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Motion estimation in ultrasound imaging applied to the diagnostic of pelvic floor disorders

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Abstract—**The main purpose of this paper is to show the potential of tissue motion estimation in ultrasound imaging for the diagnostic of pelvic floor disorders. We propose to evaluate the tissue motion using a method based on a local deformable model and on image features (local phase and orientation) extracted from the monogenic signal. The proposed method is well adapted to the pelvic organ deformations and estimates motion with subpixel precision without the need for interpolation. The estimated motion is used to visualize the bladder local deformation and to extract quantitative figures such as the deformation parameters and the bladder angle variation. These results could potentially be interesting to characterize the degree of the pelvic organ prolapse.**

I. Introduction

Pelvic floor disorders affect up to one third of women worldwide, with a peak incidence at age 60 [1]. Although conservative management should be systematically proposed [2-3], surgery remains the most relevant treatment for symptomatic pelvic organ prolapse (POP). POPs, severe enough to warrant surgical correction, have been noted in approximately 11% of the general population women. This figure is anticipated to increase in prevalence as our population ages [4-5]. Unfortunately, 30% of these surgeries are repeat procedures [6], suggesting that a better understanding of the causative mechanisms underlying clinical presentations might improve the surgical approach. Indeed, pelvic organ support defects are mostly assessed by a clinical quantification system, the so called POPQ, created in 1996 and updated in 2010 to S-POP, using the hymeneal plane as reference landmark [7]. While the POPQ appears to be very specific, objective and reproducible, it is interpreted as being difficult to learn and incorporate into daily practice. Furthermore, clinical examination only evaluates POP through the motion of vaginal walls. Therefore, dynamic pelvic MRI has been introduced in the beginning of the nineties [8], providing images at rest and during maximum strain and using the mid-pubic line as reference landmark. Although MRI is useful to assess POP qualitatively and to describe its components in terms of anatomy [9], it suffers from some limitations, mainly: considerable discrepancies between POP quantification using MRI and clinical examination, no real-time assessment, cost and access restrictions [9-10].

Ultrasound (US) imaging is a noninvasive, relatively inexpensive and real-time technique. Its increasing availability in the clinical setting, and the recent development of 3D and 4D US, have renewed interest in using this modality to image pelvic floor anatomy as a key to understanding dysfunction [11]. Some recent studies have shown encouraging results using transperineal (TP) ultrasound [10, 11]. However it has been evaluated as an alternative tool to clinical and MRI examinations in the quantitative and qualitative assessment of pelvic floor disorders in only one study [12].

Thus, the main objective of this paper is to evaluate the possibility of using the tissue motion estimated from US image sequences for the diagnostic and evaluation of the degree of symptomatic pelvic organ prolapsed.

Tissue motion estimation in ultrasound imaging is an active research domain, with various applications like heart motion or elastography. This intensive research is motivated by numerous challenges such as the low SNR of US images, the tissue deformation, the speckle decorrelation or the out of plane motion (when 2D image sequences are exploited) (e.g. [13] and references therein). In order to tackle these limitations several methods were proposed. The standard method in ultrasound motion estimation is the speckle tracking [14]. Based on an assumption of rigid local translations, it estimates the 2D (or recently 3D) translations using cost functions such as cross-correlation or sum of absolute differences [15]. This method has two main limitations. First, even for small blocks, the assumption of rigid translations may not be valid in tissue motion. The complexity of the displacement to be estimated requires the use of deformable methods using parametric motion models, like for example in [16-18]. Second, in almost all the applications, a subpixel motion estimation precision is required. To obtain such a precision, classical methods, such as speckle tracking, need to estimate the motion on interpolated images or signals. This represents a major limitation in terms of processing time and memory requirements. To overcome this limitation and obtain subpixel estimations without the need for interpolation, methods based on the instantaneous phase were recently proposed. Applied to radiofrequency (RF) signals or images, they use the phase of the classical analytical signal in 1D or the phases of multi-dimensional analytical signals [19] in 2-D in order to propose iterative or analytic approaches to estimate subpixel delays [20, 21]. Besides, Felsberg *et al.* presented in [22] a generalization of the analytical signal to two dimensions, called the monogenic signal. It gives access to the 2D local phase, orientation and amplitude. They showed in [23] how these measures can be applied to optical flow estimation. We have shown recently in [24] how the monogenic signal can be used to construct 2D translation estimators, included in the speckle tracking method. In addition to its performances in terms of precision and rapidity (reported to be at least ten times faster than classical methods for the same accuracy), we have shown that this estimator can be applied to all types of ultrasound images, RF, envelope and B-mode.

