Detection of Photoacoustic Transients Originating from Microstructures in Optically Diffuse Media such as Biological Tissue

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Abstract—The generation and detection of broadband photoacoustic (PA) transients may be used for on-axis monitoring or for imaging of optically different structures in the interior of diffuse bodies such as biological tissue. Various piezoelectric sensors are characterized and compared in terms of sensitivity, depth response, and directivity with respect to spherical broadband acoustic pulses. The influence on the sensor output of acoustic interference and refraction of the PA transients at the sample-sensor interface is discussed. Ring detectors are suitable for deep on-axis detection thanks to their strong directional sensitivity, and small disk sensors are most suited for 3-D imaging of microstructures such as the (micro)vascular system. Voltage and charge preamplification schemes are compared in terms of the signal-to-noise ratio (SNR). In all cases, the preamplifier noise turns out to be the limiting factor for the sensitivity. Based on experimental data, for several sensor types and optical wavelengths, the theoretical detectability of PA signals generated by blood-like absorbers in biological tissue is discussed.

I. INTRODUCTION

PHOTOACOUSTIC (PA) signal generation [1]–[7] may be used to determine tissue characteristics [8]–[15] or to construct tomographic images of biological tissues [16]–[18]. The technique is based on the generation of a pressure pulse by an optical absorber following pulsed irradiation. The PA pulses may have a bandwidth up to more than 50 MHz [7]; [19]–[21]. Detection of this acoustic response allows the possibility of mapping optical characteristics of the interior of the sample. For medical applications, the imaging of the (micro)vascular system, tissue depth profiling, or PA spectroscopy in a well-defined volume would be of utmost interest.

In principle, a wide variety of techniques is available for the detection of acoustic transients, based on the detection of the displacement, the velocity, or the density of particles in the medium. Here, only the most important techniques are briefly discussed. Piezoelectric detection turns out to be a very suitable technique for the detection of PA signals in tissue as with ultrasound pulse echo imaging. An important difference, however, is that the piezoelectric elements in the pulse echo technique are used for transmission and detection of the acoustic pulses, and, in PA imaging,

they are only used for detection. Characteristics of piezoelectric detection will be discussed, and some transducers that have been constructed and used for PA detection and imaging are characterized. We will focus on detection of PA signals with the purpose of depth profiling and 3-D imaging of microstructures in tissue. This requires reconstruction of the acoustic wave fronts and, thus, the use of small sensing areas. It turned out that for the purpose of PA measurements in tissue, commercially available detectors were not sensitive enough. This was mainly due to an insufficient tuning of the preamplification characteristics to the required frequency bandwidth of the PA signals. For these reasons, we decided to undertake a thorough investigation into the signal processing characteristics of PA detection. Signal-to-noise ratios (SNR) had to be reconsidered. As a result, we were forced to design our own line of probes. The result of this investigation is presented subsequently. As a general conclusion, we state that, provided proper care has been taken of PA signal processing and a proper mix of requirements of resolution, dimensions, depth, preamplification, and noise can be obtained, PA array detection for 3-D imaging of structures in tissue is very feasible.

II. DETECTION OF PHOTOACOUSTIC ULTRASOUND

Common techniques for the detection of acoustic waves at the surface of a sample are based on piezoelectric, capacitive, electromagnetic, or laser interferometric principles [22]. For PA imaging of tissue structures, we need broadband sensors (>50 MHz) signals with amplitudes in the kiloPascal range (corresponding with displacements in the picometer range) [7], [19]. To obtain a high lateral resolution, the sensors should approximate point detection (spot diameter <100 μm) at the tissue surface.

Laser interferometers or piezoelectrics are the most suitable sensors for PA tissue sensing. Interferometers as well as piezoelectric sensors can be built to have a flat response less than 1 up to 100 MHz [23], [24]. Broadband piezoelectric transducers may be up to about 30 dB more sensitive than laser interferometers covering a comparable bandwidth (e.g., 10 MHz) [22], [25]. For rough and optically poor reflecting surfaces such as tissue, the difference will even be significantly larger. A piezoelectric point-contact detection threshold of about 0.2 pm with a detection bandwidth of 10 MHz has been reported [23], which should be

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compared with the detection threshold for interferometric detection [22] of several picometers. From a sensitivity point of view, piezoelectric detection seems to be favorable, but, for the best lateral resolution, the laser interferometer is the best option thanks to its point-contact character. The higher sensitivity and ease of construction and operation of piezoelectric sensors make these the best choice for our purpose, although interesting results using interferometric detection of ultrasound have been reported recently [26]. Measurements with several prototypes of a PA probe intended for medical imaging and based on piezoelectric detection have been reported [8], [14], [18].

The piezoelectric effect is described by a combination of the electrical relationship $D = \epsilon E$ and the mechanical relation $S = sT$ of the material, where D is dielectric displacement, ϵ is permittivity, E is electric field strength, S is strain, s is compliance (stiffness coefficient), and T is applied stress. The equations can be combined in many ways to obtain a set of piezoelectric constitutive relations [27]. For ultrasound detection, often the piezoelectric voltage constant g and charge constant d are used with the relations:

$$D = \epsilon E + dT \quad \text{or} \quad E = D/\epsilon - gT \quad (1)$$

where $d = \epsilon g$. The coefficients g , d , ϵ and s are tensors and are generally characterized by two subscripts to indicate the components of the tensors.

