Anatomic and Functional Imaging Using Row–Column Arrays

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Abstract—Row–column (RC) arrays have the potential to yield full 3-D ultrasound imaging with a greatly reduced number of elements compared to fully populated arrays. They, however, have several challenges due to their special geometry. This review article summarizes the current literature for RC imaging and demonstrates that full anatomic and functional imaging can attain a high quality using synthetic aperture (SA) sequences and modified delay-and-sum beamforming. Resolution can approach the diffraction limit with an isotropic resolution of half a wavelength with low sidelobe levels, and the field of view can be expanded by using convex or lensed RC probes. GPU beamforming allows for three orthogonal planes to be beamformed at 30 Hz, providing near real-time imaging ideal for positioning the probe and improving the operator’s workflow. Functional imaging is also attainable using transverse oscillation and dedicated SA sequence for tensor velocity imaging for revealing the full 3-D velocity vector as a function of spatial position and time for both blood velocity and tissue motion estimation. Using RC arrays with commercial contrast agents can reveal super-resolution imaging (SRI) with isotropic resolution below 20 μm. RC arrays can, thus, yield full 3-D imaging at high resolution, contrast, and volumetric rates for both anatomic and functional imaging with the same number of receive channels as current commercial 1-D arrays.

Index Terms—Beam forming, row-column (RC) arrays, super resolution, ultrasound, velocity measurement.

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I. INTRODUCTION

Currently, 2-D ultrasound imaging is mostly conducted using 1-D array transducers with 192–256 elements, which are employed to dynamically focus on the image. Digital beamformers are used, where the signal from each transducer element is sampled at 4–8 times the center frequency for sampling rates between 12 and 60 MHz. A fully populated 15-MHz array with 256 channels will, thus, give data rates up to 30.7 GB/s, which are beamformed in real time. Currently, most arrays have a fixed geometric focus in the elevation plane (orthogonal to the imaging plane), and the focusing is often poor in this direction, underlining the necessity for 3-D focusing and imaging.

Attaining 3-D ultrasound images requires electronic steering in both the azimuth and elevation directions to allow dynamic focusing along all three directions (axial, azimuth, and elevation). Matrix arrays were early conceived as they allow full control in both directions in both transmit and receive. However, it creates another practical problem as the number of channels increased quadratically with the side length of the array assuming a square array aperture. A straightforward translation to 3-D would give arrays with 192 × 192 = 36864 elements or 256 × 256 = 65536 elements yielding data rates of 2560 GB/s, which is clearly not possible to process in real time. This has been solved by making sparse matrix probes, where only part of the elements are connected resulting in higher sidelobe levels [1]–[6]. A second approach is to make micro-beamforming in the handle to reduce the amount of data. Philips has introduced the fully sampled matrix phased array x-matrix probe shown in Fig. 1 with 9212 elements, which potentially could have 96 × 96 elements. Such a probe can be steered in both directions, and this necessitates an element size of half a wavelength \( \lambda \) given by

\[
\lambda = c/f_0
\]  

(1)

where \( c \) is the speed of sound (1540 m/s in tissue) and \( f_0 \) is the transducer center frequency. In this case for a 3-MHz probe, the element size is 250 μm and the side length of the probe is \( 48\lambda = 24 \text{ mm} \). Much of the beamforming is performed in the transducer handle to reduce the amount of data coming out of...
the probe to probably 256 channels, making this an extremely complex and expensive probe to develop and manufacture.

The focusing ability of ultrasound probes is related to their size and imaging depth. The full-width at half-maximum (FWHM) of the point spread function (PSF) is

$$\text{FWHM} = \frac{\lambda F\#}{W}$$

(2)

where $D$ is the imaging depth, $W$ is the aperture width, and $F\#$ is the F-number, in which its lowest theoretical value is 1/2 (the diffraction limit). The best possible attainable lateral resolution is, thus, $\lambda/2$. For the Philips probe, this can only be attained down to 12 mm, and after 48 mm, the resolution goes beyond $2\lambda$, which is considered as the resolution limit for an acceptable ultrasound image. In cardiology, a lower resolution has to be accepted due to the narrow space between the ribs, which prevents the use of larger probes when scanning the heart. In other applications with a wider acoustic window, such resolution is not acceptable.

In other clinical specialties, the matrix probe should be larger, further increasing the amount of elements and the complexity of the probes. In general, the lateral resolution in the two planes scales with the side length of the probe, and increasing the resolution by two will quadruple the number of probe elements. Matrix probes are, thus, not an optimal approach for attaining a high image quality, and conventional 1-D array probes have a poor out-of-plane resolution limiting their ability to visualize small objects when they are not at the elevation focus. Other solutions for optimal imaging are therefore needed.

A. History of RC Imaging

A possible solution to these problems is to employ row–column (RC) array probes [7]–[25]. Here, the matrix elements are addressed as either rows or columns, as shown in Fig. 2. The amount of connections to an $N \times N$ elements array is $2N$, reducing the amount of connections by $N/2$, which for a $256 \times 256$ elements arrays is a factor of 128. Often, only rows or columns are used in transmit and the orthogonal elements in receive, and suitable multiplexing can therefore reduce the scanner connection to $N$, a further reduction by 2.

It is, thus, possible to have very large RC arrays without the amount of connections to the array getting prohibitively large. The consequence of this is a theoretical focusing capability, which is much better than for a fully populated array, as the width of the array is larger, and the FWHM is correspondingly smaller.

The area of the array scales quadratically with the side length or element count, which is beneficial for the transmitted pressure and the received energy. RC probes can therefore have an increased penetration depth compared to other probes as demonstrated in [26].

The initial idea of RC arrays was presented by Morton and Lockwood [7] at Queen’s University, Kingston, ON, Canada, with simulations of a convex array for revealing the imaging area and PSF. Further simulations were given in [17]. Fabrication of such an array and data from its use was given in [27].

