FE MODEL FOR BIOMECHANICAL ANALYSIS AND PREDICTION OF OLDER WOMEN BREAST DEFORMATION

[1]ZHANG JUN, [2]SUN YUE, [3]NEWMAN LAU, [4]JOANNE YIP, [5]WINNIE YU

[1,3]School of design, The Hong Kong Polytechnic University, Hong Kong
[2,4,5]Institute of Textiles & Clothing, The Hong Kong Polytechnic University, Hong Kong
E-mail: [1]alice.zj.zhang@connect.polyu.hk, [2]newman.lau@polyu.edu.hk, [3]yue16s.sun@connect.polyu.hk,
[4]joanne.yip@polyu.edu.hk, [5]winnie.yu.itc@connect.polyu.hk

Abstract - The purpose of this study is to quantify the breast deformation and further to construct a kind of convenient, noninvasive and reliable finite element model (FEM) for the biomechanical analysis of the elderly breast. This method realizes the idea in using non-linear multi-component FEM to predict breast movement and provide hint of deformation in order to predict potential breast pains or injuries resulting from yoga exercise, for instance, Warrior II. In this paper, it initially was used the motion capture system to quantify the selected markers motions of breasts. A comprehensive musculoskeletal FEM is presented to simulate the breast (C cup size) deformations of an elderly woman through a combination of contraction and extension from the pectoralis major muscle (PMM) during arm abduction from 30° to 90°. From the results collected by the motion capture system, it was found that the selected distance in the upper outer quarter has almost four times change ratio larger than that in the upper inner quarter. By validating with motion capture data, this study highlights the customized material coefficients of PMM, in which Young's modulus of clavicular and sternocostal portion was ascertained as 0.1 and 0.08 MPa, respectively. In comparison to conventional FEMs for breast, the realistic one presented here features the musculoskeletal system as well as reveals the synergistic relationship between the breast and PMM, thereby providing a promising basis for the design of the sport bras in an ergonomic way.

Index Terms - Breast Deformation, Older Women, Finite Element Model, Motion Capture.

I. INTRODUCTION

Aging process has a profound influence on the female breast, including the proportion of glandular tissues and fat tissues, the large skin laxity as well as stretched Cooper's ligaments caused an inferior lateral migration of the breast tissues, those factors lead to breast sagging because of insufficient internal support to keep the breast in place [1]. The previous researches on human biomechanical analysis, kinetic and kinematic quantification has received more and more interests, kinematics researches typically are the relative measurement between the breast and human torso. Most of the researches are widely used the motion capture system alone to track the makers' movement placed on the breast the body during the participants were running or walking under different breast conditions including sports bra, daily bra and bare [2-4]. It can be found that during the gait cycle, the breast 3D trajectory followed a figure-of-eight pattern with four movement phases [5]. However, the displacement represented was the absolute displacement of breast surface, not exclude the displacement of trunk as well as not enough description. Besides, they placed only fewer markers on the breast to analyze the data which is not representative to analyze the whole breast deformation because the breast is not a homogenous material. It is well known that the breast has fat, glandular tissues, which has different components inside the breast leading to complex physical behavior during activities [3], [7], [8].

Related studies in the field of the virtual model usually provide a detailed structural organization of individual breasts, a wide range of imaging techniques including ultrasound, tomosynthesis, magnetic resonance imaging (MRI) and X-ray mammography are intensively investigated [6], [8]. However, the obtained model was under compressive state and its expenditure is far too expensive and not convenient for generalization. Extensive research has been conducted to develop biomechanical models of the breast using the FE method [9]. The initial continuum mechanics model was used for the simulation of breast deformation exerted by gravity loading, change of breast shape with the patient position [10] or compression deformation for mammography [11].