An important group of piezoelectric materials are the piezoelectric ceramics, e.g., the sintered ferroelectric ceramic lead zirconium titanium oxides (PZT), consisting of small ferroelectric domains. Another important class of piezoelectric materials are those that can be deposited as a thin film on a substrate, e.g., zinc oxide (ZnO). Also, ferroelectric polymer films have been developed, e.g., polyvinylidene difluoride (PVF₂). Generally, piezoelectric polymer films are regarded as the materials preferred for hydrophones (detectors of acoustic waves in liquids) because of their large g constant. Ceramics generally have a high piezoelectric stress constant and would be preferable for ultrasonic generation. Quartz has a high quality factor and is used for frequency stabilizers for example. PVF₂ [28] is attractive for medical imaging because its acoustic impedance closely matches the impedance of tissue, and it is mechanically flexible and better resistant to mechanical shock than ceramics [29]. Spot-poled membrane polymer hydrophones are highly desirable as detectors in pulsed ultrasound measurement studies because of their broadband non-perturbing properties [30] and have been presented as suitable sensors for ultrasonic dosimetry [31]. Polymer hydrophones with a diameter as small as 0.1 mm have been made [32]. Studies on the directional response of miniature hydrophones revealed large differences between individual hydrophones (polymeric as well as ceramic) and the measured values could not be related satisfyingly to several theoretical models [33]. However, all of the models were based on diffraction theory, and acoustic reflection and refraction were not taken into account.

A 9- μm thick PVF₂ membrane hydrophone has been reported for use in the 1- to 100-MHz frequency range [24]. In this range, the response was concluded to be flat within ca. 5 dB, which makes this film suitable for high resolution PA tissue imaging. The coefficient g was concluded not to depend on the frequency, but d and ϵ do vary (similarly) with frequency. This would make voltage detection more suitable than charge detection for broadband signals, as long as the output does not depend on the sensor capacitance. If the latter is not fulfilled, this may result in loss of sensitivity with increasing frequency [34]. However, experimental data in the range from 0.5 to 100 MHz showed a smooth and flat frequency response within 3 dB [35]. Often PVF₂ sensors are used with a voltage amplifier. Signal amplification schemes have been discussed [36], but no general conclusions could be drawn. For a 0.5-mm diameter PVF₂ needle hydrophone (thickness, 9 μm), a sensitivity of 30 nV/Pa has been reported [37]. However, to find the detection limit, not only the sensitivity, but also the noise level or the noise equivalent pressure P_{en} , is needed. Often, the P_{en} in a 1-Hz band is used as (a measure for) the minimum detectable pressure. It is related to the sensitivity η (V/ μPa), the measured noise voltage E_n (V), and the measurement bandwidth Δf (Hz) according to the relation [30]:

$$20 \log(P_{en}) = 20 \log \left(\frac{E_n}{\eta \sqrt{\Delta f}} \right) \quad [dB \text{ re } 1 \mu Pa]. \quad (2)$$

The measurement bandwidth was 300 kPa, so the total noise equivalent pressure was 550 Pa. We have obtained much smaller P_{en} values, but with a comparable total noise equivalent pressure, for a measurement bandwidth of about 65 MHz. Because the spectral noise distribution often is far from white, this is not a reliable way to predict the minimum detectable pressure for other measurement bandwidths. Not only the SNR and the absolute sensitivity are of importance, but also a good pulse response. This is much more easily obtained with polymers than with ceramics.

III. PIEZOELECTRIC THIN-DISK DETECTION OF BROADBAND PA SIGNALS

PA imaging requires small sensors to achieve a high resolution. The thickness has to be small enough to have an acoustic transit time that is smaller than the duration of the signal. The diameter has to be small enough to avoid destructive acoustic interference across the surface of the detector. A sensor may be characterized by its depth response and its directivity for a standard broadband signal. Depth response expresses the dependence of the sensor output amplitude on the on-axis detection distance. The directivity expresses the dependence of the sensor signal amplitude on the angle of incidence and, thus, on the relative translation of the source perpendicular to the sensor axis. The depth of the source is a parameter for the directivity. A measure for the directivity is the half width at

half maximum (HWHM) of the sensor output amplitude as a function of translation.

For a thin-disk piezoelectric sensor for which only the thickness mode is relevant (neglecting 1,3- and 2,3-coupling, where 1 and 2 are the transverse directions and 3 is the axial direction) and with a thickness much smaller than the characteristic wavelength of the acoustic signal, the response is determined by the normal component of the transmitted signal. The time-dependent piezo voltage $V_p(t)$ and surface charge $Q_p(t)$ can be obtained by averaging the normal component of the transmitted pressure over the detector surface:

$$\begin{aligned} V_p(t) &= \frac{g_{33}t_p}{A} \int P(\mathbf{r}, t) T_A(\mathbf{r}) \cos \theta_2(\mathbf{r}) dA; \\ Q_p(t) &= d_{33} \int P(\mathbf{r}, t) T_A(\mathbf{r}) \cos \theta_2(\mathbf{r}) dA \end{aligned} \quad (3)$$

where t_p (m) is the thickness of the disk; A (m²) is the sensor surface area; P (Pa) is the pressure and T_A is the amplitude transmission, both at the sensor surface; and θ_2 is the angle of the refracted wave in the sensor. g_{33} and d_{33} are the thickness mode piezoelectric constants. If coupling between the radial and thickness modes is taken into account, the response is reduced because the 3,1- and 3,2-coupling coefficients have inverse polarity compared with the 3,3-coupling. Because, in most cases, the lateral dimensions of the sensor are much larger than the wavelength of the acoustic signal, this contribution is expected to be small.

