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# Quantitative Evaluation of Female Pelvic Floor Muscle Biomechanics Using Ultrasound Elastography

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**ABSTRACT** Female pelvic floor dysfunction is a common disease which has become one of the five common chronic diseases threatening women's health. As the supporting structures, pelvic floor muscle is critical to support the pelvic organs, maintain continence, and prevent prolapse. The quantitative evaluation of pelvic floor muscle biomechanics is of great value for the diagnosis and treatment of female pelvic floor dysfunction. Our study aims to propose a noninvasive, quantitative, and objective method for motion analysis of the pelvic floor muscles. Ultrasound data, force data during continuous contractive activities from 37 subjects are synchronously collected using an ultrasound scanner and a self-designed intravaginal pressure acquisition device. A two-dimensional motion tracking algorithm is used to monitor the displacement of the pelvic floor muscles. The muscle displacement field is then computed. The parameter, defined as $M_{pu}$, which is the tangential displacements of one interested point near pubis shows good correlation with clinical measurement parameter ($L_{BP}$) ($r = -0.93$, $P < 0.05$). The continuous control ability of the pelvic muscles is further evaluated by measuring the average maintaining time of the maximal contraction force and the muscle thickness during this maintaining period. Both of them are well correlated with the clinical grading results of prolapse. It is concluded that the ultrasound measurements of tissue motions and biomechanics are of great value for the clinical pelvic floor prolapse diagnosis.

**INDEX TERMS** Biomechanics, pelvic floor dysfunction, ultrasound elastography, intravaginal pressure measurement.

## I. INTRODUCTION

Female pelvic floor dysfunction (FPFD) is a common disease in middle-aged and elderly women with pelvic floor organ shifting and dysfunction due to the progressive decline of biomechanical properties of pelvic floor support tissue. The main clinical symptoms including stress urinary incontinence (SUI), pelvic organ prolapse (POP), faecal incontinence (FI) and sexual dysfunction (SD) [1]. It has become one of the five common chronic diseases that threaten women's physical and mental health [2] and is a social health issue that is increasingly valued on a global scale. Among these symptoms, POP is one of the most common reasons for gynaecological surgery in women after parturition. The failure rate is relatively high, and an estimated 30% of women require re-operation [3]. POP is regarded as a multi-factor disease. Multiparity, old age, overweight, chronic straining and obstructive lung diseases are the most important risk factors [1], [8]. It is also find that women with joint hypermobility have a significant higher prevalence of POP [10].

The female pelvic floor is a complex, multi-level and interactional three-dimensional structure, consisting of musculature (pelvic diaphragm muscle and urogenital diaphragm muscle), ligaments and connective tissue [4], [5]. The existing hammock theory [5], regarded these fascia, ligaments, connective tissue and pelvic floor muscles of the vagina

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as three levels of upper, middle and lower to support the pelvic floor. It is pointed out that the pelvic floor muscle is the most important supporting structure among the support levels. Once the supporting structure occurs relaxation or injury, the pelvic organs will not be able to maintain the normal positions, resulting in pelvic floor dysfunction such as POP or SUI [6], [7].

The mechanism among these structures can be modeled as a biomechanical system. However, there is rare of studies to investigate the inherent biomechanics of pelvic floor.

The existing clinical assessments such as palpation and vaginal pressure measurements [24] are susceptible to interference from the muscle groups such as abdominal muscles, gluteal muscles and relies heavily on clinical experts. Electromyography (EMG) test, indirectly evaluate the pelvic floor muscle strength through the neural electrical signals, failed to directly and quantitatively give the parameters that can characterize the biomechanical properties of pelvic floor muscles. Conventional medical imaging technologies, such as magnetic resonance imaging (MRI) and ultrasound used in PFD were all based on the measurement of specific deformation variables of pelvic floor muscles under relaxation, contraction or Valsalva action. Thyer et al. [8] demonstrated that the area changes of levator hiatal (LH) and that the corresponding strain during contraction measured using 4D translabial ultrasound is comparable with muscle strength assessed clinically. Goh et al. [9] pointed out that the elasticity of anterior vaginal wall tissue correlates with age and inferior elasticity within postmenopausal women.