Although considerable progress has been made on the FE modeling of breast deformation in recent years, several concerns need to be addressed. Firstly, the majority of studies were mainly focused on young females. In fact, the breast structure and corresponding mechanical properties for each breast tissue vary significantly between the young and the elder. Secondly, walking and jogging were the major research focus while other exercises like Yoga and Taiji, representing a slower or more static movement with higher body extension becomes more and more popular but lacks sufficient studies. Additionally, the biomechanical behavior of large breast deformations under such exercise remains largely unknown. Thirdly, up to date, the most traditional and conventional approach is to directly establish FEMs

on breasts for the investigation of their deformation. However, we believe that the breast deformation is not solely depending on the breast tissues but highly associated with the whole musculoskeletal mechanics. Thus, the contribution of related muscles or tissues should be taken into consideration for a comprehensive and systematic analysis. Research in biomechanical behavior of the breast and breast kinematics are mainly focusing on younger participants aged from 20 to 50 years old, and running is not a good exercise choice for the elderly. Yoga and Taiji are preferred exercises for elderly which can be mainly focused.

II. MATERIALS AND METHODS

A. Experimental Procedure

Healthy older participants took part into the experiments. Those six participants' age was from 60 to 66 years old (61.14 ± 3.57) with the height of 155.29 ± 5.7cm, the weight of 58.01 ± 5.92 kg, underbust girth of 82 ± 6.99 cm and with C cup size. Those participants had breastfed history but no breast surgery history. Informed consents were obtained from the participant with regards to the whole experimental process involving 3D scanning and motion capturing of nude breasts. All experiments were performed in compliance with the guidelines of the Human Subjects Ethics Sub-Committee and approved by the ethics committee at The Hong Kong Polytechnic University.

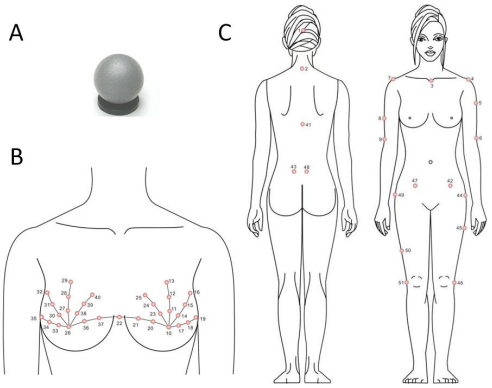


Figure 1: (A) The retro-reflective makers have been used in the motion capture process. The diagram of retro-reflective markers selected locations on the breast (B) and on women body (C).

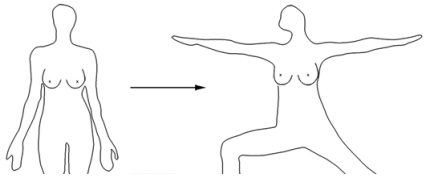


Figure 2: The yoga pose "Warrior 2" was selected for participants to do.

Motion capture is a process of collecting and recording the movements of objects. In the motion capture system, movements are captured many times per second. The motion capture (Eagle, Motion Analysis Corporation, US) uses 12 high-resolution digital cameras to track and record the trajectory of the retro-reflective markers with 9.5 mm in diameter and 1.81 g in mass (as shown in Figure 1 (A)). Before to capture the motions, the retro-reflective markers were attached to the specific position of the body. In the present study, totally 53 markers were attached on each side breast and the human body as shown in Figure 1 (B) and (C). Following, the participants were asked to perform the selected yoga posture Warrior II, which step apart the feet and at the same time arm did abduction from around 30^{O} to 90^{O} as shown in Figure 2. During the whole process, the participants were under the bare condition.

B. Anatomical analysis of women breast for building the FE model

The breasts overlay imperative chest wall muscles including the PMM, the pectoralis minor muscle, the intercostal and serratus magnus as shown in Figure 3. Primary actions of PMM include internal rotation, flexion, and adduction. Muscles cannot work independently and the passive deformation of PMM can be driven by the humerus adduction [12]. In principle, the breast motion is involved in a synergistic and musculoskeletal system and it is significantly influenced by the PMM due to the anatomical features. Thus, it is of great benefit for the establishment of PMM-based FEM to achieve better prediction of breast motion upon large movement. In order to improve the simulating efficiency and accuracy, the model should be simplified under the assumption that the breast lies on the irregularly shaped PMM without considering joints and muscle fascia. The 3D geometric model of the musculoskeletal system was acquired from TurboSquid (TurboSquid, USA). Thereof, the humerus, clavicle, the first to the sixth rib, sternum, and PMM were used in this study.

