INTERFERENCE RESISTANCE IN AUTOMATED OSCILLOMETRIC
SPHYGMOMANOMETERS

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The goal of the present work was to survey the interference resistance of channels for signal transfer and transformation in automated sphygmomanometers with oscillometric recording. To attain this goal, it is necessary to determine the characteristics of the noise signals which presently obscure valid signals, to study their interferences and their effects on the valid signals, and to outline approaches to the reduction of these effects.

A mathematical model [4] was used to determine the characteristics of valid signal, the model taking into account different types of elasticity and collapse of the humeral artery, the volume to pressure ratio of the artery, the amplitude and shape of oscillations of arterial blood pressure (taking into account the blood pressure cuff volume), and some other parameters.

The humeral artery, the arm, and a blood pressure cuff filled with air can be simulated by concentric circles (see Fig. 1). The outer wall of the cuff is immobile, and the elasticity of the cuff is fully determined by the air in the cuff chamber.

From the physical point of view, oscillation (more precisely, pulsation) of pressure in a blood pressure cuff is caused by the following processes: each pulsation of arterial pressure increases the volume of the artery, and the artery wall expands because of its elasticity, the expansion of the artery wall is translated through soft tissues of the human forearm to a blood pressure cuff filled with air. The air in the cuff obeys Boyle–Mariotte’s law for an ideal gas: the product of numerical values of pressure \( P \) and volume \( V \) under constant temperature and mass is constant: \( PV = \text{const.} \), i.e., decrease in the cuff volume causes increase of pressure in it.

Since \( PV^k = \text{const.} \), where \( k \) is the adiabatic coefficient which is equal to the ratio of specific thermal capacities \( (C_p/C_v) \) as determined under constant pressure and volume, respectively. For diatomic gases \( (H_2, N_2, O_2) \) at room temperature \( k = 1.4 \).

Therefore:

\[
\frac{P_c(t)}{P_{cd}} = \left[\frac{V_{cd}}{V_c(t)}\right]^k,
\]

where: \( P_c(t) \) is the instantaneous value of the pressure in the cuff, and \( P_c(t) = P_a(t) - P_T(t) \); \( P_T(t) \) is the instantaneous transmural pressure; \( P_{cd} \) is the pressure in the cuff under diastolic arterial blood pressure; \( V_{cd} \) is the volume of the cuff under diastolic arterial blood pressure; \( V_c(t) \) is the instantaneous volume of the cuff.

The initial volumes of cuff and artery can be determined experimentally, their values being connected by the following equation:

\[
V_c(t) = V_a \cdot V_{cn} - V_a(t),
\]

where: \( V_a \) is the volume of the humeral artery under zero value of \( P_T(t) \); \( V_{cn} \) is the normalized volume of the cuff; \( V_a(t) \) is the instantaneous volume of the humeral artery.

The system of equations can be solved by a successive approximation method. The pressure in the cuff is calculated in this method for discrete values of arterial blood pressure during each cardiac cycle. The procedure of the calculations is repeated for different values of initial pressure \( P_{cd} \), the value of \( P_{cd} \) gradually declining from 220 to 1 mm Hg.

The results of the study fall into three groups.

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Fig. 1. Schematic presentation of a human forearm: 1) blood pressure cuff; 2) soft tissue; 3) artery wall; 4) blood in the artery.

Fig. 2. Volume per unit length of humeral artery, $V_a(t)$, as a function of transmural pressure, $P_T(t)$, for different types of elasticity of the artery. Here and in Figs. 4-6: 1) high elasticity; 2) normal elasticity; 3) low elasticity.

Fig. 3. Shape of arterial blood pressure pulse wave in humeral artery: $P_{a,s} = 120$ mm Hg and $P_{a,d} = 70$ mm Hg; $s = 0.42$.

I. Characteristics of Valid Signal With Various Types of Elasticity of the Humeral Artery

The dependence of arterial volume on transmural pressure with different types of elasticity of the humeral artery was studied in [7]. The dependencies for humeral arteries with high, normal, and low elasticity are shown in Fig. 2 (curves 1, 2, and 3, respectively). For positive transmural pressure, the curves are of exponential character, and they are completely described by three parameters: volume per unit length of humeral artery, the slope of the curve under zero transmural pressure, and the asymptotic decrease in artery volume at large transmural pressure.

Results of calculations of amplitudes of valid signals ($\Delta P_c = P_{c,s} - P_{c,d}$) for different types of elasticity of the humeral artery and for different values of $V_{c,n}$ are shown in Fig. 3. Systolic and diastolic arterial blood pressure in the calculations were taken as $P_{a,s} = 120$ mm Hg and $P_{a,d} = 70$ mm Hg, respectively. The shape of the arterial blood pressure pulse wave in the humeral artery was approximated by a half-cycle sine curve (Fig. 3), and $s$ was taken as a ratio of pulsation duration ($t$) to the cardiac cycle period ($T$). For patients in the age group from 18 to 24 years, $s$ is constant and equal to 0.42. The amplitude of the pulsation is equal to the pulse pressure value ($P_{a,p} = P_{a,s} - P_{a,d}$).

Figure 4 shows that the amplitude of the valid signal rises as $V_{c,n}$ decreases, this fact being explained by variation of the elastic properties of the cuff as a whole. However, if the cuff volume $V_{c,n}$ is less than 100, the values of $P_{a,s}$ and $P_{a,d}$ will be determined with significant error.