A Feasibility Study on Aortic Pressure Estimation Using UWB Radar

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Abstract—Microwave radar based techniques have been proposed for medical applications such as heart beat and respiration monitoring, and for breast cancer detection. This paper investigates the feasibility of using microwave radar techniques to estimate aorta diameter variations. A survey of relevant medical literature shows that the aorta diameter variations can be used to estimate aortic compliance, pulse pressure and mean pressure.

This study is based on 2D simulations with a simplified geometry. The results provide for an upper bound on received power in the range (0.5, 5) GHz and show 40 dB loss at the lower end increasing to around 120 dB at the upper end, where material losses are dominant. Furthermore, notches related to aorta diameter are observed in the spectrum and may be used in a frequency-domain approach to estimation.

In the time domain, simulations indicate that two echos may be identified and associated with the aortic diameter.

In future work, we are planning to conduct further simulations with complex, realistic geometries.

I. INTRODUCTION

Non-invasive measurements of blood pressure (BP) exist such as sphygmomanometer, photoplethysmograph [1], tonography [2], pulse transit time [3]; however they all rely on peripheral measurement points. This may constitute a problem in certain situations such as when flow redistribution to central parts of the body (heavy injury, temperature) degrades these measurements; another situation where central measurements may prove advantageous is in presence of strong movement of the peripheral locations and which affect pressure measurements [4].

An interesting overview of the use of radar for medical applications is proposed in [5] and which traces research back to the late 1970’s. It seems that renewed interest has been spurred following McEwans micropower impulse radar [6] in the early 1990s which combined ultrawide band (UWB) pulses with very low power, small size and low system cost. It also seems that some of this momentum in research related to UWB pulses has been founded on dubious claims of exceptional behavior related to the impulsive nature of the signal, such as specific penetration, resonances and presumed inadequacy of a Fourier type description; most claims have been refuted [7]. The research into medical sensor applications include apexcardiography, heart rate, respiration rate, heart-rate variability, blood pressure pulse transit time (peripheral locations) and associated applications such as through rubble or walls vital signs detection [6], [8], [5]. With respect to imaging, the use of an antenna array for the early detection of breast cancer [9], [10] should be mentioned.

This article’s contribution is a feasibility study by simulations using a simplified geometrical model of indirectly estimating blood pressure based on aorta diameter variations.

II. PHYSIOLOGICAL PROBLEM DESCRIPTION

There are two distinct approaches to estimating blood pressure based on aorta diameter variations: 1) using the linear relationship between percentage changes in instantaneous blood pressure and diameter, and shown for carotid artery pressure by Sugawara et al [11]; 2) estimating the elasticity of the aorta (local compliance or incremental elastic modulus) and relating this to blood pressure [3], [4], [12] without being explicit with respect to the functional relationship. In both approaches, the radar-based method will aim at detecting the aorta walls and estimate the diameter as a function of time (d(t)). In the second approach the key point is the relationship between the elasticity of a homogeneous, circular tube and the speed of propagation of a pressure pulse along the tube presented by Otto Frank in 1926 according to [13]:

\[ v = \sqrt{\frac{K_L}{\rho}}, \]

where, \( v \) is the speed of the pulse propagating along the aorta, \( K_L \) is the bulk elastic modulus per unit length and \( \rho \) the density of the liquid filling the tube (\( \rho_{blood} = 1.05 \text{ g/cm}^3 \)). Although the medical community is most familiar with ‘Compliance’ (\( dV/dP \)), where \( V \) is the arterial system volume and \( P \) its pressure, some authors operate with a local measure more appropriate for current needs: \( C_L = dA/dP = 1/K_L \). Here \( A \) is the cross-sectional area. Hence by estimating \( v \) for a given \( \rho \), \( C_L \) follows directly.

Moen-Korteweg’s Eq. (2) relates the incremental Young’s elastic modulus \( E_{inc} \) to propagation speed, aorta radius \( r \) and wall thickness \( h \):

\[ E_{inc} = \frac{\Delta P}{\Delta r/r} \frac{r}{h}, \]

\[ v = \sqrt{\frac{E_{inc} h}{2 \rho}} = \sqrt{\frac{1}{2 \rho} \frac{\Delta P}{\Delta r/r}}. \]

\( (r, \Delta r) \) provides sufficient information for estimating the pulse pressure \( \Delta P \) and \( E_{inc} \) is related to mean arterial blood...
pressure either through (1) or (2) [3], [4], [12]. This means estimating diameter variations facilitates measuring several clinical parameters such as mean arterial pressure ($P$), pulse pressure ($\Delta P$), aortic compliance ($C_L$) and heart rate (HR).

