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Diagnostics of the human circulatory system: High resolution beat-to-beat measurement of systolic blood pressure using personalized models

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Abstract – Blood pressure and heart rate monitoring allow evaluating the cardiovascular efficiency and point out how the circulatory system responds to stress and exercise. Especially the blood pressure indicates the ability of a person to withstand emotional and physical stress. For blood pressure measurement oscillometric or Riva-Rocci/Korotkow based instruments are used as standard devices. Their severe drawback is the low time resolution preventing the registration of dynamic cardiovascular changes. We present a new high resolution continuous blood pressure measure technique based on individualized mathematical models. The method can be applied without interfering with the subjects cognitive perceptions. Beside calibration tests we present the first continuous measurements during stress and traffic situations. Characteristic pros and cons of the new method, the accuracy and new diagnostic possibilities are discussed.

Keywords: Continuous blood pressure measurement, beat-to-beat, pulse transit time, stress and traffic situations.

1. INTRODUCTION

Increased blood pressure as disease of modern civilization is not a new problem and it is assumed that each fifth adult at the age over 40 years suffers hypertension. The actual disease frequency can be only estimated, since an increased blood pressure in the initial stage causes no suffering pressure and the persons concerned therefore do not visit physicians. Consequences of an existing high blood pressure are among others hardening of the arteries which, existing during a longer period, can lead to complications such as cardiac infarct, kidney damage and brain impact. The determination of the blood pressure is therefore very important for the recognition of heart and blood cycle diseases.

Stress can make blood pressure rise temporarily and it is assumed to contribute to high blood pressure. But the long-term effects of stress are unclear yet. Stress management techniques seem to be another field of application.

Several methods exist to measure the blood pressure. They differ substantially in their theoretical background, their instrumental necessities and their accuracy. The direct invasive measuring method still is regarded as the "gold standard". This technique measures the blood pressure continuously using an intra-arterial catheter. The method is however connected with quite high instrumental expenditure and significant risks for the patient, so that it is applied only in the intensive medicine. Most frequently indirect non-invasive methods are used for blood pressure measurement like the procedure of Riva-Rocci, i.e. listening to Korotkow-sounds using a stethoscope, and the oscillometric method, both utilizing inflatable cuffs. Today the oscillometric method is employed in a variety of fully automated devices for blood pressure self measurement. The instruments have an accuracy of ± 3 mmHg and the measurement is quite simple, but it also has disadvantages. So the results become incorrect if movements and cramping of the subjects occur. In particular the blood pressure cannot be determined under load conditions, when a subject is exercising. And in addition no continuous determination of the blood pressure is possible, since the individual measurement including inflating the cuff lasts about 30 to 40 seconds. Beyond that a temporal interval of at least one minute should exist between two measurements. Thus fast blood pressure fluctuations cannot be determined.

Furthermore, there are continuously working blood pressure measuring systems, which are based on non-invasive methods. We distinguish between systems using cuffs and others, utilizing pressure sensors above exposed arteries. A different approach is the computation of the beat-to-beat blood pressure from physiological parameters like the pulse wave velocity.

The Peñáz-method measures the complete wave shape of the arterial blood pressure at the finger by means of two alternatively operating finger cuffs with inserted optical transmission sensors controlling a fast electro-pneumatic servo system [1]. After calibration the measurable absolute pressures in the cuffs correspond to the momentary blood pressure values and can be registered. Disadvantage of the method is the poor applicability at patients with reduced blood circulation in the fingers and the not always ensured correlation between the blood pressure in the finger and the pressure in the central circulatory system. Beyond that, squeezing of the fingers by the cuffs can lead to pain and deafness.

Another method applies an external pressure on an artery at the skin surface which is pressed locally against a bone. The sensor housing contains piezoelectric pressure sensors, which register the arterial pressure fluctuations. The measured pressure then can be noted as being proportional to the interarterial blood pressure. Due to a microprocessor-controlled servo mechanism the artery wall is completely relieved without the blood vessel collapsing [2]. However, most systems could not be established on the market to a larger extent yet.

We have developed a new high resolution continuous blood pressure measurement technique utilizing heart frequency and pulse transit time. It is based on former investigations of our institute [3,4]. Using an optical ear sensor and a standard chest belt known from sport activities, we determine the systolic blood pressure via an individual mathematical model from the pulse transit time $T_{\mathrm{R\_P}}$. The method can be applied without interfering with the subjects cognitive perceptions. The registrations were carried out during regular laboratory and exercise-electrocardiogram (ECG) tests as well as performing strain tests with the Vienna Test System VTS and using a car simulator.

