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3D Electrode Displacement Elastography using the Siemens C7F2 *fourSight* 4D Ultrasound Transducer

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Abstract

Because ablation therapy alters the elastic modulus of tissues, emerging strain imaging methods may enable clinicians for the first time to have readily available, cost effective, real-time guidance to identify the location and boundaries of thermal lesions. Electrode displacement elastography is a method of strain imaging tailored specifically to ultrasound-guided electrode-based ablative therapies (eg. radiofrequency ablation). Here tissue deformation is achieved by applying minute perturbations to the unconstrained end of the treatment electrode, resulting in localized motion around the end of the electrode embedded in tissue. In this paper, we present a method for 3D elastographic reconstruction from volumetric data acquired using the C7F2 *fourSight* 4D ultrasound transducer, provided by Siemens Medical Solutions. Lesion reconstruction is demonstrated for a spherical inclusion centered in a tissue-mimicking phantom, which simulates a thermal lesion embedded in a normal tissue background. Elastographic reconstruction is also performed for a thermal lesion created *in vitro* in canine liver using radiofrequency ablation. Post-processing is done on the acquired raw radiofrequency data to form surface-rendered 3D elastograms of the inclusion. Elastographic volume estimates of the inclusion compare reasonably well with the actual known inclusion volume, with 3D electrode displacement elastography slightly underestimating the true inclusion volume.

Keywords

3D; 4D; Ablation; Compression; Displacement; Elastography; Elastogram; Elasticity; Elasticity Imaging; Electrode Displacement; Radiofrequency Ablation; Strain; Strain Imaging; Ultrasound

INTRODUCTION AND LITERATURE

Ultrasound strain imaging (or elastography) is a modality in which a compression is applied to tissues, and resultant pre- and post-displacement radiofrequency (RF) echo data are used to

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form images of tissue strain (Ophir et al. 1991; Insana et al. 2001; Varghese et al. 2001; Hall et al. 2003). This map of local strains is used as an indicator of tissue elasticity variations. In general, regions with higher values of elastic moduli are strained less and those with lower elastic moduli incur more strain. Ultrasound strain images provide data on tissues that complement conventional B-mode (echo amplitude) images. There generally is good correlation between pathologic changes in tissue and its elastic properties (Fung 1981), while tissue echogenicity may be uncorrelated (Hall et al. 1997). Thus, elastography provides additional useful information not available on standard B-mode images. It might be said that the idea of imaging tissue elasticity emerged from manual palpation, a procedure in which the physician qualitatively assesses the pathologic condition of the tissue by 'feeling' for stiffness changes. Elastography is a more evolved method that quantifies tissue strains resulting from applied mechanical stimuli.

One of the potential clinical applications of elastography or elasticity imaging techniques is imaging thermally coagulated regions created by ablative therapies (Parker et al. 1998; Stafford et al. 1998; Righetti et al. 1999; Wu et al. 2001; Varghese et al. 2002a; Varghese et al. 2002b; Fahey et al. 2004; Miller et al. 2004; Boctor et al. 2006). RF ablation, a method of tissue destruction, involves the insertion of an electrode into the region of tissue to be ablated. Passage of electric current through the electrode results in ionic agitation in tissue adjacent to the electrically-active region of the electrode, causing intense localized heating. Ablation temperatures greater than 60°C lead to instant tissue necrosis while temperatures greater than 46°C may lead to necrosis depending on the duration of ablation (Solbiati et al. 1997a). Traditionally, conventional sonography has been the imaging modality used most often to guide the placement of the RF electrode in tissue and to monitor the RF ablation procedure (Solbiati et al. 1997a; Solbiati et al. 1997b; Solbiati et al. 2001), essentially due to ease of operation and real-time imaging capability, and not necessarily because of superior image quality. The contrast between treated and untreated regions on the ultrasound image is intrinsically low and varied (isoechoic, hypoechoic or hyperechoic (Gazelle et al. 2000; Liu et al. 2004)), which makes it difficult to ascertain the region of necrosis. The occurrences of hyperechogenicity can be attributed to the presence of gas bubbles generated during the ablation procedure. Locations of these hyperechoic gas bubbles do not precisely correspond to the coagulated region (Malone et al. 1994). Additionally, the gas bubbles usually disappear within an hour after the ablation procedure (Goldberg et al. 2000). The ability of elastography to differentiate between regions of tissue based on their elastic properties has rendered it useful to monitoring lesion extent in RF ablation. The high temperatures encountered in RF ablation induce protein denaturation, a phenomenon that results in elevation of the elastic modulus of the treated region of tissue (Righetti et al. 1999; Wu et al. 2001; Varghese et al. 2002b; Varghese et al. 2003).