Several studies in the medical literature propose estimates of the aortic diameter variations using techniques such as magnetic resonance imaging, ultrasound, pressure sensors etc, with disparate results. Stefanidis [14] proposes a precise and invasive measurement method based on pressure and diameter sensors introduced through catheters. The article confirms a square-root relationship between the pulse wave velocity and a pressure-radius slope measurement ($C_L \approx 2 r \frac{\Delta P}{\rho R} \propto 1/v^2$) and mentions typical diameter peak-to-peak amplitudes of 2.18 ± 0.44 mm for a normal population. This means the measurement precision of the aorta diameter variations must be at a fraction of a millimeter or less - a strict requirement also for a radar based method.

III. SIMULATION MODEL

To understand the principles of using electromagnetic waves to measure diameter variations, a simulation model has been constructed for electromagnetic simulations. Our model combines a voxel representation of the human body [15], with the material electromagnetic properties proposed by Gabriel et al. [16]. Although there is not an evident mapping between all voxel categories and the tissue descriptions, the undefined tissues are not present in the region considered in this article and as such have no influence on what follows. With diameter variations on the order of 2 mm, a set of simulations with aorta diameter ranging from 20 mm to 26 mm in steps of 0.4 mm has been conducted in a simulation space with resolution 0.1 mm. The dimensions of the simulation space are shown in Fig. 1, which imply approximately 2.8 M sampling points. Due to the large number of points, a 2D simulation space was chosen. For the sake of simplifying analysis of the backscattered field from the aorta, the complexity of the geometry is reduced; a set of structures has been replaced by a homogeneous material named “average”, see Fig. 1. The “average” material is based on a plane wave propagating through a layered material and including the effect of each layer’s contribution to a total distance $x$ by scaling the respective properties by the thickness of that layer, $x_i/x$. Hence a homogeneous average material and the layered structure would differ by reflections, but would behave equivalently with respect to propagation. However, as multiple paths exist in the simulation space, average area rather than propagation path has been used for scaling individual materials.

$$\gamma_{true-avg} = \sum_{i \in M} A_i \frac{A_i}{A} \gamma_i = \sum_{i \in M} A_i \frac{\Delta \omega}{\omega} \epsilon_i \mu_i, \quad (3)$$

$$\gamma_{avg} = \omega \sqrt{\mu} \sum_{i \in M} A_i \frac{\Delta \omega}{\omega} \epsilon_i, \quad (4)$$

where $M$ is the set of materials, $A$ is the total simulation space area while $A_i$ is the area associated with material $i$, $\mu_i$ is the material permeability, $\gamma$ is related to the square-root of the complex permittivity which implies a complex mapping between the different poles of the material characteristics and the average propagation constant as given in (3). Instead of scaling $\gamma_i$, the permittivities $\epsilon_i$ have been scaled to avoid this complex mapping and expressed in (4); by considering Fig. 2, $\gamma_{avg}$ is close to the “true” average $\gamma_{true-avg}$ compared to the intrinsic variation between the represented tissue properties.

Finally, due to simulation tool limitations, the Cole-Cole models used by Gabriel et al. [16] have been approximated by Debye models.

Finally, a point source has been used instead of a realistic antenna configuration as the focus of our article is the behavior of electromagnetic propagation in this specific problem and not how well energy may be directed towards the aorta according to antenna properties.

IV. SIMULATION RESULTS

A. Transfer function

The current source signal $J_z$ in the simulation is a $7^{th}$ order derivative of a Gaussian pulse with energy centered around...
Scaling by calculated distance. The received signal is normalized by an equivalent signal the radiated energy that is reflected back to the receiver, determined by the medium).

hence on a propagating wave whose information is contained in the received field and the radiated field was calculated. This procedure assumes that the portion of the source energy that is contained in the propagating wave in (7) in a homogeneous (γ) region reliably estimates the radiated energy in the simulations with objects present.

Considering the transfer function $H_{r,R}(ω)$ (Fig. 3), we observe the following: 1) the attenuation is a strong function of frequency, and within the relevant range is at least 40 dB. 2) notches occur at almost regular intervals whose frequencies are functions of the aorta radius. The notches vary in strength and separation with frequency. Those between 1 GHz and 1.25 GHz and between 1.9 GHz and 2.2 GHz are the most prominent with an amplitude in the range $12 - 15$ dB, while higher-order harmonics are insignificant. The frequency deviations are more pronounced with increasing harmonic order.