The voltage increases with the sensor thickness as long as the sensor is thin compared with the wavelength of the signal in the sensor. According to the superposition principle, every PA response can be regarded as a composition of point-source responses. A small spherical source generates a bipolar signal that is characterized by the peak-peak duration τ_{pp} . The maximum sensor thickness is approximated by

$$t_{\max} = \frac{1}{2} \nu_d \tau_{pp} \quad (4)$$

where ν_d is the acoustic phase velocity in the piezoelectric medium along the poling axis, and it is assumed that significant reflection at the backing material may occur. With this thickness, a bipolar PA pulse will broaden somewhat during the detection, but τ_{pp} of the sensor signal is the same as for the PA pulse itself. If the backing material is acoustically matched, the thickness can be two times larger.

Expression (3) for the sensor signal shows that the output amplitude will be related to the angle of incidence, the sensor diameter, and spatial length of the acoustic transient $P(\mathbf{r}, t)$. Upon increasing the angle of incidence, the average pressure on the sensor is reduced, and, thus, the sensor shows enhanced forward sensitivity. But also, for on-axis detection, the interference can be significant and may influence the depth response. For point detection (i.e., for vanishing capacitance), the total charge vanishes, but the voltage is at a maximum; for large sensors caused by

interference, the total charge is at a maximum, but the signals will be severely distorted, and the voltage will drop. A compromise has to be found. This will be illustrated subsequently. To maximize the average surface charge, the phase difference between the signals arriving at the center and at the edge of the sensor should be smaller than $\tau_{pp}/2$. For on-axis detection, this defines the maximum sensor radius $R_{\max}$ for maximum charge generation with limited signal distortion:

$$R_{\max} = \sqrt{\nu \tau_{pp} z} \quad (5)$$

where ν (m/s) is the acoustic phase velocity in the sample and z is the depth at which the source on-axis is located. The minimum depth for a source to be detected without signal loss determines the maximum sensor radius. As an example, for the optimal detection of a spherical PA source emitting a signal with $\tau_{pp} = 20$ ns located at 1 mm depth in water, $R_{\max} = 170$ μ m.

To be able to choose a proper detector diameter, we performed a series of simulations of the on-axis detection, considering acoustic interference. The Gaussian spherical source model [6], [7] was used with $\tau_{pp} = 20$ ns. The maximum signal amplitude and the least distortion of the piezo voltage V_p is obtained for the smallest sensors. This has, however, the disadvantage that, at the same time, the surface charge Q_p is reduced and will vanish for point detection. Assuming short acoustic pulses ($\nu \tau_{pp} \ll R$) for detection distances smaller than the sensor diameter, V_p and Q_p become independent of depth. With increasing sensor diameter, Q_p reaches a maximum, but V_p will further decrease because it is proportional to the average pressure. Without much signal distortion, the maximum surface charge and the maximum piezoelectric voltage are obtained with a detector radius of 0.1 mm for a detection distance of 1 mm. This corresponds very well with the value of 122 μ m calculated with (5) and shows the applicability of this relationship.

However, this optimization is not the only criterion for PA detection because, for PA imaging, the sensors should be as small as possible to obtain maximum resolution. This conflicts with maximum charge generation, and a compromise has to be found.

IV. CAPACITIVE SIGNAL LOSS

Because of the small sizes of the piezoelectric element, its capacitance C_p may be very small. The signal amplitude V_p is reduced because of the load capacitance, which consists of the stray capacitances C_s and the input capacitance C_i of the operational amplifier [Fig. 1(a)]. For voltage detection, this results in an effective input voltage V_d at the amplifier input. In case the piezoelectric material is coupled capacitively, this coupling capacitance C_c , which reduces the capacitance of the sensor, should be taken into account, and the effective capacitance of the piezoelectric element should be used, given by $C_p C_c / (C_p + C_c)$. This is important for sensors with a small C_p . The presence of

Fig. 1. Preamplifier configurations for piezoelectric detection of acoustic signals. A) Voltage amplification; B) charge amplification. Symbols: V_p = piezo voltage caused by bound charges; C_p = capacitance of piezoelectric sensor; C_s = stray capacitance; C_i = input capacitance of OpAmp; C_g = capacitance to reduce DC-gain; C_f = feedback capacitance; C_{f1} , C_{f2} = filter capacitances; R_s = shunt resistance; R_i = input impedance of OpAmp; R_g = gain resistance; R_f = feed back resistance; and R_{f1} = filter resistance.

$$\left(\frac{V_{out}}{V_p}\right)_{CA} = -\frac{j\omega R_f C_p}{1 + j\omega R_f C_f} \rightarrow -\frac{C_p}{C_f} \text{ for } \omega \gg \omega_0; \quad (8)$$

$$\left(\frac{V_{out}}{V_p}\right)_{VA} = \frac{j\omega R_i C_p}{1 + j\omega R_i C_i} \frac{1 + j\omega(R_f + R_g)C_g}{1 + j\omega R_g C_g} \rightarrow \frac{C_p}{C_i} \left(1 + \frac{R_f}{R_g}\right) \text{ for } \omega \gg \omega_0 \quad (9)$$

the other capacitances causes an attenuation of the signal amplitude (above the cut-off frequency) given by

$$\frac{V_d}{V_p} = \frac{C_p}{C_p + C_s + C_i}. \quad (6)$$

For small sensor capacitances, this signal loss can be significant. As an example, a 0.2-mm diameter PVF₂ sensor ($\epsilon_r = 12$) with a thickness of 9 μm is taken. The capacitance is 0.37 pF, and, with $C_s = C_i = 2\text{pF}$, this results in an attenuation by a factor of 12. In case of charge amplification (CA), there is not such a signal attenuation, but then the gain is limited by C_p as is shown later.