2. MODEL ASSUMPTIONS

The dependency of the pulse transit time from the blood pressure is well established since the first investigations of Bramwell [5] and Gribbin [6].

2.1. State of the art

A useful approach consists in using the relation between blood pressure and pulse transit time. First attempts have been accomplished approximately 80 years ago. It was shown that with higher blood pressure the artery walls are more stretched and the elasticity decreases. This results in a higher pulse wave velocity, which is inversely proportional to the pulse transit time. More concrete statements were made by [6]. At the arms of 26 persons two sensors were fastened for pressure curve registration and the pulse transit time was computed. For each subject an individual linear relation between blood pressure and pulse wave velocity was found. Individual repetition measurements confirmed this relationship. Other investigations confirmed this [7-10].

Heard, Lisbon, Toth and Ramasubramanian [7] describe the DxTek monitor of the company DxTek Inc. This apparatus for continuous non-invasive blood pressure determination also utilizes the pulse wave velocity. The temporal distance between the R-complex rise of an electrocardiogram derivative and the time of the half pulse rise is computed as pulse wave transit time. The systolic blood pressure is determined using a quadratic equation. The coefficient of the square term is given as fixed value, the additive constant is determined individually. So far the system was validated in a clinical study, however, this system also seems not to be offered on the market.

Elter [9] mentioned apart from its own developments three devices for the continuous measurement of the blood pressure using the pulse transit time [10]. However, none seems to have been put on the market. The equipment of Greubel [11] uses an electrocardiogram recording and the photoplethysmographically measured finger pulse for the computation of the pressure. In addition the equipment must be calibrated in rest and after load. The prototype system of the company Vectron functions in a comparable manner, whereby the finger pulse is obtained using a pulse oximeter. The system of the Sentinel Company is named Artrac 7000. The pulse transit time is determined via two photosensors, which are fastened to ear and finger [12]. The system is calibrated by means of oscillometric measurements, which must be repeated every five minutes. Elter judges the determination of the transit time from pulse waves, which are not determined on the same blood vessel, as critical. Likewise disturbing are the frequently necessary calibretions, in particular if they have to take place during phases with distinct changes of blood pressure.

The structure of the patented Beat-2-Beat equipment [12] of the VSM MedTech Ltd. company is comparable with the Artrac 7000 described above. The two sensors for optical volume pulse measurement can be placed in different places such as ear, toe, forehead or finger. From the registered pulse waves the transit time is determined by cross correlation. The actual blood pressure computation is carried out using an equation of the form $p = a + b \cdot \ln(T)$, where T is the pulse wave travelling time, p is the blood pressure, a and b are individual constants, determined by calibration. The equipment was validated in several clinical studies on over 100 patients.

In the institute of Electrical Measuring of the University of Paderborn together with the Vestic Children's Hospital of Datteln investigations for continuous blood pressure measurements during exercise-ECGs have been accomplished. Barschdorff and Erig [3] describe a linear relation between blood pressure and pulse wave velocity. The model coefficients were individually determined. For the R-complex-determination in the ECG necessary for the pulse transit time computation a chest belt common in the fitness field is used. The peripheral pulse signal is measured with an infrared sensor at ear and/or finger. In [4] an advancement of the procedure is presented now using a neural net as mathematical model. It was trained for exercise-ECG applications with the data of more than 50 patients.

The determination of the blood pressure from cardiovascular parameters without using a cuff, i.e. without the necessity to apply an external pressure, is a very interesting approach. It represents the only way for a strain-free and continuous measurement. The formal relation between blood pressure and pulse wave velocity is described in the following section. Subsequently, our new method for non-invasive, continuous and high resolution blood pressure determination is presented.

2.2. Formal relationship between pulse transit time and blood pressure

Following Busse [13] and others [14] the mathematical relation between the pulse wave velocity c and the inter-arterial blood pressure p can be deduced. Regarding a non-absorbing and loss free flexible hose, this relation can be derived for a cross section Q as follows. Using Young's elasticity module E

$$E = \frac{\sigma}{\varepsilon} = \frac{F}{\varepsilon \cdot Q} \qquad (1)$$

and the compression module K

$$K = \frac{dp}{dV/V} \qquad (2)$$

we yield the Weber formula for the velocity c of a longitudinal pressure wave:

$$c = \sqrt{\frac{K}{\rho}}\,. \tag{3}$$

Here ρ refers to the fluid density. Fig. 1 shows schematically the flexible, liquid filled hose as an artery model with inner radius R, wall thickness h and length l

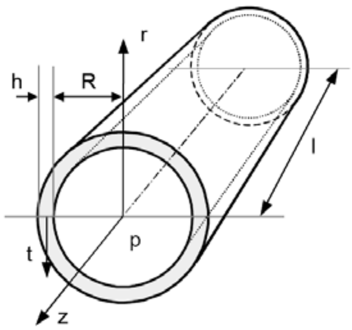

Fig. 1. Flexible hose as a schematically artery model.