**B. Time Domain**

Fig. 4 shows the time-domain responses where two distinct echos may be identified based on their radius-dependent delay. The delay between extrema is close to 1 ns.

In order for the transfer function to express the ratio of the radiated energy that is reflected back to the receiver, the received signal is normalized by an equivalent signal calculated by letting the distance $R$ in (6) become large while scaling by $\sqrt{2\pi R}$ so that the energy on the growing cylinder tends towards a constant and is hence independent of $R$. The scaling by $\sqrt{2\pi R}$ effectively refers the far field energy “back to the source”.

$$E_{z,src}(ω, r) = \mu J_z(ω) \cdot jω \cdot \frac{3}{4} H_0^2(γr),$$

$$E_{z,prop}(ω) = \lim_{r→∞} \left[\sqrt{2πr} \cdot E_{z,src}(ω, r)\right],$$

$$H_{r,R}(ω) = \frac{E_z(ω) - E_{z,scf}(ω)}{E_{z,prop}(ω)} = \frac{E_{z,σ}(ω)}{E_{z,prop}(ω)}.$$ (8)

Here, $c$ is the speed of propagation as determined by $γ$. This procedure assumes that the portion of the source energy that is contained in the propagating wave in (7) in a homogeneous (γ) region reliably estimates the radiated energy in the simulations with objects present.

**V. DISCUSSION**

In Section IV, the modulus of the transfer function between the received field and the radiated field was calculated. This function is a result of several effects and which are described in Section V-A. The time-domain equivalences are commented upon in Section V-B.
A. Transfer function

The transfer function presented in Fig. 3 incorporates several phenomena such as the reflections at the air-skin boundary, the converging (entering) and diverging (exiting) effect of this boundary, the attenuation as the wave propagates through the lossy material, the interaction of the wave with the cylindrical structure of the aorta expressed in the aorta’s radar cross section (RCS) $\sigma_{\text{aorta}}$. These phenomena are summarized in the radar equation:

$$P_{\text{Rx}} = \frac{(G_{Tx}) \cdot (\sigma_{\text{aorta}}) \cdot (A_{Rx})}{(2\pi R_{Tx} L_{M,Tx}) \cdot (2\pi R_{Rx} L_{M,Rx})}, \quad (9)$$

$$G_{Tx} = \frac{P_{\text{Propagation}}(\theta)}{F_{\text{Injected, isotropic}}} = 1,$$

$$A_{Rx} = \frac{P_{\text{incident}}}{F_{\text{Received}}} = 1,$$

where, $P_{\text{incident}}$ is the received power density, $A_{Rx}$ is set to unity in this context because the received signal and incident signal are identical and the generic loss terms $L_{M,Tx}$ and $L_{M,Rx}$ include transmission and reflection coefficients. The terms relating to spherical propagation have been replaced by equivalent terms for cylindrical propagation. Considering the antenna gain $G_{Tx}$, the approach of Section IV amounts to using an ideal, isotropic antenna as the injected power equals the isotropic, radiated power. Extrapolating from (9) we anticipate an order of $1/(2R)^2$ increased attenuation in 3D compared to a 2D geometry.

As presented in Tab. I, the air-skin-avg and avg-skin-air transmission loss $(1 - |\rho|^2)$; normal incidence) account for approximately 6.4 dB.

The attenuation of the wave as it propagates in the average material is shown in Fig. 2. Fig. 5 shows the equivalent transfer function for the front reflection from the aorta $H_M(f)$.

The RCS of the aorta, $\sigma_{\text{aorta}}$, may be estimated by disregarding loss in the surrounding material in which case an exact result is known [19, chapter 4] and illustrated in Fig. 5 for a radius $r$ of 10 mm. The notches at approximately 1.2 GHz and 2.2 GHz correspond well with that in Fig. 3. Furthermore, the notches decline in strength with increasing frequency, an effect probably due to the lossy propagation of the wave inside the aorta.

The notches occur due to interference as waves are phase-shifted by $\lambda/2$. There are two possibilities: either the front-wall reflection may combine with the rear-wall reflection or else with a wave traveling around the aorta. These two waves travel distances $4r$ and $2r + \pi r$ at velocities 0.13$c_0$ and 0.2$c_0$ respectively (Fig. 2). The ratio of delays is $\Delta t_{\text{aorta}}/\Delta t_{\text{avg}} \approx 1.23$ at 1.5 GHz, which implies that their interference will result in phase shifts and amplitude scaling depending on the waves’ relative strengths. Assuming the second echo is primarily due to round-trip propagation within the aorta, a simple explanation of the notches may be proposed. The notches occur when the phase difference is equal to $\pi k$, or equivalently when the second echo phase shift is $2\pi k$.

