Medical Engineering & Physics 36 (2014) 1515–1520

Contents lists available at ScienceDirect

Medical Engineering & Physics

journal homepage: www.elsevier.com/locate/medengphy

Technical note

Pulse pressure waveform estimation using distension profiling with contactless optical probe

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ARTICLE INFO

Article history:
Received 17 September 2013
Received in revised form 4 July 2014
Accepted 28 July 2014

Keywords:
Arterial pulse pressure waveform
Distension profile
Pulse wave analysis
Cardiac catheterization
Non-invasive measurement
Optical probe

ABSTRACT

The pulse pressure waveform has, for long, been known as a fundamental biomedical signal and its analysis is recognized as a non-invasive, simple, and resourceful technique for the assessment of arterial vessels condition observed in several diseases. In the current paper, waveforms from non-invasive optical probe that measures carotid artery distension profiles are compared with the waveforms of the pulse pressure acquired by intra-arterial catheter invasive measurement in the ascending aorta. Measurements were performed in a study population of 16 patients who had undergone cardiac catheterization. The hemodynamic parameters: area under the curve (AUC), the area during systole (AS) and the area during diastole (AD), their ratio (AD/AS) and the ejection time index (ETI), from invasive and non-invasive measurements were compared. The results show that the pressure waveforms obtained by the two methods are similar, with 13% of mean value of the root mean square error (RMSE). Moreover, the correlation coefficient demonstrates the strong correlation. The comparison between the AUCs allows the assessment of the differences between the phases of the cardiac cycle. In the systolic period the waveforms are almost equal, evidencing greatest clinical relevance during this period. Slight differences are found in diastole, probably due to the structural arterial differences. The optical probe has lower variability than the invasive system (13% vs 16%). This study validates the capability of acquiring the arterial pulse waveform with a non-invasive method, using a non-contact optical probe at the carotid site with residual differences from the aortic invasive measurements.

1. Introduction

The profile of the arterial pressure pulse shows the result of the ventricular blood ejection over the arterial system, modulated by its mechanical properties. The load on the ventricle during ejection can be described by the systolic blood pressure. The arterial pressure wave travels from the central aorta to the peripheral arteries and several changes occur. The pulse pressure wave progressively steepens and increases in amplitude, while it loses the sharpness of the dicrotic notch [1].

Another effect that contributes to the pulse pressure profile is the time of arrival of the reflected pressure wave in the arterial system, during the cardiac cycle. This will depends on the pulse wave velocity (PWV) and the distance from the individual reflecting sites. One must note that the reflected waves of the most peripheral reflecting sites will arrive earlier at the larger peripheral arteries than at the central aorta [2]. The pulse pressure waveform, measured at the aortic site, is expected to be very similar to the waveform acquired in the carotid vessel (despite their proximity) but not equal, due to their different mechanical behaviour that results from the differences in the structural properties of the vessel walls [3,4].

The arterial pulse waveform is affected by changes in the peripheral circulation or alterations in the cardiac operation. Thus, the pulse wave analysis (PWA) could provide a better understanding of the risk factors, as well as help in establishing the extent of cardiovascular diseases, diagnosis and monitoring of the therapies effects [5–8].

Direct blood pressure monitoring with an arterial catheter is currently considered to be the most accurate method, but its invasive nature has several disadvantages. Great efforts have been made in the development of new techniques for non-invasive detection

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http://dx.doi.org/10.1016/j.medengphy.2014.07.014
1350-4533/

[7]. However, all of them establish direct contact with the patient's tissues at the artery site, which may distort the waveform. Then, the clinical application of a non-invasive and non-contact method can overcome practical and technical limitations inherent to the currently used methods such as arterial applanation tonometry, ultrasound and plethysmography. They require contact of the probe with the patient skin and compress the artery throughout an entire cardiac cycle, so the Bernoulli effect distorts the shape of the pulse curve [9,10].

For instance, in the tonometry the probe needs to be placed over the widest pulsation area, and requires support from solid structures such as the bone. For this reason, it is difficult to make valid measurements with tonometry over the carotid artery, since it is involved in soft tissues [11]. Moreover, arterial tonometry must be used with great caution in patients with carotid atherosclerotic plaques. Any risk of rupture attributable to the high external pressure, imposed by probe, should be avoided [12]. Therefore, the contactless technique is obviously safer in this aspect. Several patients also feel quite a discomfort during carotid applanation tonometry, especially when their pulsation is not easily assessed (i.e. obese patients).