Our purpose in this paper is to propose a novel method for pre-operative quantitative assessment of POP, based first on 2D TP ultrasound, second on a method of motion estimation based on a deformable bilinear model and finally on the monogenic signal. We have deliberately restricted our approach to the anterior compartment (bladder prolapse) for this feasibility and preliminary study.

II. Ultrasound Acquisition and methods

A. *Ultrasound acquisition*

US imaging was performed using Voluson 8 expert ultrasound system (GE Ultrasound, Zipf, Austria) with a 4-8 MHz curved array 2D transducer. Ultrasound gel was inserted into a latex free probe cover, the probe was then inserted into the cover for hygienic reasons, and more gel was applied to the outside of the upright probe. Ultrasound was performed with the probe positioned against the vulva in the mid-sagittal plane of subjects in dorsal lithotomy position, with a filled bladder. Care was paid to avoid exert undue pressure on the perineum so as to allow full development of POP. To be satisfactory, the mid-sagittal view had to include the pubic symphysis, the urethra and bladder neck, the vagina, cervix, rectum and anal canal. Reference images were flagged at rest and after Valsava maneuver. Dynamic sequences (20 fps) were obtained using the cineloop function, recorded from rest to maximum strain.

B. *Method of motion estimation*

As explained in the introduction, numerous motion estimation methods adapted to ultrasound images have been proposed in the literature. In this paper, we propose a hybrid technique, taking advantage from the method called BDBM (Bilinear Deformable Block Matching) in [18] and the 2D delay estimator based on the monogenic signal phase presented in [24]. Moreover, the 2D dense motion fields estimated between pairs of consecutive images are added on the whole sequence [25].

The method proposed here to evaluate the motion of pelvic organs has several advantages:

- Locally, the motion of one block of pixels is approximated by a bilinear model which takes into account rigid translations, rotations, dilatations and shear deformations. The utility of such a model over a classic rigid translation model was shown in [18].
- In order to estimate the eight parameters of the bilinear model, we proposed in [18] to estimate the translations of small regions of interest. We propose in this paper to estimate these 2D local delays by using the monogenic signal phase. This is particularly interesting in our application for two main reasons: first, subpixel estimations are achieved without the need for interpolation, which allows a very small computational time; second, this estimator can be applied on B-mode images, which are the only ones available from the scanner used for the patients' examination.

The main steps of the proposed motion estimation method are given in Figure 1.

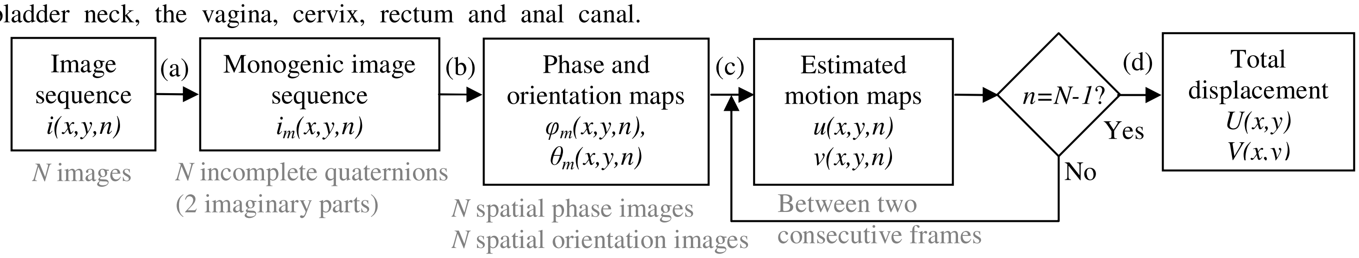


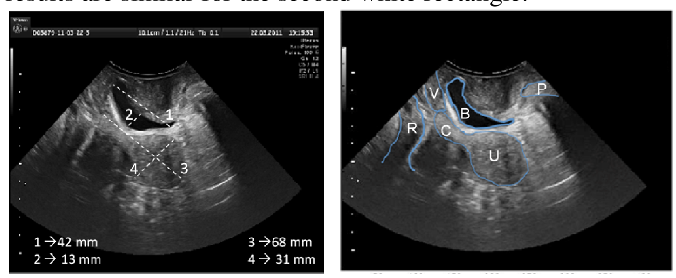
Figure 1. Flow diagram of the proposed motion estimation method. (a) The monogenic signal corresponding to each of the N B-mode images of the sequences is processed [22]. (b) From the monogenic signals i_m, we extract the N local spatial phases and the N local orientation maps, needed for motion estimation [24]. (c) For each couple of consecutive images, the dense motion fields u and v are estimated using the BDBM method and the monogenic 2D delay estimator [18,24]. (d) The N-1 estimated motion maps are transformed (registered) so that they correspond to trajectories of physical features of the first image and added afterwards (U and V) [25].

