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A Review on Analysis of Speckle reduction in Ultra Sound Images

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Abstract*: Ultrasound is an inexpensive and widely used imaging modality for the diagnosis and staging of a number of diseases. In the past two decades, it has benefited from major advances in technology and has become an indispensable imaging modality, due to its flexibility and non-invasive character. Ultrasound imaging utilizes high-frequency sound waves to picture inner structures by contrasting reflection signals created when a light emission is anticipated into the body and ricochets back at interfaces between the structures. However the convenience of ultrasound imaging is corrupted by the nearness of flag dependant noise known as dot (speckle). The speckle pattern depends on the structure of the image tissue and various imaging parameters. Speckle noise show in ultrasound picture influences edges and fine subtle elements which constrain the differentiation determination and make symptomatic more troublesome. There are two main purposes for speckle reduction in medical ultrasound imaging (i) To improve the human interpretation of ultrasound images (ii) De-speckling is the preprocessing step for many ultrasound image processing tasks such as segmentation and registration. Various strategies have been proposed for decrease speckle noise in ultrasound imaging. The objective of this paper is to give an overview of speckle noise reduction in ultrasound imaging.
Keywords*: Speckle noise, Ultrasound image, Different methods, filters analysis

I. INTRODUCTION

Ultrasound imaging assumes pivotal part in therapeutic imaging because of its non-obtrusive nature, ease and capacity of framing ongoing imaging. Medical Ultrasound imaging is done using ultrasonic waves in 3 to 20 MHz range. Ultrasound waves are created from the transducer and go through body tissues and when the wave achieves a protest or surface with various surface or acoustic in nature, it is reflected back. These echoes are received by the apparatus (the transducer array) and converted into electric current. These signals are amplified, conditioned and shown on a display device in real time. The image generated using Ultrasound imaging commonly known as Ultrasound Scanning is called an Ultra sonogram. The resolution of the image will be better by using higher frequencies but at the same time limits the depth of the penetration. There are diverse methods of ultrasound imaging.

The most common modes are (a) B-mode - the basic two-dimensional intensity mode, (b) M-mode to assess moving body parts (e.g. cardiac movements) from the echoed sound and (c) Color mode pseudo coloring based on the detected cell motion. Ultrasound imaging is a generally utilized and safe restorative indicative system, because of its non intrusive nature, minimal effort and capacity of framing ongoing imaging. To accomplish the most ideal determination it is vital that medicinal pictures be sharp, clear and free of commotion and ancient rarities. However event of spot noise is an issue with ultrasound imaging. Spot is the curio caused by impedance of vitality from haphazardly appropriated scrambling. Spot commotion has a tendency to lessen picture determination, complexity and obscure essential points of interest, consequently diminishing indicative estimation of the imaging methodology. In this way spot commotion decrease is a critical essential in ultrasound imaging. Ultrasound image is adaptable, transferable and comparatively safe, but this type of image consists of full of acoustic interferences (speckle noise) and artifacts. Speckle is a complex phenomenon, which degrades delectability of target organ and reduces the contrast. It affects the human ability to identify normal and pathological tissue.

This paper is organized as follows: section II presents the related work. Section III presents different noise models. The methods and Results to filter out the speckle noise are presented in section IV and V and followed by conclusion in section VI.