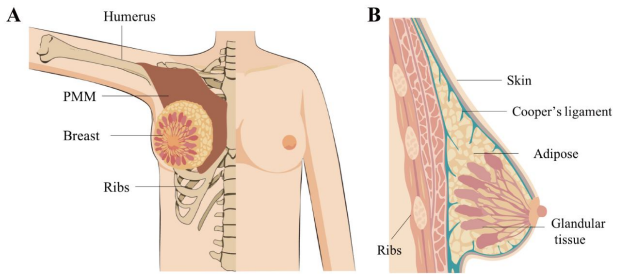


Figure 3: A simplified anatomical diagram of women torso (A) and sagittal section of breast (B).

The FE model was constituted by the comprehensive musculoskeletal system comprising a rigid skeletal supporting part (ribs, humerus, and clavicle), deformable breasts, skin, and specific 3D scanning

body geometric model. Herein, as shown in Figure 4 one marker was placed on the middle position between the left and right breast was marked as O. In the direction of the centerline in inner quadrant r at 9 cm was marked as point A, in the direction of the centerline in the upper outer quadrant at 9 cm was marked as point C.

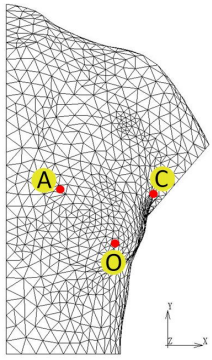


Figure 4: The diagram of the location of point O, A and C.

C. Material model and material parameters of upper body components

At first, the material parameters should be set up for different deformable element parts. Anatomically, a breast consists of various components that are adipose tissues, glandular tissues, as well as Cooper's ligaments as shown in Figure 3 (B). However, it is verified by previous researches that the breast of the elderly can be assumed as a homogeneous and in-compressible hyper-elastic material because the aging process makes the decline of breast occurs and this phenomenon becomes even more prominent on adipose and glandular tissues for females aged between 55 and 69 [13]. As a result, it was assumed that the breast deformation has a large non-linear strain, the breast accordingly should be expressed by the below strain energy density function (1):

$$W(I_1, I_2) = \sum_{i,j=0}^{n} C_{ij} (I_1 - 1)^i (I_2 - 1)^j$$

where C_{ij} is the constant which is determined by stress-strain curves, n is determined the accuracy requirements. In this study, n was equal to 2, the equation can be expressed by (2):

$$W = C_{10}(I_1 - 1) + C_{01}(I_2 - 1) + C_{20}(I_1 - 1)^2 + C_{02}(I_1 - 1)^2 + C_{11}(I_1 - 1)(I_2 - 1) \quad (2)$$

Refer to the previous reports, the material coefficients for the elderly's breast were set up as C_{10} = 4.1e-5 KPa, C_{01} = 4.3e-5 KPa, C_{20} = 6.5e-4 KPa, C_{11} = 3.1e-4 KPa, C_{02} = 5.2e-4 KPa [14]. The material parameter of the body part was using the two parameters equation, where C_{10} = 0.0167 KPa and C_{01} = 0.0017 KPa. Same material coefficients were set up for the body part. Followed by adding the boundary conditions, how to refine the loading and boundary conditions is extremely crucial for an accuracy FE model. Taken into consideration the body symmetry, the simulation was simplified using half body for improved efficiency. As above-mentioned, participant was wearing a standard bra during 3D body scanning. In order to get a realistic simulation, gravity force was exerted as compensation on the breast by 10 increments. Then a rotation center can be measured as the intersections by the extension of humerus under arm abduction from 30° to 90°. The coordinate of this rotation center in the FE model is determined as (136.01, 39.73, -10.54).

III. NUMERICAL MODEL CONSTRUCTION

The 3D body scanner (VITUS Smart XXL, Germany) was used to get an individual-specific geometric model (Figure 5). It is worth noting that the participant was required to put on a standard bra (Wacoal, Wacoal H.K. Co. Ltd.) to visualize breast roots because of the sagging problem. To achieve a smooth and cutting-out geometric model capable of constructing FE models, post-treatment has been done on the original geometric model using three different software. Firstly, the model was processed by Geomagic Design X (Rapidform XOR software, USA) to fill holes on the breast's surface, smooth uneven surfaces and delete reduplicated layers. Secondly, Rapidform (Rapidform XOR software, USA) was employed for faces subdivision, shape trimming and auto-surfacing for the following mesh creation. Thirdly, as abovementioned, PMM could be divided into two portions. One is the clavicular portion, covering half of the clavicle. The other is the sternocostal head, locating from the second to six ribs and coastal margin of the sternum. Herein, the clavicular and sternocostal portions were applied in this work for the PMM by Apex (MSC, USA).