III. Results

For this feasibility study, the shown images were recorded from a 62 year old woman suffering from a symptomatic POP warranting surgical treatment. Cervix, anterior and posterior vaginal segments were staged according the S-POP [7] and estimated as stages 2, 3 and 1 respectively. The sonographic examination lasted about 15 minutes. The procedure was performed by a gynecologic physician.

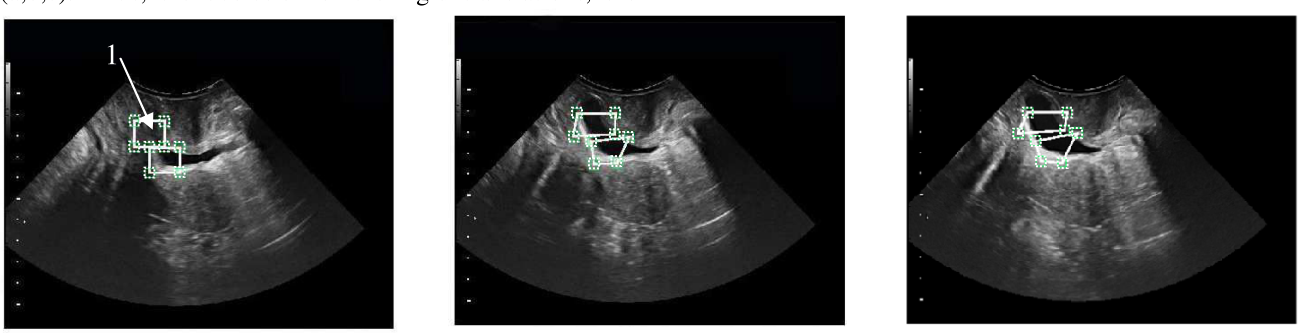
Anatomical mid-sagittal view is shown in figure 2. No abnormalities were found during sonographic exploration. In particular, both ovaries, myometrium and endometrium had a normal appearance, and bladder wall thickness was usual. Pelvic floor is generally divided into 3 compartments: anterior (bladder and urethra), middle (uterus, cervix and vagina) and posterior (rectum and anal canal). As it was mentioned earlier, our protocol only included the anterior one.

The dynamic sequence included 171 frames, of 480×640 pixels. The whole pool was used for motion estimation, using the method described in section II.B. The blocks tracked using the monogenic signal were of size 10×10 pixels (~3×3 mm). No image interpolation was performed. The computation time between two frames was 9s (Matlab implementation on a P8700 2.53GHz). The region of interest we chose was focused on anterior and middle compartments because of the displacement amplitude and direction. As it is described in figure 3, two blocks (white colored) were manually placed by the phisician on the lowest part of the bladder, where the descent was developing. Even if the dense motion field was estimated on the whole sequence and for the whole ROI, we have chosen in this paper to focus only on the displacement and the deformation of these two blocks. Around each corner of the two white blocks, we placed a white dashed block showing the size of the blocks of pixels used for 2D delay estimation by the monogenic signal. The two white blocks could also be placed upon the bladder neck to assess bladder neck and urethral mobility and to differentiate a cystocele from an urethrocystocele, which has a surgical impact. The deformation along the image sequence of the top white rectangle (noted 1) shown in Figure 3(a) is quantitatively described by the estimated parameters of the bilinear transform shown in Figure 5(a,b,c). Thus, the evolution of the rigid translations, the scaling factors and the shear parameters is given. We observe that the rigid translations are equivalent in both directions. However, the deformations (scaling factors and shears) are more important in the lateral direction than in the axial one. The lateral scaling factors shown in Figure 5(b) are positive on the whole sequence, which highlight the widening of the block visible between the images in Figures 3(a) and 3(b). The axial scaling factors are negative (smaller in absolute value than the lateral ones) and highlight an axial thickening of the block, barely visible in Figure 3. The results are similar for the second white rectangle.



