MODELING ARTERIAL AND LEFT VENTRICULAR COUPLING FOR NON-INVASIVE MEASUREMENTS

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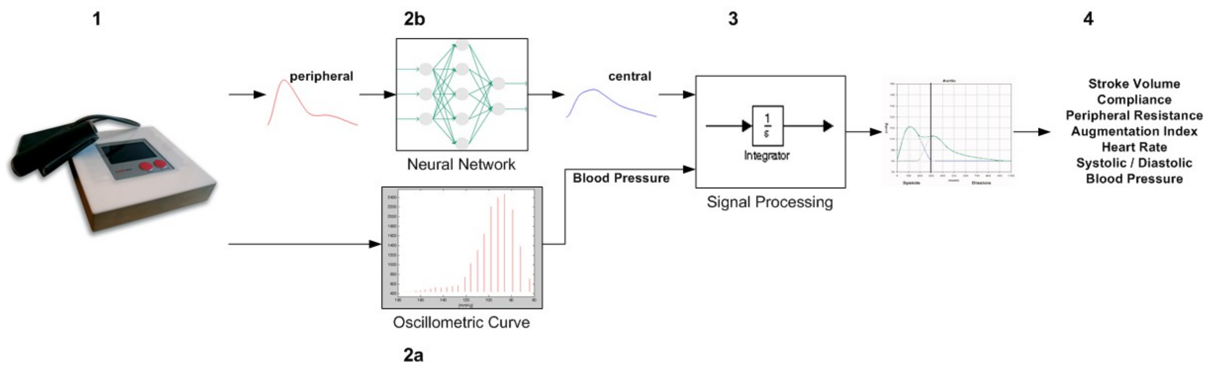
Abstract

Aim of the presented work has been the development of an algorithm for a non-invasive, portable, easy to use, and affordable device (1) for measuring systemic cardiovascular parameters like cardiac output and peripheral resistance. The data acquisition is based on a common oscillometric measurement using an occlusive blood pressure cuff (2a) and no additional calibration is necessary. The introduced novel algorithm combines several simulation techniques like neural networks (2b) or differential equations (3), which will be explained briefly. The determination of the hemodynamical parameters is based on the idea that the ejection work of the left ventricle is subject to an optimization principle (4). Finally we present some results that justify our approach. This kind of model needs no additional external parameters to work and opens therefore good perspectives for non-expert use.



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Presenting Author's biography

Christopher Mayer has graduated in mathematics at the Technical University of Vienna. Currently he is working on his doctoral thesis about cardiac output measurement at the Austrian Research Centers. His research interests are pulse wave analysis and embedded simulation using Matlab Simulink and digital signal processors.

1 Introduction

Cardiovascular diseases (i.e. cardiac infarction, stroke) are rapidly increasing in our society. Meanwhile cardiovascular diseases are jointly responsible for about 50 percent of all fatal casualties in western developed countries. They are one of the leading origins for morbidity and mortality in these countries [1, 2]. Up-to-date statistical data emphasize that these diseases cause more deaths than cancer [3].

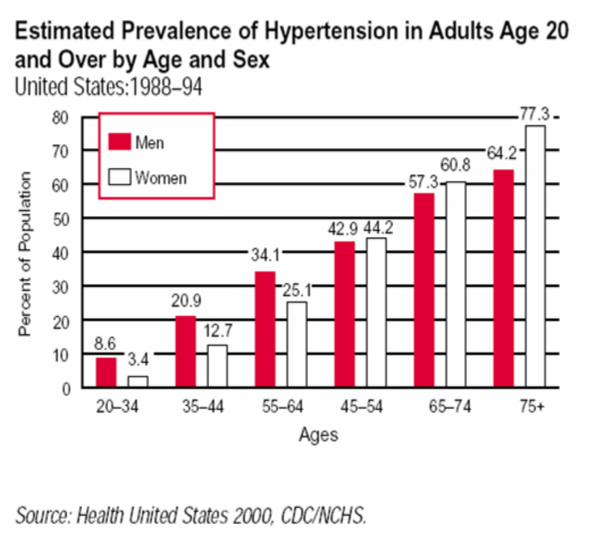

Fig. 1: Estimated prevalence of hypertension

Worldwide about 1 billion people suffer from hypertension, which causes a significant increase in risk of cardiovascular diseases. In figure 1 one can see the prevalence of hypertension depending on age and sex. Blood pressure is affected by the cardiac stroke volume (SV), the heart rate (HR), impaired arterial compliance (C_A), increased peripheral resistance (R_P), and the human body fluid volume. Anti hypertensive agents lower blood pressure by reducing these factors or combinations thereof. Consequently, measurement devices for the above mentioned hemodynamic parameters are needed to improve classical therapies.