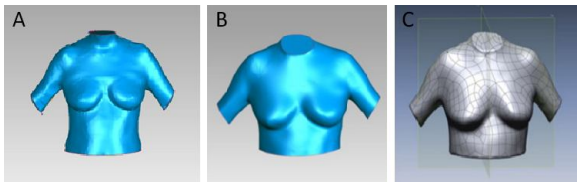


Figure 5: Image acquisition from 3D body scanner (A), the first processed by Geometric Design X (B), and further processed by the Rapidform (C).

Mesh creation of the geometric model was accomplished by Apex. The 3D 4-node tetrahedral solid element was used to discretize the model for the computational domain. The reason for using this element is that its quadratic displacement behavior is quite suitable for the modeling of the irregular shape deformation often found in the female breast and PMM. The biomechanical material model and material parameters specific for the different components in the musculoskeletal system of women

upper body were determined above-mentioned. The settings values were stated in the following Table 1. In this work, one of aims is to determine the biomechanical behavior of PMM by the analysis of theoretical and experimental data. To be specific, the Young's modulus of designated clavicular and sternocostal portion can be calculated through the variable-controlling method. Given that the Young's modulus of clavicular portion was reported to be 0.206 MPa for the young, it is reasonable to set relatively lower values of 0.2, 0.15, 0.1 and 0.05 MPa for the elderly [15]. As a result, for the material parameters of the clavicular and sternocostal portions in the PMM was estimated to set as 0.1 and 0.8 respectively. The defined material model is the isotropic-elastic model.

Table 1: Biomechanical material model and material parameters of the musculoskeletal system components applied in this work.

Material model	Component	Coefficient
Mooney-Rivlin	Breast tissue	C_{10} = 4.1e^{-5} KPa C_{01} = 4.3e^{-5} KPa C_{20} = 6.5e^{-4} KPa C_{11} = 3.1e^{-4} KPa C_{02} = 5.2e^{-4} KPa
	Body	C_{10} = 0.0167 KPa C_{01} = 0.0017 KPa

IV. RESULTS AND DISCUSSION

First of all, the motion capture results were achieved. The changes of 3D coordinates as time increasing had been recorded. Next step was to analyze the breast deformation by calculating the linear displacements of line OA and OC. The linear displacement changing the ratio of OA and OC can be calculated by the following (3):

$$\lambda = \left(\frac{\Delta L - \Delta L_0}{\Delta L_0} \right) \times 100\% \quad (3)$$

Herein, the ΔL presents the original displacement while the ΔL_0 presents the deformed displacement.

The calculated linear displacement of changing ratio on OA and OC during angle changing was derived and shown in Figure 6. It is clear to notice that the line OA (Figure 6 (A)) was contracted, whereas the OC (Figure 6 (B)) was stretched. These six participants have a similar changing trend which the average value of the final changing ratio of OA was changed by 2.5%. Compared with OA, there is a relatively larger deviation of the breast deformation as arm abducting by 60^o. OC has nearly four times larger deformation than OA. When it comes to design the sports bra, the elasticity arrangement should be considered based on this motion capture data. The motion capture results also can be used for the FE model validation. To validate the accuracy of FE model, a method of Root-mean-square error (RMSE) can be used, RMSE results reflect the data difference and the lower the better. Eventually, the optimized material parameters of the clavicular and sternocostal portions of the PMM can be determined by the minimum value of RMSE results. The rough value of material coefficient for the sternocostal portion in PMM was tuned from the 0.06 to 2.0 with a compensation value of 0.5. The RMSE (Root-mean-square error) can be expressed by (4):