The group by Daher and Yen [9] at the University of Southern California, Los Angeles, CA, USA, has also fabricated a number of arrays and extensively investigated their performance. Initial simulations of a $256 \times 256$ RC array were presented in [8] with more extensive simulations in [9]. Results from a $64 \times 64$ PZT array operating at 5.6 MHz were shown in [10] and later for an impressive $256 \times 256$ PZT array operating at 6.4 MHz with a size of $40 \text{mm} \times 40 \text{mm}$ [12]. More results from a cyst phantom were presented in [13] and a full overview of the results are given in [16]. A spatial compounding method for RC arrays used on the $256 \times 256$
A novel approach similar to RC arrays for 3-D imaging was presented by the group at Roma Tre University, Rome, Italy, using the concept of a criss-cross array [11], where a CMUT array with two spatially superimposed linear orthogonal arrays was investigated. This yields $2N$ connections and an imaging example using two emissions to reduce grating lobes was presented. A fabricated CMUT prototype with overlapping arrays was presented in [28] with $120 \pm 120$ elements, and imaging was conducted using the ULA-OP scanner [29]. Focusing in the orthogonal plane was attained by approximating a Fresnel lens using a varying bias voltage across elements.

Zemp et al. [30] at the University of Alberta, Edmonton, AB, Canada, have also developed a series of RC probes often under the name TOBE: top-orthogonal-to-bottom electrode. The feasibility and fabrication of such an array was presented in [30] for a $64 \times 64$ elements array fabricated in the CMUT technology with more details in [23]. Its use for photoacoustic imaging was demonstrated in [31] using a laser for excitation and synthetic aperture (SA) imaging for creating the image. Chee and Zemp [32] described an advanced modulation scheme, where a combination of individual elements could be acquired simultaneously for the CMUT TOBE array. Other combinations of this scheme were presented in [33] and [34], and results with sidelobes below 45 dB were attained. Recent results for a 10-MHz $64 \times 64$ elements electrostrictive array using the Hadamard encoding in transmit for an increased signal-to-noise ratio and SA focusing were presented in [35] and for 30 MHz in [36], demonstrating the good image quality of these arrays and imaging schemes.

Flesch et al. [37] at the Institut Langevin, Paris, France, have also worked extensively with RC arrays, especially for flow estimation and super-resolution imaging (SRI). A plane wave compounding scheme was described and used for power Doppler imaging (detecting the presence of flow). Using that scheme for flow imaging has unfortunately revealed fairly high grating lobes [38]. The approach has also been used for imaging a rat brain in [39] and [40] using a 15-MHz $128 \times 128$ PZT array.

Our group in Denmark has worked extensively with RC arrays for the last ten years within anatomic and functional imaging primarily based on SA sequences. We have also fabricated a range of RC probes, including PZT- and CMUT-based devices [41], and developed fabrication schemes for diverging lenses and probes with integrated lenses [42], [43]. The various results and possibilities will be presented in the following. The challenges of using the RC array are detailed in Section II, the possibilities for making anatomic images are shown in Section III, and the blood velocity estimation is presented in Section V. A method for SRI is presented in Section VI, and a discussion of the benefits, challenges, and future potential is presented in Section VII.
RC arrays, however, have several issues to address to attain high-quality imaging, including that the contrast of the images is often slightly lower than for current 2-D scanners. These issues are described in the following, starting with the general principles of SA imaging, which is needed to attain an optimal image quality and a high volume imaging rate.

III. ANATOMIC IMAGING

RC arrays can be used in two fundamentally different ways: using focused emissions or using SA imaging with circular or plane emissions. The first necessitates that the image lines are acquired one at a time, and for a volume with $100 \times 100$ lines, this often takes more than a second. The preferred method is therefore to use SA imaging, where dynamic transmit focusing is attained by emitting with a number of broadly insonifying beams and by receiving with the orthogonal array elements for dynamic receive focusing. Emitting with waves that are plane in both directions and tilted along the steerable direction, also called ultrafast imaging, has been studied in [37], [38], and [47], which, however, seems to give fairly high sidelobe and grating lobes.

The second wave type is circular waves, which are plane along the long direction of the transmit element and circular in the orthogonal direction. These virtual sources can have a negative F-number for a diverging wave, or they can be focused on increasing the transmitted pressure. Both focus placements seek to acquire data suitable for creating a synthetic transmit aperture. Synthetic transmit aperture imaging can be used for improving the image quality by having dynamic focusing in both transmit and receive [48]. The imaging is performed by circular transmission with a single or a collection of elements. The origo of the wave is therefore known precisely and can be used in the beamformation. The scattered signal is then received by all elements of the orthogonal transducer elements. The path from transmission to reception can, thus, be precisely calculated. A full volumetric image of the object is focused for each emission as the whole image volume is insonified. This is a low-resolution volume, as it is only focused in transmit. Repeating the process for a number of virtual sources and summing all the low-resolution volumes will yield a high-resolution volume, which is dynamically focused on both transmit and receive.

The spread of the virtual sources and the corresponding largest distance span will determine the FWHM attainable in the transmit direction and correspondingly, the spread of the receive elements will determine the FWHM in the orthogonal direction. The contrast for the resulting PSF is determined by the number of transmit sources and the number of receiving elements. Currently, the best image quality is attained by emitting with a virtual source in one direction and then receiving with the orthogonal elements. SA focusing will then yield the optimal PSF if the beamformer described in Section IV-A is employed. Often a group of elements is used as emitters to increase the emitted energy [49], [50], and the effective width of the aperture is then reduced by the number of elements in the virtual source.

The focusing ability is also dependent on how many receiving element that can contribute to the receive focusing, which is determined by the acceptance angle given by [51]

$$\alpha = 2 \arctan \frac{1}{2F\#}.$$  

A wide element will restrict the acceptance angle and increase the possible F-number. The minimum attainable F-number of 0.5 is obtained when the element has a size of half a wavelength. This also applies for the transmitting elements, and the ideal pitch of the RC array is, thus, $\lambda/2$. 
A. B-Mode Performance of RC Arrays

An example of the PSF and image quality obtainable from a 6-MHz Vermon 128 × 128 elements RC array is shown in Fig. 6. An SA sequence with 96 row emissions followed by 96 column emissions were made using an F-number of −0.7 with 32 elements and Hanning apodization in transmit. The scattered signals were received on all 128 orthogonal elements. This was beamformed with an F-number of 0.7 and a Hanning apodization for both transmit and receive. Imaging was conducted on a 3-D printed PSF phantom [52] with scattering cavities in a 6 × 4 × 4 grid with a 2.05-mm spacing in all three directions. The scatterers are 205 μm wide along the x- and y-axes, but only 80 μm in the z-direction. A Verasonics Vantage 256 scanner was used for the measurements shown on the top row in Fig. 6, and Field II [53], [54] was used for the corresponding simulation shown in the second row.

