**Parametric Imaging for the assessment of Left Ventricular wall motion: A Review**

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**Abstract**

The assessment of wall motion abnormalities such as hypokinesia or dyskinesia and the identification of their extent as well as their degree of severity allow an accurate evaluation of several ischemic heart diseases and an early diagnosis of heart failure. These dysfunctions are usually revealed by a drop of contraction indicating a regional hypokinesia or a total absence of the wall motion in case of akinesia. The discrimination between these contraction abnormalities plays also a significant role in the therapeutic decision through the differentiation between the infarcted zones, which have lost their contractile function, and the stunned areas that still retain viable myocardial tissues. The lack of a reliable method for the evaluation of wall motion abnormalities in cardiac imaging presents a major limitation for a regional assessment of the left ventricular function. In the past years, several techniques were proposed as additional tools for the local detection of wall motion deformation. Among these approaches, the parametric imaging is likely to represent a promising technique for the analysis of a local contractile function. The aim of this paper is to review the most recent techniques of parametric imaging computation developed in cardiac imaging and their potential contributions in clinical practice.

***Keywords:*** *parametric imaging, quantification, cardiac imaging, wall motion abnormalities.*

**INTRODUCTION**

The regional assessment of wall motion is a crucial step for the evaluation of left ventricular function in order to detect different contraction abnormalities and segmental asynchrony [1,2]. In addition, the methods dedicated to the assessment of segmental left ventricular (LV) wall motion play an important role in the identification of dyskinesia and aneurism that represent markers of many cardiac pathologies such as the Wolff Parkinson White's (WPW) syndrome and Left bundle Branch block [3,4]. In recent years, several studies have shown that the detection of wall motion abnormalities may help in the therapeutic decision by providing a valuable discrimination between infarcted areas in the myocardium that lost their contraction ability and areas at risk able to recover their contraction after revascularization [5,6].

Different cardiac imaging modalities are used for the analysis of LV function, such as echocardiography, nuclear medicine, cardiovascular computed tomography, cardiac magnetic resonance Imaging. The clinical parameters usually computed in clinical routine to assess ventricular function are the cardiac output index, the end diastolic volume (EDV), the end-systolic volume (ESV), the stroke volume (SV), the LV ejection fraction (LVEF), and the myocardial mass [7-9].

The role of the left ventricle is to ensure that the blood stroke volume ejected in each beat is well adapted to the needs of the body. In case of a drop in contractility or in presence of LV dysfunction, the EF is reduced below normal ranges that in healthy subjects are between 55% and 70% [10]. However, these values do not always indicate a normal LV wall motion, as EF represents a global functional parameter. In some cases, normal EF values may mask a regional LV dysfunction. For this reason, the assessment of normal EF values is currently a matter of debate and needs to be re-examined for some cardiac pathologies such as mitral regurgitation, where EF remains preserved [11, 12].

In addition to the global parameters described above, for LV wall motion assessment the radiologist or the cardiologist establishes his/her diagnosis by visually following the dynamic behavior of the myocardium in terms of changes in its thickness, and of synchronism between the different sectors of the LV [13,14]. Moreover, for a semi-quantitative assessment of the LV wall motion, the myocardium is divided into 16 segments according to the American Heart Association (AHA) standard (six segments for both basal and mid ventricular levels and 4 segments for the apex) [15]. A visual wall motion scoring is established by the radiologist/cardiologist using a 4 point-scale with: 1 indicating a normal wall motion, 2 designating an hypokinetic area, 3: an akinetic area, and 4: a dyskinetic area. The final score is computed by dividing the sum of the scores of all analyzed segments by the number of these segments [16, 17].

