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Simulating Capacitive Micromachined Ultrasonic Transducers (CMUTs) using Field II

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Abstract—Field II has been a recognized simulation tool for piezoceramic medical transducer arrays for more than a decade. The program has its strength in doing fast computations of the spatial impulse response (SIR) from array elements by dividing the elements into smaller mathematical elements (MEs) from which it calculates the SIR responses. The program features predefined models for classical transducer geometries, but currently none for the fast advancing CMUTs. This work addresses the assumptions required for modeling CMUTs with Field II. It is shown that rectangular array elements, populated with cells, can be well approximated by neglecting the cells. Further, it is demonstrated that scaling of the SIR translates into better computational efficiency.

I. INTRODUCTION

The currently released version of the ultrasound simulation program Field II [1], [2] has been developed mainly with the classical piezoelectric medical ultrasound transducers in mind. Thus, all of the predefined transducer geometries in Field II are based on flat, curved, or double-curved transducers with rectangular elements. So far, piezoceramic transducers have been dominating the field of ultrasound imaging. However, for new transducer technologies, such as capacitive micro-machined ultrasonic transducers (CMUTs) [5], the support in Field II has been limited.

CMUT technology has become an enabling transducer technology, which has caught significant attention not only within the medical imaging community. CMUTs consist of multiple capacitive cells with one movable top electrode (plate), all electrically connected in parallel, to form transducer elements. These elements can be used in an arbitrary array configuration. The shape of the cells (e.g. circular, square, hexagon) and elements is defined by lithographical methods, which, besides high precision, provide good design flexibility and a wide possible operation frequency range.

Each top plate of the CMUT cells deflects due to an applied dc bias voltage and the ambient pressure. Modeling of each deflected cell with Field II requires a dense grid of MEs. MEs are Field II’s subdivision of the transducer surface into smaller transducer elements of which a well known analytical solution to the SIR exists.

This paper addresses the assumptions needed for modeling the SIR of CMUT cells with Field II. It investigates the possibility of using a scaling of the SIR with the ratio of active and inactive area of a transducer element consisting of multiple cells.

II. THEORY

The field response from any flat piston transducer with a homogenous moving surface can be calculated with Field II using the SIR method. The pressure is calculated by convolving the surface acceleration with the SIR in emission. The pulse-echo SIR (PESIR) can be calculated by convolving the SIR in emission with the SIR in receive [9]. Modeling the SIR of curved surfaces, for which the analytical SIR is unknown, requires a subdivision of the surface into a dense grid of MEs. The total response is then calculated by superposition. This method is only valid under the assumption that each ME moves in the normal direction of the surface element that it represents, and it is a necessity that the wave propagation is linear. The SIR is a solution to the Rayleigh integral which limits its validity for highly curved transducers. Therefore, the integral does not account for the fact that a slightly curved transducer will introduce reflections and diffraction. However, these effects are negligible for the case of a small curvature compared to the wavelength and transducer size [10], [11], [12]. These are the underlying assumptions made for this work.

![Fig. 1. Cross sections of regular CMUT (a) and Piston CMUT (b).](image-url)
plate deflection can a simple analytical solution for the SIR be found see Arditi et al.[12].

Figure 1 shows the plate deflection for two types of CMUTs. One for regular CMUT cells and one for a CMUT that features a center mass in each plate, i.e., a piston CMUT, see [6]. Note that the plate material at the center of the regular CMUT cell is moving a larger vertical distance compared to the plate material closer to the post region. In addition, only the plate material right at the center of the cell features an entire perpendicular movement. Thus, we assume that during excitation, the ME representing the plate material closer to the post region, moves perpendicular to the surface segment that it represents. In the very near-field of the cell this assumption will not hold, thus, piston CMUT cells are more suitable for the SIR modeling approach. This is because a large fraction of the plate is moving as a real piston, and the rest of the plate will have a relatively small contribution to both the acoustic pressure field and the electrical field inside the vacuum sealed cavity. Hence, the validity of the Rayleigh integral is not so much compromised as for the regular CMUT cell. In general, the validity of the aforementioned assumptions becomes even more critical for other specific conditions or modes of operation. For example, CMUT plates that exceed a static deflection-to-thickness ratio of 20%, i.e. when also in-plane stress starts to matter in terms of static and dynamic response of the plate [7]. Another example are CMUT cells that are biased beyond the pull-in point, i.e., when the plate experiences contact with the bottom of the cavity (collapse mode of operation) [8].