An isotropic resolution of (1.05λ, 1.10λ, 0.62λ) = (x, y, z) is attained for the measured data, and a similar performance is seen for the simulated data. The data are also compared to a simulated 6-MHz linear array translated over the aperture in steps of 0.2 mm obtaining 100 images. A 12 emissions’ SA sequence was used, and the images from this volumetric scanning are seen in the third row. The linear array probe has an elevation resolution determined by the geometric elevation focus at 22 mm with an F-number of 4.4, and the four rows of point scatterers can therefore not be differentiated due to the fixed elevation focus. The acquisition of the linear array dataset necessitated 12 × 100 = 1200 emissions and mechanical translation, whereas the RC dataset used 192 emissions corresponding to a normal focused linear array image. A volume rate of 52 Hz can therefore be attained down to a depth of 7 cm.

Finally, the bottom row in Fig. 6 shows the in vivo images of a Sprague-Dawley rat kidney. The dynamic range is 60 dB and an isotropic speckle pattern is seen in all three imaging planes due to SA imaging, a constant F-number throughout the image, and the large size of the RC array.

Resolution as a function of depth is visualized on the two left columns in Fig. 7, where points are seen in the xz planes and the wires in the yz-direction. The array has also been used for scanning a tissue-mimicking cyst phantom with an attenuation of 0.5 dB/[MHz·cm], as shown in the two right most columns in Fig. 7. The cyst size diameters are 2.0, 4.0, and 8.0 mm. The 2-mm cysts can clearly be seen down to 50 mm, and the 4- and 8-mm cysts are visible down to the penetration depth of 110 mm.
The array used here is far from optimal for SA imaging. No edge apodization is included in the array and the probe pitch is $\lambda$, which limits the acceptance angle in both transmit and receive. An optimal array with both these properties does currently not exist and can therefore only be simulated. The optimal resolution possible using SA imaging for a 192 × 192 elements RC array has been simulated for two types of arrays in [55]. Both have $\lambda/2$ pitch for optimal imaging with geometries shown in Fig. 8, where the first is a traditional rectangular grid array and the other is an interwoven array for increasing the active area of the array. The second array is only possible to manufacture with silicon CMUT fabrication processes, whereas the first array can be made using the traditional PZT technology. The long elements are edge apodized to avoid the ghost artifacts after the PSFs. Imaging is conducted by emitting with one element at a time and receiving with the orthogonal elements, so a full volume uses 192 emissions, which is the same as for normal
Fig. 7. Two left columns: orthogonal wire phantom images for the Vermon RC probe for a matrix wire phantom with two rows of wires stretch out along the y-direction. Two right columns: orthogonal cyst phantom images for the Vermon RC probe for a tissue-mimicking phantom with an acoustical attenuation of 0.5 dB/[MHz·cm].

Fig. 8. Simulation setup. To the left, the straight element RC array is visualized. The cells have \( \lambda/2 \) spacing, which matches the interelement spacing. The interwoven structure is visualized to the right. The red lines symbolize the assumed acoustic center of the column elements, the green likewise for the rows (from [55]).

TABLE I

|                | FWHM | Azimuth-Range | Elevation - Range | C-plane |
|----------------|------|---------------|-------------------|---------|
| Straight       | 0.62 \( \lambda \) | 0.60 \( \lambda \) | 0.60 \( \lambda \) |         |
| Interwoven     | 0.61 \( \lambda \) | 0.62 \( \lambda \) | 0.62 \( \lambda \) |         |
| CR20dB         | 1.42 \( \lambda \) | 1.42 \( \lambda \) | 1.43 \( \lambda \) |         |
| Straight       | 1.34 \( \lambda \) | 1.34 \( \lambda \) | 1.30 \( \lambda \) |         |
| Interwoven     | 1.34 \( \lambda \) | 1.34 \( \lambda \) | 1.30 \( \lambda \) |         |

The PSFs for the arrays were simulated in Field II and the quantitative numbers for FWHM and contrast are shown in Table I. It can be seen that the resolution is very close to an ideal obtainable resolution of \( \lambda/2 \) for SA focusing and a good contrast is also obtained. Full apodization in transmit and receive has not been employed, and this is why the PSFs are not round and have edge effect artifact, which can be avoided with proper apodization. The contrast is slightly lower than for normal SA imaging 1-D arrays, due to the switching between rows and columns in transmit and receive.

IV. IMAGING PENETRATION FOR RC ARRAYS

The large active area of the RC probe is advantageous for attaining a large penetration depth, defined as the imaging depth, where the signal-to-noise ratio attains a value of 0 dB. This is shown in Fig. 9, where the 128 × 128 elements Vermon RC array was used for imaging a cyst phantom with an attenuation of 0.5 dB/[MHz·cm]. The 6-MHz PZT array attains a penetration down to 11 cm or 428\( \lambda \) when using only 32 elements in transmit using an F-number of \(-1\). Similar results have been attained in [26] for two 62 × 62 elements RC arrays, one fabricated using CMUT technology, and one traditional PZT array. The 3-MHz PZT array attained a penetration down to 14 cm when using only a single element in transmit, whereas using an F-number of 1 or \(-1\) gave predicted penetration depths of 25–30 cm, considerably more than the conventional array’s penetration of 300\( \lambda \)–400\( \lambda \) (15–20 cm). Here, it should also be kept in mind that these are first version linear array imaging, but here, the full volume is acquired and perfectly focused in all three directions.
prototype arrays, the experimental scanner SARUS [56] was used, and the arrays were fairly small (62 × 62 elements, \(\lambda/2\) pitch). A more realistic array with 256 × 256 elements would have a 16 times larger surface area, and using a better prototype and scanner would also significantly increase the penetration depth. This can be translated to imaging with a higher center frequency if the large penetration depth is not needed, which would increase resolution in all three directions proportionally to the wavelength. For the 62 × 62 elements PZT probe using an F-number of \(-1\), the measured mechanical index (MI) was 0.67 and the derated spatial-peak-temporal average intensity \(I_{spta}\) was 0.53 mW/cm. The allowable limits for MI is 1.9 and 720 mW/cm for \(I_{spta}\) for peripheral vessels [57]. It is, thus, possible also to increase the transmitted pressure by a factor of 3, and the limit on \(I_{spta}\) can essentially not be reached, showing the further potential for increasing the signal-to-noise ratio.