Several researches in literature have shown that the current visual analysis used in clinical practice suffers from inter and intra-observer variability, as the ability to detect and recognize a LV wall motion abnormality relies on the degree of expertise of the clinical observer, as well as on the severity of the abnormality [18]. Cong et al [19] demonstrated that the reproducibility of LV wall motion analysis is highly dependent on the image quality. Other studies, investigating the reproducibility of regional LV evaluation, revealed that the accuracy of LV parameters (LV volumes, EF, regional fractional area change…) in MRI as well as in echocardiography is influenced by the precision of myocardial contours delineation [20,21]. Furthermore, the measurement of myocardial deformations such as strain and strain rate using tissue Doppler method or other 2D ultrasound techniques is still angle dependent which may result in underestimating the motion [22]. These constraints represent a common limitation in the study of left ventricular function by different cardiac imaging modalities.

In order to improve the accuracy and the repeatability of LV wall motion interpretation, semi-automated or automated methods are needed to generate additional parameters to be used in conjunction with visual approach. In this context, the regional assessment of the LV function has been the main topic of several researches in the literature. The developed methods can be classified into three main families: the first includes all the techniques dedicated to the automated or semi-automated detection of myocardial contours in order to provide a quantitative measurement of regional functional parameters, such as regional fractional area change (RFAC) or regional myocardial thickening [23]. These features were used in echocardiography and MRI as an index of the performance, thus providing information about the regional LV function both in systole and in diastole. To obtain the RFAC values, the LV cavity is divided into different segments according to AHA standard [15]. For each segment, RFAC is defined as a percentage of the difference between end diastolic and end systolic areas divided by the end diastolic area. This parameter is computed after outlining the endocardial contour of the myocardium on 2D echocardiography or from cine-MRI images of the end diastolic and end systolic phases. The RFAC value is represented by a specific color using a color kinesis method or other techniques were the color degree reflects the timing and the extension of endocardial motion in that region [24, 25].

The second family dedicated to the assessment of LV wall motion is relevant to the cardiac deformation tracking approaches based on optical flow methods [26-28]. The former used the intensity or the phase information to track and measure LV displacement between two consecutive frames. These techniques fall beyond the scope of this review.

As a third kind of approaches, parametric imaging represents an additional promising tool allowing an improvement in objective analysis of regional LV wall motion, free from contour tracing and to be used in conjunction with visual assessment. In this review, a description of the different parametric imaging approaches utilized for improving the assessment of LV wall motion by combined assessment with dynamic images in visual examination, applied to MRI and echocardiography techniques, will be presented.

**Parametric Imaging methods**

The main concept of parametric imaging consists in summarizing the information included in a sequence or a series of acquired dynamic images into a reduced number of synthetic still frames (usually one or two images) [29,30]. The generation of parametric images requires the selection and the computation of a physical parameter allowing the assessment of the cardiac wall motion. This parameter is computed from each pixel videointensity by applying mathematical transformations [31].

Parametric imaging applied to cardiac imaging was first applied in nuclear medicine and more specifically in the Gated single photon emission computed tomography (SPECT) examination [32]. The main goal of parametric image computation was to improve the performance of the regional assessment of LV wall motion in Gated SPECT examination. This later requires the acquisition of several hundreds of cardiac cycles synchronized with the ECG to obtain images (a minimum of 16 frames is required) representing the different moments of the cardiac cycle [33], where the visual interpretation of each image increases significantly the processing time..

The computation of parametric images in Gated SPECT examination is based on the monitoring of the radiotracer evolution in the acquired image over the cardiac cycle. The variation of this activity reflects the change in left ventricular volume and perfusion during the different phases of the cardiac cycle. From these curves, it is possible to extract physiological parameters that can be synthesized in a reduced number of images called "parametric images" or "parametric maps".

The use of this technique as a tool for the functional study of the left ventricle in clinical practice of nuclear medicine has known a great success given its contribution to the analysis of wall motion abnormalities such as hypokinesia and akinesia associated with several cardiac pathologies [34,35]. However, several limitations inherent to the nuclear medicine modality were present. The main limitation is related to the invasive nature of nuclear medicine examination. Other pitfalls are associated to technical problems and acquisition protocol including low count scans, time of imaging acquisition, background activity and the spatial resolution of gamma cameras for the Gated SPECT examination [36].

