft tissues such as the stomach or lung, and the stiffer gelatine (3:5:10) could simulate the heart or liver [12]. However, the 3:10:10 gelatine would be most appropriate to replicate aged human skin. Additionally, the 3:10:10 gelatine exhibits repeatability in mechanical and photoelastic properties with an acceptable toughness.

The konjac showed excellent resistance to tearing when compared to the gelatine and agar (Fig.3), but the transparency was poorer. The fact that the material shows a non-linear viscoelastic response is not being seen as a drawback since tissue also displays such properties and thus the properties are being explored further.

PUNCTURE EXPERIMENTS

During needle insertion, the reaction force on the needle is the sum of contributions from tissue rupture ahead of the tip, tissue deformation in the vicinity, and friction between the tissue and needle. Furthermore, any change in geometry along the needle will also cause changes in needle-tissue response. For example, the work required to puncture the skin will be much higher than the work required to cut the tissue due to the high stress concentration on the crack tip. Therefore, the insertion of the needle cannot be considered as a single motion. The needle insertion process could be separated into three phases, which is the deflection on skin before puncture, insertion of the needle tip and insertion of the tip and shaft as shown in Fig. 5 [18].

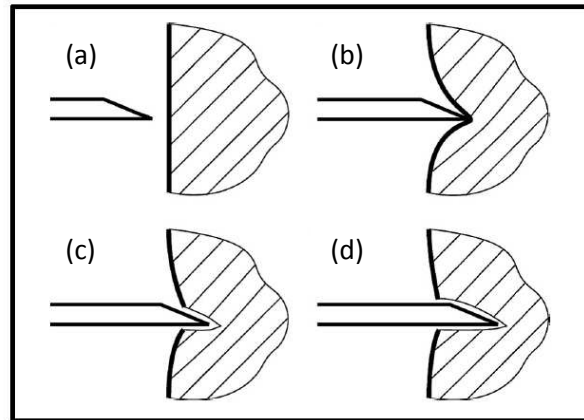


Fig. 5 Phases of needle insertion: (a) before contact; (b) boundary displacement; (c) tip insertion; (d) tip & shaft insertion [18]

Phase I: Boundary Displacement

The first phase begins as soon as the needle comes into contact with the tissue and ends when the first crack is initiated. During this phase, the insertion force comprises the force to overcome tissue elasticity and friction, and the force to initiate the first crack (puncturing of the tissue boundary) [4]. Due to the tissue's elasticity, the needle does not puncture the tissue upon contact but the boundary moves with the needle. The small area where the needle pushes the boundary is called the interfacial area. The small interfacial area creates high stress levels with minimal force, and as the stress exceeds the tissue puncture toughness, a crack will be initiated.

Phase II: Tip Insertion

The second phase begins from the first crack initiated to the position where the whole needle tip is in the tissue and the tissue boundary touches the shaft. A rupture often occurs in a split second because the large amount of strain energy stored during the boundary displacement phase is released at once to extend the crack. As a result, a large drop in force is noticed and the cutting forces can be seen to be relatively lower than the initial puncture force due to the presence of the crack. However, additional puncture events may occur as the needle pierces further into the tissue, due to tissue inhomogeneity. Additionally, the insertion force also includes the effort to push the crack apart caused by the gradual increase of area of the bevel needle tip. Hence, the increase in force is nonlinear at this stage.

Phase III: Tip and Shaft Insertion

Finally, the third phase starts as the tip and shaft travel through the tissue and stops when the tip reaches its target. In this phase the needle tip is still subjected to cutting and/or puncture force but the behaviour is much more consistent as the interfacial area of tip and tissue remain constant [18]. The increasing contact area between the shaft and the tissue causes varying friction force throughout the insertion process. Although most of the stock needles manufactured are lubricated, a certain degree of friction still persists.

Puncture Test Procedure

A new set of gelatine-glycerine-water mixture 3:10:10 cylindrical specimens were prepared for the puncture experiment. The 25° bevel tip needle was attached to the Tinius Olsen machine, placed just touching the gelatine specimen and the load cell

and displacement readings were both set to zero. The needle was then displaced and every 0.5mm its corresponding force recorded. This procedure was repeated with the 60° cone tip scaled needle. The results are shown in Fig. 6. The procedure was then repeated within a light field circular polariscope and the images recorded at every 1mm displacement are shown in Fig. 7. Since photoelasticity gives an integral effect, a two dimensional system was created so there were insignificant out of plane effects. To hold the gelatine in space, glass sheets were glued together forming a container which could hold the gelatine at a uniform thickness of 15mm.