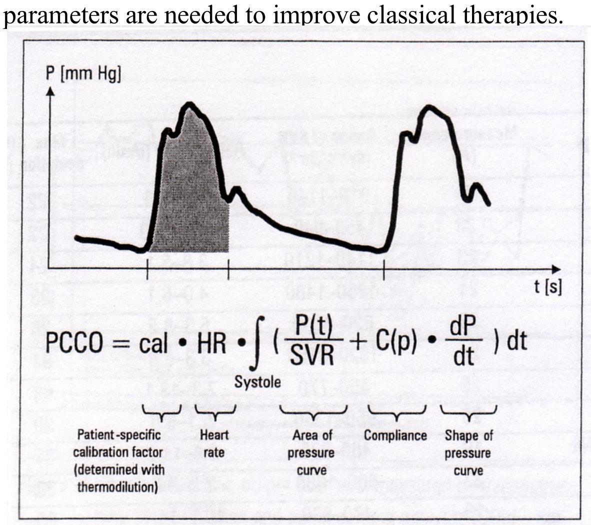

Fig. 2: Example for a pulse contour analysis method - PICCO by PULSION Medical Systems AG

Nowadays exist different measurement devices for the determination of the above mentioned hemodynamic parameters. On the one hand non-invasive methods, such as the Doppler method or impedance cardiographs, whose drawbacks are the operator dependency and the costs. On the other hand there are invasive methods, such as the golden standard thermodilution. Their disadvantage is the invasiveness and the combined risks. Existing pulse contour analysis methods are either dependent on personal data, such as age, height or weight or on an initial calibration reintroducing invasiveness. In figure 2 one can see the functionality of a pulse contour analysis method using a personal calibration factor obtained by thermodilution.

Aim of the presented work has been the development of an algorithm for a non-invasive, portable, easy to use, and affordable device for measuring the cardiovascular parameters based on a common oscillometric measurement using an occlusive blood pressure cuff.

2 Methods

The data acquisition is realized by means of an occlusive cuff applied to the upper arm resp. the Arteria brachialis. The process of acquiring data for the algorithms is divided into two separate cycles. The first one is used for calculating the systolic and diastolic blood pressure by analyzing the recorded oscillometric amplitudes.

The algorithm for determining the systolic and diastolic blood pressure is based on the analysis of the characteristics of the envelope of the oscillations' amplitudes [6]. Artifacts caused by patient's motion, by cardiovascular diseases, or by problems of the measurement device itself (i.e. motor and valve) lead to inaccuracies in the calculated blood pressure values. Therefore primarily an artifact reduction has to be made. It is simply based on the knowledge of the characteristic of the envelope. The systolic blood pressure is determined by means of determining two lines, one representing the slope of the envelope and the second one representing the basal level of the oscillations (see Fig. 3).

The diastolic blood pressure is determined by calculating the center of mass of the oscillations' amplitudes and corresponding pressure values in the interval mean pressure to minimal cuff pressure of the deflation process.

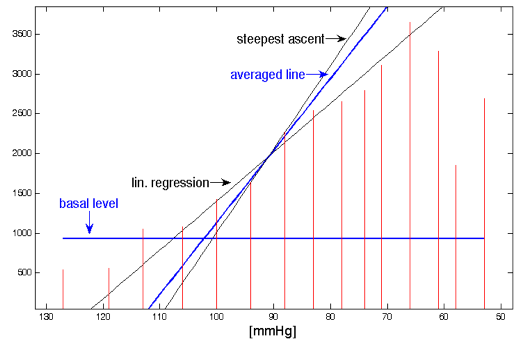

Fig. 3: Determination of the systolic blood pressure

During the second period the pulse wave is registered. By averaging and preprocessing the signal by applying various filter techniques the peripheral pressure contour is obtained.