In the current study, an optical non-contact probe developed for measuring the distension waveform of the carotid artery is compared with the pulse pressure waveform acquired by an intra-arterial catheter in the ascending aorta, during cardiac catheterization procedures, in order to validate the new optical device. This work was carried out with the aid of the Hemodynamic team from the Cardiology Unit of the Centro Hospitalar e Universitário de Coimbra (CHUC) and designed to determine the correlation between the two waveforms and analyse the differences obtained from PWA parameters, based on the time intervals of the cardiac cycle and the corresponding areas in the arterial pulse waveform of 16 subjects who had undergone cardiac catheterization [13,14].

2. Methods

2.1. Description of the optical system

The optical probe was designed to assess the arterial pulse wave profile at the carotid site, based on the optical reflectance fluctuations of the skin surface during the underlying pulse wave propagation [15].

The carotid artery is the natural probing site for the pulse waveform measurement due to the heart proximity and because it is easily accessible, i.e. it is close to the skin surface. The blood pressure wave travels across the arterial tree in a compliant way, forcing the blood vessels to distend elastically, according to the pressure wave profile, and causing a visible distension effect. This distension can be used to generate an optical signal correlated with the passing blood pressure wave. Several studies have shown that the distension waveform and the pressure waveform have an analogous wave contour and, therefore, can reciprocally be used for pulse wave analysis [16–18].

The functional structure of the optical probes allows the emission of the light and the detection of the reflected beam light. The box containing the optical probe, in an ergonomic configuration, ensures a non-contact signal acquisition at the artery site, by keeping a 3 mm distance between the probe and the skin surface. The obtained signals were digitized with a 16-bit resolution data acquisition system (NI, USB6210®) at a sampling rate of 20 kHz and stored for offline analysis using Matlab® tools.

Previously, a comparison test was carried out between an ultrasound image system GE *Vivid e*® (30 Hz), as a source of reference data, and the optical sensors that allow a higher resolution (20 kHz), adequate to feed feature extraction algorithms [16]. The clinical use of these optical probes includes vast applications, as they provide enough resolution for features extraction with higher accuracy, and they are an inexpensive and a non-invasive diagnostic method for the detection and monitoring of the pressure wave profile.

A larger study was performed in 131 young subjects and the results showed that the use of this new technique is a trustworthy method to determine PWV and PWA parameters using dedicated algorithms [19,20].

Previous studies have proved the good reproducibility of arterial pulse waveform acquired with the optical technique in 13 patients by 2 senior operators. The results show small Standard Deviation of Measurement (SEM) values for heart rate (operator A: SEM = 1.47, operator B: SEM = 1.79), Augmentation Index (operator A: SEM = 1.70, operator B: SEM = 1.89) and ETI (operator A: SEM = 1.96, operator B: SEM = 2.27) [21].

Table 1
Main characteristics of the volunteers.

Characteristics	Value
n, Males/females	16 (12/4)
Age, years	65.2 ± 12.3
Height, cm	163.3 ± 6.0
Weight, kg	72.3 ± 7.0
BMI, kg/m^2	27.1 ± 2.6

Values are numbers or means ±SD.BMI indicates the body mass index (weight/height^2).

2.2. Study population

The characteristics of the volunteered population in this study are presented in Table 1. This is a preliminary study for correlation of the two measures and not a statistical study. This is a prospective study. Similar studies of this kind present a sample of the same magnitude [14,22,23].

2.3. Study protocol

Measurements were performed in a study population of 16 patients, with a cardiovascular pathology, who had undergone cardiac catheterization. In all cases, simultaneous invasive (in the aortic root) and non-invasive measurements were performed. The study protocol was approved by the ethical committee of the CHUC.

Subjects were allowed to rest 15 min in the supine position at a temperature-controlled environment before the angiography proceeds. Each exam procedure consisted in the acquisition of a set of cardiac cycles at the carotid artery for 5 min with the optical system. The measurements were taken by a senior physician who had been trained to operate the optical probe device.