### A. Beamforming Implementations

Beamforming of RC data must consider the long and narrow elements, where the emitted field is a plane wave along the long side and a circular wave in the orthogonal direction, as shown in Fig. 3. The left figure illustrates a focused emission at \(\mathbf{s}\) in the \(yz\) plane and the emission of the plane wave in the \(xz\) plane, so the emitted field is described as a focal line and not as a focal point. The ToF calculation has to take this into account, where the transmit time is from the transducer surface to the focal line and from the focal line to the field point \(\mathbf{f}\). The time to reception is then from \(\mathbf{f}\) to the center of the receiving elements. The projection of the distances to the long elements is shown in Fig. 10. A more detailed explanation and the exact equations can be found in [24].

The importance of replacing the delay calculation with the specifics for the RC array rather than the traditional spherical delay calculation is shown in Fig. 11, where the received signal quickly attains the wrong geometric position, if the new calculation method is not employed. This can both lead to the geometric distortion shown but also leads to a diminished resolution and contrast, if not implemented properly.

Efficient implementations of this type of beamforming have been developed for a GPU in [58] and [59]. The software is written under the CUDA environment and takes in radio frequency (RF) data from the RC channels and then yields a focused line, image, or volume. The beamformer is parametric and can be used for very large volumes only limited by the RAM of the GPU card. An example of performance for a state-of-the-art Titan V Nvidia card is shown in Fig. 12. This roughly corresponds to the newer Nvidia GTX 3090 card. This GPU can attain a beamforming performance of around 40 Gsamples/s. Two intersecting B-mode images with 96 lines containing 512 samples for a 192 × 192 elements RC array can, thus, be beamformed with a frame rate around 30 Hz when 64 emissions are used for creating the volume. A full volume with 96 × 96 lines can be beamformed in 1.45 s. It is, thus, possible to make real-time scanning and plane visualization with a state-of-the-art GPU card, and the full volume can be beamformed in a reasonable time for off-line visualization and inspection.
elements and for real or complex (cplx) sample values (from [58]).

The colored lines show performance for 64 or 192 receiving volumes. The throughputs for the shallow phantom and (c) and (d) deep phantom. (a) and (c) Cross planes and (b) and (d) full volumes. The colored lines show performance for 64 or 192 receiving elements and for real or complex (cplx) sample values (from [58]).

The results are based on concave lenses attached to the 62 × 62 elements RC array have been fabricated and tested on the arrays. Results for wire and cyst phantoms have been measured and processed and are shown in Figs. 15 and 16. They both show similar results as for the simulations that an increased field of view is attained along with good focusing abilities and an acceptable contrast. More results and details can be found in [61].

The current equations in the lensed beamformer give reasonable results, but it has been shown that ray-tracing theory can further enhance the quality of the results and increase the field of view [62]. This should be further investigated and incorporated in the beamformers.

C. Convex RC Imaging

An obvious method to avoid making a lens would be to shape the RC probe in a double curved, convex shape as suggested in [17]. The SA sequence developed for flat RC arrays would be nearly directly applicable to a convex array, where the beamforming then would take the geometry into account. Such arrays would have many benefits for abdominal ultrasound imaging. Their footprint could be made quite large, which ensures a low F-number even for large depths. The large size would also ensure a large penetration depth, as the emitted energy is distributed over a large area keeping MI low but still acquiring the returned energy from a large surface. Démoré et al. [17] demonstrated that a 128 × 128 RC convex array could cover at 60° × 60° sector with a good image quality using their imaging scheme.

V. Flow Imaging

Current commercial scanners can all display the velocity of blood in the human circulation. The blood motion is detected
by estimating the positional shift between two emissions using correlation-based estimators finding either the phase or the time shift [63]. These methods are well established and widely used in the clinical for quantifying vascular diseases. They, however, have several drawbacks. The detected shift is only in the direction of the ultrasound beam, and most vessels run parallel to the skin surface, so the least important velocity component is detected. This is often remedied by tilting the ultrasound beam and introducing angle compensation methods. These are unreliable for angles close to 90°, and for complicated vessel geometries, the angle will vary as a function of space and time precluding a single angle correction factor. This has been remedied by introducing vector flow imaging (VFI) in a number of methods [64], [65]. One VFI method uses the transverse oscillation approach where an oscillation perpendicular to the ultrasound beam is introduced during receive beamforming, and the shift in the lateral direction can then be estimated yielding the full 2-D velocity vector [66], [67]. This has been shown in a number of clinical studies to give improved flow estimates, which are more consistent and easier to use for medical doctors [68].

A. Tensor Velocity Imaging

The VFI approach gives consistent results for flow in the 2-D plane but neglects the out-of-plane component. The approach has therefore been translated to RC arrays for both traditionally focused emissions [69], [70] and for a fast SA-based approach [71], [72] using interleaved SA imaging [73], [74]. Here, the full 3-D blood velocity is estimated in the volume for each time instance for full tensor velocity imaging (TVI). The probe can be placed to just cover the vessel, and the full velocity vector is estimated for any position in the volume with hundreds of estimates per second.

An example of TVI is shown in Fig. 17 for measured pulsating flow in a carotid artery phantom, where the arrows indicate the direction and the colors indicate the velocity magnitude. The velocities over time for different positions in the vessel are shown in Fig. 17, showing that the velocity components in all directions can be estimated as a function of time everywhere in the volume. The full 3-D vector velocity field can therefore be acquired for a couple of heartbeats, and the velocity for any place and time can be determined retrospectively after the acquisition has been made, thus increasing the clinical relevance.