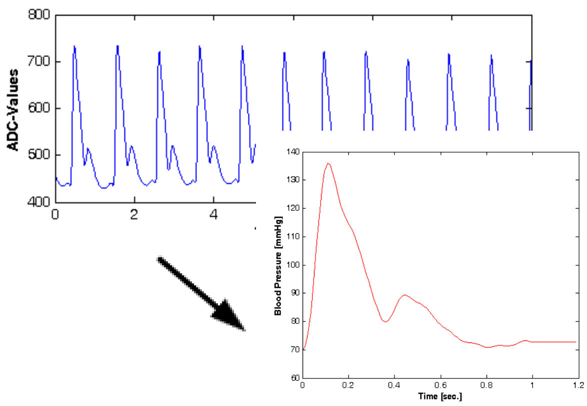

Fig. 4: Sampled Pulse wave

Afterwards an transfer function needs to be utilized for the calculation of the central pulse wave on basis of the peripheral one. The governing equations are those of Navier-Stokes and their linear solution has been published by Womersley in the 1950s. The basic idea of Womersley is that every volume flow and pressure profile can be represented by aggregated harmonic functions (sine and cosine). To compose and decompose such pressure and flow waves the so called Harmonic Analysis, introduced by Fourier in 1822 is used.

To calculate the wave propagation along an elastic tube Womersleys theory uses a specialized Telegraph (Wave) Equation, which incorporates important parameters like tube length and diameter as well as elasticity. As a consequence pulse wave velocity (PWV) has to be considered intensively, because regarding unsteady flow the wall mechanics massively influence the pulse wave velocity, which describes the propagation of the pressure waves. The fundamental relation, which is called the Moens-Korteweg equation, describes primary behaviour of the pulse wave velocity (c_0) in elastic tubes:

$$\textit{Pulse Wave Velocity } (c_0) = \sqrt{\frac{Eh}{2R\,\rho}}\ , \quad (1)$$

$$\begin{gathered} Eh = YoungModulus * wallthickness \\ R = radius \\ \rho = blood\ density \end{gathered} .$$

The most important implication of this is the fact that changes in diameter or Young modulus have a quadratic effect on pulse wave velocity. Due to the hardening of arteries by arteriosclerosis and therefore increasing stiffness, pulse wave velocity is used as a relevant risk indicator in clinical practice. The reader may also note that PWV rises along the arterial tree. As stated before using the Moens-Korteweg Equation the decreasing diameters as well as the naturally

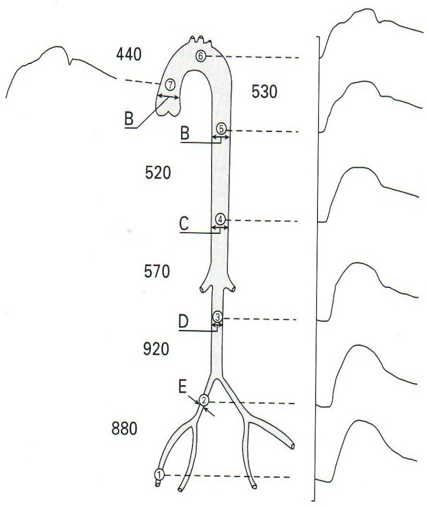

Fig. 5: Measured pulse wave propagation along the aorta.

higher Young modulus in peripheral arteries are responsible for this behaviour. One may also look at the pressure curves plotted on the above figure. The pressure curve gets smoothed by the hardening tubes and becomes also slimmer in shape with decreasing values for the vessel diameter along the arterial tree.

As already mentioned above the Womersley solution depends on harmonics. Each harmonic has a different frequency, whereas the important ones are usually those under 50 hz. Depending on the frequency and radius flow waves follow pressure waves in a certain way. To calculate this behavior we utilize a Bessel Function (FJ) which is also used to determine the skin effect in classical electro-technical applications. The function is parameterized with the so called Womersley Number (α) which is dimensionless and normalizes the above mentioned influence of radius and frequencies.

$$\text{Womersley Number } \alpha = \sqrt{\frac{R\omega}{\nu}}\,, \quad (2)$$

$$\begin{gathered} R = radius \\ \omega = angular\ frequency \\ \nu = kinematic\ viscosity(\mu / \rho) \end{gathered}\,.$$

As stated in the previous paragraph both the arterial radius and elasticity have major influence on pulse wave velocity. Cumulating all our considerations into one relation between radius, elasticity and frequency yields in a resulting term referenced as