For several piezoelectric materials [PVF₂, P(VF₂-TrFE), PZT-4, PXE-42] [27], [38]–[41], the sensitivity η_v of a disk sensor was calculated, assuming detection in an aqueous medium, according to

$$\eta_v = g_{33} t_{\max} T_1 R_2 A_C \quad (7)$$

where T_1 is the normal amplitude transmission at the water-sensor interface, R_2 is the amplitude reflection at the sensor-backing interface, and $A_C = C_p/C_i$ is the capacitive voltage attenuation with $C_i \approx C_p + C_s + C_i$. Stainless steel is taken as the backing material, but other materials that acoustically match to the sensor can be taken as well. The sensor thickness is optimized according to (4). A stray capacitance of 2 pF and a preamplifier input capacitance of 1 pF was assumed. For 0.1 to 0.5-mm diameter transducers, η_v varies between 0.2 and 3 $\mu\text{V}/\text{Pa}$ for PVF₂ and varies between 0.6 and 1.4 $\mu\text{V}/\text{Pa}$ for PZT.

The charge generation is proportional to the capacitance of the sensors, i.e., proportional to the surface area. Charge generation in the ceramics is much larger than for quartz or PVF₂. Because the dielectric constant of the ceramics is much larger, the piezoelectric voltage is strongly reduced. The polymers show the largest intrinsic voltage sensitivity as given by g_{33} [see (1) and (3)].

For the large sensor diameters, the effective output voltage of the polymer films is higher than that for the other piezos, but, for the smaller diameters, the capacitive loss is increasingly important, and the ceramics show better performance. With respect to the charge generation, the ceramics are superior.

Piezoelectric ceramics have also some drawbacks. Because of the large impedance mismatch with water, they tend to ring. This can be avoided if the backing is acoustically matched. At the sample-transducer interface, a large part of the signal is reflected back into the sample. This may be complicating if it leads to significant reflections in the sample that are also detected. Use of acoustically matching layers may reduce this problem, but the construction of the layers for broadband transducers is difficult to accomplish. For piezoelectric polymers, the reflections can be suppressed easily by matching. Another advantage of the polymers is the small thickness of the sensors, which makes them less sensitive for radial forces. As a result, the transducer can be regarded as a thin disk operating in thickness mode, in contrast to ceramic sensors with optimized dimensions for which the thickness approximates the lateral dimensions.

V. SIGNAL PREAMPLIFICATION

In principle, two preamplification schemes are possible: voltage amplification (VA) and CA. Signal preamplification for small hydrophone elements is not discussed much in literature. Sometimes facts are mentioned that are only relevant for voltage amplification (e.g., [35]). Often VA is assumed or presented as the only possibility [36]. In other cases, some advantages and disadvantages of both amplification schemes are mentioned but without thorough discussion of small signal detection or a comparison of SNR

[42]. CA has the advantage that the output is independent of the cable capacitance and stray capacitances, but the disadvantage that the noise level is directly proportional to the load (wiring) capacitance. A disadvantage of VA is the voltage coupling loss that results from a small sensor capacitance relative to the load capacitance. In a comparison of the two types of preamplifiers [42], it was demonstrated that a high sensitivity and flat frequency response can be achieved by using CA, but, in fact, there was little difference in the minimum detectable pressure level for the two amplifiers with short wiring.

The electrical circuits for VA and CA are shown in Fig. 1. Apart from the filters at the output of the operational amplifiers that are used to reduce the noise at low frequencies (below 1 MHz), the transfer functions are given by (8) and (9) (see top of previous page)

where $C_t = C_p + C_s + C_i$, $R_t = R_s R_i / (R_s + R_i)$, the high pass cut off frequency $\omega_0 = 1/R_f C_f$ for CA, and ω_0 is the maximum of $1/R_f C_f$ and $1/R_g C_g$ for VA. The capacitance C_g is used to reduce the offset at the output of the OpAmp, which results from the bias current at the non-inverting input. We see that, with CA, the signal amplification is given by $G_{CA} = C_p/C_f$, i.e., the ratio of the piezo capacitance and the feedback capacitance. For very small sensor capacitances (<1 pF), it is not possible to amplify the signal because there will always be a feedback stray capacitance of about this value. With VA, the signal is amplified with the amplifier gain $G_{VA} = (R_g + R_f)/R_g$ but also attenuated by the capacitive quotient C_p/C_t . For the relevant signal frequencies, the relative signal amplitudes with the two amplification schemes are given by

$$\frac{S_{VA}}{S_{CA}} = \frac{C_f}{C_p + C_s + C_i} G_{VA}. \quad (10)$$

This shows that the voltage amplifier gives a higher response than the charge amplifier if $G_{VA} > (C_p + C_s + C_i)/C_f$. As long as the output signal amplitude is not too small to make further amplification necessary, it is the SNR of the preamplifier that determines the detection limit.