After the optical signal acquisition, the arterial catheter was used to monitor the blood pressure during the surgery and it was positioned to record the pulse pressure at the aorta site. In these tests a 6-Fr Judkins right catheter was used that was connected to a pressure transducer using a saline infusion system. After flushing and calibration the transducer, the hemodynamic polygraph was set to 200 mmHg/10 cm sensitivity and at a 100 mm/s registry speed. The used system was a Siemens® *Artis Zee* with AXIOM Sensis hemodynamic recording system. For waveform analysis data were resampled for 10 s at 200 Hz to cover, at least, two respiratory cycles.

2.4. Hemodynamic measurements

To compare the waveforms acquired with the two systems with different amplitudes and time duration, it is convenient to normalize both over amplitude and time. After this process, the dicrotic

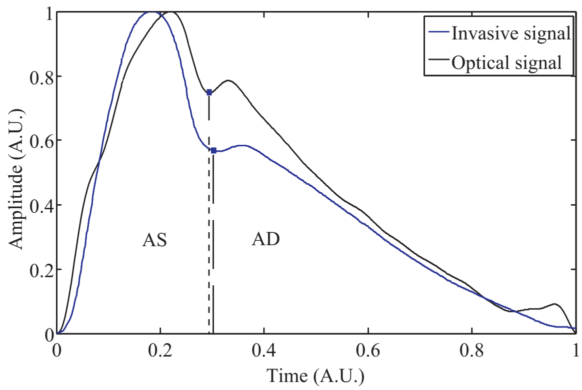

Fig. 1. Invasive aortic pulse pressure waveform and non-invasive carotid waveform, with systolic and diastolic periods marked in each pulse. The dicrotic notch is marked with a small square.

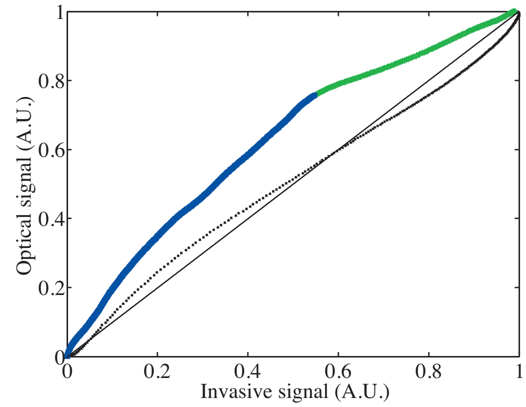

Fig. 2. Plot of the invasive signal vs optical signal. The upstroke of the pulse pressures is marked in black. The systolic period segment is comprised between the black and green segments; the diastolic period is in blue. The thin solid line represents the $x = y$ segment.

notch, which marks the end of systole, was determined for the extraction of parameters. The AUCs of the pulse waves represent the cardiac performance and the potential for coronary perfusion, enabling the cardiac workload [19]. The indices under analysis were the AUC, the AS and the AD, the ratio AD/AS and the ETI [14].

In a non-invasive approach of the global cardiac function, time intervals of the size of the cardiac cycle were used [13]. The ETI, also referred as Left Ventricular Ejection Time, is an important component for the evaluation of the left ventricular performance, determined by the ratio between systolic ejection time and the total duration of the cardiac cycle [24]. The ETI varies from 30 to 42% in healthy individuals [25].

2.5. Statistical analysis

Data is reported as mean ± the standard error, within 95% confidence intervals. Statistical analysis was performed by SPSS® software, version 19.0 (SPSS Inc., Chicago, Illinois).

3. Results and discussion

The pulse pressure waveform and the hemodynamic parameters under study were compared between the two acquired pulse waveforms (invasive and non-invasive). The mean value of heart rate (HR) obtained from the optical measurement data and the invasive measurement was 69 ± 8.80 bpm and 71 ± 7.81 bpm, respectively, with a mean difference between them of 5 bpm. HR value remained unchanged during the surgery procedures; consequently, the hemodynamic indices were also stable during the assessment period.

3.1. Comparison of the pulse pressure waveforms

The waveforms obtained with the optical system, placed externally over the carotid artery have a very similar contour when compared to the pressure curves obtained with a catheter, placed inside the aortic artery. These waveforms are normalized and presented in arbitrary units (A.U.) in Fig. 1.