Complex Wave-Propagation Velocity (c) =

$$\sqrt{\frac{\pi R^2 (1 - FJ(\alpha))}{\rho C}}\,. \quad (3)$$

Based on the above introduced relation we are able to calculate the transmission of pulse and flow waves at any point x along a cylindrical tube using the following equations [7]:

Flow Wave = (4a)

$$Q(x,\omega) = a\cos\left(\frac{\omega x}{c}\right) + b\sin\left(\frac{\omega x}{c}\right),$$

Pressure Wave = (4b)

$$P(x,\omega) = i\sqrt{\frac{\rho}{C\pi R^2(1 - FJ(\alpha))}} * \left(-a\cos\left(\frac{\omega x}{c}\right) + b\sin\left(\frac{\omega x}{c}\right)\right)$$

Using the above formalism the pulse and pressure waves within the arterial system can be described well. For practical use the parameter identification for this method is still too complex. Therefore we use a three layer feed forward back propagation network (Fig. 6), which we have trained with clinical data.

To remove reflections from this curve the augmentation index, which is defined as

$$AIx = \frac{sBP - iBP}{sBP - dBP} \quad (6)$$

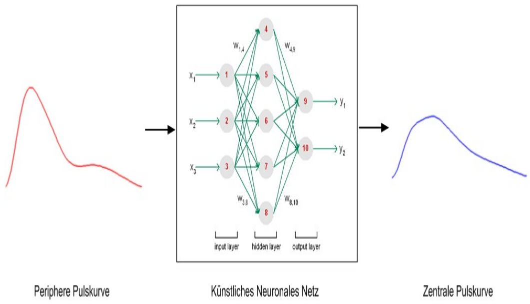

Fig. 6: Pulse wave transform

is used. The inflection pressure (*iBP*) represents the pressure at the first positive inflection point within the systole

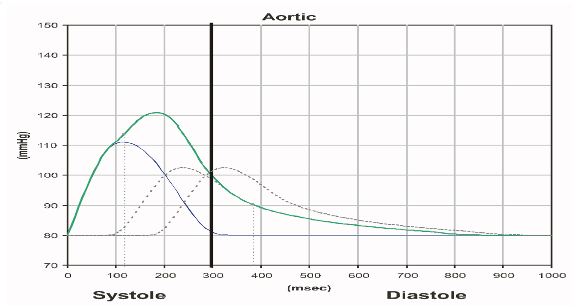

Fig. 7: Aggregated Central Pulse wave

and is determined by means of cubic spline interpolation. The augmentation index is a measurement for the superposition of the generic and reflected pulse curves. It is also used as a measure for arterial stiffness.

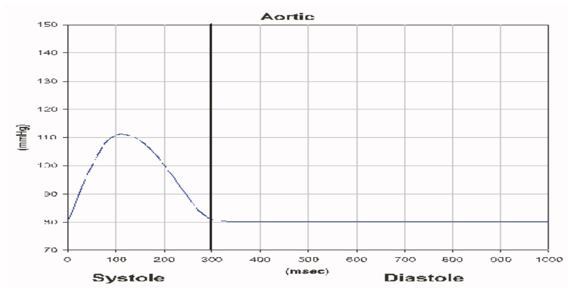

Fig. 8: Generic Central Pulse wave

Based on the generic aortic pulse wave the cardiovascular parameters are calculated by using the idea that the ejection work of the left ventricle is subject to an optimization principle. For this purpose the system dynamic of the sub model is described by the Windkessel equation (7a) [8,9] and several constraints.

$$q(t) = R_P C_A \dot{x} + x = \alpha \dot{x} + x \quad (7a)$$

$$R_P = f(\alpha) \quad (7b)$$

$$C_A = f(\alpha) \quad (7c)$$

$$x(0)=x_0 \tag{7d}$$

$$x(t_s)=x_s \tag{7e}$$

$$x_s=x_0 e^{\left(\frac{t_p-t_s}{R_P C_A}\right)} \tag{7f}$$

$$x_d=x_s e^{\left(\frac{-t-t_s}{R_P C_A}\right)} \tag{7g}$$

$$q(t_s)=0 \tag{7h}$$

$$\int_0^{t_s} q\,dt=V_s \tag{7i}$$

$$\int_0^{t_s} x\,dt+\int_{t_s}^{t_p} x\,dt=V_s \tag{7j}$$

This system gets solved by a calculus of variations. The resulting equation is subsequently a system of second order which we solve using the classical approach. *A, B, C* are the integration constants and α depends on R_P and C_A.

$$p(t)=Ae^{\alpha t}+B^{-\alpha t}+C \tag{8}$$

To identify the mentioned parameters we use a Levenberg-Marquardt algorithm. The goal is to match the area under the central pulse wave during Systole with respect to the augmentation of the wave caused by reflections.