To overcome these limitations, several studies aimed at testing parametric imaging approach on other non-invasive imaging modalities, such as echocardiography and MRI.

In this context, Cardiac Magnetic Resonance Imaging (CMRI) is considered the reference modality for the assessment of the LV function thanks to its non-invasive nature and to the fast MRI sequences allowing a dynamic analysis of one slice of the heart in one apnea, with high spatial resolution. The images, conventionally acquired along three planes of view (in long axis, one slice each representing the apical 4- and 2-chambers view, and in short-axis, by multiple parallel planes) allow a complete visualization of the heart useful for visual interpretation of LV wall motion and quantification of its function. In clinical practice, to compute the required parameters for LV clinical assessment [37-39], a post-processing software (Argus) available in the acquisition console is used to delineate the endocardial and epicardial contours of the myocardium in each slice over time of the cine MRI images (in short axis view) covering the whole left ventricle. In addition to CMRI, the echocardiography and more specifically the 2D echocardiography has long been regarded as the commonly used modality for the computation of global and regional parameters and the study of left ventricular function thanks to its simple protocol and its low cost [40-42].

Different parametric imaging methods were proposed as a tool for the regional assessment of LV wall motion using echocardiography or CMRI modalities. Some of them are already applied in nuclear medicine, while other techniques were focused on the echocardiographic and the CMRI imaging. In the next section, we will present an overview of the different studies in the literature that have adopted the parametric imaging technique in echocardiography and CMRI for the assessment of regional wall motion abnormalities.

### **Parametric images based on Covariance Analysis**

Covariance analysis is a statistical tool that allows the tracking of motion over time from a set of image data. Used for the first time in nuclear medicine and then applied to cardiac MRI, this technique allows the generation of a parametric image able to quantify the temporal behavior of the cardiac wall motion [43,44]. The principle of this technique is based on the computation of the similarity degree between the intensities of the pixels located in a region of interest called the reference region and those of the pixels located in the cine-MRI images. The contraction behavior in this reference region must be perfectly identified in order to obtain correct values ​​of the similarity degree.

The degree of similarity calculated for each pixel in the cine-MRI images is likely to provide additional information about the behavior of myocardial contraction. The degree of covariance for each pixel is computed using the following covariance function [43]:

$Cov\left(i,j\right)= \frac{1}{T}\*\left( A\left(i,j\right)-µ\_{A}\left(i,j\right)\right)\*\left( R\left(t\right)- µ\_{A}\right)$ (1)

where A (i, j) is the pixel value in the image, T is the number of frames in the Cine-MRI sequence and $µ\_{A}$ (i, j) is the mean of the pixel values (i, j) in the sequence series. R (t) is the average value of all pixels in the reference region and $µ\_{A}$ is the average value of all reference regions in the image sequence.

The calculation of covariance value Cov(i, j) for each pixel makes it possible to generate a mapping image in which the value of each pixel represents the temporal similarity with respect to the reference region. The set of the calculated similarity parameters results in a functional image able to quantify the temporal change between the different sectors of the myocardium. The sign of the covariance value indicates the degree of the contraction similarity between two areas. If it is positive, the contraction is synchronous and it follows the same direction. If the sign is negative, the two zones are out of phase. A typical example of this phase shift is the atrial and ventricular motion that follow opposite movements (i.e., the atrium is relaxing while the ventricle is contracting). When the covariance value is equal to zero, the two areas contract independently.

Although several studies in nuclear medicine have used this statistical tool for the calculation of parametric images, the integration of this tool in cardiac MRI is still limited due to some constraints. First, the quality of the parametric images obtained by covariance is very dependent on the size and the choice of the reference region [44]. Usually, this region is chosen as being small in order to preserve its homogeneity. The main difficulty is associated with the selection of the region in the image that will be used as a reference. The evolution of the movement at the level of the reference area must be known a priori in order to have exact measurements of the degree of similarity between the intensities in this region and those of the other pixels of the image [43]. The following figure shows an example of a parametric image computed from cine MRI images using the covariance analysis:



1. (b) (c)

**Fig.1** End-diastolic (a) and end-systolic (b) images of a cine- MRI sequence at the basal level in a patient diagnosed with myocardial infarction in the antero-septal and infero-septal walls, together with the corresponding parametric amplitude image (c) based on covariance analysis. (The white arrows highlight the extent of the wall motion abnormality).