II. RELATED WORK

The ultrasound-based analytic medicinal imaging method used to picture muscles and numerous inner organs, their size, structure and any neurotic wounds with continuous tomographic pictures. It is likewise used to imagine an embryo amid routine and crisis pre-birth mind. Obstetric ultrasonography is generally utilized amid pregnancy. It is a standout amongst the most generally utilized demonstrative instruments in present day prescription. The innovation is moderately cheap and compact, particularly when contrasted and other imaging methods, for example, attractive reverberation imaging (MRI) and figured tomography (CT). n Speckle noise [1] influences all intelligent imaging frameworks including therapeutic ultrasound. Inside every determination cell various basic diffuses mirror the occurrence wave towards the sensor. The backscattered cognizant waves with various stages experience a productive or a dangerous obstruction in an irregular way. The obtained picture is in this way tainted by an irregular granular example, called spot that postpones the translation of the picture content. Kidney ultrasound imaging is utilized to assess kidney size and position, and help to analyze auxiliary variations from the norm and the nearness of sores and stones. Nonetheless, because of the nearness of spot commotion in these, playing out the division strategies for the kidney pictures were extremely testing and thusly, erasing the muddled foundation will accelerate and builds the exactness of the division procedure [2]. Accordingly, this examination proposes a programmed Region of Interest (ROI) age for kidney ultrasound pictures. Firstly, for speckle noise reduction the techniques such as median filter and Wiener filter are used. The picture handling can be utilized to naturally recognize the centroid of human kidney. For the outcome, middle channel has been picked as speckle commotion diminishment systems as it is quicker and identify kidney centroid better contrasted with wiener channel, wavelet channel and speckle noise anisotropic dissemination (SRAD) channel. This product can distinguish centroid up to 96.43% of precision [3]. A calculation is depicted for cleaning speckle noise in ultrasound therapeutic pictures. Scientific Morphological activities are utilized as a part of [4]. This calculation depends on Morphological Image Cleaning calculation (MIC). The calculation utilizes distinctive systems for remaking the highlights that are lost while expelling the commotion. A morphological activity utilizes the self-assertive organizing components reasonable for the ultrasound pictures which have speckle noise. Restorative picture de-noising has turned into an extremely basic exercise all through the finding. In ultrasound pictures, the noise can control data which is significant for the general professional. A wavelet-based thresholding technique is utilized for noise concealment in ultrasound pictures. Quantitative and subjective examinations of the outcomes acquired by the wavelet-based thresholding strategy accomplished from the other spot commotion decrease strategies show its higher execution for speckle diminishment [5].

III. NOISE MODELS

An inborn normal for ultrasound imaging is the nearness of speckle noise. Speckle noise is an arbitrary and deterministic in a picture. Speckle has negative effect on ultrasound imaging. Radical diminishment interestingly determination might be in charge of the poor powerful determination of ultrasound when contrasted with MRI. If there should be an occurrence of therapeutic written works, spot noise is otherwise called surface. Summed up model of the speckle is spoken to as

$$g(n, m) = f(n, m)\, u(n, m) + \xi(n, m) \qquad (1)$$

where g, f, u and ξ stand for the observed image, original image, multiplicative component and additive component of the speckle noise. Here (*n*, *m*) denotes the axial and lateral indices of the image samples or alternatively, the angular and range indices for B-scan images. When applied to ultrasound images, only the multiplicative component of the noise is to be considered and thus, the model can be considerably simplified by disregarding the additive term, so that the equation (1) becomes

$$g(n, m) = f(n, m) * u(n, m) \qquad (2)$$

Homomorphic de-speckling methods take advantage of the logarithmic transformation, which, when applied it converts the multiplicative noise to an additive one. Denoting the logarithms of g, f and u by gl, fl, and ul, respectively, the measurement model becomes

$$g_1 (n, m) = f_1 (n, m) * u_1(n, m) \qquad (3)$$

At this stage, the problem of de-speckling is reduced to the problem of rejecting an additive noise, and a variety of noise-suppression techniques could be evoked in order to perform de-noising.

A. Speckle Noise in Medical Images

Medical image processing is very important to obtain the correct images, which facilitates the accurate observations for the given application. Medical images are corrupted by different types of noises. Generally medical images are corrupted by speckle noise. It reduces the quality of the images which makes it difficult for the feature extraction, recognition, analysis and quantitative measurement.

B. Noise in Ultrasound Images

Ultrasound imaging framework is broadly utilized demonstrative instrument for present day solution. It is utilized to do the representation of muscles, inward organs of the human body, size and structure and wounds. Obstetric ultra-solography is used during pregnancy. In an ultrasound imaging speckle noise shows its presence while doing the visualization process [6].

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C. Medical Ultrasound Speckle Pattern

Natures of Speckle design relies upon the quantity of scrambles per determination cell or disseminate number thickness. Spatial conveyance and the attributes of the imaging framework can be isolated into three classes:

1) The full grown spot design happens when numerous irregular dispersed scrambling exists inside the determination cell of the imaging framework. Platelets are the case of this class.
2) Tissue disseminates is no haphazardly dispersed with long-extend arrange.
3) When a spatially invariant reasonable structure is available inside the irregular disperse locale like organ surfaces and veins [7].

IV. VARIOUS METHODS

There are numerous speckle diminishment channels accessible, some give better visual understandings while others have great noise decrease or smoothing capacities.