It is worthy to clarify that the evaluation methods widely used at present in the pelvic floor ultrasound is proposed by Dietz et al. [11], [16], which regards the horizontal line of the lower edge of the pubic symphysis as a reference $L_{BP}$. When $L_{BP} > 10$mm, the lowermost edge of the posterior wall of the bladder is located above the reference line greater than 10 mm, which is characterized as 0 degree of POP, indicating as normal status; similarly when $0 < L_{BP} \leq 10$mm, it is characterized as I degree, and $-20\text{mm} < L_{BP} \leq 0$mm as II degree, and $L_{BP} \leq -20$mm as III degree, respectively. The evaluation method has been verified being well correlated to the POP-Q method by numerous clinical research. Please be noted that the POP-Q grading method [19] is a qualitative staging method for the diagnosis of female POP approved by the International Association of Urinary Control in 1996.

The above-mentioned methods evaluate the structural or functional change of pelvic floor muscles, but fail to take both into consideration at the same time. Elastography is considered as the best choice for the direct evaluation of inherent biomechanical properties of pelvic floor muscles, especially the acoustic-radiation-force-based techniques, which can realize an objective and quantitative assessment. However, the measurement of shear modulus of muscles will be severely affected by the angle between the propagation direction of shear wave produced by the acoustic radiation force and the muscle fiber orientation [26], and most currently available ultrasound scanners in clinic cannot be driven into acoustic-radiation-force excitation mode because of the special requirements of system hardware.

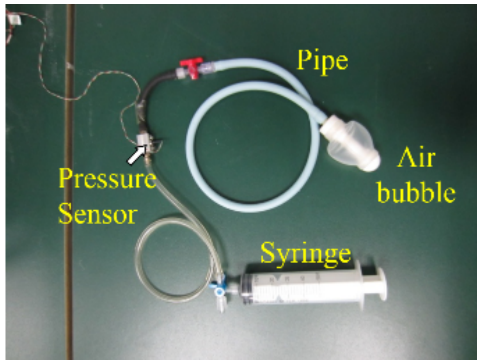


**FIGURE 1.** Picture of the intravaginal pressure acquisition device, consists of an air bubble, a pipe, a pressure sensor, and a syringe.

In this paper, we proposed a novel method, which synchronously captured ultrasound data of women puborectalis and vaginal pressure data during active contraction to evaluate the biomechanics of pelvic floor muscles. The basic principle is to measure both the displacement tracking from ultrasound image sequences and muscle force simultaneously, and then derive the biomechanical parameters from the force-displacement curves, to describe and grade the severity of POP. This method can be easily implemented with clinical ultrasound scanners and will also provide a quantitative measurement.

## II. MATERIALS AND METHODS

### A. DATA ACQUISITION APPARATUS AND EXPERIMENTAL PROTOCOL

The experimental scheme was designed to simultaneously record the muscle force and ultrasound images during the entire contraction and relaxation process when the subject was asked to follow the instructions to actively control her levator ani muscles. The levator ani muscles produce intravaginal pressure when it contracts, however, it's challenging to directly measure the muscle force. Instead, we measured the intravaginal pressure using a self-designed apparatus, which consists of an air bubble, a pipe, a pressure sensor and a syringe (Figure 1). During measurement, the air bubble was placed into the vagina, and a certain amount of water (about 50 mL) was injected into the air bubble to ensure the bubble wall completely fits the inner wall of the vagina, therefore, the pressure sensor could sense a more accurate squeezing force when pelvic floor muscle contracts. This device can also obtain real-time changes of vaginal muscle strength when the subject completes a series of actions.