Elasticity imaging methods may be classified according to the manner in which tissue motion is induced. External compression elastography is probably the most commonly used technique, where tissue is compressed axially (the axial direction refers to the direction of the ultrasound beam) using either an external plate that also houses the ultrasound transducer (Ophir et al. 1991; Insana et al. 2001; Varghese et al. 2001) or the transducer itself, for freehand compression (Hall et al. 2003). This method works well for superficial organs like the breast and prostate. However, it is not ideally-suited for abdominal organs like the liver and kidney, where our RF ablation monitoring work is directed. Since these organs are not directly accessible, compression has to be applied through the skin and other layers of tissue, often leading to nonuniform compression. This culminates in elastograms with low signal-to-noise (*SNRe*) and contrast-to-noise (*CNRe*) ratios. Other disadvantages of external compression elastography in this application include excessive lateral and elevational motion which, when coupled with physiological motion resulting from cardiovascular and respiratory processes, further degrade

elastographic image quality (Kolen et al. 2002; Varghese et al. 2002a; Varghese et al. 2002a; Varghese et al. 2002b).

When external compression elastography is used in conjunction with RF ablation, the RF electrode is an additional constraint because it could interfere with the positioning of the external compression plate. An alternative method of inducing tissue motion is via the RF electrode itself, thus obviating the need for an external compression plate (Varghese et al. 2002b; Bharat et al. 2005; Bharat and Varghese 2006; Jiang et al. 2007). Strain imaging performed using this mode of tissue displacement is termed 'electrode displacement elastography'. In electrode displacement elastography, the stress field introduced in tissue is not as uniform as that in external compression elastography; instead the displacements induced are localized around the thermal lesion. Two-dimensional electrode displacement elastography has previously been studied using finite element analysis (FEA) simulations (Jiang et al. 2007) and tissue mimicking (TM) phantom-based experiments (Bharat et al. 2005). In this paper, we examine the feasibility of obtaining 3D surface rendered elastographic boundaries of a spherical inclusion in a TM phantom. We use a 4D probe consisting of a curvilinear array that is mechanically rotated to scan a 3D sector volume. Elastographic volume estimates of the inclusion are obtained and subsequently compared with the actual known volume of the inclusion to test the accuracy of the method. The ability to visualize thermal lesion boundaries in 3D is also evaluated using thermal lesions created *in vitro* in canine liver tissue.

MATERIALS AND METHODS

3D electrode displacement elastography described in this paper was tested on a specially-designed TM phantom. This method was also evaluated on canine liver tissue *in vitro*. The tissue experiments were conducted under an approved protocol from the University of Wisconsin Research Animal Resources Center. This section is divided into various subsections, in which the TM phantom experiments, tissue preparation for *in vitro* experimentation, data acquisition procedure, strain estimation and subsequent image processing techniques are explained separately.

Tissue mimicking phantom

3D electrode displacement elastography was evaluated on a cubic TM phantom containing a spherical inclusion, which simulates the presence of a spherical lesion in normal liver tissue. A schematic diagram of the TM phantom for electrode displacement elastography is shown in Figure 1. The TM materials used in the phantom have been described previously (Madsen et al. 2003; Madsen et al. 2005a) and consist of dispersions of microscopic safflower oil droplets in a gelatin matrix. The 19-mm diameter spherical inclusion was formed and bonded round the hooked end of a stainless steel rod, the latter representing an ablation electrode. A storage modulus (real part of the complex Young's modulus) of 50 ± 2 kPa was measured at 1 Hz for the inclusion material using a Bose EnduraTEC® model 3200 system (Madsen et al. 2005b). The background, with a storage modulus of 11 ± 2 kPa, surrounds the inclusion and is bonded to it. Thus, the inclusion is approximately 5 times stiffer than the background. Oil droplets in both materials provide tissue-like ultrasonic backscatter and attenuation. The inclusion is centered in the 12-cm $\times$ 12-cm $\times$ 12-cm cube of background material.