The small effect of signal amplification in the carotid results from a phenomenon that occurs throughout the arterial tree. In clinical practice, an increase in the whole amplitude of the pulse pressure is recognized when it travels distally, i.e. a gradual widening of the pulse pressure between two sites of the arterial tree [26]. This amplification of the pressure wave with the distance (spatial amplification) does not require additional energy input in the arterial system, and so, by definition, is more a distortion than a true amplification, translated as an alteration in the morphology of the waveform [27]. Between two arteries that are close to each other, the distortion is very small but significant, due to the structural arterial differences as can be seen in the performed analysis [3].

The results show that invasive and non-invasive signals are correlated and have good overall similarity, for all patients, with a maximum of 20% RMSE. The mean value in all analysed samples was 13% for the RMSE. The coefficient of determination, R-squared value, for the profile of the signals under analysis was approximately 0.8 in all cases.

The strength of the correlation of pulse waveforms is expressed by the correlation coefficient. For the signals under study the obtained values were in the range of 0.920 to 0.996, thus demonstrating a strong correlation.

The signals frequency study showed great similarity in the modulus. The power spectral density analysis showed attenuation for the invasive signals and confirms the amplification effect in the optical signals. In the frequency window of interest for the analysed signals (below 100 Hz) there is a nearly constant attenuation for signals acquired with the invasive system for all patients. Therefore, there is no frequency modulation able to distort pulse waveforms in a significant way, in the scope of the presented analysis.

3.2. Comparison of hemodynamic parameters

The comparison between the AUCs allows the assessment of the differences between the phases of the cardiac cycle, systole and diastole. Fig. 2 shows a biparametric plot of the invasive signal versus the non-invasive signal. For both systems a mean of the pulse pressure from the sample in analysis is represented. When the two signals are the same, the result would be the bisector of the first quadrant, the line $y = x$. The plotted curve shows that in the upstroke of the waves, part of the systole period marked in black has a greater proximity to the $y = x$, which means that in this zone the signals are almost equal.

There is a close relationship between the optical (diameter)–invasive (pressure) slopes obtained for all subjects and represented by the mean curve in Fig. 2. However, comparing the change shown by the obtained slope some differences were detected.

After the systolic peak, differences between the two signals are more obvious. During the diastole, after the dicrotic notch (marked in blue), the differences are more substantial. This means that, in the diastole segment, the optical signal exhibits greater amplitude than the non-invasive signal because there is a higher difference between the structures of the two analysed waves.

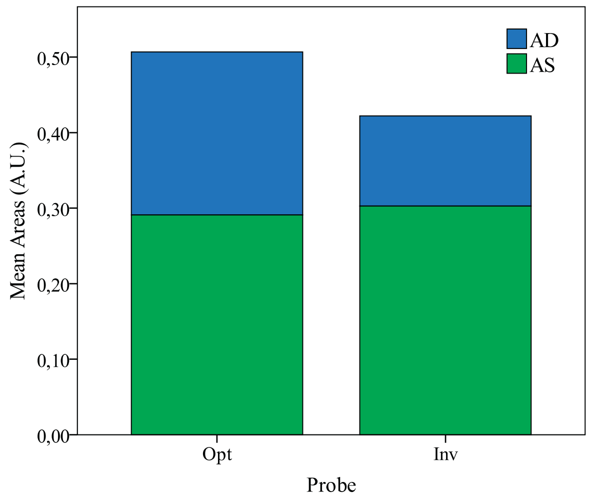

Fig. 3. Bar plot of AS and AD determined by the two systems: optical and invasive.

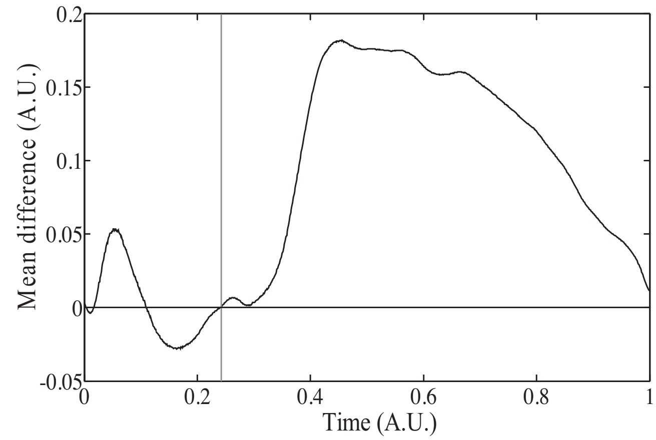

Fig. 5. Mean difference between the optical and the invasive signals during a heart cycle.