In order to derive a relationship between fluid induced pressure p and the force due to the tangential wall tension σ_t and assuming a linear dependency between tangential elasticity E_t and fluid pressure p

$$E_t = E_0 + E_p \cdot p \tag{4}$$

we obtain (5), which indicates a quadratic dependency of the fluid pressure p from the wave velocity c

$$p = \frac{2 \cdot \rho}{E_p} \cdot \frac{R}{h} \cdot c^2 - \frac{E_0}{E_p}\,. \tag{5}$$

E_0 and E_p are constants.

The blood pressure thus depends on the relationship between radius and wall thickness of the artery segment, on the blood density and particularly on the pulse wave velocity. An increase of the blood pressure leads to an increase of the pulse wave velocity. Since it is reverse proportional to the pulse wave transit time, it becomes shorter with the increase of the blood pressure.

Assuming the hose radius R and the wall thickness h as well as the blood density ρ as constant, we can compute the blood pressure p from the measured pulse wave velocity c. In general we therefore find with the constants C_1 and C_2 an equation of the general form

$$p = C_1 \cdot c^2 - C_2 \tag{6}$$

The mathematical models for continuous blood pressure determination are based on this quadratic equation.

3. BLOOD PRESSURE MODELS

The blood pressure and the pulse wave transit time are depending on a variety of physiological parameters, which are unknown and cannot be measured individually. Therefore a general square equation is used, whose coefficients must be determined by a parameter estimation procedure.

3.1. Systolic blood pressure

The following considerations are first applied to the systolic blood pressure P_{sys}. Later we will discuss to what extent these assumptions can be transferred to determine the diastolic pressure P_{dias}. The mathematical model, which is the basis for the pressure determination, possesses the pulse transit time T_{R_P} as input and the systolic blood pressure P_{sys} as output.

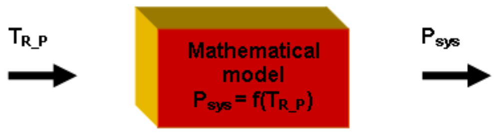

Fig. 2. Mathematical model for P_{sys} determination.

The model is separately defined for exercise/strain activities (subscript: str) as well as for recovery situations (subscript: rec) of a subject

$$\begin{aligned}\Delta P_{sys,str} &= a_{1,str} \cdot \Delta T_{R\_P}{}^2 + a_{2,str} \cdot \Delta T_{R\_P} + a_{3,str} \\ \Delta P_{sys,rec} &= a_{1,rec} \cdot \Delta T_{R\_P}{}^2 + a_{2,rec} \cdot \Delta T_{R\_P} + a_{3,rec}\end{aligned} \tag{7}$$

The a_i are the separately to determine and subject specific model coefficients. ΔT_{R_P} and ΔP_{sys} refer to the normalized input and output values, respectively.

$$\begin{aligned}\Delta P_{sys} &= P_{sys} - P_{sys,rest} \\ \Delta T_{R\_P} &= T_{R\_P} - T_{R\_P,rest}\end{aligned} \tag{8}$$

The pulse transit time and the blood pressure are not used as absolute but as normalized values with respect to the subject at rest (subscript: rest). This increases the robustness of the model coefficients concerning fluctuations of the physiological and psychological conditions. For example quite different quiescent blood pressures can be determined, depending on previous activities and the general health condition of the person under test. Also, as already mentioned, the model is divided in an exercise and a recovery part. Several measurements indicated that different relationships are describing the dependency between blood pressure and pulse transit time for the exercise as well as for the recovery phases.

In order to compute the systolic blood pressure continuously, one needs the pulse transit time beat-to-beat, the individual model coefficients and the quiescent blood pressure values. That means that for the time interval, during which the blood pressure is to be determined, the pulse transit time must be registered.

