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Modeling carotid and radial artery pulse pressure waveforms by curve fitting with Gaussian functions

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ABSTRACT

Modeling arterial pressure waveforms holds the potential for identifying physiological changes. There is a clinical need for a simple waveform analysis method with a high accuracy in reproducing the original waveforms. The aim of this study was to determine the accuracy of modeling carotid and radial pulses using Gaussian functions, making no physiological assumptions. Carotid and radial pulses were recorded from 20 normal volunteers. Ten consecutive beats from each volunteer were analyzed to determine beat-to-beat variability. Each pulse was decomposed using seven combinations of up to three Gaussian functions. The first function was always positive, but the second or third could be either positive or negative. Three positive Gaussian functions reproduced the original waveforms best with a mean absolute error (MAE) of 1.2% and 1.3% for the carotid and radial pulses respectively, and a maximum residual error of only 4.1% for both. This model had significantly smaller errors than any of the other six (all $P < 0.001$). Four positive Gaussian functions were then used to test the stability of this model. An insignificant change of the mean MAE (1.2% for both carotid and radial pulses) was obtained, showing that the stability has been reached with three positive Gaussian functions. The variability of MAE calculated as the standard deviation (SD) over the 10 beats was small at 0.2% for both pulses confirming the repeatability of using three positive Gaussian functions.

1. Introduction

Developments in arterial hemodynamics have indicated that arterial pressure waveform contains more information than is visually available from peripheral measured sites (wrist, ear, finger, or toe) [1–3]. Due to the complicated arterial topology, arterial pressure waveforms vary between different measured sites. However, all sites contain information about the general function of the cardiovascular system. This includes indices describing left ventricular systolic function [4], arterial stiffness [5], dynamics of the autonomic nervous system and heart–vasculature interaction [6]. Therefore, contour analysis of the arterial pressure waveform could be an important tool to explore and assess changes in cardiovascular system function.

Many researchers have used various waveform analysis techniques to identify specific features of the arterial pressure waveform. The most common are by derivative methods, which use the first [7], second [8] or third derivatives [9] of the arterial pressure waveform, or by wave intensity analysis [10,11], which analyzes vascular hemodynamics in terms of traveling energy waves. These techniques are simple and can be used in real-time analysis. However, none of these techniques analyzed the features of the complete arterial pressure waveform.

Researchers have also modeled the complete arterial pressure pulse using the Windkessel model from which compliance of the artery can be derived [12,13], or used distributed models of the systemic arterial tree to reproduce pressure waveforms at various locations [14–16], or used waveform fitting techniques, which decompose the arterial pressure waveform into several independent sub-waveforms. Published waveform fitting approaches include Rubins' method for analyzing simultaneously measured ear and finger blood volume pulse signals using four Gaussian functions [2], and Huotari's method for analyzing finger and toe photoplethysmographic (PPG) pulses using five logarithmic normal functions [17,18]. Both studies have demonstrated that the Gaussian function parameters were highly related to cardiac hemodynamic parameters, including the augmentation index, the reflection index, arterial elasticity and vascular aging. In terms of the effectiveness of modeling, Rubins reported that the residual error between the measured pulse and the fitted function did not exceed 10%. Huotari's study provided only some examples

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Table 1
Clinical characteristics of the volunteers participating in the study.

Characteristic	Mean ± SD
Age (year)	51 ± 11
Height (cm)	171 ± 9
Weight (kg)	69 ± 8
Body mass index (kg/m$^{2)}$	23 ± 3
Heart rate (beats/min)	71 ± 8
Brachial SBP (mmHg)	118 ± 12
Brachial DBP (mmHg)	71 ± 10
Brachial MAP (mmHg)	87 ± 9
Brachial PP (mmHg)	47 ± 11

Data are expressed as mean ± standard deviation (SD). SBP, systolic blood pressure; DBP, diastolic blood pressure; MAP, mean arterial pressure; PP, pulse pressure.

with an average maximum residual error of 4%. However, none of those studies attempted to specifically evaluate the accuracy of model fitting. Furthermore, when component separation methods are used for contour analysis of the arterial pressure waveform, it is important to determine the best combination of fitting functions.

The aim of this study was to investigate the optimum combination of Gaussian functions that make up the arterial pressure waveform without any assumption about incident and reflection waves or any other physiological factor. Gaussian functions were used in this study because ventricular pressure induced by cardiac output has been shown to contain some Gaussian features [2]. We tested this modeling approach for both carotid artery pressure waveforms (CAPW) and radial artery pressure waveforms (RAPW) using between one and three Gaussian functions with different polarities.