A translabial ultrasound probe with frequency at 6-12 MHz was put on the perineum of the subject to fully show the puborectalis muscle together with pubis and rectum during the contraction and relaxation. The subject was asked to lay on her back, and follow the standard instruction to make her every effort to slowly contract her levator ani muscles,

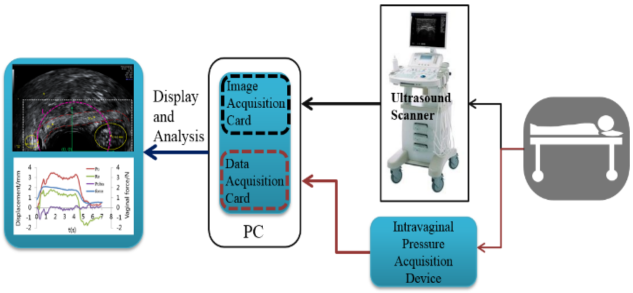


**FIGURE 2.** System diagram to simultaneously collect the intravaginal pressure and ultrasound images of levator ani muscles during active contraction and relaxation by the subject.

**TABLE 1.** Number of subjects grouped with different degree of pop evaluated with Dietz's method.

| | Degree of POP | | | |
|---|---|---|---|---|
| | 0 | I | II | III |
| Nulliparous | 7 | 1 | 0 | 0 |
| Vaginal | 0 | 4 | 11 | 5 |
| Caesarean | 3 | 5 | 1 | 0 |

hold the maximal force period for at least 5 seconds, and then relax. The ultrasound data collected by an ultrasound scanner (GE Voluson E8 Expert, USA) was recorded by an image acquisition card (NIPCI-1411, National Instruments Corp. Austin, TX, USA) and the intravaginal pressure was recorded by a data acquisition card (NIPCI-6024E, National Instruments Corp. Austin, TX, USA). A VC++ program was developed to simultaneously record and display both ultrasound image and pressure data, and then for further analysis (Figure 2).

A total of 37 women, aged at 27.05 ± 3.71 years old, weighted at 58.43 ± 5.37 kg, and with height at 161.21 ± 5.20 cm were recruited from Department of Obstetrics, the First Affiliated Hospital of Shenzhen University for this study. They were divided into three groups: normal nulliparous group, vaginal delivery primiparous group and those got caesarean section operation (Table 1). Subjects were all trained to learn to perform active contractions using only pelvic floor muscles before data acquisition, and were graded into 4 groups of POP degrees, with the Dietz's method by the sonographer. The puborectalis muscle contains a large number of type I slow-twitching muscle fibers that can maintain tension contraction continuously, and its biomechanical properties are critical for maintaining the pelvic floor organ in a normal position to avoid prolapse. In this study, we tried to evaluate such biomechanical properties, therefore, the subject was required to control the muscles slowly contracting to her maximal extent and then to maintain for a short period, then slowly relax. The entire process generally does not exceed 7 seconds. In order to avoid the interference caused by muscle fatigue, the subjects took 15-20 minutes to rest after completing the contraction mode. All the data were stored for later offline process. The study was approved by the local institutional review board and all subjects gave written informed consent.

### B. ALGORITHM OF MOTION TRACKING

One of the important procedure in our study is to precisely track the deformation of the pelvic floor muscle during the active contraction controlled by the subjects herself from the translabial ultrasound images. With both deformation and force, we can estimate the biomechanical properties of the muscles.

