Optimization of piezo-MEMS layout for a bladder monitor

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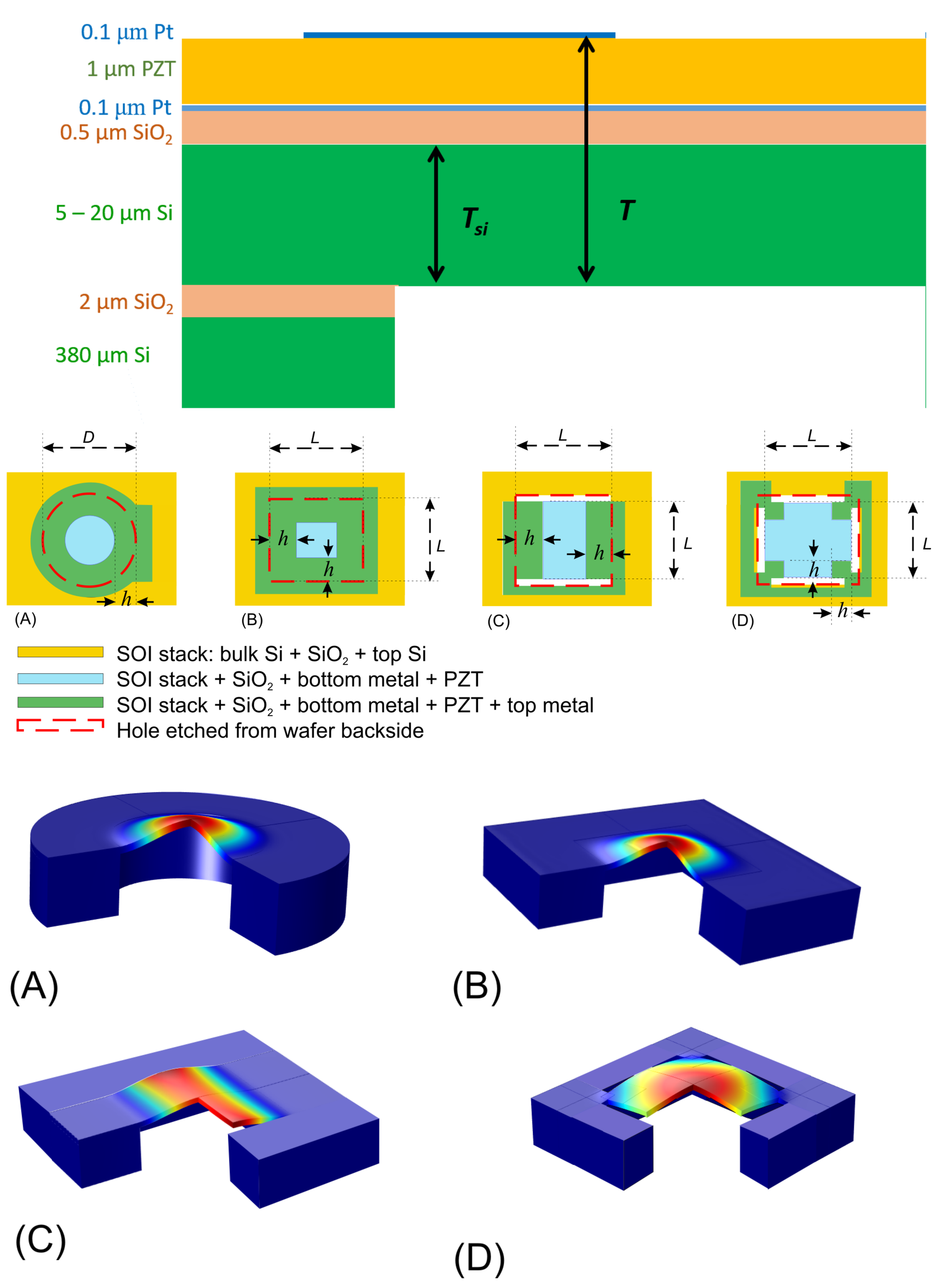
Abstract

MEMS technology has never been used for non-imaging wearable ultrasound products, where e.g. lateral resolution is a minor issue. We combine analytic modeling, finite element simulations, and measurements to optimize the performance of four piezoelectric MEMS layouts for a wearable bladder monitor. An array of square diaphragms, clamped on two sides, appears to be most favorable.