**Parametric Analysis of Main Motion Method (PAMM)**

The parametric analysis of the main motion (PAMM) has been applied to both echocardiography and MRI imaging, and consists in extracting physiological features reflecting the variation of the myocardial movement and the time of contraction from the image sequence corresponding to a cardiac cycle [45]. This approach is based on a window function, allowing the modeling of the variation of the signal intensity for each pixel of the image [46]. The PAMM exploits the difference in the intensity levels between the myocardium and the cavities on the cine-MRI or 2D echocardiograph images as well as the variation of the intensities of the pixels between the contraction and the relaxation times to estimate the myocardial movement. The gray level distribution of a pixel (x, y) located at the level of the myocardium during a cardiac cycle of duration "t" is denoted P (x, y, t). It is then possible to model this variation by a window function according to the following equation [47]:

$P\left(x,y,t\right)=A\_{B}\left(x,y\right)-A\_{v}\left(x,y\right)\*g\left(t, T\_{on}\left(x,y\right), T\_{off}\left(x,y\right)\right)+e\left(x,y,t\right) $(2)

where $A\_{B}\left(x,y\right)$ corresponds to the mean value of the gray level at the end of diastolic frame and $A\_{v}\left(x,y\right)$ is the variation of the signal intensity between the end-diastolic and end-systolic frames. The function g (t) represents the window function, which depends on the contraction starting time$ T\_{on}\left(x,y\right)$ and on the end of contraction time $T\_{off}\left(x,y\right).$ The last term e(x, y) corresponds to the residual error to be minimized.

Using this mathematical equation, four physiological parameters associated with motion and time can be extracted: AB (x, y), Av (x, y), Toff (x, y) and Ton (x, y). The computation of these parameters for each pixel located at the level of the myocardium leads to the reconstruction of four parametric images: two for amplitude parameters and two for temporal parameters.

From the modeling by a window function and applying an appropriate algorithm, the detection of the discontinuities of the curve, which reflects a transition from one state (systolic phase) to another (diastolic phase), is made possible for any pair of values (Ton, T off) such that:

1≤ Ton ≤ T off ≤Ni,

with Ni the total number of frames.

The two parameters AB and (AB-Av) can be computed, which respectively represent the number of points for which the window function takes the intensity values ​​AB and AB-Av:

$A\_{B}^{T\_{on}, T\_{off}}= \frac{1}{n\_{1}}\* \sum\_{t \notin ]T\_{on}, T\_{off}] }^{}P\left(x,y,t\right) $(3)

$(A\_{B}-A\_{V})^{T\_{on }, T\_{off}}= \frac{1}{n\_{2}}\*\sum\_{t ∊ ]T\_{on}, T\_{off}] }^{}P(x,y,t)$ (4)

where $n\_{1}= N\_{i}-n\_{2}$ and $n\_{2}= T\_{off}- T\_{on}$.

After calculating the four parameters related to the amplitude and time, the term which minimizes the error e (x,y,t) over the entire cardiac cycle is then defined using the following expression using the least squares method:

$e\_{(T\_{on}, T\_{off)}}=\sum\_{t=1}^{N}P\left(x,y,t\right)^{2 }–(n\_{1 }A\_{B }(T\_{on }, T\_{off })^{2 }+ n\_{2 }(A\_{B -}A\_{v})(T\_{on }, T\_{off })^{2 })$(5)

The process is repeated for each pair (Ton, Toff) of values, and the converging solution is that one resulting into the minimum error. After computing the amplitude images, these latter are combined into a single image using three colors: red, green and blue.