$$f_{\text{Notches}} = \frac{c_{\text{aorta}}}{4r} \left( k - \frac{\theta_0}{2\pi} \right),$$

$$\Delta f_{\text{Notches}} = \frac{c_{\text{aorta}}}{4r} \approx 0.96 \text{ GHz}, \quad (10)$$

where $c_{\text{aorta}}$ is the propagation speed in the aorta. The term $\theta_0$ represents an unknown, constant phase shift for which an explanation has not been found, and (10) shows that $\theta_0$ has no influence on the distance between notches. Fig. 3 and Fig. 5 show that this explanation agrees well with both simulated and theoretic results.

B. Time Domain

Still assuming the second echo is due to the round trip propagation within the aorta, the second pulse of energy is expected to follow after $\Delta t_{\text{aorta}}(r = 10 \text{ mm}, f = 1.5 \text{ GHz}) \approx 4r/(0.127 c_0) = 1.04 \text{ ns}$. In Tab. II, the delay between the peaks at 3.95 ns and 4.88 ns, which correspond to the peak response of the front reflection and second echo respectively, is 0.93 ns. This is relatively close to the expected value. The presence of distinct echos is clearly expressed in the peaks’ temporal shifts and may be grouped in two sets with decreasing and increasing delays. It is clear that advancing reflections are less sensitive to changes in radius than the receding reflections. While the round-trip path length of the

| TABLE II |
| --- |
| Linear regression coefficients of selected peak’s delays: \( \Delta t = ar + b \), where \( r \) is measured in mm. The strength of each peak is also indicated, and are selected from both simulation runs. |
| a[ps/mm] | b[ns] | \( t_\Delta (10\text{mm})[ns] \) |
| 1 | -31.6 | 3.91 | 3.60 |
| 2 | -32.2 | 4.09 | 3.77 |
| 3 | -32.8 | 4.27 | 3.95 |
| 4 | -33.4 | 4.46 | 4.12 |
| 5 | -34.4 | 4.66 | 4.31 |
| 6 | 71.0 | 3.98 | 4.69 |
| 7 | 73.2 | 4.15 | 4.88 |
| 8 | 75.1 | 4.32 | 5.07 |

Fig. 5. Theoretic radar cross section of a cylinder $\sigma_{\text{aorta}}$ with an incident plane wave [19] by disregarding the attenuation in the material surrounding the aorta (radius 10 mm). Transfer function $H_M(f, 19.32 \text{ cm})$ corresponding to the round trip propagation ($r = 10 \text{ mm}$) in tissue. Also, combined effect.
front reflection decreases by $2\Delta r$, the second echo is due to a round-trip internal path with increased path length of $4\Delta r$ in the aorta and a decrease in the average medium of $2\Delta r$ (12). At 1 GHz, $c_{avg} \approx 0.202c_0$ and $c_{aorta} \approx 0.127c_0$, so:

$$-2\Delta r/c_{avg} \approx -32.9 \text{ ps/mm}, \quad (11)$$

$$\Delta r (4/c_{aorta} - 2/c_{avg}) \approx 72.0 \text{ ps/mm}. \quad (12)$$

A remark on Tab. II concerns the variations of the observed delay sensitivities ($\alpha$ ps/mm): these will be affected by the interference between reflections and a crest would tend to move with the stronger signal, while being ’retained’ by the weaker. This may explain in part the entries 4 and 5 in the table and representing peaks located close to the midpoint between echos.

**VI. CONCLUSION**

The paper presents a feasibility study on measuring diameter variations of an aorta using radar based methods. A survey of medical literature shows that pulse wave velocity leads to an estimation of aortic compliance. Combined with radius and radius variations, the pulse pressure may be estimated.

Finally, based on the assumption that the non-linear functional relationship between compliance and pressure may be known, mean pressure may be estimated.

Due to strong attenuation in biological tissue, feasibility is essentially hinged on a viable power budget. Based on simulations and a simplified geometry, an upper bound on received power in the range of $\langle 0.8, 5 \rangle$ GHz shows 40 dB loss at the lower end increasing to about 120 dB at the upper end where material loss is dominant. The estimate is qualified as an upper bound due to missing reflections in our geometry. Furthermore, notches due to the aorta are present at around 1 GHz and 2 GHz and suggest a frequency-domain approach to estimating aorta radius. In the time domain, simulations indicate that two echos may be identified and associated with the aorta radius.

In future work, we are planning to conduct further simulations with complex, realistic geometries.

**VII. ACKNOWLEDGMENTS**

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