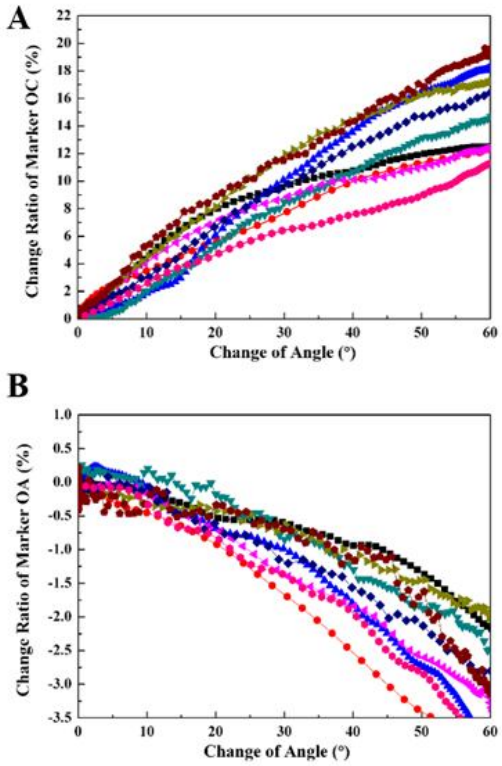


Figure 6: The changing ratio of the distance OA (A) and OC (B) during arm abducting by 60^o.

$$RMSE = 100\% \times \frac{1}{n} \sqrt{\sum_{i=1}^{n} (\Delta L_{Exp,i} - \Delta L_{FEM,i})} \quad (4)$$

Here, ΔL_{Exp} and ΔL_{FEM} are the deformation ratio obtained from the experiment and simulation, respectively; n is the number of data points generated during arm abduction. This process was calculated by MATLAB (The MathWorks, Inc. USA). The tendency is very similar, especially for the 0.2, 0.15 and 0.05 MPa. As shown in Table 2, known from RMSE results, the minimum value of OA + OC was clearly found for the 0.1 MPa (clavicular portion) and 0.08 MPa (sternocostal portion), which under the accurate requirement of error under 0.05 %. so the simulating data when the clavicular portion and sternocostal portion was 0.1 and 0.8 is the most optimized was most close to the experimental one. This result demonstrated a good fitting to the realistic experimental data. The comparison of the simulated results by FE model and the empirically experimental results of the linear displacement OA and OC was shown in Figure 7.

Table 2: The RMSE results of the clavicular portion and sternocostal portion.

RMSE%		0.2	0.15	0.1	0.05
0.2	OA	0.062	0.063	0.071	0.063
	OC	0.102	0.106	0.130	0.108
0.18	OA	0.066	0.063	0.070	0.063
	OC	0.116	0.105	0.127	0.106
0.16	OA	0.062	0.063	0.069	0.063
	OC	0.102	0.103	0.123	0.104
0.14	OA	0.062	0.063	0.062	0.052
	OC	0.099	0.100	0.101	0.048
0.12	OA	0.084	0.103	0.096	0.071
	OC	0.156	0.199	0.169	0.121
0.1	OA	0.062	0.062	0.062	0.063
	OC	0.097	0.094	0.095	0.095
0.08	OA	0.061	0.062	0.052	0.063
	OC	0.091	0.092	0.048	0.095
0.06	OA	0.119	0.101	0.089	0.089
	OC	0.074	0.182	0.071	0.158

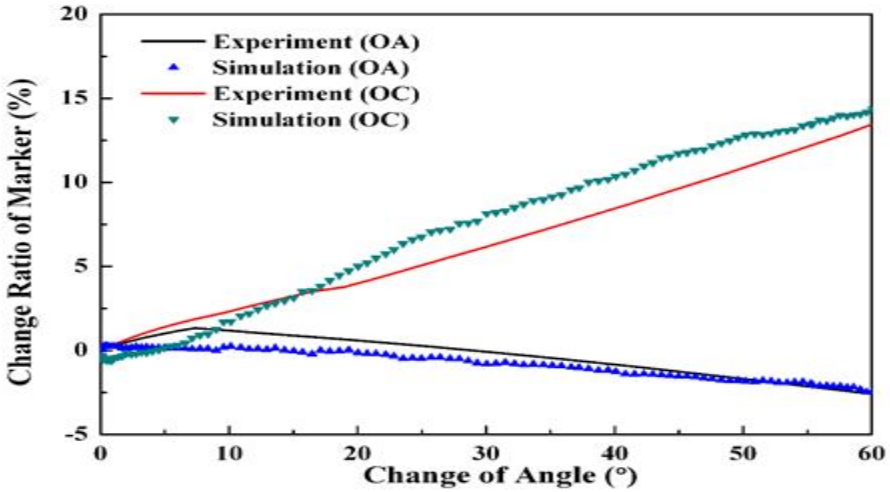


Figure 7: The comparison between the FE modelling results and the motion capture experimental results.