There are many algorithms for motion tracking of soft tissues from ultrasound images [12]– [14]. Considering the accuracy, reliability and time-consumption of these algorithms, we adopt optical flow algorithm and improve it to fulfill our requirement. At present, many optical flow algorithms are derived from the mathematical methods proposed by Horn and Shunk [15]. That is, if the gray level of the image is unchanged during motion, the following gray equations are available:

$$I\,(x, y, t) = I(x + u, y + v, t + 1) \tag{1}$$

where I is gray value, the gray value of a point $(\boldsymbol{x}, \boldsymbol{y})$ in an image at time $t$ is thought to be unchanged in the next image at the time $t + 1$. $u$, $v$ are the displacements of the point in the $x$ and $y$ directions, respectively. This processing principle of optical flow algorithm is similar to infinite differential, which is more common for ultrasound signal processing. From equation (1), the following optical flow constraint equation can be obtained:

$$I_x u + I_y v + I_t = 0 \tag{2}$$

Equation (2) is the classical Horn & Shunk optical flow formula, where $I_x$, $I_y$ and $I_t$ are spatiotemporal gradients of image brightness, which can be estimated from adjacent pixels on the image. But solving $u$ and $v$ is not enough according to the only few parameters, additional constraints are added to the developed different optical flow algorithms.

Here we use a coarse-to-fine strategy proposed by Tomas et al. [17]. This method hypothesizes that the gray space gradient is constant, as described in equation (3).

$$\nabla I\,(x, y, t) = \nabla I\,(x + u, y + v, t + 1) \tag{3}$$

where $\nabla = \left(\partial_{x,} \partial_y\right)^{\mathrm{T}}$. A smoothing hypothesis is added to accelerate the convergence of the algorithm. Depending on these assumptions, $u$ and $v$ in the axes of $x$ and $y$ by establishing the following energy formulas:

$$\begin{aligned} E_{data}\,(u, v) &= \textstyle\int_\Omega \psi \left(|I\left(x + w - I\,(x)\,/\hat{}2\right)\right) \\ &\quad + \Upsilon|\nabla I\left(x + w\right) - \nabla I\,(x)\,|\hat{}2\big)\,dx \qquad (4) \\ E_{smooth}\,(u, v) &= \textstyle\int_\Omega \psi\,(|\nabla_3\,u|\hat{}2 + |\nabla_3 v|\hat{}2\,)\,dx \qquad (5) \\ E\,(u, v) &= E_{data} + \alpha E_{smooth} \qquad (6) \end{aligned}$$

Equation (4), $E_{data}$ measures the global difference between the constant hypothesis of the gray value and the gradient gray space. $\Psi$ is a non-squared penalty function, and $\gamma$ is the weight of gray conservation and gradient conservation weight. Equation (5) is a model assumption for smoothing the optical flow field, $\nabla_3 = \left(\partial_x, \partial_y\right)^T$. Equation (6) can be obtained from equations (4) and (5), and the total energy $E(u, v)$ is the sum of the $E_{data}$ and weighted $E_{smooth}$ ($\alpha$ is the weighting factor of the smoothing term, $\alpha > 0$). The optical flow vector can be solved by finding $u$ and $v$ that minimize the $E(u, v)$ value. According to the variation principle, the minimum (1) formula, the sampling multi-resolution hierarchical calculation and the super-relaxation iteration can be used to obtain the following iterative equations to solve the optical flow vector:

$$\begin{aligned}
&\psi'\left(\left(I_{z,k+1}\right)^\wedge 2+\gamma\left(\left(I_{xz,k+1}\right)^\wedge 2+\left(I_{yz,k+1}\right)^\wedge 2\right)\right)\\
&\quad\cdot\left(I_{x,k}I_{z,k+1}+\gamma\left(I_{xx,k}I_{xz,k+1}+I_{xy,k}I_{yz,k+1}\right)\right)-\alpha\\
&\quad\cdot\mathrm{div}\left(\psi'\left(|\nabla_3\ u_{k+1}|^\wedge 2+|\nabla_3\ v_{k+1}|^\wedge 2\right)\cdot\nabla_3\ u_{k+1}\right)=0\\
&\psi'\left(\left(I_{z,k+1}\right)^2+\gamma\left(\left(I_{xz,k+1}\right)^2+\left(I_{yz,k+1}\right)^\wedge 2\right)\right)\\
&\quad\cdot\left(I_{y,k}\ I_{z,k+1}+\gamma\left(I_{yy,k}I_{yz,k+1}+I_{xy,k}\ I_{xz,k+1}\right)\right)-\alpha\\
&\quad\cdot\mathrm{div}\left(\psi'\left(|\nabla_3\ u_{k+1}|^\wedge 2+|\nabla_3\ v_{k+1}|^\wedge 2\right)\cdot\nabla_3\ v_{k+1}\right)=0
\end{aligned} \tag{7}$$