However, for cells that are arranged in a pattern to form a large aperture size transducer, the above-mentioned assumptions and approximations become less of a source of error, in particular at larger distances from the transducer surface. Hence, for medical imaging applications one can expect that the influence of the non-ideal radiation pattern from each individual cell will be very small. Further, note that acoustic cross-talk cannot be modeled for either CMUTs or piezoceramic transducers with Field II. Setting up Field II manually for modeling each cell can be a cumbersome procedure for complicated geometries. A significant simplification is possible under the assumption that the cell deflections are of minor influence to the superposition and that each cell can be considered as a plane piston. Further, this is valid for the assumption that the only physical difference between a full transducer element with each cell modeled (ECM) and a single square model (SSM), where no cells are considered, is the active area. The active area for a SSM is bigger than the one for an ECM. If it is further assumed that the cells are distributed with a pitch equal twice the cell radius, i.e., the boundary rims being in contact, the fraction between the active areas will be

\[ f = \frac{A_{\text{square}}}{A_{\text{cell}}} = \frac{2r_c^2}{\pi r_c^2} = \frac{4}{\pi}, \]

where \(A_{\text{square}}\) is the active area of the SSM, and \(A_{\text{cell}}\) is the active area of an ECM. For cell-to-cell pitches larger than the cell diameter the areas to be calculated and the fraction becomes \(\frac{A_{\text{square}}}{A_{\text{cell}}} = 1\). A good approximation of the SIR for an ECM can be made with a SSM if the SIR is scaled with the area fraction.

A few pitfalls in relation to this method have to be addressed. First, the start and the end times of the two different SIRs will be different, depending on the location of an observing point relative to the transducer. This has a significant influence if the exact time of flight is required. Second, the exact same energy of the SIR pulses is rarely achieved, i.e. this can only be an approximation. Third, near-field simulations are indeed invalid when using such an approximation.

III. Simulation

A program that defines CMUTs in Field II terminology was created. The program sets a user-defined number of circular cells in either a linear one-dimensional or two-dimensional grid to form a square-shaped transducer element. An arbitrary number of elements can be defined to form a transducer array. The plate deflection is set equal the solution of a static deflection model [13]. The cells can be placed with a minimum pitch corresponding to the diameter of the cells. All cells in a particular element are assumed to experience a uniform excitation. The number of MEs required to resolve a single cell is controlled with a resolution factor, \(n\). This factor is an integer number and corresponds to the number of MEs that fit across a cell’s radius.

To study the error in approximating the ECM with the SSM, a single cell with a radius \(r_c = 18 \mu m\) and the simulation

![Fig. 2. SIRs for a square element, square, a flat circular cell, cell, and a scaled version of square, scaled. SF 40 GHz. Only each 12th point is shown.](image1)

![Fig. 3. The mean RMS of the pressure pulses for 20 random points, each solved 1000 times. The error is calculated relatively to a high resolution solution at 10 GHz and \(n = 50\).](image2)
The same study with 20 randomly distributed pulses has shown that the response from a square element with the height and the width of twice the cell radius are scaled to 0% relative to a single flat cell and a corresponding flat rectangular cell. The RMS of the PESIR pulses are shown in Table I. The simulations are conducted at a sampling frequency (SF) of 10 MHz and with the bounding line solver of Field II’s energy preservation of the pulses. The lowest possible number of MEs is preferred.