3.2. Continuous registration of the pulse transit time

A blood volume is pressed into the aorta by each contraction of the heart, resulting in a longitudinal pressure wave, which spreads from the heart into the arteries. The travelling time - thus the pulse transit time - is then defined as quotient of the distance between the measuring points and the pulse wave velocity.

The electrocardiogram (ECG) signal is used for the determination of the mean pulse transit time. Each heart beat begins with the R-complex in the ECG signal. Its rising flank represents approximately the time of the blood ejection from the left ventricle into the aorta. The chest belt yields a burst signal synchronous with the ECG R-complex. The terminating pulse is obtained from an optical transmission sensor, when the pulse wave reaches the peripheral measuring point. Therefore, the rising flank of the arriving wave terminating the time interval must be recorded.

An uncertainty in using the ECG signal could be seen in the fact that the R-complex does not exactly indicate the beginning of the blood injection into the aorta. Correctly, the so called pre-ejection period (PEP) must be considered which characterizes the initiation of the heart muscle. This time interval is not constant. However, the changes in the PEP are highly correlated with the changes of the transit time [9], so that the influence of the PEP on the transit time determination utilizing the ECG and the peripheral pulse usually is very small. Due to the individual calibration it is considered here.

3.3. Sensors

As already shown in [3] instead of a full ECG derivation a chest belt, Fig. 3, used for pulse measurement during sport activities, is suitable for measuring the beginning of the pulse transit time interval. An optical ear sensor, also from the fitness area, serves as pulse wave arrival indicator.

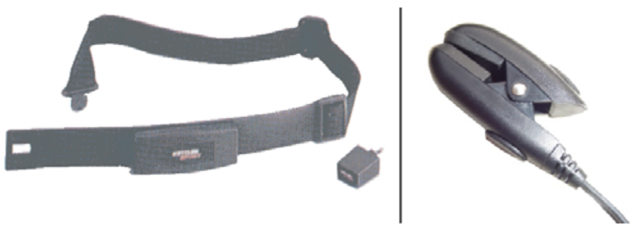

Fig. 3. Chest belt for ECG signal measurement and ear clip sensor.

The chest belt output signal is shown in Fig. 4.

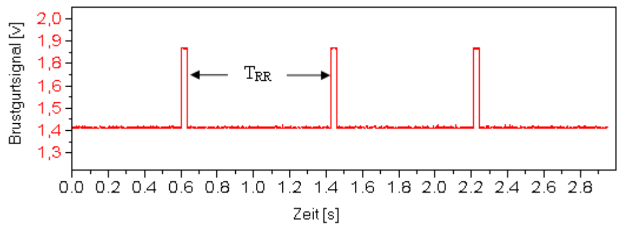

Fig. 4. Chest belt output signal.

The sensor signals are amplified and then digitized with a sampling frequency of 300 Hz using a standard LabJack USB PC front-end.

The chest belt signal can easily be evaluated by a threshold analysis, yielding the initiating trigger for the pulse transit time measurement. The heart frequency f_{HF} , which is also of importance, can be determined from the chest belt signal by computing the interval length between the rising flanks of two sequential pulses. Usually, the heart frequency is expressed in beats/minute.

If artefacts occur in the chest belt signal the determination of the heart frequency and the pulse transit time becomes quite difficult. Typical signal errors have one of the following forms:
- one or more pulses of the chest belt are missing;
- between two regular pulses additional flanks arise;
- a pulse is shifted in its position between two regular pulses

Artefacts are corrected by appropriate algorithms.

The determination of a characteristic point in the pulse signal represents a not trivial problem. This point must be clearly defined in the rising flank of the pulse wave for each heart beat. Several proposals can be found in the literature. We achieved the best results, i.e. the smallest dispersions in the transition time intervals, by using the 50 per cent point of the normalized pulse signal amplitude. Irregularities at the beginning and the end of the rising pulse flanks then do not severely affect the transit time determination.

3.4. Exercise and recovery model

To determine the model parameters, we searched for an easy to perform exercise test which does not need supporting devices. The knee-bending test proposed by Ruffier [15] could be used. However, it is not suitable for untrained persons, who could be overstrained. Better suited is a stair climbing test after Schellong [16], where the subject has to ascends and descend steps for a certain time. An immediate rise of systolic blood pressure and heart frequency with constant or slightly dropping diastolic pressure is expected. Each participant can find an own rhythm and climbing speed. Due to the simple load form and the good transferability this test is to be used for the further investigations.

At the beginning of the measurement, after fastening the sensors and a sufficient rest phase, three oscillometric blood pressure measurements were accomplished in one-minute distances. After another minute the load phase begins and the participant rises and descends the steps 30 times.