A. Averaging

The speckle can be reduced by using averaging as large as possible. It depends on the number of independent amplitude values that are measured. The scanning under exactly same conditions one obtains identical speckle patterns. There are two amplitude values independent of each other if the transducer is translated by half of its bandwidth between measurements [8]. The main disadvantage of this method is that the blurring of de-noised image and loss of details.

B. Median Filter

Median filter [9] computes the median of all the pixels with in a local window and replaces the centre pixel with this median value. This method is effective in cases when the noise pattern consists to be preserved are edges. The main disadvantage of the median filter is the extra computation time needed to sort the intensity value of the each set.

The adaptive weighted median [10] is a median filter through the introduction of weight coefficients and consequently the smoothing characteristics of the filter according to the local statistics around each pixel of the image. The more emphasis is placed on the central weights, the ability of the weighted median to suppress noise decreases but also the increases signal preservation. This is very useful characteristic because it allows the design of a space-varying filter which combines median-type properties with adjustable smoothing. One way of achieving this is to choose a family of weights which decrease as we move away from the center of the window and the rate of decrease is controlled by the local image content. The weights are adjusted according to local statistics of image by using.

$$w(i,j) = [w(k+1,k+1) - cd\sigma^2/m] \qquad (4)$$

where c is scaling factor, m,σ^2 is the mean and variance inside the $2k+1 \times 2k+1$ window and d is the distance of point (i,j) from centre of window $(k+1,k+1)$. For uniform areas constant c should be selected in such a way that $c\sigma^2/m$ is small so that maximum noise is reduced. In case of boundary, c should be selected such that $c\sigma^2/m$ should be high to preserve image details.

The weights can be further adjusted by using brightness difference with different powers. The disadvantage of this method is that the more computation when image divided into more number of angles.

C. Wiener Filter

The Wiener Filter [11], also known as Least Mean Square filter, is given by the following expression.

$$F(u,v) = \left[\frac{H(u,v)^*}{H(u,v) + \left[\frac{S_n(u,v)}{S_f(u,v)}\right]}\right] G(u,v) \qquad (5)$$

$H(u,v)$ is the degradation function and $H(u,v)^*$ is its conjugate complex. $G(u,v)$ is the degraded image. Functions $S_f(u,v)$ and $S_n(u,v)$ are power spectra of original image and the noise.

D. Frost Filter

Frost filter [12] is a spatial domain adaptive filter that is based on multiplicative noise order it adapts to noise variance within the filter window by applying exponentially weighting factors M as:

$$M_n = \exp(-(DAMP*(S/\mathrm{Im})^2)*T) \qquad (6)$$

The weight factor decreases as the variance within the filter windows reduces. DAMP is a factor that determines the extent of the exponential damping for the image. The larger the damping value, the heavier is the damping effect. Typically the value of DAMP is set to 1. *S* is the standard deviation of the filter window, I_m is the mean value within the window and T is the absolute value of the pixel distance between the center pixels to its surrounding pixels in the filter window. The value of the filtered pixel is replaced with a value calculated from weighted sum of each pixel value P_n and the weights of each pixel M_n in the filter window over the total weighted value of the image as:

$$\mathrm{Im}\, g(i,j) = \frac{\sum P_n * M_n}{\sum M_n} \qquad (7)$$

The parameters in the Frost filter are adjusted according to the local variance in each area. If the variance is low, then the filtering will cause extensive smoothing. While in high variance areas, little smoothing occurs and edges are retained.

E. Anisotropic Diffusion

Isotropic filters are able to smooth image speckle, but also reduce the clarity of features of interest. Anisotropic diffusion [13] attempts to smooth an image while preserving features edges. Applied to ultrasound images, however, anisotropic diffusion enhances the numerous false edges introduced by the speckle pattern. Speckle reducing anisotropic diffusion (SRAD), developed incorporates speckle statistics into the traditional anisotropic diffusion framework. Thus, SRAD is able to smooth homogeneous regions of speckle while preserving feature edges.

SRAD incorporates local speckle statistics into the anisotropic diffusion framework to smooth homogeneous speckle regions while preserving image features. SRAD is based on a nonlinear partial differential equation (PDE). The continuous-domain version of this PDE is

$$\frac{\partial}{\partial t} I(x,y;t) = div[c(q)\nabla I(x,y;t)] \qquad (8)$$

where (x , y) are image coordinates, t is diffusion time, $I(x,y;t)$ is the image intensity function, and $c(q)$ [0,1] is the diffusion coefficient. This PDE is typically discretised by computing finite differences in four directions at each image pixel. The SRAD diffusion coefficient incorporates local speckle statistics and is found via

$$c(q) = [1 + \frac{[q(x,y;t)^2 - q_0{}^2(t)]}{q_0{}^2(t)[1+q_0{}^2(t)]}]^{-1} \qquad (9)$$

where $q(x,y)$ is the instantaneous coefficient of variation (ICOV) and $q_0(t)$ is the speckle scale function.