Further validation of the TVI method was performed using finite element method (FEM) simulations of pulsating flow in an \textit{in silico} carotid artery phantom [75], where the ground truth is known. Motion correction was employed to improve the estimates [72], and the result is shown in Fig. 18 for both an autocorrelation estimator (left column) and cross correlation estimator (center and right column).
Fig. 17. Pulsating flow in a tissue-mimicking phantom is shown on the left. The flow is visualized using arrows, where color shows velocity magnitude and the arrows depict velocity direction and magnitude. The middle graph shows all three velocity components $v_x$, $v_y$, $v_z$ (red, blue, black) for a point in the vessel center. The right graph shows the components at a point placed near the vessel wall (modified from [71]).

Fig. 18. TVI using a $62 \times 62$ elements RC array from simulated data in the carotid artery for two different estimators compared to the ground-truth finite element data on the right. Arrows indicate direction and magnitude, which is also indicated by the color. A vortex in the upper vessel branch is seen in the top row, and reverse flow is also present in the bottom row (from [75]).

VI. SUPER-RESOLUTION IMAGING

The latest addition to medical ultrasound is SRI, where the microvasculature can be visualized down to vessel sizes of 20–50 $\mu$m. This is attained by injecting a standard contrast agent (SonoVue) intravenously and then image the motion of the gas filled bubbles. A sparse distribution of bubbles makes it possible to track individual bubbles and establish their position with micrometer precision [76]–[82]. The data are acquired over minutes and motion correction of the acquired data must be performed to maintain resolution [83]–[89], but as most of the current methods are in two dimensions, they cannot compensate for large motions and out-of-plane motion.
It has been demonstrated that RC probes can also be used for SRI [90]. A 3-MHz 62 × 62 elements RC array with a half-wavelength pitch was used with an SA pulse inversion sequence with 32 positive and 32 negative row emissions for acquiring volumetric data using the SARUS research ultrasound scanner. Data received on the 62 columns were beamformed on a GPU for a maximum volume rate of 156 Hz when the pulse repetition frequency was 10 kHz. Investigations were performed on 3-D printed point and flow micro-phantoms, where the flow micro-phantom contained a 100-μm radius tube injected with the contrast agent Sonovue. The 3-D processing pipeline uses the volumetric envelope data to find the bubble’s positions from their interpolated maximum signal and yielded a high resolution in all three coordinates. The localization precision for tracking a 3-D printed point phantom was (20.7, 19.8, 9.1) μm in the x-, y-, and z-coordinates. The flow micro-phantom had an estimated radial precision of 16.5 μm in the yz plane and 23 μm in the xz plane [90].

This approach has been translated to the Vermon 128 × 128 array for measurements on a Sprague-Dawley rat kidney. An amplitude modulation sequence with three emissions for the same virtual source and 48 virtual sources spread out over the row elements was employed at a pulse repetition frequency of 20 kHz between the three emissions and 1.3 kHz between the virtual sources for a volume rate of 24 Hz. A 1:5 dilution of Sonovue at an intravenous infusion rate of 55 μl/min was employed over the 135-s acquisition. The data were then processed with an SRI pipeline as described above, and the final 3-D images are shown in Fig. 19 for three different views. The volume rate was fairly low, to keep the data rate low to enable acquisition over a long period of time. For this shallow scanning down to 3 cm, fprf could be maintained at 25 kHz for a volume rate of 250 Hz or recursive imaging [91] could be used to raise the volume rate to 8.3 kHz.

This technique makes it possible to visualize the flow in vessels with sizes down to 20 μm, which can be used in the diagnosis of vascular diseases found along with, e.g., cancer and diabetes. The images can both reveal the anatomy of the vasculature to reveal vascular rarefaction, neovascularization, increased tortuosity, and so on and give quantitative data for the flow to identify changes caused by disease [92], [93].

VII. DISCUSSION AND CONCLUSION

It has been shown that RC arrays essentially can be used for any kind of ultrasound imaging for visualizing the anatomy, blood flow, and tissue motion and performing SRI allowing visualization of the microvasculature and measurement of flow velocities in the microcirculation. The active number of array elements is of the same order as for conventional 1-D arrays, and the number of transmitters and receivers is therefore as for conventional 2-D imaging. Demands on the transmit stage receive data rates, and storage sizes are also the same as for 2-D imaging. The number of beamforming operations depends on what should be visualized in terms on planes and volumes, but high-end GPU cards are capable of attaining real-time visualization of orthogonal planes, and 3-D solid volumes can be calculated in seconds [58], [59].

A good B-mode image quality can be attained by using SA sequences with 2 × 96 emissions on a 128 × 128 elements RC array, yielding an isotropic PSF in the region where a constant F-number can be maintained. FWHM can be close to the diffraction limit if the array is optimized for high-quality imaging with edge apodization and a pitch of λ/2. Even the first version substandard arrays with λ pitch and no edge apodization can yield high-quality in vivo images as shown on a rat kidney scan. Comparing these results to traditional 2-D imaging, it should be kept in mind that the yz and xy planes are never shown. These planes for traditional linear arrays with a fixed geometric focusing have very poor resolution, which at the optimal geometric focusing often is 3–5λ. and away from this focus can be 10λ–20λ instead of 0.6λ attained here. With the RC arrays, it is, thus, possible to attain an isotropic resolution, and much better imaging with a uniform speckle pattern is possible, where any slice and orientation can be attained retrospectively after the data have been stored.

The large size of the arrays, and the use of the full aperture during reception and synthetic transmission, makes the signal-to-noise ratio high. The penetration depth is above 550λ even for low-intensity and low MI transmission and can be increased to be above 800λ for higher pressure transmission surpassing that of conventional 1-D arrays. This is also surpassing 2-D matrix arrays, as their elements are small and
often sparse arrays have to be used to keep the element count manageable.

TVI can also be obtained using an RC array with only 62–128 receiving elements and a transverse oscillation approach. The full velocity vector in any direction and at any place in the volume can be shown as a function of time. Using SA imaging and recursive imaging makes it possible to retrospectively probe any location in the volume and see the time evolution of the flow. Vortices and complex flow are easily visualized for any slice, making quantification easier. The method is complex with interleaved emissions, transverse oscillation, motion correction, and dedicated beamformers and estimators resulting in a high computational load. The demands are a factor 62–128 times lower than for a fully populated array, and modern GPUs offer thousands of processing units to make real-time beamforming and estimation possible.