Ten seconds after exercise end the next blood pressure measurement is started. This delay is chosen in order to avoid measuring errors due to high pressure changes immediately after exercise ending. Further measurements follow in one minute intervals. In Fig. 5, upper part, typical curves of the heart frequency and the pulse transit time are shown. The times for blood pressure measuring are marked black.

It is evident, that the model coefficients can easily be determined, if blood pressure values have been obtained both, during the exercise and the recovery phase. With oscillometric instruments a measurement during exercise activities is not possible. In the following a new procedure is presented to determine the model functions if only resting and recovery values are present.

The maximum blood pressure at the end of the exercise phase is estimated using a polynomial approximation with the measured pressures during recovery as interpolation values. This maximum pressure is used together with the other during the recovery phase obtained values as additional point for the approximation function, yielding the recovery model.

As no measurements with the oscillometric device are possible during the exercise phase, we estimate the gradient of the pressure dp/dt for $t = t_A$ at the very beginning of the step test. Using this starting time t_A and the length of the exercise time interval, we obtain the pressure slope with the previously determined maximum systolic pressure.

$$\frac{d\left(\Delta P_{\text{sys}}\right)}{d\left(\Delta T_{\text{R\_P}}\right)}\left(\Delta T_{\text{R\_P}}=0\right)=\frac{\dfrac{d\left(\Delta P_{\text{sys}}\right)}{dt}\left(t=t_{\text{A}}\right)}{\dfrac{d\left(\Delta T_{\text{R\_P}}\right)}{dt}\left(t=t_{\text{A}}\right)} \qquad (9)$$

The model function for exercise can then be determined with the basic values at beginning and end of the exercise interval and this pressure gradient. We can now evaluate the systolic pressure during exercise and recovery using these both model functions, Fig. 5, middle.

In addition, we obtain a mean pressure model by averaging the coefficients of the both models, yielding a mean blood pressure model for the person under test. The inevitable error is to be analysed, but remains in most cases within the error of the oscillometric measurement devices. Fig. 5, lower part, shows an example for the pressure evaluation with the mean model function.

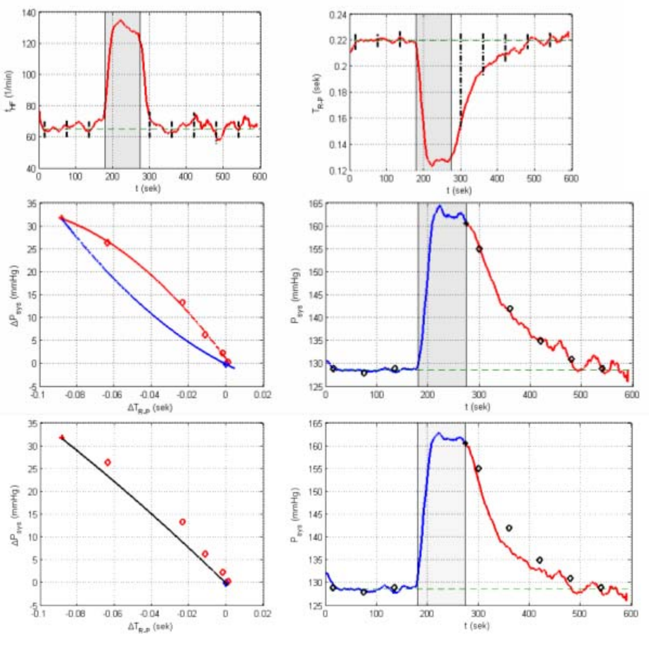

Fig. 5. Calibration measurement with heart frequency and pulse transit time (upper part), separate models for exercise/recovery and computed systolic blood pressure (middle) and mean model with computed systolic pressure (lower part).

However, as in most practical applications, especially those to be analysed later, we can not distinguish between exercise and recovery phases. We therefore have to use the mean model approach for the blood pressure determination. But in any case, as mentioned, one has to prove, that the pressure deviations are still within the accuracy of the oscillometric devices. For instance, when evaluating the blood pressure during driving a car, we cannot a priori distinguish between time intervals of excitement or calming down of the driver.

Exercise and recovery models can be of very different curvature, bending convex or concave, depending on the individual blood circulatory system or the physical training condition of the subjects.