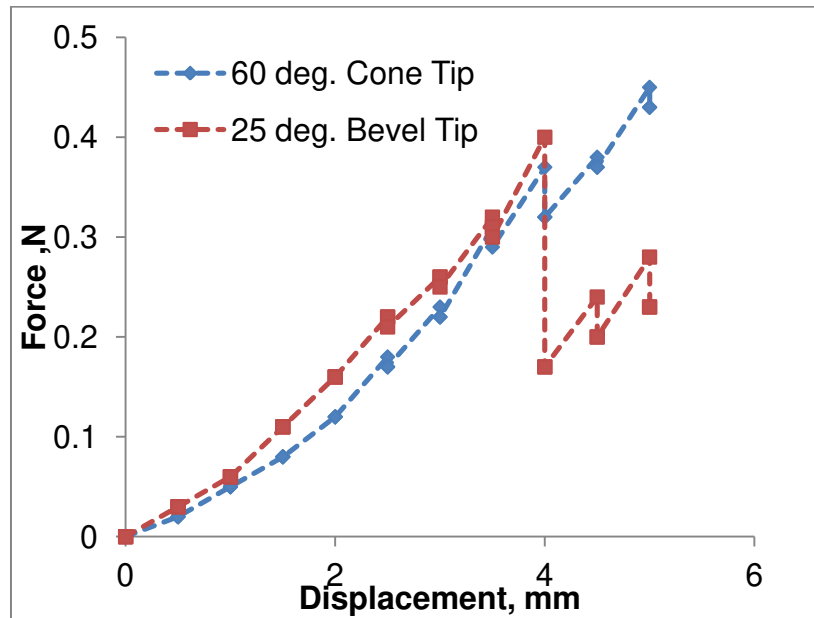


Fig. 6 Force-displacement response of two needle tips in the 3:10:10 gelatine specimen.

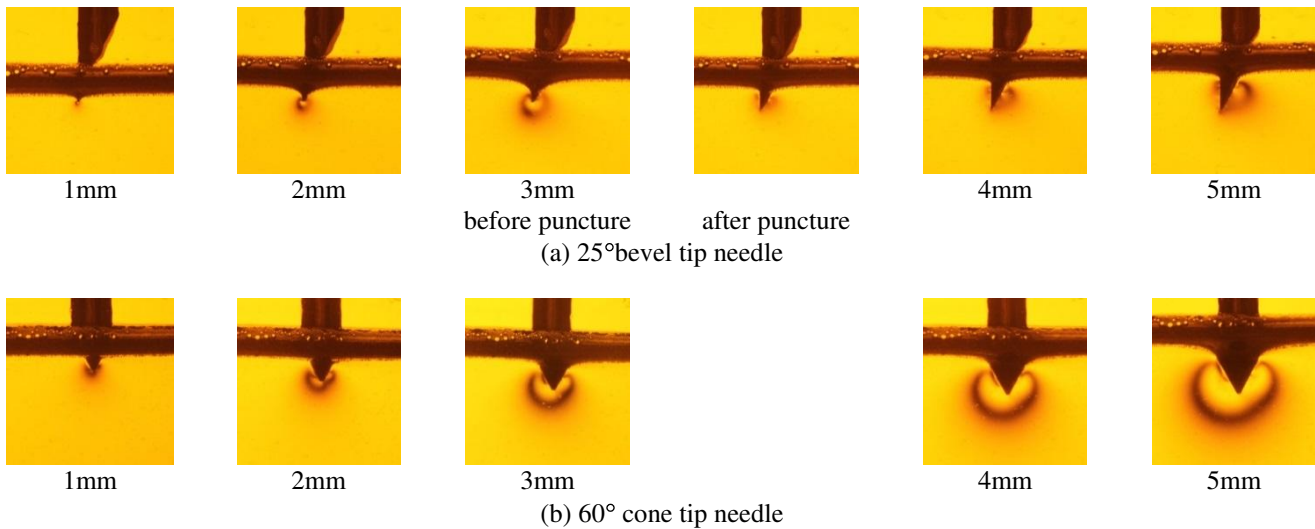


Fig. 7 Photoelastic fringes, representing the shear stress in the 3:10:10 gelatine; showing the (a) 25° bevel tip and (b) 60° cone tip with increasing insertion into the material

Discussion of puncture test

The 25° bevel tip insertion in Fig. 6 shows a steady force increase which drops abruptly at 4mm. The peak force at 4mm displacement in this case could be described as the puncture event. At this point, the first crack is initiated and the strain energy accumulated was relaxed resulting in a sharp drop in force. So the puncture event signifies the end of phase one and the start of phase two. In phase two, the variation in force is due to the continuous cutting action of the tip as well as to overcome friction and internal stiffness. The 60° cone tip produces a more moderate puncture event.

In Fig. 7 the half order photoelastic fringes represent the shear response around the needle tips. It can be seen that the fringe concentration, and hence stress, increase with insertion depth. However for the 25° bevel tip needle, the puncture event is seen in the reduction in fringes after 3mm displacement. Other than the relaxation in stress, the fringe pattern also changed after the puncture event. Before the puncture, the fringe could be seen to surround the tip uniformly, but after the material had failed, the accumulated stresses at the tip were relieved. The figure highlights the difference between symmetrically shaped tips and the bevelled tip needles. Besides showing an increase in shearing force as the needle is inserted, it confirms the hypothesis of an asymmetric force distribution on bevel tipped needles.

CLOSING REMARKS

This work is still in preliminary stages and further research is needed to develop a robust full-field experimental method for validating computational models of needle insertion. Although the gelatine mixed with glycerin was used for the photoelastic tests, and showed good optical properties and controllable stiffness, its tearing characteristics are quite different from those of biological tissues. However, the konjac demonstrates potential as a tissue surrogate which doesn't tear, and experiments are underway to quantify its non-linear viscoelastic behaviour for comparison with the non-linear properties of tissue. It is also acknowledged that the development of an isotropic surrogate material is unrealistic in the human body, but is a starting point to a much wider research programme.

The current study only considered phase one of needle insertion and the second and third phases are to be considered in the future. Another factor not discussed in this work is how the magnitude of insertion force or its rate influences the magnitude of needle deflection. Besides the cutting force, the shearing caused by friction between the needle shaft and the tissue is also damaging to the material. Since photoelasticity provides a map of shear strain, the technique appears ideal for investigation of friction and results to be presented at the conference show some interesting behaviour.

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