Introduction

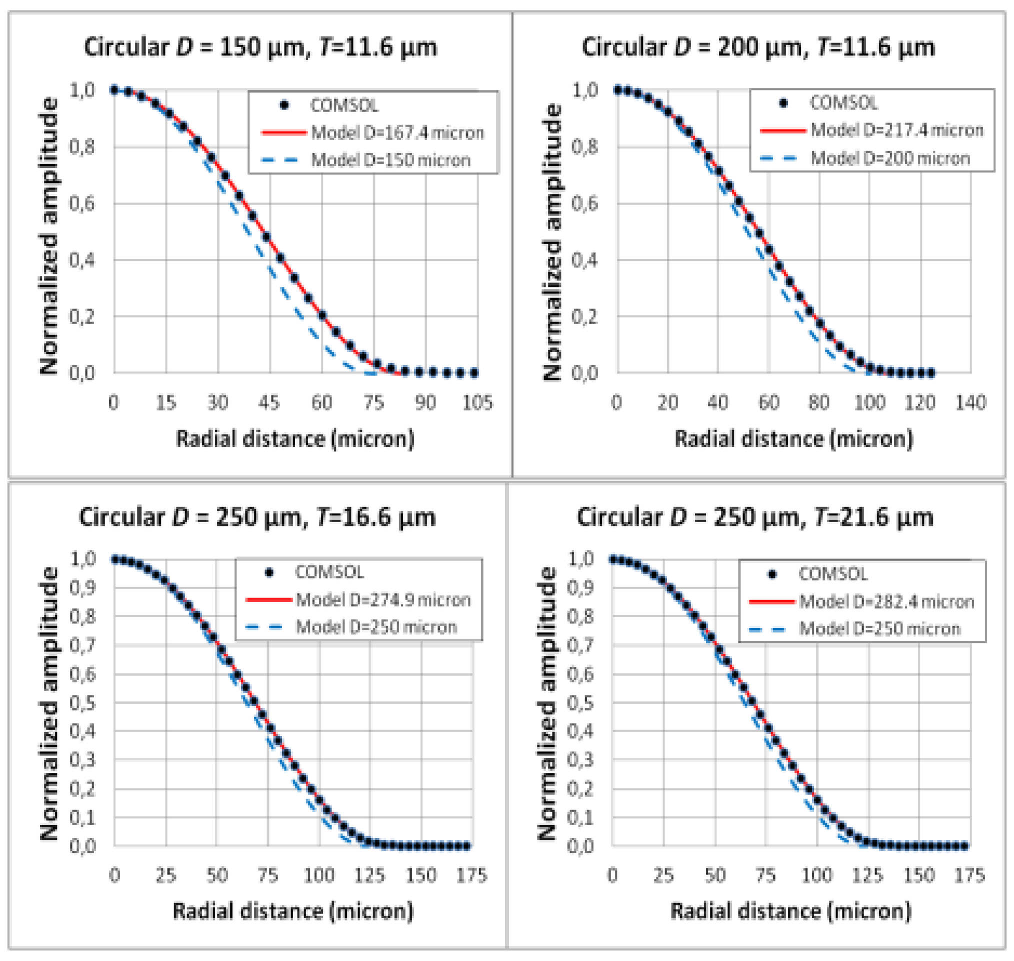
Urinary incontinence is a huge problem affecting children, disabled people and elderly, costing \$65 B/y in the US. Diapers and bed-wetting alarms do not solve this problem. Ultrasound is often used to measure bladder size, but current hand-held equipment prevents 24/7 monitoring. Novioscan BV develops a wearable ultrasound system to monitor bladder size 24/7 and to warn when it should be emptied. A MEMS (Micro electromechanical system) will enable a small size bladder monitor, which is more comfortable for the user.