Here $k$ is the number of iteration layers, and $w^0 = (0, 0)$ as the initial value. With repeated trial and verification experiments, it is found that, with $\alpha = 30$, $\gamma = 80$, the obtained deformation map can meet our requirements for the following data processing.

### C. DATA PROCESSING AND STATISTICAL ANALYSIS

Steps of processing the ultrasound image sequences to extract the muscle deformation during active contraction and relaxation are as follow: (1) segment the region of interest, here we focused on the puborectalis muscle together with pubis and rectum; (2) using the improved optical flow algorithm to obtain the displacement fields $u$ and $v$, where $u$ and $v$ are lateral and longitudinal displacement of puborectalis, respectively; (3) coordinate transformation from $u$ and $v$ to $r$ and $q$ where $r$ and $q$ are radial and tangential displacement of puborectalis, indicating the deformation direction forward or backward to the pubis; (4) select feature points $Pu$ which is close to the pubic symphysis, at 1/4 of total length of puborectalis and $Re$ which is near the rectum, at another 1/4 of total length of puborectalis, as two indications, to describe the deformation pattern of puborectalis muscle; (5) calculate the thickness of the puborectalis during all the active process.

Since the ultrasound data and the muscle force data are acquired synchronously, it's applicable for us to observe the change of muscle thickness $M_d$, muscle deformation at both directions ($M_{pu}$ and $M_{re}$ for feature points $Pu$ and $Re$, respectively) during contraction, retaining and relaxation period.

Each subject was required to repeat the contraction-retaining the maximal contraction for around 5 seconds-relaxation process for 3 times, and then the force/deformation data were averaged for analysis. All the parameters extracted from these process were correlated with the clinical parameter $\boldsymbol{L_{BP}}$ to investigate if our new biomechanical parameters can grade the degree of POP.

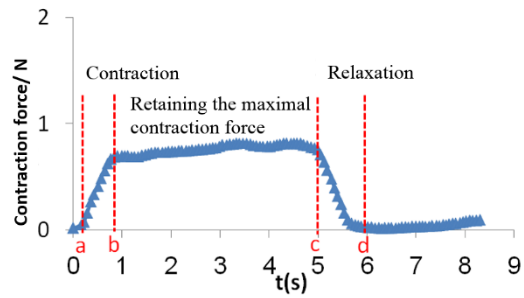


**FIGURE 3. A typical curve of slow-contraction including the contraction period (from point a to b, retaining period (from point b to c) and relaxing period (from point c to d).**

**TABLE 2. Displacements of puborectalis during slow-contraction and values of $L_{BP}$ for each group of pop.**

| Grades of POP | $L_{BP}$/mm | Tangential Disp/mm | | Radial Disp/mm | |
|---|---|---|---|---|---|
| | | $M_{Pu}$ | $M_{Re}$ | $M_{Pu}$ | $M_{Re}$ |
| 0 | 18.93± 0.98 | -4.86±0.31 | 5.07±0.76 | 2.21±0.35 | 0.89±0.87 |
| I | 4.44±1.34 | -3.60±0.34 | 6.51±1.09 | 1.99±0.74 | 1.84±0.62 |
| II | -11.55±1.65 | -2.27±0.31 | 7.20±0.74 | 2.20±0.73 | 1.81±0.39 |
| III | -23.14±1.02 | -1.05±0.21 | 8.12±2.23 | 3.23±0.66 | 0.44±0.77 |