The major noise sources in VA are the thermal resistance noise current $I_{n,R}$ from R_t , the noise current $I_{n,+}$ at the non-inverting input, and the noise voltage U_n of the OpAmp [43]. Although the detection circuit before the input of the OpAmp is a high pass filter for PA signals, it is a low pass filter for the noise generated by $I_{n,+}$ and $I_{n,R}$. A high pass filter at the output of the OpAmp effectively reduces this noise. It is taken as a second-order filter with time constant RC . Using (9), the amplitude SNR is

$$SNR_{VA} = \frac{\left(\frac{C_p}{C_p + C_s + C_i} \right) V_p}{\langle V_n \rangle_{VA}}. \quad (11)$$

The nominator is equal to η_v . For $C_p \ll (C_s + C_i)$, the signal is proportional to C_p , and $\langle V_n \rangle$ decreases with increasing C_p . In this case, the best SNR is obtained with a ceramic sensor. The choice of R_t for a given C_t is not obvious and depends on the relative noise contributions. The

Fig. 2. The noise equivalent piezo voltage as a function of the piezo capacitance for voltage and charge amplification. The following parameter values were used: $C_i = 2$ pF, $C_s = 2$ pF, $C_f = 1.4$ pF, $f_0 = 1$ MHz, $U_n = 1$ nV/ $\sqrt{\text{Hz}}$, and $I_{n,-} = I_{n,+} = 1.6$ pA/ $\sqrt{\text{Hz}}$. The integral is taken from 0 to 100 MHz.

contribution to the final detector noise of $I_{n,+}$ is minimized by minimizing R_t ; the contribution from the thermal noise is minimized by maximizing R_t (i.e., shifting the thermal noise to the lower frequencies). The lower limit of R_t is determined by the RC -time needed for detection of the PA signals (ca. 1 MHz).

The major noise sources of the charge amplifier are the thermal noise from R_f , the noise current $I_{n,-}$, at the inverting input, and the noise voltage U_n of the OpAmp [43]. Of the three noise sources, only U_n is amplified. Using (8), the SNR is

$$SNR_{CA} = \frac{(C_p/C_f)V_p}{\langle V_n \rangle_{CA}}. \quad (12)$$

With increasing C_p , the SNR increases until, at a large amplifier gain ($C_p \gg C_f$), the output noise is dominated by U_n and also becomes proportional to $G_{CA} = C_p/C_f$. For CA too, it is advantageous to use a ceramic as sensor material.

A useful quantity is the noise equivalent piezo voltage, i.e., the piezoelectric voltage that generates a signal at the output of the detector with $SNR = 1$. This quantity is found using (11) or (12). A numerical evaluation of the noise equivalent piezoelectric voltage V_p is shown in Fig. 2 for both amplification schemes as a function of C_p . It was assumed that all of the time constants were equal to $1/(2\pi \times 10^6)$ s, corresponding to a frequency f_0 of 1 MHz. For low noise OpAmps with a sufficiently large bandwidth, the following parameter values are realistic: $U_n \approx 1$ to 3 nV/ $\sqrt{\text{Hz}}$, $I_{n,+} \approx 0.5$ to 2 pA/ $\sqrt{\text{Hz}}$, and $I_{n,-} \approx 1$ to $15 \times I_{n,+}$. The parameter values were chosen in these regions.

The figure shows that for very small sensor capacitances, the noise equivalent voltage (or the SNR) is nearly the same for the two amplification schemes. The performance of the charge amplifier is somewhat better. For the

TABLE I
DIMENSIONS AND PREAMPLIFICATION PARAMETERS OF THE CONSTRUCTED DETECTORS.

Type	I.D. (mm)	O.D. (mm)	Ampl.	C_p (pF)	$C_i + C_s$ (pF)	C_f (pF)	V_n rms (mV)	η_v ($\mu\text{V}/\text{Pa}$)
Disk		0.2	VA, $G = 10$	0.37	4		0.98	2.0
Small ring	0.37	0.67	VA, $G = 10$	2.9	3.7		0.33	
Large ring	2.6	3.0	CA, $G = 15$	20.8		1.4	0.25	

parameter values chosen, the maximum benefit of the CA is obtained for a sensor capacitance of ca. 30 pF. At this C_p value, the signal to noise amplitude ratio is ca. 1.5 times larger than that for VA. For larger capacitances, the benefit decreases again. So, in practice, there is hardly any difference between CA and VA with respect to the SNR. The preferable amplification scheme is determined by maximizing the signal amplitude as expressed by (10). For very small piezo capacitances, VA is preferred because of the higher signal, but, for larger piezo capacitances, CA is preferred because of the higher SNR. But what is important is that the limit of detection decreases approximately inversely proportional to C_p for $C_p < 10$ pF. Although the piezo voltage generated by polymers is 5 to 20 times larger than for ceramics, the capacitance is 40 to 130 times smaller. For sensors with a diameter below about 1 mm, ceramics have a better SNR.

VI. EXPERIMENTAL PIEZOELECTRIC HYDROPHONES

Three hydrophones with different characteristics have been used for PA measurements. For the numerical calculations of the performance of the detectors, using the Maple package [44], well-known expressions for the signal transmission at the water-piezo interface [27], for the piezo voltage (3), and for the preamplification (8), (9) are used.

Because of the ease of handling and the low acoustic impedance, the sensors were made of 9- μm thick biaxially stretched PVF₂ film. For the imaging experiments, a disk-hydrophone was used; for PA detection of sources at larger depths, two-ring detectors were constructed. For some experiments also, a 1-mm diameter PVF₂ sensor was used with a film thickness of 28 μm .

For the disk and the small ring detector, a voltage amplifier was found to be somewhat more advantageous; for the large ring detector, a charge amplifier resulted in the best SNR. The diameter of the disk detector was a compromise between sensitivity, or SNR, and the requirement of point detection for optimal imaging resolution. The dimensions of the ring detectors were chosen to obtain two different depth ranges where the sensors have their optimum sensitivity combined with a narrow aperture. The main characteristics of these detectors are summarized in Table I.

