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Modelling and characterisation of a ultrasound-actuated needle for improved visibility in ultrasound-guided regional anaesthesia and tissue biopsy

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ABSTRACT

Clear needle visualisation is recognised as an unmet need for ultrasound guided percutaneous needle procedures including regional anaesthesia and tissue biopsy. With inadequate needle visibility, these procedures may result in serious complications or a failed operation. This paper reports analysis of the modal behaviour of a previously proposed ultrasound-actuated needle configuration, which may overcome this problem by improving needle visibility in colour Doppler imaging. It uses a piezoelectric transducer to actuate longitudinal resonant modes in needles (outer diameter 0.8–1.2 mm, length > 65 mm). The factors that affect the needle's vibration mode are identified, including the needle length, the transducer's resonance frequency and the gripping position. Their effects are investigated using finite element modelling, with the conclusions validated experimentally. The actuated needle was inserted into porcine tissue up to 30 mm depth and its visibility was observed under colour Doppler imaging. The piezoelectric transducer is able to generate longitudinal vibration with peak-to-peak amplitude up to 4 μm at the needle tip with an actuating voltage of 20 V_{pp}. Actuated in longitudinal vibration modes (distal mode at 27.6 kHz and transducer mode at 42.2 kHz) with a drive amplitude of 12–14 V_{pp}, a 120 mm needle is delineated as a coloured line in colour Doppler images, with both needle tip and shaft visualised. The improved needle visibility is maintained while the needle is advanced into the tissue, thus allowing tracking of the needle position in real time. Moreover, the needle tip is highlighted by strong coloured artefacts around the actuated needle generated by its flexural vibration. A limitation of the technique is that the transducer mode requires needles of specific lengths so that the needle's resonance frequency matches the transducer. This may restrict the choice of needle lengths in clinical applications.

1. Introduction

Percutaneous needle procedures are common clinical operations for diagnosis and local therapy. Their most widespread applications are regional anaesthesia [1] and tissue biopsy [2], with approximately 1 million tissue biopsies in the USA performed under regional anaesthesia each year to diagnose cancer [2] and a survey showing approximately 50% of anaesthetists in the USA perform more than 100 spinal and epidural nerve blocks each year [3]. In these procedures, a needle is inserted deep into soft tissue to reach the target, with the accuracy of the insertion a key determinant of the success of diagnosis or effectiveness of treatment [4]. Ultrasound imaging is increasingly used for guidance in such procedures, to visualise anatomical structures of interest as well as the advancing needle. However, there is often a problem identifying the needle because the smooth surface of the shaft reflects incident ultrasound waves away from the imaging probe and the waves scattered back to the probe are too small to be detected. This problem is further exacerbated when needles are inserted at steep angles [5]. Percutaneous procedures with inadequate needle visualisation may result in failed or inaccurate diagnosis and serious complications such as vascular, neural or visceral injury [6,7]. In ultrasound-guided peripheral nerve block, repeated failure to visualise the needle tip was reported even by experienced clinicians who had performed more than 100 such procedures [8]. According to Liberman [9], 9–18% of biopsies performed by experienced operators are inadequate because of poor visualisation.

The risks associated with poor needle visibility have motivated a lot of research. Proposed improvements include advanced

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imaging technologies such as beam steering [10] and 3D ultrasound imaging [11] to improve imaging quality; echogenic needles to increase the ultrasound waves scattered back to the imaging probe [12]; mechanical [13] and optical guidance [14] to optimise needle-beam-alignment; and observation of a vibrating needle under colour Doppler imaging [15]. All these technologies have specific abilities to improve needle visibility but they also all suffer from significant limitations and require further development. The use of colour Doppler imaging combined with a moving needle tip was accepted as one of the most promising methods to improve needle visibility [8]. The needle tip movement can be generated by manual motion of the needle [16], rotating a bent stylet within a needle [17], or piezoelectric actuators [15,18]. Gardineer and Vikomerson [19] patented a device termed a 'VIBRA', which uses a piezoelectric diaphragm to excite flexural vibrations of a needle and was later named ColorMark (EchoCath Inc., Princeton, NJ, USA). The ColorMark has been extensively used for laboratory and clinical trials, and significantly improved the needle visibility [15,20,21]. However, it was also found that the Doppler image tended to bleed into the tissue beyond the needle, resulting in difficulty in determining needle position accurately. The most probable reason for this is the nature of flexural vibration: the displacement is perpendicular to the propagation axis and thus the Doppler imaging of the needle expands beyond it. A similar device, which was based on the flexural vibration of a needle actuated by a piezoelectric buzzer, was reported elsewhere [18] and integrated into a 3-D tracking system [22].