Volumetric SRI can also be attained by RC arrays with a resolution down to 15 μm. A long observation over minutes is essential to ensure imaging of the smallest vessels, and the data rate from the arrays should therefore be low. This is very difficult to attain for fully populated or sparse matrix arrays due to the many elements, and they are also difficult to manufacture due to the small \( \lambda/2 \) pitch elements when the frequency is high. RC arrays, therefore, have distinct advantages for SRI as the data rate corresponds to normal linear arrays, and high-frequency arrays are easier to manufacturer and attain the needed signal-to-noise ratio for the low MI emissions demanded for contrast agents. Here, SA is also beneficial as the emitted MI is low, and a good SNR is attained when all emissions are combined.

RC arrays currently also have a number of drawbacks. The imaging is now performed by switching between the transmitting and receiving aperture, and this necessitates more emissions for SA 3-D imaging than what is currently used for 2-D SA imaging. Often only 8–12 emissions are needed for very high-quality 2-D SA images, whereas 48–96 times two emissions are needed for an optimal volumetric image quality. New methods for improving this are currently being investigated [47]. It should also be possible to develop combined sequences, where data are acquired for both anatomic and functional imaging with an optimal image quality at fast volume rates and where both high and low velocities can be reliably estimated from the same data. This is an area of active research.

The contrast in 3-D imaging is also poorer than for 2-D imaging, and this should be further optimized. This problem is also related to the lack of proper arrays. Rasmussen et al. [24] showed that edge apodization of the elements is vital for avoiding ghost echoes, and the imaging also benefits from having \( \lambda/2 \) pitch elements, which very few RC arrays have. Having better arrays with the correct geometry will obviously improve both image quality and frame rate to mature RC technology. The imaging region of current flat arrays is also limited to the rectangular region of the transducer footprint. This can possibly be solved by employing lensed RC arrays or convex RC arrays, but again, proper arrays are lacking and should be developed.

Overall, it can, however, be stated that RC arrays can fulfill all the demands for fast, high-quality volumetric ultrasound imaging. Anatomic, flow, functional, and SRI have all been demonstrated for simulations and phantom measurements and a few in vivo examples. It is our hope that the great potential of general RC imaging will be demonstrated in future clinical trials using optimized arrays. The combination of having a large array capable of having a good focusing, contrast, and penetration depth can, especially for abdominal imaging, lead to high-quality 3-D anatomic and functional images.

REFERENCES

[1] R. E. Davidsen, J. A. Jensen, and S. W. Smith, “Two-dimensional random arrays for real time volumetric imaging,” Ultrason. Imag., vol. 16, no. 3, pp. 143–163, Jul. 1994.
[2] S. S. Brunke and G. R. Lockwood, “Broad-bandwidth radiation patterns of sparse two-dimensional Vernier arrays,” IEEE Trans. Ultrason., Ferroelectr., Freq. Control, vol. 44, no. 5, pp. 1101–1109, Sep. 1997.
[3] J. T. Yen and S. W. Smith, “Real-time rectilinear volumetric imaging using a periodic array,” Ultrason. Med. Biol., vol. 28, no. 7, pp. 923–931, Jul. 2002.
[4] A. Austeng and S. Holm, “Sparse 2-D arrays for 3-D phased array imaging-design method,” IEEE Trans. Ultrason., Ferroelectr., Freq. Control, vol. 49, no. 8, pp. 1073–1086, Aug. 2002.
[5] B. Diarra, M. Robini, P. Tortoli, C. Cachard, and H. Liebgott, “Design of optimal 2-D nongrid sparse arrays for medical ultrasound,” IEEE Trans. Biomed. Eng., vol. 60, no. 11, pp. 3092–3102, Nov. 2013.
[6] A. Ramalli, H. Boni, E. Roux, H. Liebgott, and P. Tortoli, “Design, implementation, and medical applications of 2-D ultrasound sparse arrays,” IEEE Trans. Ultrason., Ferroelectr., Freq. Control, early access, Mar. 25, 2022, doi: 10.1109/TUFFC.2022.3162419.
[7] C. E. Morton and G. R. Lockwood, “Theoretical assessment of a crossed electrode 2-D array for 3-D imaging,” in Proc. IEEE Symp. Ultrason., Oct. 2003, pp. 968–971.
[8] N. M. Daher and J. T. Yen, “Rectilinear 3-D ultrasound imaging using synthetic aperture techniques,” in Proc. IEEE Ultrason. Symp., vol. 2, Aug. 2004, pp. 1270–1273.
[9] N. M. Daher and J. T. Yen, “2-D array for 3-D ultrasound imaging using synthetic aperture techniques,” IEEE Trans. Ultrason., Ferroelectr., Freq. Control, vol. 53, no. 5, pp. 912–924, May 2006.
[10] C. H. Seo and J. T. Yen, “SA-5 64 × 64 2-D array transducer with row-column addressing,” in Proc. IEEE Ultrason. Symp., Oct. 2006, pp. 74–77.
[11] A. Savoia et al., “P2B-4 crisscross 2D cMUT array: Beamforming strategy and synthetic 3D imaging results,” in Proc. IEEE Ultrason. Symp., Oct. 2007, pp. 1598–1601.
[12] C. H. Seo and J. T. Yen, “PSF-4 256×256 2-D array transducer with row-column addressing for 3-D imaging,” in Proc. IEEE Ultrason. Symp., Oct. 2007, pp. 2381–2384.
[13] C. H. Seo and J. T. Yen, “Recent results using a 256 × 256 2-D array transducer for 3-D rectilinear imaging,” in Proc. IEEE Ultrason. Symp., Nov. 2008, pp. 1146–1149.
[14] S. I. Awan and J. T. Yen, “3-D spatial compounding using a row-column array,” Ultrason. Imag., vol. 31, no. 2, pp. 120–130, Jan. 2009.
[15] A. S. Logan, L. L. Wong, and J. T. W. Yeow, “2-D CMUT wafer bonded imaging arrays with a row-column addressing scheme,” in Proc. IEEE Int. Ultrason. Symp. (IUS), Sep. 2009, pp. 984–987.
[16] C. H. Seo and J. T. Yen, “A 256 × 256 2-D array transducer with row-column addressing for 3-D rectilinear imaging,” IEEE Trans. Ultrason., Ferroelectr., Freq. Control, vol. 56, no. 4, pp. 837–847, Apr. 2009.
[17] C. E. M. Démoré, A. Joyce, K. Wall, and G. R. Lockwood, “Real-time volumetric imaging using a crossed electrode array,” IEEE Trans. Ultrason., Ferroelectr., Freq. Control, vol. 56, no. 6, pp. 1252–1261, Jan. 2009.
[18] A. S. Logan, L. L. Wong, A. I. H. Chen, and J. T. W. Yeow, “A 32 × 32 element row-column addressed capacitive micromachined ultrasonic transducer,” IEEE Trans. Ultrason., Ferroelectr., Freq. Control, vol. 58, no. 6, pp. 1266–1271, Jun. 2011.
[19] A. I. H. Chen, L. L. Wong, A. S. Logan, and J. T. W. Yeow, “A CMUT-based real-time volumetric ultrasound imaging system with row-column addressing,” in Proc. IEEE Int. Ultrason. Symp., Oct. 2011, pp. 1755–1758.
M. F. Rasmussen and J. A. Jensen, “3D ultrasound imaging performance of a row-column addressed 2D array transducer: A simulation study,” *Proc. SPIE*, vol. 8675, Mar. 2013, Art. no. 86750C.