F. Adaptive Nonlinear Filters

same window size. In contrast to a recursive linear filter, the recursive median filter is inherently stable since the value of the median is per definition equal to one of the input samples. Therefore, the median value is bounded if the input signal is bounded.

With respect to an appropriate window size W, two effects must be taken into account. The noise suppression increases with increasing window size W; conversely, the space resolution decreases with increasing window size. In particular, a median filter removes all objects on a flat background which contain less than (W+1)/2 pixels. To solve this problem, commonly a small window (e.g., 3 X 3) is used, and the median filtering is repeated, until the resulting output is sufficiently smooth or until no further changes occur. Thereby, the performance of smoothing and space resolution is controlled by an appropriate weight factor as a function of the local signal-to-noise ratio is estimated from the noisy image.

G. Generalized Homomorpic and Adaptive order static Filters.

The generalized Homomorphic transformation approximately transforms the signal-dependent noise and additive noise into a normalized additive noise, which has a unit variance. For a monotonic homomorphic transformation, the transformed impulses still have the maximum (minimum) value in the dynamic range: We can now search for a nonlinear filter which can suppress impulsive noise components while preserving sharp edges and can smooth out additive noise sufficiently. Then an inverse transformation g^{-1}

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used to restore the filtered image. As the filter, which is adopted between two transformations, is a nonlinear filter, we call it the generalized homomorphic nonlinear filter [15].

H. Mean Square Error Filter

Local statistic filter based on MMSE was proposed by Lee [16]. The original multiplicative signal model to the sum of signal and adaptive noise and derived the local statistics of original image from the noisy one. This MMSE filter is not demanding the distribution of original image and a better filtering result is obtained with much lower sample calculation compared with the MAP filter. So MMSE filter is widely used in speckle filtering.

The coefficient of variation was chosen as a test of region characteristic by Lopes and the MMSE filter can be enhanced by comparing the test with two predefined thresholds [17]. The enhanced filter outputs average pixel value in a moving window for homogenous area and the central pixel for heterogeneous one. This can smooth speckle more efficiently as well as keep edge and fine details. The coefficient of variation defined by the ratio of local standard variation to the mean in filtering area of noisy image is regarded as a test for homogeneous analysis. If the ratio is lower or equal to speckle noise standard deviation, it means the appearance of homogeneous area. So speckle noise standard deviation is used as lower threshold of coefficient of variation without any doubt. The enhanced MMSE filter based on area segmentation theory introduced above is expressed as follows

(a) (b) (c)

Fig. 1. Input ultrasound images.

(a) (b) (c)

Fig. 2. Output images for Average Filtering

(a) (b) (c)

Fig. 3. Output images for Median Filtering

(a) (b) (c)

Fig. 4. Output images for Wiener Filtering

(a) (b) (c)

Fig. 5. Output images for Frost Filtering

(a) (b) (c)

Fig. 6. Output images for Anisotropic Diffusion Filtering

(a) (b) (c)