Schematic cross section of the MEMS (top), layouts of the four geometries (middle) and artist impression of the MEMS (bottom). Geometries: circular clamped (A), square clamped (B), square clamped on 2 sides (C), square clamped at 4 corners (D).

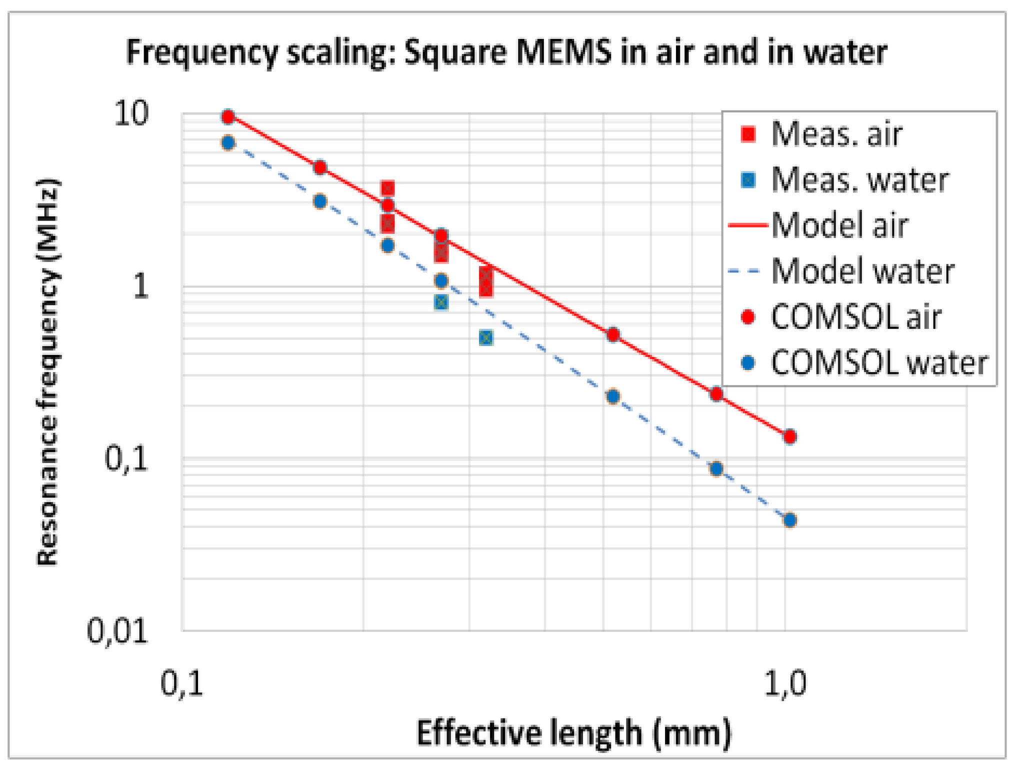
Single MEMS diaphragms

Our scaling rules result from analytic models, based on Kirchhoff's thin plate theory*, verified by simulations in COMSOL. Our MEMS diaphragms are not rigidly clamped, since the supports are made of the same silicon as the diaphragm, placed on a less stiff SiO_2 layer. We correct the lateral dimensions D for the non-rigid clamping: $D_{eff} = D + 1.5T$, where T is the thickness of the diaphragm.