M. Rasmussen and J. Jensen, “3-D ultrasound imaging performance of a row-column addressed 2-D array transducer: A measurement study,” in *Proc. IEEE Int. Ultrason. Symp. (IUS)*, Jul. 2013, pp. 1460–1463.

J. T. Yen, “Beamforming of sound from two-dimensional arrays using spatial matched filters,” *J. Acoust. Soc. Amer.*, vol. 134, no. 5, pp. 3697–3704, May 2013.

A. Sampaleanu, P. Zhang, A. Kshirsagar, W. Moussa, and R. J. Zemp, “Top-orthogonal-to-bottom-electrode (TOBE) CMUT arrays for 3-D ultrasound imaging,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 61, no. 2, pp. 266–276, Feb. 2014.

M. F. Rasmussen, T. L. Christiansen, E. V. Thomsen, and J. A. Jensen, “3-D imaging using row-column-addressed arrays with integrated apodization—Part II: Transducer fabrication and experimental results,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 62, no. 5, pp. 959–965, May 2015.

H. Bouzari, M. Engholm, S. I. Nikolov, M. B. Stuart, E. V. Thomsen, and J. A. Jensen, “Imaging performance for two row-column arrays,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 66, no. 7, pp. 1209–1221, Jul. 2019.

A. W. Joyce and G. R. Lockwood, “Crossed-array transducer for real-time 3D imaging,” in *Proc. IEEE Int. Ultrason. Symp.*, Sep. 2014, pp. 216–220.

A. Stuart Savoia et al., “A 120+120- element crisscross CMUT probe with its real-time switchable electronic and Fresnel focusing capabilities,” in *Proc. IEEE Int. Ultrason. Symp. (IUS)*, Oct. 2018, Art. no. 8580084.

E. Boni et al., “ULA-OP 256: A 256-channel open scanner for development and real-time implementation of new ultrasound methods,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 63, no. 10, pp. 1488–1495, Oct. 2016.

R. J. Zemp, W. Zheng, and P. Zhang, “Feasibility of top-orthogonal-to-bottom-electrode (TOBE) 2D CMUT arrays for low-channel-count 3D imaging,” in *Proc. IEEE Int. Ultrason. Symp.*, Oct. 2011, pp. 498–502.

R. K. W. Chee, A. Sampaleanu, D. Rishi, and R. J. Zemp, “Top orthogonal to bottom electrode (TOBE) 2-D CMUT arrays for 3-D photoacoustic imaging,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 61, no. 8, pp. 1393–1395, Aug. 2014.

R. K. W. Chee and R. J. Zemp, “Feasibility of modulation-encoded TOBE CMUTs for real-time 3-D imaging,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 62, no. 4, pp. 771–775, Apr. 2015.

C. Ceroici, T. Harrison, and R. J. Zemp, “Fast orthogonal row-column electronic scanning with top-orthogonal-to-bottom electrode arrays,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 64, no. 6, pp. 1009–1014, Jun. 2017.

K. Latham, C. Ceroici, C. A. Samson, R. J. Zemp, and J. A. Brown, “Simultaneous azimuth and Fresnel elevation compounding: A fast 3-D imaging technique for crossed-electrode arrays,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 65, no. 9, pp. 1657–1668, Sep. 2018.

C. Ceroici, K. Latham, B. A. Greenlay, J. A. Brown, and R. J. Zemp, “Fast orthogonal row-column electronic scanning experiments and comparisons,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 66, no. 10, pp. 1093–1101, Jun. 2019.

K. Latham, C. Samson, and J. Brown, “A new 3-D imaging technique integrating ultrafast compounding, Hadamard encoding, and reconfigurable Fresnel lensing,” *IEEE Trans. Ultrason.*, *Ferroelectr.*, *Freq. Control*, vol. 68, no. 5, pp. 1618–1627, May 2021.

M. Flesch et al., “4D in vivo ultrasonic imaging using a row-column addressed matrix and coherently-compounded orthogonal plane waves,” *Phys. Med. Biol.*, vol. 62, no. 11, pp. 4571–4588, Jun. 2017.

J. Sauvage et al., “A large aperture row column addressed probe for in vivo 4D ultrasonic Doppler ultrasound imaging,” *Phys. Med. Biol.*, vol. 63, pp. 1–12, Oct. 2018.

J. Sauvage et al., “Ultrafast 4D Doppler imaging of the rat brain with a large aperture row column addressed probe,” in *Proc. IEEE Int. Ultrason. Symp. (IUS)*, Oct. 2018, pp. 1–4.