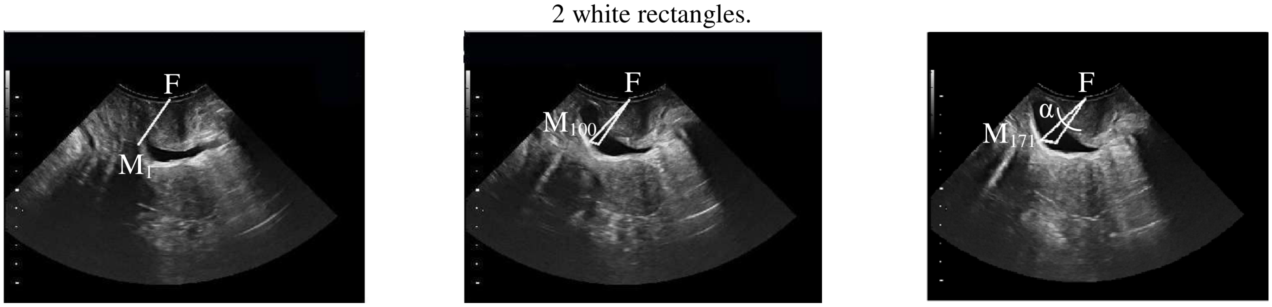
(a) (b)

Figure 2. B-mode mid-sagital view of pelvic organs during valsava maneuver. (a) measurements of bladder (1,2) and uterus (3,4); (b) handmade segmentation of pelvic organs by a gynecologist surgeon: B-bladder, C-cervix, U-uterus; P-pubis; R-rectum; V-vagina.



(a) (b) (c)

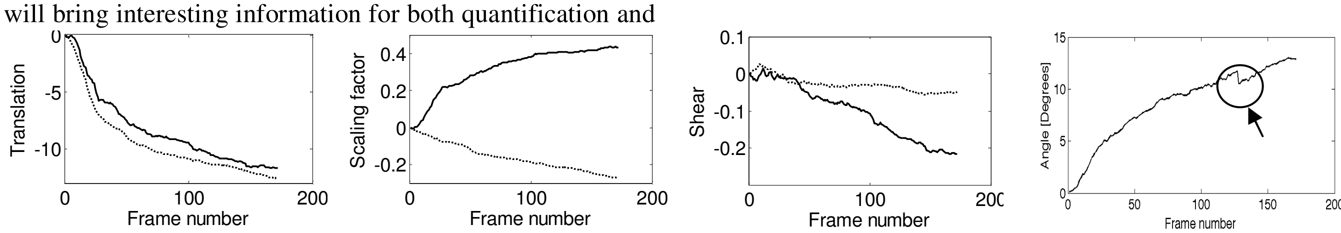
Figure 3. Bladder motion estimation during valsava maneuver, identifying the prolapse (cystocele): (a) rest (frame 1), (b) intermediate (frame 100) and (c) final (frame 171) acquisitions. Bladder displacement is illustrated by the deformation of the 2 white rectangles.



(a) (b) (c)

Figure 4. F represents the chosen fixed landmark, located on the perineum skin. M_n represents the tracked structure, belonging to bladder posterior wall. Representation of angle (α) variation between F and M_n at its initial position, and F and M_n at different positions while prolapse is increasing: (a) rest (α = 0°), (b) intermediate (α = 8°) and final (α = 12.9°) acquisitions.

As shown in Figure 4, our algorithm was upgraded with an angle estimator, allowing us to assess the prolapse quantitatively. From rest position, it calculated the angle generated by the block displacement, according to a fixed landmark. In this example, we have chosen perineum skin as reference (see Figure 4). However, it might be applied to mid-pubic or H lines [8,9], commonly used in pelvic dynamic MRI. We believe that calculating angle variation will bring interesting information for both quantification and qualitative assessment of POP. Figure 5(d) summarizes angle value variations according to each frame of the sequence. We notice, as expected, that the angle value increases while the descent is developing.



(a) (b) (c) (d)

Figure 5. Estimated parameters of the bilinear transform (a-translations, b-scaling factors, c-shears) for the white rectangle annotated 1 in Figure 3(a) (solid lines correspond to the lateral direction and dashed lines to the axial direction), (d) α values according to each frame of the sequence. On Figure (d), the arrow points out a slight discontinuity which might be caused by the patient relaxing the effort.

IV. CONCLUSION

Our protocol seemed to be efficient for pre-operative assessment of POP. Because of the use of a motion estimation algorithm, it could represent a new tool for quantification of pelvic floor disorders. Before that, it has to be compared with clinical examination and dynamic pelvic MRI. Nevertheless, we have the feeling that it could easily perform the metric calculation supported by the POPQ, with the key advantage to propose a quasi-automatic approach and a direct and real-time observation of concerned organs. We also believe that it could be applied to 3D acquisitions in order to explore the volumetric dimension. Further studies are thus required to define its true efficiency.

V. REFERENCES

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