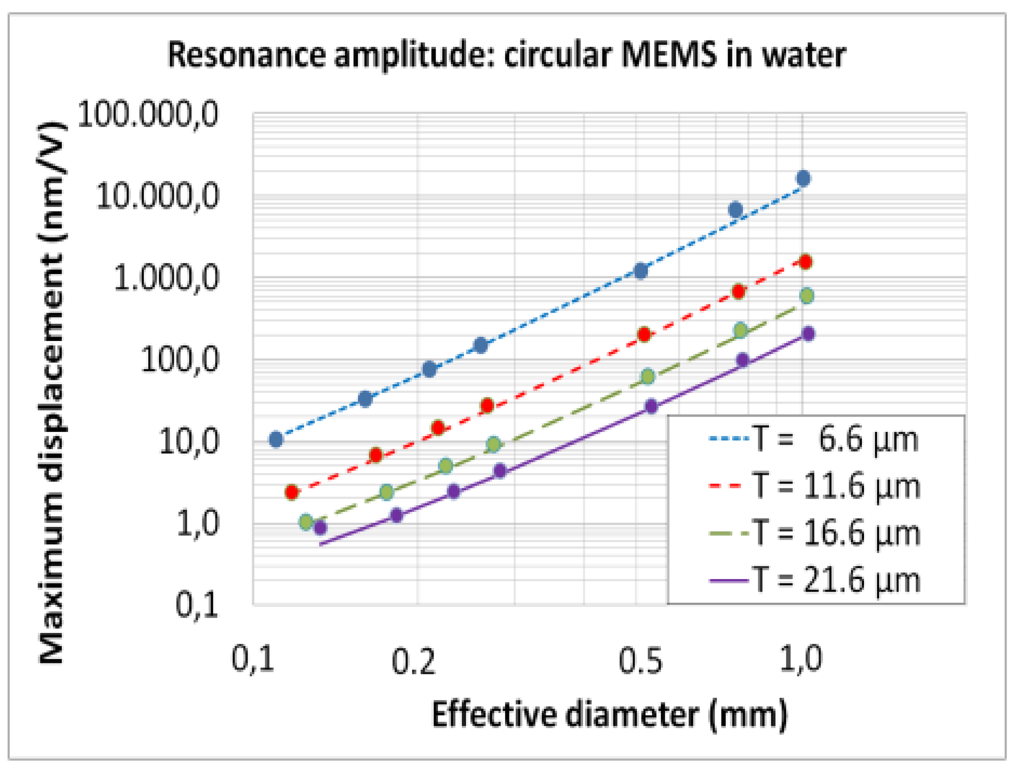


Resonant amplitude in air for circular MEMS, simulated in COMSOL (dot) and modeled analytically either using for the radial distance the diameter of the diaphragm, D (dashed), or $D_{eff} = D + 1.5T$ (solid line), where T is the diaphragm's thickness.

We take all layers in the diaphragm stack into account in second order approximation of the thickness of the top layers, compared with the silicon layer.