The parameter AB is coded by the green color while the coding of the parameter Av is dependent on the movement orientation of the endocardial wall. It will be coded by the red color for the pixels having a decreasing signal in systole and then increasing in diastole, thus reflecting a movement oriented towards the inside of the cavity, whereas it will be coded by the blue color if the movement is directed outward the cavity [45]. Similarly, the images Ton and Toff will be combined to form the mean transition image denoted Tm, corresponding to the average of these two parameters. The coding of this image is dependent on the sign of the parameter Av in order to distinguish the relaxation time from that of the contraction.

The two images must be analyzed together on the basis of two main criteria: the thickness of the myocardial walls and the color coding. Reading and interpreting the final image is subject to well-defined rules. A normal contraction is manifested by a large red wall inside the cavity whereas a hypokinesia is revealed by a thin red wall inside the cavity.



**Fig.2** Examples of mean transition time images obtained on the apical 4-chamber view (A4C), apical 2-chamber view (A2C), and apical 3-chamber view (A3C) in a patient with normal wall motion (top), a patient with RWM abnormalities (middle), and a patient with global LV dysfunction due to dilated cardiomyopathy (bottom). Adapted from [69] with permission.

**Parametric images based on Fourier analysis**

The parametric image based on Fourier analysis uses the gray-level properties that characterize cine-MRI or 2D echocardiography images and the natural contrast between the myocardium and the cavities for the estimation of the ventricular movement. Indeed, the analysis of the distribution of the gray levels for each pixel through the various images can play the role of an estimator of the variation of the ventricular volume during the cycle. Pixel positions representing the myocardium in same frames and the LV cavity in others are characterized by a big variation of videointensity over time that reflects the spanning of the LV endocardium during the cardiac cycle. On the other hand, pixel positions representing the LV cavity in each frame of the cardiac cycle are characterized by curves of videointensity with little variation [48-51].

 Fig.3 shows an example of this behavior in sequence of cine MRI images for two pixels located in different regions of the heart. The first pixel is located in the left ventricle near the endocardial wall (red pixel), while the other pixel is located in the center of the left ventricular cavity (blue pixel). Their

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**Fig.3** Example of videointensity over time computation for two pixels located in two different areas of the LV. Cine MRI image at end-diastole (a) and at end-systole b); time intensity curves for a pixel located in a region close to the LV endocardium (in red) spanned by its motion during the cardiac cycle (c) and for a pixel (in blue) in the center of the LV cavity (d).

From this observation, it is possible to formulate the following hypothesis: the evolution of the videointensity signal over time for each pixel during the cycle plays the role of an indicator of the local variation in left ventricular volume over time. Based on this hypothesis, for each pixel of the cine-MRI image, a temporal curve is extracted following the variation of gray intensity over the cardiac cycle. Then, each temporal series will be approximated by a sum of sine or cosine function, based on the hypothesis that the signal is periodic in the cardiac cycle. Therefore, each signal can be approximated by a sum of sinusoidal functions according to the following equation [52]:

$f\left(t\right)= a\_{0}+\sum\_{n=1}^{\infty }(a\_{n}\*\sin((nx +b\_{n})) )$ (6)

with f (t) is a signal of any form, but periodic with period T, and a0, an and bn are the Fourier coefficients. In order to retain the model which is the closest possible to the real curves, an iterative process is applied to minimize the error between the real and the estimated parameters based on several approximation techniques, such as the least squares methods. The following figure shows examples of intensity-time curves with approximations by sinusoidal functions [53].



(a)

**

(b)

**Fig.4** Example of intensity curves before and after fitting with a sine function using the least-square function for (a,b) two pixels located in different areas of the heart. Adapted from [54]*.*

Each obtained time curve is approximated by a sinusoid and a quantitative parameter: the amplitude or the phase is extracted from these curves to define the parameter from which it is possible to generate a parametric image [54,55]. In addition to amplitude and phase features, other previous studies used a normalized amplitude parameter defined as the amplitude of the videointensity curve of each pixel divided by its mean value to generate a parametric amplitude image able to assess the hypokinetic and akinetic segments of the myocardium [56].