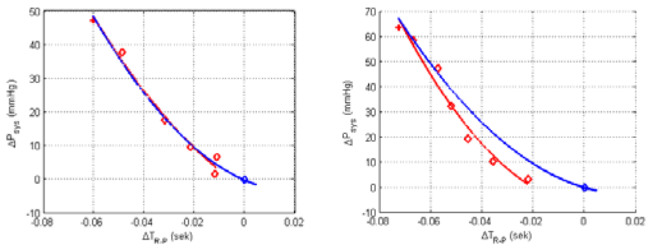

Fig. 6. Exercise and recovery models from different persons.

While in the left example of Fig. 6 there is practically no difference between exercise and recovery functions, the other example, right in Fig. 6, shows significant deviations. Figs. 5 and 6 therefore clearly indicate, that an individual determination of exercise and recovery as well as mean model functions are unavoidable in order to obtain blood pressure values as accurate as possible during changing strain conditions of persons under test.

In Fig. 7 we analyse a treadmill (exercise-ECG) test, where accurate measurements even during the exercise phase were obtained by an experienced medical assistant.

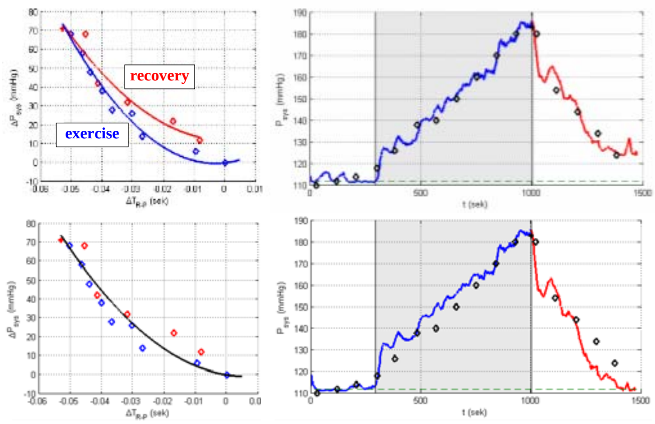

Fig. 7. Evaluation of a treadmill exercise test using exercise and recovery models (upper part, left) and averaged mean model (lower part, left), right: computed and measured systolic pressure P_{sys} .

It can be seen, that the steeper slope of the mean model function at $\Delta T_{\text{R\_P}} = 0$ overestimates the pressure values at the beginning of the exercise phase. The mean deviations, however, are in the accuracy range of an oscillometric or even a stethoscope based manual blood pressure measurement (± 3 mmHg). It still has to be demonstrated, that the model functions obtained via Schellong test calibration can be applied in different test situations. This will be shown in the second part of our investigation.

4. DISCUSSION

With our approach of utilizing individual exercise-recovery or averaged model functions we could determine the systolic blood pressure continuously and non-invasively with high temporal resolution. Also brief fluctuations are indicated, which remain unidentified if classical cuff based instruments are applied. The measuring system must be calibrated individually for the subjects under investigation using the Schellong stair climbing test. Since the pressure values during the calibration are measured using an advanced oscillometric device, a comparable accuracy can be achieved.

By using separate models for exercise and recovery the least deviations between computed systolic blood pressure and measured basic values can be achieved. For measurements, where no blood pressure values were obtained during exercise – which is the rule in practical applications - a suitable model can be determined by estimating the initial blood pressure gradient at exercise begin.

The calibration measurement for the model evaluation is to be accomplished only once for each person. Then the blood pressure can be computed using this model function with the individually derived parameters. In order to apply an exercise-recovery model to new tests, additional standardisation steps are necessary. However, in all practical applications one cannot differentiate between strain and recovery, so that a use of the double model is excluded anyway. The best possibility for these cases is the use of a single model, the coefficients of which are determined by averaging.

Due to the relatively small pressure changes and the dispersion of the measured values of P_{dias}, no generally suitable procedure for the determination of the diastolic blood pressure could be found. This is a topic for further investigations.

For a detailed failure analysis of the model based blood pressure determination comparative invasive measurements would be meaningful. Particularly in critical ranges of the blood pressure variations this comparison would be very useful. According to statements of medical experts, however, such comparative measurements can not be carried out due to the dangers for the subjects.

In the second part of this investigation [17] we demonstrate the first blood pressure measurements in stress situations. These are induced to the tested persons by carrying out experiments with the Vienna Test System [18] and during driving a car in a simulator stand. Because the sensor technique here is not at all influencing on the persons attention, the continuous model based blood pressure measurement proves as valuable additional indicator for the psychological conditions and the stress levels of the test participants.

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