## III. RESULTS

Figure 3 shows a typical muscle force curve, and was regarded as the contraction period (from point a to b), retaining period of maximal contraction force (from point b to c) and relaxing period (from point c to d). The averaged muscle displacement $M_{pu}$ and $M_{re}$, were calculated during the period between (a + b)/2 and (c + d)/2. Table 2 describes the displacements of puborectalis during slow-contraction and values of $L_{BP}$. It is found that the tangential displacement of feature point is generally greater than the radial displacement, and for the tangential displacement, the displacement of $Re$ is greater than that of $Pu$. For the correlation analysis, we found that both the radial displacement of feature points $Pu$ and $Re$, have no significant correlation with $L_{BP}$, while the tangential displacement, especially that of $M_{pu}$, is significantly correlated with $L_{BP}$ ($r = -0.93$, $P < 0.01$), indicating that smaller $L_{BP}$, larger tangential $M_{pu}$, and then severer POP.

To assess the correlation of muscle thickness and clinical POP grade, we calculated the muscle thickness curve for each subject, and found that the thickness change of muscle completely reflects the whole stage of muscle contraction, retention and relaxation, and is synchronized with the change of muscle contraction force.

As described previously, the main function of type I slow-twitching muscle fibers is to maintain the pelvic floor organ in a normal position to avoid prolapse. Therefore, the average time of puborectalis retaining the maximal contraction force ($t_0$) and the muscle thickness in this retaining period ($M_d$) were regarded as important biomechanical parameters

**TABLE 3. The average time of retaining the maximal contraction force and mean muscle thickness during the retaining period in each group.**

| Groups | 0 | I | II | III |
|---|---|---|---|---|
| $t_0$/s | 4.37±0.32 | 3.92±0.54 | 3.56±0.70 | 3.16±0.32 |
| $M_d$/mm | 13.32±0.24 | 11.86±0.18 | 9.88±0.24 | 5.34±0.38 |

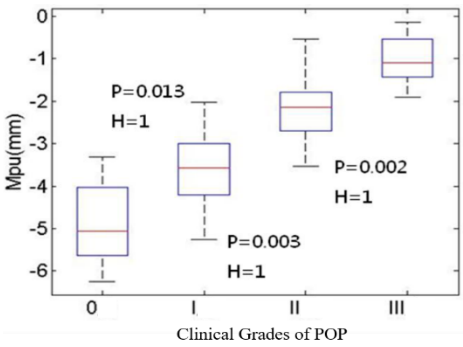


**FIGURE 4. Box plot of the tangential displacement $M_{pu}$ during the retaining period for each group of POP.**

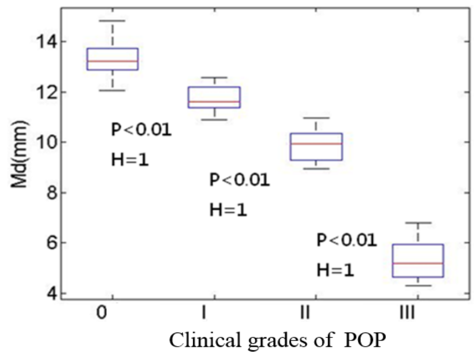


**FIGURE 5. Box plot of muscle mean thickness $M_d$ during the retaining period for each group of POP.**

to reflect this function. Table 3 describes $t_0$ and $M_d$ for each group of POP.

It is can be seen from Table 3 that as the degree of POP increases, the average retaining time of maximal contraction force decreases, that is, the ability of muscle to continuously control the maximal force decreases, and the corresponding average value of muscle thickness decreases. In other words, subjects without POP who have the best overall control, and the maximum thickness of the muscles in the holding area is 13.32 ± 0.24 mm. However, those who had severely POP with weaker ability to maintain their control, the average thickness of and the minimum muscle thickness was only 5.34 ± 0.38 mm. Figure 4 shows the box plot of tangential displacement $M_{pu}$ during the retaining period, and Figure 5 shows the muscle mean thickness $M_d$ during the retaining period. Both of $M_{pu}$ and $M_d$ have significant correlation with clinical grades of POP. That means, $M_{pu}$ and $M_d$ are good indicators to differentiate the degree of POP.