J. Sauvage et al., “3D functional imaging of the rat brain using a large aperture row-column array,” *IEEE Trans. Med. Imag.*, vol. 39, no. 6, pp. 1884–1893, Jun. 2020.
S. Harput, J. Pierre, O. Couture, and M. Tanter, “A new method for estimation of velocity vectors,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 45, no. 3, pp. 837–851, May 1998.

J. A. Jensen, “A new estimator for vector velocity estimation,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 48, no. 4, pp. 886–894, Apr. 2001.

A. H. Brandt et al., “Vector flow imaging compared with pulse wave Doppler for estimation of peak velocity in the portal vein,” Ultrasonics Med. Biol., vol. 44, no. 3, pp. 593–601, 2018.

S. Holbek, T. L. Christiansen, M. B. Stuart, C. Beers, E. V. Thomsen, and J. Arendt Jensen, “3-D vector flow estimation with row–column-addressed arrays,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 63, no. 11, pp. 1709–1814, Nov. 2016.

S. Holbek et al., “Common carotid artery flow measured by 3-D ultrasonic vector flow imaging and validated with magnetic resonance imaging,” Ultrasonics Med. Biol., vol. 43, no. 10, pp. 2213–2220, 2017.

M. Schou et al., “Fast 3-D velocity estimation in 4-D using a 62 × 62 row–column addressed array,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 68, no. 3, pp. 608–623, Mar. 2021.

L. T. Jorgensen, M. Schou, M. B. Stuart, and J. A. Jensen, “Tensor velocity imaging with motion correction,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 68, no. 5, pp. 1676–1686, May 2021.

J. A. Jensen, “Estimation of high velocities in synthetic-aperture imaging—Part I: Theory,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 66, no. 6, pp. 1024–1031, Jun. 2019.

J. A. Jensen, “Estimation of high velocities in synthetic-aperture imaging—Part II: Experimental investigation,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 66, no. 6, pp. 1032–1038, Jun. 2019.

L. T. Jorgensen, M. S. Traberg, M. B. Stuart, and J. A. Jensen, “Performance assessment of row–column transverse oscillation sensor velocity imaging using computational fluid dynamics simulation of carotid bifurcation flow,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 69, no. 4, pp. 1230–1242, Apr. 2022.

O. M. Viessmann, R. J. Eckersley, K. Christensen-Jeffries, M. X. Tang, and C. Dunsby, “Acoustic super-resolution with ultrasound and microbubbles,” Phys. Med. Biol., vol. 58, no. 18, pp. 6447–6458, Sep. 2013.

M. A. O’Reilly and K. Hynynen, “A super-resolution ultrasound method for blood vessel mapping,” Med. Phys., vol. 40, no. 11, Oct. 2013, Art. no. 110701.

Y. Desailly, J. Pierre, O. Couture, and M. Tanter, “Resolution limits of ultrafast ultrasound localization microscopy,” Phys. Med. Biol., vol. 60, no. 22, pp. 8723–8740, Nov. 2015.

K. Christensen-Jeffries, R. J. Browning, M. X. Tang, C. Dunsby, and R. J. Eckersley, “In vivo acoustic super-resolution and super-resolved velocity mapping using microbubbles,” IEEE Trans. Med. Imag., vol. 34, no. 2, pp. 433–440, Feb. 2015.

C. Errico et al., “Ultrafast ultrasound localization microscopy for deep super-resolution vascular imaging,” Nature, vol. 575, no. 7579, pp. 499–502, Nov. 2015.

D. Ackermann, G. Schmitz, and S. Member, “Detection and tracking of multiple microbubbles in ultrasound B-mode images,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 63, no. 1, pp. 72–82, Jan. 2016.

K. Christensen-Jeffries et al., “Super-resolution ultrasound imaging,” Ultrasonics Med. Biol., vol. 46, no. 4, pp. 865–891, 2020.

K. B. Hansen et al., “Robust microbubble tracking for super resolution imaging in ultrasound,” in Proc. IEEE Int. Ultrason. Symp. (IUS), Sep. 2016, pp. 1.

J. Foiert, H. Zhang, T. Ilovits, L. Mahakian, S. Tam, and K. W. Ferrara, “Ultrasound localization microscopy to image and assess microvasculature in a rat kidney,” Sci. Rep., vol. 7, no. 1, Dec. 2017, Art. no. 13662.

V. Hingot, C. Errico, M. Tanter, and O. Couture, “Subwavelength motion-correction for ultrafast ultrasound localization microscopy,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 65, no. 5, pp. 803–814, May 2018.

M. Piepenbrock, S. Dencks, and G. Schmitz, “Reliable motion estimation in super-resolution U.S. by reducing the interference of microbubble movement,” in Proc. IEEE Int. Ultrason. Symp. (IUS), Oct. 2019, pp. 384–387.

T. M. Kierski et al., “Superharmonic ultrasound for motion-independent localization microscopy: Applications to microvascular imaging from low to high flow rates,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 67, no. 5, pp. 957–967, May 2020.

I. Taghavi, S. B. Andersen, C. A. V. Hoyos, M. B. Nielsen, C. M. Sorensen, and J. A. Jensen, “In vivo motion correction in super-resolution imaging of rat kidneys,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 68, no. 10, pp. 3082–3093, Oct. 2021.

J. A. Jensen et al., “Three-dimensional super resolution imaging using a row-column array,” IEEE Trans. Ultrason., Ferroelec., Freq. Control, vol. 67, no. 3, pp. 538–546, Mar. 2020.

S. Nikolov, K. Gammelmark, and J. A. Jensen, “Recursive ultrasound imaging,” in Proc. IEEE Ultrason. Symp., vol. 2, Oct. 1999, pp. 1621–1625.

I. Taghavi et al., “Automatic classification of arterial and venous flow in super-resolution ultrasound images of rat kidneys,” in Proc. IEEE Int. Ultrason. Symp. (IUS), Sep. 2021, pp. 1–4.

S. B. Andersen et al., “Evaluation of 2D super-resolution ultrasound imaging of the rat renal vasculature using ex vivo micro-computed tomography,” Sci. Rep., vol. 11, no. 1, Dec. 2021, Art. no. 24335.
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