Liver tissue containing thermal lesions

In addition to testing on the TM phantom, the method was also evaluated on canine liver tissue *in vitro*, after the creation of an RF-ablated thermal lesion in the liver. The RF ablation procedure was performed on excised canine liver, using a Valleylab Cool-tip™ ablation electrode (Valleylab, CO, USA). The electrode used was a 17-gauge single electrode, with a 2–3 cm long electrically-active region at the end embedded in tissue. The Cool-tip™ electrode

offers the option of internally circulating chilled water during the ablation procedure, which prevents the charring of tissue adjacent to the electrically-active region of the electrode (we refer to this mode of operation as the 'cooled' mode). The system can also be operated without water circulation (referred to here as the 'dry' mode). The 'dry' mode is clinically operated prior to the removal of the electrode for stopping bleeding. In this experiment, the ablation was performed for 8 minutes (5 minutes in the 'cooled' mode followed by 3 minutes in the 'dry' mode). The 'dry' mode was used to ensure firmer contact between the electrode and tissue (due to tissue coagulation) for electrode displacement elastography. After the RF ablation procedure, the liver with the RF electrode intact was encased in a cubic block of gelatin for performing electrode displacement elastography.

3D data acquisition

Displacements ranging from 0.1 to 0.2 mm in either direction (towards or away from the transducer face) were applied to the RF electrode with precise control using a stepper motor. For each position of the electrode, a 3D sector volume of raw RF data was recorded using the C7F2 *fourSight* 4D ultrasound transducer (Siemens Medical Solutions, USA, Inc.) shown in Figure 2(a). This transducer is curvilinear and contains 192 elements. The array can be mechanically rotated in the elevational direction to scan different planes, each at a specified angle to the previously scanned plane. The maximum sector angle over which the array can be rotated (or *wobbled*) is approximately 75°. The transducer can be operated at the following center frequencies: 2.9 MHz, 3.6 MHz, 4.4 MHz, 5.3 MHz and 6.1 MHz. In our experiment, the center frequency was set at 4.4 MHz. The transducer can be utilized in 2 modes, namely 'continuous' and 'stepped' modes. In the 'stepped' mode, the array is first moved to the angular position as read from a look-up table and a data acquisition frame sequence is triggered after movement of the array has ceased. Since it is essential to synchronize the data acquisition with the position of the RF electrode, the transducer was operated in the 'stepped' mode for electrode displacement elastography.

Acquiring data from the TM phantom—The set-up for imaging the TM phantom using the mechanically rocked array (MRA) transducer is described in Figure 2. Note that the transducer is placed adjacent to the RF electrode, on the upper surface of the phantom. A schematic diagram illustrating the data acquisition procedure is shown in Figure 2(c): each dotted line represents the profile-view of a sector plane of data acquired by the transducer. The transducer acquires data starting at one of the extreme angular positions, steps through the specified sector volume at pre-determined angular increments and acquires a frame of data at each step, before returning to the initial position. This 3D data set constitutes the pre-displacement RF data. Each frame (sector plane) of this 3D data consists of 236 A-lines, with 5192 data points per A-line, acquired at a sampling frequency of 40 MHz. This high sampling rate applied to channel data, with subsequent fine control of time delays during beamforming, allows the system to modify the direction and location of the beam axis in fine steps (Freeman et al. 1999). For this paper, the settings used provided 236 A-lines per frame, leading to an inter-line spacing that is finer than the inter-element spacing. The angular increment between adjacent frames is 0.54° and 39 frames were needed to encompass the entire inclusion for a total of 21.06°. For this geometry, the inter-frame spacing at the center of the inclusion was found to be 0.45 mm. The RF electrode is then displaced vertically using the stepper motor and the data acquisition procedure is repeated for this new position of the RF electrode, providing the post-displacement RF data set. Note that the RF electrode is not displaced more than 0.2 mm from its initial position, so as to avoid damage to the phantom by way of loss of contact between the electrode and the TM material. Various such data sets were acquired at different positions of the RF electrode in the range of ± 0.2 mm from its original position. Eight different independent combinations of these data sets were used as pre-displacement and post-displacement data to generate strain images.