The arterial wall has both elastic and viscous properties. These differences in the pressure change–diameter change are related to the time-dependent response to the stress–strain relationship. In a purely elastic artery, this relationship is time-independent and, after the removal of stress, the arterial diameter dimensions would return to their initial dimensions. In the presence of wall viscosity, the arterial wall retains part of the deformation. The arterial viscous properties are responsible for part of the left ventricular energy dissipation characterized by the hysteresis of the pressure–diameter loop represented in Fig. 2 [22,28,29]. The surface between ascending and descending parts of the loop is hysteresis representing the energy dissipation due to the viscous properties of the arterial wall, which could indicate an alteration of the mechanical properties of the arterial vessel [30].

Arterial compliance reflects the ability of an artery to expand and shrink according to cardiac pulsation and relaxation, respectively [28]. The increasing age of the sample individuals at study leads to changes in the arterial system, worsening the structural mechanical components of the aorta walls, and reduces compliance. This change in compliance has significant implications in the pulse type contour analysis and contributes to the widening of the cycle hysteresis cycle verified in Fig. 2 [28,31].

Evaluating the whole AUC, there is an over-estimation of the values for the non-invasive method. The optical system presented an AUC mean value (±SD) of 0.51 ± 0.04, while the invasive system showed 0.42 ± 0.04, as represented in Fig. 3.

The area during the diastole is the responsible component for the difference between the AUCs (AS + AD) in the two systems. During the systolic period the AS mean values are very similar: 0.30 ± 0.05 for the optical system and 0.29 ± 0.04 for the invasive one. The variability of the measured values by the two systems, for each subject under study, was compared and is represented in Fig. 4. The greater variability was observed during AD.

The variability over the measurements was assessed by the variation coefficient, which is defined as the ratio of the SD of the measurements divided by their mean value. For the optical and the invasive systems, these coefficients are quite similar (13% vs 16%) but with lower variability for the optical system.

Fig. 5 represents the curve of the mean differences between the waveforms of the two systems for all subjects, during the cardiac cycle. The analysis of the pressure pulse samples shows that the average systolic peak occurs approximately at 0.2 s (marked in a grey line). After that instant, the difference begins to be more significant, as a result of the energy dissipation, due to viscous properties of the arterial wall. After the removal of the blood pressure (stress on the arterial walls), the arterial diameter would return to almost