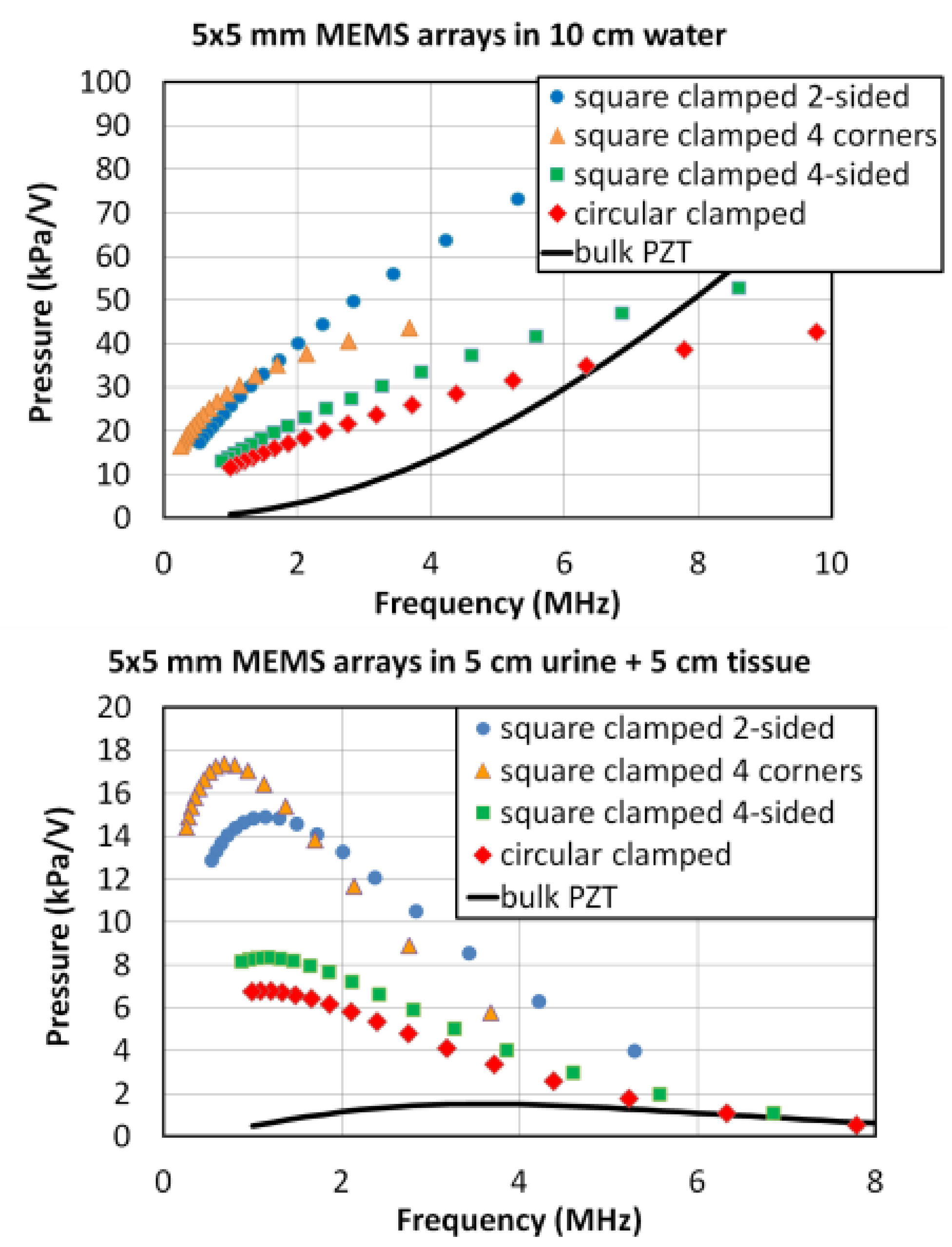
Resonance frequencies versus effective length for $T_{si} = 10\,\mu m$ clamped square MEMS, simulated (dots), modeled (lines) and measured with a Laser Doppler Vibrometer (squares).



Resonance amplitude versus effective diameter for circular MEMS, simulated (dots) and modeled (lines).

Transducer = MEMS array

An ultrasound transducer will consist of an array of MEMS diaphragms. We modeled the pressure generated by $5{\times}5\,mm^2$ arrays of our MEMS at 10 cm distance in water (similar acoustic properties as the human body) and, more realistic in half urine, half tissue with an attenuation of $0.5\,dB\,MHz^{-1}\,cm^{-1}$. We modeled the pressure at resonance, for four layouts with 80 – 240 µm diaphragms with 1.0 µm PZT and 8.0 µm silicon thickness. As a reference, we plot the pressure generated by a $5{\times}5\,mm^2$ bulk PZT5H transducer ($d_{33} = 550\,pm\,V^{-1}$).



Models for the pressure generated by four types of $5{\times}5\,mm^2$ MEMS arrays (symbols) with a $5{\times}5\,mm^2$ bulk PZT5H transducer as a reference (line).

Clearly, the attenuation in tissue has a major impact at higher frequencies. Thus we strive for as low as possible frequency of operation. However, the divergence of the beam increases at lower frequencies. These two constraints make that 3 – 4 MHz is the optimal operating frequency for our wearable bladder monitor. In the figure, we see that a square diaphragm clamped at four corners gives the highest pressure, but at too low frequency; this causes worse acoustic matching and strong divergence of the beam. The square diaphragm clamped on two sides appears to be optimal: fairly high displacement at 3 – 4 MHz, where the higher attenuation in tissue is compensated by the optimal acoustic performance.

Conclusions

We conclude that for the bladder monitor a piezoelectric MEMS transducer is preferably made of an (square) array of square MEMS diaphragms, clamped at two sides. This conclusion is based on a combination of finite element simulations (COMSOL) and thin-plate analytic modeling, partly verified by optical measurements on the MEMS.

*L.E. Kinsler, A.R. Frey, A.B. Coppens, and J.V. Sanders., Fundamentals of acoustics. New York: John Wiley & Sons, 4th edition, 2000