Acquiring data from liver tissue in vitro—A photograph of the liver tissue encased in a gelatin block is shown in Figure 3(a). The transducer was placed adjacent to the RF electrode on the upper flat surface of the gelatin block (Figure 3(b)). The data acquisition sequence is similar to that for the TM phantom and the representation shown in Figure 2(c) holds for this case as well, the only differences being that the electrode is straight (not U-shaped) and the lesion may not be exactly spherical. Nineteen (19) ultrasound frames in the elevational direction were needed to span the entire lesion (as judged from the B-mode images) with an inter-frame spacing of 1.98° . Similar to use of the TM phantom, data sets from the liver sample were acquired for different electrode positions in the range of ± 0.2 mm from its original position. Different combinations of these data sets were utilized as pre- and post-displacement data sets to form elastograms.

Strain estimation and image processing

Corresponding frames from each pre-displacement and post-displacement data set were analyzed in pairs to compute 2D elastograms. Each 2D elastogram was obtained using cross-correlation followed by linear least squares estimation. 1D cross correlation was performed on each pair of A-lines, with 3 mm windows and an overlap of 75%, to form a displacement map. A 5-point linear least squares estimator was then used to estimate the strains from this displacement map. For the TM phantom, 34 such strain maps were estimated, corresponding to the 34 pairs of RF data frames. These strain maps were then stacked together to form a 3D surface rendering of the inclusion (this process is described in detail in the next sub-section). For *in vitro* imaging of the canine liver, the 19 strain image frames required to span the entire lesion were stacked together for 3D surface rendering.

The strain maps obtained from the linear least squares estimator were found to contain artifacts within the boundaries of the inclusion. These artifacts (regions displayed as having high strain) were most likely caused by motion of the RF electrode within the lesion, which would lead to increased decorrelation and hence the falsely elevated strain values. Knowing the cause and location of these artifacts afforded the possibility of correcting for them. In this case, strain within the inclusion was expected to be lower than the high strains surrounding its boundaries. To remove high strain artifacts within the inclusion, a rectangular region of interest (ROI) that enclosed the lower strain pixels belonging to the inclusion was utilized. The mean (μ) and standard deviation (σ) of pixels belonging to the ROI were calculated. An upper threshold was set and any pixel within the inclusion exceeding this threshold was replaced by a random number belonging to a Gaussian distribution with mean ' μ ' and standard deviation ' σ '. A similar procedure was carried out for the thermal lesion strain images.

3D inclusion surface rendering

The 2D strain images obtained in this manner are utilized to form a 3D surface rendering of the inclusion. From each strain image, the boundaries of the inclusion are traced out using manual segmentation. These segmented boundaries are stacked together to form a 3D surface rendered visualization of the inclusion. From this representation, an estimate of the volume of the inclusion can be made. The volume estimate is calculated as the summation of the product of the area of the inclusion on each frame and the inter-frame spacing. Volume estimates were obtained in this manner for 8 surface renderings, and the mean and standard deviation of these estimates determined.

RESULTS

A sample B-mode image frame of the TM phantom showing the inclusion is shown in Figure 4(a). The corresponding strain image acquired from the same echo data field as the B-mode image is shown in Figure 4(b). Observe the close agreement between the strain and B-mode

images with regard to the depiction of the boundaries of the inclusion. Also note that there is an intermediate image processing step prior to obtaining the strain image in Figure 4(b), which is necessitated by the presence of the high strain artifacts within the inclusion due to motion of the RF electrode. The strain image shown in Figure 5 represents the strain image prior to removing these artifacts (the white rectangle represents the ROI used for artifact removal), while Figure 4(b) illustrates the strain image after image correction for the RF electrode artifacts. After removing the strain artifacts, the elastographic depiction of the inclusion is smoother and easier to distinguish from surrounding background strains. The 3D surface rendering of the inclusion, obtained by stacking together segmented versions of these individual 2D strain image frames, is illustrated in Figure 6. The elastographic volume estimates were calculated from 8 such 3D surface renderings, as described in the previous section. The mean volume estimate was 3414.9 mm^3 and the standard deviation of the estimates was 102.6. The actual volume of the inclusion is 3592.8 mm^3 . The underestimation of the inclusion volume with electrode displacement elastography was about 4.95 percent.