The construction of the sensor, consisting of four disk detectors, is shown schematically in Fig. 3. Four copper wires (diameter, 200 μm) are embedded in Stycast in a

Fig. 3. Schematic drawing of the PVF₂ disk detector. Copper wires (200 μm) are embedded in Stycast and are polished. The piezo film is glued on top of this surface, and the copper wires act as signal electrodes that determine the diameter of the sensor. The preamplifiers are mounted inside the housing.

grounded brass housing. The surface (perpendicular to the wires) is polished, and the piezofilm is glued on top of it, coupling the signal electrodes capacitively to the piezofilm. At the free surface, the film is metallized and connected to the brass housing by a conductive acrylate resin. This connection is shielded by a thin layer of silicon resin. The voltage preamplifiers, for which the low power video OpAmps AD810 and AD811 were used, are mounted inside the detector housing. The amplifier circuits act as a high pass filter to eliminate spectral noise contributions below ca. 1 MHz, at which point the spectral signal amplitudes are negligible. The bandwidth was always more than 65 MHz. For the charge amplifiers, as well as for the voltage amplifier, the noise was concentrated at the lower frequencies in quite good agreement with theory. The total noise, however, was experimentally about a factor 1.5 larger than the theoretical values. The disk detector has a sensitivity of 2.0 $\mu\text{V}/\text{Pa}$ as followed from extensive comparison with a calibrated needle hydrophone (Precision Acoustics Ltd., UK) with an rms noise amplitude of about 1 mV.

The ring detectors are constructed in the same way. A stainless steel tube is embedded in Stycast epoxy resin to replace the copper wires and to act as the signal electrode. Furthermore, after the attachment of the piezo film, in the large ring detector, a hole was drilled in the center of the metal tube, and a 600- μm core diameter glass fiber was mounted in the sensor. The fiber facet was located just above the sensor surface. The hole was closed with

silicon resin. For both ring detectors, the OpAmp CLC425 was used, which has somewhat better characteristics than the AD video OpAmps. The rms noise of the small ring detector was 0.33 mV and, for the large ring detector, was 0.25 mV.

VII. TRANSDUCER CHARACTERIZATION

The performance of piezoelectric thin disk sensors very much depends on the reflection, refraction, and interference across the interface with the sample. The PA signal generated by an extended source can be described as a summation of the PA signals from smaller elements (superposition principle), so the signal from a small spherical source is used for the transducer characterization. Fig. 4 shows some experimental PA signals that were detected with the 200- μm diameter PVF₂ transducer. The effective source geometries correspond with a homogeneous sphere, a homogeneous cylinder, and a cylindrical shell. Intralipid dilutions and Evans Blue solutions were used to mimic tissue optical characteristics. Also, the PA signal generated by real tissue containing artificial vessels with whole blood is shown. This is the kind of signal that is used for 3-D PA imaging of tissue samples. For the characterization of the transducers, we use the depth response and the directivity of the sensors, defined as the amplitude maximum of the response for the detection of a bipolar acoustic pulse as generated by a spherical PA source [7]. The source is characterized by a τ_{pp} of 10 to 20 ns. The depth response is the signal maximum as a function of the depth of the source for on-axis detection. The directivity is the set of signal maxima plotted as a function of the lateral displacement of the source with the depth as a parameter and is normalized with the signal at normal incidence.

Measurements have been performed to reveal the relative depth responses of the various detectors. This is shown in Fig. 5. The experimental setup has been described in large detail elsewhere [18]–[21]. We used a Nd:YAG laser at 532 nm with 10-ns pulses of about 100 μJ each. As the turbid medium, we applied either a 10% Intralipid-10% dilution or real chicken breast tissue. As the absorber, we used either whole blood or an Evans Blue solution with similar absorption. A 10- μm diameter absorbing thread irradiated over a small length acted as the PA source. Theoretically, this signal has a τ_{pp} of 12 ns. Typically, a τ_{pp} between 13 and 23 ns was measured, the broadening of which may be partially due to frequency-dependent absorption in water [7]. In the calculations, the theoretical value was used. Because the source strength was not well reproducible, the experimental depth responses of the three detectors were normalized to correspond with the theoretical curves. The figure shows the relative large sensitivity of the ring detectors at a larger depth. For a depth $z > 6$ mm, the large ring detector has the highest sensitivity. At distances close to the surface ($z < 1$ mm), the disk detector has the highest sensitivity, and, in between, the small ring detector is favorable. The significant deviations of the experimental

Fig. 4. Experimental PA signals detected with a 200- μm diameter piezoelectric transducer. The upper panel shows the signals (averaged 64 times) generated by spot illumination of a 10- μm diameter absorbing strand in water and by homogeneous illumination of this fiber in a 10% dilution of Intralipid (10%). Detection distance is 1.5 mm. The middle panel shows the PA signal from an Evans Blue solution with $\mu_a = 300 \text{ cm}^{-1}$ flowing through a translucent nylon vessel (inner diameter, 750 μm ; outer diameter, 940 μm), which is located in a 10% dilution of Intralipid (10%) and illuminated from aside with a glass fiber at a 1.5-mm distance. The dashed lines indicate the transit times from the nylon-water interfaces to the transducer. The pulse energy is 85 μJ , and the signal is averaged 16 times. The lower panel displays the signal recorded just in front of a 6-mm thick chicken breast sample that contains several 280- μm inner diameter nylon capillaries (at depths between 3 and 4 mm) with flowing whole blood. The pulse energy is 300 μJ ; the signal is averaged 16 times.

sensitivities at small depths from the theoretical curves are probably due to the fact that the source actually is a short line source. For the ring detectors, a better preamplifier was used than with the disk detector. Using the same OpAmp for this detector will increase the performance by about a factor 1.5. The results clearly show the advantage of the higher sensitivity of ring detectors for deep measurements.