1. (b) (c) (d)



 (e) (f) (g) (h)

**Fig.5** The end-diastolic (a,e) and end-systolic images (b,f) of a cine-MRI sequence respectively for a healthy subject and for a patient diagnosed with myocardial infarction at the level of anterior wall. The corresponding parametric amplitude without (c,g) and with (d,h) the schematic division into six sectors (Anterior (A), Anterolateral (AL), Inferolateral (IL), Inferior (I), Inferoseptal (IS) and Anteroseptal (AS)) According to AHA standard, (d,h). (The red arrows specify the extent of the hypokinesia). Adapted from [54]*.*

The parametric images computation based on Fourier analysis suffers from several limitations. First, the computation of these latter images is based on the periodicity assumption of the signals. However, the periodicity or the non stationarity of the cardiac signals is still at present an open subject of discussion [57]. Indeed, if we only rely on the mathematical formula of the periodicity, we can affirm that these signals are considered as periodic since they all have the same period "T" which corresponds to the duration of the cardiac cycle. From another point of view, the signals extracted from the cine-MRI images for each pixel are the approximation of the variation of the ventricular volume during the cardiac cycle. The stationarity of the cardiac cycle is itself a subject of debate. To overcome this ambiguity regarding this hypothesis, further studies are needed.

Another limitation of the parametric images based on the Fourier analysis is the requirement of a well-defined geometric form. Indeed, Fourier analysis is the modeling of signals by a sum of sinusoidal functions or by a set of sine and cosine functions [58]. These functions are periodic and symmetrical whereas this is not the real case for the signals extracted from the cardiac MRI images. These signals are mostly asymmetrical as the duration of the diastole exceeds that of systole in a cardiac cycle. The sinusoidal approximation can then lead to errors due to the non-conformity of the shape of the real signals with those estimated by a sinusoid.

**Parametric images based on the Hilbert Transform**

To overcome constraints of the Fourier approach, a new parametric imaging method based on the Hilbert transform has recently been proposed by Benameur et al [59]. This tool is well suited to periodic and non-periodic signals and thus makes it possible to overcome the stationarity limit imposed in previous work on the extracted signals.

The concept of the analytic signal was initially introduced by Gabor [60] as a one-dimensional complex signal. This latter is calculated from the real signal by canceling the negative frequencies of its spectrum and multiplying by two the amplitude of the positive frequencies so as to conserve the energy of the signal. The equation of the analytic signal denoted sA (t) is the following [61]:

$s\_{A}\left(t\right)=s\left(t\right)+j [s\_{H}\left(t\right)]$ (7)

Where s (t) is the real signal and $s\_{H}\left(t\right)$ is the Hilbert transform of the real signal s (t) defined in the time domain by the following equation [62]:

$s\_{H}\left(t\right)= \frac{1}{ πt} \* s (t)$ (8)

The Hilbert transform sH (t) can be represented in the frequency space in this way:

$s\_{H}\left(u\right)=s\left(u\right)\* (- i sign(u)) $ (9)

With s (u) is the Fourier transform of the real signal s (t). The sign of u is defined such that:

-1 si u < 0

0 si u = 0

1 si u > 0

Sign (u) =

The analytic signal is composed by a real component, which is the signal itself, and an imaginary component that corresponds to a frequency shift of Π/2 with respect to the original signal [63]. This analytical representation is well adapted to the stationary and non-stationary signals and represents a useful and efficient tool for the calculation of the module and the argument of the random signals.