Fig. 7. Output images for Adaptive Filtering

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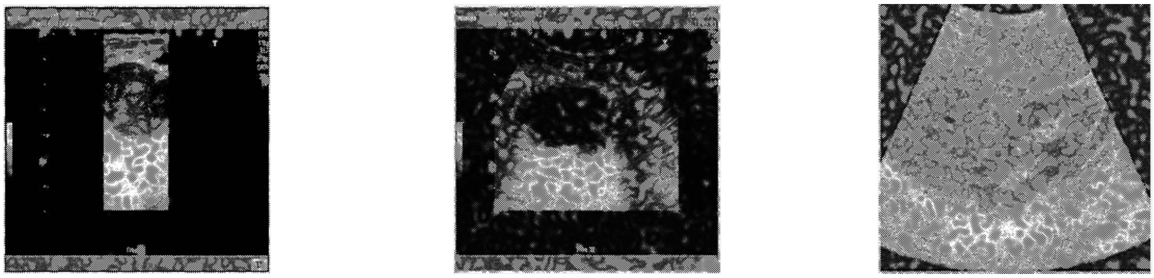
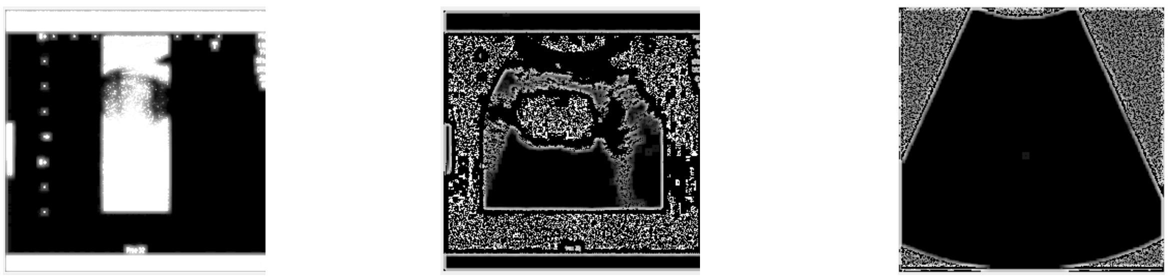


(a) (b) (c)

Fig. 8. Output images for Homomorphic Filtering



(a) (b) (c)

Fig. 9. Output images for MMSE Filtering

where C_y denotes the coefficient of variation in speckle image. C_{min} and C_{max} denote the lower and upper threshold of coefficient of variation. It is noticeable that these filters use fixed window calculating local statistics, but we hope that the window size is changeable according to region characteristic. That is to say, larger window size is chosen in homogeneous area and smaller in heterogeneous [18, 19]. The current window size is increasing gradually if the coefficient of variation calculated in speckle image satisfies the condition of homogeneous introduced above until maximum window predefined reaches or structure features appear. An appropriate window size is decided and the average filtering is chosen to smooth speckle. The threshold should be changed according to window size and speckle.

V. EXPERIMENTAL RESULTS

The following figures exhibits the comparison of all the filters defined above for an input images shown in Fig. 1 (a) – (c). The corresponding output of each filters are shown in 2(a)-(c) through Fig. 8 (a) to (c). The table 1 shows the mean square error (MSE), peak signal to noise ratio (PSNR), correlation and SSIM (Similarity Index between two images which has values between 0 and 1) for all the speckle noise filters.

TABLE I COMPARISON OF ALL FILTERS FOR SPECKLE NOISE

	Ultrasound Image 1				Ultrasound Image 2				Ultrasound Image 3			
Method/ Parameter	MSE	PSNR	Correlation	SSIM	MSE	PSNR	Correlation	SSIM	MSE	PSNR	Correlation	SSIM
Average Filter	17.26	35.5	0.95	0.88	17.26	35.74	0.96	0.88	22.83	34.5	0.97	0.82
Median Filter	21.1	35.0	0.86	0.89	21.41	34.82	0.91	0.87	39.65	32.1	0.95	0.70
Wiener Filter	18.11	35.7	0.97	0.89	19.22	35.28	0.97	0.85	27.09	33.8	0.94	0.75
Frost Filter	16.82	35.8	0.97	0.91	18.95	35.34	0.96	0.87	33.48	32.8	0.94	0.70
Anisotropic	16.8	35.8	0.97	0.91	18.2	35.5	0.97	0.87	21.2	34.8	0.94	0.82

Diffusion	2				2	4			2			
Adaptive Filter	19.54	35.2	0.91	0.83	20.51	35.0	0.85	0.82	25.71	34.0	0.94	0.82
Homomorphic Filter	25.82	34.0	0.70	0.60	36.85	32.44	0.61	0.26	82.41	28.9	0.39	0.03
MMSE Filter	67.97	29.8	0.70	0.70	107.8	27.78	0.06	0.31	226.1	24.5	0.45	0.11

VI. CONCLUSION

Some of the filter techniques used to minimize speckle noise is analysed. Table 1 summarizes the output from all filters used to minimize speckle noise for three input ultrasound images. The MSE, PSNR, correlation of input and output images, SSIM are compared. The limitations of these algorithms are sensitive to the size and shape of the window considered. If the window size is too large, over smoothing will occur, a small window size will decrease the smoothing capability of the filter and will not reduce speckle noise. Some of the de-speckle methods based on window approaches require thresholds to be used in the filtering process, which have to be estimated empirically. Most of the existing de-speckle filters do not enhance the edges; they only inhibit smoothing near the edges. When an edge is contained in the filtering window, the coefficient of variation will be high and smoothing will be inhibited. Thus, while developing an efficient and robust de-noising method for ultrasound images, number of factors to be considered. The choice of de-speckling filter and speckle model plays an important role in the design of speckle noise reduction methods and it differs from one application to other. With above analysis and review of this work, our future work will be development of a robust and efficient algorithm to minimize speckle noise in the edge of the image by maintaining the smoothing in edges.