The response of a detector depends on the angle of incidence, the signal shape, the duration, the detection distance, and the diameter(s) of the detector. Together, these parameters determine the normally transmitted signal amplitude and the acoustic interference. For the different detectors, the sensitivity for lateral translation of a PA source is quite different. These relationships have been determined experimentally and theoretically (by numerical evaluation). A 10- μm thick absorbing fiber illuminated over 70 μm acted as the PA source, generating a bipolar PA signal with $\tau_{pp} = 13$ ns. For the three detectors, the source was located at different depths: for the disk detector, at 1 mm; for the small ring detector, at 3 mm; and for the large ring, at 10 mm. The result is shown in Fig. 6.

Fig. 5. Comparison of the depth response of a disk detector (O.D., 0.2 mm), a small ring detector (0.37 mm $\times$ 0.67 mm), and a large ring detector (2.6 mm $\times$ 3 mm) for a spherical source. Calculations with $\tau_{pp} = 12$ ns. The experimental data are normalized to fit to the calculated curves.

The figure shows that all of the detectors have about the same HWHM for the signal amplitude as a function of the relative translation (disk, 0.2 mm: HWHM = 0.2 mm; small ring, 0.37 $\times$ 0.67 mm: HWHM = 0.25 mm; large ring, 2.6 $\times$ 3 mm: HWHM = 0.2 mm), but this is obtained at very different depths. The figure shows that a larger ring has a (much) more pronounced forward sensitivity. Ring detectors, however, show side lobes in their directional sensitivity at a translation equal to the radius of the ring, and the signal shape is distorted. For line sources, this effect is even much more pronounced, and the side lobe can have a higher amplitude than the sensitivity for on-axis detection. If concentric rings are used for the detection, this phenomenon can be reduced. This is shown in Fig. 7. The HWHM for the double ring detector is 60 μ m (for a source at a 5-mm depth), and, for the single ring detector, the HWHM is 76 μ m. The directivity of the double ring detector is the result of the sum of the individual ring signals. Using time shifts may further increase the forward sensitivity. With multiple concentric rings, a very strong forward sensitivity can be obtained with a variable depth of focus. This may be used for deep on-axis monitoring of PA signals in tissue. Because of the large diameters, these detectors are not very suitable for integration in a 3-D imaging sensor, unless the detector is scanned across the skin surface.

VIII. DETECTION LIMITS

The relative capabilities of the three different hydrophones to detect a spherical PA source in an optically tissue-like medium as a function of the depth of the source is shown in Fig. 8. The source, with a diameter of 370 μ m, has blood-like absorption. Experiments with a cylindrical source of this diameter in an Intralipid dilution with added

Fig. 6. Directional sensitivities of the large ring detector (LR), the small ring detector (SR), and the disk detector. The normalized signal maximum is given as function of the translation of the point source relative to the detector. The source was located at $z = 1$ mm for the disk detector, 3 mm for the small ring detector, and 10 mm for the large ring detector.

Fig. 7. Directional dependence of a single ring detector (2.6 mm $\times$ 3 mm) and two concentric rings (2 mm $\times$ 2.1 mm and 3 mm $\times$ 3.1 mm) with a point source at a 5-mm depth as a function of the relative translation. The experimental data are for the single ring detector. Calculations are for $\tau_{pp} = 12$ ns.

absorber (Evans Blue) to obtain a reduced scattering coefficient μ'_s of 0.75 mm^{-1} and an absorption coefficient μ_a of 0.1 mm^{-1} (corresponding to tissue) revealed that, for a 180- μ J laser pulse at 532 nm with the large ring detector, the PA signal, averaged 128 times, was detectable at a depth of 7.8 mm with a power SNR of 10. The rms noise was about 8.8 mV. For other optical parameter values, the diffusion approximation for an incident pencil beam is used to approximate the light fluence and to calculate the appropriate SNR values. For a spherical source with the same diameter as a cylindrical source, the PA signal is about five times smaller. These data were used to determine the detection limit for several configurations. The theoretical

Fig. 8. Theoretical detectability of a PA signal from a 0.37-mm diameter spherical source with blood-like absorption located in phantom tissue for three different detectors. Detection is in reflection mode. The calculations are for 1-mJ laser pulse energy and for signals averaged 128 times. The optical coefficients of the medium are given in Table II.

depth responses of the three detectors were used to find the relative sensitivities. The noise of the amplifiers is assumed to be the same. The absorbed energy E_a is proportional to μ_a , so the signal at 800 nm ($\mu_a = 5 \text{ cm}^{-1}$) is about 60 times smaller than that at 532 nm ($\mu_a = 300 \text{ cm}^{-1}$) for equal laser pulse energies reaching the target. This effect will be partly compensated by decreased scattering upon increasing wavelength. The optical parameters for bloodless tissue were taken as shown in Table II. The absorption is likely to be about a factor 3 too large compared with in vivo measurements in tissue and should be regarded as worst case.