From the obtained analytical curves, it is possible to extract the well-defined physical parameter as the amplitude (AI) or the instantaneous phase using the following equations [64]:

 $A\_{I}\left(t\right)=\left|s\_{A}\left(t\right)\right|= \sqrt{s(t)^{2}+ s\_{H}(t)^{2} } $ (10)

$Փ\_{I}(t)=\arctan(\frac{s\_{H}\left(t\right)}{s\left(t\right)})$ (11)

From these equations, it is possible to calculate a quantitative parameter for each pixel located in the region of interest. A color-coding is used to represent all the calculated values in the form of a parametric image in which each color reflects the level of contraction in a well-defined region. Fig.6 shows an example of a parametric amplitude image calculated using the Hilbert transform:



1. (b) (c)

**Fig.6** The end-diastolic (a) and end-systolic (b) images a cine- MRI sequence in a patient diagnosed with myocardial infarction related to the septal wall, together with the corresponding parametric amplitude image (c). Adapted from [59].

The parametric images based on the Hilbert transform represents a promising tool for the regional assessment of wall motion abnormalities such as hypokinesia and akinesia associated with some cardiac pathologies. In addition, the use of the Hilbert transform to extract instantaneous amplitude has proved its ability to generate a parametric amplitude image from stationary signals as well from non-stationary signals. However, this method needs to be validated in larger populations with different cardiac pathologies. The main limitation of the parametric image based on the Hilbert transform is inherent to the presence of trabeculations in the myocardium, which is likely to confound the wall motion interpretation. Finally, as with other parametric imaging techniques used for the assessment of LV wall motion, the interpretation of the parametric image using Hilbert transform requires a training before its use in clinical practice.

**DISCUSSION**

In recent years, several studies have been dedicated to the regional evaluation of LV wall motion. Among the methods proposed in literature, parametric imaging has emerged as a promising tool allowing an objective measurement of regional parameters able to assess the LV wall motion abnormalities, when used in conjunction with conventional visual interpretation. This technique is based on the computation of physiological features (such as amplitude, phase, covariance, value...) from signal intensity curves extracted from a series of images that represent the different frames of the cardiac cycle. These quantitative parameters are used to generate parametric images able to assess the extent and the degree of severity of wall motion abnormalities relevant to different cardiac pathologies. Several studies showed the accuracy of the parametric amplitude image in the detection of hypokinesia, manifested by a drop in regional contraction. Similarly, this image is able to quantify the extent of akinesia. In addition to the parametric amplitude image, other parametric images based on the phase features play an important role in the assessment of the extent of dyskinesia in myocardial segments, which is revealed by an opposite movement during the contraction. The review of literature reveals the existence of a several techniques for the computation of parametric images, with the common purpose of improving the accuracy and reducing inter- and intra-observer variability in the visual assessment of LV wall motion abnormalities, in particular for less experienced readers. Table 1 shows an overview of the literature on this topic, where the technique used to generate parametric images, the imaging modality, the number of patients studied, and the accuracy, sensitivity, specificity and variability of the interpretation, without and with parametric images is summarized.

**Table 1.** Overview of parametric imaging techniques proposed in literature; n= number of patients ; nw = number of women ; nm = number of men ; PI = parametric image