2. Methods

2.1. Data acquisition

Twenty normal volunteers were enrolled in this study at Qilu Hospital of Shandong University (8 female and 12 male, mean age 51 years, range 32–73 years). The volunteers had not participated in any other 'clinical trial' within the previous three months. The basic clinical characteristics including age, height and weight were measured by an experienced operator. Manual auscultatory systolic and diastolic blood pressures (SBP and DBP) were also recorded from the right upper arm at the beginning of the signal recording. The mean arterial pressure (MAP) and pulse pressure (PP) were then calculated using the classic formula: MAP = DBP + (SBP − DBP)/3 and PP = SBP − DBP. The overall clinical information is summarized in Table 1.

All measurements were undertaken in a quiet, temperature-controlled measurement room (25 ± 3 °C). Before recordings started, the volunteer lay supine on a measurement bed for a 10 min rest period to allow cardiovascular stabilization. Electrocardiogram (ECG) electrode clamps were attached to the right wrist, left and right ankles to acquire a standard limb lead II ECG. Piezoresistive sensors were attached to the neck and the left wrist to acquire simultaneous CAPW and RAPW signals respectively.

For each volunteer, the ECG, CAPW and RAPW signals were recorded for more than 1 min and converted into digital signals simultaneously using a 16-bit data acquisition card (National Instruments, USA) at a sample rate of 1000 Hz. Subsequently, off-line analysis was performed using a custom designed computer program developed from MATLAB (Version R2009a, MathWorks Inc., USA). First, the slow varying components (0–0.05 Hz) were removed from the ECG, CAPW and RAPW signals. Second, the R-wave peaks of the ECG were detected using the Wavelet Transform Modulus Maxima method [19]. Ectopic beats were identified and excluded using our previously developed method [20]. After the location of R-wave peaks, the corresponding pulse feet (start of pulse) were found. Sola et al.'s method was used to detect the pulse feet [21], which was based on the parametric modeling of the rising edge of a pulse waveform. The pulse signals were then segmented between the starting points of two consecutive pulses. The first 10 successive pulse segments without ectopic beats were used for subsequent analysis. Using 10 pulses ensured the variation over a respiratory period was included. Fig. 1 shows an example of the three signals with the features identified. Because the recorded pulse waveforms varied in amplitude and pulse period (cardiac cycle length) between different subjects. Each pulse segment was then normalized in period and amplitude to allow comparison of the effectiveness of modeling between different combinations of Gaussian function, with the period to 1000 sample points, and the amplitude to unity between baseline and peak.

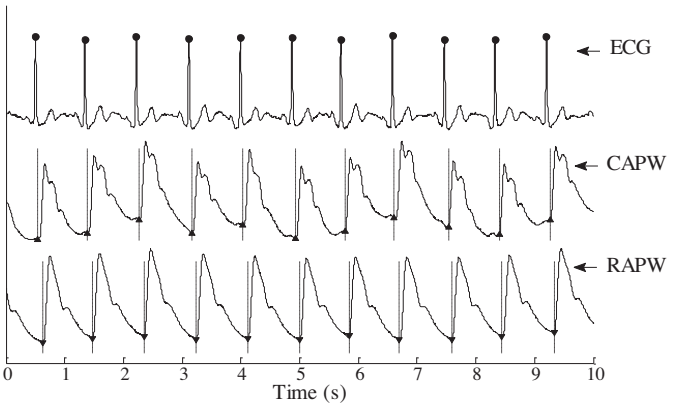

Fig. 1. An example of the simultaneous ECG, CAPW and RAPW signals. The detected R-wave peaks were denoted as "●", and the starting points of CAPW and RAPW signals were denoted as "▲" and "▼" respectively.

2.2. Waveform fitting with Gaussian functions

Each normalized pulse was decomposed into several independent Gaussian functions. The number of Gaussian functions employed ranged from one to three, since it is generally accepted that there can be up to three components to the arterial pressure waveform. These three Gaussian functions are denoted as $f_1(n)$, $f_2(n)$ and $f_3(n)$. Each Gaussian function $f_k(n)$ (k = 1,2,3) had 1000 points (n = 1,2, ..., 1000) and was determined by three independent non-dimensional parameters: waveform area A_k, halfwidth W_k and the center position C_k (within the range of 0–1000). The Gaussian function was defined as follows:

$$f_k(n) = \frac{A_k}{W_k \cdot \sqrt{\pi/2}} exp\left(-\frac{2(n-C_k)^2}{W_k^2}\right), \quad k = 1, 2, 3, \quad n = 1, 2, \ldots 1000 \tag{1}$$

The first Gaussian function $f_1(n)$ denotes the forward wave and $f_2(n)$ and $f_3(n)$ may denote reflection waves or other waveform features. The influence of different polarities for the second and third functions was tested. This resulted in seven sets of Gaussian functions (a single Gaussian function or the addition of two or three Gaussian functions). The number of functions for each pulse were 1, 2, 2, 3, 3, 3, 3, described as '+', '+−', '++', '+−−', '++−', '+−+' and '+++'. The symbol '+' denotes a positive Gaussian function