Parameters that may be changed are the laser pulse energy E , μ_a , and μ'_s of the medium, and the detection limit. The latter depends on the SNR characteristics of the detectors. Reducing the noise of a detector with a factor 2 or increasing the sensitivity of the sensor with a factor 2 (while keeping the noise constant) will shift the curve upward with the same factor. The same holds for an increase in light pulse energy. If the optical properties differ from the ones used here, the curves are changed with a depth-dependent factor that is defined by the change in the fluence:

$$\frac{\Phi}{\Phi_0}(z) = \frac{\mu_{tr}}{\mu_{tr,0}} \exp \left\{ -\mu_{eff}(z - 1/\mu_{tr}) \right\} / \exp \left\{ -\mu_{eff,0}(z - 1/\mu_{tr,0}) \right\} \quad (13)$$

where the subscript 0 refers to the coefficients used here (see Table II); $\mu_{tr} = \mu_a + \mu'_s$ and $\mu_{eff} = (3\mu_a\mu_{tr})^{1/2}$.

The figure shows that for detection distances larger than 0.5 mm, the small ring detector has higher sensitivity than the disk detector, and, for depths larger than 5 to 6 mm, the large ring detector has the highest sensitivity. For the given assumptions with the disk detector, the spherical source can be detected with 532-nm light up to a depth of 6 mm. For the ring detectors, this is ca. 8 mm. At larger depths, the ring detectors are increasingly favor-

able, but, to detect the source, the parameter values have to be changed to get the curves above the detection limit. At 800 nm, the detectors show the same relative behavior, but the SNR of a measurement does not decrease as fast with increasing depth as at 532 nm. This is the result of the larger effective optical mean free path. In principle, it should be possible to increase the light energy until the absorbed energy near the tissue surface is equal for the various wavelengths as long as there is no cavitation or plasma formation. Then, the curves for 532 nm and 800 nm start at the same value for small depth, and the source can be detected up to about 20 mm with the large ring detector (at 800 nm) if the other parameters are kept constant. If, at 800 nm, a 10-mJ laser pulse is used (i.e., 10 times larger than at 532 nm), then, at a depth of 6 mm, the SNR of the sensors are the same for both wavelengths, and, at larger depths, the near infrared light is favorable.

For cylindrical sources, the PA signal decrease caused by the acoustic propagation is proportional to $r^{-1/2}$, in contrast to the r^{-1} dependence for spherical propagation. For cylindrical sources, the curves in the figure are shifted upward with a factor $(r/R)^{1/2}$, where r is the detection distance (depth) and R is the radius of the source.

IX. CONCLUSIONS

In the introduction, we stated that, for applications with tissue, we had to reconsider the signal processing of PA detection. We have seen that the sensitivity of the piezoelectric PVF₂ hydrophones is limited primarily by the noise sources of the operational amplifier that is used for preamplification of the sensor signal. There is not much difference between VA and CA, but, for the OpAmps used, the SNR is somewhat larger with CA than with VA. The maximal difference was found for a sensor capacitance of 30 pF. For sensors with a small capacitance, VA was preferred because of the higher output signal, reducing the need of further signal amplification. The noise equivalent pressure amplitudes of these hydrophones are lower than those of commercially available needle hydrophones or than the values reported in literature.

The influence of acoustic interference (spatial pressure averaging) on the hydrophone output was demonstrated, and criteria for the dimensions of the disk detector were developed. This detector has a large angle of acceptance. The ring detectors have a much more forward peaked sensitivity and may, therefore, be used for on-axis monitoring, especially if a number of concentric rings is used to reduce side lobes.

Because of the broadband characteristics of polymer films and because of the easy handling of these films, PVF₂ was used as the sensor material. From the curves for the detectability of 0.4-mm diameter spherical PA sources in tissue with blood-like absorption, for various hydrophones, the following SNR parameters for PA detection can be derived. At a depth of 2 mm, typical SNR values range between 40 and 200 for 532 nm and between 0.7 and 3

TABLE II
OPTICAL PROPERTIES USED TO CALCULATE THE CURVES IN FIG. 8.

Wavelength (nm)	μ_a (mm ⁻¹)	μ'_s (mm ⁻¹)	μ_{tr} (mm ⁻¹)	μ_{eff} (mm ⁻¹)	l'_t (mm)
532	0.1	1.5	1.6	0.69	0.63
800	0.03	0.75	0.78	0.27	1.28

for 800 nm; at a depth of 6 mm, these values range between 0.9 and 4 for 532 nm and between 0.1 and 0.5 for 800 nm. These values are for a laser pulse energy of 1 mJ. However, because of the much lower absorption of near infrared light in tissue, a much higher laser pulse energy may be applied, which results in a much higher SNR value for this case. We have shown that with a 200- μ m diameter disk, transducer microstructures with blood-like absorption in media such as tissue can be detected up to depths of 6 mm using 532-nm laser pulses. This transducer is very suitable for application in a matrix sensor for 3-D PA tissue imaging because of its small lateral dimensions and its higher SNR compared with commercially available needle hydrophones. For PA imaging at depths up to about 5 mm, the shorter wavelengths are preferred because of the large contrast and the high resolution that can be obtained; for depths larger than about 6 mm, near infrared light is preferred. Larger depths can be reached with larger sensor diameters and a larger piezoelectric film thickness, both at the cost of resolving power. The radiant fluence attenuation with depth was taken for a worst case (rather high absorption was assumed), and the maximum detection depth is likely to be larger than indicated by the calculated curves. For the detection of cylindrical sources, the dependence on the depth is reduced. In practice, with the appropriate parameter values, it should be possible to detect sources with a diameter of several tenths of a millimeter up to about a 2-cm depth without the use of large focusing transducers. For a spherical source with this diameter, the maximum depth of detection is about 1.5 cm under the same conditions. Broad beam irradiation may further increase the maximum detection depth.

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