<table>
<thead>
<tr>
<th>Area deviation</th>
<th>Pressure deviation</th>
<th>Pulse-echo deviation</th>
<th>Mean area deviation</th>
<th>Mean pressure deviation</th>
<th>Mean pulse-echo deviation</th>
<th>Mean pulse pressure deviation</th>
<th>Mean pulse-echo deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>{0, 0, 4.7} mm Lines 0.02 %</td>
<td>{0, 0, 4.7} mm Lines 0.032 %</td>
<td>{0, 0, 4.7} mm Lines 0.064 %</td>
<td>20 points Lines 0.021 %</td>
<td>20 points Lines 0.05 %</td>
<td>20 points Lines 0.10 %</td>
<td>20 points Rect 0.022 %</td>
<td>20 points Rect 0.051 %</td>
</tr>
</tbody>
</table>

Mean pulse-echo deviation 20 points Lines 0.10 %
Mean area deviation 20 points Lines 0.10 %
Mean pulse pressure deviation 20 points Rect 0.051 %
Mean pulse-echo deviation 20 points Rect 0.103 %

Table I
Deviation of the Area of SIRs, Pressure Pulses, and PESIRs between a Flat Cell and a Scaled Version of the Response from a Square Transducer Element. Two Different Field II Solvers were Applied.

<table>
<thead>
<tr>
<th>Points</th>
<th>Solver</th>
<th>Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>20 points</td>
<td>Lines</td>
<td>1.85 %</td>
</tr>
<tr>
<td>20 points</td>
<td>Lines</td>
<td>5.62 %</td>
</tr>
<tr>
<td>20 points</td>
<td>Lines</td>
<td>11.12 %</td>
</tr>
<tr>
<td>20 points</td>
<td>Rect</td>
<td>1.87 %</td>
</tr>
<tr>
<td>20 points</td>
<td>Rect</td>
<td>3.99 %</td>
</tr>
<tr>
<td>20 points</td>
<td>Rect</td>
<td>7.92 %</td>
</tr>
</tbody>
</table>

Table II
Deviation of the Area of SIRs, Pressure Pulses, and Pulse-Echo Pulses between a Deflecting Circular Cell and a Scaled Version of the Response from a Square Transducer Element.

<table>
<thead>
<tr>
<th>Vbias</th>
<th>Young’s module</th>
<th>V</th>
<th>GPa</th>
</tr>
</thead>
<tbody>
<tr>
<td>36</td>
<td>169</td>
<td>V</td>
<td>GPa</td>
</tr>
<tr>
<td>0.29</td>
<td>0.43</td>
<td></td>
<td></td>
</tr>
<tr>
<td>150</td>
<td>1.125</td>
<td>μm</td>
<td></td>
</tr>
<tr>
<td>0.43</td>
<td>18</td>
<td>μm</td>
<td></td>
</tr>
<tr>
<td>101.3</td>
<td>kPa</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table III
Simulation Parameters Used for Calculating the Cell Deflection.

SIRs calculated at a point \((x_p, y_p, z_p) = (0, 0, 4.7)\) mm relative to a single flat cell and a corresponding flat rectangular cell with the height and the width of twice the cell radius are shown in Fig. 2. The simulations are conducted at a sampling frequency (SF) of 10^4 GHz and with the bounding line solver of Field II. In Fig. 2, cell, square, and scaled are the responses from the circular cell, the square element, and the square element scaled with \(f\), respectively.

As seen in Fig. 2, the response from a square element is, as expected, spread along a longer time interval. If the pulses of a single cell and a corresponding flat rectangular cell with the height and the width of twice the cell radius are shown in Fig. 2. The simulations are conducted at a sampling frequency (SF) of 10^4 GHz and with the bounding line solver of Field II. In Fig. 2, cell, square, and scaled are the responses from the circular cell, the square element, and the square element scaled with \(f\), respectively.

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A. Simulating a populated transducer element

Ultrasound imaging simulation requires at least one transducer element fully populated with cells. Consider a single transducer element populated with 5x150 cells. It was previously shown that the response from the rectangular element can be scaled with (1). It is therefore assumed that within certain limitations this will be applicable for an entire element, where the cell-to-cell spacing prevents the cell boundaries from touching each other, as well. The scaling factor used is
In proximity to the transducer, time of flights will become of visible influence on images for high frequency excitation and good care has to be taken on convergence of the solution in the very near-field if the cell responses are approximated.

**REFERENCES**