|  |  |  |  |  |  |  |  |
| --- | --- | --- | --- | --- | --- | --- | --- |
|  | **Cardiac imaging modality** | **Parametric Imaging technique** | **Number of patients** | **Specificity****(%)** | **Sensitivity****(%)** | **Accuracy****(%)** | **Variability inter-observator** |
| WithoutPI | WithPI | WithoutPI | WithPI | WithoutPI | WithPI | WithoutPI | WithPI |
| *Boudraa et al, 1999[65]* | Nuclear medicine | Covariance Analysis method | n = 21 | - | - | - | - | - | - | - | - |
| *Caiani et al, 2002 [51]* | Echo | Fourier Analysis | n =30nw = 12nm = 18 | 59\* | 83\* | 86\* | 87\* | 68\* | 83\* | 54\* | 32\* |
| *Caiani et al, 2004 [66]* | CMRI | Fourier Analysis method | n =18nw = 8nm = 10 | 76 | 80 | 80 | 83 | 77 | 81 | 52 | 33 |
| *Ruiz et al, 2005 [45]* | Echo | PAMM method | n = 10 | - | - | - | - | - | - | - | - |
| *Caiani et al, 2006 [67]* | CMRI | Fourier Analysis method | n =28 | 76 | 77 | 80 | 84 | 77 | 79 | - | - |
| *Redheuil et al, 2007[48]* | CMRI | PAMM method | n =33nw = 6nm = 27 | 95 | 97 | 90 | 94 | - | - | 15 | 9 |
| *Kachenoura et al, 2007[68]* | CMRI | PAMM method | n =22nw = 2nm = 20 | - | - | - | - | - | - | - | - |
| *Kachenoura et al, 2009[69]* | Echo | PAMM method | n = 45 | 60\*\*\* | 74\*\*\* | 92\*\*\* | 96\*\*\* | 75\*\*\* | 84\*\*\* | - | - |
| *Kachenoura et al, 2010[70]* | Echo | PAMM method | n=37nw = 9nm =28 | - | - | - | - | - | - | - | - |
| *Benameur et al, 2017[59]* | CMRI | Hilbert transfom method | n = 20nw = 9nm =11 | 92\*\* | 96.9\*\* | 64.3\*\* | 84.4\*\* | 87.2\*\* | 94.6\*\* | 12.8\*\* | 5.6\*\* |

**(\*)** study interpretation made by three inexperienced readers in comparison with the gold standard.

**(\*\*)** study interpretation made by two radiologists with little experience in comparison with the gold standard.

**(\*\*\*)** study interpretation made by one inexperienced reader in comparison with the gold standard.

Most of studies on parametric imaging reported in literature show a significant improvement in the accuracy of regional detection of wall motion abnormalities by adding the parametric images to the clinical visual interpretation of dynamic MRI and echocardiography examinations [45-49]. The outcome of these studies demonstrate that the interpretation of standard images in conjunction with the parametric images is likely to increase the detection of abnormal segments allowing a significant improvement in terms of sensitivity, specificity, and accuracy. Another important finding is the reduction of inter-observer variability by adding the parametric images to the standard protocol of the LV wall motion interpretation. Indeed, several studies show the clinical impact of these parametric images in the improvement of agreement between different radiologists and cardiologists in terms of segmental interpretation [48,51,59,66].

Although the parametric images are commonly used in clinical practice of nuclear medicine, their integration in cardiac MRI as well as in echocardiography examination is still limited by several factors inherent to the image quality and the lack of a standard protocol for the interpretation of these parametric images. The main limitation is that parametric images can be affected by the cardiac translation due to breathing or body motion. Another limitation lies in the absence of a standard database and protocol from which it is possible to compare the different methods of parametric images computation, in order to extract the most reproducible method in terms of accuracy and sensitivity [71-74]. In addition, some additional training is needed in order to explain the observer how to properly interpret the content in the parametric images, to exclude possible artifacts.

In addition to the methods described above, which are mainly dedicated to the quantification of cardiac contraction abnormalities, other parametric imaging approaches have been developed in cardiac MRI for the characterization of myocardial tissues and particularly for the detection of the extent of interstitial fibrosis [75-79]. These imaging techniques are the relaxation parametric maps T1 and T2. The parametric imaging represents therefore a promising method not only for the regional assessment of wall motion abnormalities but also for the characterization of myocardial tissues.

 **CONCLUSION**

In the current paper, we reviewed the most commonly parametric imaging techniques developed in the literature to evaluate the regional LV wall motion abnormalities. Previous work has revealed the increase in accuracy of visual interpretation and thus the clinical usefulness of this approach in the identification of the location and the extent of wall motion abnormalities. However, the implementation of this approach in clinical practice is still confronted with other challenges inherent to the lack of a standard protocol for the interpretation of parametric images. Further studies on larger cohorts are needed to provide guidelines on how to validate results and to define the most reliable method among the different parametric imaging techniques.

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