## IV. DISCUSSION

In this study, we aimed to develop a new method to evaluate the inherent biomechanical properties of pelvic floor muscles so that we can grade the degree of POP with an accessible and reliable technique. We described a system which employed a self-designed intravaginal pressure acquisition device and a routine-used ultrasound scanner, to simultaneously record both the contraction force and displacement of levator ani muscles during the active contraction controlled by the subject. The results from clinical trials demonstrated that our system can provide reliable evaluation of different degree of POP subjects.

Some evidences to support the validity of our experiment platform to quantify POP are as follows. First, the design of synchronization optimized the data acquisition procedure, and allow us to monitor the contraction period of the subject in an intuitive and objective way. Second, average tangential displacement of feature point *Pu* and *Re* were all have a significant negative correlation with clinical POP evaluation parameter $L_{BP}$. Third, the mean thickness of muscle during the retaining period of the maximal contraction force can accurately reflect muscle mechanical properties, and $M_d$ distinguished the different degree of POP even better than $M_{Pu}$.

Our results are in agreement with the findings of previous work regarding to the investigation of differences in deformation of the *LH* between different stages of prolapse, subjects with significant prolapse (POP stage ≥ II) demonstrated a lower strain, in other words a less compliant *LH* at the level of the pubourethralis subdivision, than women with mild or no prolapse ($P = 0.03$). Derpapas et al. [17] measured the variation in dimension of leavtor ani muscle (LAM) during Valsalva manoeuvre, using translabial 3D/4D ultrasound in women with PFD, and found that muscle thickness also correlated well with clinical parameter $L_{BP}$. Such finding also corroborates numerous studies reporting other muscles, such as biceps brachii [19], [20], brachialis [21], and abductor digiti minimi [22], together with another study providing substantial evidence of a linear relationship between muscle stiffness, muscle activation and force [18].

We also found that the tangential displacements of feature points were generally greater than the radial displacements, and in the tangential direction, the displacements of *Re* were greater than those of *Pu*. This may be explained by the complex structure of pelvic floor muscles with multi muscle components, and the biomechanical anisotropy due to muscle fiber direction [23]. On the other hand, the slow-contraction strength conveyed to the puborectalis portion of levator ani muscle may be uneven at its different anatomical subdivision [17]. Thickness of muscle reflected well muscle biomechanics *in vivo* [19]– [22]. With increase severity of POP, the mean maintenance time of muscle maximal force decreased, and the muscle thickness also decreased. This results show the fact that for subjects with POP, the ability to continuously control the muscles gets worse.

## V. CONCLUSION AND FUTURE WORK

Our study showed the potential value of ultrasound elasticity imaging together with simultaneous contraction force measurement as a non-invasive, object and quantitative method to assess the biomechanics of female pelvic floor muscle. In comparison with those acoustic-radiation-force based techniques, our technique is more friendly for routine clinical use, which means it is applicable for all the current ultrasound scanners.

On the other hand, shear wave elastography based on acoustic radiation force has been studied on various tissues of human, which can provide shear modulus that directly reflects the biomechanical properties of muscle itself [25], [26]. It's better for us to compare our results with those obtained from shear wave elastography, to investigate if the active contraction properties of the pelvic floor muscles are correlated with or different from the passive excited biomechanics. In addition, deep learning methods, such as convolutional neural networks (CNN), can effectively mine the characteristics and laws of data, which are widely used in the research of segmentation and classification. One of our future works is attempting to archive the deep neural network for the grading tasks of POP.

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