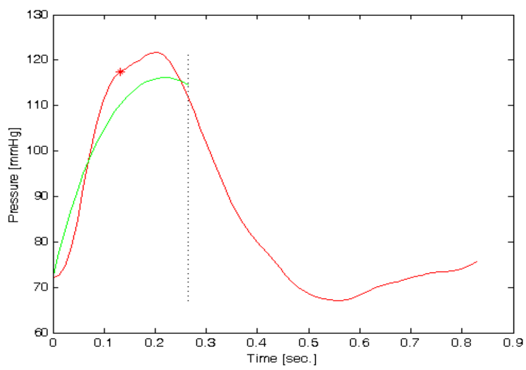

Fig. 9: LM - fit for Central Pulse wave

3 Results

For validating the algorithm and for fulfilling the requirements of the Medical & Healthcare Products Regulatory Agency for the approval as a medical technical device class IIa measurements to compare the results of the algorithm to a reference instrument have been performed. As reference instrument the Task Force® Monitor by CNSystems, an impedance-cardiograph, has been chosen.

The measurements have been performed simultaneously, whereas the electrodes for the reference instrument and all blood pressure cuffs have been placed in advance. After starting and stabilizing of the reference instrument 2-3 measurements have been taken by the CardioMon™ to show agreement and internal reproducibility.

Fig. 10: Age distribution of probands

The number of probands taking part in this study has been 67, whereas 3 have been excluded due to medical reasons. The age distribution and the distribution of the diastolic and systolic blood pressure of all probands are shown in Fig. 10 and Fig.11 and some statistical data in table 1.

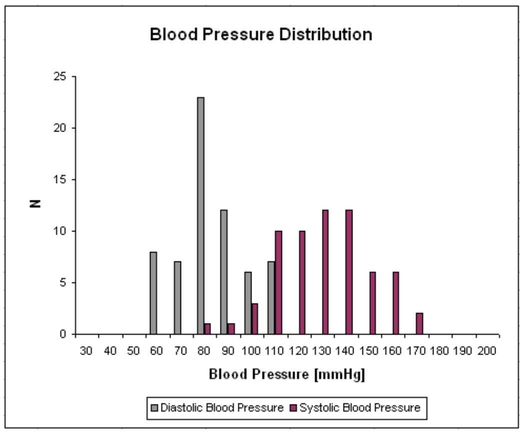

Fig. 11: Blood pressure distribution of probands

Tab. 1: Statistics of probands

Probands	p = 67
Measurements	n = 64
Age	49 ± 14 years
Gender	38 Male, 29 Female
Height	1.73 ± 0.09 m
Weight	76.00 ± 16.93 kilos

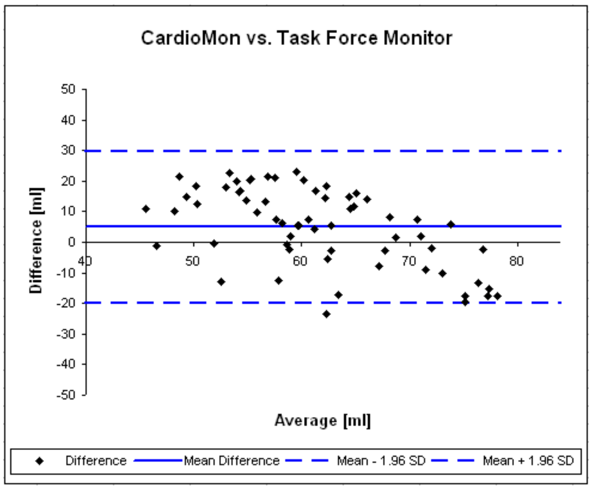

Fig. 12: Results of comparison of CardioMon to reference instrument

The results of the study are graphically presented by means of a Bland-Altman-plot in Fig. 12. The mean difference between the CardioMon^{TM} and the reference instrument is 5.03 ml with a standard deviation of 12.66 ml.

4 Conclusion

The measurements show good agreement between the algorithm presented in this paper implemented on a digital signal processor and the reference method. Therefore the application can be used as an alternative to standard cardiac output determination methods.

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