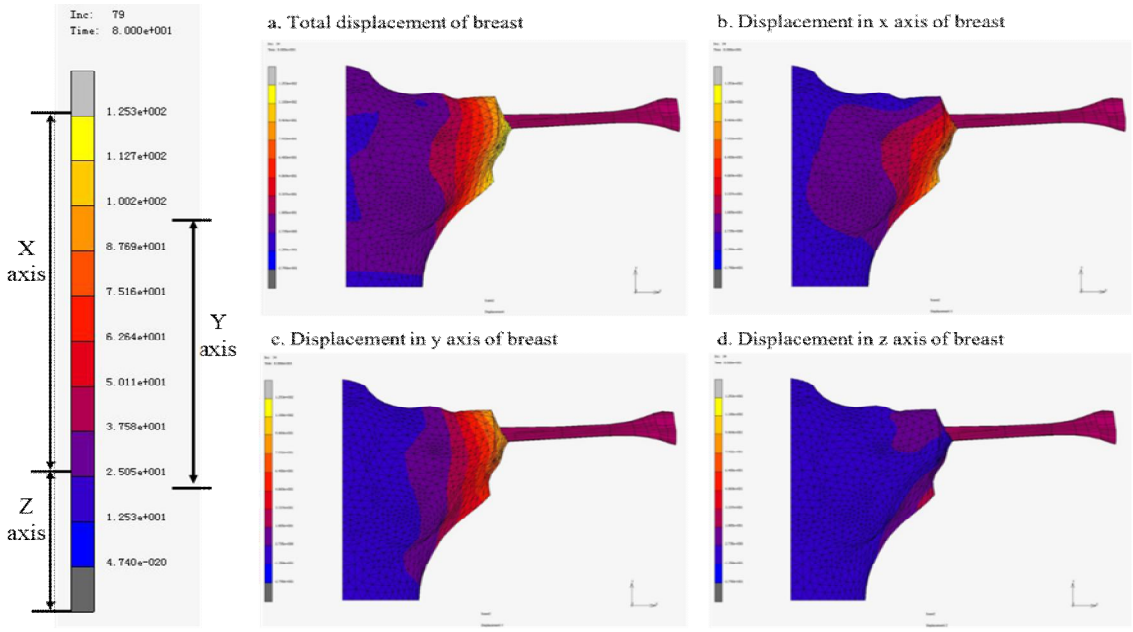


Figure 8: The displacement change ratio of breast in total (a), X (b), Y (c) and Z (d).

Using the optimal sets of material model and material parameters into the FE model, the 3D breast deformation from initial arm angle to ultimate angle was simulated and used for the prediction of whole breast deformation situation instead of just few interested points collected by the motion capture system. The displacement of breast in x, y and z directions is shown in Figure 8. It could be seen that the largest displacement change occurs in the x direction, followed by the medium and smallest one in the y and z direction, respectively. It suggested that the related breast deformation was mainly based on 2D plane and controlled by the PMM in the horizontal direction.

CONCLUSIONS

Based on the prediction results and experimental data, the Young's modulus of the PMM was accurately determined with clavicular portion of 0.1 MPa and sternocostal one of 0.08 MPa. In addition, the RMSE is as low as 0.085 %, which reflected a realistic FEM with precise material parameters. By independently evaluating the simulating results of breast and PMM, a synergistic effect was observed. In other words, the contraction and extension generated by the PMM made significant contributions to the final breast deformation, allowing the prediction of large breast motions for the elderly simply by the FEM of PMM. Besides, the data derived from FEM provided potential benefits in the ergonomic design of sports bra customized for elderly women in a convenient way. The proposed FE model laid a foundation for the elderly breasts' mechanics and can be potentially used for further comprehensive sport activities. The predicted data coupled with the questionnaire results can be used for the sports bra design. This not only fulfill the requirements of their physical but also physiological. All in all, this study provides an understanding and insight on sports bra design for the wellbeing of elderly in doing exercises.