The 3D electrode displacement elastographic imaging technique was also evaluated on RF-ablated canine liver tissue *in vitro*. Corresponding B-mode and strain images along a plane of tissue encased in gelatin are shown in Figure 7(a) and 7(b) respectively. Note that the lesion can be clearly demarcated from the surrounding normal tissue on the strain image while no such clear demarcation is visualized on the B-mode image. In fact, echogenicity changes on the B-mode image suggest a much larger lesion than is evidenced from the stiffer coagulated region depicted on the elastogram. We believe the latter depiction is more accurate based on *in vitro* studies (Liu et al, 2004; Varghese et al, 2003). Even in the heavily shadowed region seen on the B-mode image, the elastogram provides nearly complete depiction of the lesion boundaries. As in the strain images of the TM phantom, the motion of the RF electrode results in decorrelation within the boundaries of the lesion on the strain image. Strain images before and after removal of these strain artifacts are shown in Figure 8 and 7(b). The appearances of the lesion on the B-mode and strain images are compared closely in Figure 9, where the border of the lesion's depiction on the strain image is traced on top of the B-mode image.

Figure 10(a) illustrates the 3D surface rendering for the liver lesion created in the *in vitro* RF ablation experiment. The variation in the area of the lesion, as seen on the strain images formed by elevationally stepping through one of the acquired 3D data sets, is shown in Figure 10(b).

DISCUSSION AND SUMMARY

Volume estimates of the spherical inclusion in the TM phantom obtained using 3D surface rendering indicate that electrode displacement elastography underestimates the actual volume of the inclusion. This result is consistent with earlier findings based on area estimates of the inclusion, computed using 2D electrode displacement elastography (Bharat et al. 2005) and external compression elastography on RF ablated lesions (Varghese et al. 2003). Note that a constant value of 0.45 mm was used for the elevational spacing between successive frames. In reality, the elevational spacing is smaller in the near field (regions closer to the transducer) and progressively increases with distance from the transducer; however, this does not affect the volume estimates since the slight overestimation in the near field compensates for the slight underestimation in the far field. Results from the *in vitro* tissue experiment further confirms the ability of electrode displacement elastography for imaging lesions created using RF ablation. The repeatability of lesion depiction on strain images at different elevational planes afforded the possibility of 3D surface rendering of the lesion.

Figure 9 illustrates the fact that strain images provide complementary information to that available from B-mode images. It is apparent that the boundaries of the lesion cannot be ascertained from the B-mode image. Also, this image provides credence to the observation that

the location of gas bubbles on the B-mode image does not exactly correspond to the location of the lesion.

The advantage offered by 3D surface rendering over standard 2D strain imaging is essentially information on the extent of the thermal lesion in the third dimension. The importance of this information can be understood from the lesion size variation in the elevational direction, shown in Figure 10(b). This information may be critical in actual RF ablation procedures where lesion formation may not be uniform. 2D images of the ablated region may not be indicative of maximum lesion size if the lesion formation is non-uniform. In previous 2D experiments involving electrode displacement elastography on the same TM phantom using a linear array transducer, it was found that the inclusion plane imaged was off-center with respect to the RF electrode (Bharat et al. 2007). In other words, the maximum inclusion size was not visualized and it required the incorporation of a correction factor to provide more accurate inclusion dimension estimates. This is feasible when the actual inclusion dimensions are known *a priori*; however, in an actual RF ablation procedure, adding the correction term may be difficult since the lesion dimensions are unknown. This problem is circumvented using the 3D data acquisition technique described in this paper, since the mechanical rotational motion of the transducer element array effectively covers the entire inclusion from one end to the other.

Currently, there are certain restrictions to using this method for real-time 3D strain imaging. The acquisition and storage time for each 2D frame takes from a few seconds to about a minute - a complete data acquisition sequence in the TM phantom imaging experiment (39 frames) took approximately 20–25 mins. Manual segmentation is currently used to delineate the inclusion boundaries, essentially due to discontinuities along the inclusion boundary. Shadowing from the RF electrode sometimes contributes to discontinuities in the boundary, necessitating user input to extrapolate the boundary in such regions, thus limiting the use of automated segmentation. The above concerns currently preclude the use of this transducer for real-time 3D strain imaging. However, this form of 3D imaging may be used as an effective follow-up imaging tool to provide the clinician lesion volume estimates post-ablation. Currently the speed of data acquisition is limited by procedures that ensure RF data integrity during its transfer from the computer memory to the hard disk. Therefore, this drawback can be easily avoided if direct access of the data is available. Also, single electrode configurations of the RF electrode (as opposed to umbrella-shaped or multi-pronged configurations) would reduce the occurrence of shadowing. Preliminary evidence to support this claim is available from Figures 5 and 8. The strain image resulting from *in vitro* RF ablated-lesion (single electrode configuration) exhibits reduced decorrelation when compared to the elastogram obtained from the TM phantom (U-shaped electrode configuration). This may ultimately afford the use of automated segmentation methods to delineate lesion boundaries on elastograms. Together, all these improvements may help realize the goal of real-time 3D strain imaging in the future.