Previously, one of the present authors proposed a ultrasound-actuated needle (USAN) to enhance needle visibility in cancer biopsy and regional anaesthesia procedures [23]. Unlike the aforementioned devices based on flexural vibration [18,19], the USNA uses the longitudinal vibration of a needle actuated by a piezoelectric transducer. Since the vibration in a longitudinal mode is confined within the needle, the Doppler response of the needle is substantially confined to the needle as well. However, flexural vibration of the needle was also excited and was found to induce strong artefacts in the needle images indicating that knowledge of the modal behaviour of the needle is required to tune the longitudinal mode, the topic on which this paper is focused. The present study was carried out with finite element modelling (FEM) and validated with experimental characterisation. It is demonstrated that a needle actuated with a tuned longitudinal mode produces clearer images with fewer artefacts than one operating in a flexural mode. Furthermore improved needle visibility was maintained while the needle was continuously advancing in a test specimen, indicating that real time tracking of the needle position is possible.

2. Ultrasound-actuated needle

The ultrasound-actuated needle (USAN) (Fig. 1) comprises a piezoelectric transducer coupled mechanically with a conventional surgical needle. In the piezoelectric transducer, two oppositely-polarised piezoelectric rings (PZ26, Meggitt Sensing Systems, Kvistgaard, Denmark) with electrodes on the flat surfaces, with outer diameter OD = 10 mm, inner diameter ID = 5 mm and thickness t = 2 mm, are sandwiched between a front mass and a back mass with a pre-stressed bolt. The main selection criteria for the piezoelectric rings were commercial availability and appropriate dimensions to give a compact size. The dimensions of the front mass were minimised, while the length of the back mass (22 mm) was chosen so that the displacement node of the transducer was positioned at the flange of the back mass. The purpose of the pre-stress is twofold: it applies a compressive stress to the piezoelectric material to avoid tensile stress in application since piezoelectric materials has low tensile strength (around 44–49 MPa [24]) and it ensures effective physical coupling between adjacent parts of the transducer. Two brass shims (not shown) serve as electrodes to supply electric signals to the transducer. The centre of the bolt was drilled out to a diameter D = 1.5 mm so that the needle can pass through the transducer. The front mass was designed with a collet which forms a collar around the needle and exerts a strong clamping force when it is tightened. It can accept needles in the range 0.8 mm < OD < 1.2 mm. The USAN is attached to a housing (not shown) through the flange on the back mass. When excited, the piezoelectric transducer actuates the needle to oscillate in a longitudinal vibration mode. As the minimum target depth reported for needle intervention procedures is 10–20 mm [15,25], and the transducer length is 55 mm including the housing (transducer itself 40 mm), this solution is applicable to needles greater than 65 mm in length. It will be possible to reduce the transducer length in future designs, thus reducing this constraint.

3. Finite element modelling

3.1. Modelling approaches

FEM was performed using Abaqus (Dassault Systems, Velizy-Villacoublay, France) based on frequency analysis, a standard procedure which extracts eigenvalues to calculate the natural frequencies and corresponding mode shapes of a system. When modelling the transducer, the brass shims used to supply the electrical signals were not included as their thickness (0.1 mm) was much less than the piezoelectric rings (2 mm). It was checked that this did not have a significantly deleterious effect on the accuracy of the results. As stated in Section 2, the main purposes of the pre-stress are to reduce the tensile stress to prevent the failure of the piezoelectric material and to ensure intimate mechanical contact between the components. As the failure mechanism of the piezoelectric material was not included in the FEM and all the components of the transducer were bonded together ideally, the pre-stress was not considered. Opposing poling directions were specified for the two piezoelectric rings, as required in this multilayer design, and all electrodes were specified with zero potential. The input material properties are listed in Table 1.

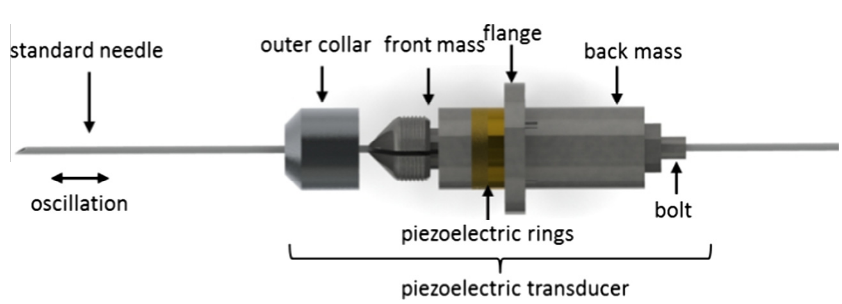

Fig. 1. A schematic of the ultrasound-actuated needle configuration.