Table 4.4: Range of values of investigated parameters in the various ARFI models that were studied.

Parameter	Symbol	Values	Units
ARFI interrogation frequency	f	1, 2, 4, 6	MHz
Transducer width	w_{trans}	4, 8, 10	cm
ARFI pulse cycles	n_c	3, 100, 300, 500, 700	–
ARFI source pressure	P_{source}	4, 5, 6, 7, 8	MPa
Lesion depth	d	1, 2, 3, 4, 5, 6, 7, 8, 9	cm
Lesion diameter	$\varnothing S$	0.5, 1.0, 2.0, 2.5	cm
Lesion stiffness ratio	E_{rel}	0.32, 0.56, 1.80, 3.20	–
Blurred lesion blur radius	b_r	1.0, 2.5, 5.0, 7.5	mm
Clustered lesion density	b_ρ	10, 20, 30, 40	cm^{-2}
Clustered lesion radius	r_{bl}	0.5, 1.0, 1.5	mm
Visible human lesion width	$\varnothing L$	0.5, 1.0, 2.0, 2.5	cm

In order to calculate the measured lesion stiffness ratios that are presented in Section 4.3.3, equations 4.14 may be applied. Assuming a constant force applied to the both the lesionous region and the soft tissue reference point, the stiffness ratio of the lesion may be calculated as the ratio between the measured tissue deformation and the measured lesion deformation. As the acoustic radiation force impulse interrogation process is highly dynamic, the maximum induced deformation in the region of interest after application of the acoustic radiation force ceased was used in all characterizations.

$$\sigma = E\varepsilon \tag{4.14a}$$

$$\sigma_{lesion} = \sigma_{tissue} \tag{4.14b}$$

$$E_{lesion}\varepsilon_{lesion} = E_{tissue}\varepsilon_{tissue} \tag{4.14c}$$

$$E_{rel} = \frac{E_{lesion}}{E_{tissue}} = \frac{\varepsilon_{tissue}}{\varepsilon_{lesion}} = \frac{\Delta L_{tissue}}{\Delta L_{lesion}} \tag{4.14d}$$

4.2.5 Physical Phantom Validation

The same CIRS Elasticity QA Phantom model 049 that was used in the quasi-static studies described in Chapter 3 was used to experimentally validate a subset of the ARFI simulations described here. Using a Siemens ACUSON S2000™ portable ultrasound machine with a Siemens 9L4 transducer, ARFI images were acquired of lesions within the phantom. The 9L4 transducer is a compounding transducer which operates from 4 MHz – 9 MHz and with the ACUSON S2000™ is capable of performing quasi-static elastography, ARFI imaging, and shear wave speed quantification. The stiffness ratios of these lesions according to the acquired ARFI telemetry across 10 trials for each nominal lesion stiffness were then compared with their simulated counterparts in an effort to validate the work completed. The results of this characterization are presented in Section 4.3.4. The detailed experimental protocol that was followed for these validations is given in Section C.2 in Appendix C.

4.3 Results

Using the k-space pseudo-spectral model of ultrasound acoustics described in Section 4.2.1, acoustic radiation force distributions were acquired and analysed for the range of input parameters give in Table 4.4. These force distributions were then fed into the time-domain finite-element model of soft tissue deformation described in Section 4.2.3 to examine the difference in relationships between the true and measured tissue stiffness ratios due to the various lesion and transducer parameters that were investigated. The result of these characterizations are presented here.

4.3.1 K-Space Pseudospectral Models of Acoustic Radiation Force

In order to adequately simulate complete ARFI imaging sequences, the magnitude and distribution of acoustic radiation force impulses was simulated according to the procedure outlined in Sections 4.2.1 and 4.2.2. By calculating the temporal average of the intensity distributions, the spatially-varying acoustic body load was obtained. A sample generated force distribution is depicted in Fig. 4.3 for the case when a 2 MHz beam focused at a depth of 4 cm was applied to the tissue for 300 pulse cycles, or 150 µs.

As expected, the force distribution is strongly concentrated around the focal point, extending axially below the focal point as per typical b-mode ultrasound acoustic beams. The net effect of the force distribution depicted in Figs. 4.3a and 4.3b is to push the tissue deeper axially and toward the focal point laterally. This resulted in a peak acoustic radiation force of approximately 175 $\mathrm{kN\,m^{-3}}$ located at the focal point.

Since the k-space pseudo-spectral models employed to simulate acoustic radiation force impulse distributions included absorption and attenuation of the ultrasound waves according to the effects seen in real soft tissues, the depth at which the probe is focused at becomes a critical parameter—the greater the focal depth, the more tissue that the ultrasound must pass through and therefore the more attenuated the signal becomes. This effect is less noticeable with lower frequency ultrasound waves as less energy is absorbed when the particle motion is limited. The resulting acoustic body force generated in the tissue at the focal point for a range of depths of interrogation frequencies is presented in Fig. 4.4.