To summarize, 3D surface rendering using electrode displacement elastography was performed on a TM phantom containing a spherical inclusion, which was designed to simulate an RF ablation experimental set-up with an RF electrode embedded in the inclusion. 3D sets of RF data were acquired using a curvilinear array transducer (C7F2 *fourSight* 4D ultrasound transducer, Siemens Medical Solutions), before and after small perturbations (0.1 – 0.2 mm) of the RF electrode. These 3D data sets consist of multiple 2D frames and elastographic processing was carried out on pre- and post-displacement pairs of these individual frames. 2D electrode displacement elastograms clearly show that the inclusion can be demarcated from its surrounding background. 3D surface rendering of the inclusion were obtained by stacking together segmented versions of these 2D strain images. Volume estimates of the inclusion using these 3D surface renderings indicate that electrode displacement elastography slightly

underestimates the volume of the inclusion. 3D electrode displacement elastography was also evaluated on an RF-ablated lesion in canine liver *in vitro*.

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Figure 1. Schematic diagram of the single-spherical inclusion phantom with an electrode embedded in the inclusion. The ultrasound transducer is placed adjacent to the electrode. The dotted line represents one plane of the 3D sector volume imaged by the transducer.

Figure 2.

(a) Photograph of the C7F2 *fourSight* 4D ultrasound transducer (Siemens Medical Solutions, USA, Inc.). This curvilinear transducer contains a 192-element array that can be mechanically rotated to scan a sector volume of the target. (b) Illustration of the geometry of the set-up for imaging the TM phantom. The transducer is placed adjacent to the electrode and held firmly in place by a clamp (not in picture). (c) Schematic of the 3D data acquisition sequence. The dotted lines represent the different planes imaged by the transducer. Starting at one end, the transducer elements are mechanically 'rocked' at pre-specified angular increments to step through the entire sector volume, acquiring a 2D frame of data at each position.

Figure 3.

(a) The picture shows liver tissue encased in a gelatin block after an RF ablation procedure using a Valleylab Cool-tip™ ablation electrode. (b) Illustration of the placement of the transducer for acquiring 3D data on the liver encased in the gelatin block. The top end of the RF electrode is clamped to a holder attached to a stepper motor, to provide precise displacements.

Figure 4.

(a) B-mode image of the central plane of the inclusion. (b) Corresponding strain image, in which the inclusion is depicted as a circular dark region in the center. The decorrelation (halo-like appearance) around the inclusion serves to delineate the inclusion from the background.

Figure 5. Elastogram showing strain artifacts within the inclusion, arising from the motion of the RF electrode. The ROI shows the image pixels used to correct for these artifacts using the procedure described in the text.

Figure 6.
3D surface rendering of the inclusion obtained by stacking together segmented versions of 2D strain images of all the planes of the TM phantom scanned by the transducer.

Figure 7.

(a) B-mode image of approximately the central plane of the thermal lesion contained in the liver which is encased in the gelatin block. (b) Corresponding strain image, in which the lesion corresponds to the dark region surrounded by a halo.

Figure 8.

Elastogram showing strain artifacts within the lesion, arising from the motion of the RF electrode. The ROI shows the image pixels used to correct for these artifacts using the procedure described in the text.

Figure 9. Overlay of the lesion boundaries (from the strain image) on the corresponding B-mode image, illustrating the poor contrast on the B-mode image. Also note that the gas bubbles on the B-mode image do not correspond to the exact location of the lesion.

Figure 10.

- (a) 3D surface rendering of the lesion obtained by stacking together segmented versions of 2D strain images of all the planes of the liver tissue in the gelatin block scanned by the transducer.
(b) Variation in the size of lesion depiction on successive strain images as we step through the 3D data set in the elevational direction.