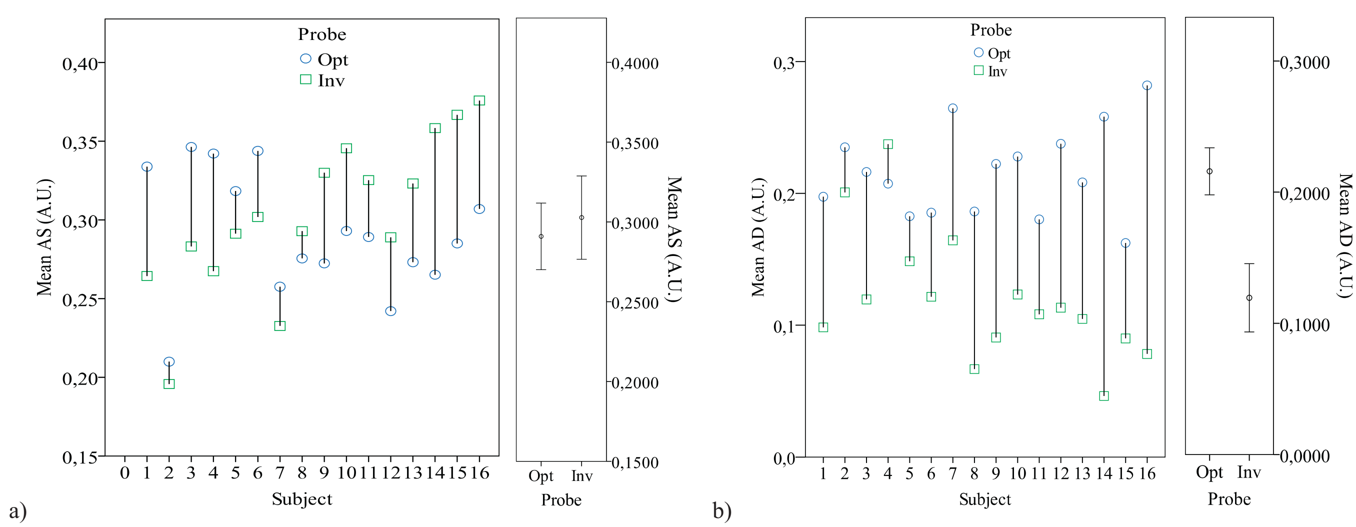

Fig. 4. Variability of the two measurement systems for each individual subject and the mean values according to the acquisition method, with error bars, representing the 95% of confidence interval for the sample mean. (a) Variability in AS. (b) Variability in AD.

their initial dimensions because the arterial wall retains part of the deformation.

The average difference between the waveforms from the two systems, optical and invasive, along the cardiac cycle is 8.75%. A Pearson correlation analysis between the areas of signals shows a significant correlation of 0.847 at a 0.01 level (2-tailed).

The ratio AD/AS demonstrates, again, the differences (optical system 0.76 ± 0.19; invasive system 0.42 ± 0.25) explained by the discrepancies originated during the diastole.

The ETI values obtained are correlated, but there is a trend to have systematic lower values from the optical probe device. The ETI mean value (±SD) is $45.64 \pm 7.80\%$ for the invasive measurements and $41.40 \pm 6.53\%$ for the optical signals. The two obtained values are slightly higher than the expected ones. However, patients with systolic dysfunction have been found to have ETI higher values. The prolonged ejection time is primarily due to the increased afterload [24,32].

4. Conclusions

The comparative study has been carried out using some of the most important clinical parameters. The used parameters evidence the differences between the structures of the waves. Nevertheless, the blood pressure and pulse pressure are the most widely accepted parameters for the analysis of the arterial system. The calibration method is, currently, still a challenge for the optoelectronics probing techniques such as the one hereby presented [33].

The major limitation of this type of technology (optical system) is the inability to direct calibration of the waveform obtained by the optical probe. Moreover, there is a limitation of the comparison procedures (optical and catheter) that is related to the fact that it is not possible to synchronize the signals obtained by the two systems. To eliminate these limitations the data were normalized over both, amplitude and time scale.

The invasive system measured the aortic pressure waveform and the optical method measured the pulse waveform by the carotid artery. There are small differences in the profile of the pulse pressure waveform of these two vessels that are the consequence of the intrinsic differences in their mechanical structural properties.

This study emphasizes the effect of the energy dissipation due to the viscous properties of the arterial wall that occurs during the heart cycle characterized by the hysteresis in the pressure–diameter loop. This effect is visible at the arterial vessel, as the diameter dimensions would return to initial dimensions. In the presence of wall viscosity, the arterial wall retains a portion of the deformation that results in part of energy dissipated by the arterial viscous properties.

The displacement of the pressure–diameter loop, indicating changes in the vessel function, can be used to show the response of a large elastic vessel to the blood pressure changes within a cardiac cycle and thus on large elastic arterial mechanical properties.

The results show that the waveforms obtained by the two systems have a strong correlation and the mean values of the areas during the systolic period are very similar. The diastole time period presents the biggest differences between the two waves, which might be explained by the mechanical energy dissipation during the heart pulse cycle.

This study, while preliminary in nature, validates the capability to estimate the arterial pulse waveform in the carotid vessel using a non-invasive way, with the novel contactless optical probe. The differences between the two systems are due to the different measurements being studied: vessel diameter and blood pressure.

The cohort size is 16 subjects but it is enough to draw conclusions about the correlation between the vessel distension profile and the pulse pressure waveform. However, the present results and their clinical implications need to be confirmed by larger studies. Those studies must include the population samples with heterogeneity characteristics such as age, blood pressure and BMI.

Declarations

Funding: support from Fundação para a Ciência e a Tecnologia (FCT) for funding the PhD grant (SFRH/BD/79334/2011).

Ethical approval

The study protocol was approved by the ethical committee of the CHUC, Portugal. All subjects were volunteers and gave a written informed consent.

Conflicts of interest statement

No conflicts of interest are declared for this paper.

Acknowledgments

The authors acknowledge the clinical collaboration of the Hemodynamic Team from the CHUC of Cardiology Unit.

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