REFERENCES

[1] D. Risius, "An investigation of breast support for older women," (Doctoral dissertation, University of Portsmouth), 2012.

[2] J. Scurr, J. White, and W. Hedger, "Breast displacement in three dimensions during the walking and running gait cycles," Journal of Applied Biomechanics, vol. 25, no. 4, pp. 322-329, Nov 2009.

[3] J. C. Scurr, J. L. White, and W. Hedger, "The effect of breast support on the kinematics of the breast during the running gait cycle," Journal of Sports Sciences, vol. 28, no. 10, pp. 1103-1109, Nov 2010.

[4] J. C. Scurr, J. L. White, and W. Hedger, "Supported and unsupported breast displacement in three dimensions across treadmill activity levels," Journal of sports sciences, vol. 29, no. 1, pp. 55-61, 2011.

[5] G. Gehlsen, and M. Albohm, "Evaluation of sports bras," The Physician and Sports Medicine, vol. 8, no. 10, pp. 88-97, Dec 2017.

[6] J. Zhou, W. Yu and S. P. Ng, "Methods of studying breast motion in sports bras: a review," Textile Research Journal, vol. 81, no. 12, pp. 1234-1248, May 2011.

[7] D. E. McGhee, and J. R. Steele, "Optimising breast support in female patients through correct bra fit. A cross-sectional study," Journal of Science and Medicine in Sport, vol. 12, no. 6, pp. 568-572, Nov 2010.

[8] N. A. Lee, H. Rusinek, J. Weinreb, R. Chandra, H. Toth, C. Singer, and G. Newstead, "Fatty and fibroglandular tissue volumes in the breasts of women 20-83 years old: comparison of X-ray mammography and computer-assisted MR imaging," American journal of Roentgenology, vol. 168, no. 2, pp. 501-506, Feb 1997.

[9] L. H. Chen, S. P. Ng, W. Yu, J. Zhou, and K. F. Wan, "A study of breast motion using non-linear dynamic FE analysis," Ergonomics, vol. 56, no. 5, pp. 868-878, Feb 2013.

[10] L. Han, J. Hipwell, T. Mertzanidou, T. Carter, M. Modat, S. Ourselin, and D. Hawkes, "A hybrid fem-based method for aligning prone and supine images for image guided breast surgery," IEEE International Symposium on Biomedical Imaging: from Nano to Macro, pp. 1239-1242, Mar 2011.

[11] T. Hopp, M. Dietzel, P. A. Baltzer, P. Kreisel, W. A. Kaiser, H. Gemmeke, and N. V. Ruiter, "Automatic multimodal 2D/3D breast image registration using biomechanical FEM models and intensity-based optimization," Medical image analysis, vol. 17, no. 2, pp. 209-218, Feb 2013.

[12] S. W. Wolfe, T. L. Wickiewicz, and J. T. Cavanaugh, "Ruptures of the pectoralis major muscle: an anatomic and clinical analysis," The American Journal of Sports Medicine, vol. 20, no. 5, pp. 587-593, Sep 1992.

[13] S. L. Dong, T. C. Chu, G. Y. Lan, Y. C. Lin, Y. H. Yeh, K. S. Chuang, "Development of an adjustable model breast for mammographic dosimetry assessment in Taiwanese women," American Journal of Roentgenology, vol. 196, no. 4, pp. 476–481, Apr 2011.

[14] Y. Sun, W. Yu, N. Lau, L. H. Chen, W. Z. Jiao, "Analysis of breast deformation in different leaning angles using finite element method," Proceedings of 14th Asian Textile Conference (ATC-14), pp. 98-101, 2017.

[15] R. D. Sacks, and R. R. Roy, "Architecture of the hind limb muscles of cats: functional significance," Journal of Morphology, vol. 173, no. 2, pp. 185-195, Aug 1982.

★★★