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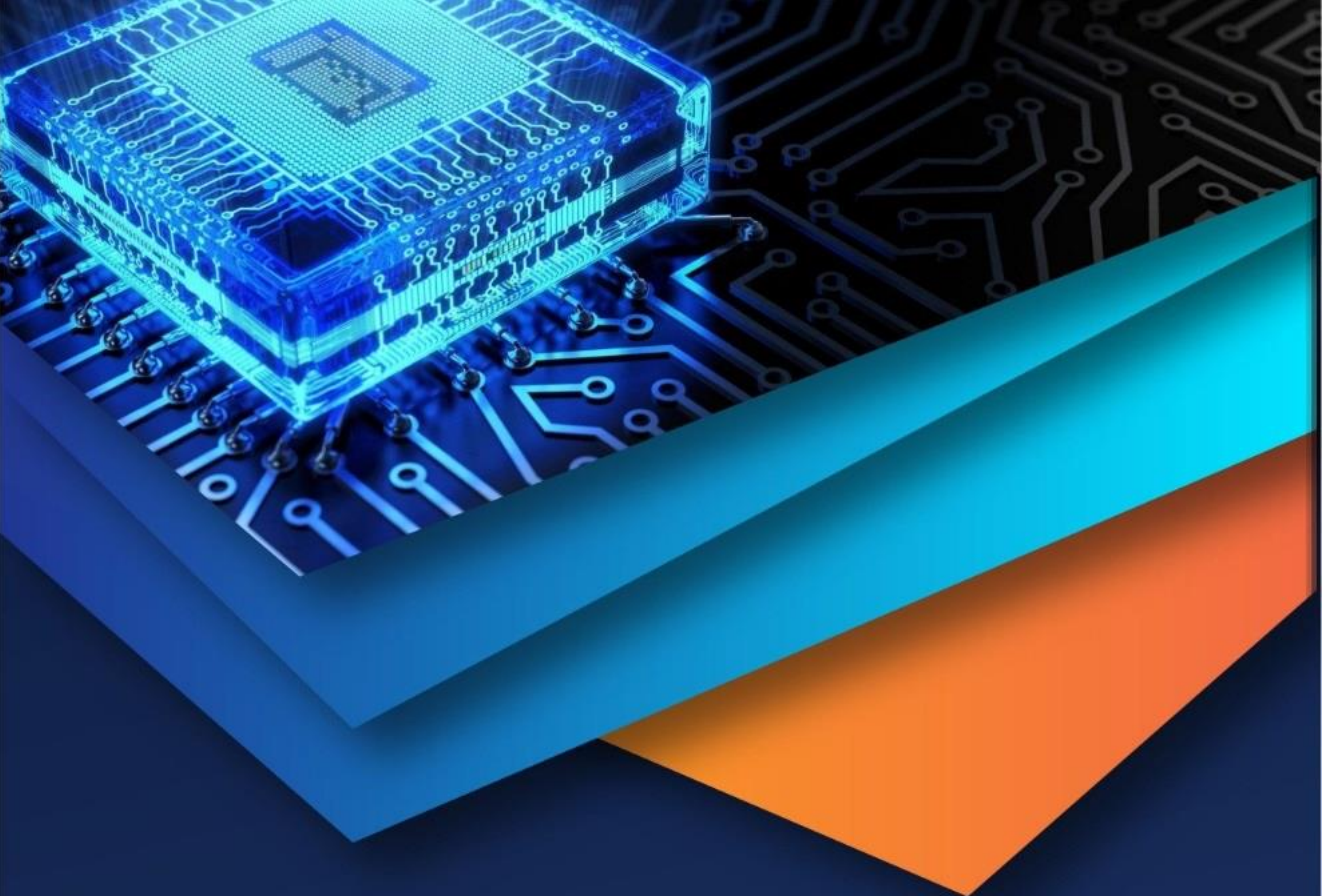

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